THE EFFECT OF AGEING ON THE CONTROL OF TOE CLEARANCE DURING WALKING

by

Simon Barrie Taylor

(M.App.Sc. – Biomechanics; B.App.Sc – PE)

School of Sport and Exercise Science,

Faculty of Arts, Education and Human Development, and

Institute of Sport, Exercise and Active Living (ISEAL),

Victoria University

Submitted in the total fulfillment of the requirements of the degree of Doctor of Philosophy

January, 2012

Abstract

Increased tripping frequency and the associated risk of falling is an international health problem in older adult populations. The average distance between the toe and the ground surface is approximately 1.3cm at the event where the toe reaches a minimum clearance height during leg swing of walking (Winter 1991, Begg et al. 2007). Little is known about how the lower limbs are controlled and coordinated to perform safe (risk-averse) toe-to-ground clearance. This thesis investigates the kinematic interactions of the lower limbs during unperturbed walking with a specific emphasis on examining the cooperative interactions of the stance and swing limb.

Two groups of 28 female participants with different mean age (25(±6) and 69(±3) yrs) were recruited to walk on a motorised treadmill for a period of ten minutes unperturbed at their self-selected walking speed. Three-dimensional lower limb segment kinematics captured from an OPTOTRAK motion capture system served as the input for a six degree of freedom biomechanical model. To investigate the toe clearance task, the lower limb segment chain was generalised as three cooperative effector systems: a stance limb effector; a swing limb effector; and the combined effector representing the collective swing and stance effectors. The swing phase of the gait cycle was the primary interest and this was separated into two normalised time regions which centred upon two toe-to-ground reference states, one at early swing and the second at mid-swing. The behaviour of the effector systems and the component segments of which they span was described using normal distribution statistics, non-linear serial correlation analyses, correlation analysis, and multi-dimensional analyses of covariance. This was performed at each time state within the two swing phase regions.

The results of the thesis reveal age-related control and coordination changes of the toe clearance task. The elderly control the toe clearance task by assigning greater level of control effort to the stance limb. The consequence is an altered representation of the toe clearance task that is dependent on the goals of the stance limb. The young group demonstrate toe clearance control statistics that indicates optimal sharing of task details between the stance and swing limbs.

Student Declaration

Doctor of Philosophy Declaration

I, Simon Taylor, declare that the PhD Thesis entitled “Control of toe clearance during walking” is no more than 100,000 words in length, including quotes, and exclusive of tables, figures and appendices, bibliography, references and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. All work in this thesis was conducted by me, with the exception of UCM and DFA software programs in MATLAB that were developed with the collaboration of Victoria University’s research assistant, Barrie Taylor.

Simon Taylor

January, 2012

Dedication

I dedicate this work to my dear wife, Belinda, and my two children Charlie and Ava. My family of cheerful faces, full of love and happiness, bring out the best in me. To all the suffering partners of PhD candidates, Belinda is well qualified to know your pain. Thankyou honey, for being my test pilot, walking many miles on the treadmill, and walking with me through this up-and-down process.

Acknowledgements

Thanks to my two supervisors, Dr Russell Best and Prof. Rezaul Begg, for your tolerance, support and careful reviews. Thanks for the wise advice during challenging times. You provided many opportunities for me to expand my teaching and research experience and ultimately provided an opportunity for me to realise a career vocation that I love doing.

Dr Daniel Lai

Thanks for your time that you spent discussing with me the deeper insights of my research and opening my eyes to the interesting concepts of coordination and control.

Mr Barrie Taylor

Thanks for all the long hours of Matlab technical support.

Biomechanics team at VU

Thanks everyone for always stepping up to help out.

Table of Contents

List of Tables

List of Figures

Introduction

The capacity to walk is important to life's activities, longevity of health and well-being; however, in older adults walking can pose an injury risk because it reduces the stability limits for maintaining upright balance. At least one in three people aged over 65 years suffer a fall; furthermore, accidental falls in older adults usually occur while walking. Falls are a major problem within this population because the immediate injury trauma sustained results in longer term consequences of reduced mobility and quality of life (Lord, Sherrington, Menz and Close, 2007). For an older adult, a serious injury from a fall can therefore represent a tipping point into a negative lifestyle. For this reason, falls in older adult populations has become an epidemic problem worldwide. The increasing ageing population worldwide is causing the incidence rate of falls to rise steadily, and this is compounding the financial burdens put upon worldwide health and aged-care systems. Current trends indicate that annual cost of falls to the Australian community in 2051 will reach AUD$1,375 Million (Cripps and Carman, 2001). The financial cost of falls in America is expected to reach US$32.4 Billion by 2020 (Englander, Hodson and Terregrossa, 1996). The cost of falls is a serious worldwide problem; however, maintaining lifestyle activities like walking are critical for older adults, even though they may increase their risk of a fall.

There are many causes of a fall; however, tripping while walking is cited as the most common. Tripping on uneven and flat terrain is cited as causing more than 50% of the falls in older adults (Berg, Alessio, Mills and Tong, 1997). A trip-related fall while walking is most common because the body enters regions of upright instability at a time when the toe is travelling close to the ground. When walking, the toe trajectory characterising the swing limb endpoint is fast moving in the direction of progression, and low to the ground at the middle of the leg swing cycle. The toe reaches a minimum ground clearance height at mid-swing (i.e. known as MTC, or minimum toe clearance). There are simultaneous tasks occurring at this phase of the gait cycle that places the upright body in a state that is susceptible to a trip-related fall following a destabilising perturbation. The body is supported by a narrow base area the size of one foot width. Maintaining upright balance must be performed dynamically within the narrow limit provided by the support foot area, and this is difficult because the body centre-of-mass is moving with a high forwards momentum with medio-lateral instability (Winter, 1992). The consequence of a trip creates a mechanical perturbation that is usually large enough to destabilise the gait pattern. Regaining gait stability is not so easy for older adults, their body faces a significant mechanical challenge in response to preventing a fall (Pavol and Pai, 2007; Pijnappels, Reeves, Maganaris and van Dieen, 2007). For example, if the toe is obstructed from its natural swing path, the moving trunk will develop an increase in forward angular momentum. Arresting this augmented trunk motion will require the appointment of a new movement plan with significant joint and muscle strength in the support limb (van Dieen, Pijnappels and Bobbert, 2005). Simultaneously, a swift movement of the swing limb to provide a timely foot position will also be required to create a new support base area from which forces can be directed to recapture the falling upper body. This strategy is difficult for some elderly persons to execute and several repositioning steps are usually required before upright balance is restored (Pijnappels et al., 2007).

The holy grail of falls prevention is to find the key markers that identify people with high falls-risk. On one hand, some older adults may display poor trip-response attributes and be classified as being at risk of falling if they sustain a trip. Another scenario is that older adults might be more at risk of initiating a trip because of a particular gait pattern. For the prevention of trip-related falls, an argument can be made that increasing trip avoidance probability is equally as important as improving trip-responses. Minimising the likelihood of a trip occurring in older adults will mean that a trip response plan will be less frequently needed. Certainly, a combined improvement outcome in trip-avoidance and trip-response will certainly have a positive effect for worldwide communities. However, when compared to trip response studies, the rationalised merit of investigating trip prevention has received slightly less focus (Barrett, Mills and Begg, 2010). While trip-response studies demonstrate significant differences between young and older adults, so far, research has not convincingly demonstrated that older adults are more likely to trip (i.e. toe-to-ground/obstacle contact) any more than a young adult (van Dieen and Pijnappels, 2007; Barrett et al., 2010). Identifying gait markers that indicate an increased likelihood of tripping is difficult because of the internal complexities of the sensory and motor systems underlying walking (i.e. the loco-sensorimotor system).

It might seem that the simplest option to reduce the probability of tripping would be for the loco-sensorimotor control system to raise the height of the toe trajectory (Begg et al., 2007; Best and Begg, 2008). However, high toe clearance is an energy expensive gait pattern; and, optimal design processes of the loco-sensorimotor control system following years of walking practice and evolution resist energy costly walking patterns. Therefore, one proposed hypothesis related to swing-toe trajectory is that the loco-sensorimotor system seeks to achieve a balanced compromise between the competing needs of both energy efficiency and trip avoidance (Begg et al., 2007). This competition is hypothesised to constrain the trajectories of the swinging-toe so they remain within a bounded region in space. During steady state walking on flat terrain, it is plausible that a narrow region of highly repeatable toe trajectories is sustainable. However, the loco-sensorimotor system is open to internally manifested perturbations that can potentially propagate into 'divergent' toe trajectories. When the loco-sensorimotor system gets to this state and toe trajectories threaten to exceed a nominal region, there will need to be some capacity within the system to allow flexible re-adjustments of sub-optimal trajectories. Therefore, evaluating the probability of tripping is arguably about measuring the performance of the loco-sensorimotor control system to dynamically manage toe-trajectories. This is the important area of trip-avoidance research requiring more attention.

The current literature detailing how the loco-sensorimotor system manages toe-trajectories, and the redundant multi-degrees-of-freedom system from which it emerges, is limited because of research confined to several areas. First, most research has been limited to the analysis of a single discrete event within the toe-trajectory, namely the minimum toe-to-ground clearance event (MTC). This is a critical event along the trajectory, however, this does not provide information about how the trajectory is managed as it approaches MTC. Second, most details of gait parameters describe the average performance of the toe trajectory across multiple repetitions, by applying descriptive statistics. While this approach provides meaningful distribution statistics and probabilities, it does not reveal whether some trajectories are selectively managed by the control system. The third area of research literature that is limited is how a redundant multi-degrees-of-freedom system cooperates to control the toe trajectory. For example, when determining how a task variable (i.e. toe trajectory) is controlled by element variables (i.e. joint or segment angles), analyses have been limited to correlations of variance among element variables, and this has no direct association with the task variable (Bianchi, Angelini, Orani and Lacquaniti, 1998; Ivanenko, Grasso, Macellari and Lacquaniti, 2002; Mills, Barrett and Morrison, 2007; Shemmell, Johansson, Portra, Gottlieb, Thomas and Corcos, 2007). Furthermore, when research studies have assigned joint angle (element) variables with the toe trajectory (task) variable, the methods have assumed a non-redundant system (Winter, 1992; Moosabhoy and Gard, 2006). Therefore, there is no information in the literature about how the system manages redundancies in the multi-degree-of-freedom system that controls toe trajectories. The last characteristic of current research that limits an assessment of how the toe clearance task is managed relates to how posture control of the stance limb cooperates with movement control of the swing limb. For example, most studies describe the toe trajectory without considering details accounting for a dual task contribution of the supporting limb (i.e. maintaining upright posture as well as contributing to control toe trajectory). The discussion above demonstrates current limitations about understanding how the toe clearance performance is managed by the loco-sensorimotor control system. While the details extracted from current methods may potentially identify a person at a high risk of tripping, an appropriate assessment of how toe trajectories are ‘managed’ by the system will be limited. The studies of this thesis aim to contribute knowledge about the control of the toe clearance task by applying new methods and new theoretical frameworks.

Gait trajectories can be characterised as stable by the convergence strength of trajectories tending towards a nominal region in state space. A loco-sensorimotor system of a healthy older adult can potentially demonstrate stable gait trajectories; however, when the loco-sensorimotor system enters into uneven terrain, a stable trajectory might not be optimal for the environment conditions. Similarly, when walking along even terrain, stable trajectories might not be conducive when gait trajectories have succumbed to internal perturbation. This emphasises the need for a toe trajectory to have the capacity to flexibly switch between stable trajectories to accommodate the environment or the internal conditions. The research to date does not indicate how the loco-sensorimotor system can switch to a nominal optimal toe trajectory if given a condition where the current trajectory is sub-optimal. Explaining how optimal gait trajectories emerge from the movement system, and are managed or controlled by the movement system, has been a major challenge for researchers in the field of robotics, neuroscience, motor control, and biomechanics. However, while all fields may follow different theories and apply different methods for describing how gait trajectories emerge and are controlled, they all consistently agree that walking involves gait trajectories displaying non-periodic limit-cycle behaviour. Because of the limit-cycle feature, nonlinear tools have been universally employed; however, some recent approaches have found that integrating specialised linear algorithms with nonlinear approaches can help describe the formation of limit-cycle trajectories (Manchester, Mettin, Iida and Tedrake, 2011). These cutting-edge mathematical approaches are exceedingly complex. However, there have been two commonly used basic tools that describe limit-cycle behaviour in walking. For example, Lyapunov analysis typically describes the general convergence behaviour of trajectories across the whole limit-cycle. Hence, Lyapunov analysis provides an indication of how well trajectories can settle upon a stable region of attraction in state-space. The limit with simple Lyapunov analysis is that it does not detail stability at specific points along the cycle. One tool that can provide limit-cycle details at discrete points of the cycle is Detrended Fluctuation Analysis (DFA). This method describes the serial correlations of the 'limit-cycle behaving' toe trajectory at discrete intersecting sections of the cycle. While the DFA does not detail the whole trajectory, it can be applied at sequentially occurring 'task-relevant' states of the cycle. In this way, the DFA can characterise the serial correlations of multiple gait trajectories at different points of the cycle. Some researchers propose that the DFA quantifies how values from a time series will be likely to continue persisting in a way that is consistent with its recent history. Similarly, the DFA demonstrates anti-persistence when trends are frequently prevented from persisting, due to some over-correcting mechanism. In this thesis, DFA values are proposed to reflect persistence-likelihood and the reduction in persistence is related to control intervention (e.g. Dingwell, John and Cusumano, 2010).

The state of the toe is determined from a kinematic chain of inter-linked segments, which forms an effector system composed of multiple-degrees-of-freedom. Although walking involves multiple effector systems, where each effector can give rise to the state of the variable it serves to control, toe trajectories arise from the collective behaviour of elements (e.g. segments and muscles) within two general effector systems of the lower limbs. The stance and swing limbs can each be defined as an independent effector sub-system, and they link at the hip position of the swing limb. A vector between a point under the support-foot to a point defining the swing-hip can describe the final kinematic configuration of the stance effector; while the swing effector can be described by a vector from the swing-hip to the swing-toe. The advantage of describing the behaviour of a stance and swing effector is relevant to the details required by the central nervous system to plan movement. A convincing argument is that the central nervous system is interested in the effective limb configuration, more so than specific element details of limb components (Bosco, Eian and Poppele, 2005; Bosco, Eian and Poppele, 2006). When considering the state of the toe position relative to the ground, it is the collective configuration of the stance and swing effectors that determines the toe position. Information from research about how these two effectors cooperate is limited with respect to toe trajectory control. The stance and swing effector systems can combine to form a redundant set of configuration solutions that result in an equivalent toe state. For creating a unique toe state, both effectors have some freedom to vary because inter-dependently, a cooperation solution can be found that meets a goal state. At a within-effector level, the segments are also redundant with respect to a goal-state in the vector orientation, as there can exist a solution set to meet this goal that is comprised of multiple configurations of the within-effector elements. The benefit of redundant systems is that there is scope for the system controller to facilitate sharing of resources between elements of the independent effector systems, particularly when multiple task goals arise simultaneously (White and Diedrichsen, 2010). For the control of toe trajectories, it is unknown how the loco-sensorimotor controller allocates task sharing between the stance and swing effector systems. There is likely to be inter-dependent cooperation that exists at two levels: (i) between-effectors; and (ii) within-effectors. In addition to dual-goal tasks (e.g. toe clearance and upright posture stability), exploiting this feature of redundancy is also critically important for a loco-sensorimotor system that has neural transmission noise contaminating the motor commands and sensory feedback (Faisal, Selen and Wolpert, 2008). The issues raised above describe the nature of the loco-sensorimotor system from which toe-trajectories emerge. So far, there has been limited information in the literature detailing how the loco-sensorimotor system manages these issues to satisfy task-relevant goal-states of toe trajectories.

In addressing the cooperation of effector elements at the within-effector level of stance and swing limbs, this thesis will apply a new analysis tool in gait analysis. The Uncontrolled Manifold (UCM) hypothesis is a linear geometric model that describes how a movement system controls against the undesirable variability (including system noise) that threatens the stability of the effector due to mis-coordination of the within-effector components. A proposed beneficial outcome of variability structuring ensues that an effector system will be stable, and therefore provide greater certainty for meeting it's assigned goal (Latash et al., 2008). A proposed disadvantage of variability structuring is that the system will be more robust to responsive actions that seek to change the effector goal (Latash et al., 2008). In the context of toe trajectory control, the outcomes of this analysis may provide age-sensitive information about the loco-sensorimotor control system.

Collectively, the issues raised above underlie the approach taken in this thesis to assess the performance of toe clearance. It is important to observe the dynamics under steady state walking conditions because self-reports of trip-related falls have been associated with walking under 'apparently' preferable conditions (Berg et al., 1997). It is possible that in such reported tripping cases, internally manifested perturbations have created sub-optimal toe trajectories that permeated without appropriate correction. If research can observe how toe-trajectories are managed during the approach to the critical minimum toe clearance event when walking under steady state conditions, we can advance our knowledge about trip avoidance. So far, only one study has applied the DFA hypothesis to describe the vertical toe trajectory at its minimum clearance state; findings show that older adults with previous history of a trip-related fall apply significantly less 'management' of toe trajectories during steady state walking when compared to healthy older adults (Khandoker, Taylor, Karmakar, Begg and Palaniswami, 2008). No studies have explored how redundancy in the system is exploited to manage movement variability and to ultimately control toe trajectories.

In summary, this thesis aims to extend knowledge about the performance characteristics of the toe clearance task and how it is affected by the normal ageing process, by investigating:

- the average kinematic details of the stance and swing limb effector systems, at task-relevant events of the swing phase;
- control intervention of the effector systems, at task-relevant events of the swing phase;
- within-effector coordination, at task-relevant events of the swing phase;
- between-effector coordination, at task-relevant events of the swing phase.

1.1 Glossary

To help the readability flow of the thesis by preventing digressions, this section will provide background definitions that may be unfamiliar to the reader.

TERM	**DEFINITION**
Allometric control	Multiple agents from 'bottom-up' and 'top-down' sources that are influencing in a non-linear way the dynamic coordination of 'agents' within a system.
Anti-persistence	The extinction of cycle-to-cycle local trends in a time-series of 'states', due to subsequent over-corrections of previous error states.
Attractor – fixed point	Final state in phase space and physical space that a dynamical system settles toward.
Attractor – limit cycle	As time approaches infinity a cyclic trajectory will stabilise if it is in the neighbourhood of a limit-cycle.
Bayesian inference	A model of probability applied for determining an optimal representation of information. A motor command requires optimal state-estimation and this reflects a 'Bayesian inference process' to account for 'noisy' representations of sensory information and feedforward (motor command) information.
Closed-loop control	Rely upon measured sensory information from external environment which allows integration with internal model for acquiring the state estimate. This information is processed by the relevant levels of the controller.
Complexity	Diverse layers of actions and interactions amongst system components which operate across multiple time scales
Control parameter	An influential 'agent' that causes change to the behaviour of the order parameter (or collective parameter).
Cost-to-go	On-line selection of an optimal trajectory of physical states that minimises a task-relevant cost
Cost function	An algorithm composed of goal-relevant variables whose values require minimisation during the process of optimising the outcome of task performance. These variables generally take the form of 'energy', 'movement variability', 'outcome risk'.
(DFA) Detrended Fluctuation Analysis	A method for quantifying persistence, or the structure of serial correlations in a time-series. A nonlinear statistical measure of persistence-likelihood. The persistence-

	likelihood parameter, 'α', is derived from the DFA process.
(DST) Dynamic Systems Theory	Time dependence of a point in geometric state-space. Observing change in state over time.
Effector	Elements making up a functional system, a kinematic chain described by a limb of inter-connected segments, can be defined as a kinematic model with end-effector mobility defined by the degrees of freedom in the system
Elemental variable	All the degrees-of-freedom components which are relevant to the performance of a movement task
Embodiment	Term to describe the interplay of information and physical processes complex dynamical system all the various components and subcomponents making up the structured anatomical (Pfeifer, 2007)
End-effector	A variable representing the distal position on the last link in a kinematic chain and can be used to characterise the execution of task goal involving an effector system
Equifinality	Allows the accomplishment of complex actions reliably and repetitively while allowing a relatively high variability in the movement particulars (Cusumano and Cesari, 2006).
Feed-forward model	An online predictive capacity of the sensorimotor system by anticipating the future states based upon the perceived current state. This is distinguished from inverse model.
Forward dynamics	Optimisation process for mapping muscle activity onto kinematic and kinetic experimental data. The solution is compared against experimental EMG data. Kinetic data is usually GRF data because of errors typically found by using inverse dynamics.
(GEV) Goal Equivalent Variance	Describes the strength of a motor synergy and any variance along this dimension will satisfy the task goal. Also referred to as task-irrelevant variance, because variations do not cause change to performance goal. From DST, this parameter describes the stability of the system.
Inverse kinematic solution	A unique solution for a kinematic chain. A redundant kinematic chain has no unique solution to the inverse kinematic problem.
Inverse model	Estimation in advance of consequences of planned motor command (*a priori* control). Distinguished from feed-forward model that is related to online prediction.

Limb/Leg (stance and swing)	Defined by the segment linkage of the foot, shank and thigh segments. The stance limb is defined by the supporting foot, shank, thigh and pelvis. The swing limb is defined by the thigh, shank and foot.
Limit cycle	A periodic-type trajectory in state-space that does not converge exactly onto a fixed path but remains asymptotically stable. For example, if a given trajectory with initial conditions lies close to a fixed region, the trajectory will forever remain attracted to the fixed region of stable equilibrium. Limit cycles describe the typical behaviour of a cyclic system, from which it repeats from relatively equivalent initial conditions and finds equivalent stable states through the cycle. Limit cycles have the ability to dissipate small instantaneous perturbations across time to ensure the cycles remain asymptotically stable.
Long-range correlations	Refers to temporal order of serial correlations in a time series that exist over medium to long time scales. Other terms are long-term dependence, persistence, fractal-like structure, *1/f* noise. The correlations are hypothesised to be derived from long-term dependent sources within a system; however, long-term correlations have shown to occur from short-term dependence.
Motor equivalence	Ability to perform a task with various task-equivalent effector configurations, such that co variations of the system elements do not effect the task outcome. Motor equivalence is a necessary requirement of redundancy.
Motor primitives	Basic foundation of co-variant muscle combinations that can be modulated by the central nervous system to perform task-specific motor actions, i.e. forming a movement synergy.
MTC (Minimum Toe Clearance)	The vertical state of the toe reaches a local minima with respect to the ground surface at the middle period of the swing cycle.
MX1 (Maximum toe clearance)	Where the vertical state of the toe reaches a local maxima with respect to the ground surface during the initial period of the swing cycle.
Nash Equilibria	An equilibrium state reached by multiple agents acting in cooperation to reach a share/mutual goal. The state of each agent cannot gain any further when the system has achieved a state of Nash equilibrium.
NGEV (Non-Goal Equivalent	

Variance)	When co-variance of an effector system causes a change to the state of the end-effector. This change is task-relevant because the end-effector state is hypothesised to be a variable requiring control. This is the variance perpendicular to the 'plane' of the uncontrolled manifold (UCM).
Noise	The unpredictable neuro-physiological variations within the sensorimotor processes that contribute to state estimate error and motor actuation errors. Noise contributes to variability of a task variable, however, it is not the only contributing source.
Order Parameter	A 'meaningful' variable that expresses the collective coordination of the system. The MTC can be rationalised to be an order parameter expressing meaningful coordination of relational subsystems inherent to the locomotor system. An order parameter expresses synergy behaviour.
Open loop feedback control	Rely upon internal models of control, such as inverse model and feed-forward model
(OFCT) Optimal Feedback Control Theory	Based on a hierarchical model of motor control and incorporates key principles of coordination and control – one of the few motor control mechanistic models that have reflected self-organisation and embodiment concepts from dynamic systems theory.
Passive control	Using the biomechanical attributes of a system – e.g. inertia and musculo-skeletal properties – to stabilise movement. Generally receives limited neural input from the control hierarchy. However, muscle stiffness is also a feature of 'passive' control, but this requires prior neural command intervention.
Performance variable	A variable that defines the performance outcome of a task. A variable belonging to a cost function that defines optimal task performance.
Persistence	The cycle-to-cycle local trend of a time series that decays slowly across cycles.
Perturbation	Disturbance that disrupts a trajectory stability of a parameter away from its preferred state
Redundancy	Multiple configuration options within a system that lead to one unique solution
Self-organising system	Dynamic coordination and stabilisation between multi-coupled oscillators driven by local agents and allometric control, rather than direct hierarchical control.

(SCA) Serial correlation analysis	Analysis of temporal structure in a time series
Stability	Ability to preserve and maintain a stable behaviour in a variable environment - system ability to offset external perturbations by returning or finding a suitable alternate stable state. Capacity of a system to restore its state after a phasic perturbation away from that state.
State	A physical property that describes a body. In this thesis, it is related specifically to a kinematic property that describes an effector system in this thesis.
State estimate	Estimated position and orientation of a task-relevant body part with respect to a movement plan and the external environment
State-space	A vector space that describes the combined properties of motion and configuration/position of a physical system
Synergy	A functional concept of coordinating elements of a system to meet a meaningful goal. A hypothetical control structure in the motor system that organises components of various levels into a single coherent unit.
Task space	The space that details the physical properties of the components contributing to the task performance
Task-relevance	Details that affect the goal of the task variable, i.e. details that cause change to the end-effector state.
Task variable	A parameter that represents the task outcome of a collective system
Toe-swing	The state of the toe relevant to the swing limb during walking
(UCM) Uncontrollable Manifold hypothesis	A 'planar space', existing in multi-dimensional space, that contains a set of solutions where each solution solves the task goal. Specifically, it is a method that maps effector co-variance to end-effector variance in task-space to create a linear estimate of a manifold 'plane' in n-dimensional task-space. The manifold has two components of variance that is hypothesised to be controlled differently. Variance parallel to the manifold is termed goal equivalent, where any effector co-variance in this dimension is considered task-irrelevant, i.e. it does not change the state of the task variable. Variability along this parallel dimension is suggested to be 'uncontrolled' because the task goal is not affected.
Variability	The change in a task variable from one repetition to another. This can be due to many sources that can be predictable and non-predictable. For example, sensory

	noise and noise in the motor actuators make inference of state-estimate difficult and this leads to variability. More predictable sources of variability within the systems are dynamic changes to cost policies, motor-control management strategies, and non-linearities in the musculo-skeletal system.
Workspace dimensionality	Relates to the embodied loco-sensorimotor system, which contains the operational repertoire of redundant solutions for satisfying task goals in an optimal way. The operational repertoire relates to dimensionality of motor primitives available to exploit during task performance.

Chapter 2

Literature review

This section explores the research field on areas related to the problem of controlling the toe trajectory during the swing phase of walking. The sections will be divided into four themes:

- reducing the problem of trip-related falls in older adults;
- the embodied system from which toe-trajectories arise;
- theories of how toe-trajectories are shaped;
- quantifying toe clearance performance.

Problem of falls and trips in older populations:

This first section relates to the worldwide problem of falls and the context of trip-related falls. This section also discusses what research approaches have been undertaken to identify persons or groups at risk of trip-related falls.

Context from which toe-swing performance arises:

This second section brings a holistic context to the toe clearance task. First, the general purpose of walking is discussed and explores sub-tasks that underscore this purpose. The toe clearance task is then defined as a dynamic performance task, such that it is the expression of an adequate solution arising from a given body-environment situation. It is also demonstrated that toe clearance is shaped by competing constraints and competing sub-tasks of walking. Toe clearance is determined by toe trajectory, which in-turn is shaped from a complex and redundant loco-sensorimotor system. This provides a difficult challenge for researchers to solve a tractable cause-effect map through the neuro-muscular-skeletal system. In concluding this section, it will be argued, that theories in line with embracing these issues should be considered as a means to assess toe clearance performance in an appropriate context.

Theories of how coordinated and controlled movement trajectories emerge from the sensorimotor system:

This third section describes two different theories, dynamic systems theory and optimal feedback control theory, to explain how the toe clearance task is potentially controlled by the sensorimotor system. General perspectives of both theories have been applied in biped robot designs, however, dynamic systems theory has been the main philosophical theory for explaining human walking. These two theories will also provide context for describing how the age affected sub-systems affects the process underlying toe trajectory control. The theories described and their applications will provide a rationale for selecting tools that can capture the performance of the toe clearance task.

Quantifying toe clearance performance:

The fourth section explores previous methods that have examined toe trajectories of the swing phase. Building upon this current information, two general methods are proposed to add further insight into how the toe clearance task is performed because they represent a holistic appraisal of the loco-sensorimotor system. One method is serial correlation analysis, which is proposed to reflect the integrating dynamics of the passive autonomous system and the actively controlled volitional system. The second method is the uncontrolled manifold hypothesis, which is designed to capture a coordination structure of the redundant system that leads to control of the goal-directed toe clearance task.

2.1 Falls, trips and gait in older adults

Falls in older people is a serious national and international health concern with major consequences to the health and well being of the older person who has suffered a fall, the burden placed on their family members, and the strain placed on community resources. Understanding the cause of falls is important. This section will outline the nature of falls in older people by reviewing studies that have used report surveys and observational tests. Specific attention will be given to the risk factors associated with falls. In concluding this section, the implication of the findings, that walking variability is associated with falls, has for future investigative research identifying persons with high risk of trip-related falls is discussed.

2.1.1 Epidemiology and aetiology of falls

In Australia at least a third of people aged 65 years and over ('older adult' group) fall at least once each year (Lord, 1994). This is also believed to be an underestimate due to non-reporting of non-injurious falls and some people suffer multiple falls (Lord et al., 2007). This incidence rate of falls is commonly reflected in studies for both rural and urban areas (Lord, Ward, Williams and Anstey, 1993; Luukinen, Koski, Hiltunen and Kivela, 1994) and throughout other parts of the world (Tinetti, Speechley and Ginter, 1988; O'Loughlin, Robitaille, Boivin and Suissa, 1993; Luukinen et al., 1994). A fall occurring during walking has been defined as "unintentionally coming to ground or some lower level, other than as a consequence of sustaining a [external] blow, loss of consciousness, sudden onset of paralysis as in stroke or epileptic seizure" (Gibson, Andres, Isaacs, Radebaugh, and Worm-Petersen, 1989).

Older community-dwelling people report that at least 50% of their falls occur during walking-related activities (Tinetti et al., 1988; Campbell, Borrie and Spears, 1989; Berg et al., 1997; Hausdorff, Rios and Edelberg, 2001) and only a smaller proportion of falls result from other activities such as stair climbing within the home (Lord et al., 1993). It is reasonable to predict that foot contact on unseen obstacles along the walking path, or an inability to adequately negotiate obstacles in the walking path environment can potentially lead to falls. The proportion of falls which can be attributed to a trip during walking varies between studies: Berg et al. (1997) reported 34%; Blake et al. (1988) reported 53%. Two other studies reported 28% and 44% when grouping trips, stumbles and slips in the same category (Tinetti et al., 1988; Roudsari, Ebel, Corso, Molinari and Koepsell, 2005). In Australia, two reports by the Australian Institute of Health and Welfare (AIHW) investigated hospital separations due to fall injury in persons older than 65 years, and the specific mechanism of the fall. Cripps and Carman (2001) found that 39% of this reported falls came under the slips, trips and stumbles category. The Cripps and Carman (2001) report suggested that the rate of falls belonging to the general category of trips, stumbles and slips is potentially much higher than reported because a large proportion of the classified fallers do not specify the type of fall. In a more recent AIHW that included a much larger cohort of reported falls in persons older than 65 years (*n* = 22,801), evidence showed that 61% of the those persons sustaining a 'fall on the same level' was due to tripping, and 27% due to slipping (Kreisfeld and Harrison, 2010). Furthermore, females accounted for a 75% proportion of these trip-related falls (Kreisfeld and Harrison, 2010). This finding that there is a high proportion of females that display a higher rate of falling is supported by other research (Campbell et al., 1989; Lord, 1993). The rate of falls due to slips, trips and stumbles rises exponentially and begins much sooner in females compared to males. The rate of falls has been shown to increase exponentially with increasing age (Cripps and Carman, 2001). The rate of falls in older females between the age of 65-69 years is comparable to males at ten years their senior (Kreisfeld and Harrison, 2010).

2.1.2 Immediate and long term consequences of falls

It is likely that the differential proportion of fall rates between the sexes does not reflect the frequency of trip-related falls in males, because males may not be as likely to be at risk of bone fractures of the wrist or hip due to their moderately higher ratio of 'bone strength' to 'fall load magnitude' (Riggs, Melton, Robb, Camp, Atkinson, Oberg, Rouleau, McCollough, Khosla and Bouxsein, 2006). Therefore, in the 'apparently' fit and healthy female at post-menopause, it is particularly important to minimise the risk of a fall because their wrist and hip bone structures generally indicate a reduced ability to withstand the impact load brought about by a fall (Riggs et al., 2006). Studies have reported slight differences in injury consequences of falls which are largely dependent on sub-group ages within the ‘older adult’ population (e.g. 65-74, 75-84, 85+ years of age), where between 22% and 60% of fallers suffer injury (Lord et al., 2007). The average cost per injured elderly person is nearly double that for injured persons of other ages (Rice and Mackenzie, 1989). In the US a report to congress revealed that persons aged over 70 years will spend 70% of their total health care costs on injuries related to falls (Rice and Mackenzie, 1989).

The Australian AIHW report by Kreisfeld and Harrison (2010) indicated that the most common bone fracture injury reported is the hip (26%) and the second most common is the wrist (8%). The age-related deterioration of bone strength at the hip appears to match the incidence of hip fractures (Riggs et al., 2006). The consequential nature of a hip fracture is much more severe in nature, as it not only requires a longer stay in hospital compared to a wrist fracture, but they are at risk of a poor recovery. In a large cohort of admitted hip fracture patients (from a sample of n = 674), it was reported that between 18-33% died within 1 year of the incidence (Magaziner, Hawkes, Hebel, Zimmerman, Fox, Dolan, Felsenthal and Kenzora, 2000). It was also found that between 25-75% of elderly hip fracture patients that were previously living independently, did not regain the same independent level of living prior to hip fracture incidence (Magaziner et al., 2000).

People who have experienced more than one fall show a strong prevalence of subsequent falling (Maki, 1997; Friedman, Munoz, West, Rubin and Fried, 2002). Figure 2.1.2.1 illustrates the interactive consequences of falls experience. Ultimately, immobilisation is the consequence older adults must avoid, in preserving their quality of life. Experiencing a fall, or having an associated experience such as witnessing someone else suffering a fall, can consequently lead to a ‘fear of falling’ that has a disabling long-term effect on an active lifestyle. For example, a developed fear of future falls can cause self-imposed mobility restrictions, leading to morbidity and loss of independence (Tinetti et al., 1988; Marottoli, Berkman and Cooney, 1992; Friedman et al., 2002; Lord et al., 2007). Fear of falling is evident in 48% of those people who have suffered a fall (Tinetti, Mendes de Leon, Doucette and Baker, 1994). A fear of falling can result in referred admissions to an institutional setting (Lord, 1994; Tinetti and Williams,

1997) and thus elevate dependency and long term care. A fear of falling can be a realistic appraisal of functional ability that accurately reflects the likelihood of future falling. Alternatively, a fear of falling could potentially be an irrational appraisal of a healthy functional ability resulting in an unnecessary curtailing and restriction to lifestyle. Additionally, the irrational appraisal of poor functioning could cause the gait patterns to become unnecessarily unstable. This remains an unknown outcome, whether a fear of falling can cause neuro-mechanical abilities to cause mis-coordinated and unstable movements and cause a higher probability of future falls.

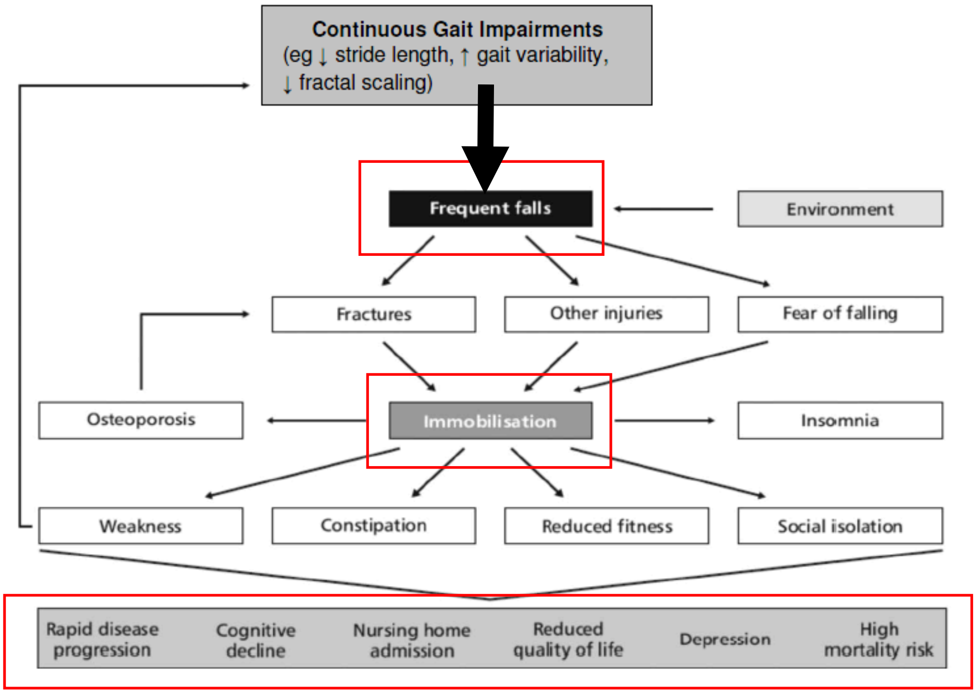


Figure 2.1.2.1.

Flow diagram demonstrating the factors involved in a 'viscous cycle' effect of gait impairments on falls, falls on immobilization, immobilization on quality of life and level of care. The final outcome of falls and immobilization is listed in the grey box below. Adapted from Hausdorff et al. (2009).

Comparatively, figures show that the direct cost of falls to the community is not only the highest ranked among all injury categories, but it is five times greater than the second ranked, road traffic accidents (Potter-Forbes and Aisbett, 2003). A compounding future projection predicts that this cost will increase due to population growth trends for the 65 years and over population group. The 65 years and older age group in Australia have increased by 72% over the last 20 years (Australian Bureau of Statistics., 2009). The 65 years and older group currently represents 13% of the population . Projections by the Australian Bureau of Statistics (2009) estimate a growth of this group to represent 23-25% of the total population by 2051. For the older elderly (above 85 years); projections estimate that this group will increase from 1.2% to between 4.9-7.3%. If current trends in rate of fall-related injury continue in Australia, by 2051 the direct annual health care cost will be AUD$1,375 Million (Potter-Forbes and Aisbett, 2003). Internationally, the financial cost of falls in America is expected to reach US$32.4 Billion by 2020 (Englander et al., 1996). The holy grail of falls research is to find a parameter which can identify in advance those people at risk of falling so that preventative measures can be undertaken.

2.1.3 Intrinsic and extrinsic risk factors of falls

Generally, the cause of falls may be categorized into two groups: intrinsic and extrinsic factors. Increased age- and pathology-associated changes to the multiple systems that control balance and locomotion are intrinsic mechanisms contributing to falls. Extrinsic factors elevate the risk of falls related to environmental hazards, such as poor lighting, footwear, home settings (bathrooms and floor rugs), uneven and/or slippery footpath surfaces. Evidence for such factors with published evidence for having a ‘strong’, ‘moderate’, ‘weak’, or ‘little’ association with falls have been outlined by Lord et al. (2007). Table 2.1.3.1 summarises the intrinsic and extrinsic factors associated with falls. These factors are classified into intrinsic factors (socio-demographic; postural instability; sensory and neuromuscular; psychological; medical) and extrinsic factors (medication; environmental). Interestingly, there has shown to be an effect of footwear on balance ability, but the type of shoe characteristics has yet to demonstrate a conclusive relationship with falls risk (Menz, Morris and Lord, 2006; Menant, Steele, Menz, Munro and Lord, 2008). Following several research studies and systematic reviews of the research literature, it is proposed that shoes with high heels or soft soles should be avoided to ensure stable walking patterns (Menz et al., 2006; Menant et al., 2008; Menant, Steele, Menz, Munro and Lord, 2009).

Table 2.1.3.1

Intrinsic and extrinsic associations with falls (Modified from Lord et al., 2007).

Intrinsic factors
Socio-demographic: STRONG - advanced age, ADL/mobility limitations, history of falls MODERATE - female gender, race, living alone, inactivity, walking aid use
Postural instability: STRONG - impaired sit-to-stand transfer ability, reduced gait velocity/cadence/step length MODERATE – impaired stability when standing, impaired stability when leaning and reaching, slow voluntary stepping, increased step timing variability
Sensory and neuromuscular: STRONG – poor visual contrast sensitivity, decreased depth perception, reduced vibration sense, reduced tactile sensitivity, reduced muscle strength, poor simple reaction time, poor choice reaction time MODERATE - poor visual acuity, reduced proprioception
Psychological: STRONG – increased fear of falling MODERATE – impaired selective attention
Medical: STRONG – impaired cognition, stroke, Parkinson's disease, number of chronic conditions MODERATE – depression, abnormal neurological signs, incontinence, acute illness, arthritis, foot problems, dizziness
Extrinsic factors
Medication: STRONG – use of multiple medications, benzodiazepine use, antidepressant use, anti-psychotic use, psychoactive medication use
Environmental: WEAK ASSOCIATIONS ONLY

There doesn't appear to be any consistently reported environmental hazard that has an association with falls and this is possibly due to the transient nature and contingency of such exposure to certain hazards. It has been reported that certain home hazard reduction interventions are effective when targeting persons with a history of falls (Nikolaus and Bach, 2003).

2.1.4 Interventions that show success for improving falls risk

There has been a recent ‘Cochrane report’ that reviewed 110 research studies, across 15 countries, where the outcomes of the reviewed studies was relevant to falls-risk / falls-occurrence reduction in community dwelling older adults (Gillespie, Robertson, Gillespie, Lamb, Gates, Cumming and Rowe, 2009). In terms of exercise interventions, it was found that programs which incorporated a combination of at least two components of exercise, balance, flexibility, and endurance were found to be successful at reducing falls-risk / falls-occurrence. For persons without a significant physical impairment (e.g. following hip fracture or stroke), the effectiveness of ‘exercise’ intervention programs has been found to be successful under different settings, such as individualised prescription of home exercise classes, Tai-Chi (considered as a multiple component exercise intervention because it targets strength and balance), or targeted exercise classes in a group. Functional training with multiple modalities has greater likelihood of reducing falls compared to general resistance training program (Gillespie et al., 2009). The effect of a single intervention, such as general walking groups, does not improve the reduction of falls-risk / falls-occurrence. In summary, the ‘Cochrane report’ by Gillespie et al (Gillespie et al.), effectively demonstrated that certain exercise interventions are a successful means for reducing falls.

2.1.5 The gait cycle

This section is a slight diversion but provides a necessary description of the general phases of the walking cycle that are being discussed in the proceeding sections. There are various ways of describing these phases because of the many repeating events which can be used to define walking sub-phases. A common description of gait relates to whether the size of the body’s base of support is spanned by one foot or two feet, i.e. double support phase and swing (single stance) phase. A step includes a period of double support (i.e. pre-swing) followed by single support (i.e. swing). A stride is defined by two consecutive steps, indicating that a stride is a true measure of one limb’s cycle.

Falls in older adults often occur during walking because the upper body mass is deftly balanced above a small support base, where balance can be easily perturbed. Body parts during walking can be described as gait trajectories in space and time. During temporal moments of the step cycle (period between the timing of consecutive, opposite-limb, heel contacts), gait trajectories representative of balance conditions can approach regions of instability. Small internal or external mechanical perturbations can lead to unbalanced upright posture and a fall. Describing the conditions for balance during this step period is similar to describing the very basic mechanical features of a 'passive dynamic walker' (e.g. McGeer, 1990). A 'passive dynamic walker' is a machine device composed of parts and joints that walk stably down a slope by transferring potential energy to kinetic energy, without any motor actuation or control. The 'walker' has a centre-of-mass positioned above two oscillating pendulums (sometimes designed with curved feet) that act to transfer body mass by acting as energy conserving inverted pendulums. The mass of the 'passive dynamic walker' is stabilised by an envelope region roughly the size of the changing contact area of the feet of the pendulum. In the 'passive dynamic walker' the centre-of-mass 'vaults' upwards and then 'falls' downwards, but maintains upright 'posture' at the conclusion of the step cycle when the point mass is re-stabilized by the deft timing and positioning of the foot from the swift forwards moving swing limb (pendulum). Following this very brief double-contact period, where the base of support formed by both pendulum feet have temporarily created a large envelope area, the roles of the pendulum are then alternated and the point mass begins its next cycle of 'vaulting' and 'falling' forwards. The trajectories of the pendulum angles show stable limit-cycle behaviour, demonstrating that gait can be stable using passive properties alone. However, falls in the passive walking robots are frequent, being generally accredited to a perturbing energy transfer from foot-to-ground collisions that are too large to be corrected by 'passive' control alone. Within the last decade, advanced designs of actuated biped robots have began to show human-like dexterity when walking over level ground by using minimal actuation and very general control laws (Collins, Ruina, Tedrake and Wisse, 2005). However, the current challenge of replicating a 3D biped robot to walk with efficient dexterous autonomy over uneven terrain has proven too difficult to solve. The problem faced by designers is determining how sub-optimal gait trajectories can flexibly switch to a nominal trajectory from a set of candidate stable gait trajectories. In humans, the older adult body seems to face a parallel challenge when traversing in uneven terrain. Understanding the challenges faced by older adults can be explored through some of the challenges of biped robot designers.

Leg swing performance is preceded by the initial conditions during the double support phase. A view for double support performance depends upon the initial conditions created by the terminal motion of the leg swing. The double support phase is a task where the initial conditions of balance and momentum need to be created for facilitating the task needs of leg swing control.

2.1.6 Gait parameters associated with falls

The following research approaches have investigated the intrinsic risk factors to ascertain which parameters are best predictors of fallers. Hill et al. (1999) compared measures of balance, muscle strength, gait, fear of falling and general health in a multifactorial investigation. Included were healthy active community dwelling women aged at least 70 years (n = 96). In this study Hill et al. excluded participants who showed decreased sensory and cognitive functioning. From multivariate logistic regression analysis, measures of gait performance (gait asymmetry and double support time) were most strongly associated with multiple fallers. This supports the usefulness of gait as a movement task to screen for falls risk.

Hausdorff et al. (2001) conducted a 1 year prospective study by assessing multiple intrinsic factors similar to those listed in Table 2.1.3.1 and also conducted an examination of specific walking parameters using average and standard deviation measures. Although the study did not report whether the sample of community-dwelling older adults presented a history of falls prior to the study, the group displayed relatively normal measures of healthy functioning in a variety of attributes. During the follow up period, 40% of the group reported a fall and 75% of the falls occurred while walking. The authors did not report whether the fall was the result of a trip or slip. Measures of gender, age, mental health (depression), health status (number of medications), physical ability level (activities of daily living) were not significant delineators between fallers and non-fallers. Most of the administered functional tests, such as the physical performance test, average gait parameters (e.g. walking speed and swing time), functional balance and reach tests, and a timed up and go test did not show significance between fallers and non-fallers. However, the authors found similar results to Maki (1997), that increased gait variability was a significant indicator of future fall risk. An odds ratio analysis by Hausdorff et al. (2001) showed that a small increase in stride time and swing time variability can increase fall risk by a factor of five, and two, respectively. Of further interest, correlation analysis revealed that many of the intrinsic factors were significantly associated with gait variability, such as mental and general health status, balance ability, upper and lower body strength, ability to perform daily activities and functional tests. This study strongly supports the use of gait variability measures for fall risk assessment and the follow up of thorough investigations into the nature of gait variability. Hausdorff et al. (2001) conclude that gait variability appears to reflect disease processes rather than normal aging.

Studies show fear of falling is associated with reduction of walking speed (Tinetti, Richman and Powell, 1990; Maki, 1997) and a higher risk of falling (Maki, 1997). However, it is unknown whether the outcome of a fear of falling produces a 'safer' walking pattern in the form of stable gait trajectories. The reduction in stride length and walking speed from a fear of falling has not demonstrated a reduction in subsequent falls (Maki, 1997; Hausdorff, 2007; Menz, Lord and Fitzpatrick, 2007). The general older adult population walks more slowly than young adults (Winter, 1991b; Prince, Corriveau, Hebert and Winter, 1997), older adults with falls history walk even slower (Hausdorff, Edelberg, Mitchell, Goldberger and Wei, 1997a; Hill et al., 1999; Hausdorff et al., 2001). There are contradictory beliefs as to whether slower walking is actually beneficial for improving walking stability. Studies have shown that a slower walking speed appears to dampen destabilising perturbations existing in walking patterns (Dingwell, Cusumano, Sternad and Cavanagh, 2000; Pavol, Owings, Foley and Grabiner, 2001; Weerdesteyn, Nienhuis, Geurts and Duysens, 2007). However, because studies show that walking slowly increases gait variability (Dingwell and Marin, 2006), there is a reasonable view that older adults might benefit from an increase in gait stability if they walk faster. In a follow-up investigation to clarify this issue, the same authors found that age-related changes in walking variability were independent of walking speed

(Kang and Dingwell, 2008a). The authors proposed that the cause of walking variability is not due to the gait regime, but stems from internal ageing processes in the musculo-skeletal systems, namely reduced muscle strength and joint range of motion.

In summary, the consequences of a fear of falling depend upon the severity invoked upon immobilisation. At best a fear of falling is necessary to reduce walking speed so as to help stabilise gait trajectories, but at worst it disables mobility and curtails quality of life. General measures of gait instability have a strong association with falls risk and it is difficult to know whether an 'irrational' fear of falling will negatively affect gait stability. The consequence of a reduced walking speed has conflicting outcomes because speed will affect the variability of different gait parameters differently (Brach, Studenski, Perera, Vanswearingen and Newman, 2007a; Moe-Nilssen, Aaslund, Hodt-Billington and Helbostad, 2010).

2.1.7 Summary

This section has emphasised a strong association between gait and falls (Hausdorff et al., 1997a; Brach, Berthold, Craik, VanSwearingen and Newman, 2001; Owings and Grabiner, 2004; Brach, Berlin, VanSwearingen, Newman and Studenski, 2005; Brach et al., 2007a). However, these parameters of gait do not have a specific functional link with types of falls. Trip-related falls result in a high proportion of falls in older adults. So, how does increased step timing variability or step width variability have a functional link with tripping? It is unknown how gait time-distance parameters are associated with tripping risk or toe trajectories during leg swing (Barrett et al., 2010). There needs to be more detailed observations of the gait pattern to be able to arrive at specific solutions as to what is causing trip-related falls. For example, a recent study by Dingwell et al. (2010) explored how time-distance parameters are highly variable but the nature of inter-dependent cooperative interactions allows a goal outcome to be controlled. This redundancy philosophy was nicely illustrated in a theoretical paper by Cusumano et al. (2006). The paper by Dingwell et al. (2010) emphasises that variability details of a parameter can potentially be irrelevant because it is unknown whether such variability is beneficial for controlling a functional task goal, because it could theoretically help cooperative interactions with other parameters.

From this perspective, the question of why older adults are potentially more at risk of falls because of general gait variability requires an approach that is in-line with a redundancy philosophy. This is the approach considered in this thesis for examining toe trajectory differences between young and older adults.

2.1.7.1 Plausible parameters of gait specific to tripping risk

As a response to the issue of specificity when linking gait parameters to falls, research has began focusing on analysing the task-specific performance of the toe-to-ground clearance parameter during the swing phase of walking, termed minimum toe clearance (Winter, 1992; Mills and Barrett, 2001; Moosabhoy and Gard, 2006; Begg et al., 2007; Mills et al., 2007; Best and Begg, 2008; Khandoker et al., 2008). This thesis aims to build upon these studies by applying a redundancy philosophy to the way the task is controlled.

2.2 The context from which toe-swing performance arises

The biomechanical conditions during walking make the body relatively unstable at the time of mid-swing, when the foot is most likely to collide with the ground or obstacle (Winter, 1992). These unstable mechanical conditions are the likely basis for tripping frequency being linked with falls frequency (Pavol, Owings, Foley and Grabiner, 1999). For older adults, the consequences of a fall pose a relatively greater health risk due to reduced bone density and therefore a higher probability of bone fracture (Chen, Simpson, March, Cameron, Cumming, Lord, Seibel and Sambrook, 2008b; Chen, Simpson, March, Cameron, Cumming, Lord, Seibel and Sambrook, 2008a; Chen, Sambrook, Simpson, Cameron, Cumming, Seibel, Lord and March, 2009). Therefore, tripping should be avoided by controlling the trajectory of the toe during leg swing. However, the loco-sensorimotor system is complex with redundant features for satisfying multiple task-goals. When multiple task goals appear simultaneously it introduces a performance compromise among the multiple systems. This major sub-section (2.2) of the literature review will show why an assessment of toe trajectory control needs to consider the context of the task.

2.2.1 The-goals of walking

The evolution of walking would suggest that the general purpose of walking is to mobilize the body towards an intended direction by using limited energy and sparing cognitive resources. For beneficial reasons, upright bipedal gait was the preferred mode of locomotion probably because it allowed secondary tasks to be performed, such as sensing the environment and performing secondary movement tasks. The goals of walking within a gait cycle are therefore related to meeting the general purpose of minimising cognitive demand, minimising energy expenditure, and maintaining upright posture. All other mechanical tasks of walking are likely to have emerged so that these general purposes can be performed optimally. In doing so, the neural-muscular-skeletal systems underlying the walking process (i.e. loco-sensorimotor system), have also evolved optimally to precede the accomplishment of these overriding needs.

In motor control, the performance of a movement goal is achieved by a subgroup of task-specific anatomical components belonging to a functional unit in the sensorimotor system, termed effectors (Diedrichsen, 2007; Latash et al., 2008). The neuro-anatomical system underlying walking is termed the loco-sensorimotor system. There are two primary effectors of the loco-sensorimotor control system that mechanically define the toe trajectory, the stance and swing effector mechanical systems. Whether implicitly, or explicitly, gait analysis is generally linked around the performance of several movement sub-tasks that provide upright posture and mobility (Winter, 1991a; Neptune, Clark and Kautz, 2009a):

1) providing energy for upright body support;

2) providing energy for forward progression of the body;

3) providing energy to accelerate the leg into swing;

4) controlling the body centre of mass and base of support relationship;

5) controlling the foot trajectory during leg swing.

The above sub-tasks are not performed in isolation and there are moments in the gait cycle where goals of the sub-tasks are performed simultaneously and other times where goals of the sub-tasks are performed sequentially. Mechanically, the stance and swing limbs are connected through the pelvis segment causing the kinematic task-goals of the swing limb to be dependent upon the kinematics of the stance limb. To adequately assess performance of the toe clearance task, there needs to be evaluation of inter-dependent task goals separately assigned to the stance and swing limbs. When two effectors can collaborate together to achieve a common goal, the performance outcome becomes optimal (Diedrichsen, 2007; Gorniak, Zatsiorsky and Latash, 2007; Diedrichsen and Dowling, 2009). This raises the issue of when during the gait cycle, the stance and swing effectors have separate goals and shared goals. This section will explore these issues.

As outlined above by Winter, the mechanical goals of walking relate to foot control during leg swing, controlling balance, and providing energy for upright posture and raising the centre of mass ‘upwards and forwards’ over the support limb. The two general biomechanical conditions related to tripping are linked to two simultaneous tasks of (i) control of the vertical path of the toe during leg swing and (ii) control of whole-body balance.

In real life situations, the environmental conditions in which we walk are changing. The ground surface can change, the lighting and visual field changes and, therefore, our sensory information about the world needs to update as we encounter these changes. The visual, vestibular and somatosensory systems measuring this information combine multiple sources of information to ensure greater certainty of the body state motion within the environment (Patla, 2003; Wolpert, 2007).

During the initial conditions for leg swing at the pre-swing phase of double support, the initial body state undergoes a fast medial-to-lateral shift in momentum, and there is an associated change in whole body angular momentum (Winter, 1991a). At this stage the locomotor control system must plan a coordinated strategy to address dual-control tasks: balance of upright posture and mobilising the swing leg (Pandy, 2001; Ivanenko, Poppele and Lacquaniti, 2004). Optimal performance of both tasks is enhanced when the task goals are presented sequentially so that information about the goal can be shared among effectors (Diedrichsen, 2007; Howard, Ingram and Wolpert, 2008; Diedrichsen and Dowling, 2009). No studies have considered investigating the collaboration between the stance and swing effectors in the assessment of toe clearance task.

2.2.1.1 Balance and upright posture

In walking, balance is commonly assessed using a linear description of the COM-BOS relationship, but balance also involves rotational control of the whole body angular momentum (Hof, 2007). Walking is a whole body movement task involving rotations of body segments in all of the three planes of motion; sagittal, frontal and transverse. In healthy gait, segmental motion of the arms and legs is predominantly in the sagittal plane, whereas pelvis and trunk range of motion occurs equally in the transverse and sagittal planes (Whittle and Levine, 1999).

Medio-lateral balance during walking is a highly regulated task-goal (Bauby and Kuo, 2000). An example of when medio-lateral balance requires regulation is when the horizontal position of the body centre of mass exceeds the lateral limits of the ground surface description of the support envelope (e.g. during single stance of early swing phase). At the beginning of double support the body's base of support can be described by an enveloped area in the horizontal plane which is as large as the ground contact area formed by the feet (Winter, 1991a). After transitioning from pre-swing (terminal double support) to the subsequent swing phase period, the base of support envelope is reduced to the ground contact area of the stance foot (Winter, 1991a). During this period, in the event of the centre of mass moving outside the medial limits, the base of support envelope can increase in size medially by simply lowering the swing foot to make contact with the ground, and effectively regain control the body centre of mass. In an alternate case, where the vertically projected body centre of mass exceeds the lateral limits of the support envelope, the role of the swing foot has a dilemma between two options. One option is to re-align the body centre of mass within the lateral limits of the support envelope by increasing whole body angular momentum of the body (Popovic, Hofmann and Herr, 2004). The second option is the more difficult task of repositioning the swing foot onto the lateral side of the support envelope. This task is more difficult to execute during early swing when the swing foot is posterior to the stance limb. This is also the period of the gait cycle where angular momentum in the frontal plane is reported to be largest (Kaya, Krebs and Riley, 1998).

Angular momentum during walking is almost conserved, even though there are large three-dimensional forces occurring from ground collisions and large angular momentum of body segments (Herr and Popovic, 2008; Neptune and McGowan, 2011). When angular momentum of the whole body approaches zero, posture control is effectively becoming re-established. Balance on the other hand, relates to the relationship between the body centre of mass and the support envelop established by the limits of the centre of pressure. It has been found that regulating whole body angular momentum is a means to control the position and velocity of the body centre of mass (Hof, 2007; Herr and Popovic, 2008). Minimising angular momentum helps maintain balance when the body centre of mass and support envelope are stable. In contrast, when balance conditions are unstable, increasing angular momentum is invoked to re-establish balance conditions, and in-turn upright posture (Hof, 2007; Herr and Popovic, 2008). Hence, balance and posture control are goals determined by the motions of the whole body and not just the stance limb (Popovic et al., 2004; Herr and Popovic, 2008; Hofmann, Popovic and Herr, 2009).

There hasn't been a study investigating whole body angular momentum in elderly populations. There hasn't been an association of whole body angular momentum regulation with falls. In a very small sample size, Simoneau and Krebs (Simoneau and Krebs, 2000) did not find any differences between fallers and non-fallers when measuring whole body angular momentum. The limited research is possibly because determining full body angular momentum requires participants to undergo extensive testing procedures that can provide an appropriate whole body biomechanical model. A simplified study of trunk and whole body angular momentum was compared in a group of healthy older adults and balance impaired elderly (Kaya et al., 1998). This study found that angular momentum in the transverse plane was highest during mid swing, and angular momentum of the whole body and trunk was highest around heel strike in the sagittal and coronal planes. Balance impaired participants had higher peak linear and peak angular momentum in the coronal plane which occurred in the phase between initial stance and mid stance (i.e. mid swing) for both preferred walking speed and when walking speed was matched between groups (Kaya et al., 1998). It is reasonable to theorize that the ability to regulate whole body angular momentum has an association with fallers. From the research so far, there is no indication that this is the case.

The description of balance control during walking indicates that the leg swing has a contributing role for balance regulation and maintaining upright posture. The issues of task co-dependence are currently unknown between leg swing performance and upright balance performance.

2.2.1.2 Toe clearance

As stated in section 2.2.1, maintaining balance and upright posture is among the five primary goals of walking and generally acknowledged as such in most gait studies (Owings and Grabiner, 2004; Brach et al., 2005; Hausdorff, 2005; Moe-Nilssen and Helbostad, 2005; Brach, Studenski, Perera, VanSwearingen and Newman, 2007b; Dingwell, Robb, Troy and Grabiner, 2008; McAndrew, Dingwell and Wilken), and toe clearance is a task that is expected to be subservient to this goal. Maintaining balance and upright posture depends upon an unconstrained swing leg, and this certainly means trip avoidance. Therefore, the task of posture control has been theoretically linked with leg swing cooperation (Mills et al., 2007; Schulz, Lloyd and Lee, 2010).

The vertical path of a ‘toe point’ - located under the distal inferior surface of the shoe-outsole - throughout the swing phase is illustrated in Figure 2.2.1.2.1. The risk of tripping is greatest during the mid-swing phase, where the toe approaches a minimum clearance height. This swing phase event is called the minimum toe clearance (MTC) and it is an hypothesized task goal of walking that requires precise control to avoid the risk of tripping (Winter, 1991a; Karst, Hageman, Jones and Bunner, 1999; Begg et al., 2007; Schulz et al., 2010).

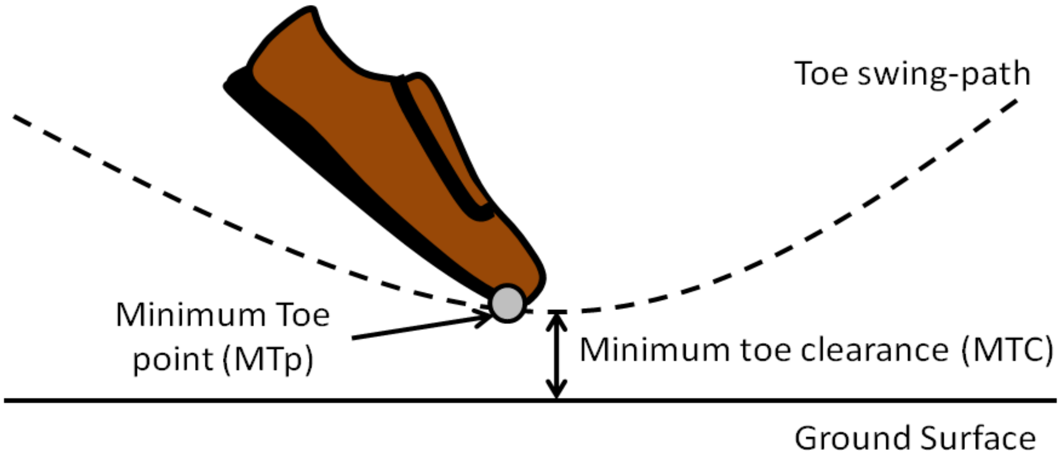


Figure 2.2.1.2.1

The top diagram shows the path of the distal inferior surface of the shoe out-sole (MTp) across the middle portion of the swing phase. The vertical arrow refers to the minimum toe-to-ground clearance height, between the minimum toe point and the ground surface.

Among a range of different experiment designs, the research literature reports that the toe height at MTC is between 10mm-20mm with a standard deviation of 2-6mm (Winter, 1991a; Dingwell, Ulbrecht, Boch, Becker, O'Gorman and Cavanagh, 1999; Karst et al., 1999; Moosabhoy and Gard, 2006; Begg et al., 2007; Mills et al., 2007; Osaki, Kunin, Cohen and Raphan, 2007; Schulz et al., 2010). MTC typically occurs between 45-55% of the swing phase duration (Winter, 1992; Moosabhoy and Gard, 2006; Osaki et al., 2007). Differences in reported MTC and MTC_{time} can be attributed to the different types of methods that describe the toe point location with respect to the foot (Begg et al., 2007; Schulz et al., 2010). These different methods are described further in section 3.5.

Qualitatively, the cause of a trip is credited with the inability to satisfy toe-to-ground clearance height requirements during the swing phase, so it might appear critical that walking with an average clearance height sufficiently high from the ground is required. The research literature involves studies that have explored how MTC height changes due to walking conditions, and how it is altered in different population groups. There have been mixed reports of MTC changes due to altered walking conditions. For example, the reported toe clearance for normal overground conditions are 12.9mm (Winter, 1992), and 10.0mm (Schulz et al., 2010), and with slow and fast walking speed 8.9mm and 12.3mm (Winter, 1991a). Normal treadmill walking conditions report average MTC as 15.6mm (Begg et al., 2007) and 14.9mm (Mills et al., 2007), and this has been reported to reduce by 4.3mm per 1 m/s of increasing walking speed (Miller, Feiveson and Bloomberg, 2009). This contrasts with Winter's (1991a) overground walking study, that indicated there was no MTC change when walking speed increased above a preferred speed. Another study found that walking speed did not alter the maximum swing limb length at the MTC event (Osaki et al., 2007), however, this study did not account for the action of the stance limb length which ultimately contributes to the collective MTC parameter. MTC height appears to be dependent upon dual task conditions during overground walking. For example, the MTC height increases if body is simultaneously involved in a postural control task (e.g. carrying a tray with full glass of water on it), whereas if the dual task is cognitive (e.g. answering questions) the average MTC is reduced (Schulz et al., 2010). Two recent studies found that toe clearance is reduced during treadmill walking when compared to overground walking conditions (Miller et al., 2009; Nagano, Begg, Sparrow and Taylor, 2011). From the above literature, it appears that MTC height is dependent on the type of terrain and the presence of a dual task, while the effect of walking speed is inconclusive.

There have been mixed reports of MTC differences between population groups for steady state walking and when walking conditions have been modified. In most studies comparing between healthy young adults and healthy older adults, the MTC height remains relatively unchanged (Elble, Thomas, Higgins and Colliver, 1991; Winter, 1991a; Mills and Barrett, 2001; Begg et al., 2007; Mills et al., 2007; Khandoker et al., 2008). However, MTC was found to be significantly higher for a group of elderly with a history of trip-related falls when compared to healthy elderly without a history of falls (Khandoker et al., 2008). In a study where vision was occluded in three levels, MTC increased significantly, but only for the largest degree of occlusion (Graci, Elliott and Buckley, 2009). No differences in MTC height were found in population groups who have incurred some sensory loss (e.g., peripheral neuropathy, Dingwell et al., 1999).

Limb end-point control is measured by MTC variability, and the most commonly reported is the standard deviation. This average variability statistic is found to range between 3mm to 7mm for various populations and walking conditions (Winter, 1992; Dingwell et al., 1999; Karst et al., 1999; Moosabhoy and Gard, 2006; Begg et al., 2007). It has been presumed in the research literature that the control of MTC control will be affected by certain types of dysfunction, or age-related co-morbidities that affect the locomotor system. Therefore, variability has also been compared between population groups. Larger variability of foot clearance has been found in healthy older adults compared with young adults when walking on a treadmill at preferred speed (Begg et al., 2007), while other studies from overground walking report no differences between age groups (Winter, 1991a). There was found to be significantly higher MTC variability in a group of elderly fallers when compared to an age-matched healthy group (Khandoker et al., 2008). In a vision occlusion study of young participants, the MTC variability did not change (Graci et al., 2009). In controlled studies investigating the effect of peripheral sensory loss on MTC variability, there were no MTC variability differences found (Dingwell et al., 1999). In older adults walking on a treadmill, as the walking speed increased there was a decrease in MTC height and a small increase in MTC variability (Karst et al., 1999). In contrast, for young adults increasing walking speed on a treadmill, the variability of the swing limb length at MTC remained invariant (Osaki et al., 2007). Further studies are needed to investigate whether the elderly have a poor ability to control MTC when their walking speed increases, because, there are no studies that have compared MTC variability between young and older adults when increasing walking speed.

MTC represents the vertical dimension of foot trajectory and variations in this dimension are relevant for toe-to-ground clearance task. From this implicit viewpoint of previous studies, anterior-posterior or medio-lateral variations in toe position during swing phase can be considered irrelevant for toe-to-ground clearance task based upon the limited reported information. To support this viewpoint, evidence of variability across all three dimensions should confirm that toe-swing variability in the vertical dimension is less than both of anterior-posterior or medio-lateral variability. By contrasting the variability in each direction of three dimensional space, the form of a loco-sensorimotor control policy can be interpreted. For instance, variability is likely to be regulated only in the vertical dimension, because it can be argued that the precise location of the toe in the fore-aft and medial-lateral directions is relatively irrelevant to the task goals of walking.

With respect to control, variations in toe clearance that are exceedingly high from an average intention can be deemed energetically inefficient (Begg et al., 2007). Variations that are exceedingly lower from an average intention will indicate poor control and increase the relative risk of toe-to-ground contact. Optimal performance of toe clearance by the locomotor system is designed to find a control policy that minimises both energy inefficiency and risky behaviour. Studies following this proposed rationale have therefore investigated the average variations of MTC. This leads to one of the more interesting findings, and probably the most important, in the literature investigating the descriptive features of MTC. This is the ubiquitous nature of positive skewness found when characterizing MTC distributions (Begg et al., 2007; Graci et al., 2009; Schulz et al., 2010). The finding of positive skewness suggests that a control strategy is being adopted whereby low clearance values are penalized at a greater extent when compared to higher toe clearance values (Begg et al., 2007). There was a general finding that MTC distributions with low clearance heights were related to low standard deviations or high positive skewness, implicitly suggesting a strategy by people with low clearances to exert relatively higher level of control (Begg et al., 2007). There have not been any reports in the literature which have explored distributions of toe clearance heights along different points of the swing cycle to reveal how the skewness and standard deviation could potentially change in a state-dependent manner. This feature of the toe-swing can potentially reveal control strategies being applied in a task relevant way and demonstrate possible differences between young and older adults.

2.2.2 Toe trajectory

For people to avoid low MTC values and reduce the probability of tripping, having control of the toe trajectory prior to MTC is critical. Altering the trajectory of the toe during the initial region of the swing phase can be considered an important task to help control the vertical state of the toe trajectory as it approaches MTC.

In normal healthy gait, the vertical state of the minimum toe point (MTp) during leg swing follows a characteristic path that includes two local maxima and one minima (Figure 2.2.2.1). There have been a number of different techniques used for quantifying the toe-path during leg swing. Most studies have quantified the toe-to-ground clearance height at the MTC event (Winter, 1992; Karst et al., 1999; Mills and Barrett, 2001; Begg et al., 2007; Graci et al., 2009) and there has been only limited reports of toe-to-ground clearance details at MX1 or MX2 (Redfern and Schumann, 1994; Moosabhoy and Gard, 2006; Osaki et al., 2007).

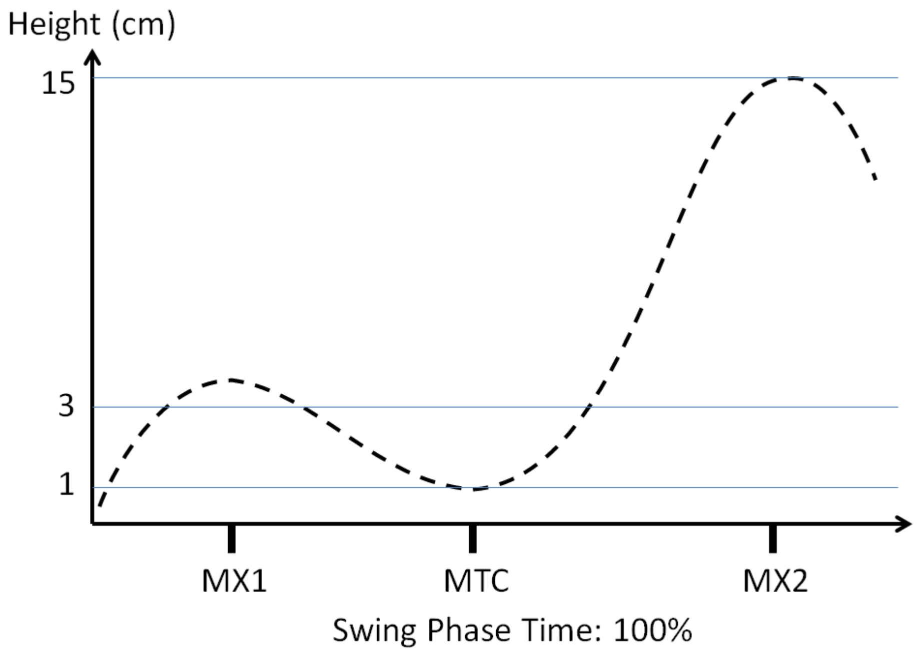


Figure 2.2.2.1.

Profile of the toe trajectory in the vertical dimension across a swing cycle.

2.2.2.1 Redundancy of swing limb and toe trajectory

Coordinating the toe-swing task is complex when considering the redundancy of a two-dimensional, three degrees-of-freedom, swing leg effector system (Figure 2.2.2.1.1). The pelvis, thigh, shank and foot (elements of the stance effector) orientation and length define the position of the foot end point (toe; end-effector). The joint positions of the hip, knee and ankle are assumed to be perpendicular to the page. The toe trajectory is state-dependent with respect to the walking sub-task. Although the planning of the task can begin prior to the onset of lift-off (Goldberg, Ounpuu and Delp, 2003; Osaki et al., 2007), the trajectory expresses task performance which begins at lift-off and is finalised at the minimum toe clearance event (MTC). The toe clearance task is to satisfy two competing constraints of reducing the probability of ground contact and reducing movement energy. The task-space of the swing effector represented in the sagittal plane is defined by three degrees-of-freedom (3-DoF). The end-effector position has 2-DoF. The effector system has many configurations that can equate to the same toe point (end-effector) position. Hence dimensionality is undetermined and the system is redundant. The initial conditions of the task can be represented by the inertial state of the system, the configuration of the system and the position of the proximal hip position. The pelvis position and orientation determines the proximal hip position of the swing effector. The pelvis is a function of stance effector mechanics and the angular momentum of the trunk. The swing limb axis can be represented as a vector from the proximal swing hip joint centre to the end-effector toe point. This vector represents the effector configuration with respect to a fixed reference axis at the proximal swing hip (i.e. somatic reference frame).

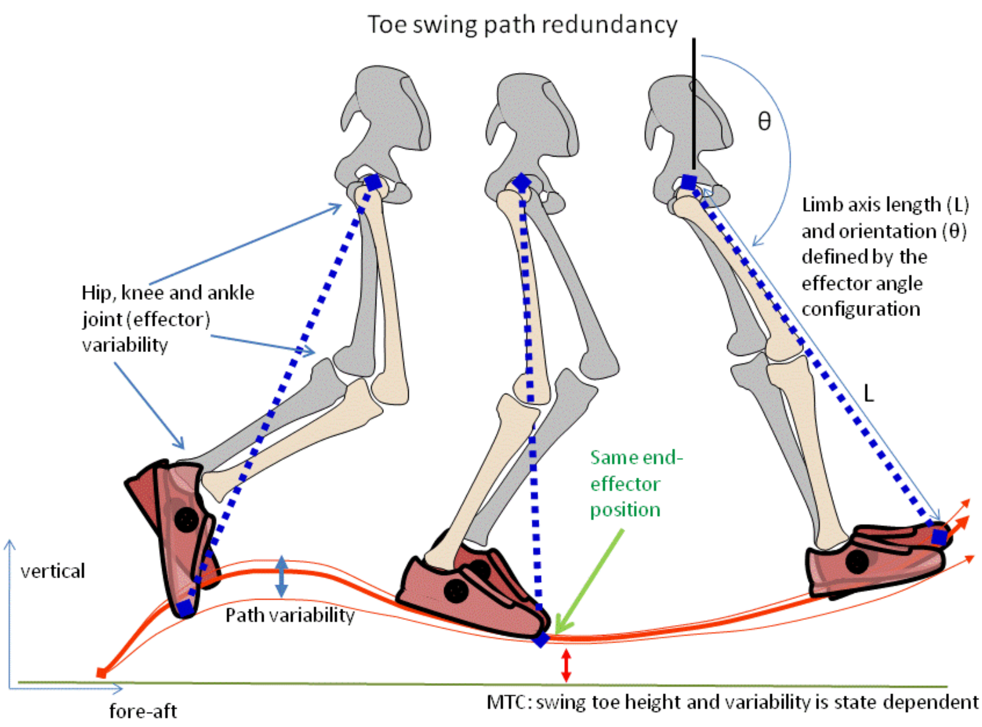


Figure 2.2.2.1.1

Redundancy of toe trajectory in the swing effector system represented in the sagittal plane.

Redundancy means that the same outcome goal can be obtained from many different ways. A fixed toe trajectory may be achieved by exploiting a variety of muscle contractions spanning the joints of the hip, knee and ankle, which was given the term equifinality by Bernstein (Bernstein, 1967). For example, a target position of the toe end point can be achieved through infinitely many joint configurations made up by the swing limb segment articulations. Given the abundant joint configurations and muscle activation and co-activation patterns, the final trajectory pattern of the toe is very stereotypical both within and between individuals (Karst et al., 1999; Osaki et al., 2007) and remains consistent regardless of stability and sensory feedback (Dingwell et al., 1999; Ivanenko, Dominici, Cappellini and Lacquaniti, 2005). These findings indicate the 'equifinality' of the task.

In an open loop control scheme the movement details are planned without requiring information about the state of the body in the environment. An open loop controller issues a series of planned commands for an entire movement task without receiving any feedback about the state of the task (see inverse models in section 2.3.4.1.5). The controller can use inferred states of the body under this scheme to improve the performance of the movement details required for meeting the task goal. From a dynamical systems perspective, a low level open loop controller may be likened to limit cycle behaviour, whereby the influence of destabilizing perturbations is dissipated across consecutive cycles. With open loop control of movement, there is no higher level intervention by the controller. There are many research articles which have addressed the issue of how muscles are grouped to simplify the task of coordinating the swing-leg during gait (Ivanenko, Poppele and Lacquaniti, 2006; Neptune et al., 2009a).

2.2.3 Embodiment of the loco-sensorimotor control system

The last section suggested that the toe clearance task requires integrative performance of the loco-sensorimotor systems. The embodied anatomical substrate provides the context for the way the motor system responds to body-environment conditions. Integrative performance of the loco-sensorimotor systems is based upon body-state perception, task control and movement action. The problem facing movement analysts when evaluating movement performance of dynamic tasks like toe clearance, is that the loco-sensorimotor system has many sources providing perceptual information, addressing multiple task goals, and planning movements from multiple effector components. Physically, there are multiple equivalent solution options available to the loco-sensorimotor system for controlling and coordinating muscles and joints to achieve a goal-directed toe clearance height. As demonstrated in the previous section, there are multiple levels of neural control from different feedback loops augmenting (~10^{12} neural connections in the body) upon the musculature (~10^3 muscles in the body) and joints (~10^2 joints in the body). This gives rise to large motor redundancy, which is the set of all possible kinetic and kinematic options that produce an equivalent final outcome. For example, there is a set of joint configurations capable of placing the limb endpoint in an invariant position. Neural redundancy is a set of possible activation patterns that can act upon multiple muscles which provide a desired joint torque. Within a muscle, there are sets of multiple firing patterns which provide the desired muscle force. The neural pathway to the motor unit has inputs from multiple feedback loops operating across different time scales. In the human movement system, sets of motor equivalence are abundant, and this has traditionally been viewed as a coordination problem to be overcome by the human motor control system.

For a well organised system, this embodied redundancy is considered to be a luxury for movement flexibility, robustness and adaptability (Pfeifer, 2007; Latash, 2008). These three components represent the major embodied features of the loco-sensorimotor control system: afferent sensory processes; central nervous system control; efferent muscle actuator subsystem.

The question of this thesis revolves around effector coordination and end-effector control, or how the toe trajectories can be dynamically altered to fit task dynamics, and whether these details are different between young and older adults. Toe trajectory performance takes place within the context of sensorimotor processes and therefore a review of the biological and mechanistic interpretations of this process will be made in the following parts of this section. At the conclusion of this section there will be links made between mechanistic and biological perspectives of loco-sensorimotor control. This is important for reviewing the position of the current state of research for developing hypotheses about the nature of toe-swing performance relevant to trip avoidance.

2.2.3.1 A complex nonlinear structure

Embodiment is a term used to describe biological details making up the different layers of interacting components between body and environment (Pfeifer, 2007). The position and state of the body as it interacts with the external world is sometimes referred to by biomedical engineers as the 'musculoskeletal plant'. The 'loco-sensorimotor' plant is controlled by the peripheral nerve pathways, the spinal cord, the sub-cortical area of the brain (e.g. cerebellum) and the cortex, making up the embodied neural control system (Duysens, Clarac and Cruse, 2000; Rossignol, Dubuc and Gossard, 2006). The efferent and afferent communication system relays information from the neural control centres to the motor-neurons and motor units attaching to muscle sites, and from sensor organs back to the central command centres (Figure 2.2.3.1.1). There are various sensor modalities, stemming from different sensor organs (e.g. optic, auditory, skin, muscle spindles) via various afferent neural pathways, with independent time delays and synaptic junctions (Duysens et al., 2000; Rossignol et al., 2006). The effects of ageing will alter the structures and function of these embodied components: sensor organs, nerve axons and cell endings, muscle mass, bone mass, skeletal articulations and joint range of motion, and the sub-cortical and cortical regions of the brain (Spirduso, Francis and MacRae, 2005). In effect, ageing causes the loco-sensorimotor control system to adjust to the altered embodied components and find residual resources of redundant capacities to regulate the goals of walking.

The components of interest for the task of walking are related to the sensorimotor system. The 'network anatomy' of this system is described in Figure 2.2.3.1.1. The diagram illustrates the complex structure of the sensorimotor system and a proposed circuitry of feedback loops and information sharing with muscles at different levels of the control 'hierarchy'. There are lower level circuits acting at the muscle-spinal level and higher level circuits acting between the spinal and supra-spinal levels (i.e. lower and higher level controllers). At the intermediate level there are central pattern generators. The intermediate level has direct access to the sensory inputs of the peripheral aspects of the body. There are a variety of loops at the low level that allow for quick motor response based upon sensory information, however these actions are less refined and less tuned to the task requirements (Kurtzer, Pruszynski and Scott, 2008; Pruszynski, Kurtzer and Scott, 2008; Kurtzer, Pruszynski and Scott, 2009). The higher level interactions are more goal-directed. The complexity of the system is represented by the neural anatomy of the body and the stochastic nature of the world.

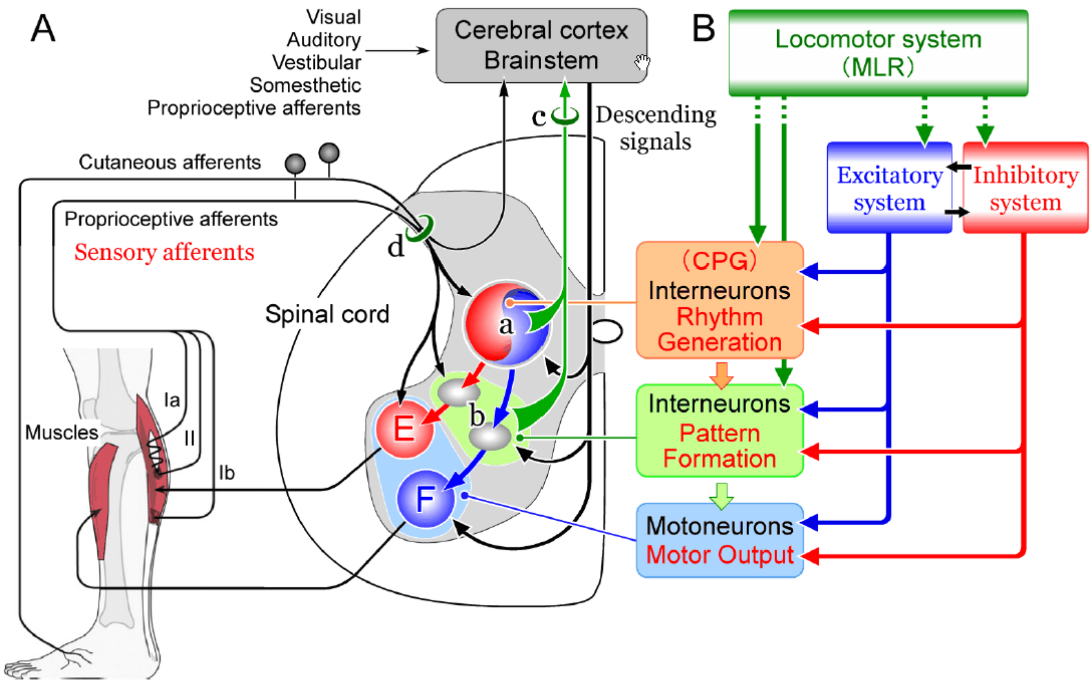


Figure 2.2.3.1.1

Different levels of looped neurophysiologic pathways making up the entire efferent (descending) and afferent loco-sensorimotor system. See text for details. The diagram was taken from Takakusaki (2008).

The three blue arrows represent different afferent return loops arriving at different levels of the central nervous system. The longer loop involves longer time delays. The brain stem interprets the 'analogue-type' afferent signals in a lower dimensional 'digital' representation. These afferent signals arrive with added noise (Bays and Wolpert, 2007; Faisal et al., 2008). It has been proposed that the existence of the central pattern generators (CPGs) are located in the spinal cord and have strong association with mid-level control (e.g. cerebellum). The flexor or extensor muscle actuators of the CPG, interneurons, motorneurons can be mediated by higher and lower level control. These components are integrating centres for incoming higher level control input and sensory information inputs direct from proprioceptive and cutaneous afferents. The final pathway of generating muscle flexion and extension force is derived through this general process. The efferent muscle commands are subjected to additive noise which is dependent upon several factors, but generally associated with the physiology and orderly recruitment of the motor-unit pool (Jones, Hamilton and Wolpert, 2002; Todorov and Jordan, 2002; Faisal et al., 2008). The neural system operates with noise and time delays (Faisal et al., 2008).

The structures of the embodied system encompass many diverse components and subcomponents, which act and interact dynamically with reciprocal interests. The outcome of these actions is an expression of coordination at various levels: neural, muscle, skeletal. It is a matter of debate between biological and mechanistic views about how these actions and interactions are controlled into a state of coordination (Loeb, Brown and Cheng, 1999; Kelso, 2004; Scott, 2004; Latash, 2008; Torre and Wagenmakers, 2009). The effect of muscle activation on the limb mechanics has been successfully modelled and this highlights the non-linear nature of movement output from neural input because of the elastic properties within muscle (Zajac, Neptune and Kautz, 2002; Zajac, Neptune and Kautz, 2003).

2.2.3.2 Components of the loco-sensorimotor system

Biological systems are recognized as containing various components (or agents) that are dynamically coupled between physical sub-systems both internal and external to the organism, a property also defined as embodiment (Pfeifer, 2007). When placed in an environment, the biological components become embodied with their environment as components between body and environment mutually perturb each other and give rise to dynamical interactions. The dynamical relationship between a loco-sensorimotor system and the environment defines embodied features specific to walking. The morphology of biological systems has evolved through a process of embodiment, and the evolution of human beings and bipedal walking is an example of such a process.

One of the mysterious areas of understanding loco-sensorimotor system is associated with the problem of deciphering between the interface of active and passive control of walking. Arguably, the precise motor control task of walking could be performed with limited neuro-mechanical actuation and rely mostly upon anthropometric properties. For example, biped robots can walk in a stable manner down a shallow slope without any control input or motor actuation (McGeer, 1993). However, with a small amount of control, the stability limits of passive walking models are increased (Verdaasdonk, Koopman and van der Helm, 2009). When biped models utilize their passive dynamic structures they not only substantially reduce their energy usage (Collins et al., 2005; Verdaasdonk et al., 2009), but they appear to move with greater dexterity (Collins et al., 2005). Decerebrated cats, i.e. without supra-spinal control, can also manage to walk with dexterity using only sensory reflex loops (Rossignol et al., 2006). Both examples highlight the capabilities of embodied passive-type features at different biological levels within the loco-sensorimotor system. The addition of active control from supra-spinal control input brings about a further complexity to embodiment and the problem becomes even less clear for determining when passive control requires active intervention.

This section examines the components of the embodied loco-sensorimotor system for the purpose of exploring how this concept can relate to controlling toe clearance during walking. There are three general areas that the loco-sensorimotor controller will exploit when controlling the states of the toe trajectory. These three control areas include aspects of passive and active control. The three areas are: predictive control (active); reactive control (reflex-like and therefore mostly passive); and biomechanical control (derived from both active and passive means). These three areas of control are affected by ageing of the systems described below.

2.2.3.2.1 Limb Biomechanics

Limb mechanics during walking can be an outcome of both passive and active muscle contributions (Neptune, Sasaki and Kautz, 2008). The limbs have natural passive features of inertial dynamics and joint structures that can be exploited by the loco-sensorimotor controller to influence movement behaviour with energy efficiency. Additionally, the limbs can be actuated by neuromuscular activity. The integration of both processes are utilised by the loco-sensorimotor control system for planning and execution of movement. During walking, most of the muscle activity occurs during the beginning and end of the stance and swing phases (Ivanenko, Cappellini, Dominici, Poppele and Lacquaniti, 2007; Neptune, McGowan and Kautz, 2009b).

The actions for movement require neural commands sent to the musculature so that the muscle can produce force on the skeletal segments which articulate at differently orientated joint surfaces and structures with other segments. The forces and moment arms create joint torques, which give joints their stability or movement. Inertial motion of the limbs is influenced by segment anthropometric properties. The elastic features of muscles incur non-linear outcomes for the joint torques. The level of muscle (co)activation can alter the damping property during segment collisions which influence muscle firing and sensory measurements. The combination of uni-articular and bi-articular muscles influences the way energy is transferred from one body segment to the next. The process of ageing can affect all of the above processes.

Changes to the internal dynamics of the muscle-skeletal system usually happen slowly, such as reduced muscle mass and bone density, and the loco-sensorimotor control system has time to adapt. Furthermore, the level of noise during generation of muscle activity, due to neural degradation and synaptic degeneration, usually occurs slowly.

Limb movements occur from muscle forces which cause joint torques. A kinematic chain of bone segments moves as a function of the joint forces at the articulations and the direction of the internal forces created by the muscles. Bone segments move according to Newton's angular law of acceleration: $T = I\ddot{\theta}$, where T, is the net torque, I, is the moment of inertia of the segment, and $\ddot{\theta}$, is the angular acceleration of the segment. In multi-segment systems with many degrees of freedom and the addition of external forces, like ground reaction forces, this version becomes more complex and the direct relationship between torque and motion now is dependent upon segment lengths, segment masses, and segment centre of mass locations (Winter, 2005). Forward dynamics using multi-segmental biomechanical models and simulations show that energy flows between segments, and muscle actuations can affect motion at joints not spanned by that particular muscle (Zajac et al., 2002; Zajac et al., 2003).

Muscle modelling studies have investigated the use of passive muscle-tendon properties of the limbs to swing the leg during walking. Selles and colleagues (Selles, Bussmann, Wagenaar and Stam, 2001) investigated four models of the swing phase to assess ballistic passive and active processes and they determined that all models underestimated the swing length and time data. This is in agreement with Piazza and Delp (Piazza and Delp, 1996) who also found that modelling the swing phase demonstrated that it is not a purely passive process. There is a suggestion that swing phase trajectory involves a greater contribution of active muscle work at slower walking speeds (Selles et al., 2001). Studies are generally in agreement that limb mechanics during the push-off phase in pre-swing are providing the necessary conditions for minimising active control of leg swing (Goldberg et al., 2003; Anderson, Goldberg, Pandy and Delp, 2004; Reinbolt, Fox, Arnold, Ounpuu and Delp, 2008). In summary, there is a degree of muscle activation during the swing phase which indicates active control, while the amount seems to be determined by energy restoring conditions during pre-swing.

2.2.3.2.2 Altered changes in brain structure and function due to healthy ageing

Research investigations into how the sensorimotor system changes during the healthy ageing process have focused mostly on the peripheral changes. However, impaired movement function is associated with reduced structure and function of the brain. 'Ageing well' simply means that any decline is a system is due solely to the singular process of ageing. Alternatively, pathological ageing is not due to an age-associated change per-se, but rather the additional affects of disease, functional impairments, and ill health. While the term 'ageing well' is associated with relatively few co-morbidities and functional limitations, the collective process of systems 'ageing well' can place limits on sensorimotor function and physical performance, however the ultimate affect on lifestyle and independent living is relatively minor. In contrast, pathological ageing affects movement function to a greater extent and in-turn the consequential restrictions on lifestyle and independent living will reflect this magnitude.

It is proposed that the loco-sensorimotor control system is structured according to a control hierarchy, such that a higher level controller mediates a lower level controller (Loeb et al., 1999). This structure will be explained in more detail in the following sections, but for now the broad details will provide context to this initial section. Briefly, the proposed higher level controller exists in the supra-spinal neural network structures and the lower level controller resides in the neural system structures of the mid-brain, central pattern generators, and reflex networks (for more detail see Figure 2.2.3.2.2.2). From an assumption that the controller of a healthy loco-sensorimotor system exploits passive control, this in-turn suggests that the low level controller leads most of the neural processing effort. It can also be expected that this same argument applies to healthy older adults, such that they also prefer to rely upon the neural structures of the low level control system.

While the ageing process in the brain can occur without pathology, there are a number of subtle changes occurring within the brain that alters the ability to perform movements optimally. Executive function is associated with the neural structures of the forebrain area and this brain area is responsible for monitoring the lower-level controller of movements. In healthy older adults with no known cognitive dysfunction, their decline in executive function is a strong prospective predictor of future falls (Herman, Mirelman, Giladi, Schweiger and Hausdorff, 2010). Other studies also suggest that occasional falls in older adulthood are more likely to be due to subtle changes of executive control, whereas older adults who suffer multiple falls are related to more advanced decline that is evident in cognitive dysfunction (Anstey, Wood, Kerr, Caldwell and Lord, 2009).

Measures signalling that the sensorimotor system has aged well, including general measures of cognitive function, will not necessarily detect healthy brain function. Healthy ageing will still demonstrate affects to the size of the white and grey matter as indicated by decreases in volume, where the prefrontal and parietal cortex is most affected (Seidler, Bernard, Burutolu, Fling, Gordon, Gwin, Kwak and Lipps, 2010). The cerebellum also demonstrates a significant reduction in healthy older adults (Raz, Lindenberger, Rodrigue, Kennedy, Head, Williamson, Dahle, Gerstorf and Acker, 2005). Rosano et al. (2008) tested a cohort of 220 older adults and demonstrated a strong positive relationship between grey matter volume and gait performance (i.e. measures of step length and double support time). Healthy older adults can perform complex motor tasks as competently as young adults, but there is evidence that this requires significantly greater activity in the following brain regions: bilateral anterior lobe of cerebellum, premotor cortex, parietal cortex, prefrontal cortex and anterior cortex (Wu and Hallet, 2005). In summary, older adults engage in greater brain activation to compensate for areas which have atrophied or become impaired (Seidler et al., 2010) (Figure 2.2.3.2.2.1). This indicates reliance on greater cognitive processing when carrying out movement tasks, but this comes from a reduced reservoir of processing capacity, and therefore creates a ‘bottleneck’ in the prefrontal cortex when performing movement tasks which require intervention of a high-level controller.

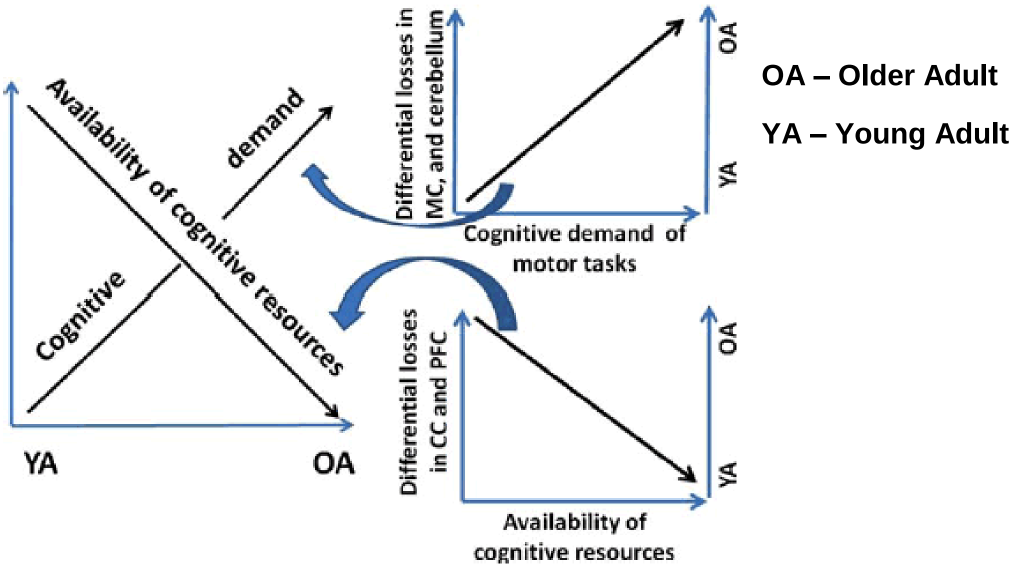


Figure 2.2.3.2.2.1

Supply and demand – changes in the older adult brain (from Seidler et al. 2010). Cognitive demand increases in older adults as the supply of cognitive resources diminishes (i.e. attention, working memory, visual spatial processing, other brain engagements when performing motor control tasks). Abbreviations: CC (cortex); PFC (prefrontal cortex); MC (motor cortex).

Evidence suggests that walking patterns are mediated by the basal ganglia whose role is to influence the afferent processes at the mid-brain (brainstem) level (Figure 2.2.3.1.1.1). The basal ganglia has the mediator role between the volitional and emotional regions of the forebrain. These two regions could also be referred to as active and autonomous (i.e. passive) control processes. The cerebellum has the role of integrating, contrasting and distributing all the information. In combination, these general regions of the forebrain determine the level of control-influence that forge the behaviour of the limb trajectories during walking.

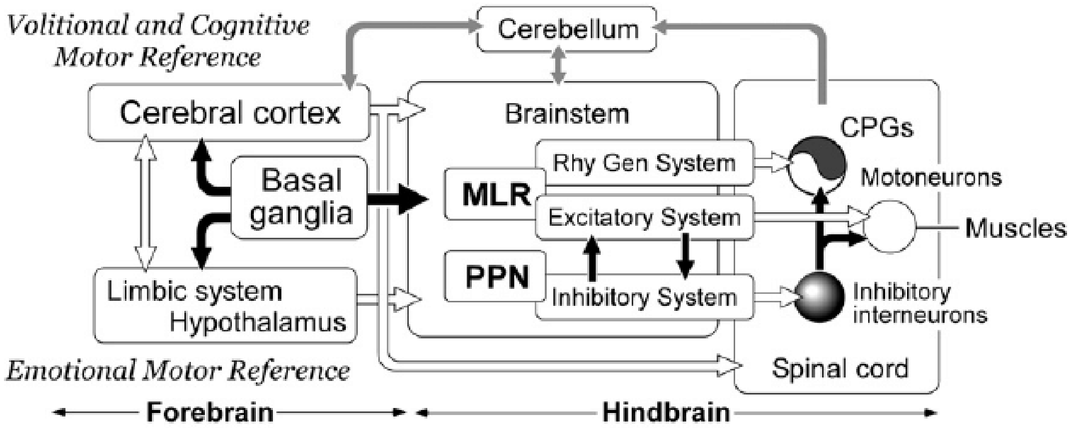


Figure 2.2.3.1.1.1

A close inspection of the mediating role of the basal ganglia for influencing the brainstem behaviour and in-turn the spinal cord neurons during locomotion. In the brainstem, MLR represents the midbrain locomotor region and PPN represents the pedunculopontine nucleus (muscle inhibitory region). Taken from Takakasuki (2008).

2.2.3.2.3 Sensorimotor noise

When a goal-directed movement task is performed, the precision of the outcome is largely dependent upon 'noise' in both afferent and efferent neuro-physiological systems (Harris and Wolpert, 1998; van Beers, Haggard and Wolpert, 2004; Faisal et al., 2008; Orban and Wolpert, 2011). During walking, when older adults perform the toe clearance task, the precision of control will also be subject to noise in these systems. The integration of sensory information and a motor action plan will become more challenging for older adults because sensory and motor systems degrade with age (Roos, Rice and Vandervoort, 1997; Spirduso et al., 2005; Christou and Tracy, 2006; Callisaya, Blizzard, Schmidt, McGinley, Lord and Srikanth, 2009; Christou and Enoka, 2011) and the likely outcome is an augmentation of 'noise'. Decision making mediates the integrated process of sensory inference and motor commands through the central nervous system. The movement plan mediated by decision making is likely to be altered with the increase in system noise (Nagengast, Braun and Wolpert, 2010). The increase in cognitive load described in the previous section is possibly coupled with the deterioration of the sensorimotor system. As system noise increases with ageing processes, the elderly are faced with greater uncertainty when controlling precise movements at low muscle intensities. These conditions are characteristic of the toe clearance task. This section briefly reviews the age-related changes to the afferent and efferent systems which contribute to toe clearance performance.

Efferent System

Older adults display many age-related changes to the motor-neural pathways, including extinction of cortical motor neurons and spinal motor neurons, and there is a general slowing of the transmission along the corticospinal and reflex pathways (Roos et al., 1997). The changes to motor unit size and discharge functioning is also altered due to ageing. When spinal motor neurons deteriorate, the motor units adapt by innervating with more muscle fibres. The outcome is larger but fewer motor units (Roos et al., 1997). The discharge rate of motor units in elderly is found to be more variable (Christou, Shinohara and Enoka, 2003) and the variability of force at low maximum voluntary contraction levels is more variable than younger adults (Christou et al., 2003; Enoka, Christou, Hunter, Kornatz, Semmler, Taylor and Tracy, 2003).

Errors in goal-directed movements are dependent upon the intensity of the motor command signals (Harris and Wolpert, 1998). The noise in the efferent system is dependent upon the number and magnitude of the motor units connecting with the muscle. In healthy young adult participants, Hamilton et al. (2004) found that command signal noise contributed up to 6% of the movement variance. It is likely that older adults contribute much more due to age-related changes. Indeed, the physiological changes described above suggest that noise in the elderly efferent system will contribute more towards movement variance.

The large body of research investigating the link between motor commands, muscle forces and movement error in elderly populations comes mostly from a research group led by Evangelos Christou and Roger Enoka. The major relevant findings from their work is summarised below. At low muscle contraction intensities (loads within 10-15% of maximum voluntary contractions), older adults have amplified variations when controlling the force and timing of muscle contractions (Christou and Tracy, 2006; Christou and Enoka, 2011). Typically, the outcome is greater movement error (Christou and Enoka, 2011). In contrast to determining between age differences in studies of single joint movement tasks and investigating a single muscle, when coordinating tasks involving a multi-segment limb, where multiple muscles span across many segments, older adults demonstrate larger motor variability (Christou and Tracy, 2006). Recently, Christou et al. (2011) addressed the limitations of previous research using single muscle designs which have brought about mixed hypotheses related to motor command variations found in older adults. In line with this perspective, a refined study investigated co-antagonist muscles between young and older adults (Christou and Enoka, 2011).The outcome of the recent study demonstrated that older adults showed less accuracy of targeted movements performed at sub-maximal load. The difference in accurate performance was attributed to two independent sources of executing noisy motor commands and inflexible motor planning (e.g. invariant activation patterns of the antagonistic muscle).

The implications of the findings for older adults performing the toe clearance task during walking are suggesting that older adults will exhibit larger errors in toe positioning. The source of the errors can potentially come from noisy motor commands or an ineffective plan to coordinate an appropriate muscle strategy. The above studies by Christou et al. were careful not to invoke sensory factors, and this introduces a third potential source of error when performing targeted movements.

Afferent System

The afferent system represents sensory information supplied to the central nervous system about the state of the body with respect to the environment. These feedback signals are noisy and delayed, displaying various levels of precision dimensionality depending upon the context and origin of the input source, and arriving after the event has occurred. Scott (2004) outlined context-dependent differences in the quality of feedback information existing between and within the general sources of proprioceptive, visual and vestibular systems. At the peripheral level, the signals are analogue with additive noise and are being processed on a continuous basis (Faisal et al., 2008). Therefore, the state of the body and the environment is only as precise as the inferred signal details by the central nervous system (Wolpert, 2007). To compensate for feedback delays, an internal model is created, but this is based upon planned outgoing motor commands that are themselves noisy. Therefore, sensory feedback displays varying degrees of uncertainty which are context dependent.

The effects of 'healthy ageing' on the sensory system show a strong association with gait variability (Callisaya et al., 2009; Callisaya, Blizzard, McGinley, Schmidt and Srikanth, 2010; Moe-Nilssen et al., 2010). Noise in the sensory system represents greater uncertainty for movement planning because the state of the body and environment becomes more ambiguous and this creates errors for goal-directed movements (Kording and Wolpert, 2004; Orban and Wolpert, 2011). The increase in sensory noise in healthy older adult populations can be expected to contribute towards movement errors. It is well documented that sensory systems of vision, vestibular functioning and proprioception deteriorates with the process of ageing well (Spirduso et al., 2005; Lord et al., 2007).

2.2.3.2.4 Low level control: spinal cord and short-latency reflexes

Correcting movement errors based upon immediate feedback from the sensory system is limited due to transmission timing delays. Figure 2.2.3.1.1 described the different sensory feedback loops of the afferent system. Time delays in feedback are solved by the short-latency reflex which comes from sensitive muscle spindles which respond to minor muscle length changes. During walking, the short-latency response has a functional role to provide immediate joint extension to the support limb.

There are state- and phase- dependent spinal reflexes from cutaneous and proprioceptive inputs (Rossignol et al., 2006). The change in afferent information is phase dependent (i.e. stance/swing) whereby cutaneous information is primarily a limb loading information source and proprioception is primarily a limb position information source (Rossignol et al., 2006).

During movement the brain can elect to open various afferent (supraspinal and spinal derived) channels to the appropriate motor-neurons to influence correction to walking sub-tasks without interrupting progression (Rossignol et al., 2006).

2.2.3.2.5 Mid-high level control: primary motor cortex and long-latency reflexes

The volitional motor control system is predominantly structured around the primary motor cortex (M1), and this coordinates through neural activity with other areas of the central nervous system. The primary motor cortex (M1) has the ability to control all movement details, such as distance, direction, velocity, accuracy, force, and impedance. Because of the time delays involved with afferent signals reaching M1, regular feedback loops arriving at M1 may be too slow to make required adjustments to correct movement errors from perturbations. Sophisticated reflex loops, or long-latency reflex loops R2 and R3 (50-100 ms), can learn to replicate the approximate behaviour of M1. These long latency rapid motor responses demonstrate learning contexts of the task and the goals of the task by altering their tuning to obtain fast corrective motor patterns (Pruszynski et al., 2008).

The stretch reflex, or short-latency reflex loop cannot be modified by voluntary intent (Pruszynski et al., 2008), however, the short latency rapid response loops are useful to initiate the less specific aspects of a movement to prepare the movement so that less work is required by the longer latency motor responses which arrive later. With limited training the longer latency motor response loops are able to be tuned in a sophisticated way to match the properties of the task and therefore provide flexibility and precision when they arrive (Pruszynski et al., 2008). With decreased predictability of movement perturbations, R2/R3 responses are delayed, indicating that they rely upon current sensory information and prior history of perturbations. The R2/R3 motor response loops have been found to contain an internal model of the limb dynamics (Kurtzer et al., 2008). This demonstrates that task specific fast motor responses can be achieved without the sophistication of processing from M1.

2.2.3.2.6 Central pattern generators and inter-neuronal pathways

One way the body addresses the problem of coordinating embodied elements during rhythmic movements, like walking, comes from the hypothesized role of neural oscillator components which form central pattern generators. Some researchers argue that central pattern generators solve the coordination problem at the low hierarchical level of the CPG network because within their structures they are capable of generating rhythmic movements without any input (Rossignol et al., 2006). It is suggested that CPG units produce low dimensional motor primitives (Tresch, Saltiel, d'Avella and Bizzi, 2002; Bizzi, Cheung, d'Avella, Saltiel and Tresch, 2008) and a simple low dimensional efferent input via higher level control can combine these motor primitives to create rich movement patterns (Ijspeert, 2008).

The existence of central pattern generators in humans (Dimitrijevic, Gerasimenko and Pinter, 1998) is supported by evidence from studies on vertebrates (Rossignol et al., 2006) which share similar central nervous system properties with humans. Many researchers share a strong belief of CPGs existing in humans (Kuo, 2002b; Choi and Bastian, 2007; Ijspeert, 2008). The finding that humans and animals can learn various patterns of gait indicates that CPG circuitry is likely to be adaptable (Rossignol, 2000; Choi and Bastian, 2007).

It is likely that CPG behaviour is expressed in the form of limit cycle patterns of walking trajectories (Kuo, 2002b). Limit cycles are patterns which have an attraction to regions of stability (discussed in more detail in section 2.4). Under this assumption, perturbations in walking patterns are dissipated by ‘passive’ or ‘low-level’ control, rather than requiring the regulation of ‘active’ or ‘higher-level’ control (e.g. input from primary motor cortex). In this case, small perturbations will be allowed to persist until damped by low-level ‘reflex-like’ CPG control, and supported by the natural inertial and compliance properties of the musculoskeletal system. The predominant role of low level autonomous mechanisms underscore many theoretical beliefs about how walking is controlled (Ashkenazy, Hausdorff, Ivanov and Stanley, 2002; Collins et al., 2005; West and Latka, 2005; Kuo, 2007).

2.2.4 Summary

This section has introduced the issues which surround the operation and integration of the components which make up the loco-sensorimotor system. Evaluating the performance of the toe clearance task is a considerable challenge and approaches about how this can be achieved will be discussed in the following sections.

Falls occur predominantly from trips during the act of walking. However, research so far has not demonstrated empirical evidence to indicate where to attribute the error in gait performance. This is due to two main reasons, first the parameters being investigated in most large scale falls studies have not had theoretical link to tripping mechanics, and second, the limited studies which have investigated parameters of toe-swing control have not validated the parameters with trip frequency. To support the development of research protocols for investigating these links further, general knowledge about the processes governing the task performance of the toe-swing is required. For instance, avoiding a trip occurrence requires a three-stage response process of locomotor planning and execution: sensory perception of possible threat; selection of an appropriate corrective response; effective response execution. A trip occurrence can involve poor negotiation of a seen obstacle or it can involve the swinging toe being obstructed by an undetected obstacle. In both cases, the sensorimotor system undergoes the three-stage process. What isn't known is when a trip occurs, which aspect of the effector system is responsible for the error? The exploration of the problem requires consideration of how the sensory and motor systems are integrated. This is indicative of the need to view the problem from a more holistic perspective. This thesis will apply analysis methods which explore the underlying processes contributing to the toe-swing task. The outcomes of these analyses will provide implicit information about the locomotor strategy for performing the toe-swing task.

2.3 Theories of how coordinated and controlled movement trajectories emerge from the sensorimotor system

There is no ‘cutting-edge’ robot, or simulated model of bipedal walking, which demonstrates dextrous movement through uneven terrain in a naturally stable manner. There has been no biological or sensory-motor research which has detailed cause-effect mapping and principles about how goal-directed movements are controlled through the complex system. Movement control is something that seems to be derived from multiple interacting components that seem to have an indirect and non-linear influence on the task outcome. Various theories have been proposed as to how control is possible in such a system.

2.3.1 The problem of redundancy in the loco-sensorimotor system

Large variations at the joint or muscle level of an effector does not imply that the goal-state end-effector variance will be proportionately large because of the existence of motor redundancy (Bernstein, 1967). Similarly, large variance inherent within both the stance and swing effector does not imply that the final toe position will have proportionate variability. There has been observations that the average size of variations of the joints in the lower limbs is comparatively larger compared to the variations of vertical toe position at MTC (Winter, 1992; Mills and Barrett, 2001). Motor equivalence, or equifinality, describes the many different motor configurations leading to an invariant outcome. Bernstein (1967) demonstrated how the goal-directed end-effector position repeats reliably when elements of the effector exhibit much greater movement variability. Recent views of ‘Bernstein’s coordination problem’ by many motor control researchers agree that the control system maximises performance by exploiting task-relevant structure in the degrees of freedom (Scholz and Schoner, 1999; Todorov and Jordan, 2002). While there has been an attempt to describe the sensitivity of the toe position to isolated angular changes of the lower limbs (Winter, 1992; Moosabhoy and Gard, 2006), there are no known studies examining how embodied redundancy of the system influences toe clearance control.

The following section will review two general theories of how complex systems perform goal-directed tasks by controlling redundancy.

2.3.1.1 Hierarchical control verses self-organisation

There are two philosophically different theories related to how human movements are coordinated and controlled. The difference is in the way they treat the control process which allows a system to reach a goal state. One theory suggests an indirect emergent form of control, while the other theory is based upon hierarchical sources that impose a form of direct control. These are now briefly outlined below.

Dynamic systems theory (DST) takes the view that coordination of a multi-degree-of-freedom system emerges from self-organised dynamic couplings among the multi-level system components (Kelso, 1995; Temprado, 2004). This field of thought is considered to be most viable as a biological substrate (Turvey, 2007; Latash, 2010a). The DST philosophy of the control process does not presume direct control, but rather it is hypothesised that volitional intention is dispersed through the system as chain of energy interconnecting the coupling relations among components within the system, which shapes coordination and the reaching of a goal state (Kelso, 2004).

In contrast, optimal feedback control theory (OFCT) views the emergence of control and coordination as an outcome from an optimised solution which is directly mediated by a 'hierarchical control structure' (Loeb et al., 1999; Todorov and Jordan, 2002). The benefit of applying OFCT models has been their ability to explicitly investigate how different sensorimotor sources can influence the goal-directed motor actions (Scott, 2004; Shadmehr and Krakauer, 2008). Both OFCT and DST approaches are discussed below.

2.3.1.1.1 Direct hierarchical control

Li et al. proposed a two level hierarchy of control in their optimal feedback control model (Figure 2.3.1.1.1.1) and this hierarchy is supported by other researchers (Loeb et al., 1999; Scott, 2004). Figure 2.3.1.1.1.1 below illustrates two levels: 1) low level controller which leads by monitoring progress and guiding the biomechanical plant towards the task goals; and 2) high level controller operating in the background to support with corrections. There is a theory which suggests that the low level controller acts on the high dimensionality of the state estimate of the body in the environment, x, and forwards a low dimensional copy, y, to the high level controller. In this case the higher levels of the central nervous system mitigate background vigilance on the lower level of the central nervous system (Li et al., 2005). Based upon the neurophysiological model presented in section 2.3.1, it is likely that a two-level hierarchy (as illustrated in Figure 2.3.1.1.1.1) could have additional levels associated with the rhythmic networks of the CPG. The term control hierarchy has also been applied by Laquaniti (1989), who used this structure to imply that different gait parameters reveal different levels of the control hierarchy, such that foot trajectory parameters were related to the highest level of control, and joint angles and muscle activities were related to lower levels of the control hierarchy.

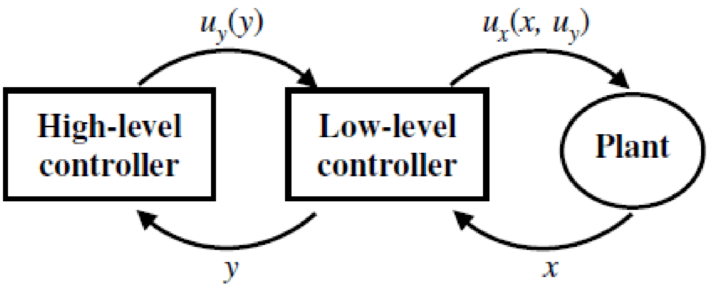


Figure 2.3.1.1.1.1

Hierarchical order of a control structure coping with the redundancy problem (taken from Li, Todorov and Pan (2005)). A low level controller is at the spinal level, while the high level is supra-spinal level.

2.3.1.1.2 Self-organisation and dynamic systems theory

The hierarchical control structure presented above is different to the DST proposal of emergent order and stability created from self-organised processes occurring within a system. A complex system is embodied with diverse components, and in DST language of biological systems these components are described as agents that represent individual micro-systems. Agents respond energetically to the dynamic conditions evolving in their environment, as well as the local interactions of immediate neighbouring agents. Through the interconnected complex system, agents influence the local and therefore the global state of the system. The philosophy of self-organisation comes from the view that agents have a natural tendency compelling them towards preferred stable states of coordination. This concept has been explored experimentally through bimanual oscillations of the hands (Haken et al., 1985; Schoner, Haken and Kelso, 1986). Figure 2.3.1.1.2.1 illustrates an example of the Haken-Kelso-Bunz model (Haken et al., 1985) of coordination stability, as a function of the relative phase oscillations between agents (between -180 to 180 degrees). The graph on the left shows agents oscillating at their 'preferred' frequency, and stability at two regions: 0 degrees relative phase; and ±180 degrees relative phase. At 0 degrees relative phase, stability is maximal as indicated by the size of the 'trough' (i.e. energy potential). For this simple collective system of two agents, the potential required to drive the system out of this coordination state will require relatively more energy at a relative phase of 0 degrees, in comparison to coordination states at regions of ±180 degrees. This example demonstrates that for this system, for maintaining stable relations between these agents, the 'in-phase' oscillations will be preferred over 'anti-phase' oscillations. When the oscillating frequency increases beyond the preferred rate of 1Hz, coordination becomes unstable for 'anti-phase' relations.

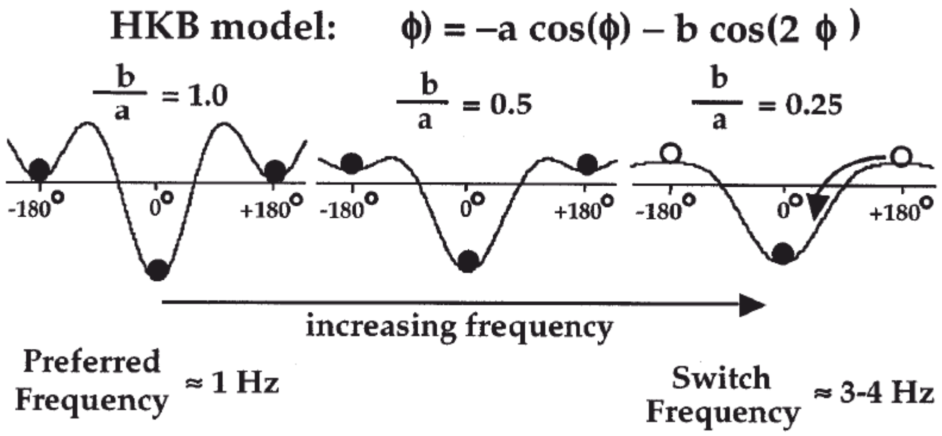


Figure 2.3.1.1.2.1

The Haken-Bunz-Kelso model of agent coordination stability as a function of relative-phase and oscillation frequency (Haken, Kelso and Bunz, 1985).

Agents are determined by the conditions within their environment. The environment conditions by which an agent is acted upon is created by alternate groups of assembled agents from potentially remote regions in the system. Agents are also directly and indirectly influenced by other agents resulting in a nonlinear cause-effect process. Complex systems are open, being subjected to flows of multiple fluctuating influences perturbing the state of the agents within the system. Positive feedback in complex systems is a term to describe how small perturbations become amplified as they propagate throughout the system, creating a global effect. The alternative is negative feedback, where small perturbations are damped and fail to have any ultimate effect on the system's preferred equilibrium state. Both positive and negative feedback variants underscores the unpredictable state of complex systems and suggest that complete control of complex systems is near impossible because no internal or external agent plays a critical role for gaining complete control. So, how are complex systems controlled given that coordination of agents appears chaotic?

For open complex systems that are self-organised, the term control is often replaced by the term 'order' which is a form of functional coordination. Through the process of self-organisation, order within the system spontaneously emerges from the collective local coupling interactions of agents. In a sufficiently large system, the inclusion or exclusion of agents does not deter from the spontaneous order derived from the collective assemblage. A major assumption, seemingly contradictory to the DST perspective, is that order is not arbitrary, but it is goal-directed. Heylighen (2008) proposes that order emerges from agents acting so that they are always attempting to close-in on their 'preferred' goal state. This process requires some compromise between agents because not all agents (of independent motives) can maximally achieve their goal. The collective multi-agent solution becomes a compromise to reduce destabilising tension between competing agents. The assumption that agents can develop goal-directed actions suggests that DST does share 'OFCT-like' views on coordination. Recently there have been further views of self-organised multi-agent systems proposing that agents follow independent goal-directed actions (Turvey and Fonseca, 2009).

Assembled configurations of mutual adapting local agents, through positive feedback, result in preferred states of global coordination and this suggests the emergence of a functional synergy (a coherent stable state). The interacting agents are proposed to assemble far quicker into a coherent 'synergy' when they share like qualities. This is because like-agents share in a 'niche' solution which is mutually beneficial. Unlike-agents assemble less readily as they explore each other for a mutual fit. This assumption of how quickly synergies can form has relevance to age-related deteriorating systems like those of the elderly (Latash, 2008).

The behaviour of the complex multi-agent system fluctuates and explores its state-space searching for a stable region. Self-organisation means that the system has settled upon a preferred attractor (e.g. similar to the 'trough' in Figure 2.3.1.1.2.1). Complex systems can have multiple attractor states for which it can enter from different initial conditions. It can be likened to an analogy of a multi-lane highway, where the system explores options to find the most suitable stable 'lane' that satisfies the fluctuating multi-agent 'needs'. The efficiency of self-organisation is enhanced by increasing the initial variations so that it visits different states of its state-space before settling into an attractor state. Once the system finds an attractor it has constrained the agents into a collective utility. Small perturbations can force the system out of a stable attractor, but self-organised systems are flexible by either returning the same attractor or moving to an alternate attractor. This describes complex systems as adaptable because the general multi-agent organisation is robust to internal perturbations or external environment changes.

There is a functional purpose for the enslaved multi-agent system which is in the form of a functional synergy. The discussion above suggests that a complex system finds order in the form of low-dimensional functional solutions from high-dimensional chaotic interactions. Capturing the details of low-dimensional solutions in the motor output workspace has been employed for researching posture control (Hsu, Scholz, Schoner, Jeka and Kiemel, 2007) and postural control of walking (Roberts & Latash 2009), however no studies have applied this to toe clearance control.

Dynamic systems theory has been the predominant application to gait analysis because its basis is conducive to the robust autonomous rhythms generated by the complex locomotor system (McGeer, 1993; Garcia, Chatterjee, Ruina and Coleman, 1998; Collins et al., 2005). Using trajectories in state-space to represent system parameters, the DST tools explain regions of gait stability as a means of indirectly identifying the nature of control. Through this philosophical framework of self-organising nonlinear systems, DST researchers faced the difficult challenge of creating large enough stability regions when gait trajectories are naturally non-periodic (e.g. as caused by foot collisions at heel contact). To overcome this problem, researchers propose a form of 'active' control to help stabilise trajectories when they appear to diverge from a region of stability. When considering discrete goal-directed tasks which require high-level control, like targeted reaching or striking tasks, OFCT has become a preferred model to directly test sensorimotor control hypotheses (Loeb et al., 1999; Scott, 2004; Wolpert, 2007). The OFCT philosophy has been applied in a limited way to investigate how walking is controlled (e.g. Pandy, 2001; Kuo, 2002b). The benefit of discussing OFCT is the mechanistic framework from which it can examine both sensory and motor redundancies involved in movement control (Harris and Wolpert, 1998; Hamilton and Wolpert, 2002; Diedrichsen, 2007; Liu and Todorov, 2007; Nagengast et al., 2010). Recently, there has recognition by DST proponents to incorporate the linear form of OFCT-type models into the mechanical simulation of human walking (Schaal, Mohajerian and Ijspeert, 2007; Iida and Tedrake, 2010; Manchester et al., 2011). The recent reconciliation of the two methods indicates that both DST and OFCT should be reviewed when evaluating how the toe clearance task is controlled during walking.

2.3.1.2 Redundancy in walking

The task functions which need to be satisfied for walking were indicated in section 2.2. The ability to simultaneously perform the dual control tasks of toe-swing and posture stabilisation requires solving the problem of coordinating the many degrees-of-freedom. This problem of redundancy also has a more positive perspective when assuming that an appropriate control structure exists, whereby the high dimensionality of the loco-sensorimotor system has an abundant repertoire of available resources to be able to optimally perform simultaneous movement tasks (Latash, 2008). The problem for the loco-sensorimotor control system is how to coordinate the subcomponents and resolve multiple and simultaneous sub-tasks of walking. Some questions can be put forward that are relevant to coordinating the redundancy of the locomotor system for satisfying the dual control tasks of toe-swing and whole body balance. Are the effectors of these walking control tasks coordinated independently? What is the effect of toe-swing on posture stability? At what stage in the cycle do the tasks share common mechanical goals and therefore allow the effectors to act under a mutually shared coordination structure? To what extent is the toe-swing performance improved when these tasks are coordinated together? It appears from the research literature that these questions are yet to be explored in gait analysis.

The flexibility afforded to the locomotor system is a beneficial feature provided by its high redundancy but this becomes a problem for motor command decision making by central nervous system control. To make this process easier and less taxing for the central nervous system, it is proposed that control policies could exist differently at different levels of the control hierarchy (Loeb et al., 1999; Li, Todorov and Pan, 2004). This follows on from the proposed view of a high and low level controller as described in section 2.3.1.1. This indicates that the framework of motor control described by the loops in Figure 2.3.1.1.1.1 are proposed to be acting at multiple levels of the central nervous system and in parallel. The type of feedback received is dependent upon the controllers but also the communication between controllers at the different levels is part of the proposed hierarchical control structure representative of the spinal cord and motor cortex areas (Li et al., 2005). The low level controller receives full dimensionality of the body state in the environment and sends a copy in a low dimensional form to the high level controller. This relayed information to the high level controller is in a low dimensionality form because it carries only the task-relevant information, and therefore provides the high level controller with only the relevant details to optimise performance (Liu and Todorov, 2009). The theory of optimal controllers is based upon evidence of how variance is structured according to the principle of minimal intervention (Li et al., 2004; Todorov, 2004) and also demonstrated by the uncontrolled manifold hypothesis (Scholz and Schoner, 1999; Latash) (see section 2.4.4). The task-redundant variance does not require control because variance in this subspace is task-irrelevant and therefore the high level controller does not require information about the body state in this subspace. The controllers only need to monitor variance in the task-relevant subspace because these variations in body state estimate will affect the task goal. It is proposed that the central nervous system does this by setting sensory features (group of sensory measures, i.e. modules or synergies) which extract and optimise the task-relevant features and then map onto task-relevant motor synergies which are in accordance with the task goal (Liu and Todorov, 2009).

2.3.1.3 General motor patterns selected to achieve the tasks of walking

While the manner in which system redundancy is resolved remains debatable so far, there have been numerous attempts to describe details of coordination occurring within the locomotor system. These will be described below, and from these investigations it is apparent that coordination is related to sub-task goals of walking. Empirical investigations into the effect of muscle activity on the movement kinematics and kinetics of walking has revealed modules of muscle groupings at certain phases of the gait cycle (Ivanenko et al., 2004). This has been further supported in simulation studies (Neptune et al., 2009a).

There have been gait studies implying synergies from gait biomechanics that are associated with sub-tasks of the gait cycle (Ivanenko et al., 2006; Shemmell et al., 2007; Neptune, Zajac and Kautz, 2009c; Sasaki, Neptune and Kautz, 2009). Muscle simulation studies have identified muscle 'modules' believed to be functional because of how their combined actions are shown to affect the mechanical state of gait (Ivanenko et al., 2006; Neptune et al., 2009c). These studies only report average details of muscle activations, and therefore cannot explicitly make a link to the task goals of walking. For example, simulation studies can only approximate the physical attributes of the loco-sensorimotor system, and therefore any gait kinematic details derived from the simulation will generally have poor accuracy (Neptune et al., 2009c). However, muscle modules identified in walking provide the only insight into muscle synergies (task-relevant) found in human walking. Neptune (Neptune et al., 2009a) has shown that there exists muscle modules associated with swinging the leg during walking. Yuri Ivanenko (Ivanenko, Grasso, Zago, Molinari, Scivoletto, Castellano, Macellari and Lacquaniti, 2003; Ivanenko et al., 2004; Ivanenko et al., 2006; Ivanenko et al., 2007) has applied both electromyography and kinematic analysis to probe more specifically into the link between coordination synergies and the toe-swing task (walking, kicking, crossing obstacles). While the electro-myography studies have provided supportive evidence of muscle modules, the kinematic studies by Ivanenko have discovered links between synergies of kinematic (effector) details and task goal of toe position (end-effector) using principle component analysis (Ivanenko et al., 2007). The outcome proposed that the task details of limb length and orientation modulate the co-variance of the effector kinematics. This is the only body of work which has attempted to make a general link between toe-swing task and lower limb coordination. The specific details of effector synergy relations with end-effector position at critical events of the swing phase have not been addressed.

2.3.2 Control of toe states from optimal feedback control theory framework

So far the review of the literature has covered the anatomy of the complex embodied system and the issue of coordinating redundancy of this system. This section will describe optimal feedback control theory (OFCT) because it has been used successfully to represent how the complex sensorimotor system functions to precisely control movement tasks. There have been only very recent works that have applied this theory to walking, however, recently there have been slight variations of the framework that demonstrate successful simulations of walking (Tassa, Erez and Todorov, 2011).

Controlling the state of the toe position during leg swing requires the loco-sensorimotor system to develop a strategy that performs three related functions. First, it needs to optimally infer the state of the body and environment. Second, it needs to determine a management policy for minimising unwanted behaviours associated with the task goal. For example, such behaviours might relate to variable states of the body, or the expense of energy from either cognitive or metabolic demands. This is the form of a cost function, or cost policy. Third, from an available set of motor actions, it needs to select the motor commands that achieve states that are associated with minimising the unwanted behaviours determined by the cost policy. These three functions listed above represent the Optimal Feedback Control Theory (OFCT) framework.

Understanding the processes that control toe clearance are best defined through these three functions of OFCT.

2.3.2.1 Optimal estimate of toe states

The first requirement for controlling toe clearance is to obtain accurate details about the state of the toe position. Section 2.2.3.2.3 explored the nature of sensorimotor uncertainties, describing that these are mostly due to neural noise existing in both afferent and efferent processes (Faisal et al., 2008). For dynamic tasks like the toe-swing, the ability to act optimally requires an accurate state estimate of the toe position, and this is limited to the uncertainty of information gathered by the central nervous system. To optimise the state-estimate, the central nervous system processes the information according to Bayes rule, which has now become a statistical theory 'Bayesian inference'. The application of Bayes rule has been predominantly researched in novel pointing tasks of the upper limb (Harris and Wolpert, 1998; Hamilton and Wolpert, 2002; Kording and Wolpert, 2004). A study by Kording and Wolpert (2004) showed that by modifying grades of sensory feedback results in a progressive increase in error in final position of goal-directed finger pointing tasks.

Bayes rule is described in the equation below, describes how a optimal state-estimate [*posterior;* P(state|sensory input)] is obtained from prior experience [*prior;* P(state)] and the probability of receiving the sensory information given the hypothesized state of the world [*likelihood*; P(sensory input|state)]:

$$\overbrace{\mathrm{P(state \mid sensory\ input)}}^{\text{Posterior}} = \frac{(\overbrace{\mathrm{P(sensory\ input \mid state)}}^{\text{Likelihood}}\ \overbrace{\mathrm{P(state)}}^{\text{Prior}})}{\mathrm{P(sensory\ input)}} \qquad (2.3.2.2.1)$$

The process of acquiring the optimal state-estimate is basically a distribution of weighted probability distributions of the maximum likelihood estimate [P(sensory input|state)] and the distribution of the prior [P(state)]. The 'prior' reflects the probability distribution of previous states, and is independent to the 'likelihood'. For example, the toe location at MX1 is not equally probable at MTC, the probability distribution of priors will be dependent upon the state of the swing phase. The CNS appears to store this prior experience information (Kording and Wolpert, 2004). Evidence from research also has shown that after people learn and store the prior probability distribution of a task they then combine this with sensory evidence, according to Bayes rule (Kording, Ku and Wolpert, 2004). The CNS needs to know what the probable likelihood is for observing the sensory input information given the specified toe location. The evidence of Bayes rule in sensorimotor tasks indicates the central nervous system can represent the uncertainty of its sensors (Kording and Wolpert, 2004). For example, the CNS will consider the probability of receiving the observed proprioceptive information given a specified toe location.

Most processing of motor commands from proprioceptive and cutaneous input requires a time delay. Furthermore, these time delays can be expected to be longer in older adults (Sosnoff and Newell, 2007). The time between toe off and MTC is approximately 0.2 seconds for walking speeds of approximately 1.1 ms^{-1}. An internal forward model can improve the state-estimate by having the predictive ability of knowing the sensory feedback of future states (Wolpert and Flanagan, 2009). This is achieved by the CNS using an efferent copy of information relayed by outgoing motor commands. The body has shown to apply Bayesian inference and a Kalman filter process for integrating time delays and forward predictions (Kording and Wolpert, 2004; Vaziri, Diedrichsen and Shadmehr, 2006).From this view, the state estimate at a time instant will combine the predicted forward model information with delayed sensory feedback information (Wolpert, Ghahramani and Jordan, 1995; Wolpert and Flanagan, 2009). This is illustrated in Figure 2.3.2.1.1A.

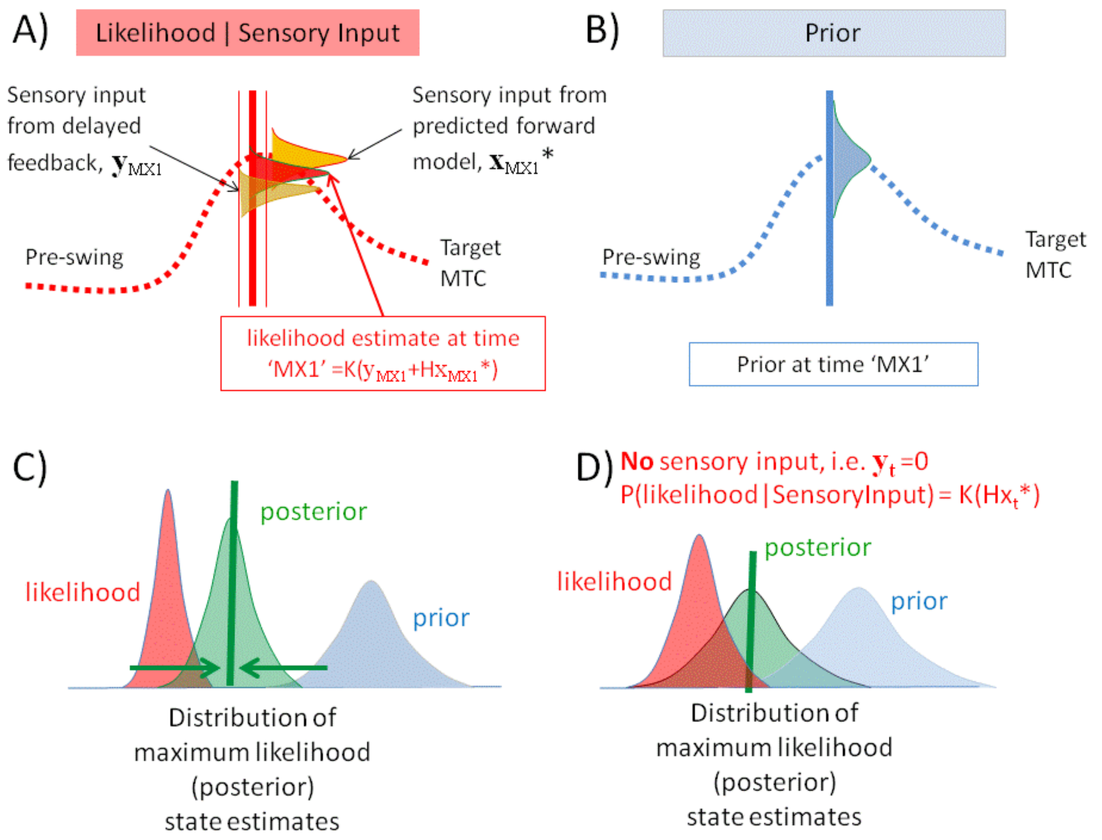


Figure 2.3.2.1.1

Estimating the vertical toe position at time 'MX1' using Bayesian inference from uncertainties in sensory feedback, uncertainties in outgoing motor commands, and uncertainties of prior states. A) Coping with feedback delays, the 'dynamic likelihood' is obtained by applying a Kalman filter process, which weights the uncertainties (distributions) of all sensory inputs, $\mathbf{y}$, and uncertainties in the feedforward model (predicted state based on outgoing motor commands), $\mathbf{x}$*, according to K(y+Hx*). B) The distribution of 'priors', uncertainty of known previous states of the toe at MX1. C) The figure represents the 'posterior', or optimal state estimate (green distribution), from the weighted combination of the 'prior' (blue distribution) and the 'likelihood' (red distribution) according to Bayes rule. D) The effect on the 'posterior' uncertainty when sensory feedback is not available and the 'likelihood' uncertainty is increased because it now relies upon the distribution of the forward model, outgoing motor commands.

Figure 2.3.2.1.1C illustrates the toe position estimate at MX1. If sensory feedback is degraded or becomes more uncertain, then the likelihood distribution will be more uncertain and reflect only the forward model distribution. The 'posterior' state-estimate (green curve) will be weighted more closely towards the 'prior' distribution (blue curve). This demonstrates a possible strategy for how an older adults toe trajectory state will behave if sensory information is degraded. Hypothetically, if feedback is unavailable then the state estimate will be weighted towards the distributions of the 'forward model' and the 'priors' (Figure 2.3.2.1.1D). While the CNS could use the forward model without the actual feedback to estimate the 'sensed' body state, an inaccurate/uncertain forward model will cause the state estimate to drift over time. This can be alleviated somewhat by forward models learning to predict future observations of sensory information by comparing the forward model sensory outcomes to actual sensory outcomes (Wolpert et al., 1995). The errors in the difference can be attributed to prediction and updating the forward model will improve future state estimates and this can be further improved with increasing experience (i.e. movement task practice).

Figure 2.3.2.1.2 illustrates the hypothesis that older adults will have greater uncertainty of toe states compared to young adults. This is based upon the findings from section 2.2.3.2.3 which suggests the increase in 'noise' existing in the elderly afferent and efferent processes. The relevance of the state uncertainty has implications for the design of a control policy. With increased uncertainty, there exists a greater risk that the movement outcome is less precisely known. Therefore, management of the risk will exist in the form of a control policy.

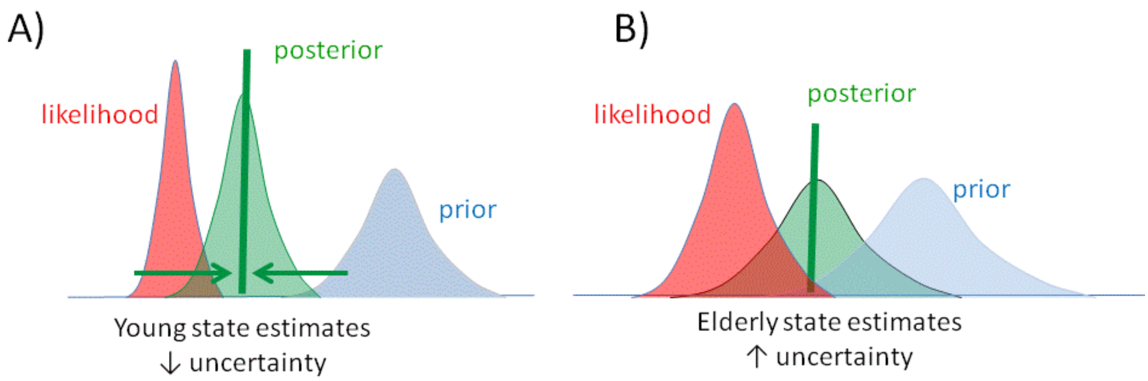


Figure 2.3.2.1.2

A hypothesis about the comparison between the young and older adults when they make predictions about the toe state at a given time instant along the swing-toe trajectory during walking. A) The young group will have less uncertainty when estimating toe states because the prior distribution and the likelihood distribution will be more peaked. B) The elderly group will have increased uncertainty when estimating to states because the likelihood and the prior will have expanded.

2.3.2.2 Control policy

The Bayesian inference of the posterior estimate of the body state now allows the CNS to decide upon an appropriate action. This section combines the second and third functions of the optimal feedback control theory model, cost policy and motor command selection. For example, to avoid tripping and increase safety, the CNS might decide upon changing the vertical state of the toe. Alternatively, the state of the toe location might be high and a change to a lower location might result in less energy expended. Therefore, the decision to act will depend upon what the CNS determines to be of potential reward and of potential cost (or loss). For the toe-swing performance, there is likely to be certain degrees of individual differences when assigning a reward/cost strategy because of individual perceptions of value associated with the assigned outcome. Some actions may prove to be more risky than others because of the associated cost/reward uncertainty, however, the risk might outweigh the potential reward value.

The CNS ‘controller’ considers the current state estimate of the body, with respect to the cost/reward policy, to then estimate in advance the optimal motor command selection to achieve the goal state. Optimization of command selection operates on the premise that a set of sequential actions will be selected that is the result of an ‘a priori minimum’ solution from a set of results that have been estimated from the repertoire of cumulative state-to-state actions that can transpire for all possible future states in the world (Todorov and Jordan, 2002; Todorov, 2008). The action with the least loss gets selected and this is updated again at the next state. Hence, optimisation computations are exhaustive for

$$\sum_{states} L(action, state)\ P(state|sensory\ input)$$

The function plays the important role by summarising all relevant information at each state about the future, revealing the cost that will accumulate from this state to the end goal state. It therefore indicates the most optimal control solution to make at each state, and thus the optimal motor command.

2.3.2.2.1 Idiosyncratic movement patterns

It has been reported that similar muscle commands are repetitively chosen to perform sub-tasks of walking between individuals (Ivanenko et al., 2004), indicating that the ‘gait’ controller has aspects that are common across people. People show idiosyncratic behaviour when selecting a favoured set of muscle commands from the highly redundant repertoire when walking (Neptune et al., 2009a) and when completing novel movement tasks. The walking pattern has learned to avoid energetically costly disturbances (Donker, Daffertshofer and Beek, 2005; Dingwell and Marin, 2006) and it is believed that this has evolved simply by selecting movement patterns which minimize energy expenditure (Pandy, 2001; Anderson and Pandy, 2003; Kuo, 2007; Neptune et al., 2008). While, these studies have overlooked the ability of the loco-sensorimotor ability to use feedback and attend to an external goal, the studies do demonstrate the weighting afforded to the regularization term of energy expenditure. Furthermore, resolving the redundancy problem by attaining coordination of effectors has been able to be explained by the minimization of muscle energy when processing the minimum 'cost-to-go' function.

The minimization of muscle activity also has an effect on movement variability because motor noise is increased when the size of the motor pool is increased (i.e. motor command signals) (Harris and Wolpert, 1998). The controller therefore, learns to optimize performance by exploiting the available redundancy of effectors and appears to prefer a coordination strategy which distributes the work across a set of multiple muscles according to the minimization of effort and the consequences of motor noise on movement error (Harris and Wolpert, 1998; Haruno and Wolpert, 2005; O'Sullivan, Burdet and Diedrichsen, 2009). This distribution across multiple effectors is described as cosine-like tuning, which means that a select muscle group is recruited to satisfy the cost function in counter-intuitive ways, as there are muscles within this group which activate to satisfy the cost function but they perform pulling directions which are somewhat adjacent to the movement goal (Todorov and Jordan, 2002; Haruno and Wolpert, 2005). One reason why the central nervous system selects cosine tuning is because this strategy distributes the muscle work across the effectors in an optimal way, reducing the sum of squared motor commands (Todorov and Jordan, 2002). Others have suggested this tuning is a strategy to reduce movement variability (Haruno and Wolpert, 2005).

The term effort is argued to be a better term than energy (mechanical work + heat) because optimization models have found that the minimization of the sum of the squared muscle commands provides a more accurate account of the movement details compared to the sum of the these commands (Todorov and Jordan, 2002). One view is that the nervous system accounts for energy in a conservative way by attributing it with a higher cost than it actually represents (Diedrichsen, 2007). Harris and Wolpert (Harris and Wolpert, 1998) found that the noise associated with end-point variability of the arm and eye increases monotonically with muscle command signal, i.e. noise increases with muscle force. Research has generally found that movements become more variable as the muscle signal applied increases due to the physiology of the motor pool (Bays and Wolpert, 2007; Faisal et al., 2008). When the control function optimizes the minimization of variability cost, due to signal dependent noise, the sum of the squared motor commands has also demonstrated to be the optimal solution. However, both components are not the same. When optimizing a simple performance task of the upper limbs the simultaneously considered cost contribution of both effort and variability were shown to be different, such that effort is weighted by a factor of x7 relative to variability (O'Sullivan et al., 2009). These relative cost components will influence the way the central nervous system distributes work across the redundant effectors.

2.3.3 Framework of Optimal Feedback Control Theory (OFCT)

After presenting the components of the framework, Figure 2.3.3.1 describes the OFCT model. The 'plant' (body state within the environment) is represented by the vector x. Sensory information, y (e.g. proprioceptive and cutaneous mechano-receptors, vision, etc), of the current plant state is optimised using Bayesian inference (see section 2.3.2.1) and then integrated with the efferent-copied forward model, x^* (also optimised by Bayesian inference). The optimal state estimate of the plant $\hat{x}$ is determined by sensory integration process using a Kalman filter, K, computation (see section 2.3.2.1). For situations where external feedback is time delayed, when rapid processing of state estimates is required, the sensory integration process is too slow. At the top of the OFCT loop is the controller which can be represented at different hierarchy levels of the central nervous system neurophysiology. The controller computes the required motor commands according to a cost/reward function (i.e. -*L*), or control policy (cost function), which captures the task goal. The controller also determines what feedback details from the state estimate are most relevant to the task goal. The outgoing motor commands, u, are selected based upon an inverse model (Wolpert et al., 1995), which accounts for noise in the plant, the task goal, the cost/reward function, the current state estimate. The inverse model exploits the redundancy of the motor system to produce commands which optimise the composition of the cost function, -*L*.

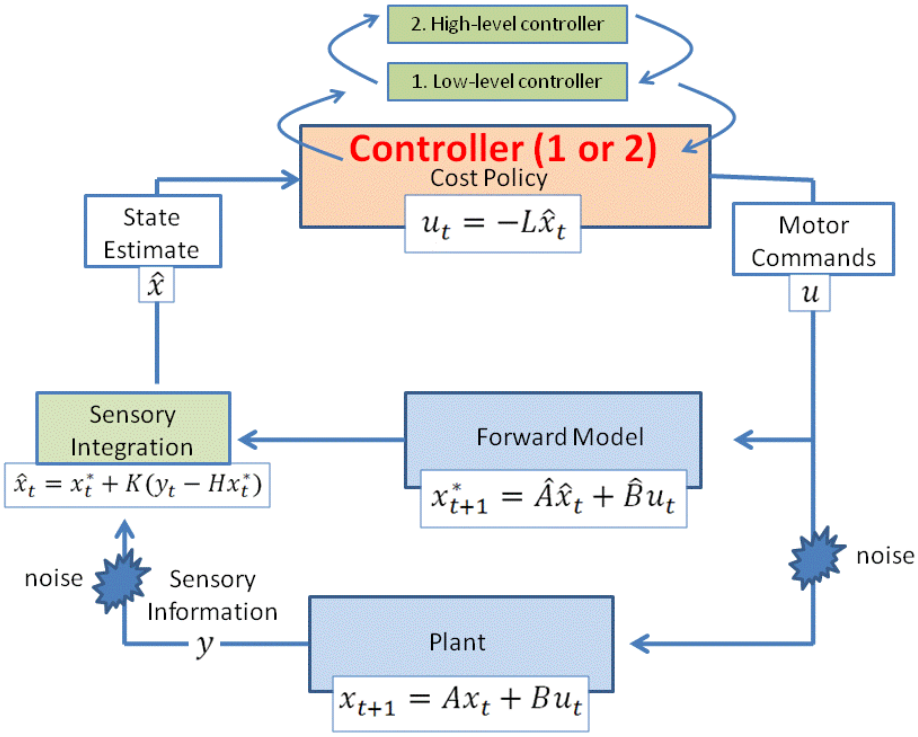


Figure 2.3.3.1

Illustration of the computational aspects of Optimal Feedback Control Theory (adapted from Diederichsen et al. (2010)).

Figure 2.3.3.1 illustrates several loops, the top of the loop is the controller which can exist at possibly several hierarchical levels of the embodied central nervous system (Rossignol et al., 2006), but here the levels are described generally as low and high level controllers. At a time t, the current state estimate, $\hat{x}$, is mapped to the motor command, u, by the cost policy or cost function, $-L$; where $u(t) = -L(\hat{x}(t))$. The outgoing motor commands are governed by how the controller views the state estimate based upon its cost policy.

The model is based upon a computational perspective which in a positive way allows for tractable explanations of empirical findings (Scott, 2004; Doya, 2009). The model has become an important computational analysis tool for developing optimal performance principles that have allowed elaborate predictions on how the body minimizes uncertainty and movement variability. It is argued that optimisation models have accounted for more empirical phenomena than any other class of model (Liu and Todorov, 2007; Todorov, 2009). Biological based research observations provide important insights into the development of specific OFCT models but biological research alone does encounter limits when probing the depths of knowledge of the neurophysiological embodied components underlying the loco-sensory -motor system (Scott, 2004; Rossignol et al., 2006; Srinivasan and Ruina, 2006). Both mechanistic and biologically based research approaches provide mutual information in which to advance both approaches to investigate the motor control system. The benefit of the model for this thesis is in the way that it serves as a general qualitative framework and guide for understanding the parameters detailing the performance of the toe-swing task.

Studies on walking have been applying mechanistic models to simulate neuro-musculoskeletal behaviours because controlled experimental protocols can be too complex or difficult to appropriately assess research questions which relate to locomotor embodied complexity (Lin, Walter, Banks, Pandy and Fregly; Chou, Song and Draganich, 1995; Anderson and Pandy, 2001; Kuo, 2002a; Anderson et al., 2004; Ruina, Bertram and Srinivasan, 2005; Srinivasan and Ruina, 2006; Emken, Benitez, Sideris, Bobrow and Reinkensmeyer, 2007; Scafetta, Marchi and West, 2009). The success of these mechanistic models is based upon optimizing certain task related features for obtaining close simulation with available experimental observations. The form of these models extends only as far as the details addressing the research question. Most optimization models have assessed the average behaviour of gait mechanics from an open loop control perspective (Chou et al., 1995; Anderson and Pandy, 2001) and only few have broadened the model to include mechanical responses to closed loop, sensory feedback information (Kuo, 2002b; Emken et al., 2007; Pham, Hicheur, Arechavaleta, Laumond and Berthoz, 2007). The optimal feedback control theory model developed by Todorov and Jordan (Todorov and Jordan, 2002) has had only limited application to locomotor studies (Emken et al., 2007; Pham et al., 2007).

2.3.3.1 Cost policies for walking

The cost function, '-L', can be composite in its form and is related to general homogeneous costs, such as energy consumption (Pandy, 2001) or endpoint variability

(Harris and Wolpert, 1998). The minimization of these costs have been able to account for kinematic trajectories (Harris and Wolpert, 1998; Todorov and Jordan, 2002), joint torques , muscle activity (Valero-Cuevas, Venkadesan and Todorov, 2009), and task-redundant and task-relevant variability structure (i.e. emergence of muscle synergies) (Todorov, 2004; Todorov, 2005).

When considering the above theories for toe trajectory during walking, the application of a cost function can be hypothesised for the MTC event. The cost function can stipulate where toe locations need to be assigned a penalty value. High values represent greater penalties and therefore contribute against the minimisation of the cost function. Similar to a regression line fitted to data by considering the squared error of the points which the line passes through, a cost function can also represent a similar relationship with performance and penalties for errors. For the MTC event, the likely penalty which is assigned a cost by the central nervous system is the state estimate of toe trajectories which travel close to the ground, or at the lower boundary of the estimated safety zone. When this is compared to toe trajectories which are at the higher boundary, the cost partitioned would be expected to be modulated differently. Given two consecutive toe-swing tasks during unobstructed level walking, the controller might award a penalty based upon the vertical toe location being below the desired nominal threshold height. An error at the MX1 event (error relative a nominal value) for the one set of two trials, of -1.2cm and -1cm, versus another set of two trials of -1.5cm and -0.8cm. The total cost for the first set of trials will be $(-1.2)^2 + (-1)^2 = 2.44$.; whereas the alternate performance sequence represents a total error cost of $(-1.5)^2 + (-0.7)^2 = 2.74$, demonstrating an increase in error cost by a factor of 1.1. This demonstrates that the sum of the control cost rises for extreme errors. In this example, the penalty assigned for negative errors is a squared function, however, positive errors may take on a different function that assigns a lower penalty. This represents an ‘asymmetric’ cost function, and can be reflected by a skewed distribution (Figure 2.3.3.1.1).

The central nervous system is likely to penalise MTC errors differently across cycles and between individuals. It is interesting to see that the typical MTC histograms of both healthy young and healthy elderly subjects reported by Begg et al. (2007), show a positively skewed distribution, and this was greater in the elderly. It remains to be investigated whether positive skewness is the outcome from a cost policy assigned to toe position errors, and whether such a policy is modified with ageing. The MTC histogram distribution might be reflecting the outcome of a cost function that awards higher penalties to values below the desired toe location relative to locations above the nominal state (Figure 2.3.3.1.1).

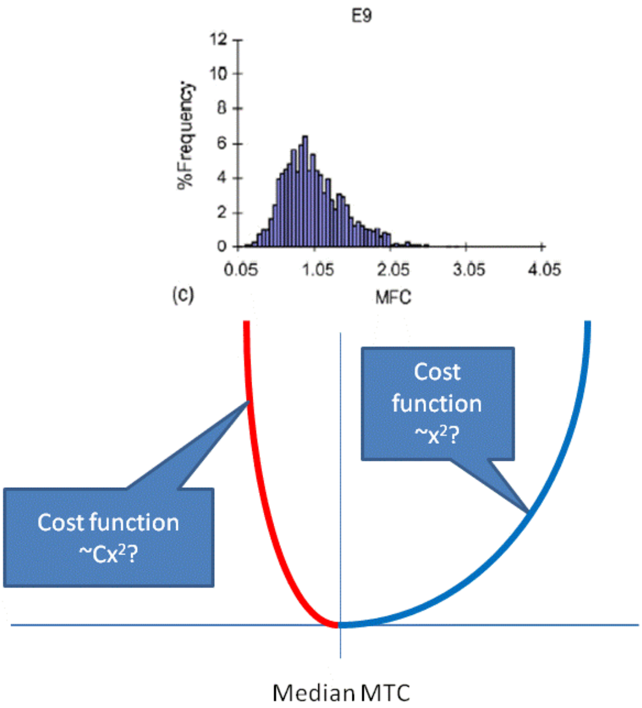


Figure 2.3.3.1.1.

Decision of toe swing trajectories are based upon what outcomes the brain assigns a penalty. Based upon the common finding of positive skewness in MTC distributions (Begg, Best, Dell'Oro and Taylor, 2007), a hypothesized cost function strategy in the form of how the central nervous system internalizes penalties of accuracy errors in the state of toe trajectory height at regions either side of the median/intended MTC. The strategy of the cost function in this case is likely to reflect a greater weighting of accuracy cost relative to energy/effort cost.

The OFCT model can account for control schemes with multiple costs which may switch priority in a flexible way, or include composite costs which may have proportionate weightings which are more stable across trials (Liu and Todorov, 2007; Diedrichsen and Dowling, 2009; Nagengast et al., 2010). With comparison to empirical data, various forms of control functions have explained how people will accordingly adjust their planning and execution strategies to achieve optimal performance of a task goal (Liu and Todorov, 2007; Diedrichsen and Dowling, 2009; Nagengast et al., 2010). The outcomes from OFCT research also show that tracking of a criterion behaviour is ‘effort’ inefficient, such as the error tracking from state-to-state of an end-effector trajectory, but the performance of movement details can be explained much better by controlling for a task relevant cost like minimising final end point variability (Harris and Wolpert, 1998; Todorov and Jordan, 2002; O'Sullivan et al., 2009; Nagengast et al., 2010). The control regulation of the toe-swing task has already been discussed as a dual (possibly multi) control task, with multi-effector redundancy, and the flexibility of the control functions of the OFCT model thus have practical relevance as a framework to investigate the control principles of the toe-swing task.

What would the cost function look like for walking? What might be the gait parameters which express the outcome of such an applied cost function? It has already been demonstrated that minimising energy expenditure during walking is related to step length and step width. Could it be possible that the squared error of these parameters is sought to be minimized by the controller? These parameters might just be a by-product of the general regularization term of energy cost, but how might the external goal term be described? What gait parameters reflect these external goals? The discussion in section 2.2 indicated the toe-swing and balance are likely to feature in such a description of external goals.

2.3.4 Risk sensitive cost function

Nagengast et al. (2010) applied optimal feedback control theory to investigate control policies that consider variations of the movement costs, rather than just the average accrued costs, when performing a goal-directed movement task within a ‘noise’ affected system. The study has relevance to strategies adopted by older adults when controlling the toe-trajectory in an uncertain/noisy system. Movement trajectories were compared to simulated data derived from different cost policies in an OFCT model. The cost policy was designed to penalise target ‘error’, and the accumulated ‘control’ effort applied along the trajectory. The empirical task was to perform movements that reduced the accumulated penalties in a trial. Simulated data was compared to empirical data to explain what cost policies are being employed. The optimal solution task, therefore, was to minimise control effort and minimise target errors. In a noisy system, however, this sets up a dilemma for the controller, because target errors will be expected to increase with minimal control effort. Nagengast et al. (2010) explored the task under high noise and low noise conditions. In addition to the cost of online 'control' effort and final target 'error', there was a third cost variable added which was 'uncertainty' of the accumulated cost, and this represents risk. In noisy systems such as the human sensorimotor system, error uncertainty is difficult to control, whereas control effort is volitional. Nagengast et al. (2010) considered this uncertainty, and set up the optimal solution to minimise 'online control effort', 'final target error' and the variance of the cost associated with the outcome, i.e. 'uncertainty'. This presents a nice insight into the type of control policy issues faced by older adults when controlling minimum toe clearance states.

The outcome of the study demonstrated that people accept control cost when systems have increased noise. When there was a greater penalty for control effort, the control effort decreased and the subsequent target error increased. There was increasing control effort applied as the state of the movement trajectory approached the target goal. Nagengast et al. (2010) proposes that in accepting penalties of increased control effort, the payoff was a minimisation of uncertainty in the accrued cost of error and control. So, while this strategy is in general more costly, the outcome uncertainty of the penalty is minimised. Therefore, this study indicates that people place a greater value on minimising the uncertainty of costs, rather than explicitly optimising all costs. The relevance to an older adult performing the toe clearance task is this, that they can be expected to have a policy that accepts movement errors associated with a set level of control. This policy will typically expend more control effort than generally required, but the policy will minimise the uncertainty of the accrued penalties. Another important outcome suggests that control effort penalty increases, there is less control applied. Because older adults have a limited resource pool for central nervous system processing (i.e. section 2.2.3.2.2), the penalty assigned to 'control effort' by the elderly cost policy can be expected to be higher when compared to the young. Therefore, the cost policy of the elderly will need to address a ‘noisy’ loco-sensorimotor system, with a premium value on control effort. The walking pattern adopted will reflect these issues.

2.3.5 Passive control of toe trajectories: biomechanical control

The problem being assigned in this thesis is that of how to control the toe trajectory optimally. The cost policy described above can potentially reveal a central nervous system strategy for trip risk avoidance. The policies and task goals recognized by the controller during different states of the swing phase will be outlined below.

It has been found that the motor control system considers a trade-off between increasing impedance control and minimizing end-point variance (due to signal dependent noise) when a movement task involved going around obstacles (Selen, Franklin and Wolpert, 2009). To reduce kinematic variability of the end-point trajectory, stiffness/impedance control (effect of muscle co-activation) at the effector end point is aligned orthogonal to the direction of mechanical instability (due to increased SDN). This is because increasing stiffness in the same direction would only contribute to the SDN and be a counterproductive compensation strategy to reduce end-point variance. The trade-off indicates that increasing stiffness too much will be suboptimal as it would begin to contribute proportionately more towards SDN instability. Furthermore, stiffness reduces the adaptability of the effector group (Liu and Todorov, 2007). The more optimal alternative to going around obstacles is to rely upon efficient perception-action matching (Wolpert and Flanagan, 2010). This will require tuning the sensors to the task relevant information and training the mapping of this information to ‘ready to go’ motor commands carried out by slow reflex loops which are customized to match output to the task-relevant information (Liu and Todorov, 2007; Johansson and Flanagan, 2009). It has been reported that large end point errors increase stiffness (Franklin, Burdet, Tee, Osu, Chew, Milner and Kawato, 2008) and therefore by improving predictive ability, there will be less end point error because of a subsequent decrease in the SDN property.

2.4 Quantifying toe clearance performance

This review of the literature explores the leg swing of walking as a performance task which can be quantified by considering the effector system involved and the context of other task goals. The framework of optimal feedback control theory and dynamic systems theory contribute information about different levels of control intervention when performing this task. Performance of this task must be considered with respect to the three general areas which an adaptable movement system contains: perception, control and action. To take steps forwards in minimizing tripping risk in older adults and therefore to reduce the rate of falls, the valid assessment of this task is important. The current state of the research literature on this important task is under resourced and probably limited in contextualizing its performance.

2.4.1 Research approaches to investigating toe-swing performance

The approaches adopted by experimental research to explore how a trip is manifested from biomechanics of walking can be grouped into three areas:

1) preferred unobstructed walking and analysis of the toe-to-ground trajectory (Begg et al., 2007);

2) obstructed gait and tripping response (van Dieen and Pijnappels, 2007); and

3) obstacle avoidance and crossing (Begg and Sparrow, 2000).

The first area has a focus on the behaviour of the locomotor system when sensory information in the environment remains familiar and stable. The locomotor system therefore doesn't require frequent support from the higher level centres (i.e. supra-spinal) about changes in the environment but rather the walking pattern evolves from the predominantly lower level (i.e. spinal) autonomous processes. The second area focuses on the ability to adapt and generate new motor commands which lead to a successful trip recovery. This form of research doesn't explore the cause of tripping frequency, which has been found to have a stronger association with falling in comparison to the ability to recover from a trip (Pavol et al., 1999). The third research area considers how gait patterns are modulated when the context of the task changes significantly due to obstacle(s) in the walking path. Because of the nature of the tasks, all three approaches described above measure different aspects of the loco-sensorimotor control system. All three approaches have contributed knowledge in separate areas, but all have contributed important information to help understand the link between walking and tripping.

Experimental designs that perturb walking stability have been conducted to explore the response of the locomotor system to tripping (Pijnappels, Bobbert and van Dieen, 2005) as well as walking over obstacles (Chou and Draganich, 1998; Lam and Dietz, 2004). So far, there is yet to be any research linking falls with deficient swing limb mechanics when negotiating obstacles in the environment. When older adults are negotiating obstacles, the general findings are that older adults need more time for selecting the appropriate foot trajectory (Lowrey, Watson and Vallis, 2007; Weerdesteyn et al., 2007) and that they need to use a better visual gaze strategy for foot placement (Chapman and Hollands, 2007). From observational studies, trips in older adults generally result from unseen obstacles inside the home environment and on a level surface (Campbell, Borrie, Spears, Jackson, Brown and Fitzgerald, 1990). This might suggest that trip-related falls occurring in the context of obstacle negotiation is not linked to trip-related falls as strongly as the context of tripping over an unseen obstacle. This can be argued to be a strong case for conducting appropriate experimental approaches to examine foot swing characteristics during level walking in an invariant unobstructed environment. This thesis is in line with the first research area and investigates the existence of a baseline difference between young and older adults when controlling the toe-swing.

2.4.2 Gait variability

There have been three general methods applied in gait research which have described the different characteristics of gait variability. One method describes stability of the hybrid limit-cycle behaviour of gait trajectories using Lyapunov Exponents (LyE). Another method describes the dimensionality of the workspace, or complexity of the system, using Detrended Fluctuation Analysis (DFA). The third method describes the average state of noise in the system by using standard deviation (SD). Theoretically, all three forms of gait kinematic variability can be independent of each other. Dingwell and Marin (2006) found that stability (LyE) and variability (SD) of joint angle trajectories are independent measures of walking, such that large variability of joint angles can be associated with either high or low stability of the joint angle trajectories. However, joint angle stability does appear to be dependent upon walking speed (Dingwell and Marin, 2006). With respect to time-distance parameters of walking, Gates and Dingwell (2007) found that of complexity (DFA) and variability (SD) are independent measures. Complexity is maximised when walking is without external stressors (Hausdorff, Purdon, Peng, Ladin, Wei and Goldberger, 1996; Scafetta et al., 2009). Interestingly, complexity appears to increase during gait regimes of un-preferred speed (i.e. slow or fast) (Hausdorff et al., 1996; Jordan, Challis and Newell, 2007). From the above evidence, the relationship between stability, complexity and variability seem to be representative of different aspects of the loco-sensorimotor control system. The results from above indicate that the relationship between complexity (i.e. workspace dimensionality) and stability appears to behave differently at slow speeds compared to faster speeds (Dingwell and Marin, 2006; Jordan et al., 2007). In a follow-up study by Jordan et al. (Jordan, Challis, Cusumano and Newell, 2009) the authors generalise that an increase in complexity is associated with a decrease in stability. This argument contrasts with the above discussion. The argument also overlooks evidence from metronome walking (Hausdorff et al., 1996; Scafetta et al., 2009), and the finding that metronome walking under non-preferred walking speeds causes a further reduction to complexity (Scafetta et al., 2009). A proposition which is in line with the discussion above is that the increase in DFA when walking at faster speeds is likely to be related to a control policy which loosely controls instability of the passive dynamic limit-cycle.

2.4.3 Serial correlations and persistence

When a task is performed repetitively, such as sub-tasks of walking, the serial variations from trial-to-trial reveal information about the function of the control system (Hausdorff, Zemany, Peng and Goldberger, 1999; Hausdorff, Lertratanakul, Cudkowicz, Peterson, Kaliton and Goldberger, 2000) and the policy of the control system (Dingwell and Cusumano, 2010). Serial correlation analysis has not been applied to observe the control of states at discrete time-points along a gait trajectory. A time series analysis of a parameter that describes a time series of cyclic state-to-state fluctuations can indicate whether the processes controlling the states are long-term dependent, by quantifying long-range correlations in the time series. Depending upon whether the perspective is from an emergent viewpoint or a hierarchical viewpoint of control, the quantification of changing fluctuations of a performance parameter can be attributed to either: (i) the type of corrective feedback processes being applied; or (ii) the richness of dimensionality in the system from which coordination dynamics arise.

Serial correlations found in the performance history can reveal a level of persistence in behaviour amongst local fluctuation changes. Persistence was a term coined by Mandelbrot and van Ness (1968) when local fluctuation behaviour exhibited a future trend based upon a previous trend. This results in a positive correlation for this time lag. When there exists many positive correlations for different time lags, the time series is said to be persistent, and long-range correlated. The converse of this is when a past local trend is followed by a future trend in the opposite direction, resulting in a negative correlation for this time lag and is called anti-persistence. If such a system is hypothesized to be governed by an open loop, or a closed loop feedback control does not intervene or regulate the performance, therefore the persistent behaviour is analogous to a positive feedback system where behaviour continues in trends with limited corrections (or uncorrected). This could also be likened to limit cycle behaviour, as fluctuations are slowly dissipated across successive cycles until stability is resumed. Conversely, negative feedback is analogous to trends being corrected, or over-corrected. This represents a hierarchical view of serial correlation behaviour, such that the nature of a performance parameter will determine whether it will be allowed to persist with limited corrections, or whether it will be tightly regulated by a control policy.

Serial correlation behaviour has most recently been in vogue with the dynamic systems theory perspective and that relates to coupling of neural oscillators. Under this perspective, persistence of a performance parameter represents a self-organised system, where the interacting (dynamical and reciprocal coupling) components are operating at multiple time scales or frequencies. The outcome behaviour of this parameter is expressed on a power-spectrum plot, or a log-log plot of temporal scaling, and there is no evidence of a dominant time scale or frequency observed (Torre and Wagenmakers, 2009). The view from this perspective is that when there exists a rich source of redundant components there will not be a dominant time scale in the underlying processes (Newell, Mayer-Kress and Liu, 2009). The outcome is observed on a power spectrum which demonstrates frequency contribution representative of multi-scaled timing processes (Newell, Deutsch, Sosnoff and Mayer-Kress, 2006). This is the hypothesized signature of 1/f noise commonly found in biological time series analysis because it can be modelled from mechanistic processes (Peng, Hausdorff, Havlin, Mietus, Stanley and Goldberger, 1998; Newell et al., 2006). From this perspective, a quantification of the noise structure represents the rich diversity of involved processes contributing to the performance behaviour and how constraints of this multi-scale and multi-component resource can affect the serial correlations (Peng et al., 1998; Goldberger, Amaral, Hausdorff, Ivanov, Peng and Stanley, 2002; Newell et al., 2009).

There have been attempts to model the persistence found in timing parameters using models from dynamic systems perspective (Gates, Su and Dingwell, 2007; Scafetta et al., 2009; Torre and Wagenmakers, 2009). Strangely, there has not been an attempt to use serial correlation analyses to assess the existence of fractal-like variance in the generated data from optimal feedback control theory models exploring redundancy of effector control and end-effector accuracy. The fact that optimal feedback control theory is based upon redundant stochastic properties of the incoming and outgoing components, and the model can be represented as a rich source of multi-scaled and multi-component 'measurements and commands', which suggests that this model can make a contribution towards explaining the origins of the signature 1/f noise property found in parameters of human movement. It can also serve as a procedure to strengthen claims of support for the optimal feedback control theory model. Using a mechanistic approach along the same line of view that serial correlations can lend support for exploring the central nervous system intentions Dingwell et al. (2010) demonstrated that serial correlations are parameter specific, and that they are dependent upon the contribution of the performance parameter to the task goal. This in some way refutes the claim that serial correlations are an indication of complexity and that changes in complexity represent a break-down in the multi-scale interactions. This has previously been suggested as the basis for stride timing showing random-like correlations during synchronized (timed to metronome) walking (Hausdorff et al., 1996) and also an explanation for ageing and disease (Goldberger et al., 2002). The suggestion of this break-down in complexity in gait has been refuted by Delignieres and Torre (2009), who demonstrated that the complexity still exists, however, the persistence in serial correlations are reduce due to a two-level control structure composed of a 'hierarchical control parameter' which can force the coupling changes to a 'limit cycle' oscillator model.

The gait patterns of healthy young adults demonstrate long-range correlations within the variability structure of the stride interval (Hausdorff, Peng, Ladin, Wei and Goldberger, 1995). Following on from preliminary studies of gait and heart rate variability, a number of studies conducted on various movement tasks have confirmed the ubiquitous property of 1/f noise inherent in the sensorimotor system (Jordan, Challis and Newell, 2006; Gates and Dingwell, 2007; Torre and Delignieres, 2008). The rationale for applying such measures comes from the break-down of aperiodic *fractal-like* dynamics in stride timing, i.e. random-like correlations, and falls risk in older adults (Hausdorff, 2007). This parameter has been able to distinguish between older adults at risk of falls when average measures of standard deviation were not (Hausdorff et al., 1997a).

The choice of parameters will have an influence on the type of persistence found. Stride time intervals involve a range of gait sub-tasks included within the phases of double support, single support and leg swing. Stride timing intervals and stride length performance history show relatively high levels of persistence compared to step length and step time data (Jordan et al., 2007). Step timing and step length are related to phases involving double support and swing and these parameters require a shared level of control for determining walking speed (Dingwell et al., 2010) which in some ways indicates these parameters to be 'synergistic' for meeting this task. Step timing and step length along with other spatial-temporal parameters of the gait cycle demonstrate a minimum degree of persistence at the preferred walking speed (Hausdorff et al., 1996; Jordan et al., 2007; Schablowski and Gerner, 2010). Step timing demonstrates a higher level of persistence in comparison to step length (Jordan et al., 2007). From a hierarchical perspective, or optimal feedback control theory perspective, the difference can be attributed to the possibility that one of the parameters is control-regulated, whereas the other parameter is not. Because step length has an influence on head stability (Latt, Menz, Fung and Lord, 2007; Latt, Menz, Fung and Lord, 2008) and energetic expenditure (Donelan, Kram and Kuo, 2002), it is most likely to be a parameter under task-relevant monitoring by a controller. The results from Jordan et al. (2007) indicate this by showing that step length is regulated at the preferred walking speed and this regulation is not as ‘tight’ for slower or faster walking speeds. The authors however attributed the change in persistence to be an indicator of neuromuscular constraints made available to the system due to the task demands. This is along the same perspective as restriction of redundant components with multiple time scales, but the inverted U doesn’t agree with the theory that increasing redundant components with multiple time scales underlies the presence of long-range persistence. This is further evidence to revise the causes of *1/f*-type noise.

The fractal-like behaviour raises the question about whether walking may be state dependent as task goals will arise and introduce task-related perturbations arising at certain points along the cycle, and the response of a performance parameter may demonstrate idiosyncratic behaviour which can be picked up by serial correlations. To test the hypothesis of state dependent serial correlations in walking, Dingwell and Kang (2007) applied orbital stability (a non-linear measure of limit cycle stability) to body parameter details (sagittal plane ankle, knee and hip joint angles) and a nominated goal parameter (3-D trunk accelerations) and could not confirm the presence of state dependence in either overground walking or treadmill walking. This is surprising in some respects, given the changing effect of consecutive goal-directed movement sub-tasks in the gait cycle. However, the selection of the parameters detailing movement performance may not represent the task goals of walking in a sensitive way. This suggests there needs to be further studies to explore this area.

2.4.4 Structure of task redundant and task relevant variability

The uncontrollable manifold hypothesis stemmed from dynamic systems theory concepts (Schoner, 1995; Scholz and Schoner, 1999). The recognition of coordination dynamics in the locomotor system raises the issue of stable coordination states and flexibility of the system to switch coordinative states in an energy efficient way which allows the system to be task adaptable. Coordination patterns of the redundant effectors need to be both stable and flexible to control the trajectory of the toe. Self-stabilising neural oscillators suggest that the coordination pattern is able to resist small perturbations. Flexibility amongst the states of the neural oscillator coordination dynamics allows the ability to change between states and effect change in the end-effector trajectory. For the task of toe-swing, the trajectory of the toe (end-effector) is dependent upon the coordination at the lower limb joint (effector) level of the swing and stance limbs. The performance of the toe-swing can be described in relation to this coordination at the effector level. There are many ways to characterize the performance details of a movement parameter, but most of these methods are restricted because of the isolated representation of the parameter with respect to the system of which it belongs to. The uncontrollable manifold hypothesis is a concept which sought to address the issue of solving the problem of redundancy in an effector system by quantifying the variance of when the effector coordination is stable relative to task objective and when the effector coordination is unstable and causes change to the task objective (either unwanted or wanted, so flexibility can be equated with instability).

The functional relationship between the body variance and goal variance (Cusumano and Cesari, 2006) is provided by geometric equation which explicitly defines the relationship. The geometric relationship is used to explore all possible solutions in the redundant effector system for meeting the goal solution. Variations of the uncontrollable manifold hypothesis method are the minimum intervention principle (Todorov and Jordan, 2002), the goal equivalent manifold (Cusumano and Cesari, 2006) and the solution equivalent manifold (Muller and Sternad, 2004).

The uncontrollable manifold hypothesis quantifies two variance parameters that describe the distribution of variance relative to a manifold (equation describing the body-goal relationship) in n-dimensional effector (linear) space. One variance parameter describes variance parallel to the manifold plane and a second parameter describes the variance orthogonal/perpendicular to the manifold plane. The variance parallel to the manifold has various terms which mean the same thing, such as goal equivalent variance (Cusumano and Cesari, 2006), task-redundant (irrelevant) variance (Todorov and Jordan, 2002), good variability (Latash, 2008). The variance along this component of the manifold is free to persist because any variance along this dimension of the plane does not affect the task goal. This can be likened to the stability of the effector system, such that the greater the variance along this dimension, the greater is the stability of the effector 'synergy' upon the task goal. This is an important definition of stability because it incorporates a relationship with a task goal rather than being isolated to the behaviour parameter without system context. The compliment variance in the orthogonal direction also has terms which describe the same thing, such as task-relevant variance (Todorov and Jordan, 2002), non-goal equivalent variance (Cusumano and Cesari, 2006) and 'bad' variance (Latash, 2008). The task goal will be sensitive to variance along this dimension, so that if the variance is large, this will be reflected in the task goal behaviour.

The concept and definition of a synergy is best defined through the uncontrolled manifold hypothesis (Latash, 2010b). Figure 2.4.3.1.1 below illustrates a simple two degrees of freedom effector system and their relationship with a task goal defined by the UCM line (this becomes a n-dimensional manifold in higher dimensional effector space). When the variance along the UCM (Vgood) is greater than the variance perpendicular to UCM (Vbad) then the effector system is suggested to be a synergy because the variance parallel with the UCM satisfies the task goal. When the Vgood = Vbad, the stability of the synergy is not strong.

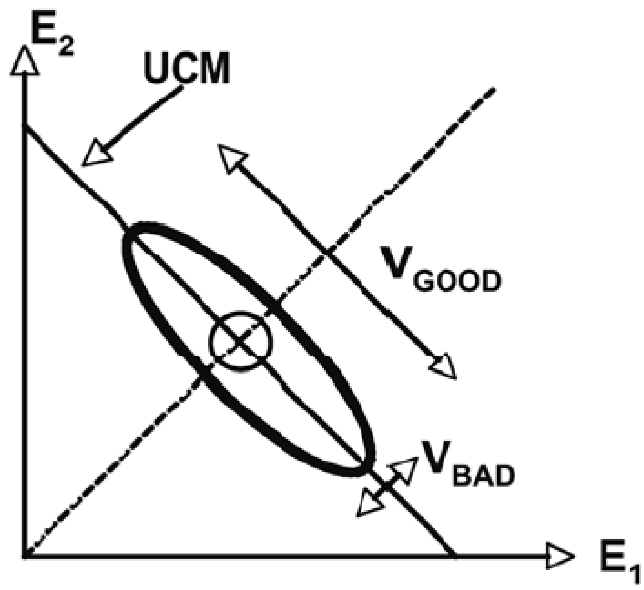


Figure 2.4.3.1.1

The concept of the uncontrolled manifold hypothesis in low dimensional space (figure taken from Latash (2008)). The 2-dimensional space is defined by two effectors, E1 and E2. Both of the effectors are co-related with a task goal which is defined as the line represented by the UCM. This gradient indicates that whenever the E1 and E2 can co-vary such that they lie upon this manifold/line then the required solution is achieved. The values by which E1 and E2 can take on for satisfying the solution are able to extend all along this line. If either E1 or E2 do not cooperate together the outcome will be reflected by a point outside of the UCM line. Any variance along the UCM is good (Vgood) in the respect that the task goal will always be attained, whereas any variance distributed outside, or perpendicular to the UCM (i.e. Vbad) is an indicator of the task goal not being satisfied. Latash describes the definition of a synergy when the effectors co-vary in a task-redundant way. This same concept can be applied to higher dimensional effectors and end-effectors.

The uncontrollable manifold hypothesis provides an intuitive interpretation of how the effectors co-vary to satisfy the hypothesized task goal. There are similar methods to investigate co-variance amongst effectors by applying factor analysis (Daffertshofer, Lamoth, Meijer and Beek, 2004; Ivanenko et al., 2007) and by describing effector variance on a plane created in three-dimensional space (Borghese, Bianchi and Lacquaniti, 1996; Ivanenko, Cappellini, Dominici, Poppele and Lacquaniti, 2005). These methods have demonstrated patterns of coordination amongst the effectors, however, these methods do not have the ability for describing how coordination influences hypothesized movement outcomes. The uncontrolled manifold hypothesis is dependent upon the identification of a task goal because of the geometric relationship between a hypothesized end-effector (goal) and the effectors (body), and therefore what the 'controller' identifies as an important task variable might not be reflected in the geometric equation. Therefore, the uncontrolled manifold hypothesis has been used for identifying between a hypothesized task goal with an alternate goal that shares the same degrees of freedom, i.e. belonging to the same effector, but involving a different dimension (Scholz and Schoner, 1999; Black, Smith, Wu and Ulrich, 2007). The exploration of the variance structure of a UCM using a different selection of task goals in an equation allows stronger confirmation of the organized synergy among the effectors acting to satisfy a task goal.

The notion of synergies represented by the uncontrolled manifold hypothesis is in line with a controller selecting motor commands which reduce the task-relevant variance (Vbad) and maximize the task-redundant variance (V_{good}). The other notion is that a hierarchical controller doesn't need to be concerned about variance along the task-redundant dimension and any sensory information about the state of the task can be tuned in to address only the details of task-relevant parameters. This aspect of the uncontrolled manifold hypothesis was also applied by Todorov and Jordon (2002) for assessing a control model which linked sensory feedback with motor commands. They used the term Minimum Intervention Principle (Todorov and Jordan, 2002) because of the way it defined the workings of their closed loop feedback control model (introduced earlier in section 2.3.1.1.1).

The uncontrolled manifold has not been applied to the toe-swing task of walking. It is well suited to investigating trajectory control of the toe and how the coordination of lower limbs influences this performance. The application can also be applied to investigating this performance relationship when the toe trajectory is low, medium or high, and at different states of the swing trajectory. The geometric equations can also be applied to investigations of other contributing task goals, such as controlling the stance hip position by the stance limb, or representing the stance limb and swing limb as an effector system which is a two dimensional synergy with the toe clearance height. These are all new areas for the uncontrolled manifold hypothesis to explore. By exploring different effector systems and task goals, the insights into the central nervous system intentions for controlling the redundancy towards achieving a task goal can be more effective. For the benefit of understanding the concept of a task-relevant structure of the variance of the UCM, the minimum intervention principle adopted by Todorov (2002) will be adopted. This view assumes that the system has a hierarchical controller operating at two levels. The next section will show how a closed loop feedback control model can demonstrate how the variance is structured in the highly redundant locomotor system.

The benefit of reducing variance orthogonal to the UCM and allowing variance to persist parallel to the UCM could be related more so to the need for the system to perform multiple tasks simultaneously, rather than to minimize the end-point error/variability in an optimal way (Gera, Freitas, Latash, Monahan, Schoner and Scholz, 2010).

2.4.4.1.1 Serial correlations and uncontrolled manifold hypothesis

Serial correlations have been applied primarily with timing interval details of rhythmic human movement tasks, and they have not been a tool for investigating the temporal structural expressed within the uncontrolled manifold dimensions. The UCM hypothesis proposes that the variance parallel along the manifold is task redundant and therefore requires less regulated control. In the context of open loop and low level closed loop control scheme, the expected persistence in this dimension would be relatively high. In contrast, the persistence associated with the perpendicular component of the manifold would require the higher control levels of the closed loop system. Therefore, the degree of negative feedback due to intervention by the controller would be relatively frequent and not allow the correlations to persist to the same extent as expected to occur in the parallel dimension.

The application of long range persistence with UCM ‘type’ variance has only been applied to the ‘end-effector’ task of walking speed and the effector system was represented by step interval timing and step length (Dingwell et al., 2010). Walking speed was fixed using a treadmill protocol. Therefore walking speed was determined to be a regulated control task. The ‘effectors’ were considered task redundant. The results of persistence demonstrated anti-correlations for walking speed and long-range persistence for step length and step timing. When the effector system was plotted in a simple two-dimensional effector space, the task-redundant variance component and the task-relevant variance component were considered parallel and perpendicular to the manifold. Temporal correlations along the manifold were found to contain relatively higher persistence in comparison to the correlations perpendicular to the manifold (Dingwell et al., 2010).

2.4.5 Minimising tripping risk

Tripping risk will be minimized by an effector system which demonstrates functioning at all three levels of an optimal closed loop feedback control system. This is best achieved by optimally setting body sensors to tune in and receive the most relevant information to estimate the current body state, and to have this information matched to a set of 'ready to go' motor commands which produce task relevant movement requirements to attain a particular task goal (Li et al., 2005).

2.4.5.1 Assessing toe-swing performance and identifying markers of trip risk

To assess these components during walking in a thorough manner is a difficult task because of the embodied complexity of closed loop locomotor control. What can be observed in the movement details analysed is the outcome of all three components interacting and tracing insight back to these components using a framework of performance. The analysis tools of the uncontrolled manifold hypothesis and serial correlation analysis provide an understanding of how the central nervous system resolves redundancy and what tasks it considers important to maintain a vigil of regular control. The toe-swing task has not been explored from this approach and this thesis contributes to the first steps.

2.5 Aims of the thesis

The general aim of the thesis is to investigate the performance characteristics of the toe-clearance task during steady state walking and how performance characteristics change in older adults.

The specific research questions of this thesis relate to the performance of effector systems describing the swing limb, stance limb and toe trajectory across four studies:

- Study 1: Will statistical parameters of the three effector systems reveal age group differences?

This will be investigated from the following aims:

  - Determine if there is an <u>effector configuration</u> difference between age groups across selected toe-trajectory states of early-to-mid swing region
  - Determine if there are <u>segment configuration</u> differences between age groups across selected toe-trajectory states of early-to-mid swing region
  - Determine if there is a <u>control policy</u> difference in effector configuration between age groups across selected toe-trajectory states of early-to-mid swing region

- Study 2: Will serial correlation values, proposed to represent the effector systems dynamic control management, reveal differences between age groups?

This will be investigated from the following aims:

  - Determine if there is evidence that serial correlation values are state-dependent across swing cycles

- Determine if there is evidence that serial correlation values are effector-dependent across swing cycles
- Determine if there is evidence that serial correlation values are age-dependent across swing cycles

♦ Study 3: Will the uncontrolled manifold hypothesis reveal differences between age groups?

This will be investigated from the following aims:

- determine if there is evidence of variability management by the redundant stance effector at events of the swing phase;
- determine if there is evidence of variability management by the redundant swing effector at events of the swing phase;
- determine if there is an age difference in the way variability is managed by the stance and swing effectors;
- determine if variability management is trajectory-dependent (i.e. trajectories defined by initial conditions at MX1 and subsequent response made at MTC);
- determine if ageing differences exist in variability management.

♦ Study 4: Will the stance and swing effectors cooperate differently between age groups?

This will be investigated from the following aims:

- determine if there is evidence of cooperation between redundant stance and swing effectors for managing toe trajectories during the swing cycle;
- determine if either the stance or swing effector is predominantly responsible for causing a change in toe trajectory.

Chapter 3

Methodology

3.1 Participants

There are two major reasons for limiting this study population to female participants. The first is to control for gender differences. Second, elderly females have a higher falls incidence compared to males (Lord et al., 2007; Kreisfeld and Harrison, 2010). Twenty eight healthy young adults between the ages of 18 - 35 years (mean age 25.2 $\pm$ 5.9 years) and 28 healthy older adults aged 65 years and older (mean age 68.9 $\pm$ 3.9 years) participated in the study. The sample size of twenty eight was found to be the cut-off number from a power analysis comparing mean toe clearance variability between healthy young and healthy elderly groups with sample size of ten.

Young healthy adults were randomly selected from within the university community. Healthy older adults were recruited from within the community of Melbourne by advertisement in local 'seniors newspaper' that had a high readership index. Interested volunteers made contact by phone and a database of interested older adults was collated. The initial phone conversation included a structured screening evaluation about the participant's general health status. The information collected from participants over the phone included the following: age; free of any neural/muscle/skeletal diseases likely to affect walking; list type of medications used; list type of activities that posed problems or pain; community dwelling and carried out an independent lifestyle; whether able to walk independently for 20-30 minutes without walking aid; and any history of falls. Participants were also notified that to gain official consent, they would require the signed approval of their General Practitioner, who would be asked to assess the participant's health status. After collating the database, a group of 'healthy' participants were contacted by phone to discuss the study details and further clarify the inclusion criteria and the nature of the testing protocol. Participants were then forwarded via mail the documentation of the study protocol and exclusion criteria, with an accompanying letter of consent for their GP to approve and sign. Their General Practitioner was asked to review the listed 'healthy' ageing criteria to determine whether their client met these conditions (see Appendix A and B) and to also consider the nature of the test protocol conditions before signing approved consent for their client to participate.

The specific conditions for GP assessment were: requirement of a walking aid; displayed clinical cardiopulmonary, neurological, musculoskeletal, or orthopaedic conditions; not living independently within the community; difficulty performing daily living activities; signs of dementia or cognitive disease; severe depression; medications affecting their balance and gait; uncorrected vision impairments; a history of falls;. Participants that were deemed within the healthy population criteria, by the GP, returned the GP-authorized evaluation form in the mail to be received by the 'investigator' prior to testing day. Similar basic screening methods have been applied in other studies to obtain a healthy sample of the older adult population (Herman, Giladi, Gurevich and Hausdorff, 2005; Kang and Dingwell, 2008a; Watt, Franz, Jackson, Dicharry, Riley and Kerrigan, 2010).

In addition to the GP evaluation and self-reporting from phone conversations, the elderly volunteers underwent additional screening of physical and mental functioning on the day of testing. The purpose was to provide further confidence that the selected sample of elderly participants were representing a healthy ageing cohort. Participants completed an abbreviated mental test score (AMTS) (Hodkinson, 1972); Visual Contrast Sensitivity (VCS) using the Melbourne Edge test (Verbaken and Johnston, 1986); a timed up and go test (TUGT) (Whitney, Lord and Close, 2005); a modified falls efficacy scale questionnaire of 14 items (MFES) (Hill, Schwarz, Kalogeropoulos and Gibson, 1996). From a group of 31 volunteers presented, a group of 28 participants were selected because they achieved the following cut-off scores: above 7 for the AMTS; above 17dB for the VCS test; below 11 seconds for the TUG test; above 9 for the MFES.

3.2 Experiment protocol

3.2.1 Participant characteristics

All participants were measured for height and mass. Height was measured according to the 'stretch stature protocol' described by Bloomfield et al. (1994). Body mass was measured on a force platform. The weight of the external materials carried by each participant included a light-weight backpack and a connection box with NDI® strober devices (Figure 3.2.1.1). This was attached to a harness supported by a spring loaded retractable pulley which reduced the load effect but negligible in terms of creating an un-weighting effect on the participant's body. Dominant mobilizing limb was determined by the participant's preferred leg when kicking a ball (Hart and Gabbard, 1998).

Figure 3.2.1.1

Experimental set up. Backpack of NDI strobe connection boxes which are supported by spring loaded harness device. Camera set up is described in detail in section 3.3.1. Marker set up is described in detail in section 3.3.2. All subjects were given the same set of clothing but participated with slight variations of rubber-soled footwear (see section 3.2.2).

3.2.2 Participant preparation

Participant preparation took approximately 60 minutes to complete. Participants were fitted with 'non-slip/anti-migration' lycra exercise shorts and long sleeved T-shirt available in five different sizes to cater for body shape differences. Participants brought their own footwear, the only condition being that they be rubber-soled comfortable walking shoes. The fitting of the markers and EMG took approximately 40 minutes to complete. A static anatomical landmark (AL) calibration procedure was then administered to collect relevant ALs using NDI digitizing probe and NDI ToolBench software (described in section 3.3.2). The AL calibration procedure took approximately 20 minutes to complete. This was followed by the treadmill warm up period which is described in the next section.

3.2.3 Walking test

A treadmill walking warm up period preceded the experiment data collection protocol. There were three walking protocols prescribed to all participants. In all testing and pre-testing conditions, participants walked wearing their own rubber soled shoes and were required to walk swinging their arms naturally without the use of hand rails. Participants were freed from all distractions during walking tests. Prior to any testing taking place participants were asked to familiarize with the treadmill for between 8-12 minutes. During this period participants explored different walking speeds between 1.5 km/h and up to 6.0 km/h, and were also encouraged to explore different step frequencies and step lengths. This was to minimize unfamiliar sensory effects when walking on a stationary treadmill.

3.2.3.1 Treadmill walking test

A 10 minute bout of walking, at a constant treadmill speed, was set at the participant's preferred walking velocity. The preferred walking speed (PWS) was defined by the participant after undertaking a speed adjustment procedure that determined when walking was uncomfortable. Each participant was asked to imagine "walking by yourself to a destination". The treadmill speed increased from an initial set speed equal to a proportion of their Froude velocity. [NB: The Froude velocity is a walking speed derived from limb length and gravity, where 0.42 of Froude velocity has been found to be corresponding to human preferred walking speed (Cavagna, Heglund and Taylor, 1977; Kram, Domingo and Ferris, 1997)]. The treadmill speed increased gradually until the participants reported the speed had began to be uncomfortably fast in relation to their perceived task. The 'concealed' speed of the treadmill then increased with a constant change in velocity until they perceived the treadmill speed being uncomfortably slow. This was repeated in descending speed until the speed was perceived to be uncomfortably slow. The PWS was determined from the average of three 'uncomfortably fast' speeds and three 'uncomfortably slow' speeds. A similar approach has been applied elsewhere by Dingwell and Marin (2006), and England and Granata (2007).

3.3 Measuring gait kinematics

An accurate three-dimensional kinematic analysis of human walking is dependent upon undertaking methodological steps which minimise errors when reconstructing a 3D (three-dimensional) model. The minimisation of errors address two general areas of the data collection procedure: 1) a method to accurately estimate a biomechanical model which describes the instantaneous body segment details in spatial coordinates (see section 3.3.2); and 2) an accurate motion capture measurement system (see section 3.3.1).

A kinematic model of inter-linking bone segments is reconstructed in 3D spatial coordinates from noisy body-mounted markers which are designed to represent the motion of the underlying bone. Unfortunately, the reconstructed model is only an estimate because the surface marker trajectories cannot precisely mimic the motion of the underlying bone segments because of the overlaying soft tissue deformation (e.g. skin sliding and muscle wobble). For the same reason, the exact description of the anatomical landmarks which represent the bone segment end-points and inter-segment joint centres in the reconstructed model are also only an estimate. Both anatomical landmark mis-location and soft tissue artefact are the major contributors to kinematic modelling errors (Cappozzo, Della Croce, Leardini and Chiari, 2005; Della Croce, Leardini, Chiari and Cappozzo, 2005; Leardini, Chiari, Croce and Cappozzo, 2005).

There are numerous methodological procedures proposed to minimise reconstruction errors when recreating a 3D kinematic model from stereo-photogrammetry and surface markers (Cappozzo, Cappello, Della Croce and Pensalfini, 1997; Lu and O'Connor, 1999; Charlton, Tate, Smyth and Roren, 2004; Della Croce et al., 2005; Leardini et al., 2005). However, there is yet to be a method for completely overcoming all errors. The current study took the pragmatic approach by applying methods which considered the thesis aims and the error-contributing issues. The outline of this approach will be detailed through the remainder of this section.

3.3.1 Instrumentation and lab set-up

This section describes the set-up of the lab for the walking protocols and the measurement system used.

The best performing non-invasive measurement systems for capturing kinematic data is stereo-photogrammetry. A nine camera motion capture system was set up around the treadmill. There were three NDI (Northern Digital Inc., Waterloo, ON, Canada) optoelectronic position sensor 'towers' positioned bilateral to and behind the participant's walking progression. The camera system was composed of two camera tower models from NDI; two towers were OPTOTRAK CERTUS® (Far Focus) and the other was an OPTOTRAK 3020®. Each position sensor 'tower' contained three cameras which are set in a rigidly fixed array and calibrated at the NDI factory.

The markers were tracked from active strobing of infrared light emitting diode markers (iREDs) using First Principles software (NDI). The three-dimensional coordinates of the tracked markers were relative to a three-dimensional LAB-based reference system (i.e. lab coordinate system LCS). Each iRED marker was 11mm in diameter. Both the CERTUS and 3020 units have been reported to have the same level of resolution accuracy, with a maximum RMS accuracy (at 2.25 meter distance from camera) of 0.10mm for 2D motion, and 0.15mm for 3D motion (http://www.ndigital.com/research/certus-techspecs.php). The CERTUS system has maximum accuracy within a volume of 2.2 x 2.7 x 1.7 m^3, positioned 2.25 meters away from camera. During camera registration procedures performed across separate testing days, the camera system generally returned an RMS error value of between 0.20mm and 0.35mm.

3.3.1.1 Sampling rate

Most of the kinematic details of gait are within a signal frequency of 6Hz (Kirtley, 2006; Winter 1974). The minimum required sampling rate normally recommended for evaluating gait kinematics is at least 50Hz (Winter, 2005). This current study applied the maximum sampling rate available from the camera system which was 70Hz for motion tracking iRED markers. The maximum was set because the CERTUS system sampling frequency gets reduced when the 3020 model is co-connected in the system. Generally, both the Optotrak camera systems have a sampling rate determined by the marker frequency and the number of markers connected. For ‘n’ markers, and ‘m’ marker frequency, the maximum sampling rate is determined by: f = m / (n+2). This formula comes from the way NDI sequentially ‘strobe’ each iRED marker within a sampling frame. The time interval within a sampled frame limits the number of sequentially firing markers. The CERTUS system has a maximum marker frequency of 4600, however when connected in series with the 3020 model, the maximum marker frequency, m, is limited to 2500. Therefore, when n = 33 markers are employed, the available sampling rate is restricted to 70Hz [i.e. f = 2500 / (33+2)].

The sampling rate of the NDI motion capture system is off-set by the employed number of tracking markers. This presented an issue for defining the number of body segments to be analysed in this study. By including the trunk segment in this study meant that there was a decision to represent the foot as one rigid segment rather than two. This has limitations for accurately describing the distal-inferior surface of the foot. Essentially, the foot segment is non-rigid because multiple segments can move within the foot. An analysis of the distal-inferior surface of the foot was conducted in section 3.6.

Because the firing rate of the active iRED strobe markers acts in succession, and 33 markers were used, the NDI First Principles applies a time interpolation algorithm to transform the position coordinates so that within a sampled frame of data the marker positions are aligned at a simultaneous instant in time. This algorithm is patented NDI intellectual property which is not released for public knowledge.

3.3.1.2 Marker filtering

Tracking the surface markers on the body is affected by two types of random noise: 1) system capture noise and 2) digitising noise. Unfortunately, the frequency of the overlaying soft-tissue is of a similar frequency to the bone movements and cannot be considered random noise to be filtered.

The issue of attenuating the effect of random measurement noise on iRED marker position tracking was investigated using a residual analysis method described by Winter (2005). The analysis investigated various cut off frequencies between 1-20Hz and using a fourth order, single pass Butterworth filter. Eight participants were assessed for walking at both slow and fast speeds. The outcome of this analysis resulted in the optimal determination of cut off frequencies for the head and trunk at 6 Hz, while the pelvis and lower limbs were 7 Hz. These findings were subsequently applied during the marker filtering process (see section 3.4). Kepple, Siegel and Stanhope (1997) also applied a 4th order Butterworth filter with cut off frequencies to the foot (6Hz), leg (5Hz), thigh (4Hz) and pelvis (3Hz) for a 50Hz camera system. However, their early model passive marker motion capture system is recognized as a considerably less accurate system compared to the active marker system in this study (Durlach and Mavor, 1995). The cut off frequencies applied in this study are similar to other studies investigating kinematics of walking which also used the OPTOTRAK motion capture system: 7Hz cut off at sampling rate of 80Hz for full body marker set (MacLellan and Patla, 2006); 6Hz cut off at sampling rate of 60Hz for lower limbs marker set excluding foot segment (Houck, Yack and Cuddeford, 2004); 7Hz cut off at sampling rate of 50Hz for foot markers (Begg et al., 2007); and 5Hz cut off at sampling rate of 50Hz for full body marker set excluding foot segment (Bruijn, Meijer, van Dieen, Kingma and Lamoth, 2007).

3.3.1.3 Lab set up

All participants were tested at the Victoria University Biomechanics Lab. The lab set-up is illustrated in Figure 3.3.1.3.1. There were three NDI (Northern Digital Inc., Waterloo, ON, Canada) optoelectronic position sensor 'towers' positioned bilateral to and behind the participant's walking progression. Each position sensor 'tower' contained three cameras which are set in a rigidly fixed array and calibrated at the NDI factory. The tester operated the control station during data collection. The control station included NDI First Principles operating platform software.

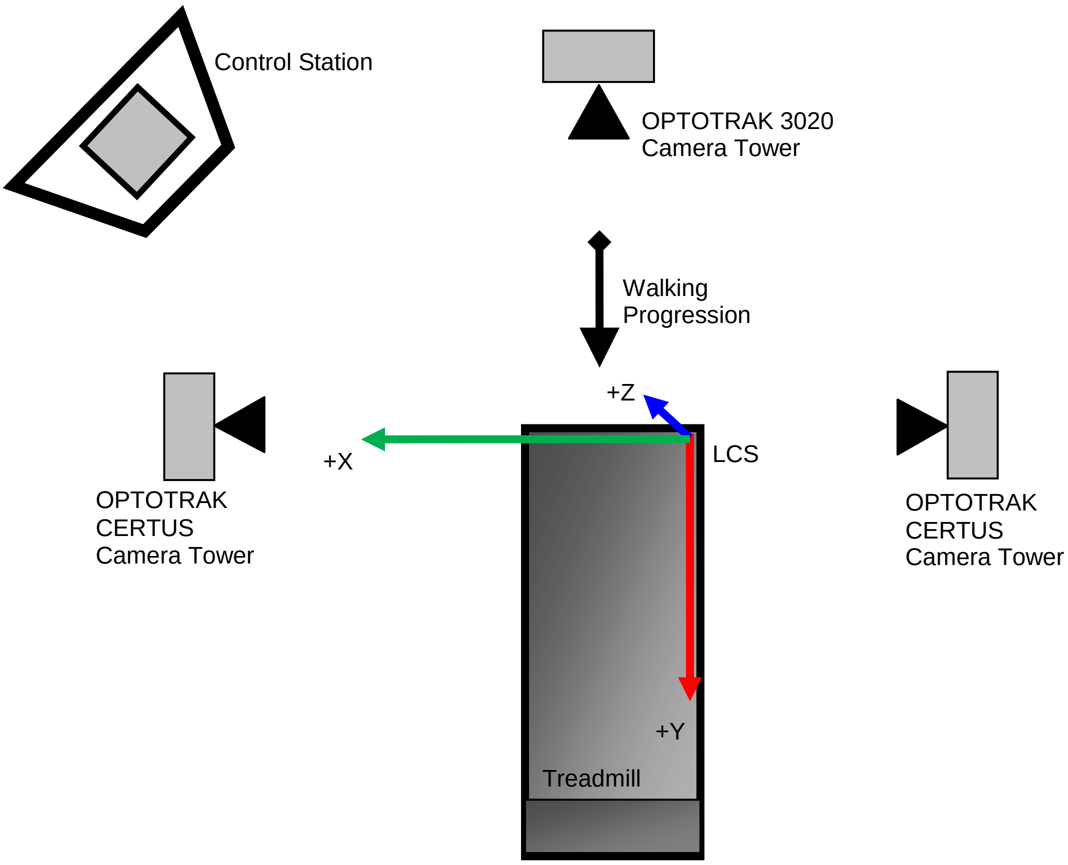


Figure 3.3.1.3.1

Lab and equipment set up. Transverse view of the lab set up. Three OPTOTRAK position sensor towers were used: one 3020 tower behind; and two CERTUS towers lateral to the treadmill. Each position sensor tower contained 3 cameras. The walking progression is in the direction of the +Y axis.

3.3.1.3.1 Global Reference System

The position of a lab coordinate system (LCS) was fixed with the treadmill deck. The LCS was orthogonal and corresponded to a right hand rule orientation (recommended for Visual3D software users). The origin was located at the posterior left corner of the belt surface (Figure 3.3.1.3.1) and the XY plane was level with the treadmill surface. Positive Y was in the direction of walking progression (posterior-to-anterior), positive X was directed from left-to-right relative to the direction of walking progression, and positive Z directed upward (Figure 3.3.1.3.1).

3.3.2 Reconstructing a biomechanical model

This section outlines how the three-dimensional biomechanical model was reconstructed from a photogrammetric measurement system which tracked the positions of body mounted markers. The approaches taken to minimise surface marker error addressed three areas: 1) marker set-up strategy to minimise the effect of soft-tissue artefact; 2) procedure to accurately identify inter-segment axes of rotation, or joint centre locations; and 3) algorithms which accurately estimate the six-degrees-of-freedom segment POSE (i.e. 3D position and 3D orientation) through optimisation.

Rather than use a hierarchical-type model (e.g. conventional gait model), the current study modelled the pose of each individual body segment independently (6-DoF model). The benefit of a 6-DoF model makes no a priori kinematic assumptions about how the segment should behave and is not subject to a cascade of marker error effects which can propagate across segments when applying conventional gait models (Buczek, Rainbow, Cooney, Walker and Sanders, 2010). The 6-DoF approach was somewhat determined by the constraints of the NDI measurement system, whose active marker sets require adjoining wires, and also the iRED markers need to project in a relatively direct line towards the camera towers (section 3.3.1). However, these constraints did not limit a 6-DoF marker set up procedure which was based upon marker clusters attached to rigid plates, and in-turn these plates were attached to the skin of body segments.

The procedure for reconstructing the 3D biomechanical model was to first determine an arbitrary technical frame from the cluster of markers attached to body segments. This marker cluster is termed a cluster technical frame. Using a calibration anatomical system technique [CAST; (Cappozzo, Catani, Croce and Leardini, 1995; Benedetti, Catani, Leardini, Pignotti and Giannini, 1998)] local landmarks relative to the cluster technical frame provided the details for a transformation matrix which re-aligned the segment into a meaningful anatomical frame. Landmarks were based upon superficial and internal methods; palpable bony prominences and functional computations of joint centres and axes of rotation (Schwartz and Rozumalski, 2005) respectively. The anatomical frame of each segment was modelled to have six degrees-of-freedom (6-DoF) using procedures in Visual3D (C-Motion Inc., Rockville, USA), and this allowed a standard way to make inter-participant comparisons of body kinematics.

During dynamic movement tasks like gait, the effect of soft tissue movement becomes influential. Three general methods exist for minimising the affect of soft tissue artefact caused segment marker errors on the 3D segment pose: direct pose estimation; segment pose optimisation (Veldpaus, Woltring and Dortmans, 1988); and global pose optimisation (Lu and O'Connor, 1999). The global optimisation method was designed to minimise the shortcomings of hierarchical type models. Because the current study estimated the anatomical frame from the independent 6-DoF model, the global optimisation method was not considered necessary.

3.3.2.1 Tracking segment motion

NDI ToolBench software was used for instantaneous tracking of thirty-three iRED markers attached to body segments to eventually obtain 6-DoF segment motion. Segments were tracked using an array of markers rigidly clustered on a body mounted plate (Manal, McClay, Richards, Galinat and Stanhope, 2002). These plates were attached to the feet, shank, thigh and pelvis. Table 3.3.2.1.1.a-b identifies the location of the tracking markers and this is also illustrated for body locations (Figures 3.3.2.1.1-2).

Figure 3.3.2.1.1.

Marker set up. Various views (A-F) describe the marker set up: A) frontal view; B) to minimize subject distraction the wires were passed behind the subject to a control box supported by the harness; C) sagittal view; D) cluster-technical-frame plate fastened to the shank; E) cluster-technical-frame attached to the foot; F) cluster-technical-frame fastened to the pelvis.

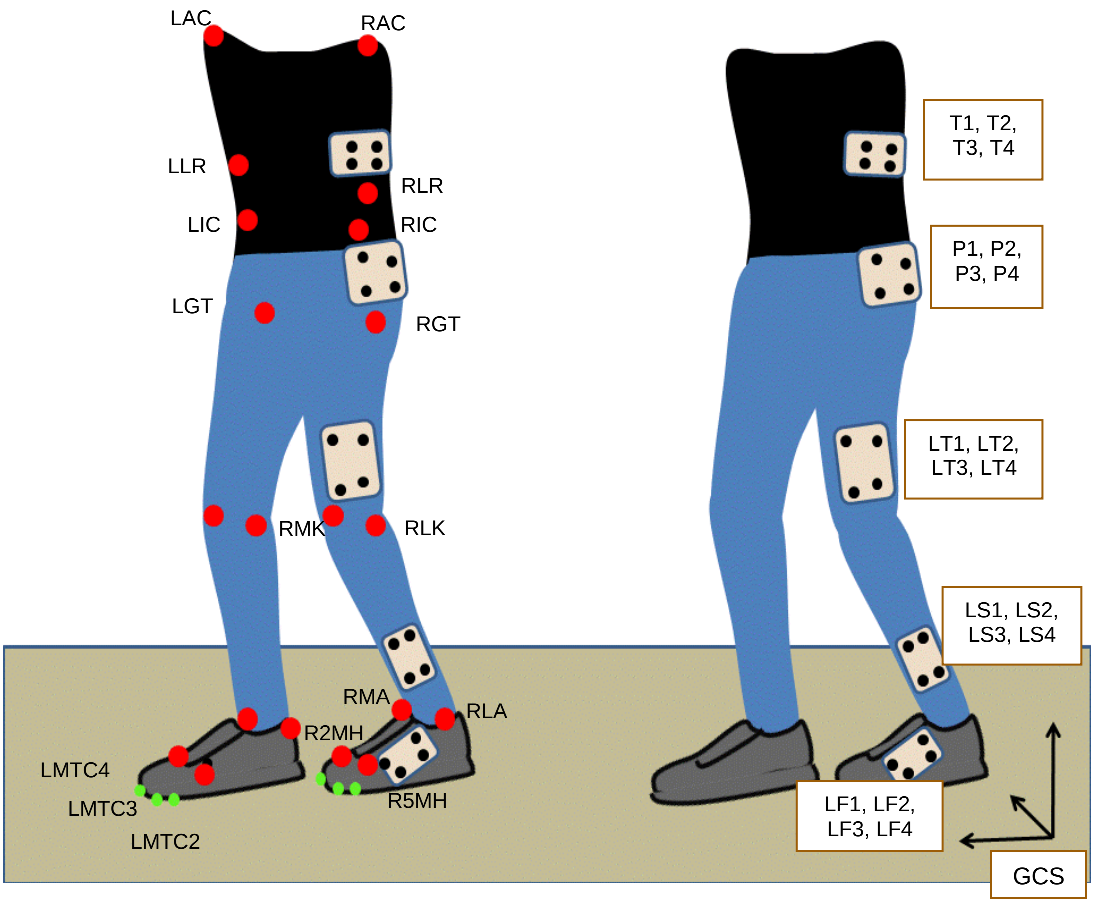


Figure 3.3.2.1.2

Animation of cluster markers and virtual markers. Note: the pelvis cluster here is displayed to the side of the pelvis for illustration purposes. Left: The segment coordinate systems are positioned and oriented relative to anatomical landmarks (in red) at the distal and proximal endpoints of the segment by a combination of virtual markers and functional joint centre computation. The virtual points representing the part of the shoe, which gets closest to the ground during mid swing are indicated in green. Right: Tracking cluster markers are located on rigid plates, which are fastened onto body segments: posterior surface of trunk and pelvis; lateral surface of left and right lower limbs. These cluster technical frames track the segments. All abbreviations in are defined in Tables 3.3.2.1.1a and 3.3.2.1.1b.

Table 3.3.2.1.1a Specific marker locations (head to hip)

Segment	Description
Head	
HD	Head target marker (lateral to occipital process)
Thorax	
Target Markers	
T1 T2 T3 T4	IRED target marker on Trunk Cluster Plate located between T11 and L2
Virtual Markers	
LTAC (RTAC)	Left Acromion process ***
LTR (RTR)	Left Lateral aspect of last rib ***
Pelvis	
Target Markers	
P1 P2 P3 P4	IRED target marker on Pelvis Cluster Plate located between L5 and S2
Virtual Markers	
LPSIS (RPSIS)	Posterior Superior Iliac Spine ***
LASIS (RASIS)	Anterior Superior Iliac Spine ***
LIC (RIC)	Iliac Crest ***
LGT (RGT)	Greater Trochanter ***
RHIP _FUNC (LHIP_FUNC)	Hip joint centre *
LIC_lm (RIC_lm)	Iliac Crest ***

* - virtual anatomical landmark (Schwartz et al. 2005)

** - projected virtual anatomical landmark

*** - virtual anatomical landmark located manually

Table 3.3.2.1.1b Specific marker locations (thigh to toe)

Segment	Description
Thigh	
Target Markers	
LT1 (RT1) LT2 (RT2) LT3 (RT3) LT4 (RT4)	IRED target markers on Thigh Cluster Plate located at distal part of segment
Virtual Markers	
LLK (RLK)	Lateral femoral epicondyle ***
LMK (RML)	Medial femoral epicondyle ***
LKnee_FUNC (RKnee_FUNC)	Landmark #1 along knee joint axis *
LKnee_FUNC_X (RKnee_FUNC_X)	Landmark #2 along knee joint axis *
LLK_lm (RLK_lm)	True lateral femoral epicondyle **
LMK_lm (RMK_lm)	True medial femoral epicondyle **
Shank	
Target Markers	
LS1 (RS1) LS2 (RS2) LS3 (or RS3) LS4 (or RS4)	IRED target marker on Shank Cluster Plate located at distal part of segment
Virtual Markers	
LLM (or RLM)	Lateral malleolus ***
LMM (or RMM)	Medial malleolus ***
Foot	
Target Markers	
LF1 (or RF1) LF2 (or RF2) LF3 (or RF3) LF4 (or RF4)	IRED target marker on Foot Cluster Plate located at lateral &medial part of segment
Virtual Markers	
LMTC1 (or RMTC1)	Minimum toe clearance point1 ***
LMTC2 (or RMTC2)	Minimum toe clearance point2 ***
LMTC3 (or RMTC3)	Minimum toe clearance point3 ***
LMTC4 (or RMTC4)	Minimum toe clearance point4 ***
L5MH (or R5MH)	5th metatarsal head ***
L1MH (or R1MH)	1st metatarsal head ***
LH1 (or RH1)	Posterior aspect of calcaneus ***
LH2 (or RH2)	Inferior and posterior aspect of shoe out-sole surface ***

* - virtual anatomical landmark (Schwartz et al. 2005)

** - projected virtual anatomical landmark

*** - virtual anatomical landmark located manually

The rigid plates were custom made from lightweight thermoplastic and moulded to the shape of the segments in which they were attached. The pelvis, thigh and shank cluster plates were positioned at areas to minimise soft tissue artefact. The body segments were circumference wrapped with anti-migration neoprene rubber (inner underlay with adhesive spray) with Velcro on the outer surface. The cluster technical frame plates were also fitted with Velcro material on the inner surface and were mated onto the neoprene bands. The plate-attached Velcro loop underlay was matted onto a body segment-attached Velcro loop. The neoprene underlay wrap served to also restrict soft tissue wobble (Manal, McClay, Stanhope, Richards and Galinat, 2000).

The cluster technical frames were defined by the four cluster markers of the respective segments (Table 3.3.2.1.2). The construction of the cluster technical frame plate design considered the recommendations proposed by Cappozzo et al. (1997) and the pelvis cluster technical frame plate attachment was an exception case. There are several methods which have been applied to describe the motion of the pelvis using a cluster technical frame (Benedetti et al., 1998; Whittle and Levine, 1999; Vogt, Portscher, Brettmann, Pfeifer and Banzer, 2003). The design of the pelvis cluster technical frame in this study considered a pragmatic approach for female participants while considering cluster technical frame plate design recommendations by Cappozzo et al., (1997). Further steps were taken to ensure the pelvis cluster technical frame could mimic the pelvic motion by designing lycra shorts with rubber underlay to prevent material migration across the skin and a large Velcro patch sewed across the buttocks for the plate attachment site. The plate was secured onto the Velcro patch against the posterior superior iliac spine by wrapping a non-slip neoprene belt at the level of the anterior superior iliac spine. This is similar to the ‘Milwaukee Brace’ design (Benedetti et al., 1998) in the sense that the current study also used a band to attach a rigid plate on the pelvis from which a cluster technical frame was embedded.

Table 3.3.2.1.2 Cluster technical frame definitions.

Segment	Description
Thorax	
Origin	T1
Vertical (z) axis	Line in direction of T2 to T1
Mediolateral (x) axis	Perpendicular to vertical axis, in plane containing T1, T2 and T3
Anterior-posterior (y) axis	Mutually perpendicular to other 2 axes
Pelvis	
Origin	P1
Vertical (z) axis	Line in direction of P2 to P1
Mediolateral (x) axis	Perpendicular to vertical axis, in plane containing P1, P2 and P3
Anterior-posterior (y) axis	Mutually perpendicular to other 2 axes
Thigh	
Origin	RT1 (same for left side)
Vertical (z) axis	In line with RT1 and RT2, direction away from RT2
Mediolateral (x) axis	Perpendicular to vertical axis, in plane containing RT1, RT2 and RT3
Anterior-posterior (y) axis	Mutually perpendicular to other 2 axes
Shank	
Origin	RS1 (same for left side)
Vertical (z) axis	In line with RS1 and RS2, direction away from RS2
Mediolateral (x) axis	Perpendicular to vertical axis, in plane containing RS1, RS2 and RS3
Anterior-posterior (y) axis	Mutually perpendicular to other 2 axes
Foot	
Origin	RF1 (same for left side)
Vertical (z) axis	In line with RF1 and RF2, direction away from RF2
Mediolateral (x) axis	Perpendicular to vertical axis, in plane containing RF1, RF2 and RF3
Anterior-posterior (y) axis	Mutually perpendicular to other 2 axes

3.3.2.1.1 Justification of cluster technical frame marker set up

The major advantage of using marker clusters on a plate is that it avoids the need to place markers at the joints which is where markers are most susceptible to artefacts from skin sliding (Cappozzo, Catani, Leardini, Benedetti and Croce, 1996). In practice, this relative motion can be up to 20mm in the lower limbs (Fuller, Liu, Murphy and Mann, 1997).

The attenuation and characterization of soft tissue artefact within individuals is an extremely difficult problem to solve because the frequency content of soft tissue artefact is similar to bone (Fuller et al., 1997). This makes it difficult to use harmonic analysis or other filtering techniques to compensate for its unwanted effect. The most affected knee joint motion is associated with abduction/adduction and internal/external rotation, while motion in the sagittal plane (flexion/extension) is less affected by soft tissue artefact (Reinschmidt, van Den Bogert, Murphy, Lundberg and Nigg, 1997). The effects of soft tissue artefact are also dependent upon the specific motor task undertaken (Cappozzo et al., 1996), such that muscle contraction, skin sliding and tissue deformation, gravity and inertial effects will all contribute independently. Soft tissue artefact errors have shown to be larger in running compared to walking (Reinschmidt et al., 1997).

In consideration of the results and the limitations expressed in the literature for analysing three-dimensional segment kinematics, the marker set up approach adopted in this investigation has considered a balanced approach to model reconstruction. The application of cluster technical frame plates, like those adopted in this study, has been performed in many biomechanical studies (Seay, Haddad, van Emmerik and Hamill, 2006; Bruijn et al., 2007) and the advantage of this marker set up technique relative to other methods has been strongly supported in terms of relative validity and pragmatism (Holden, Orsini, Siegel, Kepple, Gerber and Stanhope, 1997; Benedetti et al., 1998; Manal et al., 2000). The segment location of the cluster technical frame plates considered recent evidence which described the distal portion of the thigh and shank to contain relatively less soft tissue artefact in comparison to remaining sections of these segments (Stagni, Fantozzi, Cappello and Leardini, 2005).

3.3.2.2 Anatomical landmarks

Anatomical landmarks were listed in Table 3.3.2.4.1. This section describes how the anatomical landmarks were obtained.

3.3.2.2.1 Protocol for hip and knee functional tasks

Internal anatomical landmarks were obtained using a functional procedure (Schwartz and Rozumalski, 2005). To obtain the hip and knee landmarks which define the hip joint centre and the knee axis of rotation, a functional movement task was performed by all participants (Camomilla, Cereatti, Vannozzi and Cappozzo, 2006). The movements for computing the hip were active hip flexion/extension, adduction/abduction and circumduction. Participants were instructed to keep the upper trunk as steady as possible and to execute smooth movements. For computing the knee axis the tester flexed and extended the participant's knee passively while the participant stood on one leg holding a hand rail for support.

3.3.2.2.2 Justification for functional landmarks

Stagni et al. (2000) investigated data distortion affecting the 3D hip and knee angles and moments due to hip joint centre location error. They found that hip joint centre errors have only a small proportionate effect on the range of the signal magnitude for the hip angles (<5% of signal range), knee angles (~1.5% flexion/extension, <7% abduction/adduction and internal/external rotation, of signal range) and moments (<3% of signal range). The largest effect was upon the hip moments (~20% range).

Other studies have also reported that anatomical landmark mislocation has a large effect on the relative degree of error associated with minor rotations out of the primary joint rotation and interpretation of joint angles should be restricted to the major joint angles (i.e. plane detailing the largest range of motion specific to joint) until a standard method is established for reporting on minor angles (Della Croce et al., 2005). Identifying the joint centre and axis of rotation by improving anatomical landmark location procedures, are thus relevant for reducing inaccuracies of the anatomical frame and minimising joint angle errors.

Ehrig et al. (2006) conducted an independent analysis of currently available methods for defining the hip joint centre locations, such as regression methods (e.g., Campbell et al., 1990), sphere fitting methods (e.g., Leardini, Cappozzo, Catani, Toksvig-Larsen, Petitto, Sforza, Cassanelli and Giannini, 1999) and transformation techniques (e.g., Schwartz and Rozumalski, 2005). The analysis was performed on simulated thigh and pelvis segment anatomical frame motion typically reconstructed from a four marker cluster technical frame. The known joint centre location was compared against the computed joint centre of rotation, which included modelling noise affects on the cluster technical frames caused by soft tissue artefact. When the analyses were performed under both limited and full range of joint motion, as well as various soft tissue artefact effects, Ehrig et al. (2006) found that the Schwartz and Rozumalski (2005) method compared best.

Schwartz and Rozumalski (2005) found hip joint centre mean differences between methods indicating that the functional method is 5.8mm anterior, 15.1mm lateral and 16.6mm superior compared to regression methods that employ anatomical landmark palpation techniques. If the knee flexion/extension axis is incorrectly defined then 'cross-talk' errors will propagate both proximally to give imprecise hip joint rotation (i.e. high SD) and distally as excessive knee varus/valgus range of movement. The knee axis of rotation defined by the functional method produced significantly less varus/valgus and significantly more flexion/extension range of movement during walking ($p<.001$).

Because of anatomical landmark mislocation error effects that bony prominences do not coincide with the joint centre, functional anatomical landmarks were used to estimate the hip joint centres and knee axis-of-rotation within the Visual3D environment using methods developed by Schwartz and Rozumalski (2005). The functional tasks performed and captured using NDI was exported into Visual3D. The existing anatomical medial and lateral epicondyles of the knee, from palpated probe collection, were projected from within their sagittal plane of location onto the unit vector defined by the result of the functional knee axis of rotation. This created functionally derived anatomical landmarks forming the medial and lateral distal ends of the thigh and proximal ends of the shank. The anatomical method was used to define the ankle joint from the medial and lateral malleoli and the metatarsal-phalangeal joint from 2nd and 5th metatarsal heads. The phalanges were not included as a segment in the link model. Functional hip joint centre landmarks were used to define the pelvis segment with anatomical landmarks at the iliac crest.

3.3.2.3 Static calibration trial

A static standing trial was recorded for one second while the participant stood in the anatomical position with feet at shoulder width and the longitudinal axis of the feet was aligned parallel with the Y axis. The standing posture was used for building the biomechanical model and served as the zero reference for segment angles.

3.3.2.4 Anatomical frame

The procedure for collecting the NDI data for Visual 3D input is described in the following section. Palpable anatomical landmarks relative to the respective segment-cluster technical frame were stored as virtual markers using the ‘imaginary marker’ procedure in NDI First Principles. The tip of the digitising probe was positioned on the palpated anatomical landmarks. This tip position relative to the probe cluster technical frame has an accuracy of < 1mm. NDI First Principles was configured to capture simultaneously the superficial anatomical landmarks (virtual markers) and cluster technical frame markers (tracking markers) for all trials. Figure 3.3.2.4.1 illustrates the general reconstruction process for the pelvis and the thigh segment. Superficial anatomical landmark identification was performed by the same tester for all participants and these anatomical landmarks are listed in Table 3.3.2.4.1-2.

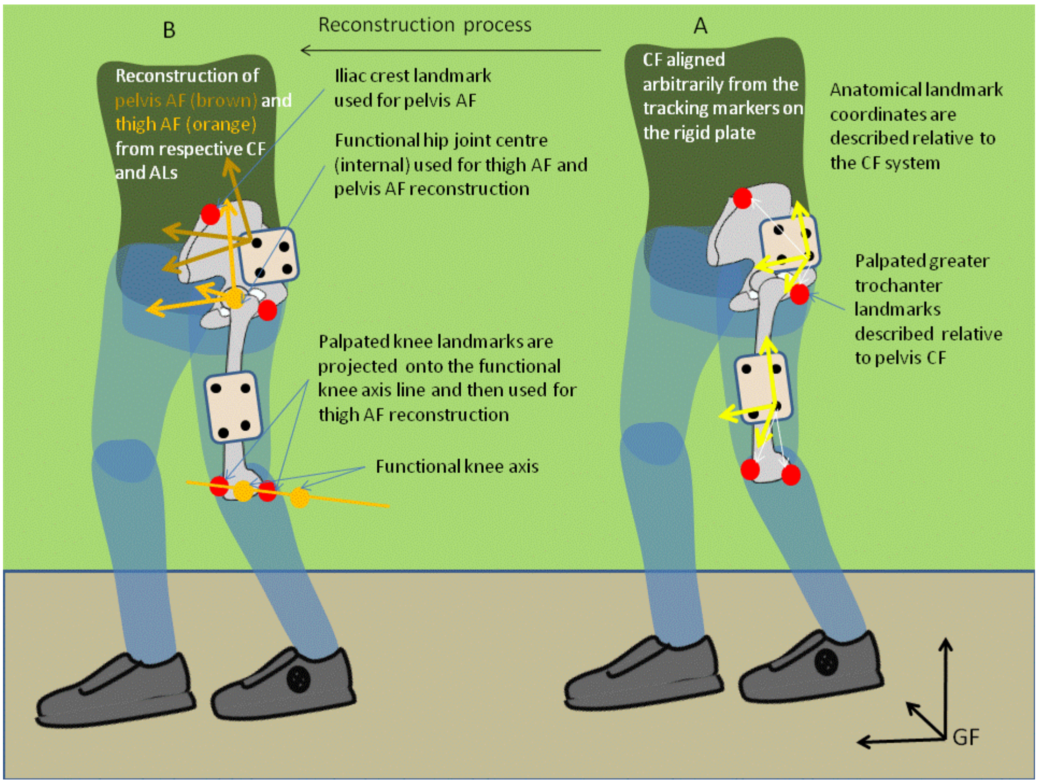


Figure 3.3.2.4.1

General process for anatomical frame (AF) reconstruction for the pelvis and thigh from their respective cluster technical frame (CF) system (yellow axes) and associated anatomical landmarks (red points). The origin of the pelvis anatomical frame (brown axes) is located at the mid-point of the line joining the proximal left and right iliac crest. The origin for the thigh anatomical landmarks (orange axes) is at the functional hip joint centre. The global reference frame (GF) is located at the bottom.

Table 3.3.2.4.1 Anatomical Frame Definitions (thorax to thigh)

Segment	Description
Thorax	
Origin	Midpoint between LIC_lm and RIC_lm
Vertical (z) axis	Line between two mid-points created by a) LIC_lm and RIC_lm mid-point, and b) LAC and RAC midpoint. Unit vector is directed from proximal to distal segment.
Anterior-posterior (y) axis	Least squares approach to firstly determine the ZX plane containing the four segment end points: proximal landmark points LIC_lm, RIC_lm and segment distal landmark points LAC, RAC. The Y axis is perpendicular and directed anterior from this ZX plane.
Mediolateral (x) axis	Orthogonal (according to the right hand rule) to the Z and Y axes
Virtual points	LIC_lm, RIC_lm
Pelvis	
Origin	Midpoint between LIC_lm and RIC_lm
Vertical (z) axis	Line between two mid-points created by a) LIC_lm and RIC_lm mid-point, and b) LHIP_FUNC and RHIP_FUNC midpoint. Unit vector is directed from distal to proximal segment. LIC_lm is created to cause a forward pelvic tilt of 12° to the anatomical frame of the pelvis (Day et al., 1984, Physical Therapy found norms of 10°, Whittle G&P 1996 found female norms were 12°).
Anterior-posterior (y) axis	Least squares approach to firstly determine the ZX plane containing the four segment end points: proximal landmark points LIC_lm, RIC_lm and segment distal landmark points LHIP_FUNC, RHIP_FUNC. The Y axis is perpendicular and directed anterior from this ZX plane.
Mediolateral (x) axis	Orthogonal (according to the right hand rule) to the Z and Y axes
Virtual points	LHIP_FUNC (RHIP_FUNC) defined relative to pelvis cluster technical frame as per Shwartz (2005). LIC_lm (RIC_lm)
Thigh	
Origin	LHJC (or RHJC) which is defined by LHIP_FUNC (or RHIP_FUNC)
Vertical (z) axis	In direction from LKJC (or RKJC) to LHJC (or RHJC)
Anterior-posterior (y) axis	Firstly determine the ZX plane which includes LLK_lm, LMK_lm and LHIP_FUNC segment endpoints. The Y axis is perpendicular and directed anterior from this ZX plane.
Mediolateral (x) axis	Orthogonal (according to the right hand rule) to the Z and Y axes
Virtual points	LHIP_FUNC (RHIP_FUNC) defined relative to pelvis cluster technical frame as per Shwartz (2005). LKJC, LLK_lm, LMK_lm, LKnee_FUNC, LKnee_FUNC_X

Abbreviations: LIC_lm (left iliac crest landmark); LAC (left acromion process); LHIP_FUNC (virtual point representing the functional hip joint centre); LHJC (left hip joint centre); LKJC (left knee joint centre); LLK_lm (left lateral knee landmark); LMK (left medial knee landmark); LKnee_FUNC (point on the functional knee axis line of left knee); LKnee_FUNC_X (lateral point on the functional knee axis line of left knee). The abbreviated terms are consistent with the right limb.

Table 3.3.2.4.2 Anatomical Frame Definitions (shank to foot)

Segment	Description
Shank	
Origin	LKJC (or RKJC) defined as midpoint between LLK_lm (or RLK_lm) and LMK_lm (or RMK_lm). Unit vector is directed from distal to proximal segment.
Vertical (z) axis	Line between two mid-points created by a) LLK_lm and LMK_lm mid-point (LKJC), and b) LLA and LMA midpoint (LAJC)
Anterior-posterior (y) axis	Least squares approach to firstly determine the ZX plane containing the four segment end points: proximal landmark points LLK_lm, LMK_lm and segment distal landmark points LLA, LMA. The Y axis is perpendicular and directed anterior from this ZX plane
Mediolateral (x) axis	Orthogonal (according to the right hand rule) to the Z and Y axes
Virtual points	LKJC, LLK_lm, LMK_lm, LKnee_FUNC, LKnee_FUNC_X. LMA, LLA.
Foot	
Origin	LAJC (or RAJC) defined as midpoint between LLA (or RLA) and LMA (or RMA)
Vertical (z) axis	Line between two mid-points created by a) LLA and LMA mid-point (LAJC), and b) LMH5 and LMH2 midpoint (LMJC). Unit vector is directed from distal to proximal segment.
Anterior-posterior (y) axis	Least squares approach to firstly determine the ZX plane containing the four segment end points: proximal landmark points LLA, LMA and segment distal landmark points LMH5, LMH2. The Y axis is perpendicular and directed anterior from this ZX plane
Mediolateral (x) axis	Orthogonal (according to the right hand rule) to the Z and Y axes
Virtual points	LKJC, LLK_lm, LMK_lm, LKnee_FUNC, LKnee_FUNC_X. LMA, LLA.

Abbreviations: LKJC (left knee joint centre); LLK_lm (left lateral knee landmark); LMK_lm (left medial knee landmark); LKnee_FUNC (point on the functional knee axis line of left knee); LKnee_FUNC_X (lateral point on the functional knee axis line of left knee); LLA (left lateral ankle); LMA (left medial ankle); LMH5 (left 5th meta-tarsal head); LMH2 (left 2nd meta-trasal head). The abbreviated terms are consistent with the right limb.

3.3.2.4.1 Anatomical frame construction

There are three general ordered steps which reconstruct the segment anatomical frame from the cluster technical frame and anatomical landmarks. These are described below and illustrated in Figure 3.3.2.4.1.1.

- Define the segment end points using a combination of at least three anatomical landmarks (or functional landmarks). In the case of the thigh anatomical frame described below, it uses a single landmark for the proximal end point, i.e. the computed functional hip joint centre which also has an accompanying proximal radius to fully describe the segment shape at the proximal end. For the distal end, the thigh anatomical frame uses the lateral and medial knee anatomical landmarks which are projected onto the prior computed functional knee axis line. The proximal and two distal landmarks are then used to define the frontal plane (ZX plane) of the segment. The thigh is a special case, but all other segments use two proximal and two distal end points of the segment to describe the ZX plane.

- Create the longitudinal (Z) axis from the mid-point between distal segment end points and the mid-point between proximal segment end points (thigh being special case). Segment position and orientation is created by a least squares procedure among the ALs (as per Spoor and Veldpaus, 1980). The Y axis is then projected anterior and orthogonal to the ZX plane, from the most proximal point on the Z axis. This creates a ZY plane.
- The X axis is created perpendicular to the ZY plane, according to the right hand rule (i.e. from left to right is the positive direction according to the Z and Y axis definitions above). Therefore, to treat the left and right lower limbs with the same convention for segment motion in the frontal and transverse planes, the X axis is negated in one of either the left or right segments. Now, the referred terms adduction/abduction, and internal/external rotation, can be consistent between sides.

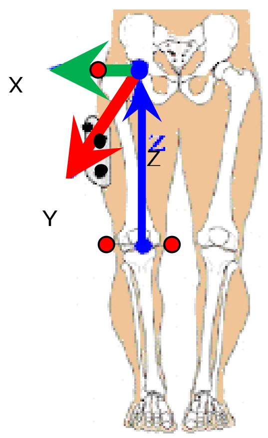


Figure 3.3.2.4.1.1

The cluster technical frame plate positioned on the lateral aspect of the thigh. Anatomical landmarks illustrated in red, tracking markers in black. The XYZ segment coordinate system axes are green, red and blue. Figure adapted from C-Motion.

3.3.3 Segment angles

Segment angles were described in three-dimensions using Cardan-Euler transformation (Cole, Nigg, Ronsky and Yeadon, 1993; Baker, 2001; Cappozzo et al., 2005). A Cardan ordered sequence of rotations X-y-z (i.e. flexion/extension, adduction/abduction, axial rotation) was used for the thigh, shank and foot, and a Z-y-x ordered sequence for the pelvis (Yaw-rotation, Pitch-obliquity and Roll-tilt) as recommended by Baker (2000). This order of rotations (i.e. Z-y-x) has also been applied elsewhere to trunk segment motion (Wu, Meijer, Lamoth, Uegaki, van Dieen, Wuisman, de Vries and Beek, 2004; Van Emmerik, McDermott, Haddad and Van Wegen, 2005; Kang and Dingwell, 2006).

Cole et al. (1993) demonstrated that the X-y-z sequence about three fixed axes is equivalent to the non-orthogonal joint coordinate system suggested by Grood and Suntay (1983) and the International Society of Biomechanics recommendations (Wu and Cavanagh, 1995). While the Grood and Suntay (1983) method is generally well accepted throughout the biomechanics society for describing joint angles, the Cardan X-y-z method is well suited for models describing segments with 6-DoF (Cole et al., 1993).

The foot segment anatomical frame pose as described by the standing trial is generally orientated at some oblique angle and is not a congruent alignment with the global reference frame. To re-align the foot segment so that during static calibration its anatomical frame is aligned with the global reference frame, a virtual anatomical frame of the foot segment can be created such that it has its XY plane parallel with the floor. When the original foot reference frame and the virtual foot reference frame are compared, the outcome is an angle that can be referred to as the foot progression angle.

3.4 Data processing and parameterization

Gait data were processed within the Visual3D environment. A quiet standing calibration trial was used to reconstruct the reference model. The static calibration model was then applied to the trials from the walking test.

3.4.1 Interpolation and Filtering

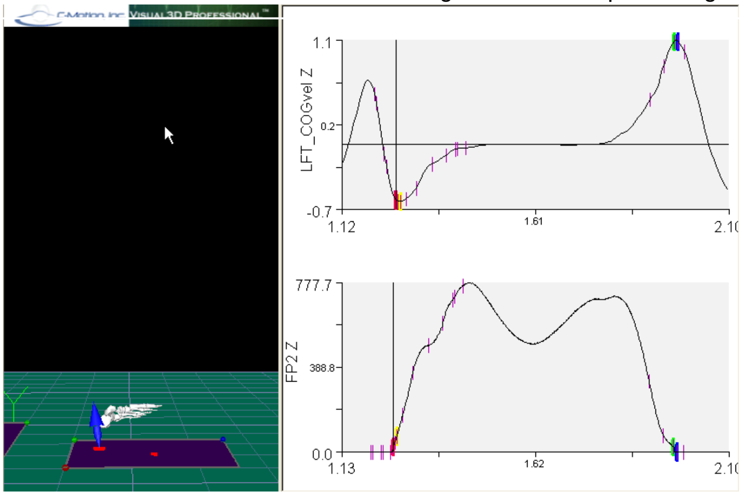
Marker occlusions during testing were rare, but there were very brief moments where markers were obstructed momentarily by loose wires or the tethered cable crossing the pelvis markers. While most occlusions were < $5/70^{th}$ of a second, a cubic polynomial fitting algorithm was used to interpolate raw data for a maximum occlusion period of 10 frames. Data filtering was then processed on the interpolated raw data at the determined optimal cut off frequency of 7Hz for the pelvis and lower limbs markers, and at 5 Hz for trunk and head markers (as discussed in section 3.3.1.2).

3.4.2 Determining heel contact and toe off events

Gait events of initial contact (i.e. heel contact) and terminal contact (i.e. toe off), were obtained by using a foot velocity algorithm by specifically determining the events of the local maxima and minima from bilateral foot velocity-time curves. This procedure was done using custom written routines in Visual3D software (C-Motion Inc.). The rationale for using the vertical foot velocity events was based upon research by O'Connor et al. (2007) who analysed foot kinematic data sampled at 60Hz with a low pass Butterworth filter with 7Hz cut-off frequency for 54 participants and several pathological participants. The method proposed by O'Connor et al. (2007) was compared with the kinematic and force platform derived events and found good agreement (Figure 3.4.2.1). The method by O'Connor et al. (2007) compared the velocity signal derived events against acceleration derived events and found a more normal distribution of the errors when comparing against force platform derived events. O'Connor et al. (2007) derived a marker representing the mid-point of the foot, specifically derived from a heel marker and a 2^{nd} metatarsal-head marker was used to obtain the 'foot' velocity profile. In this study, local maxima and local minima of the foot velocity were also used to determine the respective toe off and heel contact events. The toe off event was determined from the velocity profile of the foot centre of mass. Specifically, toe off was determined when this point reached a maximum vertical velocity at the terminal region of the pre-swing (double support) phase. When determining the heel contact event, the vertical velocity of the point representing the foot centre of mass was also used, specifically, the local minimum about the terminal swing phase indicated this event. Other methods in the literature for deriving gait cycle events during treadmill walking have been based upon algorithms using heel positions (Dingwell et al., 1999; Osaki et al., 2007). These methods however have can have potential errors for accurately defining toe-off and heel-contact timing when walking on the treadmill, because the heel rises and moves forwards during late stance, and the heel is potentially moving posteriorly relative to the body during late swing.

The method for determining toe off and heel contact events obtained force platform data of toe off and heel contact events using a threshold of 10N. Specifically, the algorithm selected the frame of data if it was greater or lower than the 10N threshold for heel contact and toe off events respectively. This meant that toe off events derived from this method represented the last sampled instant that a force was registered. Therefore, both heel contact and toe off events derived from the force platform threshold algorithm were the first and last instants where a force was registered. These force platform derived events were then compared to the events based upon the kinematic algorithm by O'Connor (2007). The results of ten subjects, each obtaining twenty samples of left and right footfalls were obtained. From approximately 400 comparisons, the toe off events defined by the kinematic algorithms always occurred after the force platform event, and in the following way: zero offset 16%; one frame offset 47%; two frames offset 31%; three frames offset 6%. For heel contact events, the data demonstrated the following: zero offset 32%; one frame offset 53%; two frames offset 15%. Each sampled frame represents 1/70 second, or 0.0143 seconds.

A. Heel contact event. Vertical animation line indicating event with respect to signal



B. Toe off event. Vertical animation line indicating event with respect to signal

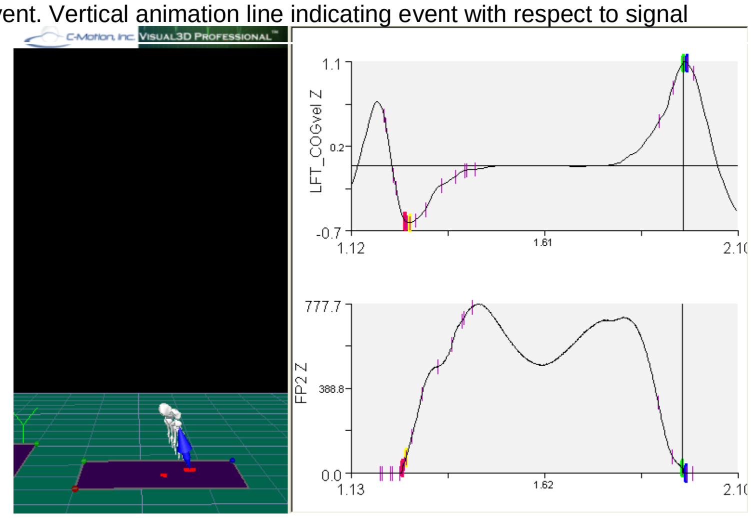


Figure 3.4.2.1

Demonstrating comparison between kinematic and force platform derived heel contact (A) and toe off (B) events. Left vertical panel is of the left foot making initial and terminal contact with force platform and the overlayed ground reaction force vector. Illustrations of events on signal on the right are for the foot centre of mass vertical velocity signal (LFT_COGvelZ) and force platform vertical GRF signal (FP2Z). Yellow and red lines represent heel contact events (force platform and kinematic algorithms respectively). The green and blue lines represent toe off events (force platform and kinematic algorithms respectively).

3.4.3 Determining the first maximum and the minimum toe clearance events of the swing phase

The goal of the swing performance is defined as the control of the end-effector path, or trajectory of the minimum toe point location as it approaches the minimum toe-ground clearance height (MTC). There were four virtual points located at the inferior surface of the distal shoe to investigate how lower limb geometry can determine which location on average gets closest to the ground surface. The point which reflected the lowest mean value at MTC was used as the MTp for that particular individual. Most commonly, this point was the most distal MTp landmark.

The first local maximum event (MX1) following toe off and the next local minimum event (MTC) were key reference events that were used to define off-set events, or toe trajectory states, during the swing phase. Events were created at MTC for the four identified MTC landmarks (Figure 3.4.3.1), such that MTp4 was the first point to enter a local minimum. Whichever of the four MTp landmarks was found to have the lowest mean MTp with respect to the ground at the first minimum inflection point (i.e. MTC) was then adopted for the remaining kinematic analysis.

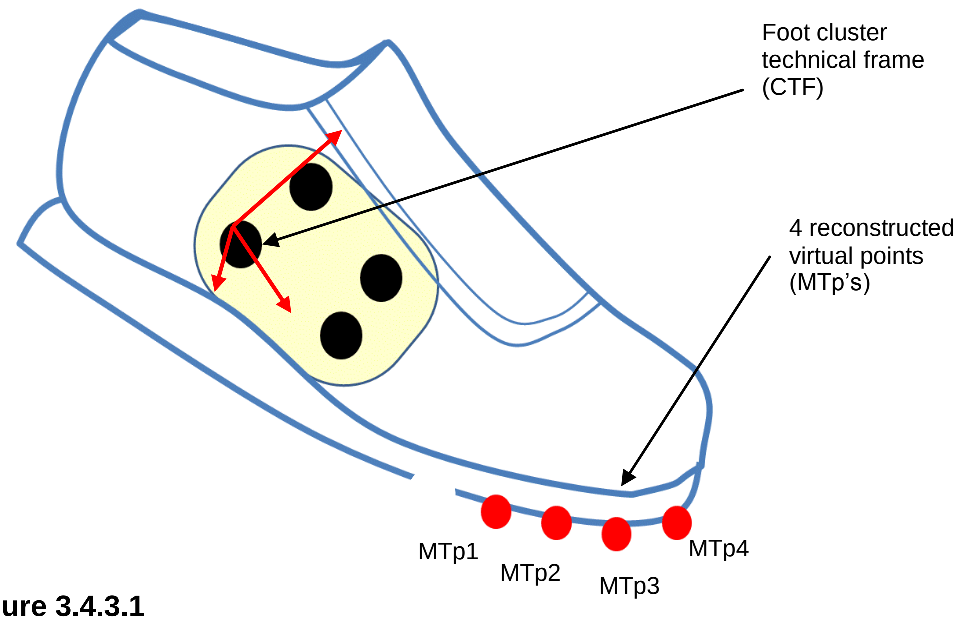


Figure 3.4.3.1

The four minimum toe point (MTp) locations on the distal inferior surface of the shoe out-sole described with respect to the foot CTF.

3.5 Gait kinematics

This section describes the processing procedures for describing the gait kinematics.

3.5.1 Walking velocity

Treadmill walking speed was calculated by the stance foot velocity (central difference method). This was determined from the average of the anterior-posterior velocity of the support foot when it was at mid-stance (NB: the average was taken from both the left and right single support phases). This was a more accurate representation of walking speed because the difference between the foot speed algorithm compared to treadmill calculation was consistently different at similar walking speeds. For example, all subjects walking at 5km/hr displayed the same discrepancy in speed, irrespective of body mass index differences. Therefore, it is most likely that the treadmill motor was not as capable as it expected! The reported treadmill speed has no influence on results because subjects walked at a speed they perceived was comfortable irrespective of the speed displayed by the treadmill.

3.5.2 Stride length, step length and step width

Step length (L_{step}), is calculated by multiplying the step time interval (T_{step}) by the average belt speed during 'foot flat' of the single stance period (v_{step}). Then adding the displacement found between the consecutive contra-lateral heel positions between contra-lateral heel strike events. So, given the 'final' heel strike is behind the 'initial' heel strike, the difference in the anterior-posterior displacement will be subtracted from the product of 'step time and mean treadmill velocity', i.e.:

$$L_{step} = T_{step} \times v_{step} + \Delta heel_{CL} \tag{3.5.2.2.1}$$

Stride length (L_{stride}), is calculated in a similar fashion, stride time (T_{stride}) is multiplied by the average belt speed for the stride time period, i.e. the average velocity calculated from consecutive 'foot flat' time periods. Then adding (or subtracting) the relative distance between successive ipsi-lateral heel strikes.

$$L_{stride} = T_{stride} \times \bar{v}_{stride} + \Delta heel_{IL} \qquad (3.5.2.2.2)$$

Stride width is calculated as the relative medio-lateral displacement between successive contra-lateral heel positions at contra-lateral heel strike events. Right step width is interpreted as positive values when the right heel contact is medial to the previous left heel contact. Contrarily, left step width is interpreted as negative values when the left heel contact is positioned medial to the right limb using the previous right heel strike as reference.

3.5.3 Describing the configuration of the three effector systems: stance limb; swing limb; and a combined system

During walking the limbs are often referred to as either stance or swing limbs because of the phase they are undergoing. In this study, the stance and swing limbs are defined as a kinematic chain of segments spanning proximal and distal end-points of a segment linkage. A kinematic chain of body segments with a task goal can be described as a kinematic effector system. The span of an effector system can be defined by a vector whose point of application is a position located on the proximal segment, and the end-point, or end-effector, is usually a distal position located on the distal segment. Typically, in motor tasks with multi-segment linkages, the end-effector is generally relevant to an outcome goal of a kinematic movement task, i.e. a task variable. In contrast, a performance variable can be defined as an end-effector state of a collective of multiple effector systems sharing a mutual performance goal.

In this current study, there were three effector systems (Figure 3.5.3.1) each forming a three-dimensional vector: a combined effector; a stance effector; a swing effector. Each effector spanned a chain of segments defined by their endpoints. The combined effector spans the collective configuration of the stance limb (foot, shank, thigh and pelvis) and the swing limb (thigh, shank and foot). The stance and swing effector vectors are separate components of the combined effector system.

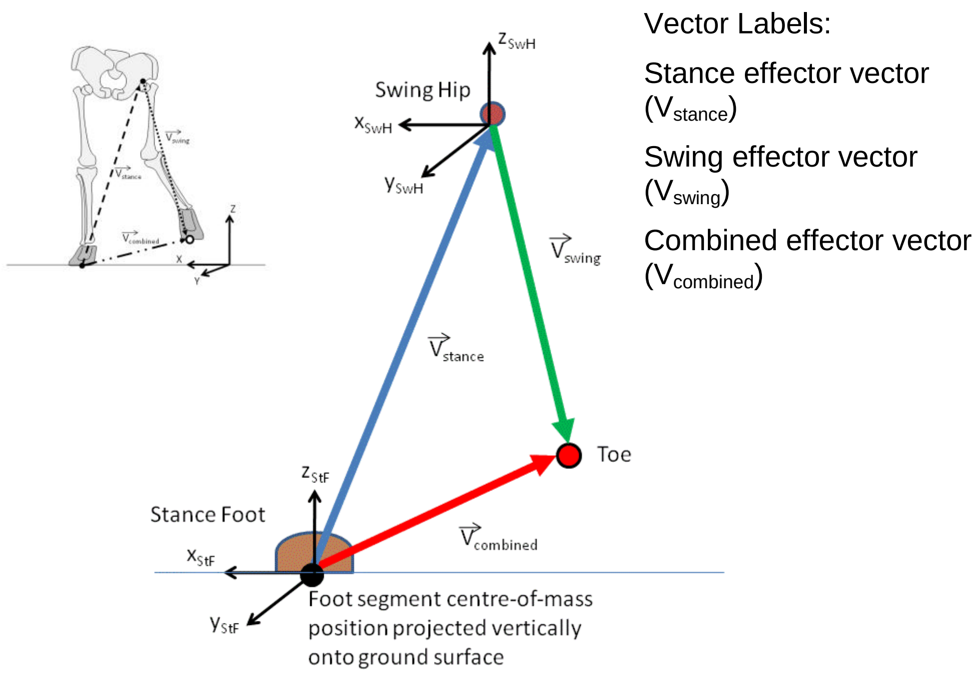


Figure 3.5.3.1

Diagram of the vector loop defining the effector systems. The three 3-dimensional vectors of stance (V_{stance}), swing (V_{swing}), and combined ($V_{combined}$) are defined by moving reference frames with origins at the stance foot, swing hip and stance foot respectively.

3.5.3.1 Reference systems for defining the toe-swing path

Describing the performance of toe control will be dependent upon the spatial reference from which it is represented. For treadmill walking, the application of an external reference system, defined by the fixed global reference frame (GF), does not adequately represent the internal or external reference space from which the loco-sensorimotor control system reconstructs the task. The effector systems have a moving reference system with the same orientation as the global reference frame (Figure 3.5.3.1.1). For the stance effector and the combined effector, the reference frame is located at a point beneath the stance foot. Specifically, the moving stance foot centre of mass point has been projected vertically downwards onto the horizontal plane defined by the global reference frame. For the swing effector, the reference frame is located at the hip joint centre of the swing limb which is defined by the moving pelvis.

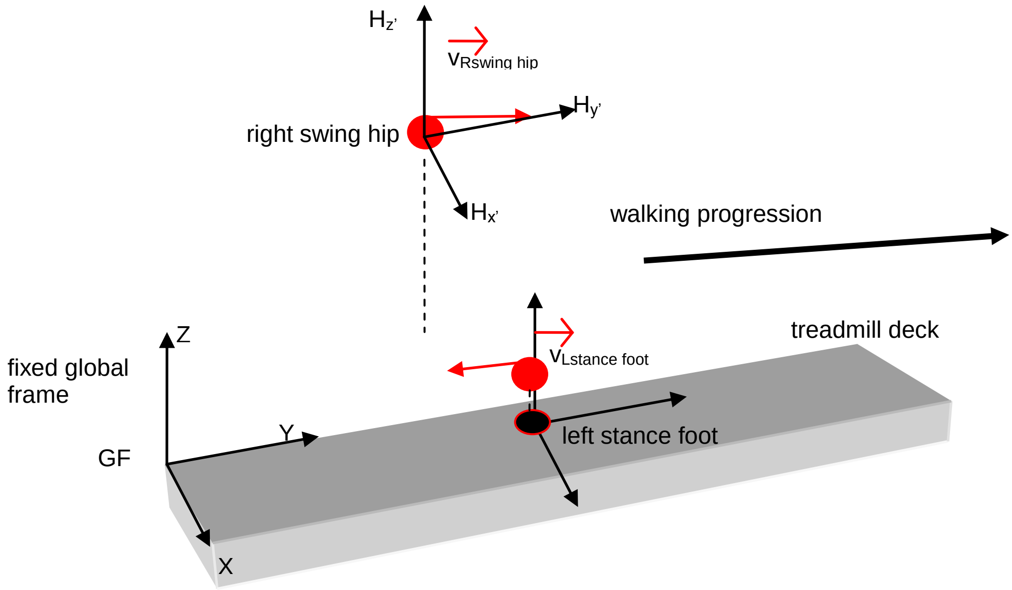


Figure 3.5.3.1.1

The moving reference system (illustrating an example case of the right limb swing phase) relative to the location of the global reference frame (GF). The two reference systems define the vector configuration of the stance (vector St) and swing (vector Sw) effector configuration. The origin of vector St is projected vertically from the stance foot centre of mass onto the XY plane of the GF. The origin of vector Sw is located at the hip joint centre of the swing limb. The orientation of the moving reference systems have a fixed orientation with the GF but move with respect to the moving body locations in the GF.

Ideally, the stance effector will have an improved functional relevance if it could be described as a vector relating the true centre of pressure location under the stance foot and the true body centre of mass position. While this does indicate a closer agreement with the balance performance of the body, it still does not explain the role of full body angular momentum. These balance parameters of body centre of mass, full body angular momentum, and centre of pressure, are details that require advanced experiment design methods. The mechanical configuration of the stance effector does not fully explain the mechanics behind the postural control task of gait, but it does provide a close approximation to the kinematic details performed by the stance limb and pelvis when carrying its role to control this task. Certainly, the description of the stance effector makes it conducive to explaining the swing hip position regardless of whether this is the task goal of the stance limb. In contrast, the swing effector can fully explain the mechanical details contributing to ground clearance of the swing limb. However, the swing effector also has competing task goals of forward progression and a contribution to whole body angular momentum.

The stance and swing effector vectors are described in three dimensions from the proximal internal reference frame. The vectors previously illustrated in Figure 3.5.3.1.1 are described below in Table 3.5.3.1.1.

Table 3.5.3.1.1.

Three-dimensional vector description of stance and swing limb effectors and the description of the minimum toe point location.

	Stance vector (S_T)	Swing vector (S_W)	Combined vector (C)
Vector direction	Projected stance foot centre of mass to swing hip joint centre	Swing hip joint centre to minimum toe point landmark (distal inferior shoe surface)	Projected stance foot centre of mass to minimum toe point landmark (distal inferior shoe surface)
Vector reference frame location	Projected stance foot centre of mass. Aligned with lab reference frame.	Swing hip joint centre. Aligned with lab reference frame.	Projected stance foot centre of mass. Aligned with lab reference frame.

3.5.4 Time points along the swing-cycle

Gait kinematics were described at points along the swing-cycle that were time-normalised from each swing-phase duration. Therefore, there were 101 equally spaced re-sampled time points along the swing-cycle. The swing-phase time periods were determined by algorithms for detecting foot-to-ground contact events written in Visual3D (C-Motion Inc., Rockville, MD, USA).

Most studies of gait describe kinematic trajectories across the stance or swing phase periods, however, this time-relevant procedure dilutes task-relevant information. There is a good argument to represent gait cycles as a sequence of mechanical states along frequent time-steps within the cycle, and observed to repeat between cycles (Forner-Cordero, Koopman and van der Helm, 2006). This has the cycle-to-cycle state-alignment advantage over common time-normalising procedures that typically use only one, or two, recurring states of the cycle. This study accepted this argument and described gait trajectories with respect to intermediate states of the swing-phase that have task-relevance with toe-clearance; i.e., events defined where toe velocity equals zero, or the first maximum and minimum turning points of the toe trajectory, MX1 and MTC respectively.

All resampled 101-point kinematic variables were then time-aligned with MX1 and MTC events within each swing cycle using custom-written event detection algorithm in MATLAB (The MathWorks Inc.). Within each cycle, the MX1 and MTC events served as a 'zero-time' reference event, such that each event was centred within its own separate sub-phase region of the swing cycle. The extent of the sub-phase region was determined by equally spaced discrete-time steps of the time-normalised swing cycle. There were 12 time-points before and after MX1, and 15 time-points either side of MTC (Figure 3.5.4.1). These two regions in essence accounted for 56% of the swing-phase cycle; whereby a 12% swing-cycle proportion surrounded each side of the MX1 event, and a 15% swing-cycle proportion surrounded the MTC event.

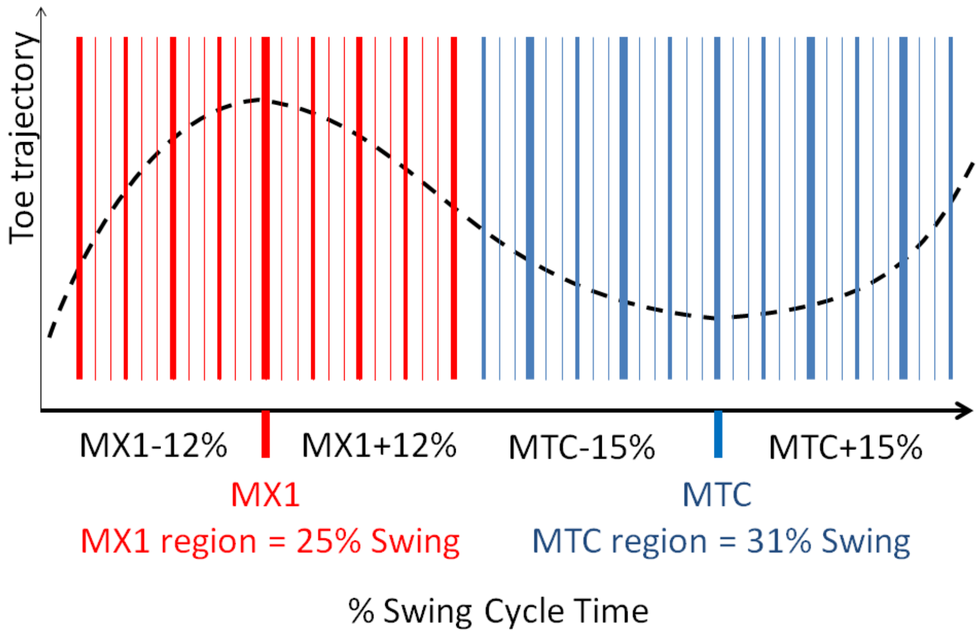


Figure 3.5.4.1.

Sub-phase regions MX1 (red) and MTC (blue) of the swing cycle. Time slice increments indicated by vertical lines that are based upon 1% time-steps relative to within-region zero-reference events MX1 and MTC. Horizontal axis represents time intervals (Δt) proportionate to 1% of time-normalised swing phase.

3.6 Uncontrolled Manifold (UCM) hypothesis

The Uncontrolled Manifold (UCM) hypothesis is addressing the problem of how the loco-sensorimotor system manages the presence of ubiquitous noise and variability found in a redundant effector system, so that the functional goal of the effector remains stable. The UCM can quantify how the variability is being managed with respect to the effect it has on the task goal by analysing data collected across multiple trial repetitions, where the task goal and initial conditions are presumed to remain relatively consistent. Specifically, the UCM gives an insight into how variability is managed by the loco-sensorimotor system through a model that decomposes variability of the redundant effector system into two components; good variability which does not cause a change to task goal performance, and bad variability which does cause a change in task goal performance.

The next sub-sections explain the specific theory and approach of the UCM hypothesis. But, here is a brief outline of how the UCM will be applied to the effector systems. The stance and swing redundant effector systems will be investigated separately. In theory, each segment in both the effector systems has 6-DoF and the endpoint of the effector is described in 3-D position. For a three and four segment chain, each multi-link effector system ultimately has 18-DoF and 24-DoF respectively, to effect the 3-D position of the end-effector. The approach taken in this thesis was to confine each effector onto a plane (i.e. reducing the system DoF), and reducing the end-effector position to one-dimension. The swing effector was analysed by the UCM in each component of the two-dimensional sagittal plane, forming two one-dimensional task variables in this plane. The stance effector was analysed by the UCM in each component of the two-dimensional frontal plane, also forming two one-dimensional task variables in this plane. The swing effector was therefore reduced to 3-DoF kinematic-link system, and the stance effector was reduced to 4-DoF kinematic-link system.

3.6.1 Mapping body segment variables to task-goal variable

A goal-directed movement can be related to a goal function which defines the task, by mapping a set of elemental variables, θ, to a task-goal variable, g (Cusumano and Cesari, 2006);

$$f(\theta, g) = 0 \tag{3.6.1.1}$$

In this study, the elemental variables are the segment angles of the effector systems, and the task-goal variable is the end-effector position. A 'goal function' is defined by a reconfiguring process of the elemental variables such that they attain the general form of the goal function (equation 3.6.1.1). The set of body state variables which satisfies the task-goal can be defined below:

$$\mathcal{G} = \{\theta \mid f(\theta, g) = 0\} \tag{3.6.1.2}$$

where $\mathcal{G}$,represents the structure of a 'manifold' in body state space ($\theta \in R^B$), where B is the dimension of the body state space. The task-variable, or the end-effector described by one component of the 3-D vector (i.e. z component; $g \in R^G$), has a dimensionality of G = 1. The dimension of the manifold is the dimensionality mismatch between the body state space (B=3) and the goal function (($f \in R^E$) where E=1), where B-E = 2 represents a 2-dimensional manifold. Therefore, the manifold is a (B-E)-dimensional surface located within the B-dimensional effector state space. For the goal variable z to be achieved, the set of body state variables will satisfy the goal function, $f(\theta, g)=0$, for every strategy belonging to the set, $\theta \in G$. However, this set of solutions is not consistent between solutions as some joint configurations can be more stable than others, even though the goal has been achieved. Therefore, a subset, $\theta^* \in G$ explains the set of superior strategies (Section 3.6.2.2.1). Therefore, for a fixed goal variable g*, a set of body state variables will satisfy the goal function, $f(\theta^*, g^*)=0$, for every strategy $\theta^* \in G$.

3.6.1.1 Task variables

The first task variable was related to controlling the vertical states of the swing-effector endpoint trajectory, or for simplicity this is defined as the vertical task of the swing-effector (Figure 3.6.1.1.1). The vector, L_{swing}, describes the first task variable and stretches between the swing-hip and swing-toe positions. Hence, this vector spans the body segments of thigh, shank and foot. The effector state-space has dimensionality of B=3. The swing limb end-effector is the vertical toe position which is one-dimensional, G=1. The body-goal mapping function is described by equation 3.6.1.1.1:

$$f(\boldsymbol{\theta}^*, g^*) \equiv$$
$$V_{swing(z)} - g^* - \left(l_{thigh} sin(\theta_{thigh(yz)}) + l_{shank} sin(\theta_{shank(yz)}) + l_{foot} sin(\theta_{foot(yz)}) \right) = 0 \quad (3.6.1.1.1)$$

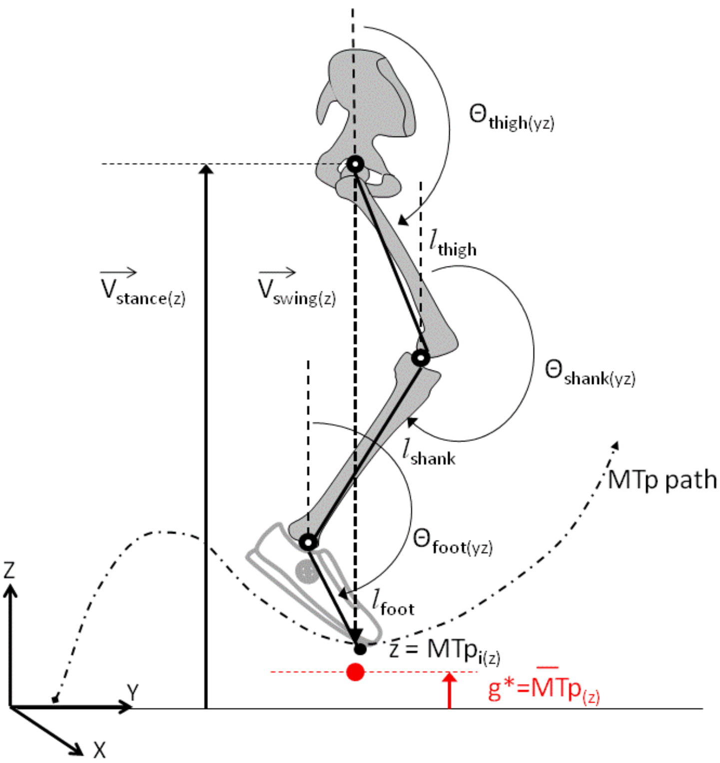


Figure 3.6.1.1.1

Illustration of the swing limb vector projected onto the YZ plane which defines the effector system that stretches between the swing hip position (y, z) to the swing toe position (y, z). Segment angles are measured relative to the positive vertical axis. A geometric model describes the vertical component of this essentially 3-dimensional vector relationship with segment length and segment angle in equation 3.6.1.1.1.

The second task variable investigated was the task-goal related to the stance effector (Figure 3.6.1.1.2). When projecting the multi-degrees-of-freedom stance effector onto the frontal plane, the body-goal mapping function makes two general assumptions. First, knee adduction/abduction is considered negligible, and therefore the shank angle can be used to describe the stance limb orientation in the frontal plane. Second, the effective length of the combined foot, shank and thigh in the frontal plane is defined by the vertical component of the thigh and shank limb lengths and angular position in the sagittal plane, where $L_{stance(xz)}$ represents the effective stance limb length.

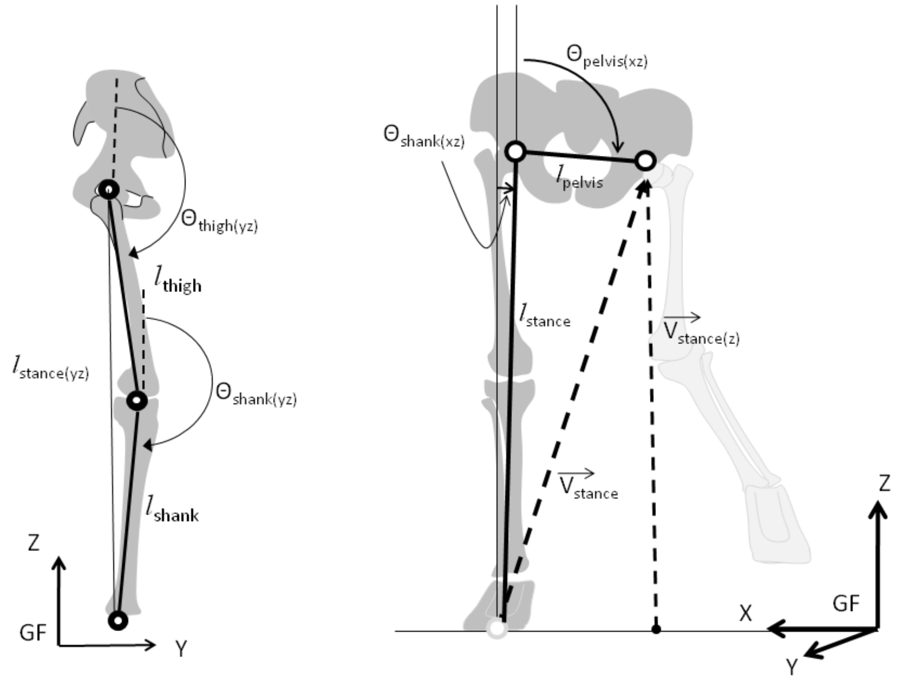


Figure 3.6.1.1.2.

Vector $V_{stance(z)}$ describes the stance effector task-goal in the vertical dimension. The length l_{stance} is defined by the sagittal plane shank and thigh. The vertical component of the vector is described by the length and orientation of the effective stance limb l_{stance} and θ_{shank}, and the pelvic width and orientation, l_{pelvis} and θ_{pelvis}.

Therefore the body-goal mapping function is described by equation 3.6.1.1.2:

$$f(\boldsymbol{\theta}^*, g^*) \equiv V_{stance(z)} - l_{stance(z)} sin\big(\theta_{shank(xz)}\big) + l_{pelvis} sin\big(\theta_{pelvis(xz)}\big) = 0 \tag{3.6.1.1.2}$$

where $l_{stance(yz)}$ is defined by equation 3.6.1.1.3;

$$l_{stance(z)} = l_{thigh} sin(\theta_{thigh(yz)}) + l_{shank} sin(\theta_{shank(yz)}) \qquad (3.6.1.1.3)$$

Therefore, substituting 3.6.1.1.2 into 3.6.1.1.3 becomes:

$$f(\theta^*, g^*) \equiv V_{stance(z)} - \left[l_{thigh} sin(\theta_{thigh(yz)}) + l_{shank} sin(\theta_{shank(yz)})\right] sin(\theta_{shank(xz)}) + l_{pelvis} sin(\theta_{pelvis(xz)}) = 0$$

(3.6.1.1.4)

Equation 3.6.1.1.4 describes a relationship between the task-goal variable and the effector system through three constants (three segment lengths) and four segment angle variables.

3.6.2 Computing the UCM components of variance

The UCM analysis was computed for three task variables. Each task variable was defined by a hypothesized end-effector with a task-goal. The task variables related to the stance and swing effector systems have different geometric models and different degrees of freedom within each model. This discrepancy in the degrees of freedom prevents making a direct UCM comparison between stance and swing task variables. The different effector systems will be evident below in section 3.6.2.1.

Each swing cycle period was interpolated to be time normalised to 101 points, therefore increasing the original number of absolute-time sample points for the swing cycle period. An event detecting algorithm in MATLAB then determined events MX1 and MTC during each swing cycle. The UCM was computed at selected percentage values of the time-normalised swing phase. These percentage values were off-set from the reference events of MX1 and MTC (see Figure 3.5.4.1). The UCM was calculated for discrete trials at six incremental time slices of 3% (swing phase time normalised) either side of the reference states, MX1 and MTC. The steps involved in computation of the UCM are described in the following sections.

3.6.2.1 Step 1: Geometric model

The first step is to define a geometric model which links the task-goal variable in effector space (e.g. segment angles of the stance or swing limb). This essentially describes the elements of the effector system in the context of the performance of the respective end-effector variable. The geometric model will be defined by the effector spanning vectors which describe the effector systems being analysed.

The swing and stance effectors are being analysed using the UCM hypothesis procedure. These two effector systems have hypothesised task variables related to end-effector position. The structure of UCM variance is being investigated in four candidate task variables; 1) medio-lateral control of stance effector; 2) vertical control of stance effector; 3) anterior-posterior control of swing effector; 4) vertical control of swing effector. The corresponding geometric models are described below:

- Two task-variables of the swing effector: vector stretching the swing limb kinematic chain between the minimum toe point and the proximal swing hip position. The projected angle of the thigh, shank and foot segment links (i.e. elements of the effector system) onto the sagittal plane describe the task space. The geometric model relating the body links with the end effector position (also see Figure 3.6.1.1.1) is described by the following equation:

$$V_{swing(z)} = l_{thigh}cos(\theta_{thigh}) + l_{shank}cos(\theta_{shank}) + l_{foot}cos(\theta_{foot}) \qquad (3.6.2.1.1)$$

$$V_{swing(y)} = l_{thigh}sin(\theta_{thigh}) + l_{shank}sin(\theta_{shank}) + l_{foot}sin(\theta_{foot}) \qquad (3.6.2.1.2)$$

- Two task- variables of the stance effector: vector stretching the kinematic chain between the stance foot centre of mass to the swing hip position. A simplified geometric model was used to define the task variables in the frontal plane. The projected angle of the thigh and shank in the sagittal plane define the stance hip position. The vertical component of this vector is used along with the shank angle to define the stance task in the vertical and medio-lateral dimensions.

$$V_{stance(z)} = [l_{thigh}cos(\theta_{ZYthigh}) + l_{shank}cos(\theta_{ZYshank})]cos(\theta_{ZXshank}) + l_{pelvis}sin(\theta_{ZXpelvis}) \qquad (3.6.2.1.3)$$

$$V_{stance(X)} = [l_{thigh}cos(\theta_{ZYthigh}) + l_{shank}cos(\theta_{ZYshank})]sin(\theta_{ZXshank}) + l_{pelvis}cos(\theta_{ZXpelvis}) \qquad (3.6.2.1.4)$$

- where the component in square brackets of equations 3.6.2.1.3-4 represents the collective length of the stance limb shank and thigh in the frontal plane, l_{stance} (Figure 3.6.1.2).

As discussed previously, there are assumptions that out-of-plane projections are considered negligible. The basis of this assumption is relevant to equations 3.6.2.1.1-4 and is described in the following analysis. With respect the thigh and shank, when the segments have an 'out of plane' orientation of approximately 10 degrees, their actual projected segment lengths is less than 1.5% of their true length. Similarly, when the pelvis segment has an estimated out-of-plane orientation of ±10 degrees when projected onto the XZ plane, the pelvis length will have an associated length change of 1.5%. Therefore, projections of the pelvis on the frontal plane, and projections of the shank and thigh segments onto the sagittal plane can potentially involve errors which project the segment lengths as a reduction of their true length by approximately 1.5%.

The other assumption is that the frontal plane stance knee (adduction-abduction) angle is zero degrees. Given that adduction could potentially involve an angle of ±5 degrees (Lafortune, Cavanagh, Sommer and Kalenak, 1992), this would potentially have an effect on the error associated with the x position of the swing hip. The magnitude of this error is therefore expected to have a negligible effect on the vertical position of the swing hip.

3.6.2.2 Step 2: Linear approximation of the UCM

Compute a linear estimate of a two-dimensional manifold in state-space, such that the set of all joint combinations that lie upon the manifold leave the task-goal variable unchanged. The desired task-goal can be defined by the average joint configuration because this is most likely to be the average of the task variable, and hence representing the desired task goal. The reference joint configuration is defined as the grand mean of the three joint angles:

$$\theta^0 = \begin{bmatrix} \bar{\theta}_{thigh} \\ \bar{\theta}_{shank} \\ \bar{\theta}_{foot} \end{bmatrix} \qquad (3.6.2.2.1)$$

The deviation matrix is obtained by the difference between the current 'stride's' joint configuration, θ_i , and the mean reference configuration, θ^0. For example, the deviation matrix is defined by $d\theta_i = \theta_i - \theta^0$, and for all joint angles the instantaneous difference is:

$$d\theta_i = \theta_i - \theta^0 = \begin{bmatrix} \theta_{thigh_i} - \bar{\theta}_{thigh} \\ \theta_{shank_i} - \bar{\theta}_{shank} \\ \theta_{foot_i} - \bar{\theta}_{foot} \end{bmatrix} \tag{3.6.2.2.2}$$

The sum of the least squares of $d\theta_i$ can provide a general indication of the joint configuration variance.

A Jacobian matrix of partial derivatives creates a link between the joint angular changes and the changes to the task variable [e.g. the vertical toe position from the hip joint of the swing leg; $V_{swing(z)}$] at each time slice (time sample) of the swing phase. The Jacobian is derived by partial differentiation of the geometric model where the partial derivatives of the task variable are taken with respect to the mean joint angle at each time slice of the swing cycle. The Jacobian matrix is obtained from the reference joint configuration matrix, by $J(\theta^0)$. So for the swing task variable, the Jacobian matrix of partial derivatives is:

$$J_{Swing}(\theta^0) = \begin{bmatrix} \frac{\partial Vswing_Y}{\partial \theta_{thigh_{YZ}}} & \frac{\partial Vswing_Y}{\partial \theta_{shank_{YZ}}} & \frac{\partial Vswing_Y}{\partial \theta_{foot_{YZ}}} \\ \frac{\partial Vswing_Z}{\partial \theta_{thigh_{YZ}}} & \frac{\partial Vswing_Z}{\partial \theta_{shank_{YZ}}} & \frac{\partial Vswing_Z}{\partial \theta_{shank_{YZ}}} \end{bmatrix} \tag{3.6.2.2.3}$$

For the task variables associated with the stance effector, the 2-dimensional end-effector solution can be described in the anterior-posterior (y) and vertical (z) dimensions. However, only the vertical (z) dimension will be computed for task variable 1 and also for task variable 2 because this is relevant to the clearance height of MTp.

$$J_{Stance}(\theta^0) = \begin{bmatrix} \frac{\partial Vstance_x}{\partial \theta_{thigh_{YZ}}} & \frac{\partial Vstance_x}{\partial \theta_{shank_{YZ}}} & \frac{\partial Vstance_x}{\partial \theta_{shank_{XZ}}} & \frac{\partial Vstance_x}{\partial \theta_{pelvis_{XZ}}} \\ \frac{\partial Vstance_Z}{\partial \theta_{thigh_{YZ}}} & \frac{\partial Vstance_Z}{\partial \theta_{shank_{YZ}}} & \frac{\partial Vstance_Z}{\partial \theta_{shank_{XZ}}} & \frac{\partial Vstance_Z}{\partial \theta_{pelvis_{XZ}}} \end{bmatrix} \tag{3.6.2.2.4}$$

3.6.2.2.1 *Jacobian of the geometric model related to the task variables*

The lengths of the segments in the geometric model of the stance and swing task variable are constants. With respect to the geometric model linking the vertical toe position with the sagittal plane angular configurations of the swing limb, the (1 x 3) Jacobian is described as:

$$J_{Swing_y}(\theta^0) = \left[l_{thigh} cos(\theta^0_{thigh}) \quad l_{shank} cos(\theta^0_{shank}) \quad l_{foot} cos(\theta^0_{foot})\right] \qquad (3.6.2.2.1.1)$$

$$J_{Swing_z}(\theta^0) = \left[-l_{thigh} sin(\theta^0_{thigh}) \quad -l_{shank} sin(\theta^0_{shank}) \quad -l_{foot} sin(\theta^0_{foot})\right] \quad (3.6.2.2.1.2)$$

The (1 x 4) Jacobian matrix for the medial component (x) of stance effector task variable:

$$J_{stance_x}(\theta^0) = \left[-l_{thigh} sin(\theta^0_{thigh_{ZY}}) sin(\theta^0_{shank_{ZX}}) \quad -l_{shank} sin(\theta^0_{shank_{ZY}}) sin(\theta^0_{shank_{ZX}}) \cdots\right.$$

$$\left. l_{thigh} cos\left(\theta^0_{thigh_{ZY}}\right) cos\left(\theta^0_{shank_{ZX}}\right) + l_{shank}\, cos\left(\theta^0_{shank_{ZY}}\right) cos\left(\theta^0_{shank_{ZX}}\right) \quad -l_{pelvis} sin({\theta^0}_{pelvis_{ZX}})\right]$$

(3.6.2.2.1.3)

The (1 x 4) Jacobian matrix for the vertical component (z) of stance effector task variable:

$$J_{stance_z}(\theta^0) = \left[-l_{thigh} sin(\theta^0_{thigh_{ZY}}) cos(\theta^0_{shank_{ZX}}) \quad -l_{shank} sin(\theta^0_{shank_{ZY}}) cos(\theta^0_{shank_{ZX}}) \cdots\right.$$

$$\left. l_{thigh} cos\left(\theta^0_{thigh_{ZY}}\right) sin\left(\theta^0_{shank_{ZX}}\right) + l_{shank}\, cos\left(\theta^0_{shank_{ZY}}\right) sin\left(\theta^0_{shank_{ZX}}\right) \quad -l_{pelvis} cos({\theta^0}_{pelvis_{ZX}})\right]$$

(3.6.2.2.1.4)

The vertical toe position (where L_{swingz} = z; Figure 3.6.1.1) which results in no-change from the intended goal can be described by:

$$dz_i = z_i - z^0 \cong 0 \qquad (3.6.2.2.1.4)$$

The null space solution of the Jacobian matrix forms the dimensionality of the linearised manifold, and links the changes in the joint configuration to changes in the vertical toe position:

$$dz_i = J(\theta^0).d\theta_i = 0 \quad (3.6.2.2.1.5)$$

The form of $J(\theta^0)$. $d\theta_i$=0, can be expressed as:

$$\begin{bmatrix} -l_{thigh}\, sin(\theta_{thigh}) & -l_{shank}\, sin(\theta_{shank}) & -l_{foot}\, sin(\theta_{foot}) \end{bmatrix}. \begin{bmatrix} \theta_{thigh_i} - \bar{\theta}_{thigh} \\ \theta_{shank_i} - \bar{\theta}_{shank} \\ \theta_{foot_i} - \bar{\theta}_{foot} \end{bmatrix} = 0 \quad (3.6.2.2.1.6)$$

Where *i* represents the time slice of the swing cycle which corresponds to the reference joint configuration used to obtain the '*d x n'* Jacobian ($J(\theta^0)$), where, 'n' and 'd' represents the joint variables and task variable(s), respectively.

3.6.2.3 Step 3: Projecting the joint configuration onto the UCM$_{parallel}$ and UCM$_{perpendicular}$

The null space of the Jacobian matrix estimates the linearised UCM. This represents the set of joint configurations which lead to no change in the task (end-effector) variable. The null space ε_i is spanned by 'n-d' basis vectors, and is solved by:

$$J(\theta^0).\varepsilon_j = 0 \quad (3.6.2.3.1)$$

So, in the case of the first geometric model, the dimensionality of the null space, ε_j, has 3-1 = 2 basis vectors. The computation of the null space, ε_i, is repeated for each time slice because each time slice requires an estimated linear manifold. The basis vector is applied to the 'mean free' deviation matrix, $d\theta_i$, to obtain the projected variance of the joint configuration onto (i) the null space, or estimated manifold parallel to the UCM, and (ii) perpendicular to the estimated manifold, by:

$$\theta_{\parallel_i} = \sum_{j=1}^{n-d} [\varepsilon_j\, T.(d\theta_i)]\varepsilon_j \quad (3.6.2.3.2)$$

Where j represents the number of (i.e. n-d) basis vectors, and i represents the time slice.

$$\theta_{\perp_i} = (d\theta_i) - \theta_{\parallel_i} \quad (3.6.2.3.3)$$

3.6.2.4 Step 4: Computing the variance of $UCM_{parallel}$ and $UCM_{perpendicular}$

Once each cycle or trial of joint configuration was projected onto the linearised manifold, or the plane orthogonal to the manifold, the total variance along each dimension is calculated for each time 'slice' of the cycle, across all cycles. The result of the total variance is normalised to the degrees of freedom (n, d, N):

$$\sigma_{\parallel}^2 = (n-d).(N_{trials})^{-1}.\sum \theta_{\parallel}^2 \tag{3.6.2.4.1}$$

$$\sigma_{\perp}^2 = d^{-1}.(N_{trials})^{-1}.\sum \theta_{\perp}^2 \tag{3.6.2.4.2}$$

To obtain an estimate for the variance along the two components of the manifold, the $UCM_{\parallel}$ and $UCM_{\perp}$ were obtained by: $\sqrt{\sigma_{\parallel}^2}$ and $\sqrt{\sigma_{\perp}^2}$, respectively. These were the main two dependent variables used for further analysis. A third dependent variable, the ratio of ($UCM_{\parallel}/UCM_{\perp}$), characterised the relative amount of task redundant variance with respect to the task-irrelevant variance.

The variables investigated are:

- $UCM_{\parallel}$ deviation normalised to degree of freedom
- $UCM_{\perp}$ deviation normalised to degree of freedom
- Ratio $UCM_{\parallel}/UCM_{\perp}$

3.6.2.5 Input data

All UCM variables were computed at time states described previously (section 3.5.4). Briefly, the UCM was computed at five sequential 3% increments which neighboured the lead-in, and lead-out, from the reference states of MX1 and MTC. The time slices used for computing the UCM are state-dependent and relevant to hypothesised task goal events of MX1 and MTC.

The UCM is computed around the mean joint configuration from a given sample of trials. This mean configuration essentially represents the goal state of the effector system. The UCM does not depend upon a temporal sequencing order and therefore trials could be categorised as discrete occurrences based upon different trial ranking criteria. Because the time slices are goal-dependent, it is assumed that the movement task goal at each time slice remains constant between trials. This is important in the computation of the UCM hypothesis. To ensure that this assumption has greater merit, the hypothesised task goals expressed by the effector trajectories within a swing trial were re-sampled into a discrete set of like trials. The criteria for re-ordering the swing trials were based upon initial conditions of the stance (or swing) end-effector at MX1 and the 'final' conditions at MTC. This ranking of trials was based upon hypotheses listed in Table 3.6.2.5.1 and illustrated in Figure 3.6.2.5.1.

Table 3.6.2.5.1.

Criteria for trial re-sampling into nine subgroups

	Re-sampling criteria 1: based upon initial condition	Re-sampling criteria 2: based upon the response made from initial condition
Stance Effector	Vertical displacement of the stance effector at MX1	Vertical displacement of the stance effector between MX1 and MTC
Swing Effector	Vertical displacement of the swing effector at MX1	Vertical displacement of the swing effector between MX1 and MTC

Trials were excluded during the first wave of trial rankings if the vertical swing hip position was outside of 5-95% of the re-ordered ranked trials (Figure 3.6.2.5.1). Within these three sub-groups, a second wave of trial re-ordering of trials based upon rankings of a second parameter, in this example, those which had a vertical MTp position outside of 5-95%. So, the first wave included three subgroups according to hip height (H_{high}, H_{ave}, H_{low}) and these sub-groups were further sub-divided into three sub-subgroups according to toe height position (T_{high}, T_{ave}, T_{low}). The number of trials employed for computing a UCM ($n \approx 50$) is consistent with other UCM studies (e.g., Tseng et al. (2006)).

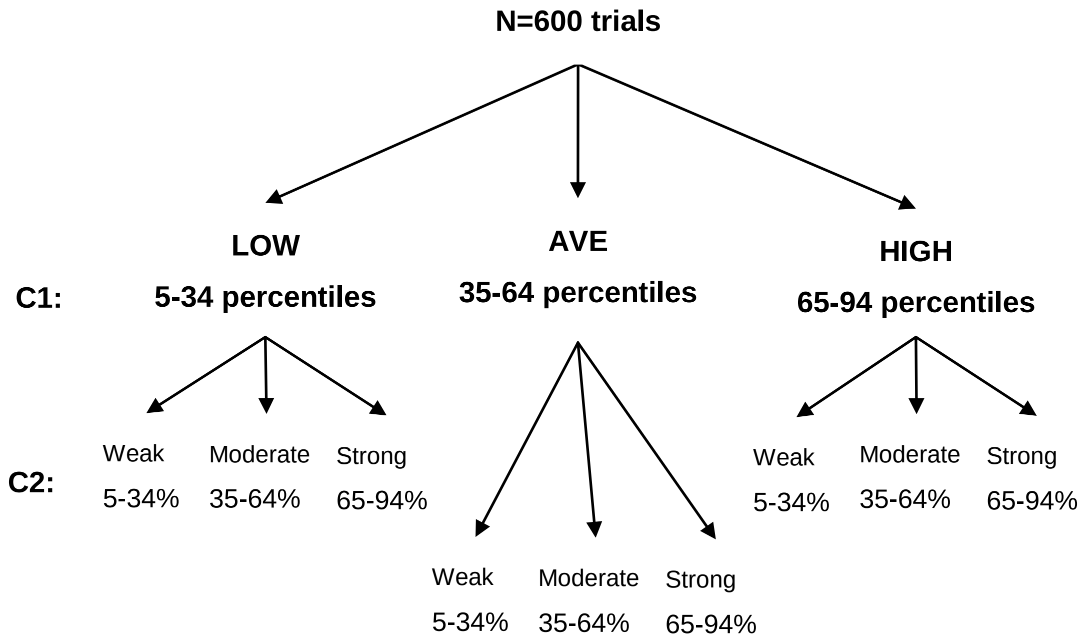


Figure 3.6.2.5.1

Subdivision of N swing trials into nine groups. From the full number of trials, the data was ranked according to a criteria listed in Table 3.8.2.5.1. Trials were subsequently ranked in order of percentiles according to the criteria, i.e. 5-34%, 35-64%, 65-94%. Data values located within lowest and highest five percentiles of the re-ordered trials were considered possible outliers to the group goal and thus they were excluded from analysis. Within each of these three groups, a second ranking criteria was performed based upon criteria 2. From this each group was further subdivided into three subsequent groups, also excluding extreme trials beyond 5-95% of the re-ordered trials. For example, nine subgroups created from N=600 trials results in approximately n≈60 trials allocated to each of the nine re-sampled groupings.

The dilemma is the identification of which control parameter does the controller pay attention to at the MX1 initial condition and which control parameter does the controller pay attention to for the necessary response. There are multiple scenarios which could represent the state estimate of interest to the controller's policy. The control parameter representing the required state at some initial condition is hypothesised to be associated with the effector state at MX1, and this could be the combination of the following: 1) the vertical displacement of the stance effector (external reference); 2) the length of the swing limb (internal reference); or 3) the toe-to-ground clearance height (i.e. combined effector - external reference). Because of the difficulty obtaining an indication of the controller's policy, the criteria was specific to the effector system and assumed that the effector was configuring its variance to control its effector endpoint.

The trials within each of the nine subgroups were used to calculate the Jacobian and null space based upon the mean of the segment angles which represent the mean 'end-effector' value.

3.6.2.5.1 Distribution of the re-sampled data

The three methods for sub-grouping the original data into nine sub categories involve data re-sampling. Following the first rank-order re-sampling criteria, each of the three sub-groups will contain a 'biased' sample of trials with distribution characteristics which are vastly different in terms of the skewed description of the data. This of course will most heavily affect those variables which are the basis of the re-sampling criteria. However, the other two variables will be also affected because of a co-dependent relationship (demonstrated by results of Section 7.3). For example, after the first criteria for re-sampling, the first and third distributions will not be normally distributed. This is because this ranking divides the distribution into three by partitioning the two 'tail-ends' from the middle section of the normally distributed. Hence, the two 'tail-ends will be strongly negatively, and positively skewed. The second distribution will be less skewed, however, it will have a kurtosis due to a peaked nature. Given that there is a variable response to initial conditions by this second 'grouping' parameter, the new rank-ordered samples will have diluted, to some extent, the bias of the 'parent' distribution even though these 'sibling' samples are taken from a non-normal 'parent' distribution. This of course will be dependent upon the variant response of the parameter to the initial conditions.

3.7 Detecting serial correlations in effector trajectory states

Various methods are available for quantifying correlations among consecutive behaviours displayed in physiologic data. These correlations might change over different time scales, such that correlations in the short term are different to correlations over longer time scales. To reliably detect long-range correlations in experimental data it is important to consider the effect of slowly oscillating trends (i.e. changing mean and standard deviation) intrinsic to the time-series data. These trends can cause the false detection of long-range correlations if not accounted for by the long-range correlation detection method. The Detrended Fluctuation Analysis method has the ability to account for such trends that might contain a different order from linearity (Kantelhardt, Koscielny-Bunde, Rego, Havlin and Bunde, 2001).

The correlations within (e.g. an effector) time series of (s_i) of i = 1, ... , N equidistant measurements, where i corresponds to a time cycle, can be determined by quantifying the correlation between s_i and s_{i+l} for different time lags, l, (i.e. correlations occurring over different time scales). Quantitatively, correlations can be calculated by the (auto-) correlation function:

$$C(l) = \langle \bar{s}_i \bar{s}_{i+l} \rangle = \frac{1}{N-l} \sum_{i=1}^{N-l} \bar{s}_i \bar{s}_{i+l} \qquad (3.7.1)$$

If the s_i is uncorrelated, then $C(l)$ is zero for $l > 0$. For short-range correlations, $C(l)$ will decline exponentially, $(l) \sim exp(-l/t_\times)$, with decay time $t_\times$. For long-range correlations, $t_\times = \int_0^\infty C(l)\, dl$ diverges and the decay time cannot be defined. In essence $C(l)$ declines as a power-law: $C(l) \propto l^{-\gamma}$, with an exponent $0 < \gamma < 1$ (Kantelhardt et al., 2001). This implies that the value at one time cycle is explicitly dependent upon a cycle occurring at potentially hundreds of cycles earlier, whose influence decays in a nonlinear fractal-like manner (Hausdorff, Mitchell, Firtion, Peng, Cudkowicz, Wei and Goldberger, 1997b). This defines long-range correlations derived from fractal-like, long-term dependence, processes. However, physiological time-series can show long-range correlations in methods quantifying serial correlations, while the time-series has not explicitly been derived from fractal-like long-term dependent processes (Wagenmakers, Farrell and Ratcliff, 2004). Regardless of the underlying processes, the quantification of long-term correlations, or persistence, reveals details about the statistical likelihood that deviations from cycle-to-cycle are directional. If deviations are likely to persist in one direction from one cycle to the next, then the correlations are defined as persistent. When a deviation in one direction is likely to be followed by a subsequent deviation in the opposite direction, the correlations are defined as anti-persistent (Dingwell and Cusumano, 2010).

3.7.1 Detrended Fluctuation Analysis method

In this thesis, Detrended Fluctuation Analysis (DFA) was the applied method to quantify the statistical persistence, or long-range correlations, in the time-series of effector trajectories (i.e. where they intersect discrete task-relevant time-slices of the swing cycle). The DFA method has been used extensively to characterise the serial correlations of gait variables since the finding that walking was not a random process (Hausdorff et al., 1995), but it is a new method applied to the characterising of time-slices along gait trajectories in three-dimensional space. The DFA method is based upon a root mean square analysis of a random walk (Hausdorff et al., 1996). It is similar to Hurst's rescaled range R/S analysis (Hurst, 1951). For a given time series $\{S(i)\}_{i=1}^{N}$ the scaling exponent 'α' can be estimated from the following steps, which have been adapted from Bashan et al. (2008):

Step 1: integration.

- To enable an analysis of time series data with weak correlations, we created the random walk profile, $Y(n)$ by integrating the original time series $X(n)$ by partial summation:

$$Y(n) = \sum_{i=1}^{n}[x_i - \langle x \rangle] \text{ for } i = 1, 2, \dots, N \qquad (3.7.1.1)$$

- where x_i is the i^{th} value of the time series, and $\langle x \rangle$ refers to the mean of the time series.

Step 2: Window fitting.

- This profile is then divided into non-overlapping 'time-window' segments ('time scale') of equal length l. For a given segment window size l, divide the integrated series $\{Y(n)\}_{i=1}^{N}$into $N_l \equiv \left[{}^{N}/_{l}\right]$ non-overlapping segments of equal length l, where $[N/4]$ is the greatest integer function. The value of N_l does not need to be a multiple of the time scale l, and therefore a short portion at the end of the time series will remain in such cases ($N_R = N - \left[{}^{N}/_{l}\right]$). The first non-overlapping segment will include the first data point in the time series, $Y1(n = 1{:}\,N_R)$. To account for this portion, a second time series can be created now from the point, $n = N_R$, i.e. $Y2(n = N_R{:}\,N)$. Therefore, $2N_l$ segments are created and hence there will exist overlapping mutually inclusive data points amongst $Y1\ and\ Y2$.

- In this thesis, the set of l segment lengths were fixed at {9, 17, 33, 65, 129}. The smallest segment length of nine was selected following a recent investigation of the DFA method (Bashan et al., 2008).

Step 3. Detrending.

- The benefit of the DFA method is dealing with the effect of non-stationarities, i.e. trends common in experimental biological time series data which can mask the correct scaling behaviour of a system. This effect is reduced by performing a detrending procedure on the integrated random walk time series to obtain a new time series $\{\tilde{Y}_l(n)\}_{n=1}^{N_l}$. This is done by estimating a piecewise polynomial trend within each segment l by a least-squares fitting function, $y_l^{(p)}(n)$. The function which estimates the polynomial trend within each segment consists of polynomials of order p which are determined separately and concatenated from segment-to-segment, i.e. polynomial curve fitting in adjacent windows are unrelated. The detrending profile function constitutes the trend series, where $N1_l \equiv \left[{}^{N}/_{l}\right]$ is the total number of data points falling within the segment boxes, i.e.

$$\tilde{Y}1_l(n) = y(n) - y_l^{(p)}(n)\ for\ n = 1, \ldots, N - N_R \tag{3.7.1.2}$$

- A second time series $\tilde{Y}2_l(n)$ will also be created to account for step 2.

$$\tilde{Y}2_l(n) = y(n) - y_l^{(p)}(n)\ for\ n = N_R, \ldots, N \tag{3.7.1.3}$$

- The degree of the polynomial can be altered to account for the best fit within the segments, i.e. linear (p=1), quadratic (p=2), or higher order (p >2) trends. The DFA is usually named after the polynomial fitting, i.e. DFA1 (p=1), DFA2 (p=2), etc. This study only applied linear fitting (DFA1, where $p=1$) to detrend the integrated time series.

Step 4: Window variance.

- In the fourth step, a data series $(v = 1{:}\, 2N_l)$ represents the variance for each of the segments $2N_l$ and is calculated by:

$$F_l^2(v) = \langle Z_l^2(n) \rangle = \frac{1}{l} \sum_{i-1}^{l} Z_l^2\,(v; l; n) \tag{3.7.1.4}$$

where,

$$Z_l^2 = \{\tilde{Y}1_l^2[(v-1)l + n]{:}\, \tilde{Y}2_l^2[(v-1)l + n]\}$$

- of the concatenated detrended time series' $\tilde{Y}1_l(n)$ and $\tilde{Y}2_l(n)$ by averaging over all data points n in the v^{th} segment.

- The variance of $\{\tilde{Y}_l(n)\}_{n=1}^{N_l}$ yields the fluctuation function, $F(l)$, for each segment l. Where, the detrended time series data values squared, summed, averaged and then square rooted (i.e. least-squares procedure).

$$F(l) = \sqrt{\frac{1}{2N_l} \sum_{v=1}^{2N_l} [\tilde{Y}1_l(n){:}\, \tilde{Y}2_l(n)]^2} \tag{3.7.1.5}$$

Step 5: obtain average fluctuations per window size.

- The steps 2-4 are then repeated for different segment sizes, $l_1, l_2, l_3, \dots, l_m$ to get a range of average fluctuations $f(l)$, as a function of m segment sizes, $f(l_1), \dots, f(l_m)$.

- Step 6: obtain scaling exponent by plotting average fluctuations per window size.

- Plot $\log f(l_i)$ verses $\log(l_i)$, for $i = 1, \dots, m$. The linear relationship on the double log plot is determined by a least squares fitted slope and provides an estimate of the scaling exponent 'α'.

The above DFA procedure was performed using a custom-written script for MATLAB (The MathWorks Inc.).

3.7.2 Auto-Correlation Function method

The Auto-Correlation Function described above (equation 3.7.1) was applied to assess the short-term correlations, or persistence, for time-lag of one cycle. Therefore, for time series s, quantifying the correlation between s_i and s_{i+l} for time lag one, we set $l=1$. Quantitatively, correlations are then calculated by the (auto-) correlation function:

$$C(1) = \langle \bar{s}_i \bar{s}_{i+1} \rangle = \frac{1}{N-1} \sum_{i=1}^{N-1} \bar{s}_i \bar{s}_{i+1} \qquad (3.7.2.1)$$

3.7.3 Input data

Detrended fluctuation analysis (DFA) and the auto-correlation coefficient of time-lag one (ACF_{lag1}) was performed on the three component dimensions that describe the three-dimensional effector vectors: stance, swing and combined. Each effector time series, s_i, of the data was extracted following the same method for obtaining discrete time-slices as per section 3.5.4.

Describing kinematic states of the effector trajectories from distribution statistics

4.1 Background

The first step towards exploring how lower limbs are managed when performing the toe clearance task is to observe the statistical features of the states of the effector trajectories in three-dimensional space. This section explores the states of the trajectories of the lower-limb effectors and lower limb segment angles (as described in section 3.6). Basically, the effector systems are being represented by the trajectory of a vector in three-dimensional space. At a discrete time event along the swing-cycle time-course the state of the segment or effector trajectory can be described. The discrete events selected were proposed to have meaning for the toe-clearance task by describing the events as task-relevant states (MX1 and MTC) and task-relevant time-slice increments. The task-relevant time-slice increments are equally spaced time-intervals between time-slices with units of time representing a percentage of the time-normalised swing phase. Because the motion state of the effectors in the vertical dimension are identical at the task-relevant events MX1 and MTC, the time-slice events selected on the trajectory represent a sequence of state-relevant time-slices. Each cycle-to-cycle ‘gait’ trajectory intersects through state-relevant time-slices in a repetitive way, providing a distribution of ‘states’ from which inferential statistics can be applied to describe the average time-course of trajectory states. This chapter investigates the statistical properties of these segment and effector ‘state’ distribution sets at MX1 and MTC swing-phase regions.

Describing the average time-course of lower-limb mechanical states is important because the control of gait trajectories depends upon exploiting optimal kinematic lower-limb configurations (Grimmer, Ernst, Gunther and Blickhan, 2008). The importance of this chapter in the context of this thesis is that it will provide a starting point for gathering further evidence about how the locomotor system manages the lower limb effectors for performing the toe clearance task. Additionally, the analysis in this chapter will contribute knowledge about the toe-clearance task that is lacking in the literature. For example, most studies investigate average gait trajectories according to normalised-time rather than state-relevant events. Averaging gait trajectories based upon normalised-time has certainly been recognised as a problem in gait analysis (Forner-Cordero et al., 2006). The approach to defining effector states in this thesis was based upon a theory that the brain predominantly receives feedback mostly related to details of limb (effector) length and orientation, and less related to segment states (Bosco et al., 2005; Bosco et al., 2006). The collective states of the stance and swing effector trajectories produce the combined effector state of MX1 and MTC, representing events of maximum and minimum limb length respectively. Therefore, MX1 and MTC are states defined by the combined effector trajectory, and these states are relevant to the task-goal of toe-clearance. Observing gait trajectories at time-slices relevant to the MX1 and MTC states is a new approach in the gait literature.

4.2 Method

This section will first detail the average trajectories of the effectors and their segment components. Second, the shape of the distribution (i.e. standard deviation and skewness) at the same state-relevant time-slices will investigate how the loco-sensorimotor system manages the statistics of the state of the effector trajectories.

Participants walked on a treadmill for ten minutes at their self-selected walking speed. The number of gait cycles from this extended period of walking varied between subjects due to step-cadence differences, but on average it provided approximately 600 swing-cycles from the left and right limbs. From each time-normalised swing-cycle, MX1 and MTC events were extracted as described in section 3.5.4. From many swing cycles, the effector trajectories formed a distribution of points at each state-relevant time-slice neighbouring MX1 and MTC (i.e. at time intervals of 1% swing phase time). The MX1 region was described by 12 equally spaced state-relevant time-slices before and after MX1. Similarly, the MTC region was neighboured by 15 equally spaced time-slices before and after MTC (see 3.5.4 and Figure 3.5.4.1). To account for individual limb length differences, all three vector component dimensions were normalised according to the combined segment lengths of the shank and thigh.

Graphical data allows an easier interpretation of ageing differences occurring within a swing cycle as compared to extensive data tables. Plots were created to describe a ‘point-by-point’ comparison between young and older adults for a measured gait parameter. Similar graphical procedures have been proposed by gait studies investigating statistical differences between pathological and normal gait (Schwartz, Trost and Wervey, 2004). The plots have been employed to detect where between-groups significant differences exist along state-relevant time-course of the swing-cycle. A failure to reach statistical significance is presumed to represent a between-group difference too small to be confidently considered an effect of ageing. In this way, the effect of ageing is analysed graphically at points along the swing phase cycle.

In the figures, statistical significance will be assumed when there is a 90% probability that the null hypothesis is wrong. This will be indicated graphically when the elderly curve (blue line) exceeds the boundary of a 90% confidence interval (red line) about the young mean. A red dashed line will represent the 90% confidence interval for a difference between independent samples (i.e. $t(54) = 1.67$); and where $\overline{X}_E \leq$ 90%CI, or when $\overline{X}_E \geq$ 90%CI, is an estimate of the 10% likelihood ($p = .1$) that a between-group mean difference is due to sampling (Welkowitz et al. 2006). The plots represent the group mean and the error bars indicate the respective 95% confidence interval ($t(27)=2.052$) about the group mean at each time-slice. (An example of the plot is described in Figure 4.2.1.; for time slice example see Figure 3.5.4.1.) This 'mean difference' confidence curve is in essence likened to a series of multiple independent sample t-tests. A conservative 90% confidence interval being chosen is less inclined to accept a 'type-II' statistical error, of falsely accepting the null hypothesis of no significance between groups. An opposite view is that there is a greater chance of a type-I error, whereby rejecting a null hypothesis is a false outcome. The view of a sample representing the ageing population is that there are diverse inter-subject differences due to diverse physical affects from ageing. With respect to gait, there is a more likely chance that gait patterns within participants – in an older adult sample of the population – has changed compared to others, whom may not have changed much at all, and this will likely exist more-so than a young subject sample. Type-II errors of making a false decision by rejecting the null hypothesis, and potentially overlooking when an ageing difference does exists, is not controlled for explicitly. The increased possibility of a 'type-II' error is anticipated to exist for a smaller sample sizes (i.e. N<28) at the 95% and above level confidence interval because the healthy older sample selected can generally be considered a section of the general older adult population. Therefore, the p-value associated with a 90% confidence interval that there is a 10% probability that rejecting the null-hypothesis is incorrect was accepted, based upon the prevention of overlooking results that could potentially carry meaningful information about ageing changes to gait.

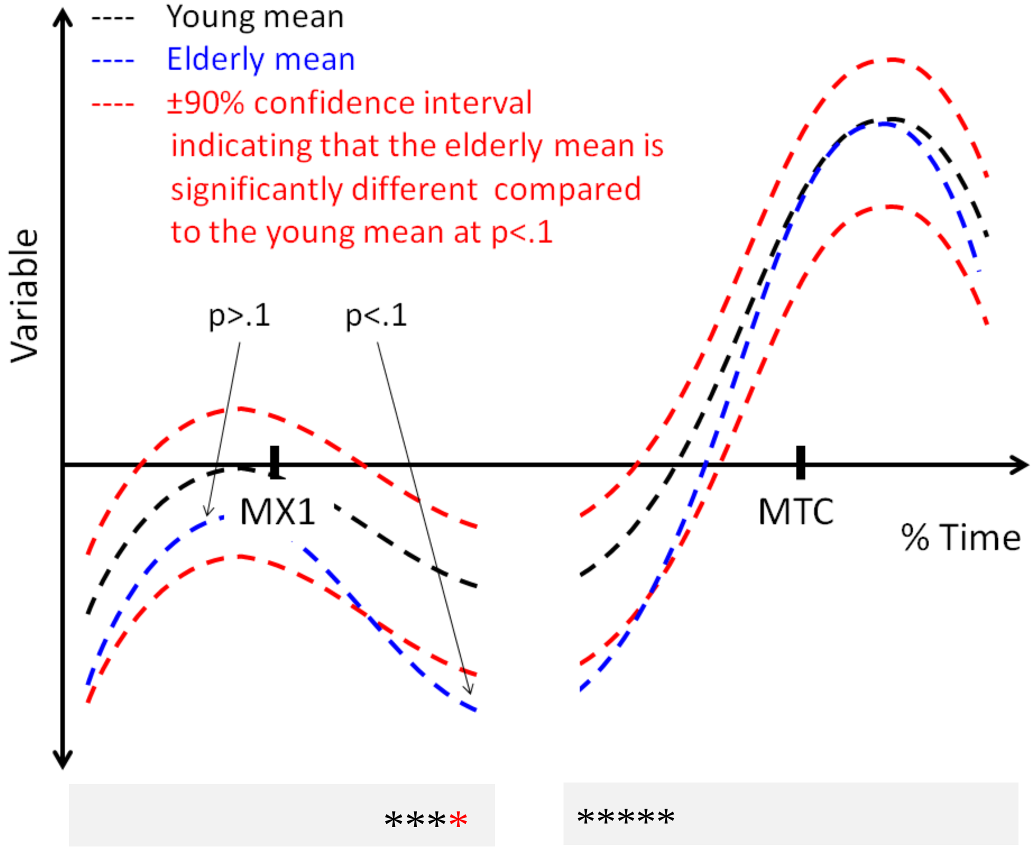


Figure 4.2.1

A hypothetical example of a plot representing a graphical form of an independent sample t-test, that compares for age differences at the $p<.1$ level. A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

4.2.1 Hypotheses and statistical design

This chapter will test the null hypotheses related to stride-to-stride distribution statistics of effector variables outlined below, which are linked to the aims described in section 2.5. For each hypothesis, the dependent and independent variables are indicated in parenthesis, (DV) and (IV) respectively. Each null hypothesis has a description of how statistical significance was tested.

4.2.1.1 Time-distance

- There will not be a significant difference in time-distance parameters (DV) of the gait cycle between age groups (IV).
    - The dependent variables pertain to parameters that capture the cycle-to-cycle step length and step width time-series: average; standard deviation; and Detrended Fluctuation Analysis.
    - Statistical design is independent sample t-tests, within each limb type (dominant and non-dominant). Significance is set at $p<.05$.

4.2.1.2 Effector-state: average

- There will not be a significant difference in segment angles (DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - Dependent variables are mean three-dimensional segment angles of the pelvis; the stance shank and thigh; and the swing foot, shank and thigh.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.
- There will not be a significant difference in effector configuration (DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - The dependent variables are mean components of three-dimensional effector vectors: stance; swing; and combined.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.

- There will not be a significantly different mean trajectory from MX1 to MTC between the elderly and young group for the vertical dimensions of the three effectors.
    - The dependent variables are standard deviations in components of three-dimensional effector vectors: stance; swing; and combined.
    - Statistical test design is analysis of a main effect for group from a two-way (group × state) mixed-design ANOVA (with repeat measures for 'state'). Significance is set at $p<.05$.
- There will not be a significant difference in mean effector velocity (DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - The dependent variables are mean components of three-dimensional effector vector velocities: stance; swing; and combined.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.

4.2.1.3 Effector-state: variance

- There will not be a significant difference in standard deviation of the effector trajectories (DV) when comparing between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - The dependent variables are standard deviation in components of three-dimensional effector vectors: stance; swing; and combined.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.

- Effector skewness (DV) will not be significantly different between the three effectors (IV-1) when comparing within swing phase states MX1 and MTC (IV-2).
    - The dependent variables are standard deviation in the vertical component of three-dimensional effector vectors: stance; swing; and combined.
    - Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.
    - Statistical test design is two separate one-way ANOVA for IV-1 and between effector differences assessed by post-hoc Bonferroni tests (significance set at $p<.05$) within each group (collapsing limb data).
- Effector skewness (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures) and within age groups (IV-3).
    - The dependent variables are standard deviation in the vertical component of three-dimensional effector vectors: stance; swing; and combined.
    - Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.
    - Statistical test design is three separate paired t-tests for IV-2 ($p<.05$) within each group.
- There will be no significant difference between young and older adults (IV-1) in the way that the standard deviation (DV) changes in the vertical dimension between effectors (IV-2) from states MX1 to MTC (IV-3; repeated measures).
    - The dependent variables are standard deviation in vertical components of three-dimensional effector vectors: stance; swing; and combined.
    - Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.

- Statistical test design is analysing an interaction effect between 'effector × state × group' using a three-way mixed-design ANOVA (with repeat measures for 'state'). Significance is set at $p<.05$. Planned comparisons between effector pairings of stance-combined and swing-combined are presumed significant at $p<.05$.

4.2.1.4 Effector-state: skewness

- There will not be a significant difference in skewness of the effector trajectories (DV) when comparing between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - The dependent variables are skewness in components of three-dimensional effector vectors: stance; swing; and combined.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.
- Effector skewness (DV) will not be significantly different between the three effectors (IV-1) when comparing within swing phase states MX1 and MTC (IV-2).
    - The dependent variables are skewness in the vertical component of three-dimensional effector vectors: stance; swing; and combined.
    - Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.
    - Statistical test design is two separate one-way ANOVA for IV-1 and between effector differences assessed by post-hoc Bonferroni tests (significance set at $p<.05$) within each group (collapsing limb data).
- Effector skewness (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures) and within age groups (IV-3).

- The dependent variables are skewness in the vertical component of three-dimensional effector vectors: stance; swing; and combined.
- Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.
- Statistical test design is three separate paired t-tests for IV-2 ($p<.05$) within each group.

♦ There will be no significant difference between young and older adults (IV-1) in the way that the skewness (DV) changes in the vertical dimension between effectors (IV-2) from states MX1 to MTC (IV-3; repeated measures).

- The dependent variables are skewness in vertical components of three-dimensional effector vectors: stance; swing; and combined.
- Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.
- Statistical test design is analysing an interaction effect between 'effector × state × group' using a three-way mixed-design ANOVA (with repeat measures for 'state'). Significance is set at $p<.05$. Post-hoc comparisons between effector pairings of stance-combined and swing-combined are defined as 'planned contrasts' and are significant at $p<.05$.

4.3 Participant characteristics

The general anthropometric characteristics and walking speed of the participants are described in Table 4.3.1. Significant differences between the groups are indicated where p>.05. Older adults (1.17 (0.04) m/s) walked at nearly the same speed as younger adults (1.19 (0.04) m/s), p = .18.

The body-mass-index (BMI) and Body Mass variables were significantly skewed within the young and elderly groups respectively. Therefore, a non-parametric independent samples t-test (i.e. the Mann-Whitney Wilcoxon U-test) was applied for these parameters. The significant difference in BMI indicates that the elderly have a larger mass with respect to body height (p<.05). Height was significantly higher for the young group (p<.05), but limb-length did not reveal a significant difference (p>.05). This appears strange, however, it is likely to be reflecting the age-reduction of inter-vertebral structures have degenerated with age (Spirduso et al., 2005).

Table 4.3.1.

Participant anthropometric characteristics.

	Group				Statistic	
	Young (N=28)		Elderly (N=28)		t	P value
	Mean	SD	Mean	SD		
Age (yrs)	24.6	5.9	68.5	3.4	34.14 (t)	**.00***
TM speed (m/s)	1.19	.04	1.17	.04	1.37 (t)	.18
Height (cm)	167.1	6.9	161.8	6.5	2.93 (t)	**.00***
Limb Length (m)	.79	.05	.77	.05	1.63 (t)	.11
Body Mass (kg)	62.4	9.9	67.6	10.4	U*	.11
BMI (kg/m^2)	22.3	3.1	25.8	3.5	U*	**.03***

TM = treadmill; BMI = body mass index; t = t-statistic from independent sample t-test; U* = only the p value is reported for the Mann-Whitney Wilcoxon U-test in IBM-SPSS-Statistics-version19 software package.

4.4 Step time-distance parameters

The step cycle time-series of time-distance parameters were characterised by the mean, standard deviation and the DFA. These were compared between the two subject groups (Table 4.4.1). An independent samples t-test showed a between-group difference for standard deviation of the step width and step length when comparing within the preferred and non-preferred limbs. There was no difference found for the step time variability defined by standard deviation or the DFA. In contrast to the standard deviation, the dynamic variability for step width, step length, step time did not identify a significant difference between the two age groups.

A repeated measures t-test was conducted to compare the dependent variable differences between the step length and step width variability. Both the young and elderly groups show a significant difference between the step length and step width variability for both the DFA and SD parameters ($p <.001$) (Figure 4.4.1). The step length parameter for both the young and elderly participants is substantially higher than the DFA value of ~0.63 found by Jordan et al. (2006) for treadmill walking across a similar time period. This is based upon the value of 0.63 existing outside of the 95% confidence intervals reported in Figure 4.4.1.

Table 4.4.1

Age differences for intra-subject time-distance parameters: Mean, Standard Deviation (SD) and Detrended Fluctuation Analysis (DFA, α).

			Group				Statistic	
			Young		Elderly		t	P value
			Mean	SD	Mean	SD		
Mean	Dominant Limb	Step Length (/LL)	.880	.074	.878	.078	.126	.900
		Step Width (/LL)	.116	.038	.122	.032	.692	.492
		Step Time (s)	**.510**	**.032**	**.494**	**.025**	**2.123**	**.039***
	Non-dominant Limb	Step Length (/LL)	.888	.072	.891	.077	.190	.850
		Step Width (/LL)	.116	.038	.122	.032	.692	.492
		Step Time (s)	**.514**	**.031**	**.496**	**.027**	**2.338**	**.023***
SD	Dominant Limb	**Step Length**	**.014**	**.003**	**.019**	**.004**	**4.385**	**.000***
		Step Width	**.025**	**.004**	**.030**	**.006**	**3.674**	**.001***
		Step Time (s)	.009	.001	.010	.002	1.510	.137
	Non-dominant Limb	**Step Length**	**.014**	**.003**	**.020**	**.005**	**5.613**	**.000***
		Step Width	**.025**	**.004**	**.030**	**.006**	**3.436**	**.001***
		Step Time (s)	.010	.002	.010	.002	.747	.458
Skew	Dominant Limb	**Step Length**	**-.132**	**.321**	**-.325**	**.393**	**2.012**	**.049***
		Step Width	.141	.065	.177	.162	1.091	.280
		Step Time (s)	.020	.207	-.017	.170	.739	.463
	Non-dominant Limb	Step Length	-.156	.322	-.341	.502	1.637	.108
		Step Width	.127	.129	.120	.141	.181	.857
		Step Time (s)	.038	.220	-.060	.253	1.560	.125
DFA	Dominant Limb	Step Length (α)	.677	.109	.680	.100	.092	.927
		Step Width (α)	.527	.066	.540	.080	.668	.507
		Step Time (α)	.644	.098	.641	.091	.110	.913
	Non-dominant Limb	Step Length (α)	.700	.096	.674	.095	1.055	.296
		Step Width (α)	.534	.065	.535	.080	.002	.998
		Step Time (α)	.625	.093	.653	.093	1.130	.263

* between group difference is significant at p<.05 level.

When comparing group average statistics of the dominant and non-dominant limbs within the groups, a paired sample t-test found no significant differences (p>.1). This doesn't indicate that participants are considered symmetric for step-time, step length and step width. This would need to be assessed by a within-subject asymmetry statistic, which is outside the scope of the aims set out for this thesis. The result simply indicates that for time-distance parameters of walking, the non-dominant limb does not perform significantly different compared to the dominant limb for distribution statistics and the DFA statistic.

Figure 4.4.1 pools the dominant and non-dominant limbs together and illustrates group mean ±95% confidence intervals for normalised step length and step width, when comparing the standard deviation and DFA statistic between the groups. In addition to supporting the information from Table 4.4.1, the graph below (Figure 4.4.1) demonstrates within the groups, that the DFA is significantly higher for step length compared to step width ($p<.001$). Additionally, within the groups, step width standard deviation is significantly higher compared to step length ($p<.001$).

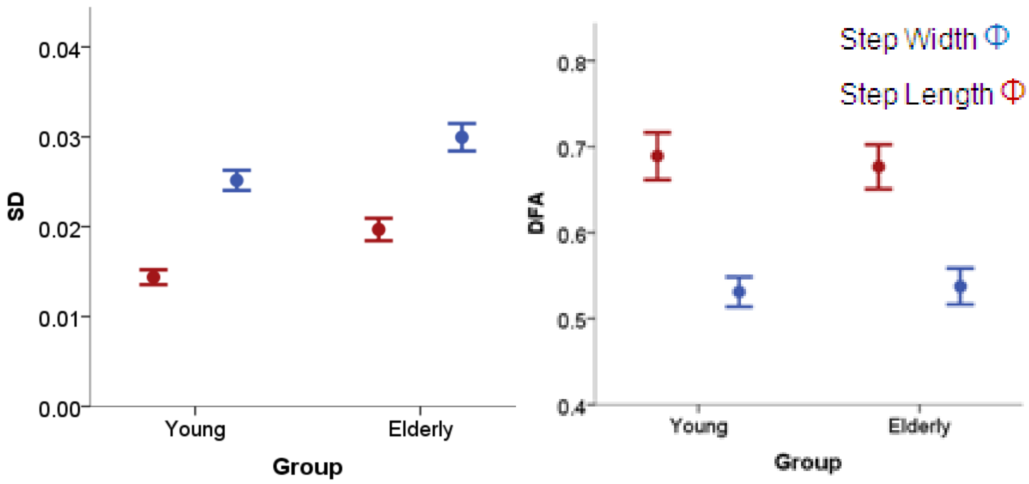


Figure 4.4.1.

A comparison between step length and step width variability within the young and elderly group. Within each group, the data from the dominant and non-dominant limbs have been pooled into one group. The symbols indicate the group mean and the error bars represent 95% confidence intervals of the group mean.

4.5 Qualitative description of the three-dimensional components of the three effectors

Figure 4.5.1A illustrates the young group mean three-dimensional signals for the three effector systems of stance, swing and combined. Each effector's position data is plotted across the two swing phase regions of interest: MX1 (26 data points) and MTC (31 data points). Although the plot is not a function of time, the gap in the signal represents the time separation between the two swing-phase regions, MX1 and MTC. So, the effectors three-dimensional coordinates at each sampled time slice of the swing phase are plotted in effector normalised units in three-dimensional space. The mean signals are zero-shifted in three-dimensional space to illustrate the shape of the effector paths. This is because in real three-dimensional space, the signal of the stance effector has a vertical difference with respect to the swing effector (and combined effector) of approximately one limb-length. Therefore a plot would diminish an illustration of the relative curve features between effectors. Figure 4.5.1B compares the young and elderly profiles of the three effectors without signal-normalising to one standard deviation of the mean signal. This preserves the effector curve for inter-group comparison purpose.

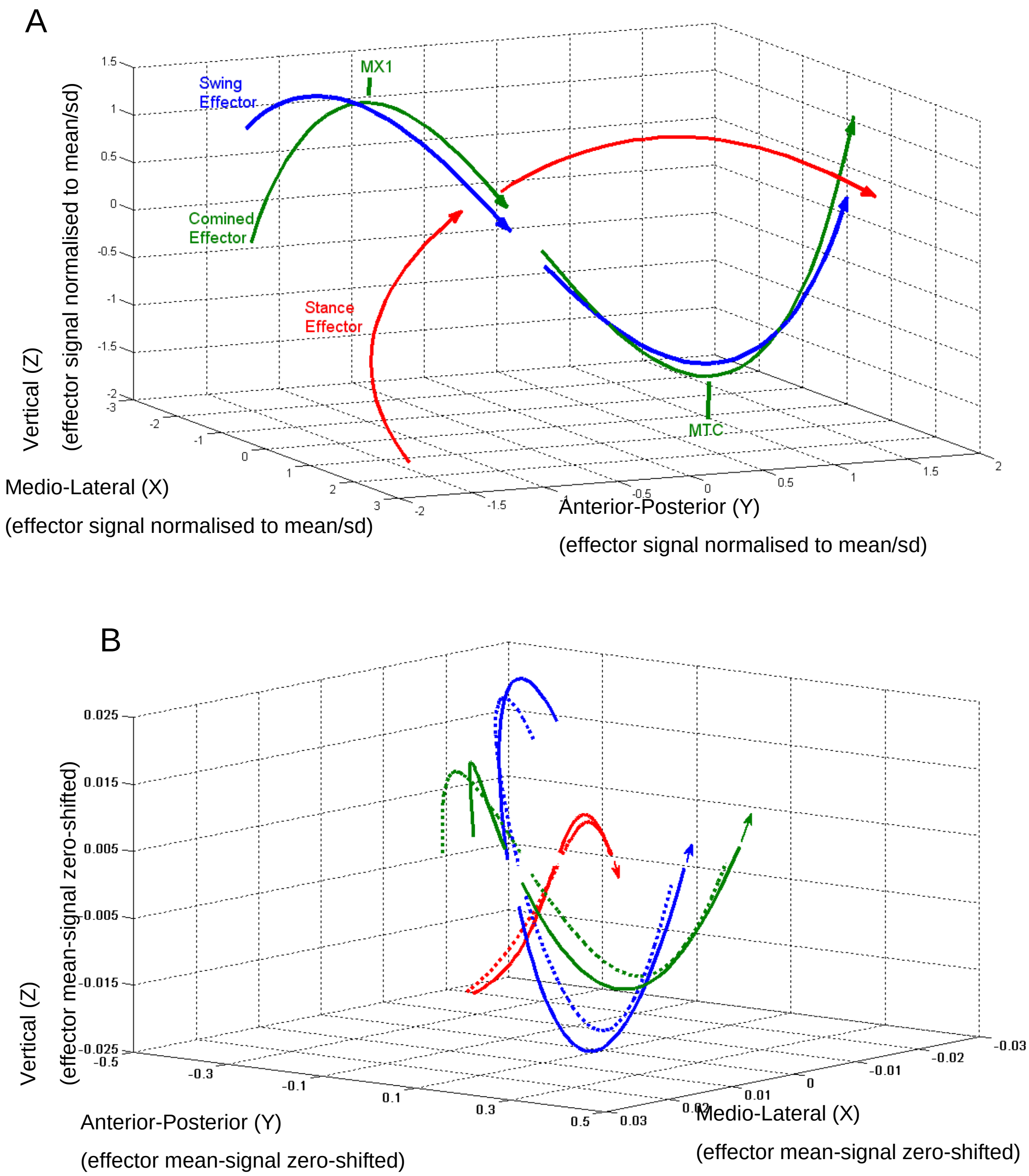


Figure 4.5.1

Two perspectives of the three-dimensional spatial plots of the effectors (see text). A) Average profile of the young group with signal normalised to mean and standard deviation. B) Comparison between young (solid line) and elderly (dotted line) when group mean signal is zero-shifted. NB: the stance effector moves medially (i.e. ‘into the page’) at MX1 region and then laterally (‘away from the page’) following MTC.

A general description of the effector behaviour with reference to Figure 4.5.1A is as follows. At the beginning of the MX1-region, the stance effector – as described by a three-dimensional vector directed from the stance-foot to the swing-hip position – moves medially or closer to the stance foot (i.e. 'into the page'). This medial motion might best be observed from a frontal view perspective in Figure 4.5.1B. Concurrently the stance effector vector initially moves slightly downward before it begins increasing its upward length as the MTC-region approaches. Alternatively, the 'inverted' swing effector vector – as described by a three-dimensional vector directed from the swing-hip to the swing-toe – is reducing its downward length at the beginning of the MX1 region. While in opposition to the medially moving stance effector, the swing effector is moving laterally (i.e. 'away from the page') as MX1 approaches. The resultant of these two effector systems is observed in the behaviour of combined effector system – as described by the three-dimensional vector directed from the stance-foot to the swing-toe. Both stance and swing effector vectors are lengthening in the anterior direction throughout the swing phase. The interaction of the stance and swing effector systems causes the combined effector vector to move upwards and slightly lateral at initial MX1-region. There is an indication in the medio-lateral view from Figure 4.5.1B that the elderly combined effector vector has a larger lateral displacement at the beginning of the MX1 region. This indicates the original step width position at pre-swing, although step width wasn't found to be significantly different between groups (see previous section 4.2). Figure 4.5.1 also demonstrates that the combined effector of the elderly travels the remaining swing phase in a more direct line, i.e. with limited medio-lateral displacement.

Section 4.6 will detail between-group differences in the three-dimensional features of the effector vectors.

4.5.1 Kinematic description of the three effector vectors.

This section describes the three vectors which represent the kinematic chain effector systems in separate three-dimensional components. Figure 4.5.1.1 describes a typical young participant's vertical displacement (A) and velocity (B) profiles for the three effector vectors. General observations of the curves show that the swing limb vector has the larger range of motion and velocity.

From the vertical displacement graphs in Figure 4.5.1.1A, two features of interest are: 1) the stance effector continues to lengthen upwards after the MX1 event; and 2) the swing effector begins to shorten in the downwards direction (or increasing upwards displacement) prior to the MX1 event. To obtain toe-clearance the vertical length of the stance vector needs to be greater than the vertical length of the swing vector. The combined effect of increasing the stance vector and decreasing the swing vector results in a positive value for the combined effector vector. This difference represents the clearance height between the ground and the swing-toe.

Figure 4.5.1.1B shows a peak maximum upward vertical velocity of the stance vector occurring between MX1 and MTC. Likewise, at a similar moment in time, there is a maximum downward vertical velocity for the vertical component of the swing effector vector.

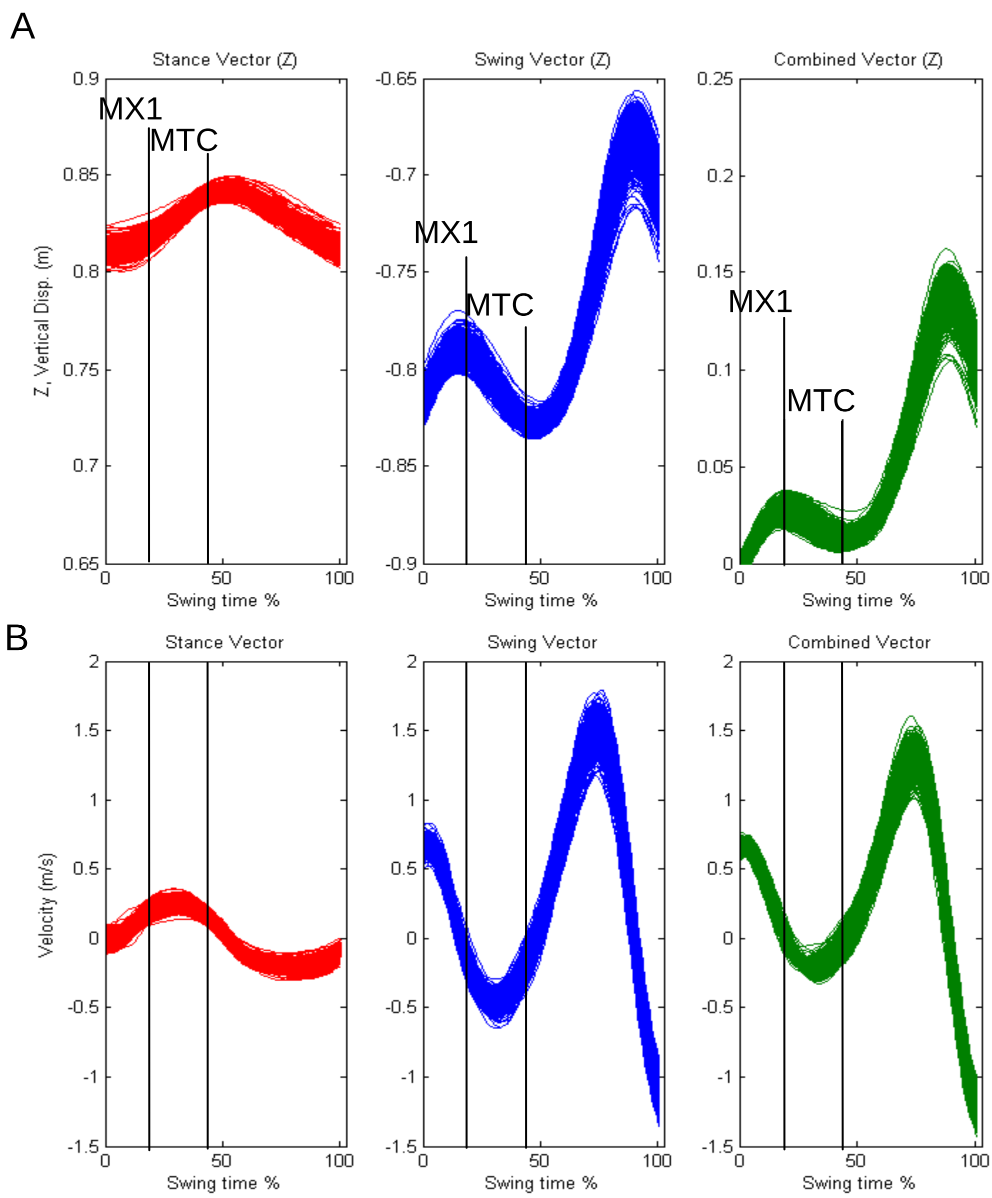


Figure 4.5.1.1

A sample plot of a young subject's vertical displacement (top row) and velocity (i.e. the first derivative of displacement; bottom row) of the three effector vectors across the swing phase of their preferred (dominant) stance-swing kinematic chain. The vertical axis of the displacement graphs are zero-shifted but the scale is the same between plots. The mean percentage time occurrence of the reference events, MTC and MX1, are highlighted. Note: the vertical displacement of the combined vector represents the ground clearance of the swing toe.

4.6 Mean length components of the effector vectors

This section investigates the null hypothesis of 'no significant between-group difference' for the distribution-mean of the effector trajectories at the discrete-time states of the two swing-phase regions, MX1 and MTC. Following on from the trajectories displayed in three-dimensional space in section 4.5, this section will explore the mean statistics of the effector trajectories in separate three-dimensional components: vertical, anterior-posterior, and medio-lateral. All components of the three-dimensional effector vector data will be presented as normalised to subject limb-length. One exception will be the combined effector in the vertical direction, which will be described in metric units and normalised limb-lengths. The data was additionally presented in metric units because this represents the 'environment space' in which the toe-clearance task is being performed. In contrast, normalised data represents 'internal/body space'.

4.6.1 Combined effector: means of the component lengths

Figure 4.6.1.1 and 4.6.1.2 display the combined effector data in the vertical dimension in normalised limb-lengths and metric units respectively. In either normalised or absolute units, there were no significant differences ($p>.1$) between the groups for the states neighbouring MX1. There are qualitative differences between the normalised and absolute curves. In the dominant and non-dominant limbs from Figure 4.6.1.1, the elderly group approach the $p<.1$ confidence interval for the initial states of the MX1 region. In comparison however, the normalised data in Figure 4.6.1.2 shows that the mean signal of the elderly is more closely aligned with the young group mean. The only states that demonstrate a significant difference ($p<.1$) is at the final states of the MTC region (i.e. MTC+15%).

In normalised limb-lengths (LL), the group mean MTC (±SD) in for the dominant and non-dominant limb was respectively: 0.0129 (±0.006) and 0.0129 (±0.005) for the young; 0.0131 (±0.005) and 0.0149 (±0.009) for the elderly. The group mean MTC (±SD) in absolute units (centimetres) for the dominant and non-dominant limb was respectively: 1.01 (±0.44) cm and 1.02 (±0.35) cm in the young group; 1.01 (±0.40) cm and 1.14 (±0.67) cm in the elderly group. There was large intra-group variance (i.e. ±0.67) in the reported MTC mean values in the non-dominant limb of the elderly. Based upon the within-group average for the dominant and non-dominant limbs, there were no between-limb significant differences found in any states of the swing-phase regions, in either normalised or absolute units.

Figure 4.6.1.3 shows the average of the combined effector trajectories in the anterior-posterior dimension. The elderly display a significantly reduced anterior position of the combined effector during the MTC states in both limbs, but particularly the non-dominant limb. This indicates that when compared to the young group, the elderly swing-toe position is significantly ($p<.1$) less forward relative to the stance foot position at MTC.

Figure 4.6.1.4 displays the medio-lateral component of the combined effector. There are no significant differences between the young and elderly at any states of the MX1 and MTC regions, in either the dominant and non-dominant limbs. The shape of the curve demonstrates that both groups display lateral motion of the combined effector throughout the swing phase.

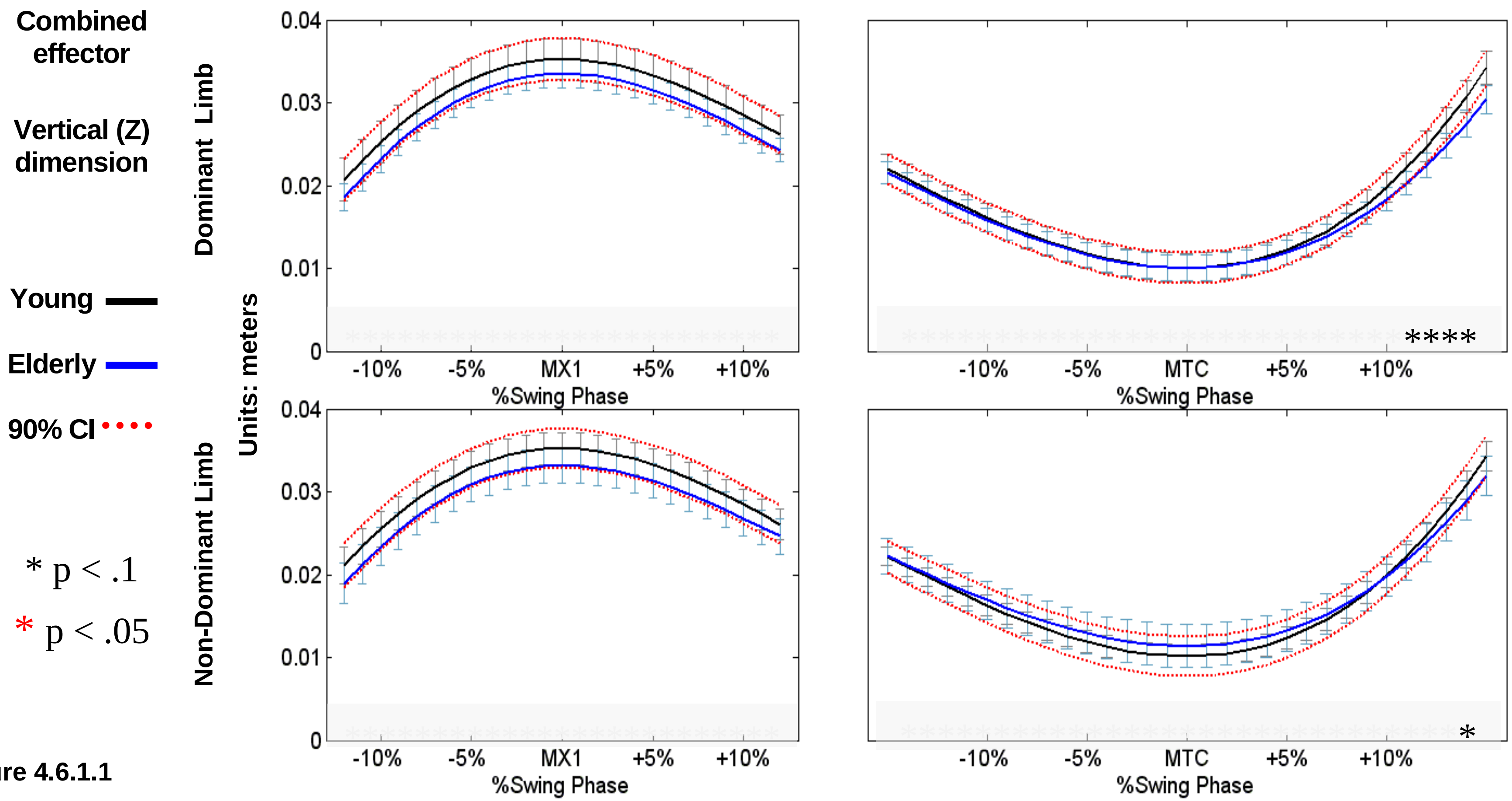


Figure 4.6.1.1

The vertical (Z) component of the combined effector vector in absolute units from the ground surface (ordinate units are meters, m). The abscissa represents units of %swing phase cycle from reference events MX1 and MTC. Error bars represent the 95% confidence interval of the young ($\bar{X}_Y$) and elderly ($\bar{X}_E$) group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exits the bounded red lines (±90% confidence interval). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

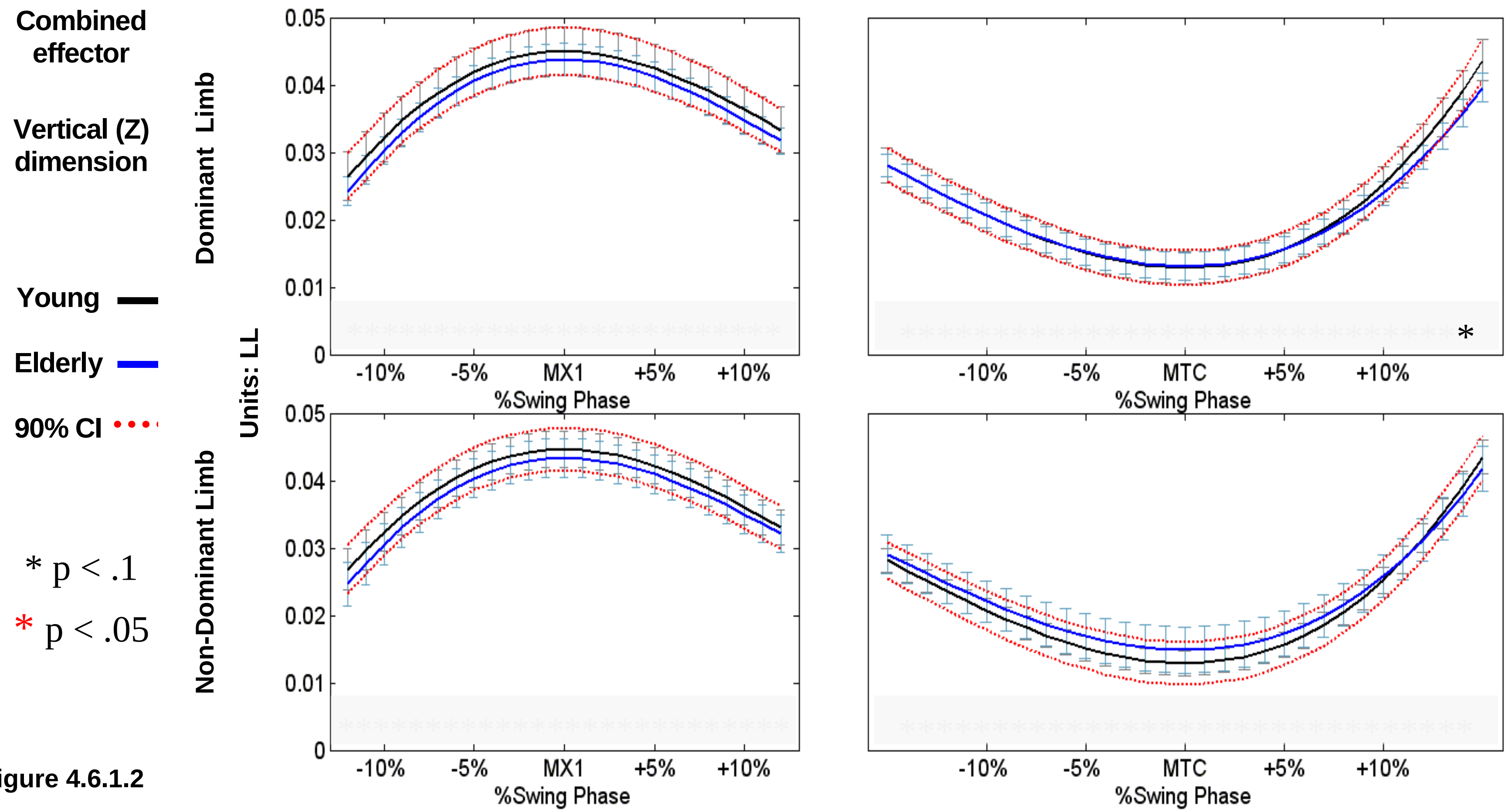


Figure 4.6.1.2

The vertical (Z) component of the combined effector vector in limb-length normalised units (LL). The abscissa represents units of %swing phase cycle from reference events MX1 and MTC. Error bars represent the 95% confidence interval of the young ($\bar{X}_Y$) and elderly ($\bar{X}_E$) group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

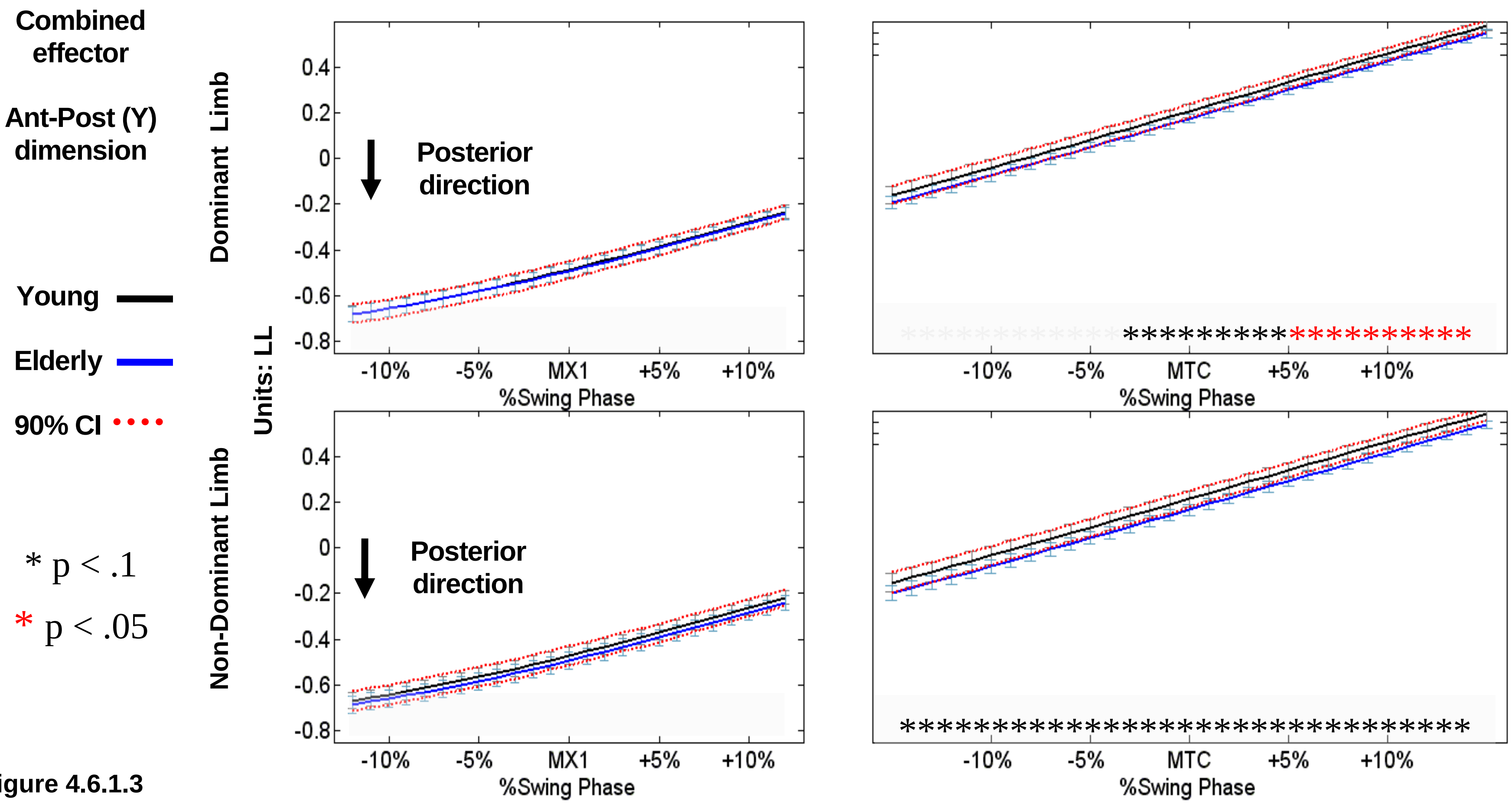


Figure 4.6.1.3

The anterior-posterior (Y) component of the combined effector vector (normalised limb-lengths, LL). The abscissa represents units of %swing phase cycle from reference events MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

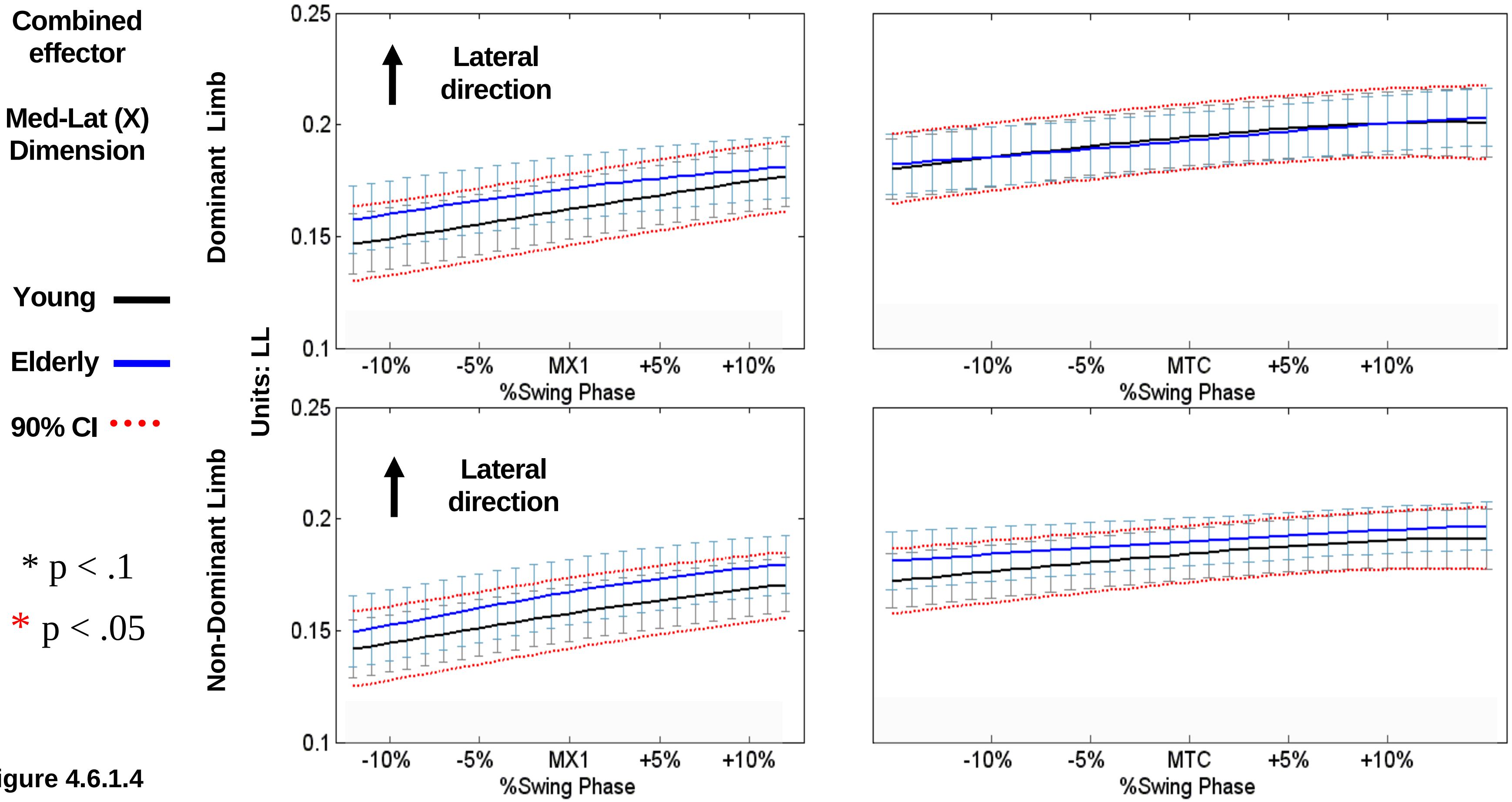


Figure 4.6.1.4

The medio-lateral (X) component of the combined effector vector (normalised to limb length, LL). The abscissa represents units of %swing phase cycle from reference events MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.6.2 Stance effector: means of the component lengths

The average configuration of the stance effector in the three-dimensions is displayed in Figures 4.6.2.1-3. Figure 4.6.2.1 shows the group average for the vertical dimension of the stance effector. The group mean curves from the dominant and non-dominant limb comparisons show that the elderly have a higher vertical displacement compared to the young mean curve throughout the swing phase. However, the confidence intervals demonstrate that between-group differences ($p<.1$) are evident when comparisons are made only within the non-dominant limb.

Figure 4.6.2.2 demonstrates that the posterior orientation of the elderly stance effector is significantly ($p<.05$) greater than the young group for the non-dominant limb during the MTC region. For the MTC region in the dominant limb, the elderly mean does not quite reach the 90% confidence intervals (i.e. $p>.1$).

Figure 4.6.2.3 demonstrates that the stance effector of the elderly has significantly larger medial displacement relative to the stance-foot during states neighbouring the MX1 region; i.e. in the medio-lateral dimension of the transverse plane, the swing-hip position of the young group is aligned significantly closer to the stance-foot position when compared to the elderly. For the non-dominant limb, the elderly continue to display significantly ($p<.1$) larger medial displacement of the stance effector through the MTC region. The general shape of the curve demonstrates that the medial displacement of the stance effector reaches a minimum prior to MTC.

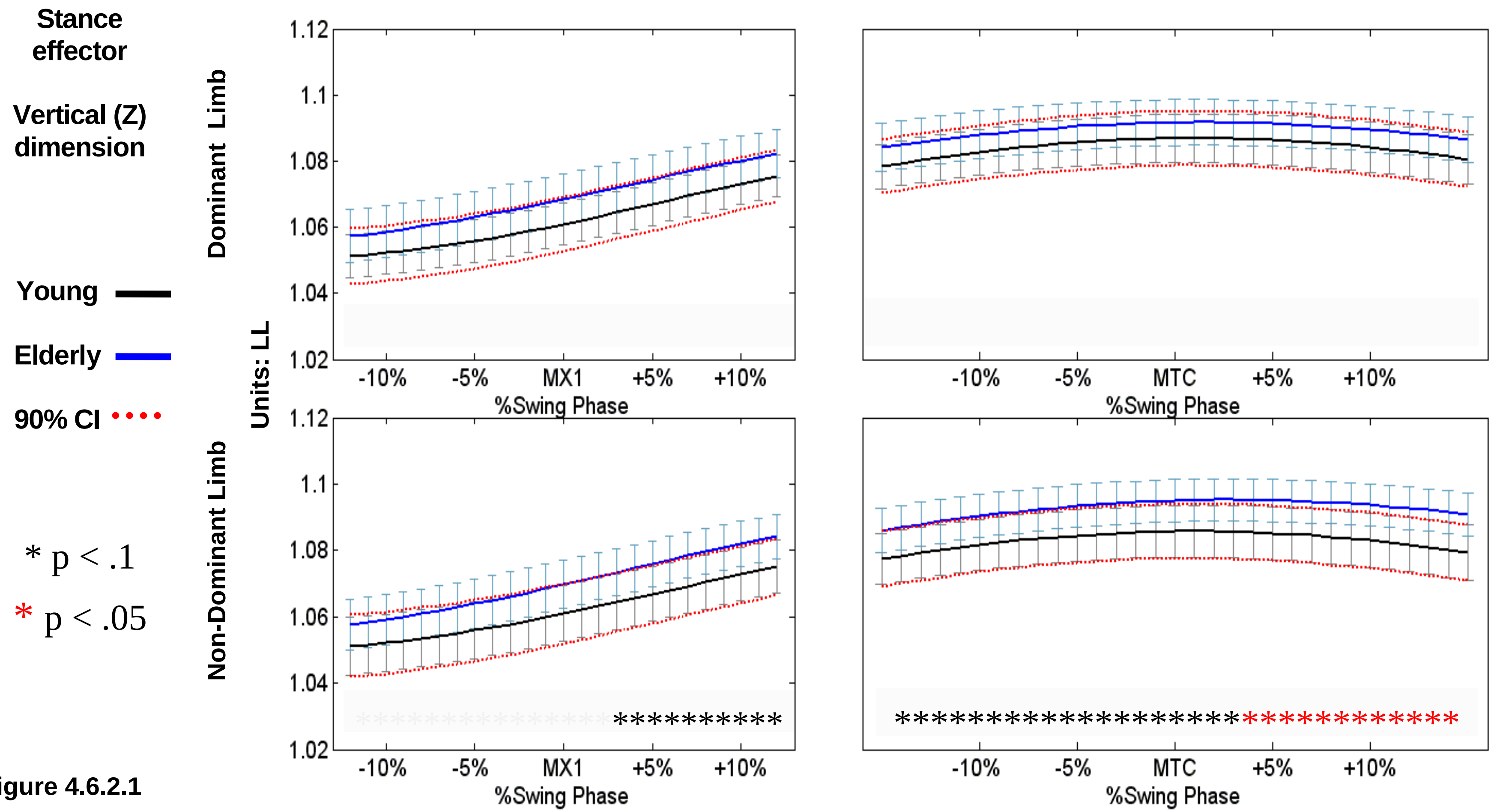


Figure 4.6.2.1

The vertical (Z) component of the stance effector vector (normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

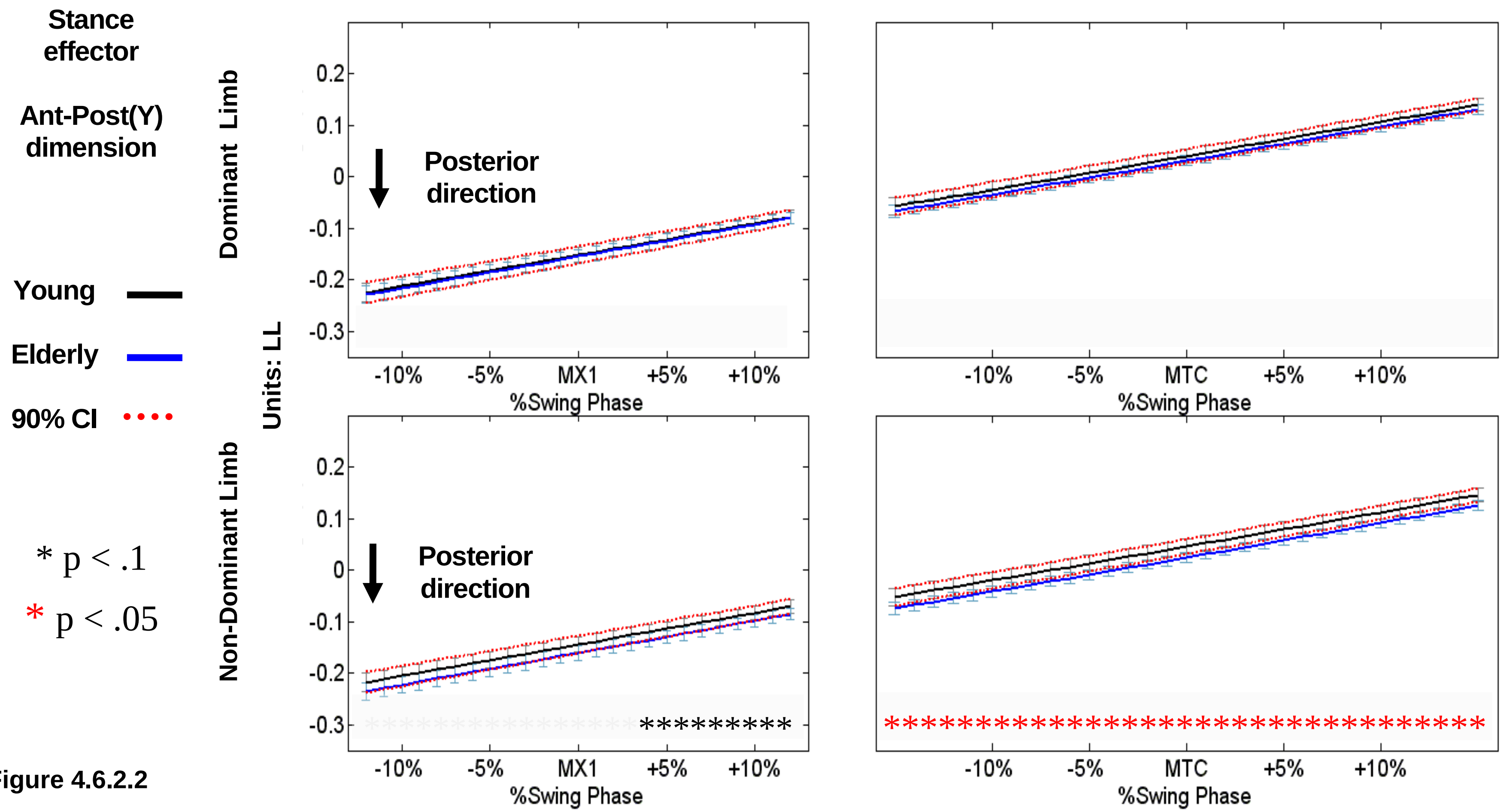


Figure 4.6.2.2

The anterior-posterior (Y) component of the stance effector vector (normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

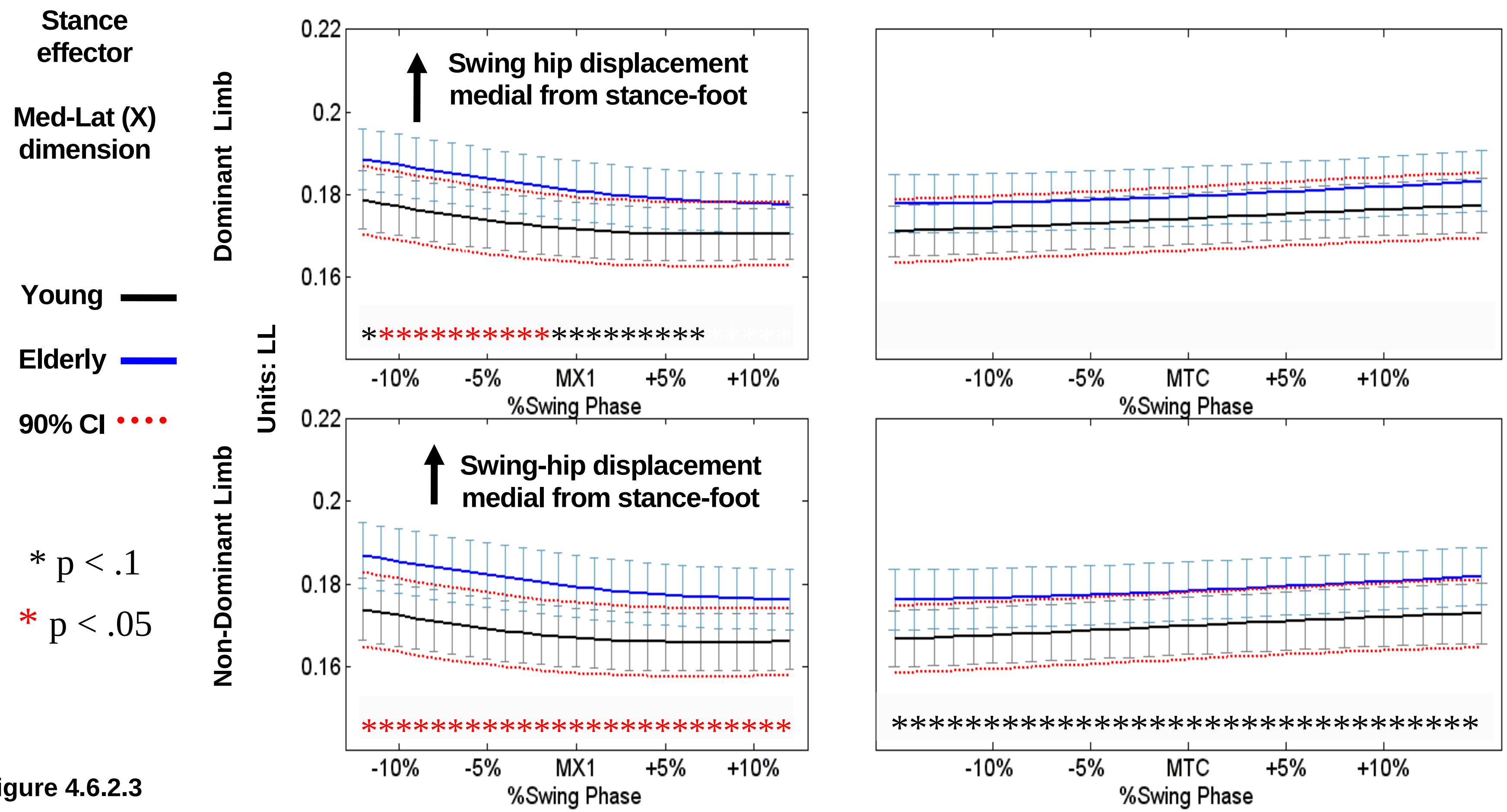


Figure 4.6.2.3

The medio-lateral (X) component of the stance effector vector (normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.6.3 Swing effector: mean of the component lengths

Figures 4.6.3.1-3 compares the young and elderly swing effector trajectories in the vertical, anterior-posterior, and medio-lateral dimensions. The swing effector is a vector directed from the swing hip position to the swing-toe position. The vertical component of the swing effector vector is therefore negatively directed. Increased negative displacement suggests an increase in limb-length of the swing effector.

Figure 4.6.3.1 displays the vertical component of the swing effector. The vertical axis scale represents negative limb-lengths, such that low values (i.e. large negative values) indicate a general lengthening of the swing-limb. For between-group comparisons in the MX1 region, the non-dominant limb shows that the elderly have a significantly ($p<.1$) longer swing effector in the vertical dimension. For the dominant limb during the MX1 region, the elderly mean does not quite exceed the 90% confidence interval ($p>.1$). Likewise, this proximity to the 90% confidence interval for the elderly also occurs during the MTC region of the non-dominant limb. If considering the argument that the dominant and non-dominant limbs can be independent cases, such that the young and elderly groups are now considered to have twice the sample size, then there is a statistically significant difference ($p<.1$) in all states of the swing cycle except for MTC neighbouring states between MTC-6% and MTC+3%. Under this limb-independence argument, this would indicate an increased vertical displacement of the swing effector in the elderly for much of the swing phase.

Figure 4.6.3.2 compares the young and elderly for the mean anterior-posterior component of the swing effector. The outcome is consistent with the anterior-posterior component of the stance effector found in section 4.6.2 (Figure 4.6.2.2). Figure 4.6.3.2 also indicates that the elderly have significantly more posterior orientation of the swing effector during the MTC region. The general shape of the curve indicates that the swing-toe position is in front of the swing-hip position prior to reaching the state MTC-5%.

Figure 4.6.3.3 shows the medio-lateral path of the swing effector begins medial to the swing hip position and moves laterally, showing a similar shape as the combined effector. Similarly to the medio-lateral component of the combined effector, there are no significant differences between the groups for the medio-lateral path of the swing effector.

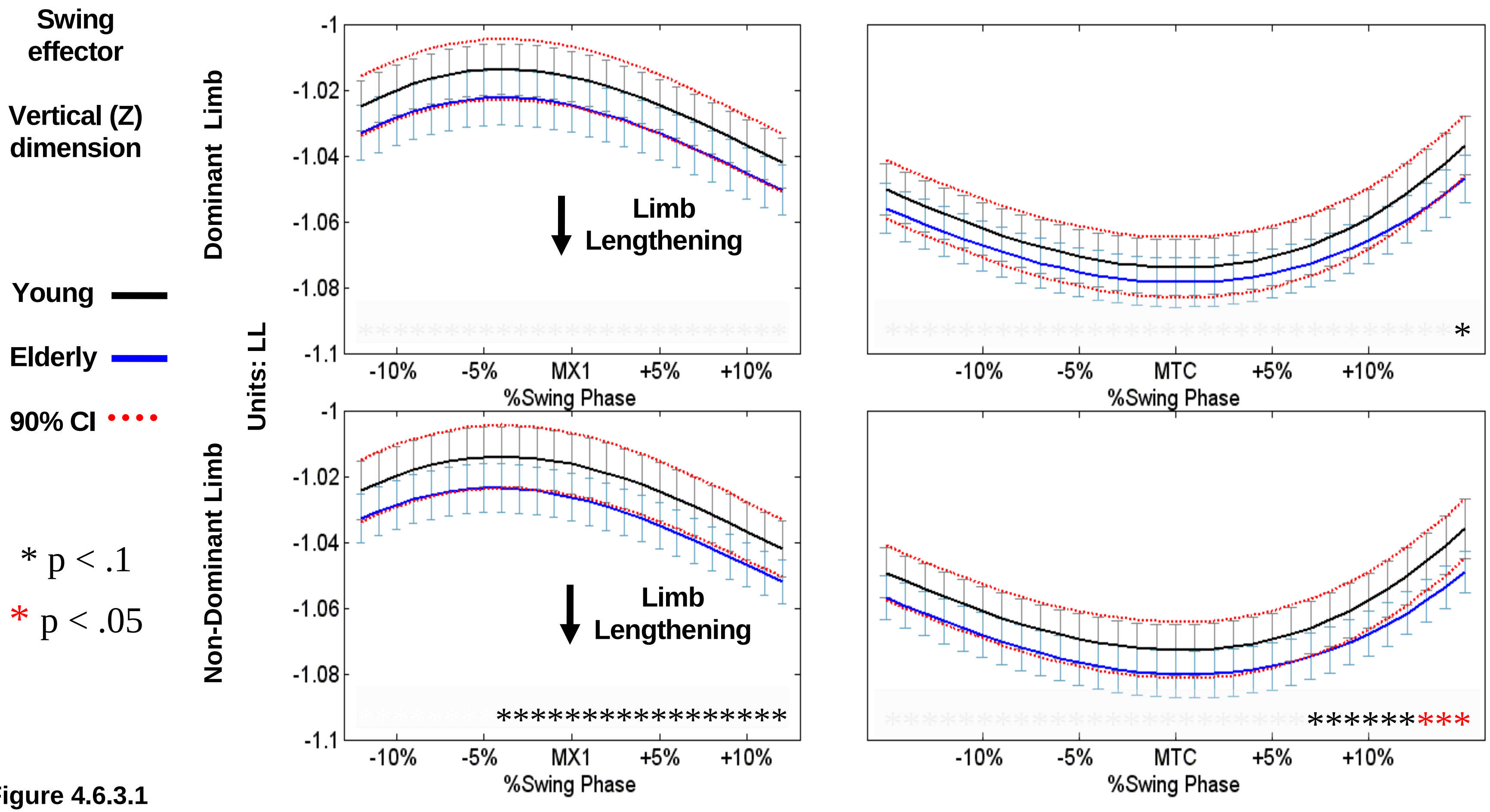


Figure 4.6.3.1

The vertical (Z) component of the swing effector vector (normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

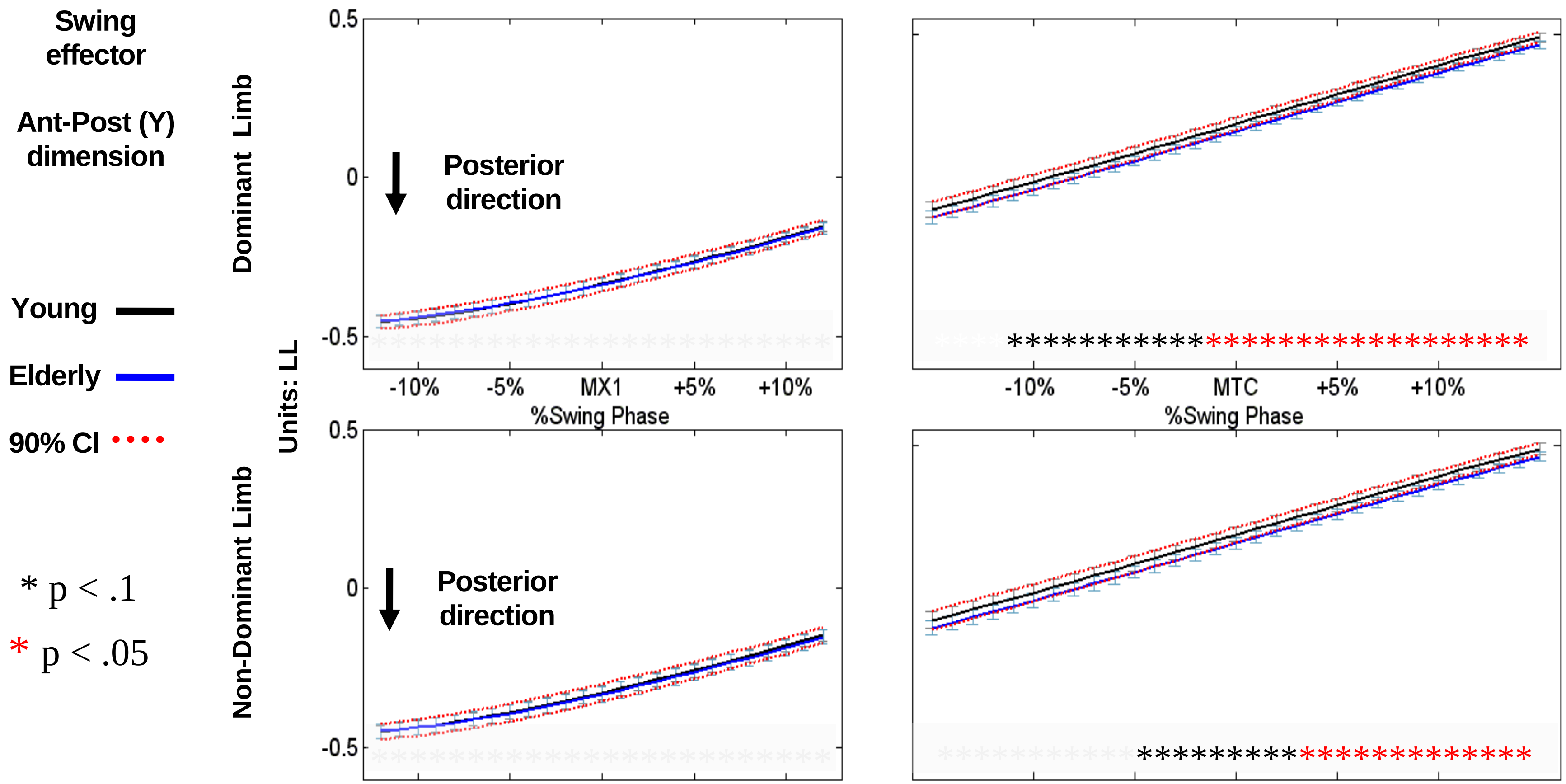


Figure 4.6.3.2

The anterior-posterior (Y) component of the swing effector vector (units normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

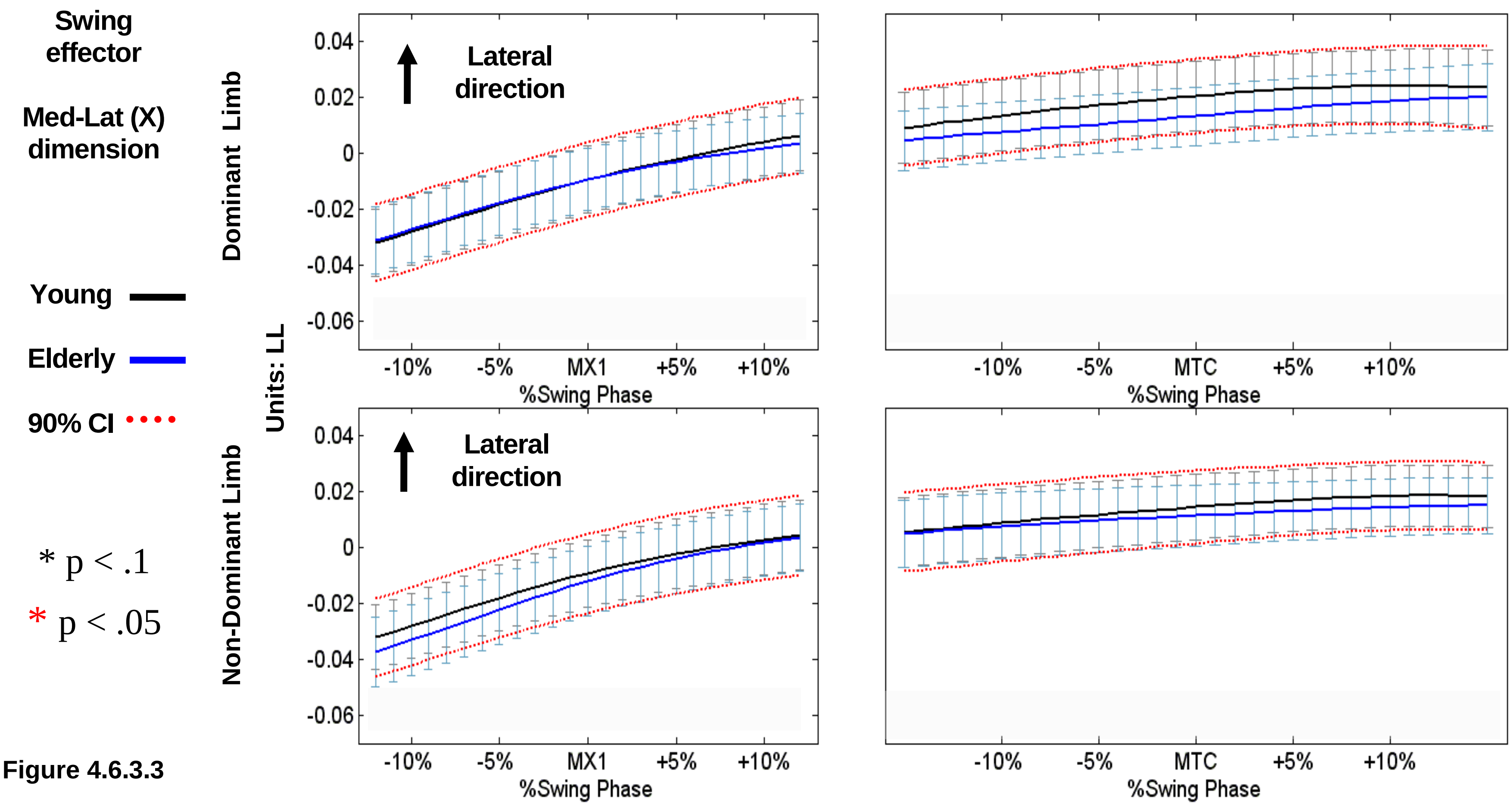


Figure 4.6.3.3

The medio-lateral (X) component of the swing effector vector (normalised to limb length, LL). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.6.4 Interaction effects: changes in effector mean between MX1 and MTC

The change in the vector length between states MX1 and MTC is summarised in Figure 4.6.4.1A-C. The graphs represent the data after pooling the dominant and non-dominant limbs into one group; group means and error bars represent the mean and the 95% confidence interval of the group mean. Visually, the lines connecting the group means show that the elderly have a higher stance effector trajectory and a lower swing effector trajectory compared to the young group. So, rather than separately compare mean trajectories of the effectors between the groups at states of MX1 or MTC, a general assessment of the trajectory between MX1 and MTC can be formed when investigating main effects from a two-way (state × group) mixed design ANOVA at the within-level of each effector. The dependent variable was the mean trajectory in the vertical dimension (normalised to limb length). The following analyses tested the null hypothesis that 'there will be no significant difference for a group main effect' (i.e. when collapsing the effect of state).

For the combined effector there is a significant interaction effect demonstrating that the groups have different vertical trajectories between MX1 and MTC ($F(1, 110) = 12.8$, $p < .05$, $\eta^2 = .06$). The main effect for group is not significant and therefore the null hypothesis that there is no significant between-group difference for height of the combined effector trajectory through the MX1-MTC region of the swing phase. However, the significant interaction suggests that the elderly have a significantly different shape in the combined effector trajectory during this region. In summary, for the elderly, the combined effector trajectory has a significantly 'flatter' shape (in the vertical range) from MX1 to MTC.

For the vertical dimension of the stance effector, there was no interaction effect for group and state [Figure 5.6.1.4.1 (B)]. There was a significant between group difference, indicating that the elderly have a higher 'limb-length scaled' stance effector trajectory ($F(1,110) = 5.01$, $p < .05$, $\eta^2 = .04$).

For the vertical dimension of the stance effector, there was no interaction effect for group and state [Figure 5.6.1.4.1 (C)]. There is a significant main effect for group

(F(1,110) = 4.53, $p < .05$, $\eta^2 = .04$) indicating that the elderly generally have a lower 'limb-length scaled' swing effector trajectory (NB: 'lower' refers to a closer proximity to the ground surface from the swing-hip position).

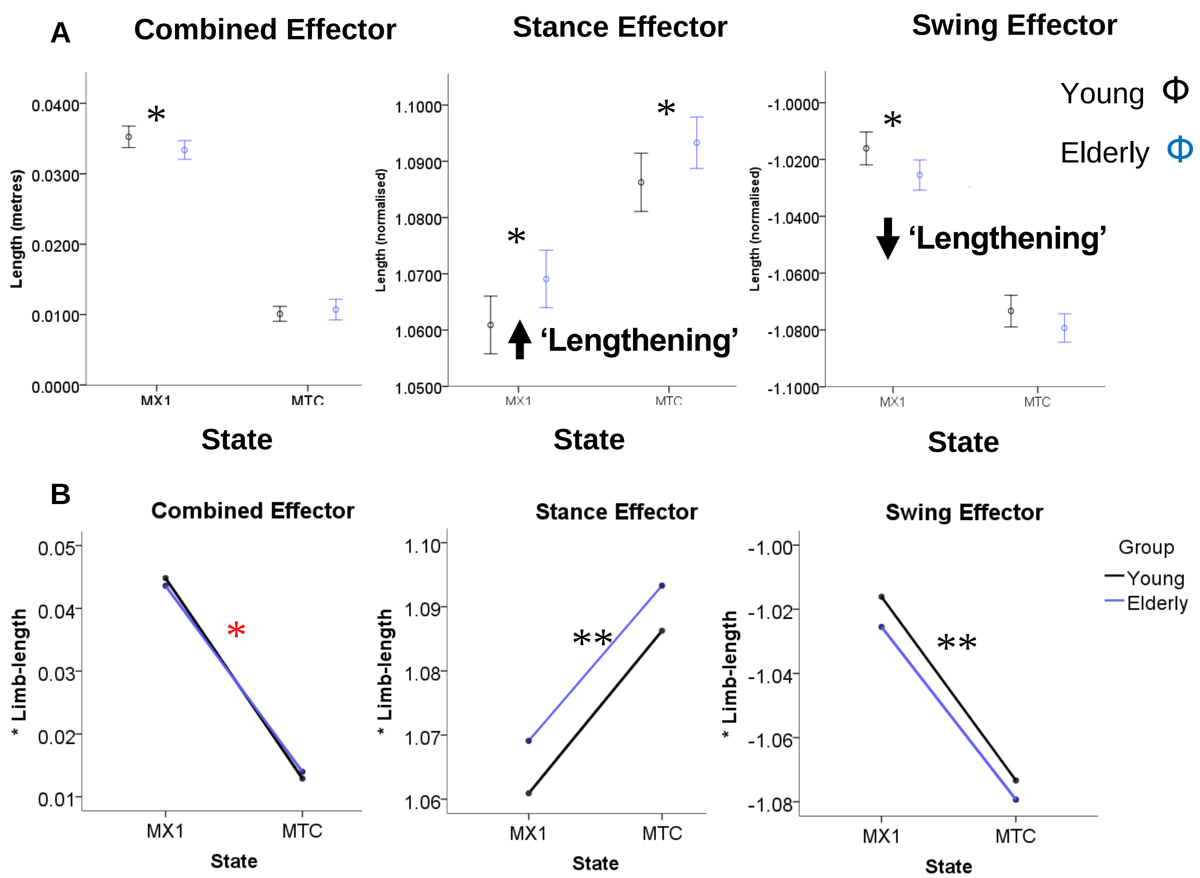


Figure 4.6.4.**1**

Summary of the mean effector trajectories in the vertical dimension. **A)** Group means ± 95% confidence intervals (error-bars) at MX1 and MTC. Note the different ordinate scales between plots: the combined effector is represented here in absolute units (metres); the stance and swing effector units are scaled to normalised limb-lengths (/LL). **B)** Graphs showing within effector interactions between state (MX1 and MTC) and group (young and elderly).

* indicates significant between group difference (p<.05).

* indicates significant interaction effect (p<.05).

** indicates significant main effect for age (p<.05).

4.7 Velocity of the components of the effector vectors

To determine the motion of the effector vectors, the first derivative of the effector displacement components were computed to provide a measure of effector velocity. As per previous sections of analysis, between-group comparisons will be made at discrete-time states of the MX1 and MTC regions.

The two reference states, by their nature, involve turning points in vertical dimension and therefore the velocity of the combined effector will converge to zero at these reference states. The stance and swing effector vectors will typically display a non-zero velocity at these states. The units for velocity are scaled to normalised limb-lengths per second (i.e. LL/s).

4.7.1 Combined effector: velocity

The velocity of the combined effector describes the motion of the swing-toe with respect to a fixed position at the stance-foot. Figures 4.7.1.1-3 displays the combined effector velocity in the vertical, anterior-posterior, and medio-lateral dimensions respectively.

Figure 4.7.1.1 indicates that the elderly display a significantly lower vertical velocity at states following MTC, for both the dominant and non-dominant limbs. General features of the curve indicate that the maximum downward velocity of the toe occurs near states MX1+12% and MTC-15%.

Figure 4.7.1.2 demonstrates that there are no significant differences in the anterior-posterior component of the combined effector. The graph indicates that the forward toe velocity nears a maximum at MTC.

Figure 4.7.1.3 compares the medio-lateral velocity of the combined effector. The elderly group have significantly higher lateral velocity in the non-dominant limb in the states preceding MX1. For the dominant limb, the elderly demonstrate a minimal lateral velocity between states MX1+5% and MTC-10%, which is significantly lower than the young group. Both groups maintain a lateral velocity of the combined effector through all states up until MTC+10%.

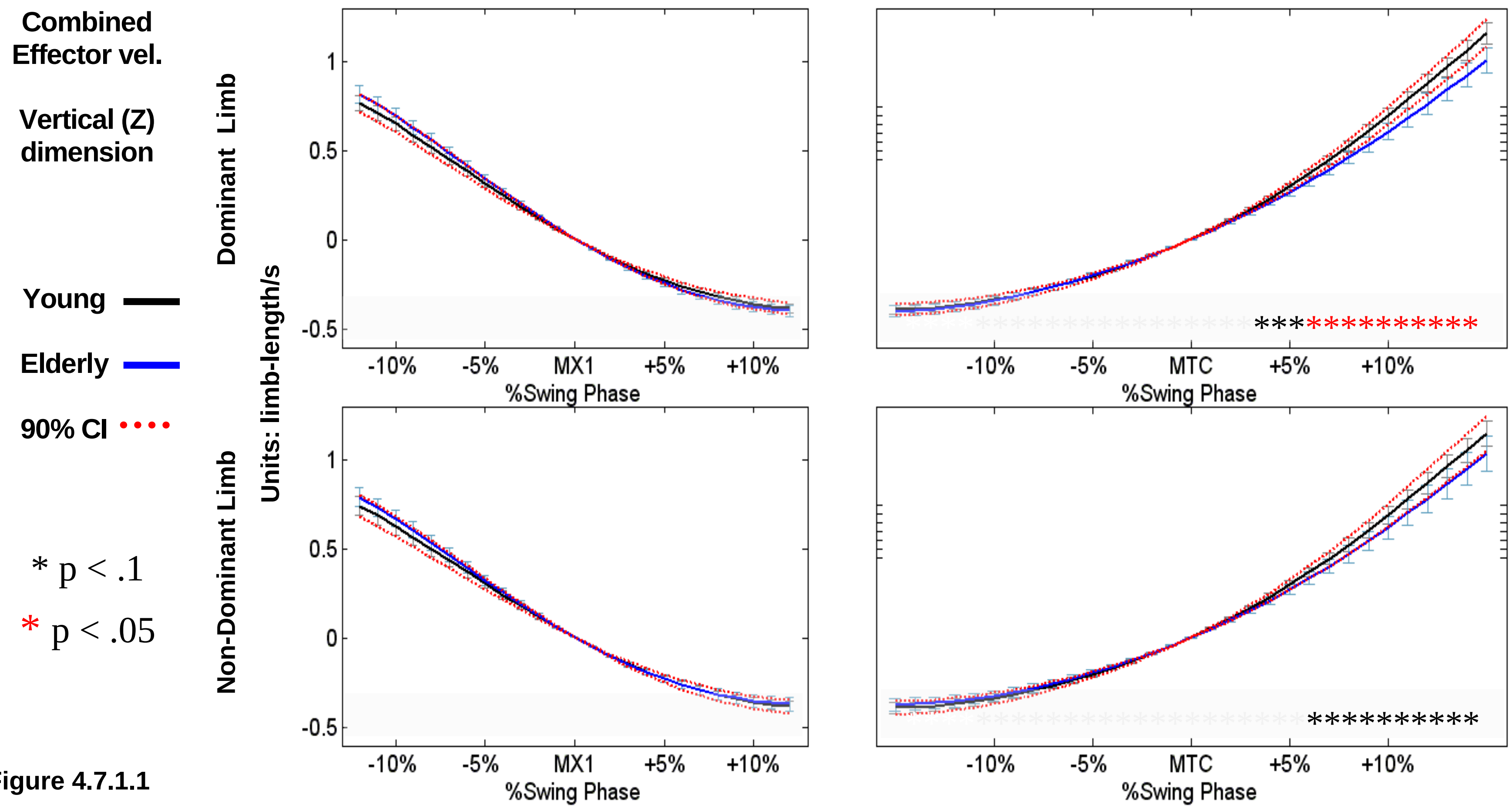


Figure 4.7.1.1

The vertical (Z) velocity component of the combined effector vector (units: limb-lengths / s). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

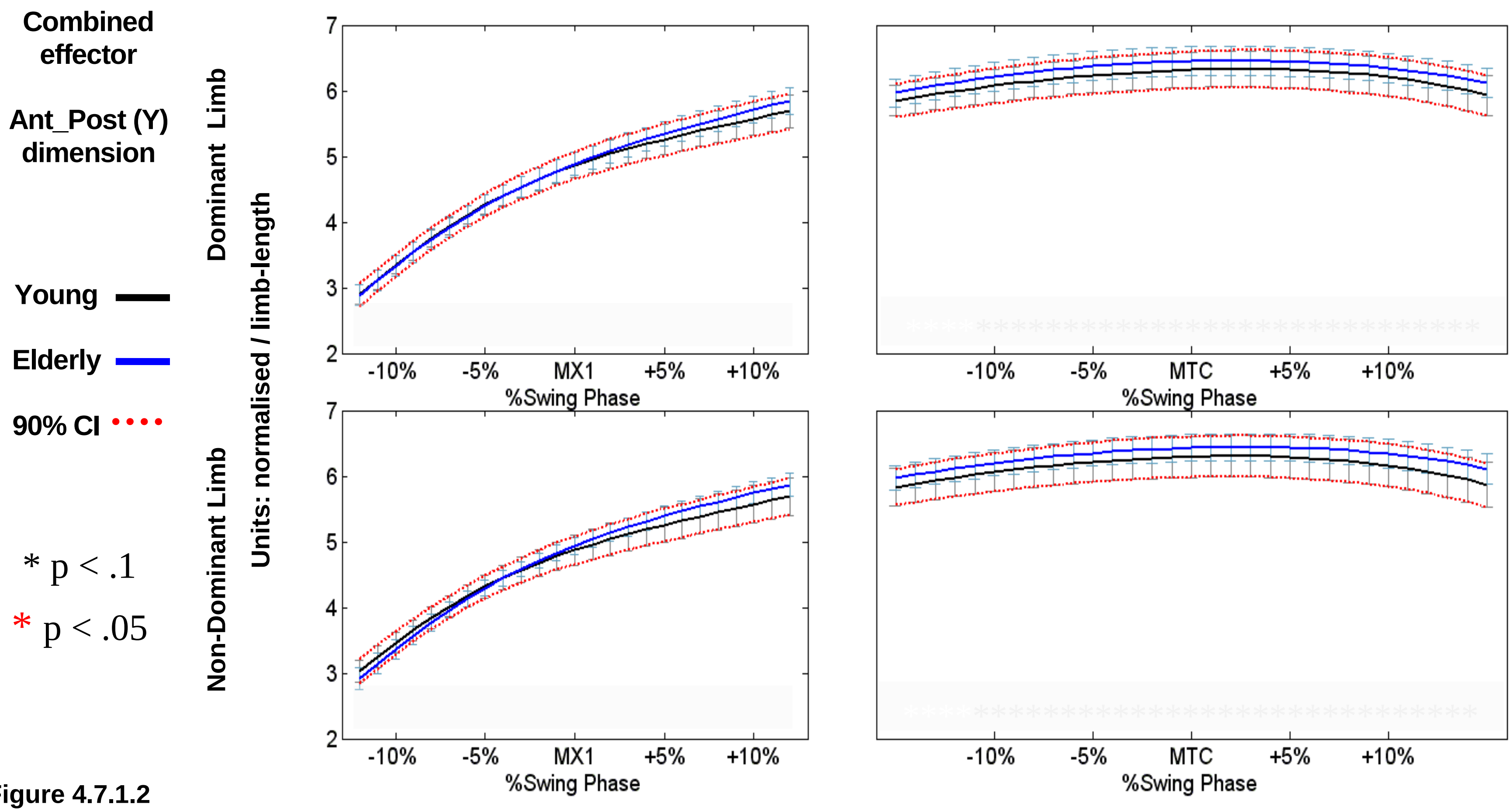


Figure 4.7.1.2

The anterior-posterior (Y) velocity component of the combined effector vector (units: limb-lengths / s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

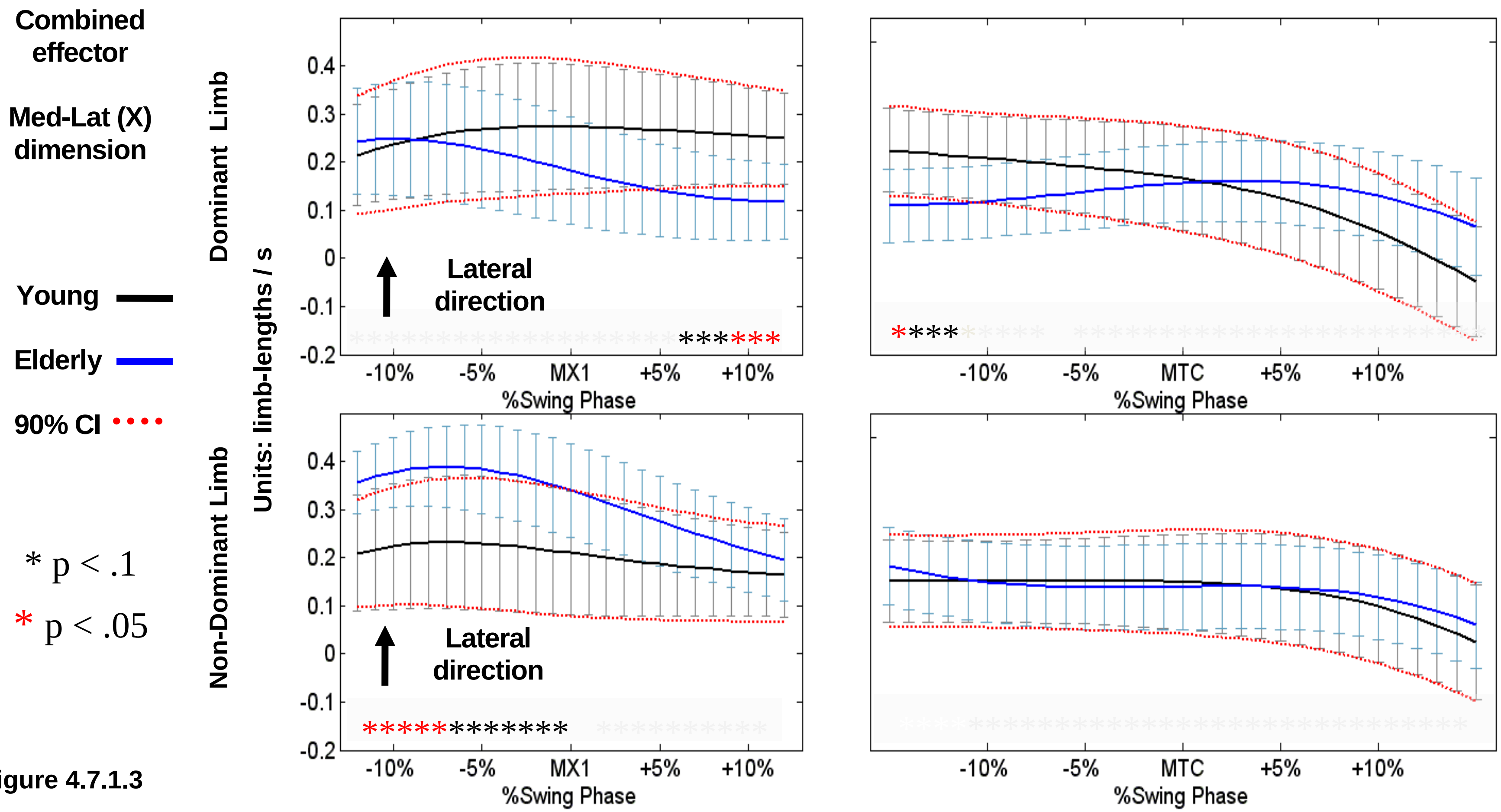


Figure 4.7.1.3

The medio-lateral (X) velocity component of the combined effector vector (units: limb-lengths / s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.7.2 Stance effector: velocity

Figures 4.7.2.1-3 displays the stance effector velocity in the three component dimensions.

Figure 4.7.2.1 shows that the velocity of the stance effector is significantly ($p < .05$) higher for the elderly group during the initial states of the MX1 region. For the non-dominant limb, the upward velocity of the stance effector is significantly higher from MX1-12% up until state MX1-2%. Both the dominant and non-dominant limbs also show that the vertical velocity of the stance effector is significantly higher in the final states of the MTC region, again, this is more significant in the non-dominant limb.

Figure 4.7.2.2 displays the anterior velocity of the stance effector. There are no between-group differences. A feature of the velocity curve profile is the minimum anterior velocity that occurs near MX1. Both groups demonstrate that the swing hip position slows down in forwards velocity until the vicinity of MX1; henceforth, the swing-hip increases in forwards velocity. This demonstrates a potential task-relevant coupling between the stance and swing effector.

Figure 4.7.2.2 displays the medio-lateral velocity of the stance effector. The general feature of the curve demonstrates that both the groups have a medial velocity, or a swing hip position that is moving closer to the stance foot position.

There is a significant difference between the groups for states between MX1+3% to MTCm11% in the dominant limb; and for the non-dominant limb, from MX1-10% to MTCm15%, and MTC+6% to MTC+15%. In contrast to the elderly, the young group demonstrate that their stance effector trajectory is significantly closer to near-zero velocity during the final states of the MX1 region.

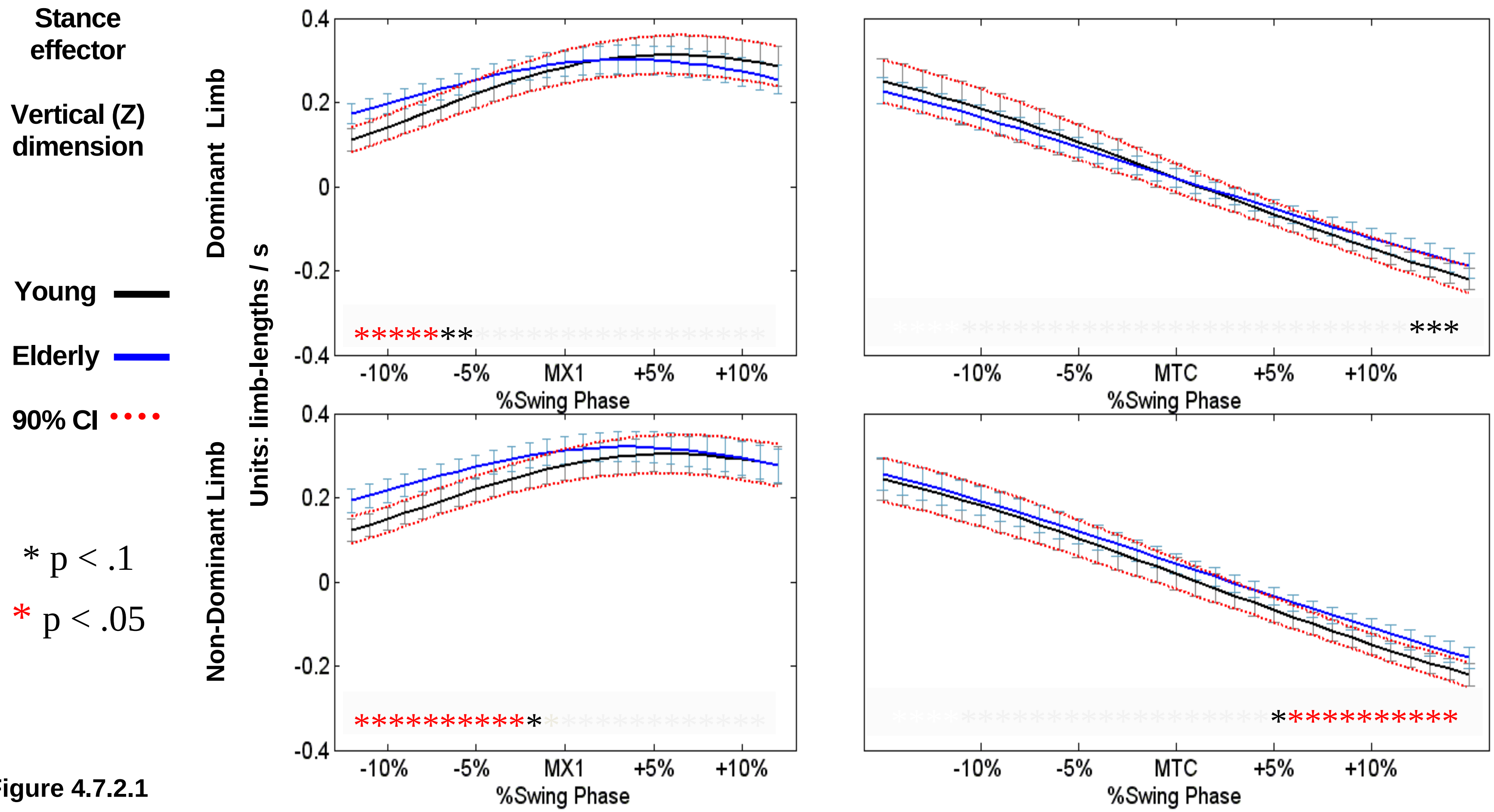


Figure 4.7.2.1

The vertical (Z) velocity component of the stance effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

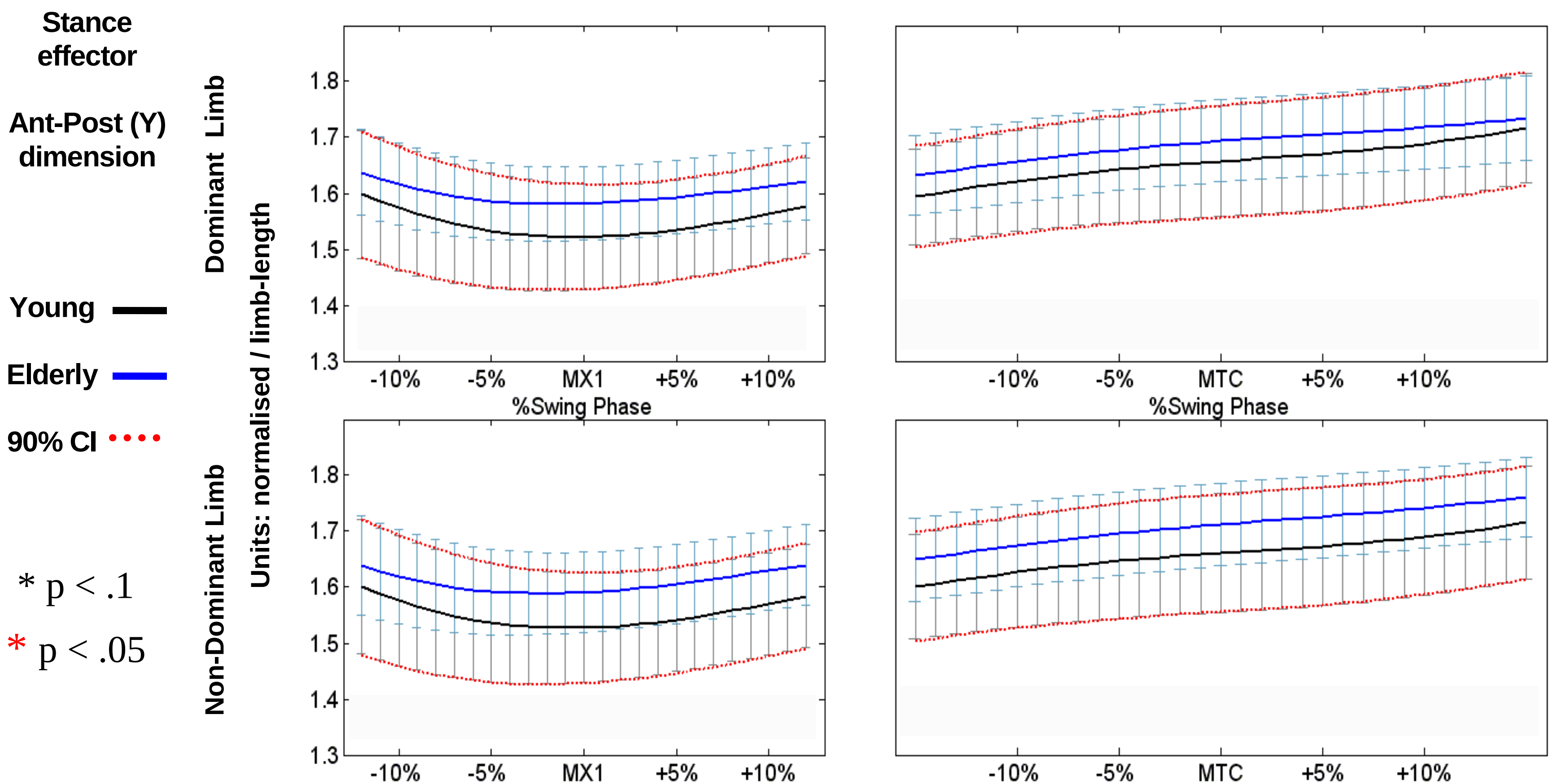


Figure 4.7.2.2

The anterior-posterior (Y) velocity component of the stance effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the young ($\bar{X}_Y$) and elderly ($\bar{X}_E$) group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

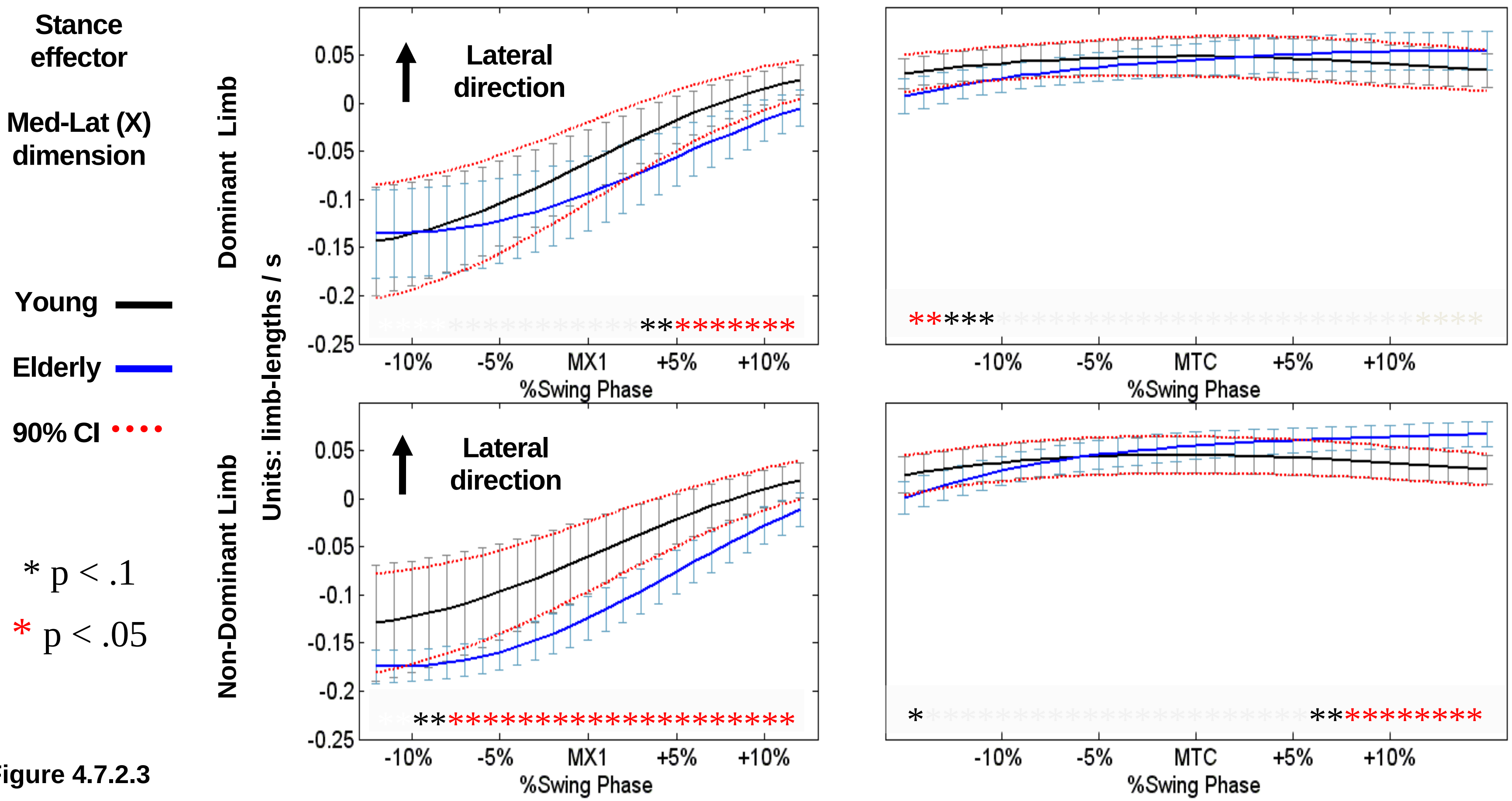


Figure 4.7.2.3

The medio-lateral (X) velocity component of the stance effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.7.3 Swing effector: velocity

Figures 4.7.3.1-3 show the swing effector velocity. Figure 4.7.3.1 shows the vertical motion of the swing effector. The elderly have a significantly lower upward velocity compared to the young group during the terminal states of the MTC region (i.e. from MTC+5% onwards). The general shape of the swing-effector (Figure 4.7.3.2) velocity curve indicates a maximum forwards velocity near MTC for both groups. There are no significant differences in the anterior velocity of the swing effector (Figure 4.7.3.2). When collapsing both limbs into one group, and making a between-group comparison at MX1+10% and at MTC, there was also no significant difference.

Figure 4.7.3.3 indicates a significantly larger lateral velocity of the non-dominant limb during the MX1 region. When contrasting this result with the finding of a significantly higher medial velocity in the elderly non-dominant stance effector (also occurring during similar states), it is suggesting a possible between-effector cooperation.

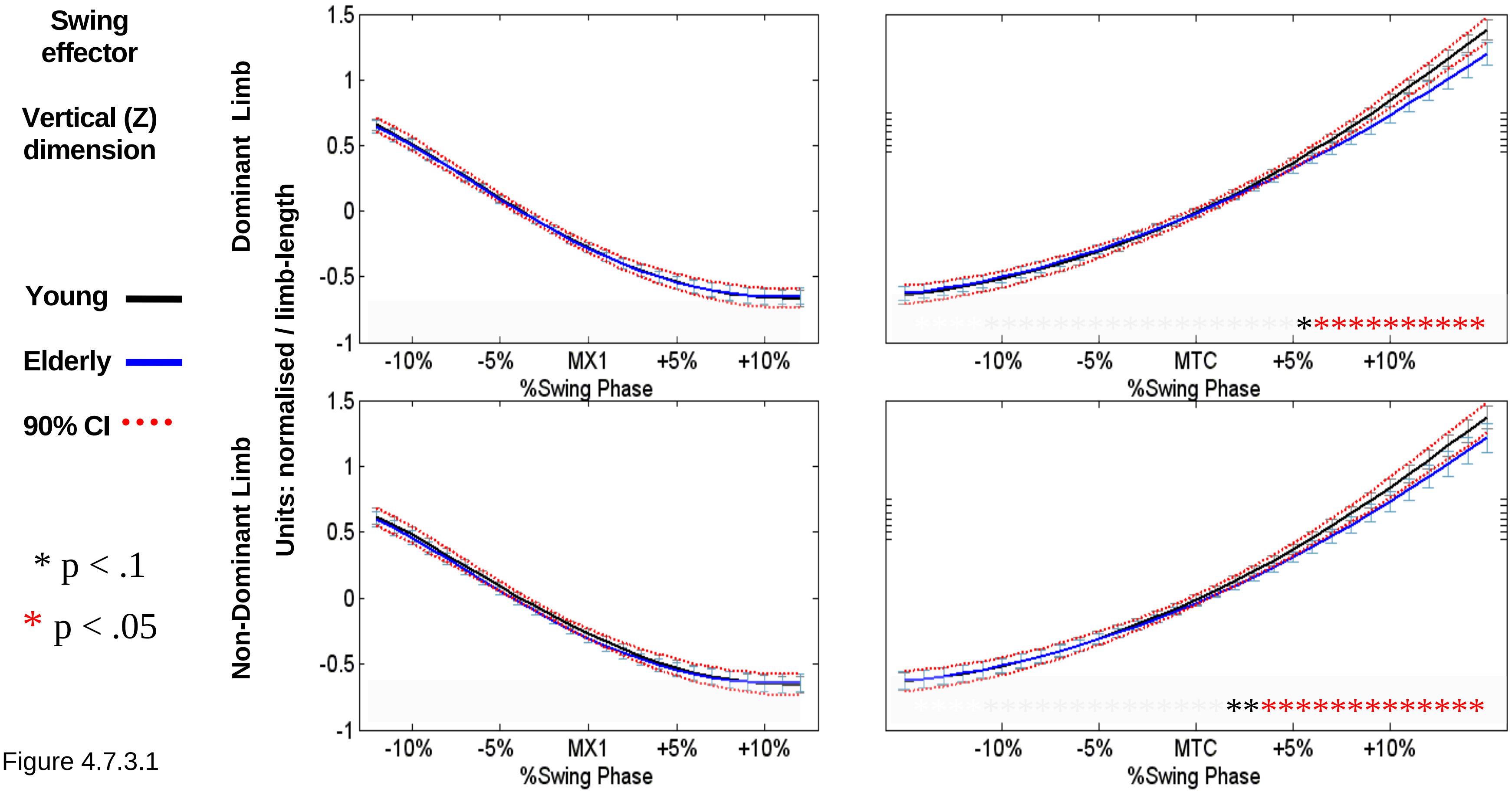


Figure 4.7.3.1

The vertical (Z) velocity component of the swing effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

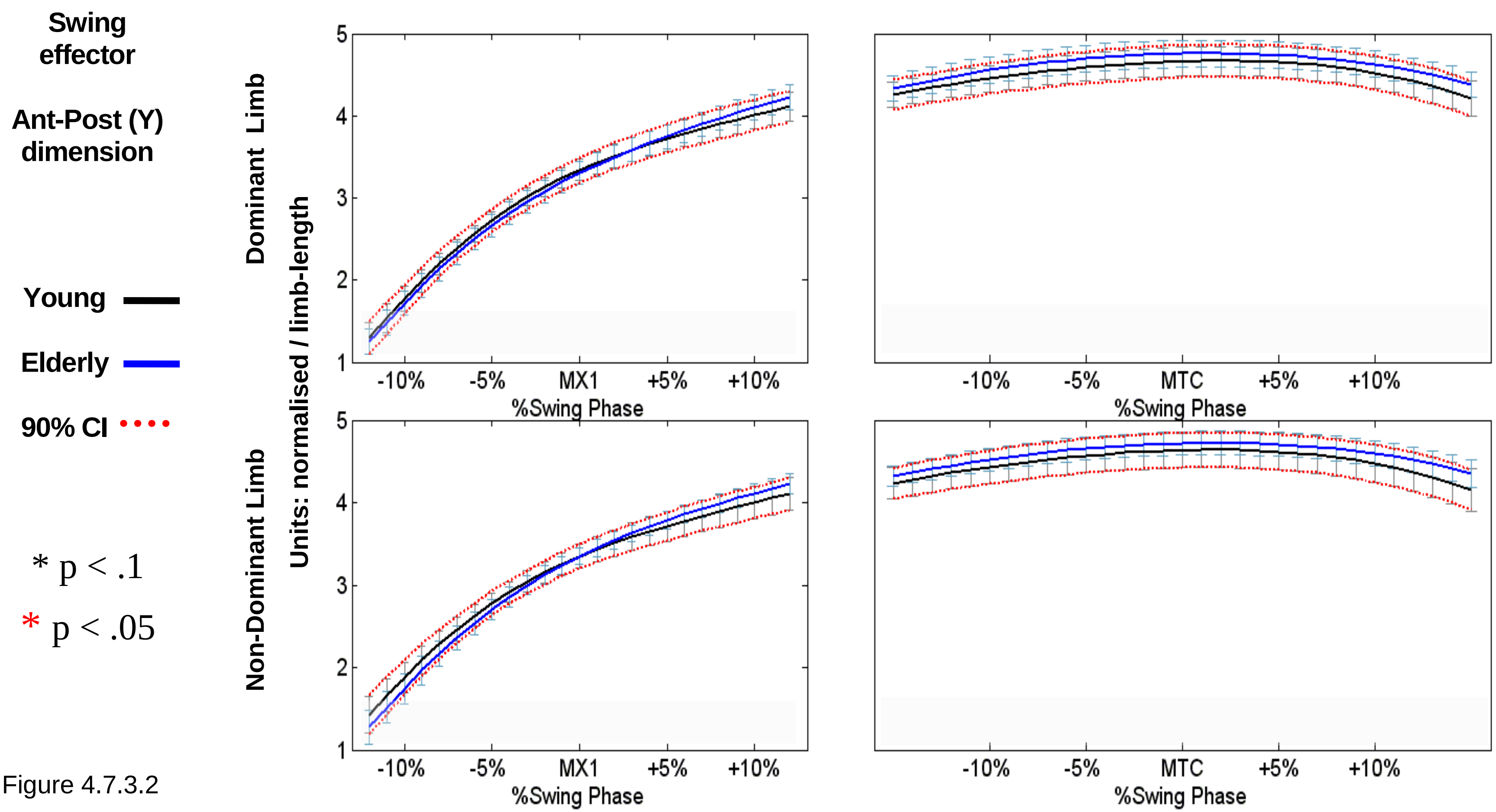


Figure 4.7.3.2

The anterior-posterior (Y) velocity component of the swing effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

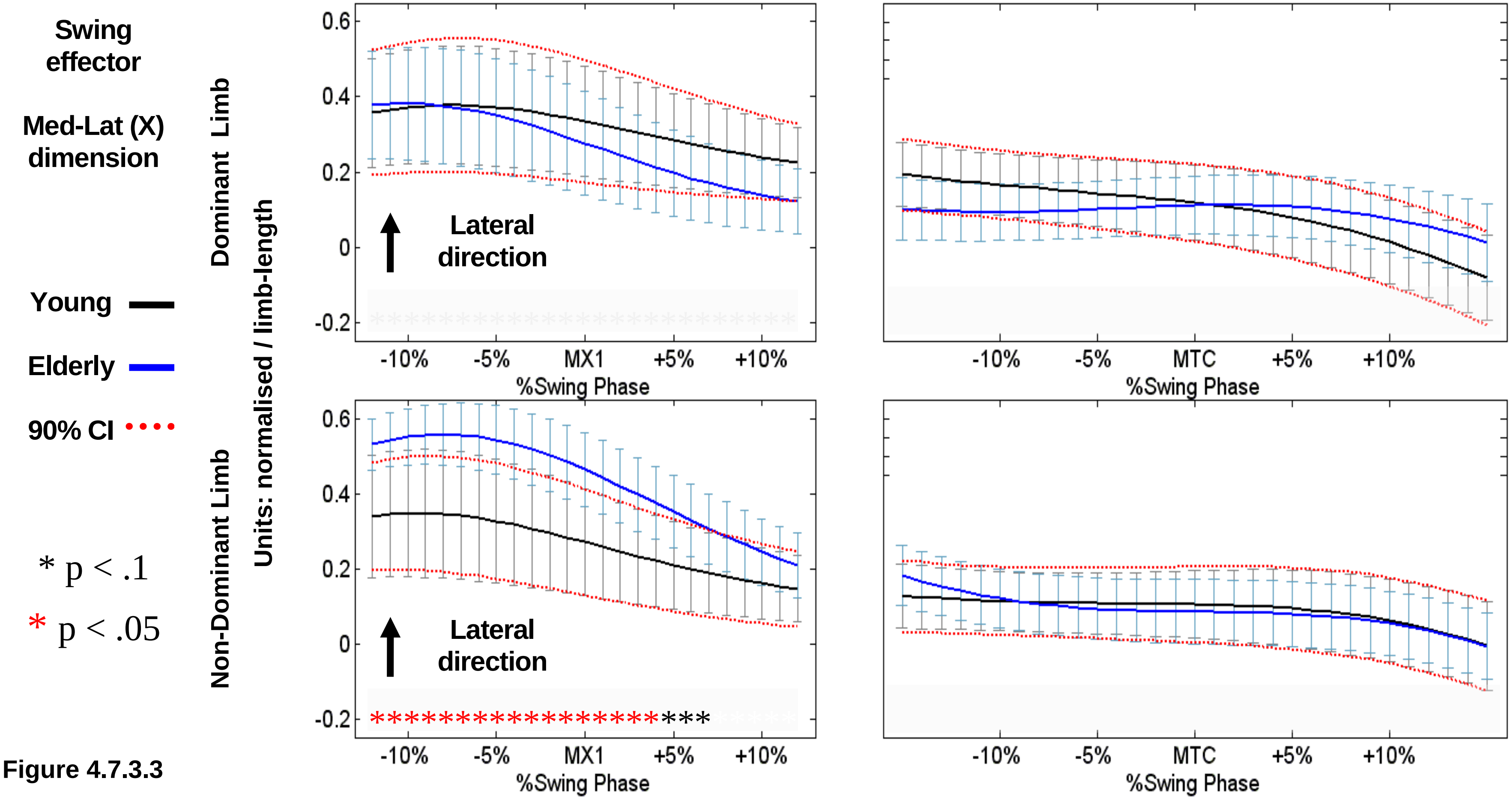


Figure 4.7.3.3

The medio-lateral (X) velocity component of the swing effector vector (ordinate units are velocity, m/s). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.8 Standard deviation of the components of the effector vectors

Most research exploring ageing changes to walking patterns has focused on time-distance parameters. The variability of effector trajectories provides specific details of where gait trajectories are performed consistently about an average state. The variability distribution of the effector states at each time-slice of the swing cycle forms a distribution representing convergence or instability of the effectors as they approach time slices relevant to MX1 and MTC. Comparing the effectors, as well as comparing between the dimension components of the variance distribution will provide new information to gait analysis. In line with the idea of ellipse fitting to describe the variance of two-dimensional data distributions, a three-dimensional sphere 'best-fit' to the three-dimensional data distribution obtained from the time-slices of the effector states would be a nice way of representing the data. Unfortunately, however, the reader will be left to visualise this three-dimensional distribution from one-dimensional presentations of the distribution.

This section will only address the variances associated within the separate effectors, and the co-variance interaction between the swing and stance effector systems upon the combined effector will be addressed in Chapter 7.

4.8.1 Combined effector: standard deviation of the vector components

The variance of the combined effector describes how the toe position is consistently controlled through the entire kinematic chain. It is expected that the elderly group will have relatively higher variance in the final toe position of the combined effector. Figure 4.8.1.1 indicates this only for the non-dominant limb. An interesting feature of the variance curve plot in the vertical dimension is the slight rise from early mid-swing and then plateaus forwards from MTC. It was expected that variance would reduce to a minimum at MTC for all participants, however, this minimum variance in the vertical dimension of the combined effector appears to occur at approximately 15% of swing phase time prior to MTC.

There is a larger variance in the elderly for the anterior-posterior dimension of the non-dominant limb at early-swing (Figure 5.6.3.1). Both the groups show an increase in variance in the anterior-posterior dimension when the swing phase progresses from early-swing to mid-swing. There is also significantly higher variance in the medio-lateral dimension of the elderly group in the region leading up to MX1 (Figure 5.6.3.1). This is evident when comparing groups within both the dominant and non-dominant limbs.

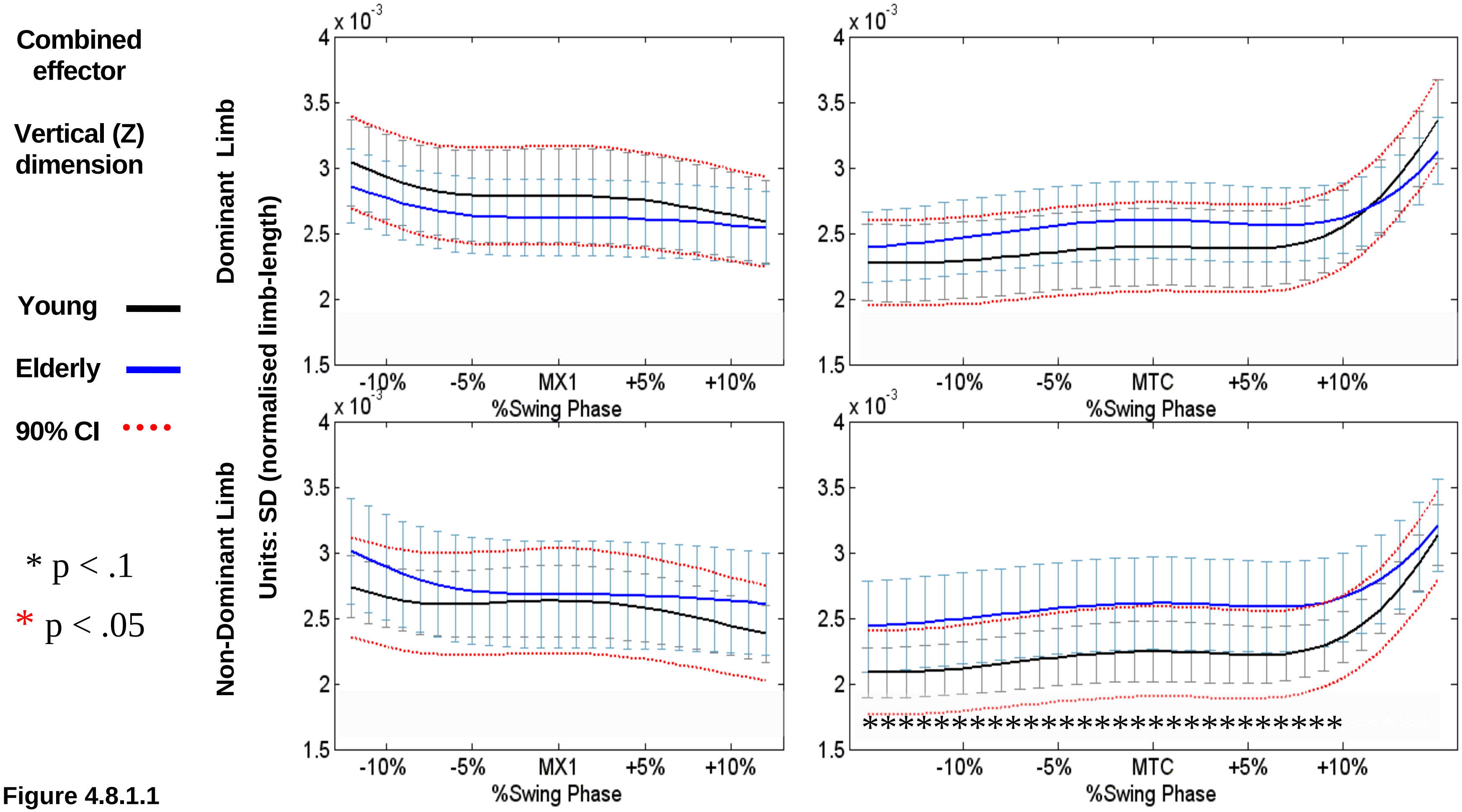


Figure 4.8.1.1

The vertical (Z) component of the combined effector vector. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

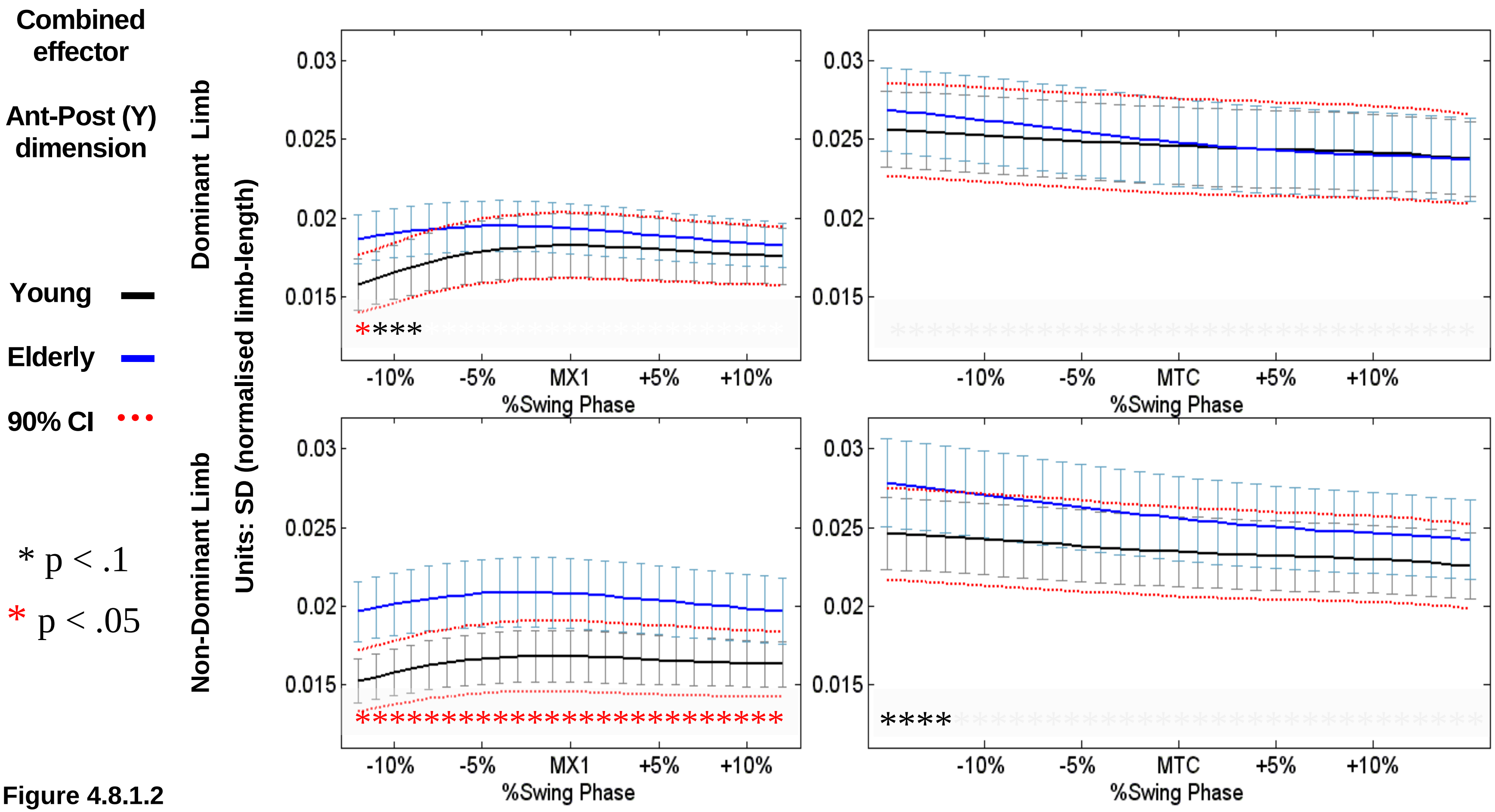


Figure 4.8.1.2

The anterior-posterior (Y) component of the combined effector vector. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1

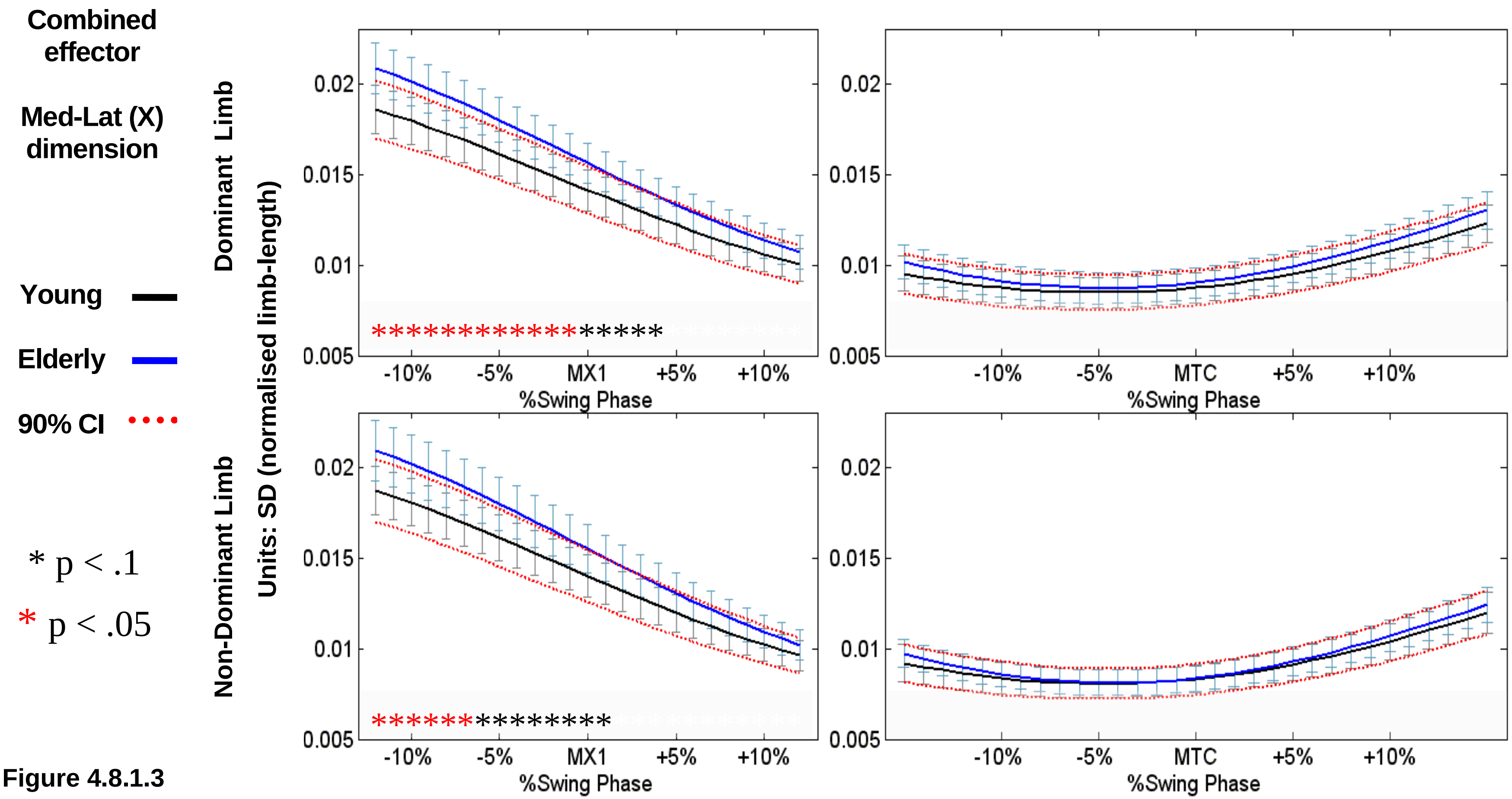


Figure 4.8.1.3

The medio-lateral (X) component of the combined effector vector. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.8.2 Stance effector: standard deviation of the component lengths

The variance in the stance effector describes the between cycle consistency in controlling the swing hip position relative to the stance foot. Figures 4.8.2.1-3 shows convincingly that the variance in the stance effector is significantly ($p<.01$) larger for the older group in all three components throughout the MX1 region in both limbs. These trends continue into the MTC region but the mean difference between groups is smaller (Figure 4.8.2.1). Generally, the larger variance occurs within the anterior-posterior dimension, followed by the medio-lateral and the vertical dimension respectively (note the different scales of the ordinates of Figures 4.8.2.1-3). For all figures, there is consistency of the group mean 95% confidence intervals. For the variance parameter associated with the stance effector, it is evident that these two groups belong to two different populations.

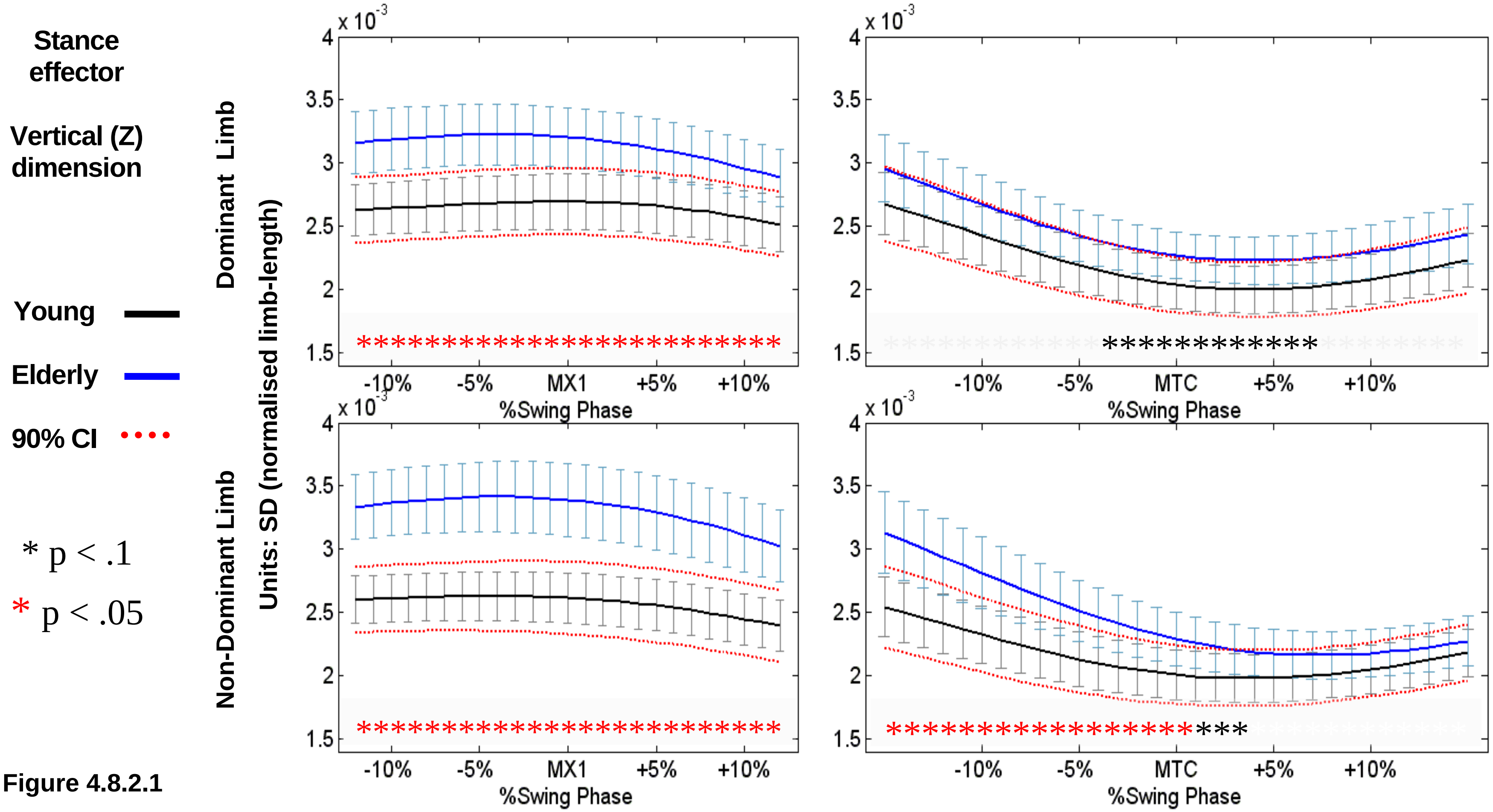


Figure 4.8.2.1

The vertical (Z) component of the stance effector vector. The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

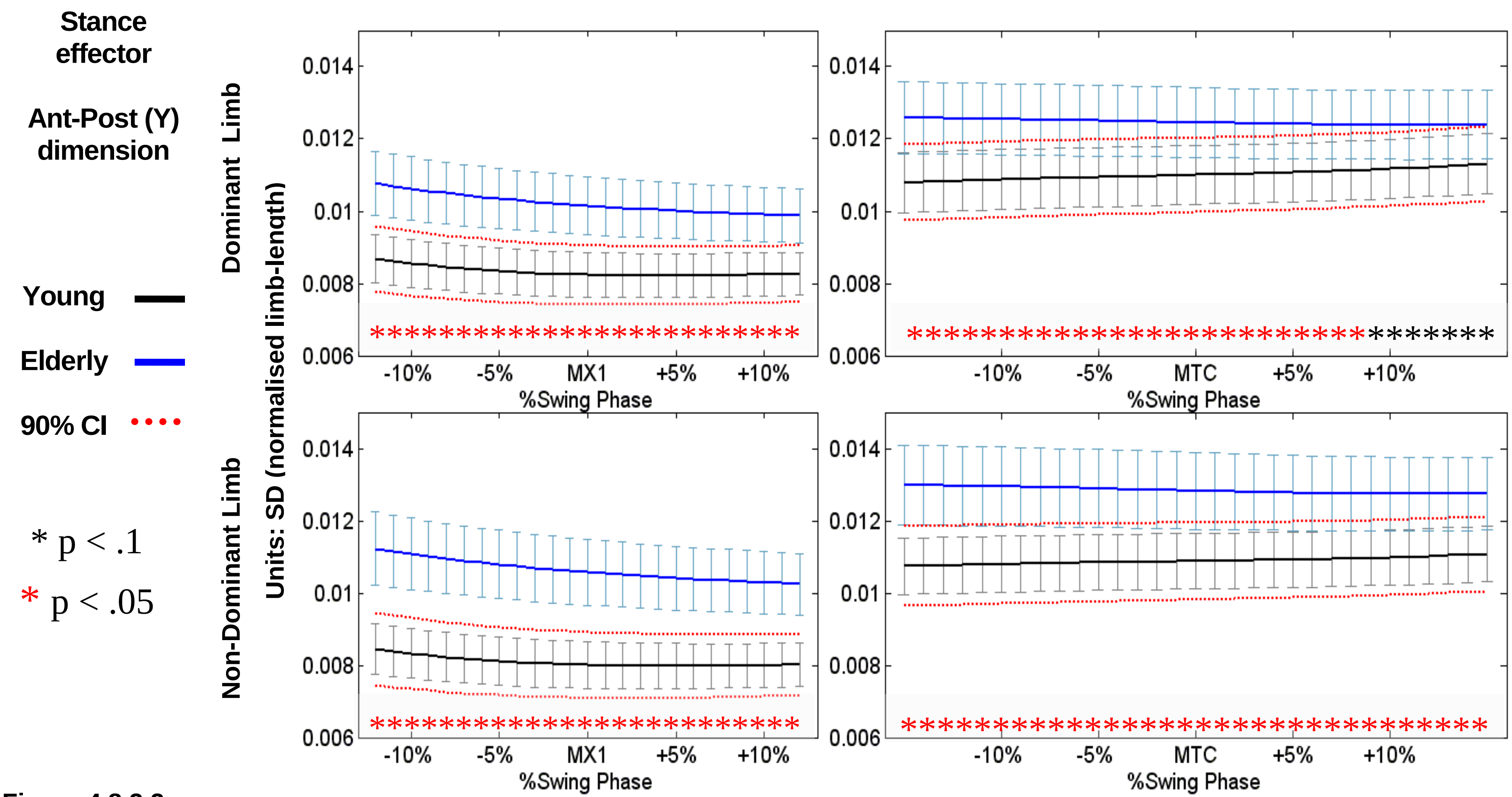


Figure 4.8.2.2

The anterior-posterior (Y) component of the stance effector vector. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

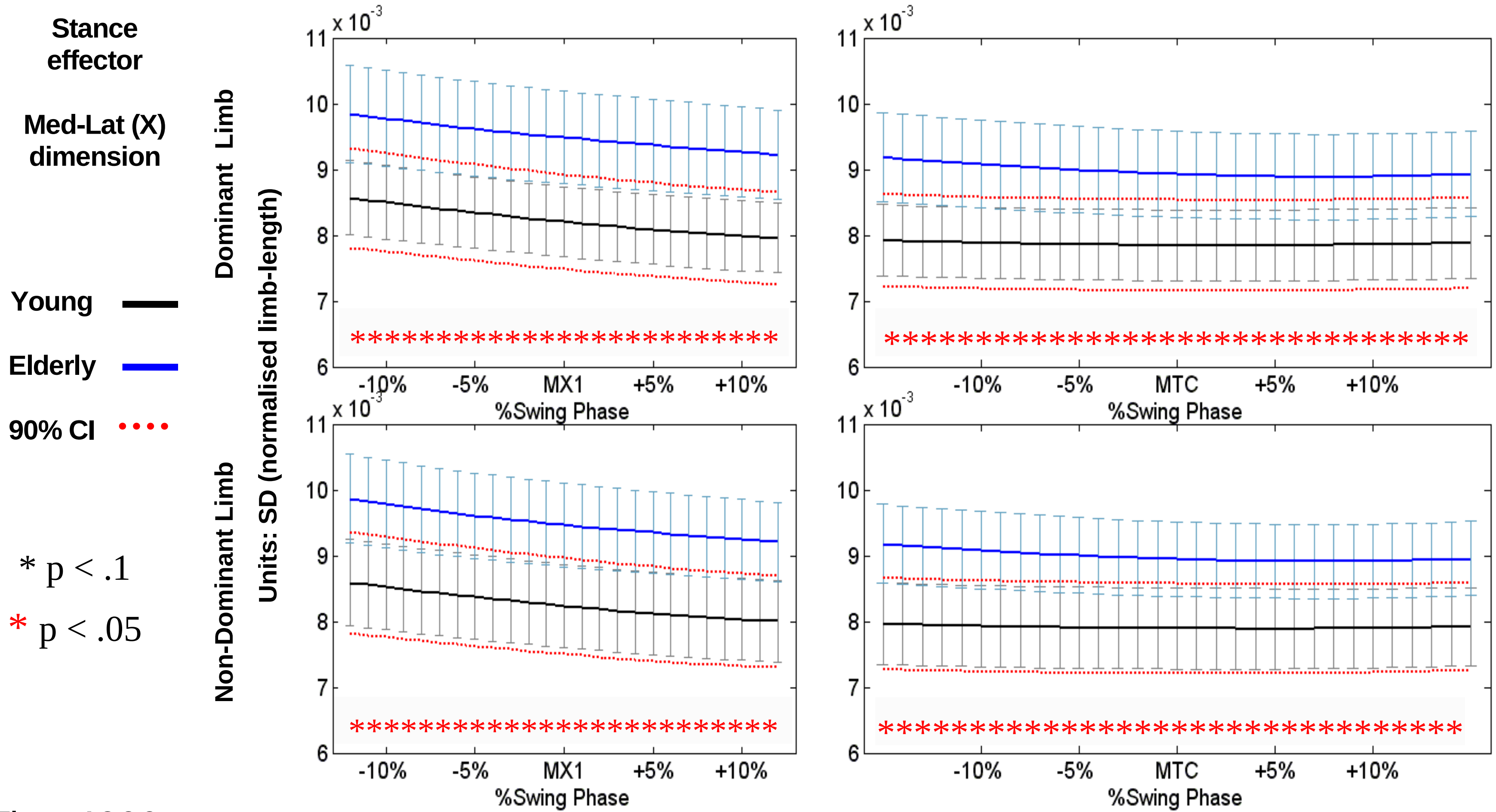


Figure 4.8.2.3

The medio-lateral (X) component of the stance effector vector. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.8.3 Swing effector: standard deviation of the component lengths

The variance in the stance effector describes the between cycle consistency in reproducing the same effector configuration. In contrast to the stance effector, Figures 4.8.3.1-3 show that the variance in the swing effector is not significantly different between groups for the vertical and anterior-posterior dimensions. The lower group mean for the elderly in the medio-lateral dimension of the MTC region is close to the 90% confidence interval (Figure 4.8.3.3). These results generally suggest that the variance in the elderly swing effector indicates that they can reproduce its configuration from cycle-to-cycle just as consistently as the young group.

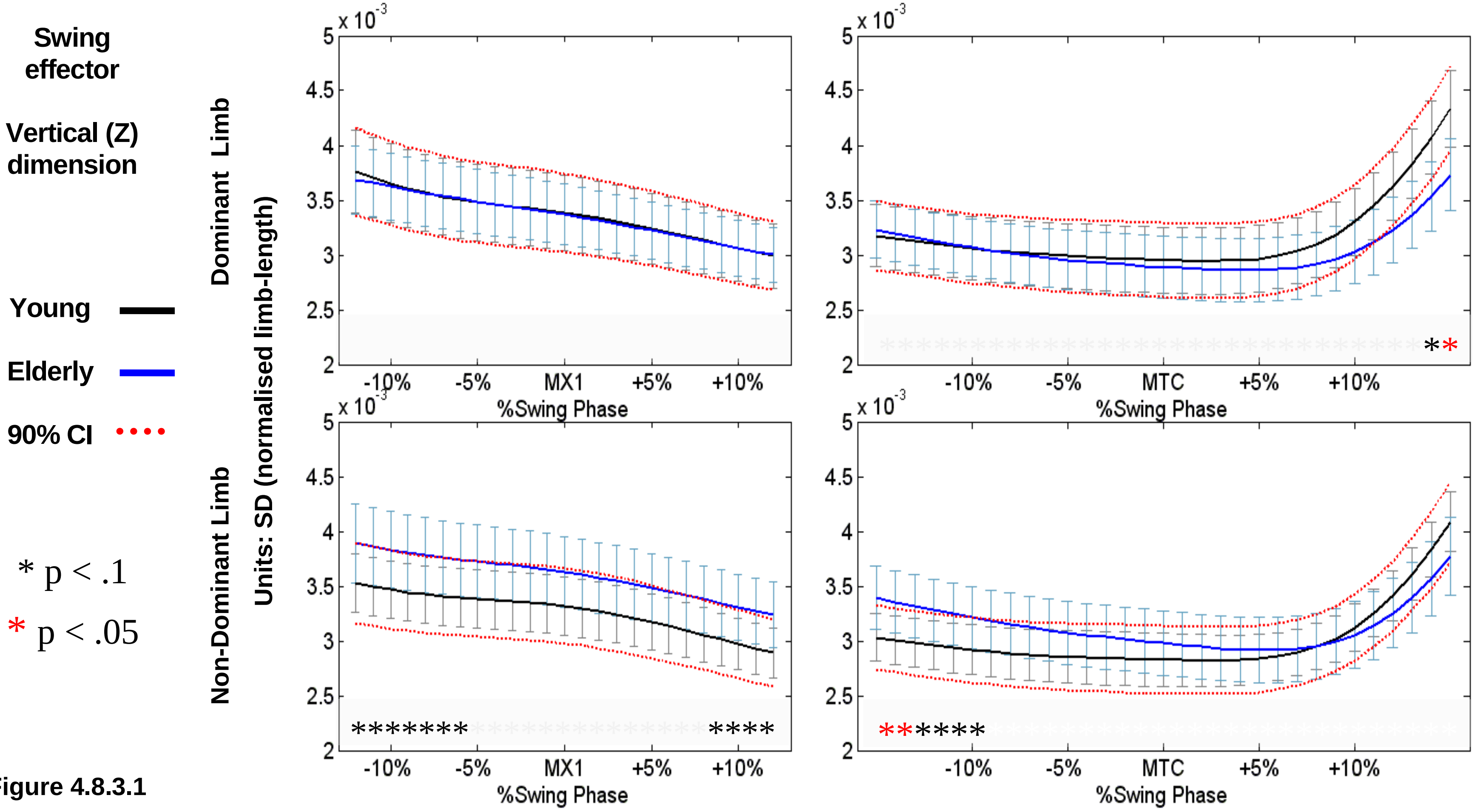


Figure 4.8.3.1

The vertical (Z) component of the swing effector vector (ordinate units are standard deviation). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

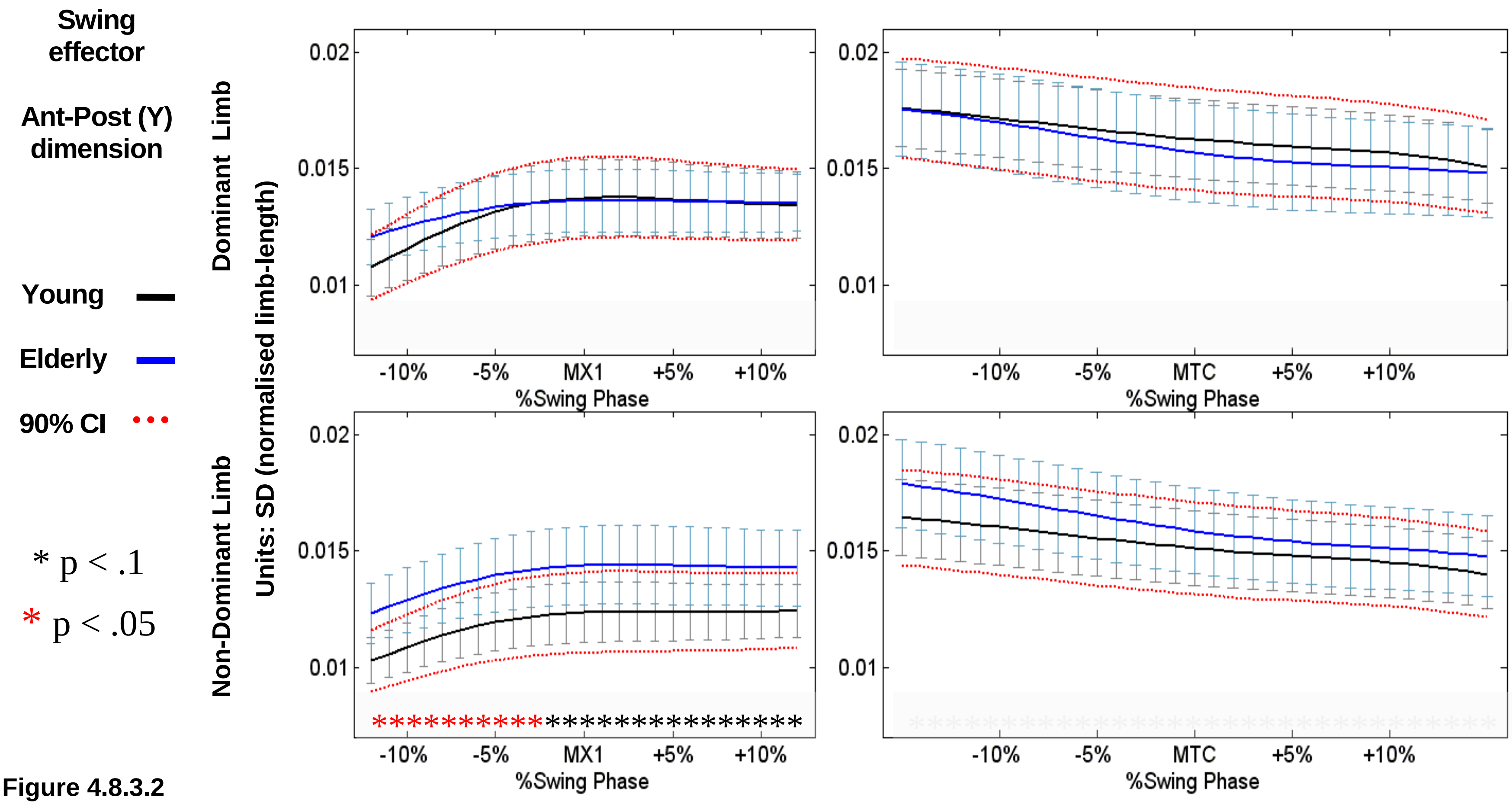


Figure 4.8.3.2

The anterior-posterior (Y) component of the swing effector vector (ordinate units are standard deviation). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

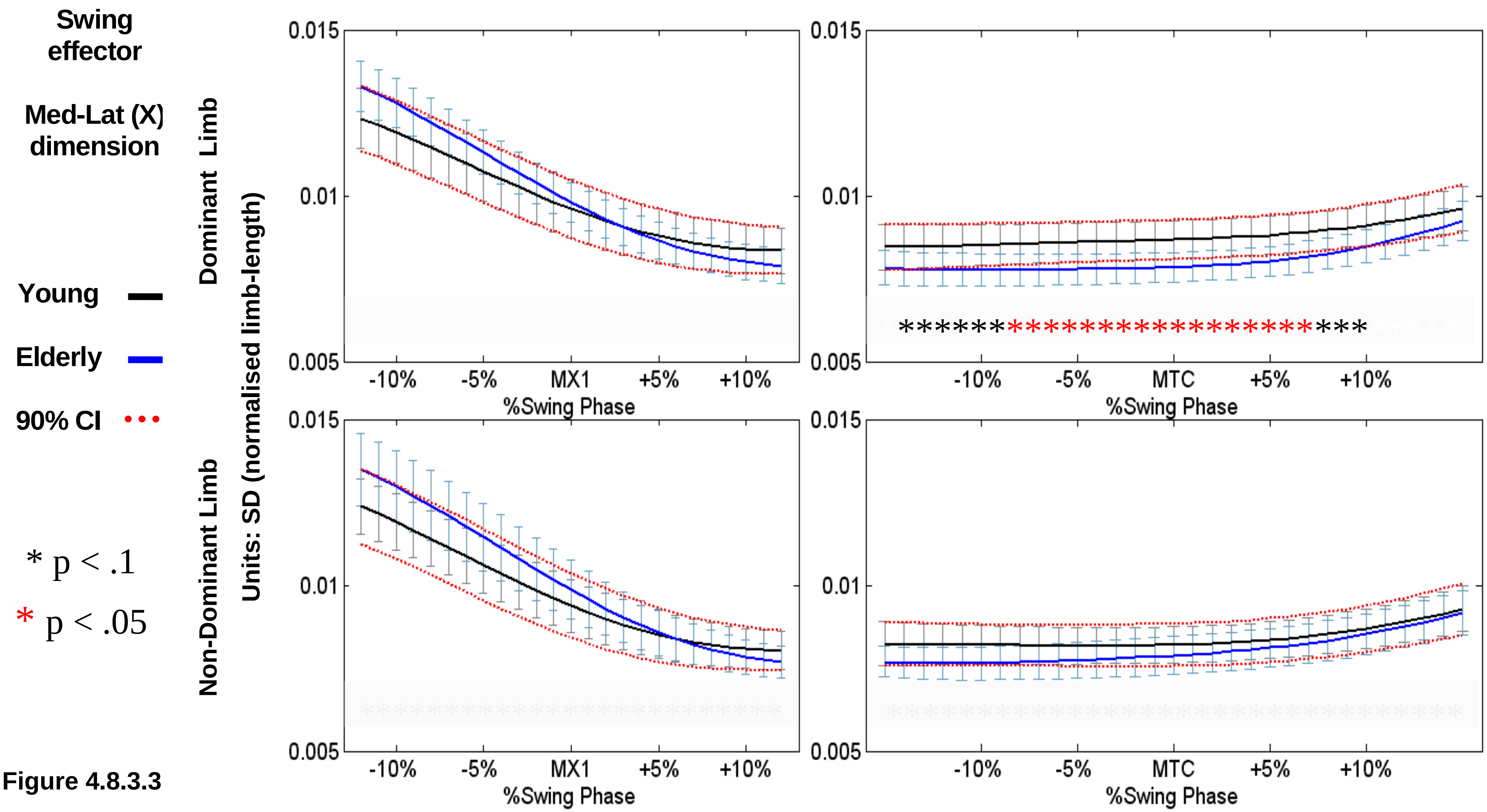


Figure 4.8.3.3

The medio-lateral (X) component of the swing effector vector (ordinate units are standard deviation). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.8.4 Interaction effects: changes in effector variance between MX1 and MTC

This section addresses the general aim of investigating effector variance between groups and task-relevant states MX1 and MTC. Three general levels of analysis were applied in different ways to analyse effector variance; the variables for analysis were standard deviation (dependent variable), and two independent variables: effector (vertical dimension) and state (MX1 and MTC). The first level was at the within effector level. The second level was at the between effector level. The third level addressed interactions between independent variables of effector, state and group. A paired t-test design was applied to test the first null hypothesis that variance will not be significantly different within an effector when comparing between states MX1 and MTC. A one-way ANOVA tested the second null hypothesis, that the effectors will not have significantly different variance at states MX1 and MTC (within each group and collapsing for limb). The third null hypothesis proposes that variance between the effectors will change uniformly between states MX1 and MTC, i.e. there will not be a significant interaction effect between effector and state. The fourth null hypothesis proposes that the young and elderly will not be significantly different in the way the variance changes from MX1 to MTC when comparing between the effectors; i.e. there will not be a significant interaction effect between effector, state and group. A three-way mixed design ANOVA with repeated measures on state (see section 4.2.1.1) was applied to test the third and fourth null hypotheses (as outlined in section 4.1.1).

Figure 4.8.4.1A illustrates the mean and 95% confidence intervals for the vertical dimension of effector variance in the three effectors at MX1 and MTC; comparisons are displayed between the young and elderly. When comparing within-effector variance in the young group, the null hypothesis proposal that 'there will be no significant difference between states' was rejected for all three effectors ($t(110) > 3.13$, $p < .005$). This result indicates that the stance ($p<.001$) and swing ($p<.005$) effector trajectories has significantly less variance (i.e. become more convergent) from MX1 to MTC. However, while the combined effector trajectories are also significantly less variant from MX1 to MTC in the young group ($t(110)=2.72$, $p<.01$), the elderly group do not demonstrate a significant difference ($t(110)=0.26$, $p>.1$). The elderly group results indicate that the distribution of trajectories in the combined effector do not become significantly more convergent from MX1 to MTC.

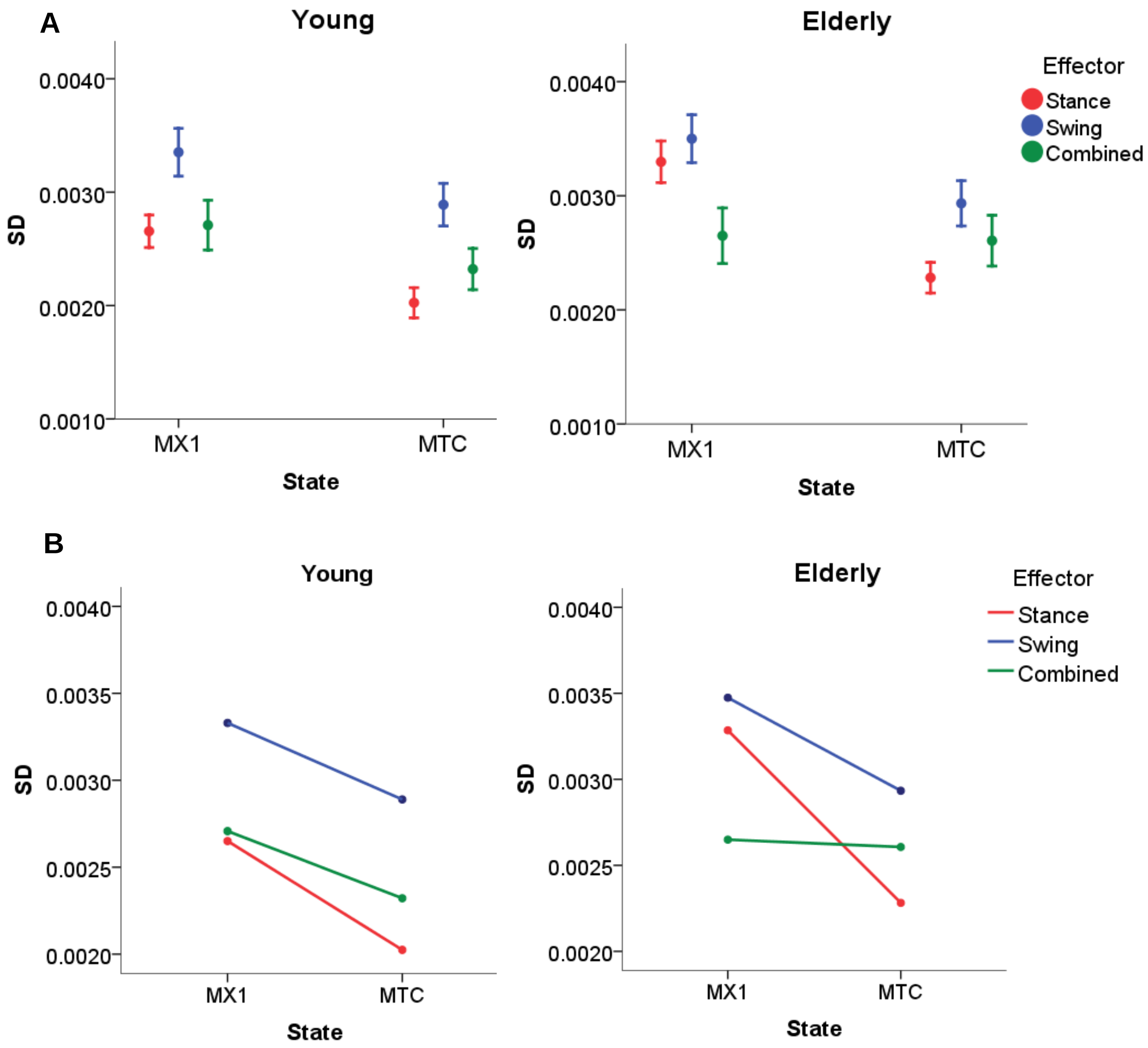


Figure 4.8.4.1

A) Group mean error-bars (95% confidence intervals) for vertical dimension of effector variance between MX1 and MTC. The data from both limbs has been pooled (i.e. n=56), because within the groups, a paired t-test reveals that there is no significant difference between limbs for effector variance (p>.1). B) Illustration of effector and state interactions for the young and elderly.

See text for significant differences at four levels of statistical comparisons.

The second level of analysis investigated variance between the effectors within each group (young and elderly) and state (MX1 and MTC). These four comparisons were analysed using one-way ANOVA with Bonferroni's post-hoc comparisons (adjusting $p<.01$). The results of the one-way ANOVA was significant in all four tests, where $F(2, 165)>11$, and $p<.001$. The post-hoc comparisons demonstrated the following between effector differences. For the young and elderly group at MTC, the stance effector variance was significantly lower than the swing effector ($p<.001$). However, at MX1, the elderly group did not demonstrate a significant difference ($p>.5$). When comparing the stance and combined effector variance at MTC, both the young and elderly demonstrate that the stance effector is significantly lower ($p<.05$). However, at MX1, the young group do not demonstrate a significant difference ($p>.5$). For both groups, the combined effector variance was found to be significantly lower than the swing effector variance at MX1 and MTC ($p<.05$).

Figure 4.8.4.1B illustrates that effector variance changes differently from MX1 to MTC when comparing between the young and elderly. Therefore, determining general behaviour of effector variance between states by pooling the groups together using a two-way mixed-design ANOVA is likely to provide misleading results. Because of this, interactions between effector variance and state were compared between groups, making this a three-way mixed-design ANOVA (with repeated measures on state). There was an interaction effect for effector × state × group ($F(2, 220) = 26.6$, $p < .001$, $\eta^2 = .195$). The combined effector was used in planned comparison with the stance and swing effectors separately, and therefore excludes comparisons between the swing and stance effectors. The justification was purely because of the hypothesis that the combined effector is the primary task-goal at MTC and obtains task-relevant contributions by the stance and swing effectors. There was a significant contrast between the combined effector variance and stance effector variance when comparing between MX1 and MTC states, and when compared between age ($F(1, 110) = 51.47$, $p < .001$, $\eta^2 =.32$). The planned contrast between the combined effector variance and the swing effector variance also revealed a significant contrast when comparing between states, and comparing for between age group differences ($F(1, 110) = 20.03$, $p < .001$, $\eta^2 =.15$). These results support graphical observations of the group means when comparing between the young and elderly in Figure 4.8.4.1B. The combined effector variance in the elderly decreases less between MX1 and MTC compared to the young group, even though the elderly decrease the variance of the stance and swing effectors from MX1 to MTC in a similar way to the young group. This raises questions about how the stance and swing effectors cooperate in the elderly group in terms of effecting performance behaviour of the combined effector trajectory.

4.9 Skewness of the components of the effector vectors

A null hypothesis about the loco-sensorimotor control system is that effector trajectories will display a normal distribution in three-dimensional space, because trajectories will be controlled everywhere, equally, in three-dimensional space. An alternate hypothesis proposed in this thesis is consistent with research theory demonstrating that the motor control system manages movement trajectories by prioritising the most relevant trajectories for achieving the task-goal (Scholz and Schoner, 1999; Todorov and Jordan, 2002). Research indicates that the motor control system applies a control policy that penalises trajectory errors in proportion to an assigned relevance for meeting the task-goal (Hamilton and Wolpert, 2002). For example, trajectory errors that compromise a task-goal have a greater penalty compared to trajectory errors that are less task-relevant. Large penalties require proportionately higher level of intervention by the motor control system to redirect the trajectory. Therefore, consistent with this theory, rather than control effector trajectories ‘everywhere’ in three-dimensional space, the loco-sensorimotor control system will manage trajectories selectively by applying a cost policy. A skewed distribution of trajectories in three-dimensional space will therefore reflect a cost policy, by awarding higher penalties for trajectory errors with most task-relevance to the toe-clearance task-goal. In relative contrast, when erroneous trajectories do not compromise the task-goal, a lesser penalty is awarded and such a policy is therefore proportionately asymmetric, thus leading to a hypothesised skewed distribution of trajectories. A positively skewed distribution at discrete-time events of the swing cycle represents more severe penalties at the low end; a negatively skewed distribution signals relatively stronger penalties at the higher end; and zero skewness (perfect normal distribution) indicates the control penalties are relatively balanced above and below a nominal trajectory.

4.9.1 Combined effector: skewness

The skewness of the combined effector trajectories for the young and elderly are compared at discrete-time events of the swing phase in Figures 4.9.1.1-3. The between-group comparisons are separated into dominant and non-dominant limbs, forming six swing-cycle plots, where each cycle is divided into MX1 and MTC regions. From the displayed p-value criterion in Figures 4.9.1.1-3, all three dimensions of the combined effector demonstrate that the elderly have significantly more positive skewness at various states of the swing phase. For the vertical dimension (Figure 4.9.1.1), the elderly have significantly more positive skewness at MX1 region of the swing cycle. This is most evident in the non-dominant limb.

For the anterior-posterior dimension (Figure 4.9.1.2), the elderly have significantly more positive skewness for all states of the MX1 region in both the dominant and non-dominant limbs. The young group have significantly greater negative skewness at the initial states leading up to MTC. There is a qualitative change towards negative skewness between the MX1 and MTC regions. This discrete disparity can be explained by the change between reference events, such that time slices are relevant to a different state. For example, effector states are made relevant to the MX1 or MTC states within the MX1 and MTC regions, respectively. There will be a natural discontinuity between MX1+12% and MTC-15% because the time-slices have made a discrete change by re-aligning to a new state-relevant reference. This will be more apparent when larger displacements occur in the anterior-posterior direction, which can be explained by larger velocities in the walking progression. This disparity does not occur to the same extent in the vertical and medio-lateral dimensions where effector velocities are reduced.

Figure 4.9.1.3 illustrates the combined effector skewness in the medio-lateral dimension. For the dominant limb comparison, the elderly show significantly greater positive skewness for states neighbouring MTC ($t(54) > 1.67$, $p < .1$). The light-blue error-bars indicate inter-subject differences in the elderly group. This is slightly higher than the inter-subject difference in the young group, however, Levene's test indicates that unequal variance is not significant at any of the swing cycle states ($p>.1$).

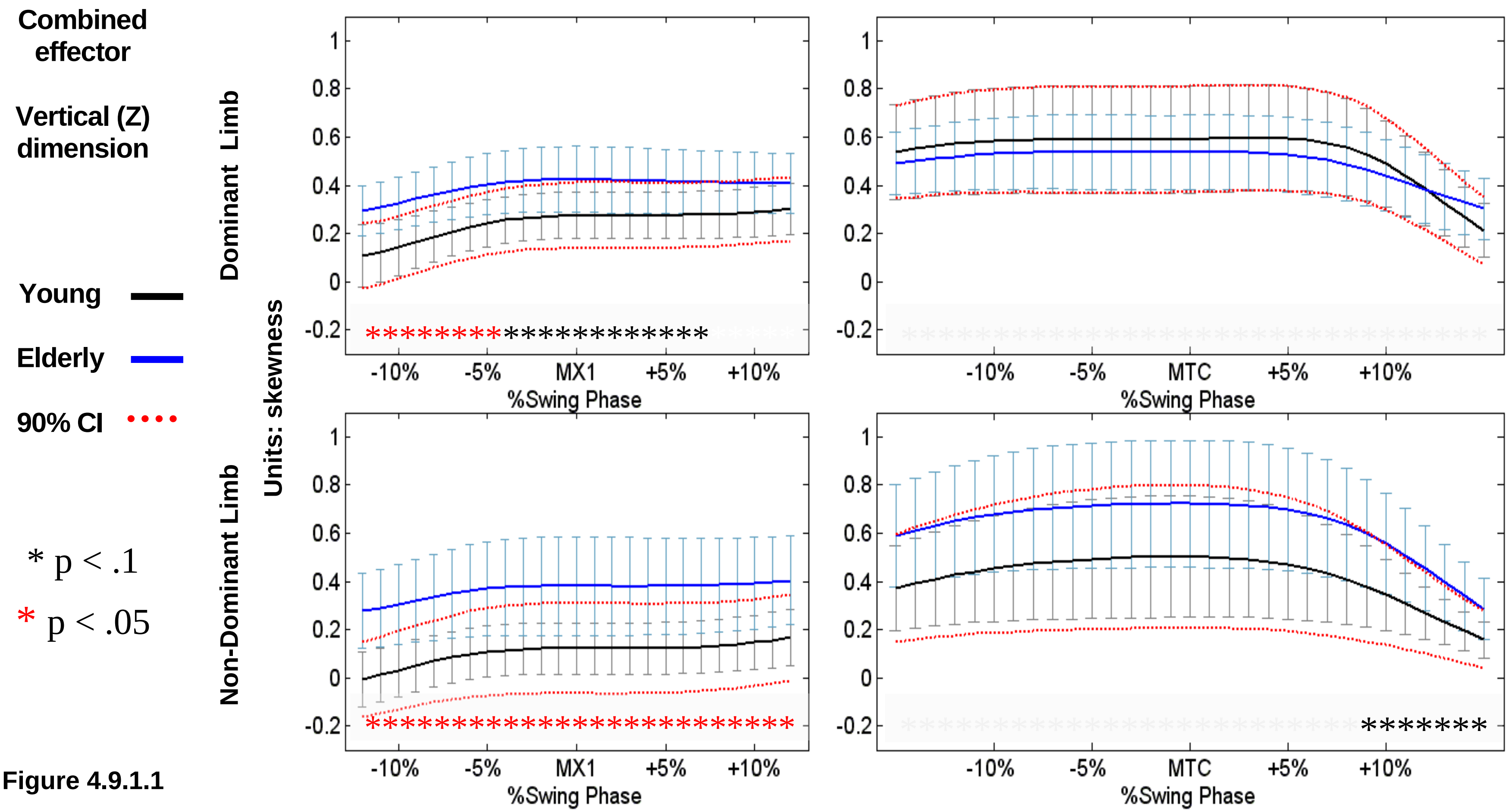


Figure 4.9.1.1

The vertical (Z) component of the combined effector vector (ordinate units are skewness). The abscissa represents the swing-time normalised percentage increments relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

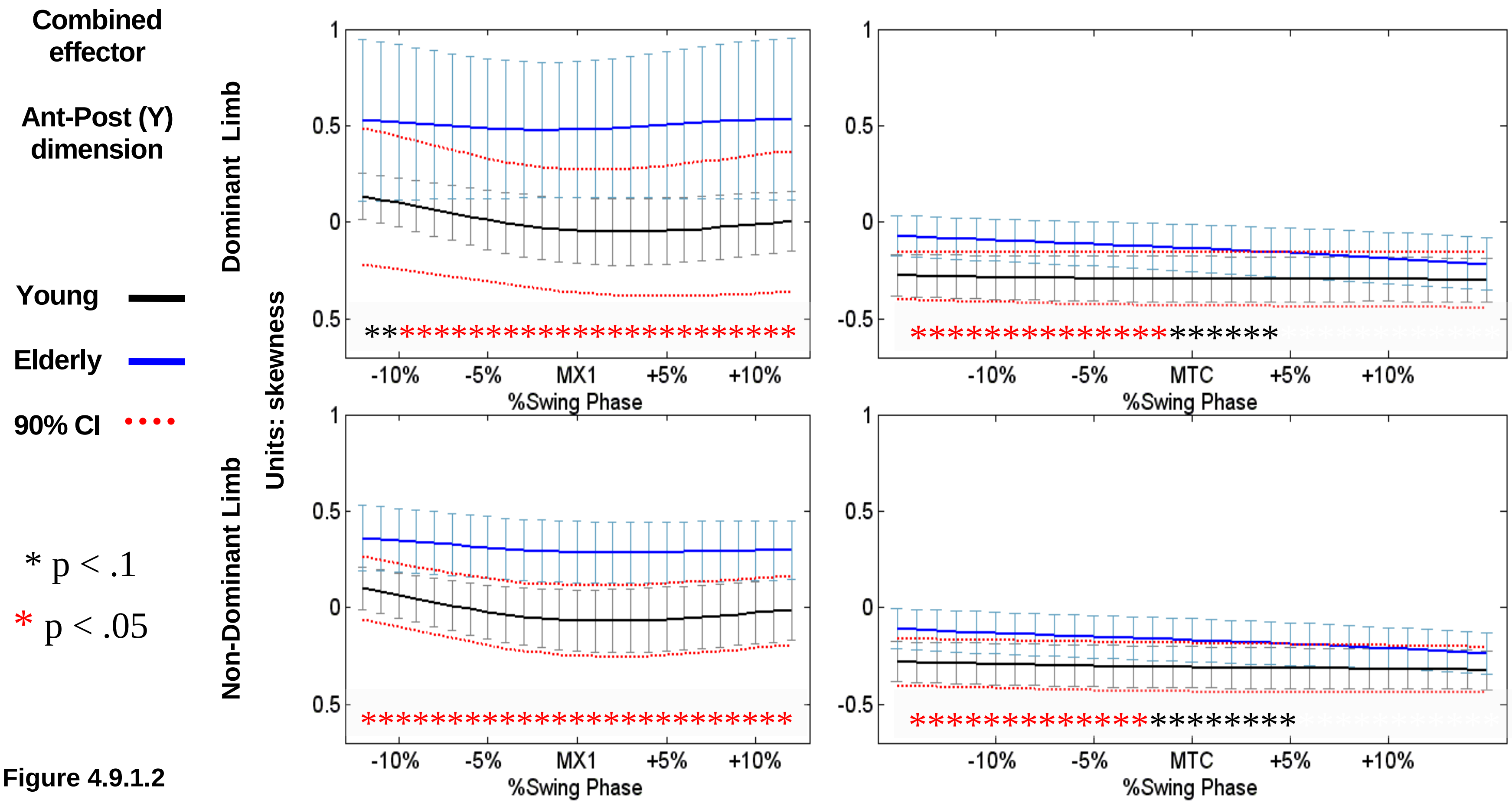


Figure 4.9.1.2

The anterior-posterior (Y) component of the combined effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

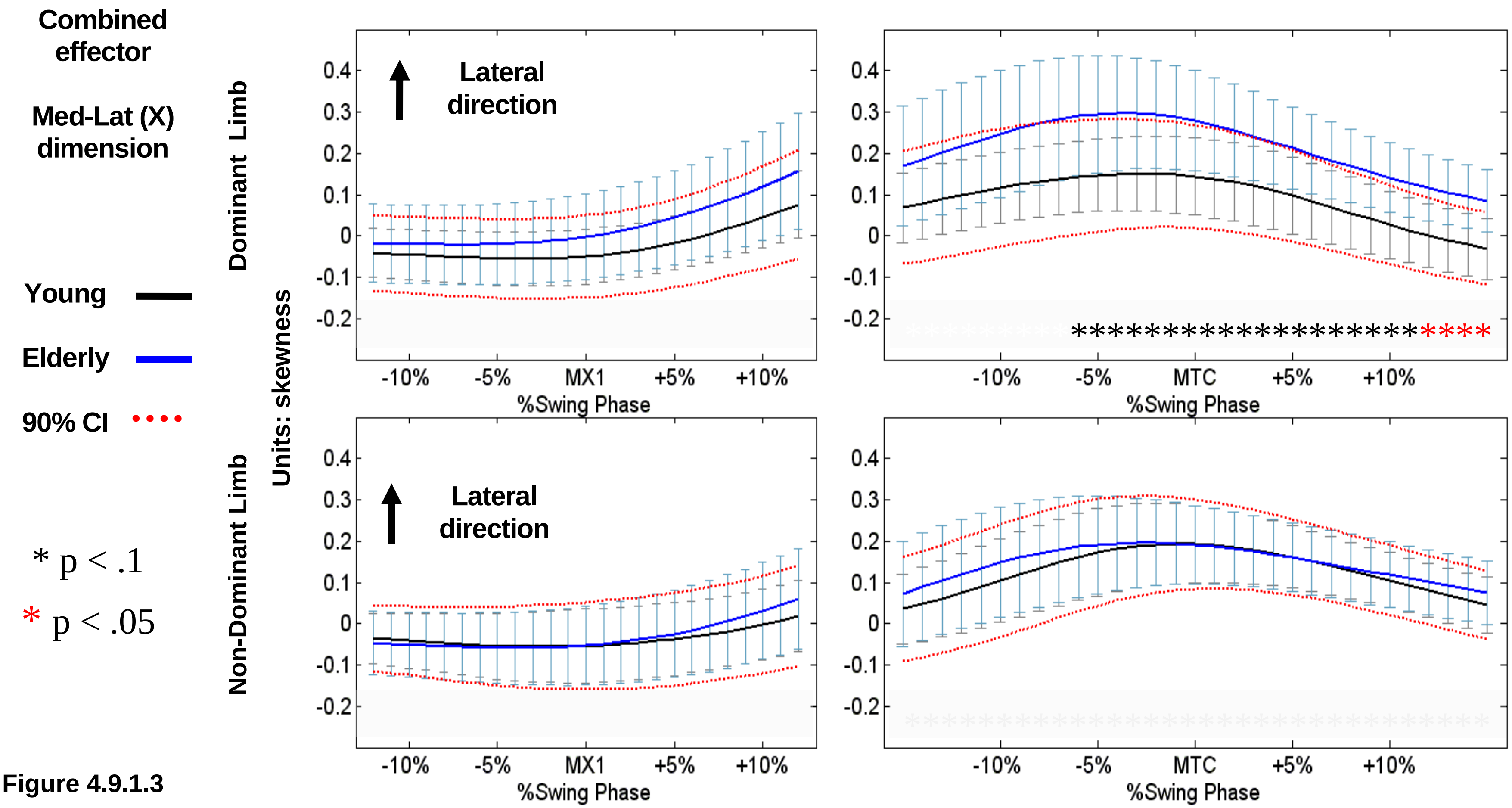


Figure 4.9.1.3

The medio-lateral (X) component of the combined effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.9.2 Stance effector: skewness

The statistical testing of stance and swing effector skewness was affected by an outlier case from the elderly group that exists in the vertical dimension of the stance and swing effector data. The outlier was considered to be behaving far from the normal group (Figure 4.9.2.1). However, this case was not an identified outlier in the vertical dimension of the combined effector. The data curve displayed in Figures 4.9.2.3-4 includes the outlier case because the case wasn't recognised as an outlier in the anterior-posterior or medio-lateral dimensions of swing or stance effectors. In the vertical dimension it has a big effect on the group mean, but more importantly the group variance, hence making Levene's statistical test of unequal variance between groups significant ($p < .001$). Figures 4.9.2.2-4 will consequently show the curve that includes the outlier. However, for the vertical dimension there is an additional symbol illustrated to represent evidence of significant between-group difference when the outlier has been removed. The discussion of statistical results of skewness in the vertical dimension of the stance and swing effector trajectories will be based upon the exclusion of case #55.

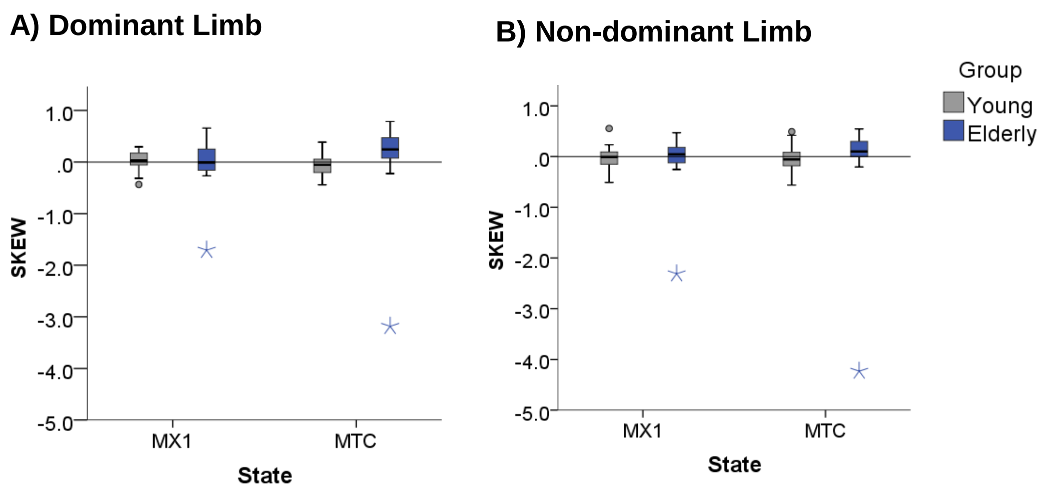


Figure 4.9.2.1

Outlier identified for skewness in the vertical dimension of the stance effector. The same case (#55) was also an identified outlier case with large positive skewness in the vertical dimension of the swing effector. This case was removed for statistical testing of skewness in the stance and swing effector comparisons.

Figures 4.9.2.2-4 show where the elderly display significantly more positive skewness during the swing cycle. For the vertical dimension of the stance effector (Figure 4.9.1.2), the elderly have significantly more positive skewness at the final states of the MX1 region and all the states of the MTC region. This is evident in both the dominant and non-dominant limbs. From Figure 4.9.2.2, there is information relative to the young group that indicates negative skewness is significantly different to 0 (i.e. 95% confidence intervals are significantly below 0, $p<.05$) at state-relevant times between MTC-15% to MTC-5%. When contrasting between groups in the dominant limb at MTC-5%, this result is opposite between groups. The dominant limb of the elderly group at this state-relevant time is significantly positive, i.e. skewness is significantly greater than 0 ($p<.05$). This tells us the policy of control for vertical stance effector states is opposite between the groups leading up to MTC. A mean negative skewness in the stance effector may represent a large penalty cost awarded to relatively large vertical displacement states on the swing hip trajectory, whereas low states are not penalised with the same proportionate cost. Such an asymmetric policy associated with the stance effector of the young group may correspond with a task-goal that is less related to the toe-clearance goal, and more-so related to posture stability or energy minimisation.

For the anterior-posterior dimension (Figure 4.9.1.3), the elderly also have significantly ($p < .05$) more positive skewness for all states of the swing cycle in both the dominant and non-dominant limbs. There is a qualitative change towards negative skewness between the MX1 and MTC regions. The young group have significantly ($p<.05$) greater negative skewness at all states of the MTC region. Figure 4.9.1.4 illustrates the stance effector skewness in the medio-lateral dimension. Both groups display mean skewness within ±0.1 of zero skewness throughout the swing cycle.

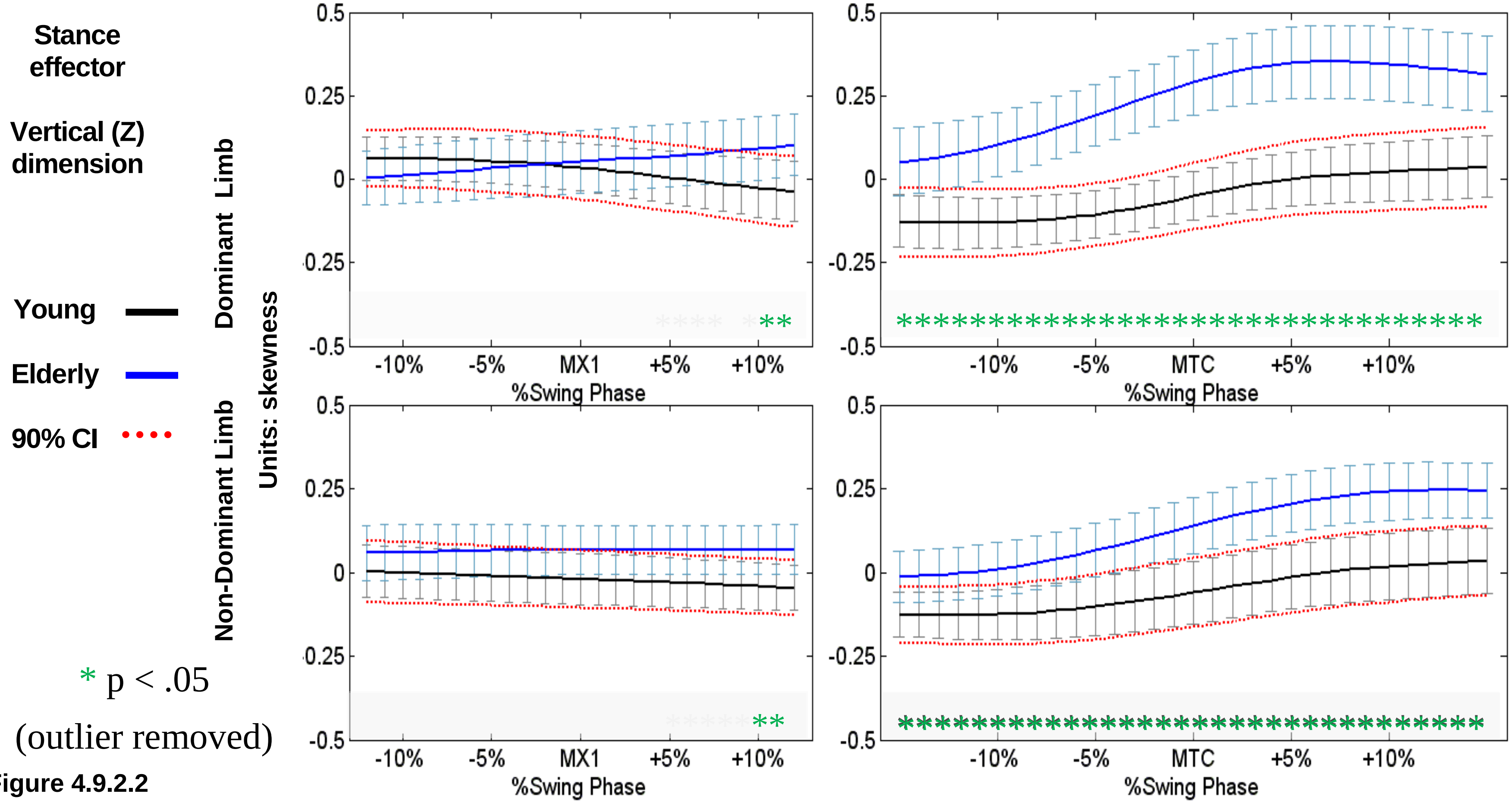


Figure 4.9.2.2

The vertical (Z) component of the stance effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1. Outlier case #55 has been removed from the elderly group (i.e. n_E=27).

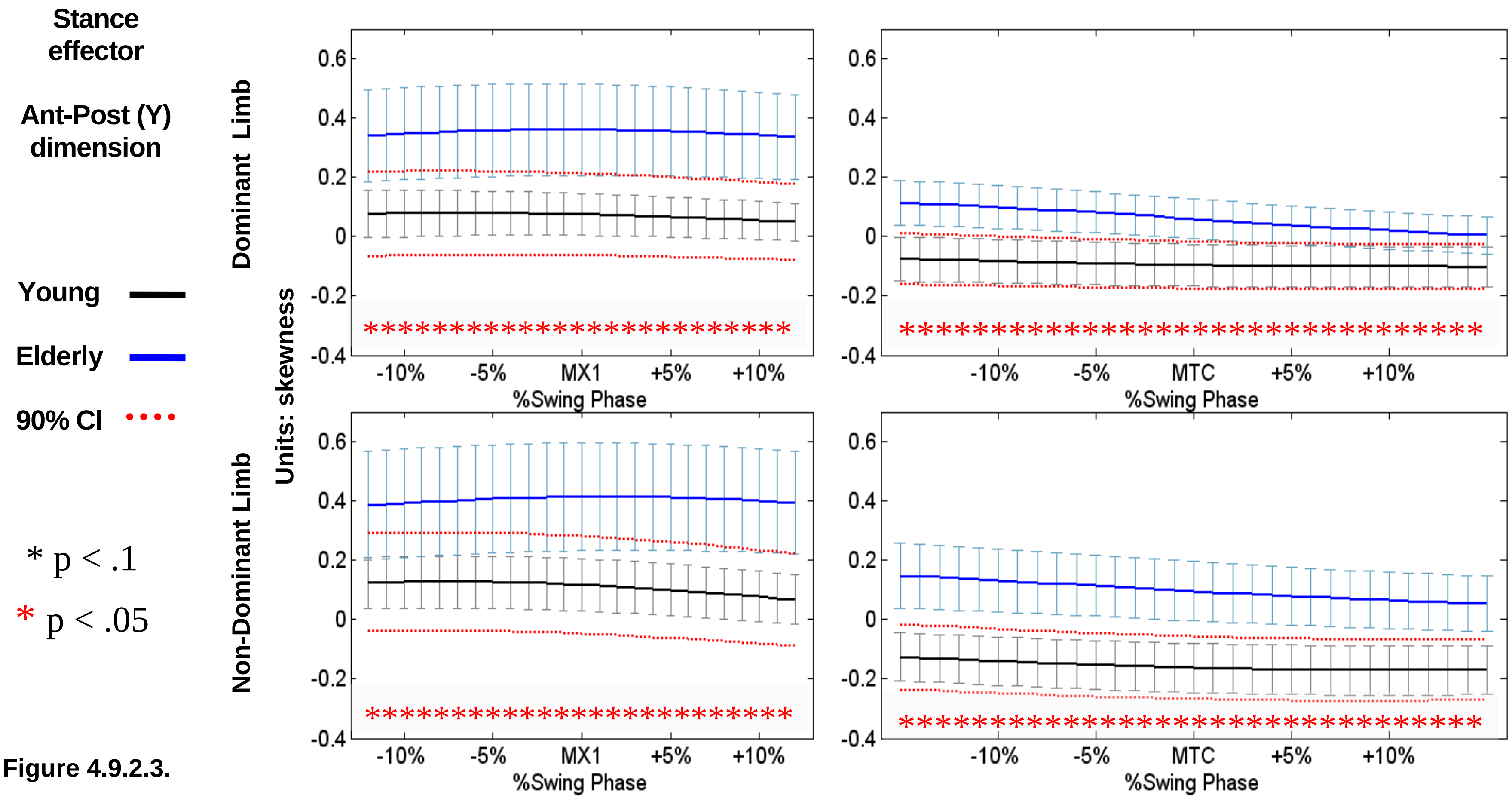


Figure 4.9.2.3.

The anterior-posterior (Y) component of the stance effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

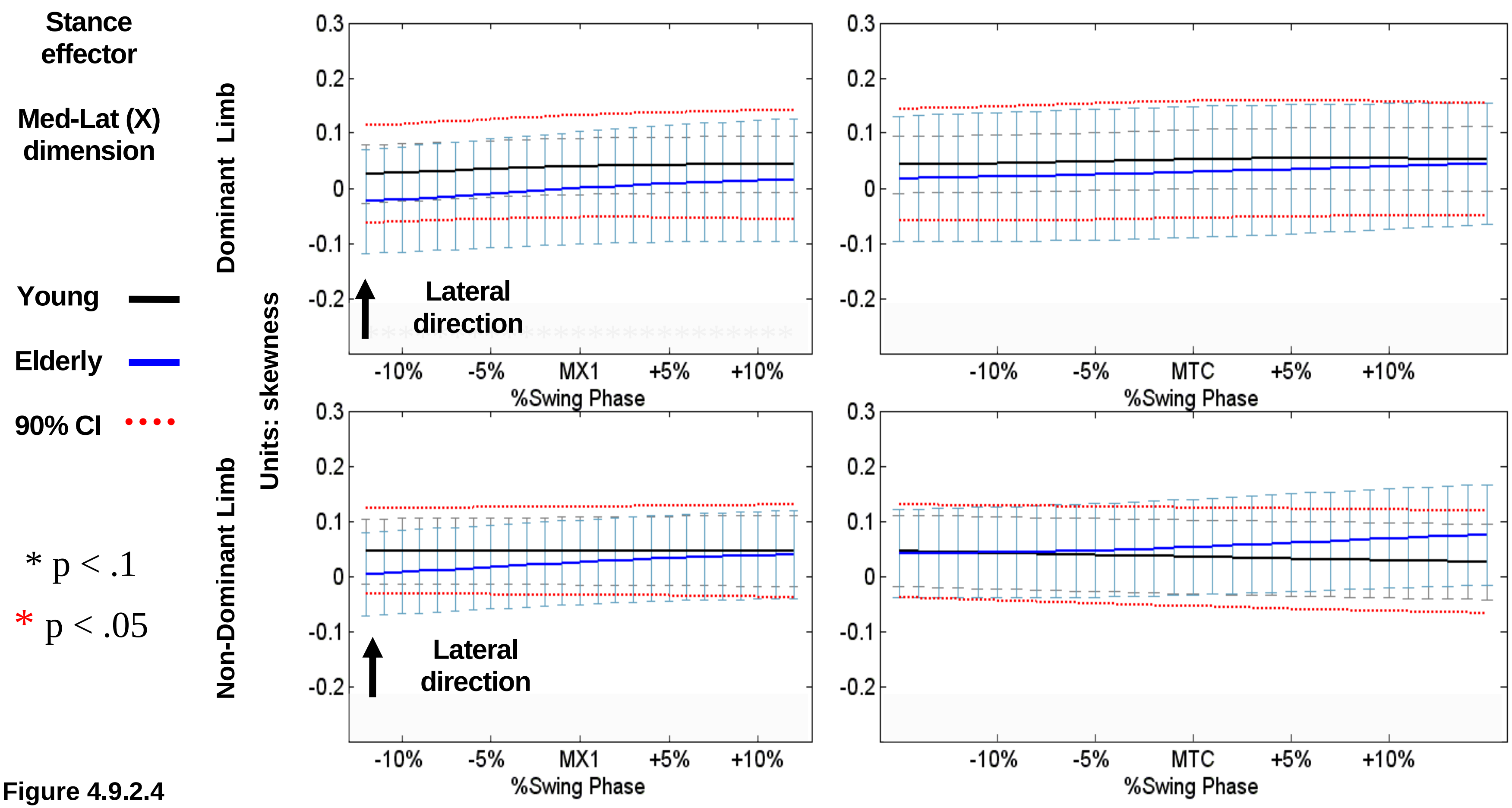


Figure 4.9.2.4

The medio-lateral (X) component of the stance effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.9.3 Swing effector: skewness

Figures 4.9.3.1-3 show swing effector skewness. The vertical dimension is displayed in Figure 4.9.3.1; the elderly group show significantly lower skewness at the state corresponding to MTC+5% of swing cycle time. There is also a significantly ($p < .05$) larger positive skewness in the elderly group at the initial states of the MX1 region (i.e. MX1-12%) for both the dominant and non-dominant limbs. Visually, there appears to be a difference between the limbs for the elderly group. Indeed, from a paired sample t-test there is a significant difference ($p < .05$) between the dominant and non-dominant limbs of the elderly, between the states MTC-5% to MTC+5%. In contrast, the young group between-limb differences are not significantly different ($p > .1$).

Figure 4.9.3.2 again shows the elderly have significantly larger positive skewness in the anterior-posterior dimension at the MX1 region. This indicates that the elderly significantly reduce (or penalise) swing effector trajectories with a posterior orientation in the states neighbouring MX1, in comparison to the young group who display a mean skewness on the negative side of zero. When observing the change in skewness from the MX1 region to the MTC region, the data of the swing effector skewness behaves similar to the stance and combined effectors. There is a noticeable shifting towards negative skewness when the discrete-time states are re-aligned with the MTC state. For the MTC region, there are significant differences between the groups for the dominant limb; the younger group show significantly ($p < .05$) larger negative skewness in the states between 'MTC-15%' to 'MTC-3%'. There is a particularly low confidence interval for both groups at MTC region. This suggests that while both groups are penalising swing trajectories in forward direction compared to trajectories in the posterior direction, the young group display this level of trajectory control significantly more than the elderly group in the dominant limb.

Figure 4.9.3.3 illustrates the swing effector skewness in the medio-lateral dimension. The intra-group variance in the dominant-limb of the elderly data was also affected by case#55; however, the shape of the group mean and between-group differences were negligible after case#55 was removed. There is a significant difference between the groups in the dominant-limb for states neighbouring MTC. In the states that follow MTC, the elderly have significantly larger positive skewness. This indicates that the swing trajectories of the elderly are penalised in the medial direction, whereas trajectories in the lateral direction are less penalised. This is significantly different to the young group. In contrast, for the non-dominant limb, the differences between the young and elderly are reversed in the states following MTC.

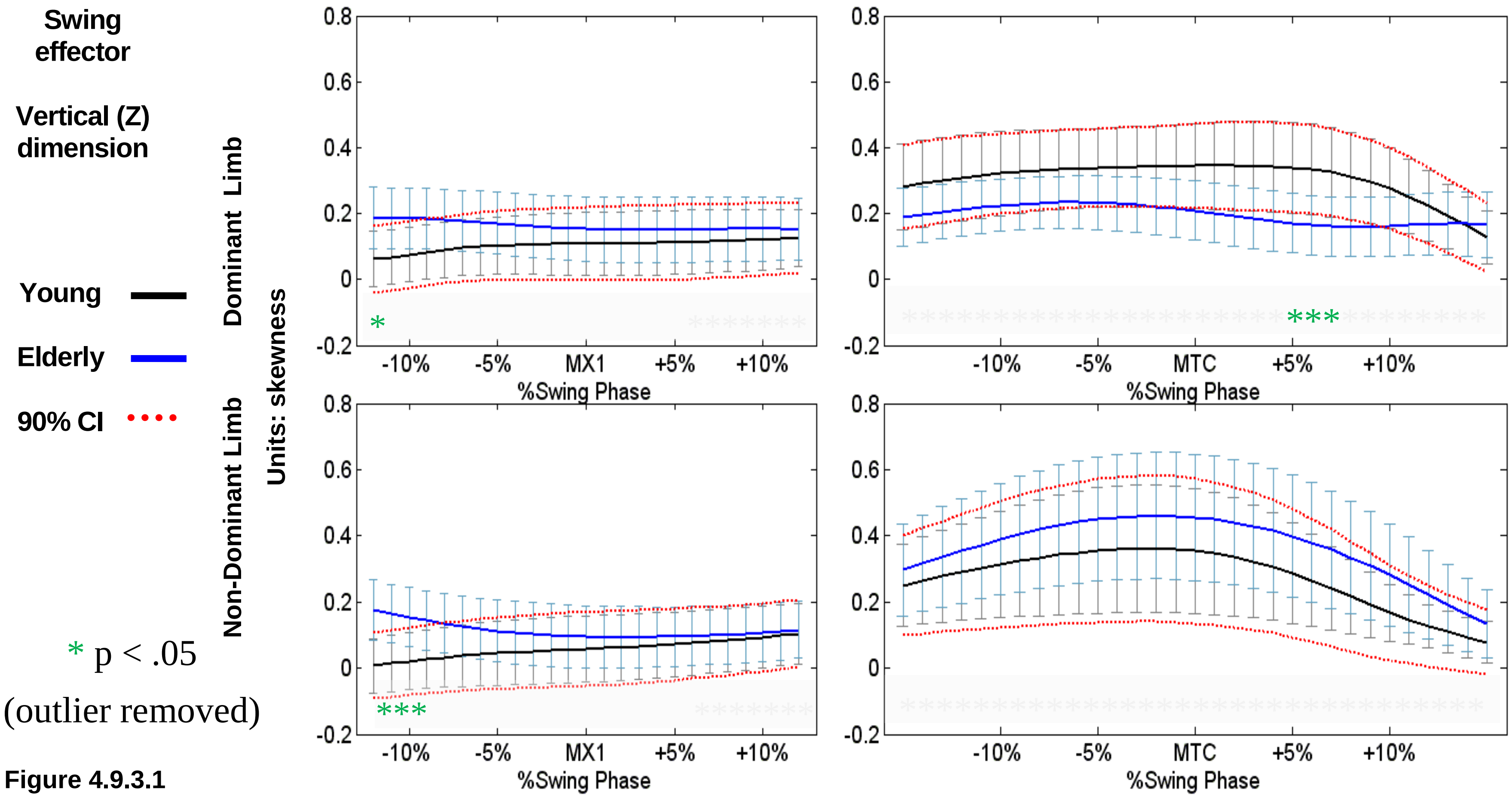


Figure 4.9.3.1

The vertical (Z) component of the swing effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the green asterisks emphasises where time-states show between-group significant differences of p<.05.

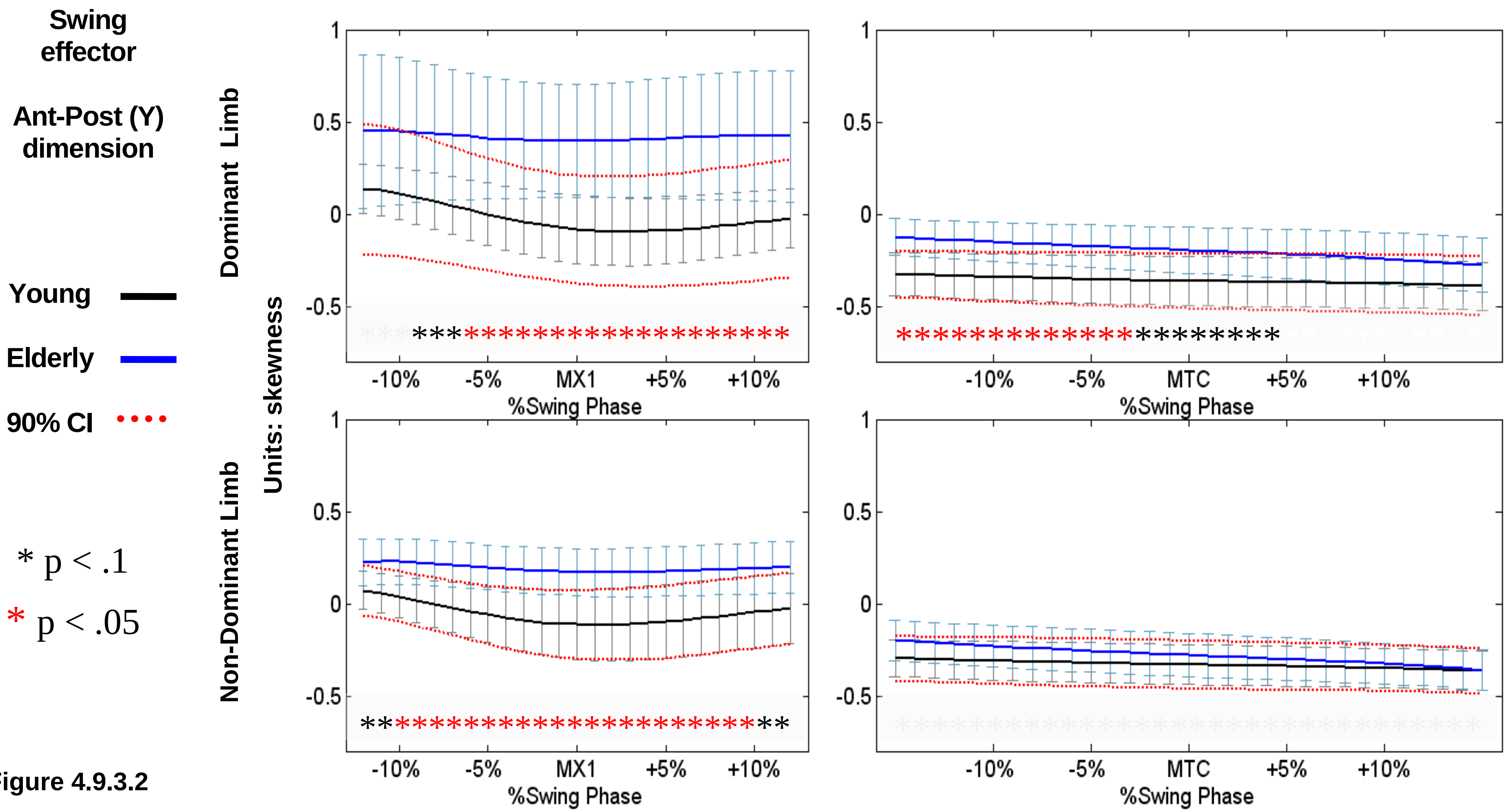


Figure 4.9.3.2

The anterior-posterior (Y) component of the swing effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

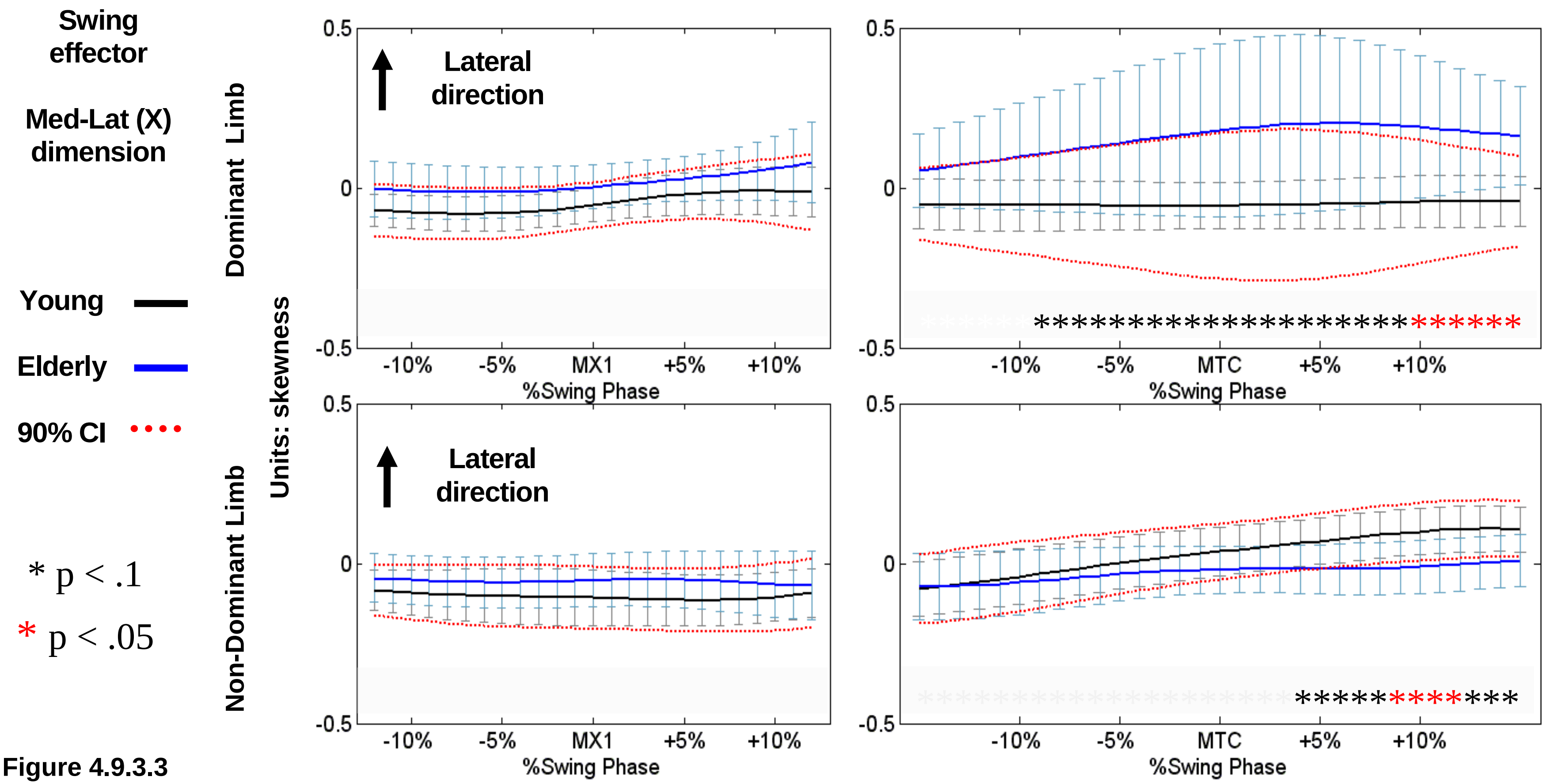


Figure 4.9.3.3

The medio-lateral (X) component of the swing effector vector (ordinate units are skewness). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.9.4 Interaction effects: changes in effector skewness between MX1 and MTC

This section addresses the general aim of investigating effector skewness between groups and task-relevant states MX1 and MTC. There were four levels of analysis applied to explore the nature of independent variables effector, state and group on the dependent variable of skewness. This was conducted in the same way as section 4.8.4, except the dependent variable in this analysis was skewness. The first null hypothesis proposes that skewness will not be significantly different within an effector when comparing between states MX1 and MTC. The second null hypothesis proposes that the effectors will not have significantly different skewness at states MX1 and MTC. The third null hypothesis proposes that skewness between the effectors will change uniformly between states MX1 and MTC, i.e. there will not be a significant interaction effect between effector and state. The fourth null hypothesis proposes that the young and elderly will not be significantly different in the way the skewness changes from MX1 to MTC when comparing between the effectors; i.e. there will not be a significant interaction effect between effector, state and group.

A paired t-test design was applied to determine the first null hypothesis within each group and collapsing for limb. A one-way ANOVA tested the second null hypothesis within each group and collapsing for limb. For the third and fourth null hypotheses a mixed design ANOVA with repeat measures (see section 4.2.1.4) was applied to test the null hypotheses specifically outlined in section 4.1.1.

Figure 4.9.4.1A illustrates the mean and 95% confidence intervals for the vertical dimension of effector skewness in the three effectors at MX1 and MTC; comparisons are displayed between the young and elderly. When comparing within effector skewness in the elderly group, the null hypothesis proposal that 'there will be no significant difference between states' was rejected for the combined-effector and swing-effector, in both groups ($t(110) > 2.1$, $p < .05$). There is significantly more positive skewness at MTC compared to MX1, within the swing effector ($p<.005$), and within the combined effector ($p < .05$). This suggests that relatively small vertical displacement (vertical 'lengths') of the combined-effector and the swing-effector are assigned a greater control penalty in comparison to 'penalties' assigned to relatively large vertical displacements. However, while the stance effector trajectories are also significantly more positively skewed between MX1 to MTC in the elderly group ($t(106)=3.49$, $p<.005$), the young group do not demonstrate a significant difference for the stance effector ($t(106)=1.65$, $p > .1$). The young group indicates that the vertical states of the stance effector trajectories do not have the same 'asymmetric' control policy employed as the elderly.

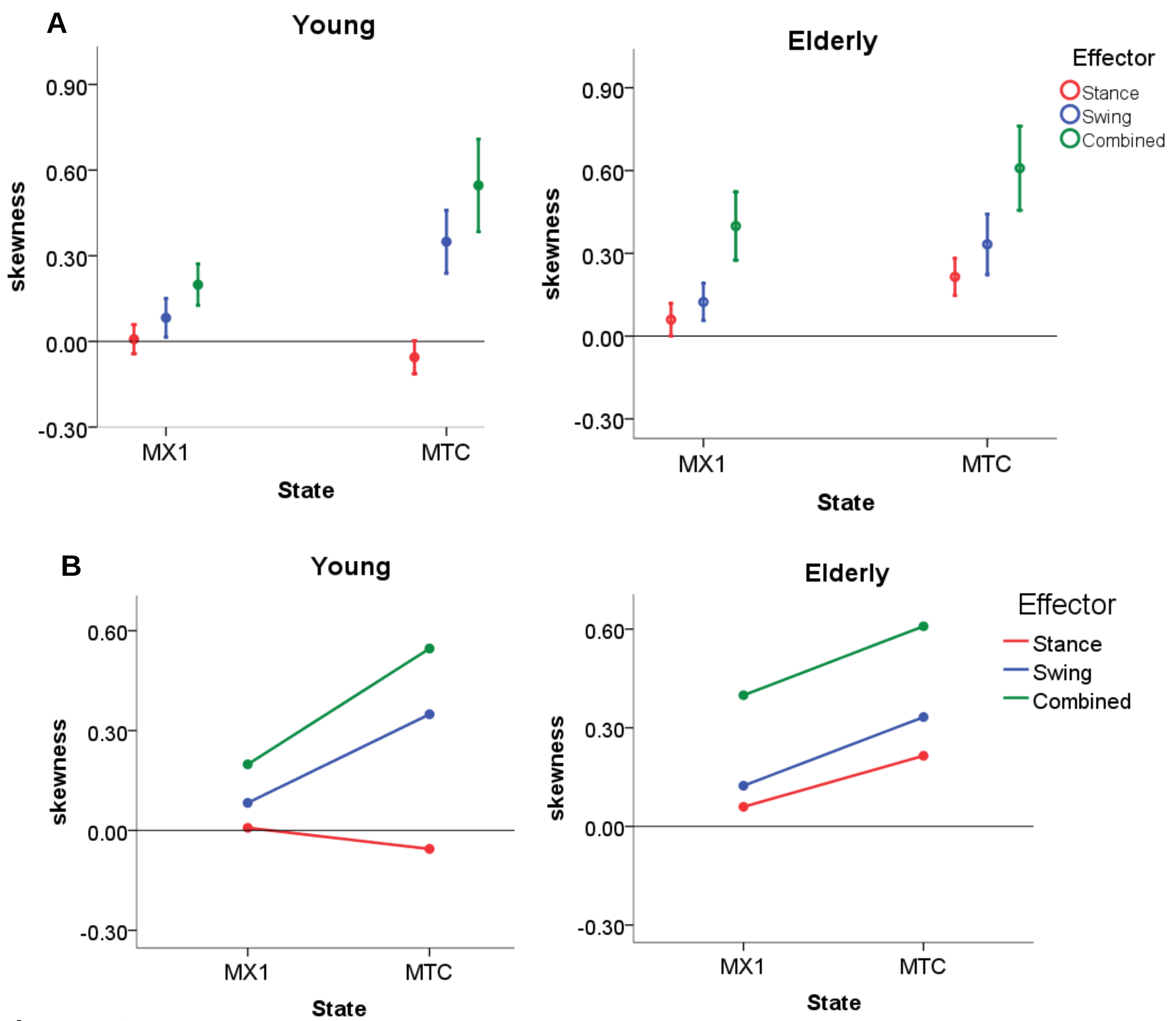


Figure 4.9.4.1

Group mean error-bars (95% confidence intervals) for vertical dimension of effector skewness between MX1 and MTC. Case#55 has been excluded from the elderly group. The data from both limbs has been pooled (i.e. $n_Y = 56$; $n_E = 54$). Illustration of effector and state interactions for the young and elderly.

See text for significant differences at four levels of statistical comparisons.

The second level of analysis investigated skewness between the effector within each group (young and elderly) and state (MX1 and MTC). These four comparisons were analysed using one-way ANOVA with Bonferroni's post-hoc comparisons (with significance set at $p < .05$). The results of the one-way ANOVA was significant in all four tests, where $F(2, 161) > 9$, and $p < .001$. The post-hoc comparisons demonstrated the following between effector differences. For the young and elderly group at both MTC and MX1 states, skewness in the stance effector was significantly lower than the combined effector ($p < .001$). When comparing the stance and swing effector skewness at MX1, neither group demonstrated a significant difference in skewness ($p > .1$). At MTC, only the young group demonstrated a significant difference ($p < .001$) between the stance and swing effector. For both groups, when comparing skewness between the combined effector and swing effector, the combined effector skewness is significantly higher than swing effector skewness at both MX1 and MTC ($p < .05$). In the vertical dimension, it appears that the loco-sensorimotor control system is concerned with penalising trajectories of the combined effector more so than the swing effector, further indicating that the combined-effector is a performance variable, whereas the swing-effector is a task variable.

An argument could be made about the vertical state of the stance effector not being a control parameter because the magnitude of skewness is significantly lower than the swing effector. This is certainly the case in both groups. It is quite likely that the stance effector has a symmetric policy, suggesting that penalties are applied equally for both low and high variations. From Figure 4.9.2.2 there is information relative to the young group that indicates a significantly negative skewness in the stance effector prior to MTC (i.e. 95% confidence intervals are significantly below 0 ($p<.05$). In contrast, the results here have shown the elderly group has a significant increase in skewness between MX1 and MTC for the vertical state of the stance effector. At this stage, results don't indicate whether the increase in elderly stance effector skewness between MX1 and MTC is an outcome of satisfying the toe-clearance task, or if it is reflecting a posture control task.

Figure 4.9.4.1B illustrates that effector variance changes differently from MX1 to MTC when comparing between the young and elderly. After accounting for violations of sphericity (Mauchley's test was significant) using Huynh-Feldt adjustment, there was an interaction effect for effector × state × group ($F(1.77, 191.03) = 6.65$, $p < .005$, $\eta^2 = .058$). Therefore, further exploring effector × state (two-way) interactions was not performed. The planned contrast between the combined effector skewness and the stance effector skewness revealed a significant interaction effect with state and age ($F(1, 110) = 10.32$, $p < .005$, $\eta^2 = .09$). These results support graphical observations when comparing between the young and elderly in Figure 4.9.4.1B. From MX1 to MTC, the combined effector skewness in the young group increases, compared to the decrease of the stance effector skewness. When comparing this to the behaviour of the elderly, this is significantly different; from MX1 to MTC, the elderly demonstrate an increase of positive skewness in the stance effector when compared to the combined effector skewness. This points to an interesting finding in the young group (and between groups), such that a change in positive skewness of the young group combined effector, between MX1 and MTC, occurs irrespective of the stance effector, which makes no change in positive skewness.

4.10 Kinematics of the segment elements within the effector systems

This section investigates the position and orientation of the segment elements comprising the effector systems about the swing phase states. Segment trajectory curves will be plotted across the swing phase regions of interest. As per the previous sections, curves will be described as 95% confidence intervals about the group mean, and between-subject group differences will be highlighted by a 90% confidence interval (the standard error in the mean difference).

4.10.1 Segment kinematics of the stance effector

There were between group differences found within the stance effector segment angles at various states of the swing phase. The largest between group differences were found for the pelvis angles. Figures 4.10.1.1 shows the sagittal plane segment angles of the pelvis (A), stance thigh (B) and stance shank (C) are significantly different between groups ($p < .1$). Generally, the sagittal plane stance effector segment angles of the elderly are further away from a vertical alignment compared to the young group. Figure 4.10.1.2 shows the frontal plane segment angles of the pelvis (A), stance thigh (B) and stance shank (C) are significantly different between groups during the initial swing. Compared to the young group, pelvic obliquity in the elderly group is significantly closer to a horizontal alignment during initial swing. In the transverse plane (Figure 4.10.1.3) the range of rotational motion of the pelvis in the elderly is smaller but not significantly different than the young. The pelvis of the elderly is displaying a more constant perpendicular orientation to the anterior-posterior axis; although, this is not significantly different when making between-group comparisons within the dominant or non-dominant limb at any point along the cycle ($p > .1$). When the data in the dominant and non-dominant limbs are combined together, treating the dominant and non-dominant cases as independent, there is a significant difference between states MX1-12% to MX1+7% ($t(110) > 1.67$, $p < .1$). Additionally, within the dominant and non-dominant limbs, the elderly show significantly greater ($p < .1$) thigh internal rotation of the stance leg between MX1-12% to MTC-6% (Figure 4.10.1.3 B).

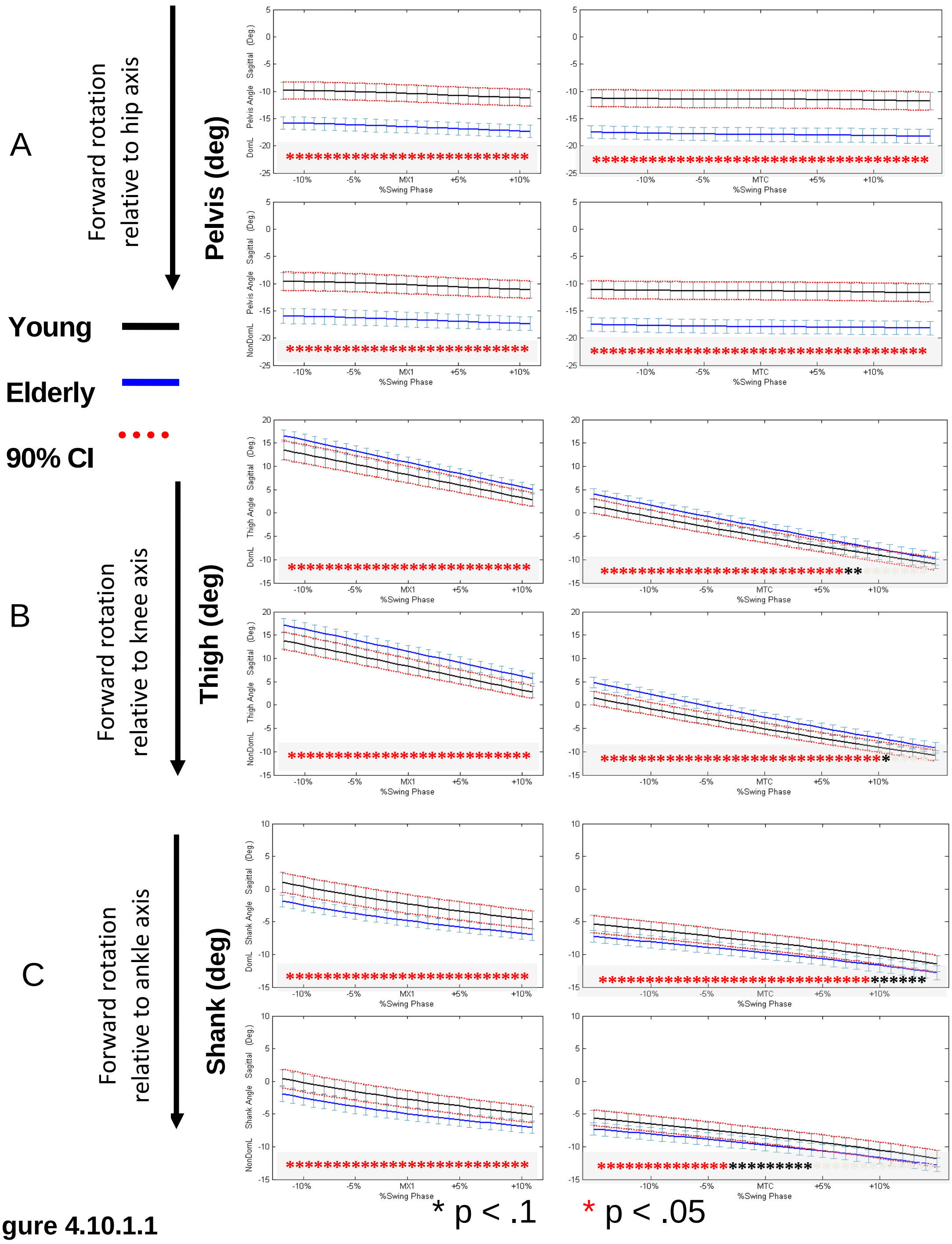


Figure 4.10.1.1

The sagittal plane segment angles of the stance effector vector (ordinate units are degrees, positive is clockwise rotation about vertical reference axis at the distal segment, NB: distal pelvis is considered to be the iliac crest). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

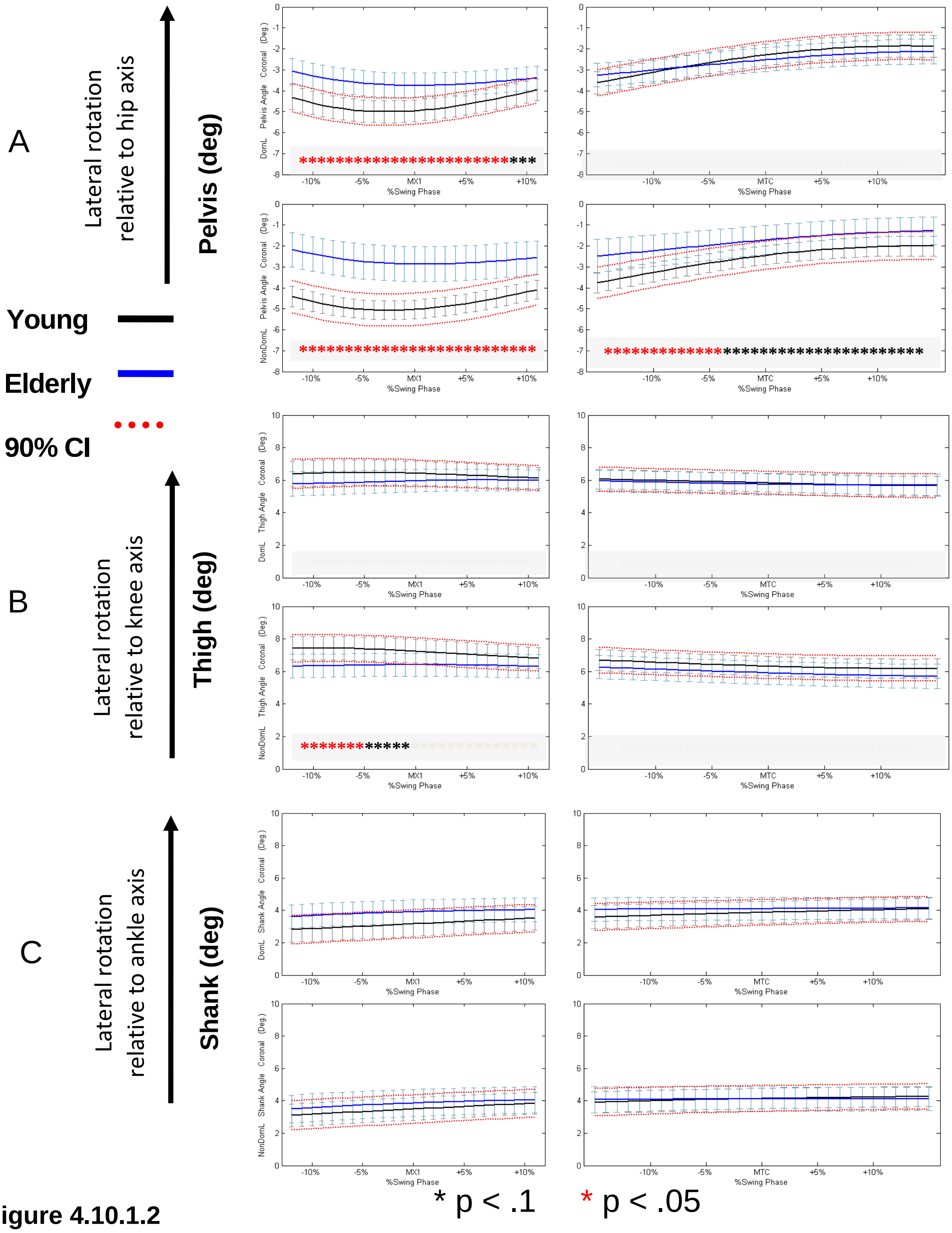


Figure 4.10.1.2

The frontal plane angular position of the stance effector segments (ordinate units are degrees). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

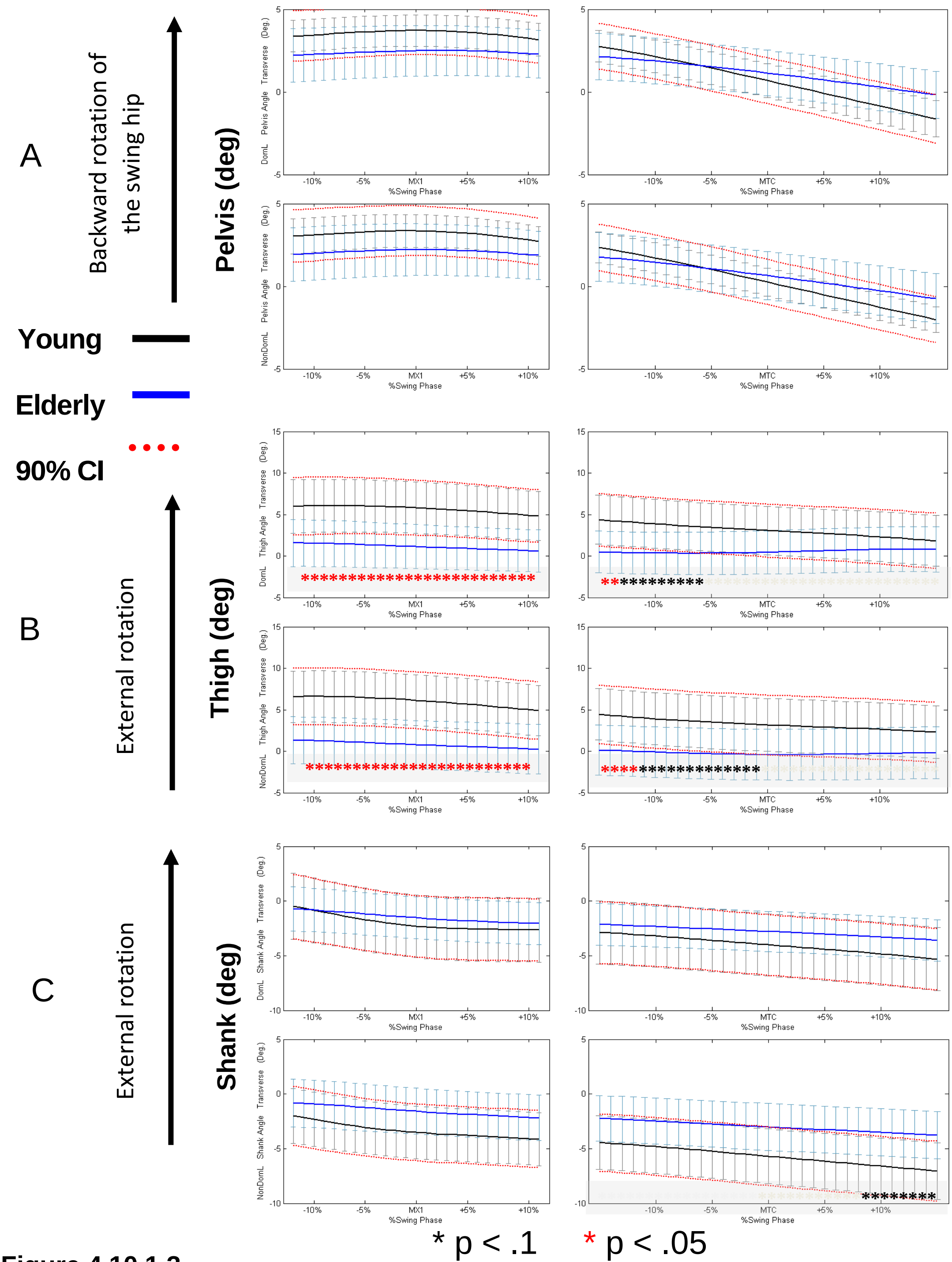


Figure 4.10.1.3

The transverse plane angular position of the stance effector vector (ordinate units are degrees). The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.10.2 Segment kinematics of the swing effector

Figure 4.10.2.1 shows the sagittal plane swing effector segment angles of the foot, shank and thigh. Between group comparisons of the curves indicate the elderly foot doesn’t approach the same relative vertical orientation as the young group during initial swing ($p<.1$). During the mid-swing region, the elderly shank and thigh angles are relatively closer to a vertical alignment compared to the young group ($p<.1$). Figure 4.10.2.2 shows a general tendency of the elderly group to favour a relatively closer vertical alignment of the swing effector segments. Figure 4.10.2.3 shows the transverse plane graphs also indicate that the elderly are relatively closer to a neutral alignment compared to the young, particularly with respect to the shank.

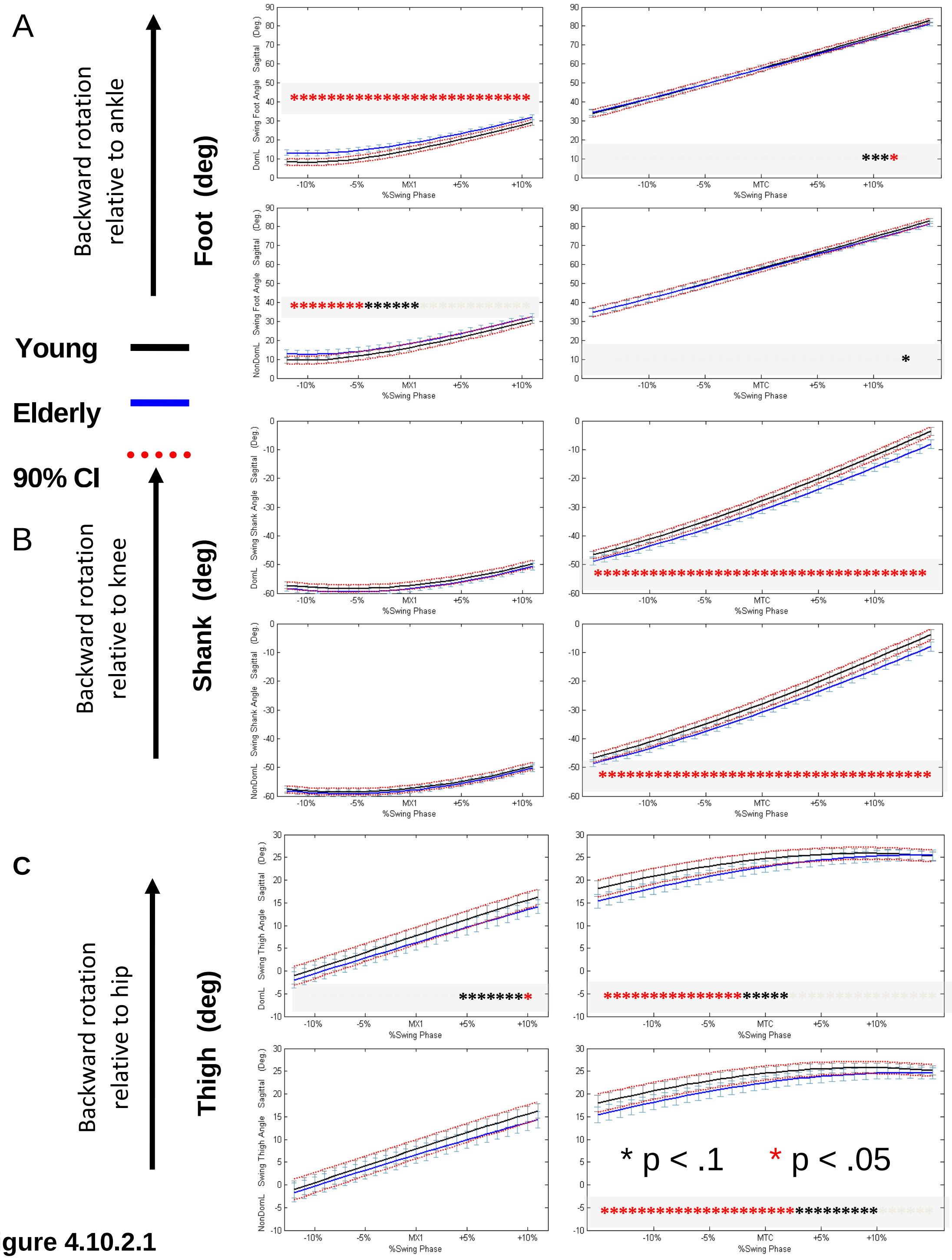


Figure 4.10.2.1

The sagittal plane segment angles of the swing effector vector (ordinate units are degrees, positive is clockwise rotation about vertical reference axis at the distal segment, NB: distal pelvis is considered to be the iliac crest). The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

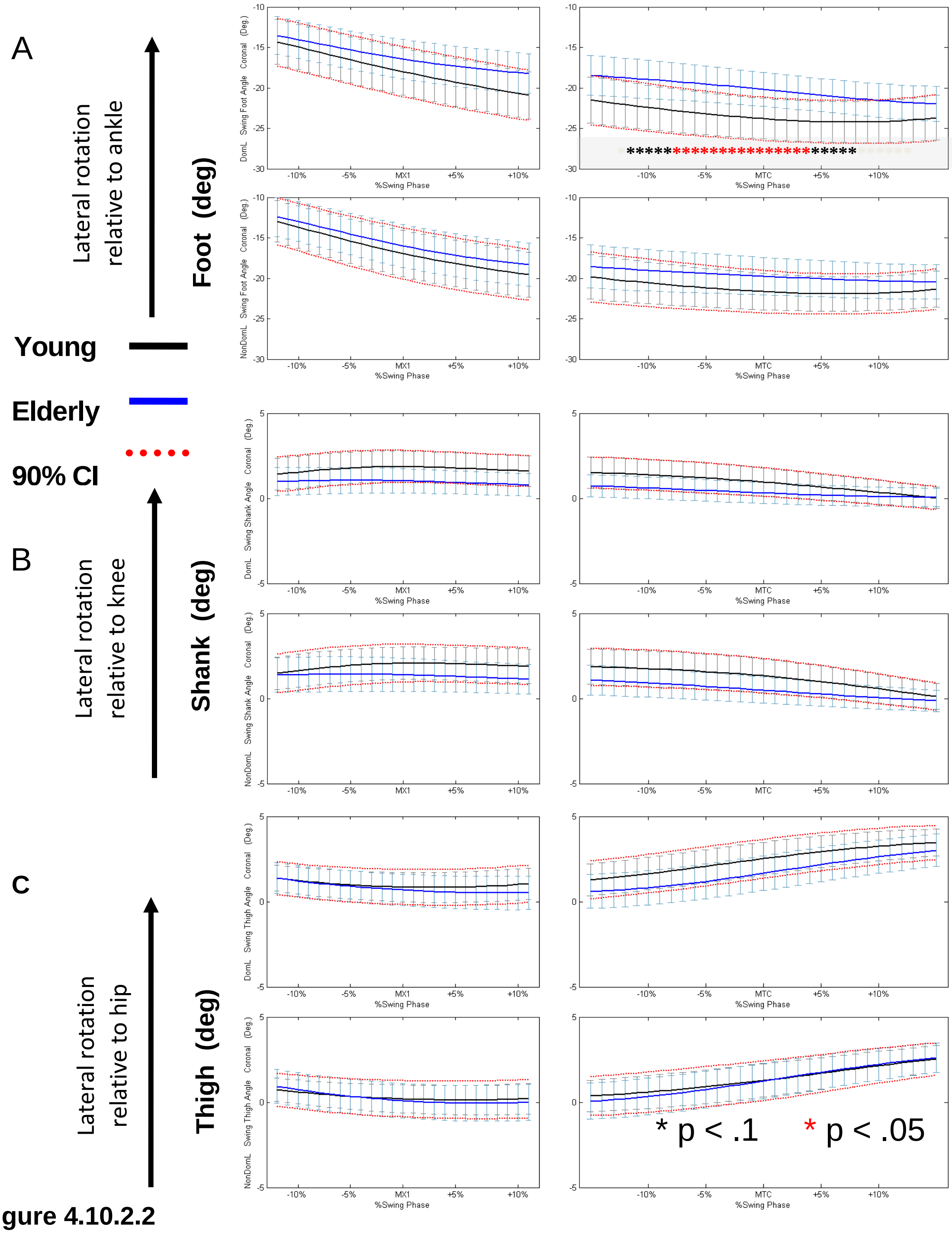


Figure 4.10.2.2

The frontal plane segment angles of the stance effector vector (ordinate units are degrees, positive is clockwise rotation about vertical reference axis at the distal segment. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

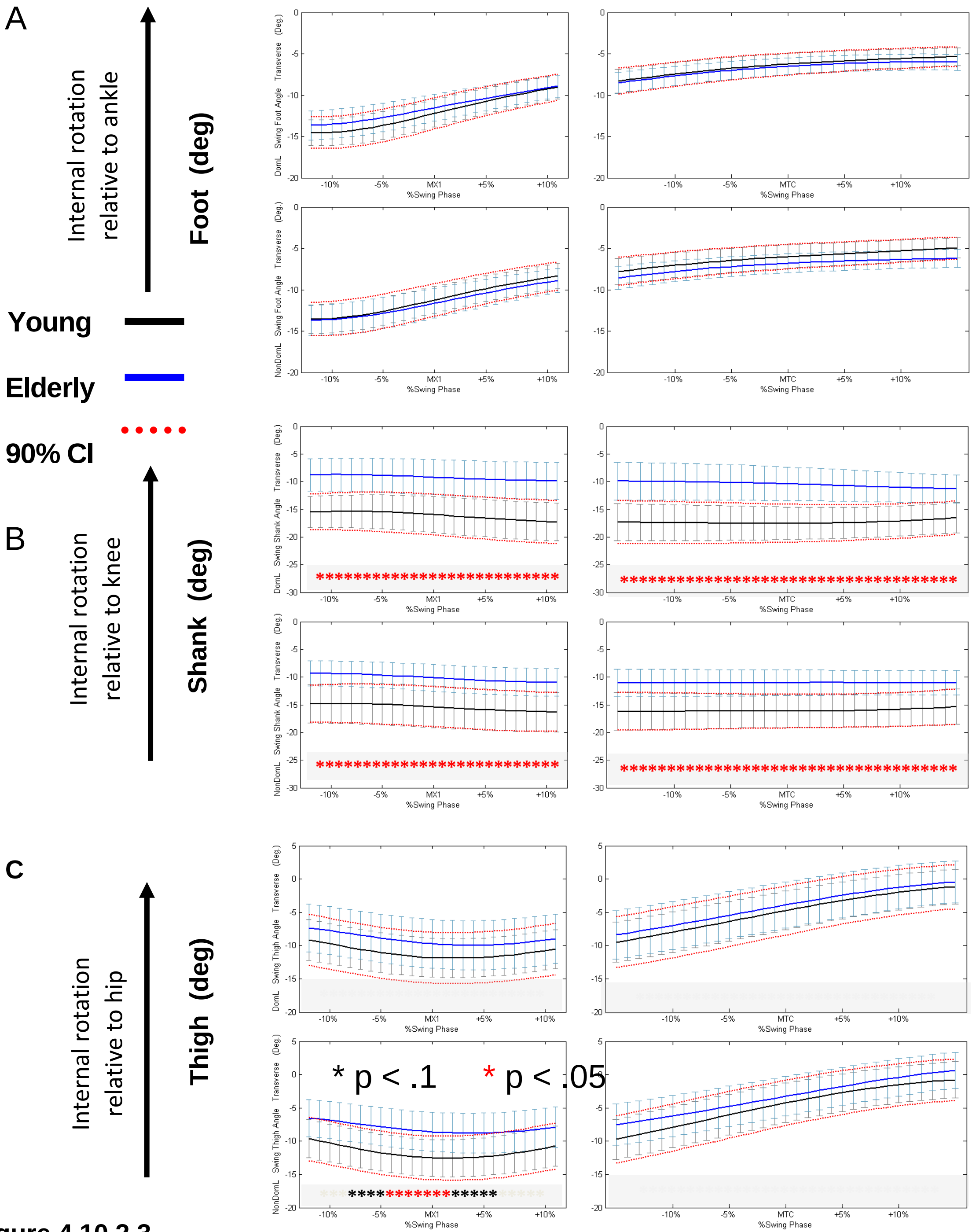


Figure 4.10.2.3

The transverse plane segment angles of the stance effector vector (ordinate units are degrees, positive is clockwise rotation about vertical reference axis at the distal segment. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

4.10.3 Segment kinematics summary

Figure 4.10.3.1 illustrates a summary of the sagittal plane segment configurations at MTC when comparing between the young and elderly. The elderly stance effector adopts more of a 'crouch' gait posture in the sagittal plane. The thigh and the shank orientation at MTC cause the elderly swing effector vector to have a greater posterior component. When contrasting the alignment between young and elderly swing effector segments, it appears that the elderly have forward rotated the effector system approximately 2-3 degrees about the swing hip position; the thigh is more vertically aligned and the shank segment has more forward rotation. The configuration of the thigh and shank in the swing effector of the elderly does not cause a statistically different foot segment to the young group, and both groups display an equal mean angle of 60 degrees. This would require the elderly to produce a higher ankle dorsi-flexion angle compared to the young group.

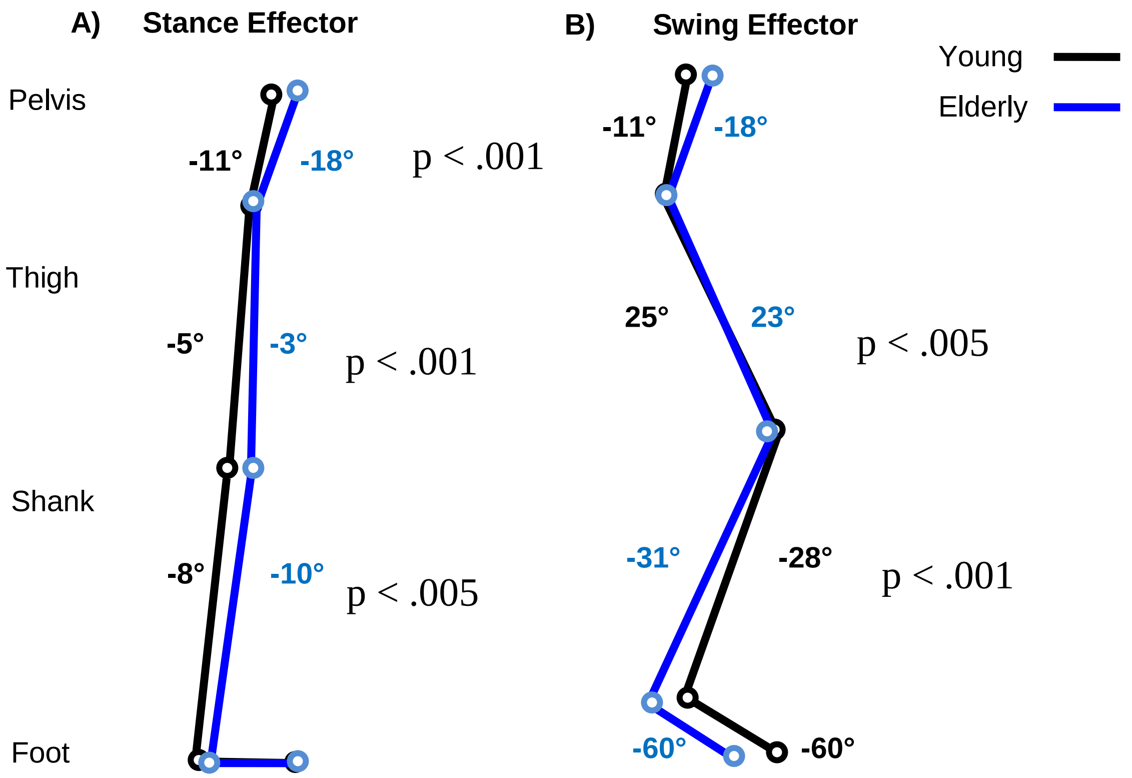


Figure 4.10.3.1

Illustration of the segment-chain in the lower limb stance (A) and swing (B) effector systems, showing between-group comparisons in the sagittal plane. The p values are computed from considering the dominant and non-dominant limbs as independent cases and combining into one group (i.e. n = 56). Positive angles represent backward rotation of the segment with respect to aligning with a vertical axis at proximal segment joint.

4.11 Chapter 4 results summary

Below are the results related to the null hypothesis listed at the beginning of the chapter.

4.11.1 Time-distance

- Null hypothesis 4.1.1 There will not be a significant difference in time-distance parameters (DV) of the gait cycle between age groups (IV).
    - For average step-time between the groups, the null hypothesis was rejected. The elderly walked with significantly lower step time within both dominant and non-dominant limb comparisons.
    - For step length and step width variability (SD) between the groups, the null hypothesis was rejected. The elderly had significantly larger variability within both dominant and non-dominant limb comparisons.
    - For step length skewness, the null hypothesis was rejected. The elderly demonstrated significantly larger negative skewness when comparing within the dominant limb.

4.11.2 Average configuration and motion of effector trajectories

- Null hypothesis 4.2.1 There will not be a significant difference ($p>.1$) in segment angles (DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
    - Figure 4.10.3.1 summarises the results.
    - The null hypothesis was rejected for the sagittal plane angles of the pelvis, thigh and shank during both stance and swing phases. When comparing segments of the stance effector at MTC, the elderly have significantly greater forward (anterior) pelvic tilt, forward tilt of the shank, and backward tilt of the thigh. When comparing

segments of the swing effector at MTC, the elderly have significantly less backward tilt of the thigh and significantly more forward tilt of the shank.

- The null hypothesis was rejected for the thigh segment of the stance effector at MX1. The elderly have significantly higher internal rotation of the thigh.
- The null hypothesis was rejected for the pelvis segment at MX1. The elderly have significantly less pelvic obliquity (i.e. sagittal axis has greater alignment with horizontal).

♦ Null hypothesis 4.2.2 There will not be a significant difference ($p>.1$) in effector configuration state (DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.

- The null hypothesis was accepted when comparing the state of the combined-effector vertical task. However, when combining limb data into one group the null hypothesis was rejected for MX1. The elderly demonstrate a significantly lower vertical state when averaged across limbs.
- The null hypothesis was accepted when comparing groups on the state of the stance- and swing- effector vertical task within limbs. However, when combining limb data into one group the null hypothesis was rejected for states at MX1 and MTC. The elderly have a significantly 'longer' vertical displacement of stance effector at both MX1 and MTC. The elderly have a significantly 'longer' vertical displacement of swing effector at MX1.
- The null hypothesis was rejected when comparing groups within limbs on all effectors configuration state for the anterior-posterior task during state-relevant times of the MTC region. The elderly group have significantly larger posterior configuration for all three effectors.

- The null hypothesis was rejected when comparing groups within limbs on the stance-effector configuration state for the medio-lateral task during state-relevant times of the MX1 region. The elderly group have significantly larger medial displacement at MX1 and MTC for non-dominant limb ($p<.05$, and $p<.1$, respectively), and for dominant limb at MX1 ($p<.1$).

- ♦ Null hypothesis 4.2.3 There will not be a significantly different mean trajectory from MX1 to MTC between the young and elderly groups for the vertical dimensions of the three effectors.
  - Figure 4.6.4.1 illustrates the configuration differences between effector, states and ‘group’ (when averaging across limbs and when effector configurations were normalised to subject limb length).
  - The null hypothesis was rejected for the vertical shape of the combined effector trajectory from MX1 to MTC. There was a significant interaction effect when comparing between the groups on the combined-effector states compared between MX1 and MTC. The elderly have a significantly ‘flatter’ trajectory approach for the combined-effector from MX1 to MTC.
  - The null hypothesis for mean trajectory between MX1 and MTC was rejected. The elderly demonstrated a significantly ‘longer’ stance effector and a significantly ‘longer’ swing effector.

- ♦ Null hypothesis 4.2.4 There will not be a significant difference in mean effector velocity (scaled to limb-lengths/sec, DV) between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
  - The null hypothesis was rejected for the vertical velocity of the combined effector trajectory for state-relevant times following MTC. The elderly have significantly reduced vertical velocity when comparing within the dominant limb ($p<05$), and non-dominant limb ($p<.1$).

- The null hypothesis was rejected for the vertical velocity of the stance effector trajectory for state-relevant times prior to MX1-6% and following MTC+10%. The elderly have significantly reduced vertical velocity when comparing within the dominant limb ($p<.1$), and non-dominant limb ($p<.05$).
- The null hypothesis was rejected for the medio-lateral velocity for several motion states of all three effector trajectories. First, the medio-lateral velocity of the combined effector of the elderly has significantly higher lateral velocity at state-relevant times prior to MX1, when comparing within the non-dominant limb. For the dominant limb, the elderly have significantly lower lateral velocity during state-relevant times between MX1 and MTC. Second, the elderly have a significantly higher lateral velocity prior to MX1 for the swing effector when comparing within the non-dominant limb. Third, the elderly have significantly greater medial velocity (relative to stance foot) of the stance-effector during state-relevant times between MX1+5% and MX1+10%.

4.11.3 Variance of effector trajectories

- Null hypothesis 4.3.1 There will not be a significant difference ($p>.1$) in standard deviation of the effector trajectories (DV) when comparing between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.
  - The null hypothesis was rejected when comparing young and elderly for effector states in the vertical dimension. The elderly have significantly higher variance in all three effectors, particularly for the non-dominant limb comparisons.
  - The null hypothesis was rejected when comparing variability of effector states in the anterior-posterior dimension. The elderly have significantly higher variability in the stance effector

throughout the swing-cycle. For the non-dominant limb of the combined-effector and swing-effector, the elderly have significantly higher variability during the MX1 region.

- The null-hypothesis was rejected for when comparing variability of effector states in the medio-lateral dimension. The elderly display significantly less variability for the swing effector during MTC when comparing within the dominant limb. The elderly display significantly higher variability for the stance effector throughout the swing-cycle. The elderly have significantly larger variability in the combined-effector states during the initial MX1 region.

The following hypotheses are summarised graphically in Figure 4.8.4.1.

- Null hypothesis 4.3.2 Effector variance (DV) will not be significantly different (p>.05) when comparing between the three effectors (IV-1), and when comparing within swing phase states MX1 and MTC (IV-2).
  - The null hypothesis was rejected for comparisons between the swing and combined effector at MX1 and MTC. The swing effector variance is significantly higher for both young and older adults.
  - The null hypothesis was rejected for the stance and swing effector variance comparisons at MTC. The stance effector is significantly less variable at MTC for both groups.
  - The null hypothesis was rejected for comparisons between the stance and combined effector. The stance effector variance was significantly lower at MTC for both groups.
- Null hypothesis 4.3.3 There will be no significant difference between young and older adults (IV-1) in the way that effector variance (DV) changes in the vertical dimension between the effectors (IV-2) when compared between states MX1 to MTC (IV-3; repeated measures).

- The null hypothesis was rejected. The elderly group behaves significantly different to the young group, for variance of the elderly combined effector (when contrasted with the stance and swing effector variance) does not decrease between MX1 and MTC in the same way as the young group combined effector variance.

♦ Null hypothesis 4.3.4 Effector variance (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures) and within age groups (IV-3).

- The null hypothesis was rejected for the young group, there is significantly less effector variance in the vertical dimension at MTC when compared to MX1.
- For the elderly group, the null hypothesis was rejected for the stance and combined effector variances. There is significantly less effector variance in the vertical dimension at MTC when compared to MX1. The null hypothesis was accepted for the elderly group.

4.11.4 Skewness of effector trajectories:

♦ Null hypothesis 4.4.1 There will not be a significant difference in skewness of the effector trajectories (DV) when comparing between age groups (IV) at each state-relevant time-slice of sub-phase regions MX1 and MTC.

- The null hypothesis was rejected for between-group comparisons of the vertical dimension. The elderly have significantly higher skewness in the combined effector during MX1. The elderly have significantly higher skewness for the stance effector during MTC. The elderly have significantly lower skewness of the swing effector at MTC+5% in the dominant limb.
- The null hypothesis was rejected for skewness in effector states of the anterior-posterior dimension for the following effectors. The elderly demonstrate significantly higher skewness for all the

combined-effector and stance-effector during all state-relevant time slices of the swing-cycle. For the swing effector, the elderly also displayed significantly higher skewness at the MX1 region, and for the MTC region of the dominant limb.

  - The null hypothesis was rejected for skewness in effector states of the medio-lateral dimension for the following effectors. The elderly have significantly higher skewness at MTC in the dominant limb for both the combined effector and the swing effector. The null hypothesis was accepted for the stance effector.

The following hypotheses are summarised graphically in Figure 4.9.4.1.

- Null hypothesis 4.4.2 Effector skewness (DV) will not be significantly different between the three effectors (IV-1) when comparing within MX1 and MTC states (IV-2).
  - The null hypothesis was rejected when comparing between the combined and swing effectors at MX1 and MTC, in both groups. The combined effector has significantly higher positive skewness. For the young group, the swing effector has significantly higher positive skewness compare to the stance effector skewness.
- Null hypothesis 4.4.3 Effector skewness (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures) and within age groups (IV-3).
  - The null hypothesis was rejected for the swing and combined effectors for both the young and the elderly groups, and for the stance effector of the elderly group. There was a significant increase in skewness between MX1 and MTC.
- Null hypothesis 4.4.4 There will be no significant difference between young and older adults (IV-1) in the way that the skewness (DV) changes in the

vertical dimension between effectors (IV-2) from states MX1 to MTC (IV-3; repeated measures).

- The null hypothesis was rejected. The effectors behave differently between the groups, when comparing between the MX1 and MTC states. Skewness in the stance effector does not increase relative to the combined effector for the young group, and this is significantly different compared to the elderly group.

Describing kinematic states of the effector trajectories from correlation statistics

5.1 Background

Chapter 4 reported the distribution statistics of the segment angle and effector trajectory states that capture effector behaviour at a select region of the swing-cycle. This chapter further investigates the behaviour of the effector trajectories. An effector trajectory can be described as a time-course of desired states that are ‘managed’ by the loco-sensorimotor controller. The state of the effector trajectory can encompass any physical property describing the effector, however, in this chapter only the effector displacement or configuration states in three-dimensional space will be examined. From multiple effector trajectories cycling through three-dimensional space, the behaviour of an effector can be explored through its serial ordering of fluctuating states at a meaningful event in time on the effector trajectory cycle. The events of the swing-cycle that are proposed to be meaningful to the toe-clearance task are at task-relevant states (MX1 and MTC) and state-relevant times of the swing cycle (e.g. MX1+6%, MTC-6%). The state-relevant times represent equally spaced time-slice steps (time-normalised to each swing cycle) that neighbour the task-relevant states MX1 and MTC. These were the same events on the trajectory that were applied in chapter 4. Chapter 4 described effector behaviour from distribution statistics of effector states at state-relevant time-slices. This chapter provides a different insight in the way that states along the effector trajectories are controlled by quantifying effector persistence using Detrended Fluctuation Analysis (DFA). Here, a scaling exponent, 'α', will define the persistence of the trajectory states at state-relevant time-slices. The scaling exponent derived from Detrended Fluctuation Analysis (DFA) has traditionally been associated with long-term correlations, but due to its inability to detect for certain the long-term dependence of a time-series, it is recommended that the scaling exponent of the DFA be associated with a more conservative measure (Wagenmakers, Farrell and Ratcliff, 2005; Delignieres and Torre, 2009). While original studies refereed to the scaling exponent as a self-similarity parameter because it correlates fluctuations across multiple time scales (Hausdorff et al., 1997b), in this thesis it suits the context to represent it as a statistic of 'persistence-likelihood' in a time-series, and is a term adopted by other studies (Dingwell and Cusumano, 2010). From this point forward, the DFA scaling exponent 'α' will be referred to as the 'persistence-likelihood' parameter.

Persistence-likelihood in a time-series is when serial fluctuations from cycle-to-cycle have a likelihood that values of one cycle will follow a directional trend established by a past-history of increasing or decreasing prior values. The term anti-persistence represents a time-series that acts in the alternate manner, whereby values from one cycle respond in an opposite manner to large or small values from the previous cycle. This thesis proposes that persistence-likelihood reflects the control effort made by the 'controller' to search through the embodied loco-sensorimotor workspace to find a desired set of movement solutions. The proposal is based upon a controller searching for a desired movement solution, and determining the acceptable level of effort to search through the workspace to realise this solution. The least 'control' effort made to find a solution will be represented by $\alpha \approx 1.0$, whereas the most 'control' effort to find a solution will be represented by $\alpha \approx 0.5$. This proposal is based upon a theory that there is a low cost for realising 'passive dynamic' solutions and a high cost when 'controlling' the search dynamic for a desired solution (Todorov, 2009).

Increasing levels of persistence-likelihood will represent the low cost, ‘passive’ dynamic solution. This indicates that movements emerge from self-organised, or passive search process through the embodied workspace that is inherent to a loco-sensorimotor system (Scafetta et al., 2009; Dingwell and Cusumano, 2010). A stronger level of persistence-likelihood is when the system can ‘run-free’ by accepting the set of solutions derived by a passive search process (Hausdorff et al., 1996; Ashkenazy et al., 2002; Scafetta et al., 2009; Dingwell and Cusumano, 2010). From this theory, persistence-likelihood, α, will range between 0.5 and 1.0 depending upon the control (‘search’) effort.

5.2 Method

The analysis of persistence in the states of effector trajectories at task-relevant time-slices was applied to the same time-series that were used in Chapter 4, where they provided input for the statistical descriptions of effector trajectory states.

Persistence was measured by the Detrended Fluctuation Analysis (DFA) method as described in section 3.7.1, which yields the persistence-likelihood parameter, 'α'. First, the time-series (of effector states at a common state-relevant time-slice) was integrated, and then fitted without overlapping segment (window) lengths of equal size. The integrated data within each local 'window' was then fitted with a linear 'trend' line. Each window length is represented by an average fluctuation calculated from all local windows. This was calculated by taking the square root of the average fluctuations from each local window. The fluctuations within each window are obtained from the point-to-point differences between the fitted trend-line and the integrated time-series. This is outlined in specific detail in section 3.7.1. The input time-series were the three separate dimensions of the effector displacement states on the effector trajectory. The DFA derived scaling exponents, or the persistence-likelihood parameter, 'α', was averaged across groups, state-times, effectors, dimensions and limbs during statistical analysis. This procedure was adapted from descriptions by Bashan et al. (2008) and ran through custom-written scripts in MATLAB (The Mathworks Inc).

The degree of short-term persistence was also investigated using an auto-correlation function (ACF) of a time-series n and its next time lag, n+1. The ACF with time lag-1 can determine anti-correlations at time-series of effector states when the ACF value is significantly below zero. The procedure for calculating the ACF_{lag-1} is explained in section 3.7.

5.2.1 Hypotheses and statistical design

As per the approach to statistical testing described in the previous section, this current section will undertake three general levels of analysis. First, between-group differences in age will be examined at discrete state-relevant time-slices along the swing-cycle. Second, within-subject effects of effector and state-relevant time-slices will be examined on dependent variable persistence. Lastly, for hypotheses about effector persistence change between age-groups along the swing-cycle a 3 × 4 × 2 (effector × state-relevant time × age) mixed-design was used (with repeated measures on state-relevant time).

Therefore, this chapter will test the null hypotheses outline below, which are linked to the aims described in section 2.5. For each hypothesis, the dependent and independent variables are indicated in parenthesis, (DV) and (IV) respectively. Each null hypothesis has a description of how statistical significance was tested.

5.2.1.1 Hypotheses of persistence in the effector states

- Hypothesis 5.1 There will not be a significant difference in persistence (DV) between age groups (IV) when comparing within effector (stance, swing, combined), dimension (medio-lateral, anterior-posterior, vertical), and state-relevant time (in sub-phase regions MX1 and MTC).
    - The dependent variables are DFA scaling exponents ('α') in components of three-dimensional effector vectors: stance; swing; and combined.
    - Statistical test design is multiple independent-sample t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.
- Hypothesis 5.2 Effector persistence (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures), within age groups (IV-3).
    - The dependent variables are DFA scaling exponents ('α') in the vertical components of three-dimensional effector vectors: stance; swing; and combined.
    - Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.

- Statistical test design is three separate paired t-tests for IV-2 ($p<.05$) within each age-group.

- Hypothesis 5.3 First, effector persistence (DV) will not be significantly different between the three effectors (IV-1) when comparing within swing phase states MX1 and MTC (IV-2). Second, and in association with previous null-hypothesis, there will not be a significant interaction when comparing between effectors (IV-1) on persistence (DV) when averaged across 'state-relevant times' within the swing-phase regions of MX1, MX1-MTC and MTC (IV-4). Third, there will not be significant differences for between-group comparisons of the above two hypotheses.
    - The dependent variables are the DFA persistence-likelihood parameter ('α') of the three effectors (vertical dimension).
    - Within each group, separate one-way ANOVA designs testing for effector (IV-1) differences on persistence within-states MX1 and MTC (IV-2). Within-group simple effects for between-effector differences to be assessed by post-hoc Bonferroni tests (significance set at $p<.05$).
    - A three-way mixed design (ANOVA) (from hypothesis 5.4) investigating a main effect for effector.
    - Collapsing time, interaction effect for effector (3) x age (2) will be investigated to determine age-group differences on between-effector persistence. Planned contrasts (simple effects) between effectors stance-swing and swing-combined will be significant at $p<.05$.

- Hypothesis 5.4 There will be no significant difference between young and older adults (IV-1) in the way that the persistence (DV) changes in the vertical dimension between effectors (IV-2) at consecutive state-relevant times (IV-3; repeated measures).
    - A three-way [effector (3) × state-relevant time (4) × age (2)] mixed-design ANOVA (with repeat measures for 'state-relevant time').

Planned comparisons (simple effects) between effector levels of stance and combined (Level 1), and also swing and combined (Level 2), will be determined for significance at $p<.05$.

- The dependent variable is the DFA persistence parameter ('α') in vertical components of three-dimensional effector vectors: stance; swing; and combined.
- Pooling the between-limb data into one group; i.e. dominant and non-dominant limbs become independent cases.

♦ Hypothesis 5.5 There will be no significant difference between effector systems and their dimensions on 'control policy' statistics of DFA (persistence), SD (standard deviation, variance), and skewness. There will be no significant differences between young and older adults when comparing significant correlations.

- An analysis of significant ($p<.05$) correlation coefficients
- Comparing independent r values was performed by taking the difference of their log normalised r scores and dividing by the combined standard error, e.g. (Field, 2009)

5.2.1.2 Statistical design

For making interpretation easier, group means and figures are based upon the pre-transformed scores, while the transformed scores were used for computing the significance and effect sizes. When Mauchley's test of sphericity was violated, the conservative Greenhouse-Geisser degrees of freedom correction of the F-statistic was applied (Field, 2009). When data distributions were significantly positively skewed, the data was transformed by taking the natural log of the raw score, i.e. $Z_{trans} = Ln(x_i)$, where x_i represents the participant score. When group data distributions were negatively skewed, the data was transformed by taking the square root of the 'raw' scores.

If a significant F statistic was reported in the three-way mixed design (with repeat measures) ANOVA tests, planned contrasts were carried out to determine simple interaction effects of group mean comparisons. Planned contrasts partition the variance in such a way during between-group comparisons, that allows the type-I error level to be maintained. When using multiple post-hoc tests, partitioning the variance for multiple tests causes associated familywise errors. Partitioning the variance in such a way, is therefore likened to planned a priori specific hypothesis testing using one-tailed significance tests, as opposed to the exploratory form of two-tailed tests. The basis for using planned comparisons in this study was due to the nature of the independent variables of 'state-relevant time' and 'effector'. The paired comparisons selected in this study was based simply upon obtaining the most important information about the data. For the independent variable 'state-relevant time', planned comparisons were made only between adjacent time-slices. For example, three comparisons of: MX1-12% and MX1-6%; MX1-6% and MX1; MX1 and MX1+6%. This was because it didn't make sense to compare non-adjacent time-slices as the data across the independent variable 'state-relevant time' can represent a polynomial curve. For the independent variable 'effector', planned comparisons were made between two effectors pairings of stance-swing and swing-combined. The rationale for these contrast pairings was based upon a presumption that the excluded combined-stance effector pairing would add the least insight. While on this basis there is some risk of losing additional inter-effector information, the alternative provides maximum information about the effector pairings that are deemed to hold most information.

5.3 Inspection of the effector time series at MTC

Figure 5.3.1 shows an example of a young participant MTC time-series for the vertical dimension of the three effectors. This sequence of 600 MTC events is approximately 10 minutes long. The distribution statistics of these time series have been investigated in the previous chapter. This chapter is focused on describing the persistence likelihood in the time-series. The figure below represents just one 'time-series' from a set of 56 time-series (i.e. 25 for MX1 region and 31 for MTC region), each representing a state-relevant time-slice along the swing-cycle. [NB: each subject has 56 x time-series per limb; therefore each subject has 112 x time-series from both limbs for each effector variable.] The time series (Figure 5.3.1) exhibits brief periods where the cycle-to-cycle states continue to either increase or decrease across consecutive strides and then this trend reverses for the next period. The fluctuations in the stance effector appear to reverse direction in a negatively correlated way with the stance effector (Figure 5.3.1 B). In contrast, the swing effector and combined effector appear to be positively correlated. These observations are confirmed by the correlation analyses in the next section.

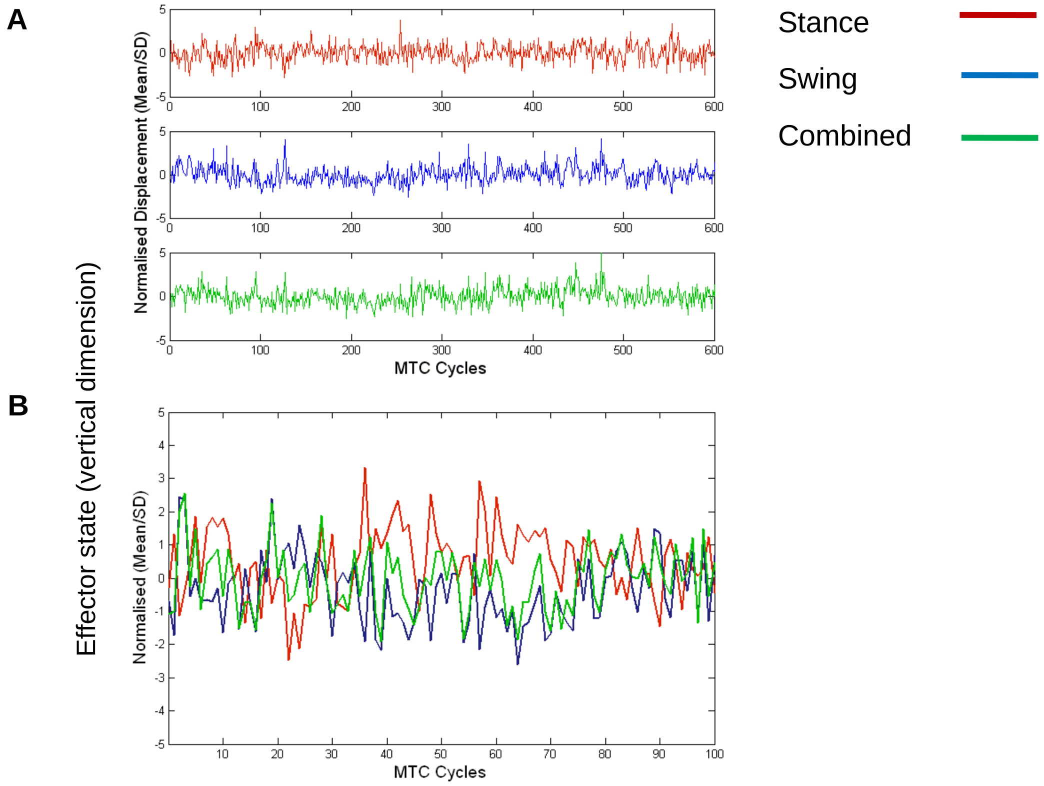


Figure 5.3.1

A) Example plot of the three effectors time series in the vertical dimension for a young subject. The vertical scale represents state values that have been mean-shifted and normalised to average variance. The signal is high frequency, however, a low-frequency drift can be envisaged along the 600 cycles. This low-frequency 'fit' represents periods of persistence in the time-series. In contrast, periods of a relatively constant 'fit' represent limited persistence, or 'anti-' persistence.

B) A sample of 100 cycles of the three effectors for close up inspection of noisy but somewhat persistent behaviour. The stance (red) and swing (blue) effectors appear to persist in relatively opposite directions. In contrast, the combined effector (green) appears to follow the same direction of persistence as the swing effector (blue).

5.4 Qualitative examination of effector state persistence-likelihood

This section illustrates the Detrended Fluctuation Analysis and Auto-Correlation$_{lag-1}$ profiles of the three effector systems. The contrast between Detrended Fluctuation Analysis and Auto-Correlation Function (lag-1) is related to the correlations made at short-range time scales (see section 3.7). The DFA applies a multi-scale correlation procedure at time scales greater than 4 cycles. In contrast, the ACF was employed to observe specific correlations at the shortest time scale, i.e. cycle lag = 1.

5.4.1 Detrended Fluctuation Analysis profiles

Figure 5.4.1.1A-B presents the group average profiles for the three effectors persistence-likelihood parameter, ‘α’, across state-relevant time-slices. A qualitative inspection of Figures 5.4.1.1A-B can provide an initial insight into the upcoming statistical analyses being conducted.

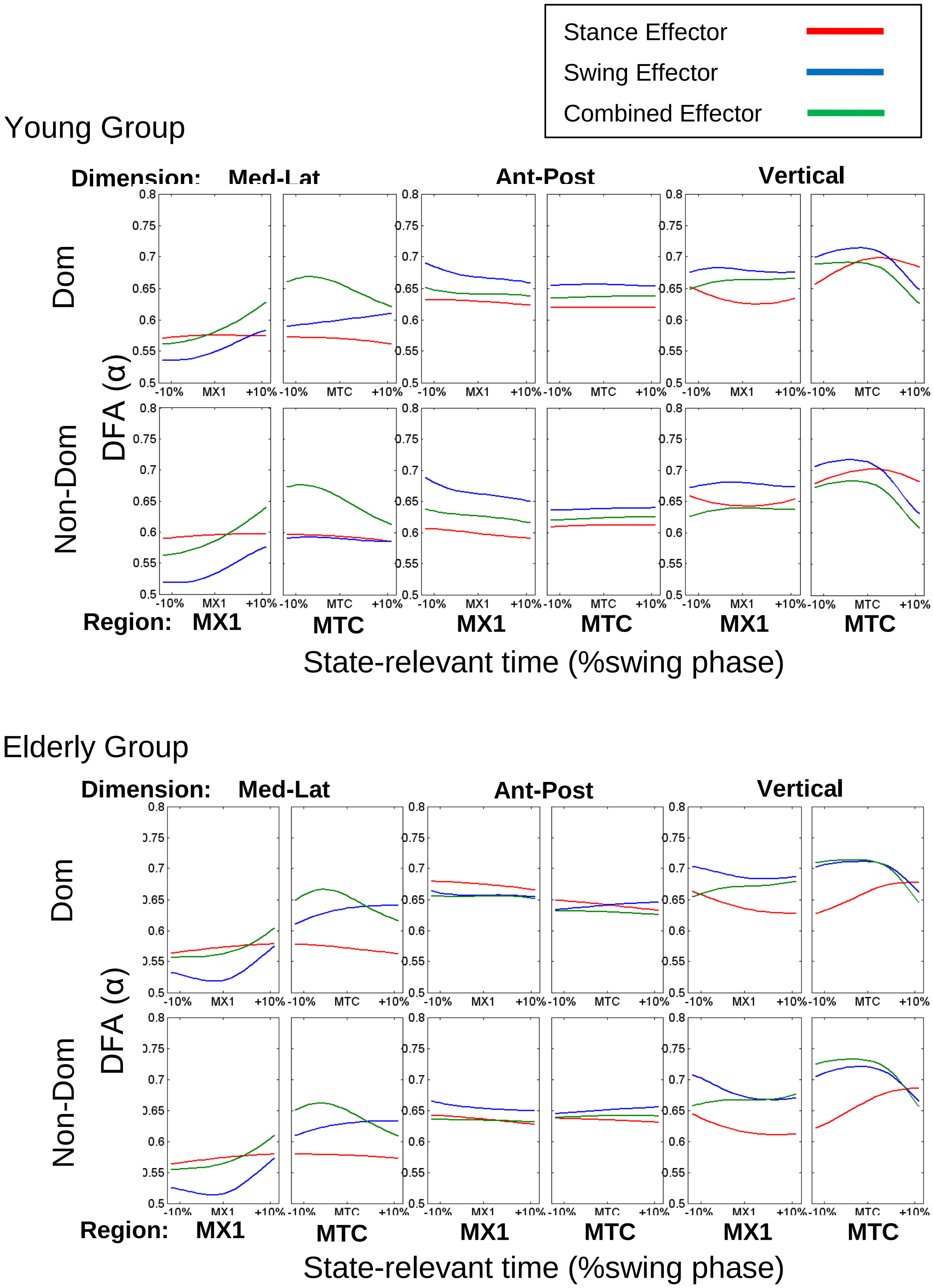


Figure 5.4.1.1

The DFA profiles of the young and older adult groups. The DFA value, α, is the vertical axis. The state-relevant time-slice (normalised to swing-phase time) is the horizontal axis scale. For each effector, the DFA group average is plotted at each time-slice within each sub-phase region (MX1 and MTC).

Figure 5.4.1.1A-B generally shows the profile of persistence-likelihood (DFA values > ~0.6) and where the serial structure becomes less ‘persistent’ (DFA values < ~0.6). A general observation from Figures 5.4.1.1A-B shows that for both age groups, the medio-lateral dimension of the three effector states has the least persistence-likelihood during the MX1 region. Apart from the elderly stance effector, the three-effector states show the highest persistence-likelihood for the vertical dimension during the MTC region. The theory of persistence when related to the general difference between the medio-lateral and vertical dimensions suggests that managing effector trajectory states in the vertical dimension at the MTC region receives less control intervention by the loco-sensorimotor control system compared to managing medio-lateral trajectory states at the MX1 region. It is an interesting finding to observe near-maximum persistence-likelihood in the vertical states of the combined effector trajectories near the MTC state.

When examining the persistence-likelihood in the three effectors across the young and elderly groups for the vertical dimension at the MTC region (i.e. right-most column), the stance effector of the elderly displays lower average persistence relative to the swing and combined effectors. Another observation in these vertical-dimension MTC-region graphs, is that the young group combined effector has reduced persistence-likelihood relative to the swing effector, whereas, this isn’t observed for the elderly group.

With respect to the anterior-posterior dimension, both groups demonstrate a more constant level of persistence-likelihood across regions MX1 and MTC. This contrasts with the wave-like change in persistence-likelihood along the swing-cycle regions observed in the medio-lateral and vertical dimensions.

When making a qualitative comparison between the dominant and non-dominant limbs within both groups in Figure 5.4.1.1, the shapes of the DFA profiles are quite similar. That is, the dominant and non-dominant limbs are more alike within a group, than any similarity between groups.

5.4.2 Auto-Correlation Function (lag-1) profiles

Figure 5.3.2.1 shows that the ACF_1 curves have almost identical profiles with the DFA curves. Graphs show very weak anti-correlations in the medio-lateral dimension for the three effectors, suggesting the presence of a corrective auto-regressive process rather than purely random correlations.

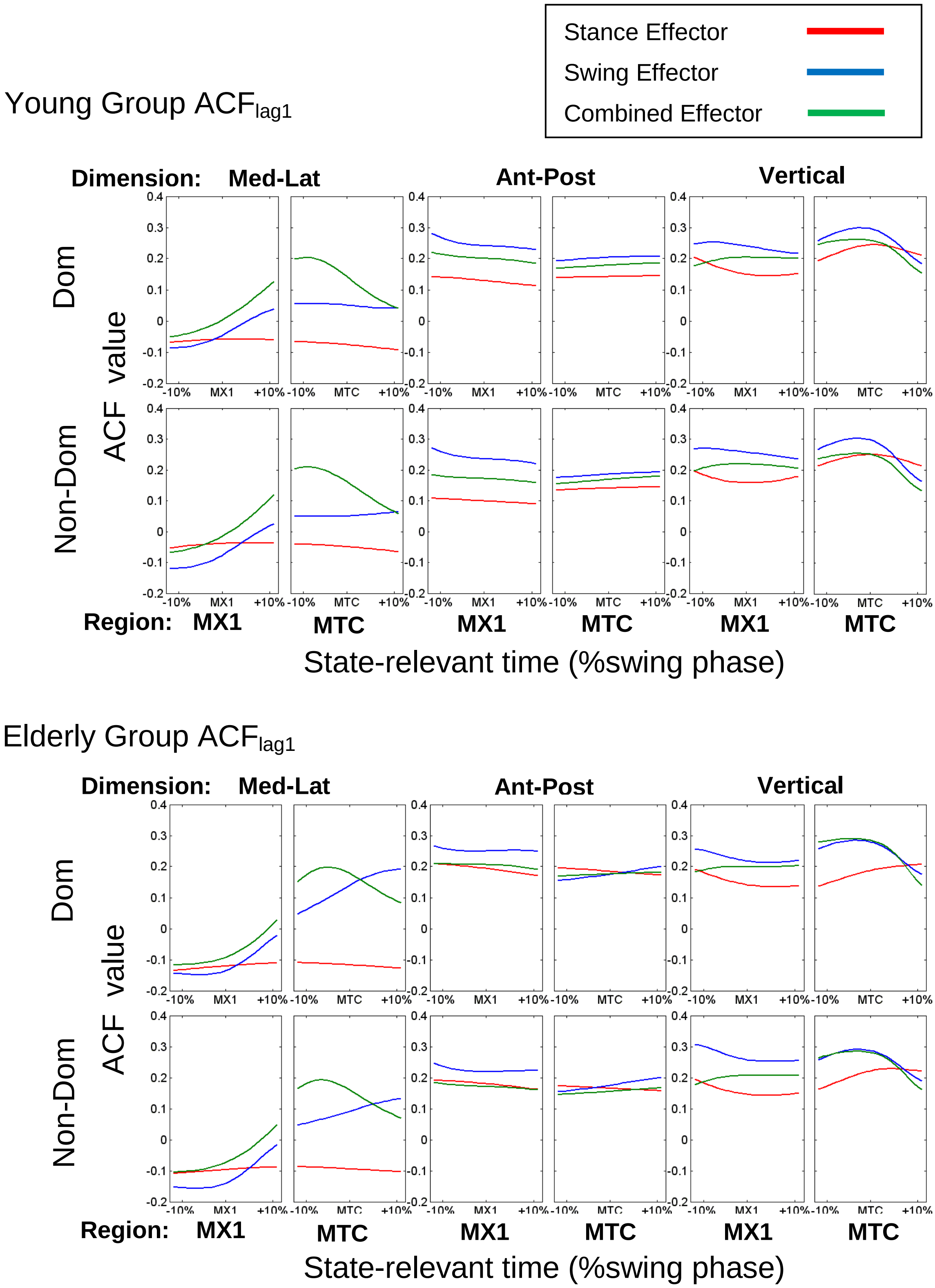


Figure 5.4.2.1

The ACF_{lag1} profiles of the young and older adult groups. The ACF_{lag1} is the vertical axis. The state-relevant time-slice (normalised to swing-phase time) is the horizontal axis scale. For each effector, the DFA group average is plotted at each time-slice within each sub-phase region (MX1 and MTC).

5.4.3 Relationship between the ACF_1 and DFA

The short-time scale analysis of ACF_{lag-1} seems to be capturing the same behaviour of the effector systems as the multi-scaled based DFA procedure. A correlation of DFA and ACF_{lag-1} (vertical dimension of the combined effector) was performed for the elderly and the young separately, where independent variables were collapsed (time, effector type, dimension, limb). This created a very high sample size (N = 3136) to perform the correlation between DFA and ACF_{lag-1}. The Pearson parametric correlation and the Spearman non-parametric correlation both showed significant correlations with different values for each participant group. The significance was expected due to the sample size. Within groups, the Pearson correlation coefficient for the young group had a between DFA-ACF_{lag-1} correlation of $r = .58$ ($p<.001$) and the elderly group had a between DFA-ACF_{lag-1} correlation of $r = .74$ ($p<.001$). Comparing between the two correlations is significant ($p<.05$, two tailed) because of the large sample (N=3136). Therefore, the elderly generally have significantly stronger association between the ACF_{lag-1} and the DFA, compared to the young group.

5.5 Persistence-likelihood of the effector states

The previous section showed that the persistence of the effector systems is dependent upon the time-state of the swing phase. Variations in the DFA across time will be compared between groups in a similar approach to that undertaken in chapter 4.

5.5.1 Vertical dimension

Figure 5.5.1.1 shows the significantly higher DFA in the elderly group compared to the young group for the non-dominant limb of the combined effector during the MTC region ($t(54) = 2.6$, $p=.012$). When comparing between groups for the dominant limb, the difference is not significant ($p = .3$). In contrast, Figure 5.5.1.3 shows the stance effector in the vertical dimension has a lower DFA value in the elderly non-dominant limb ($t(54) = 1.9$, $p=.06$) and a similar trend for the dominant limb ($t(54) = 1.63$, $p = .1$).

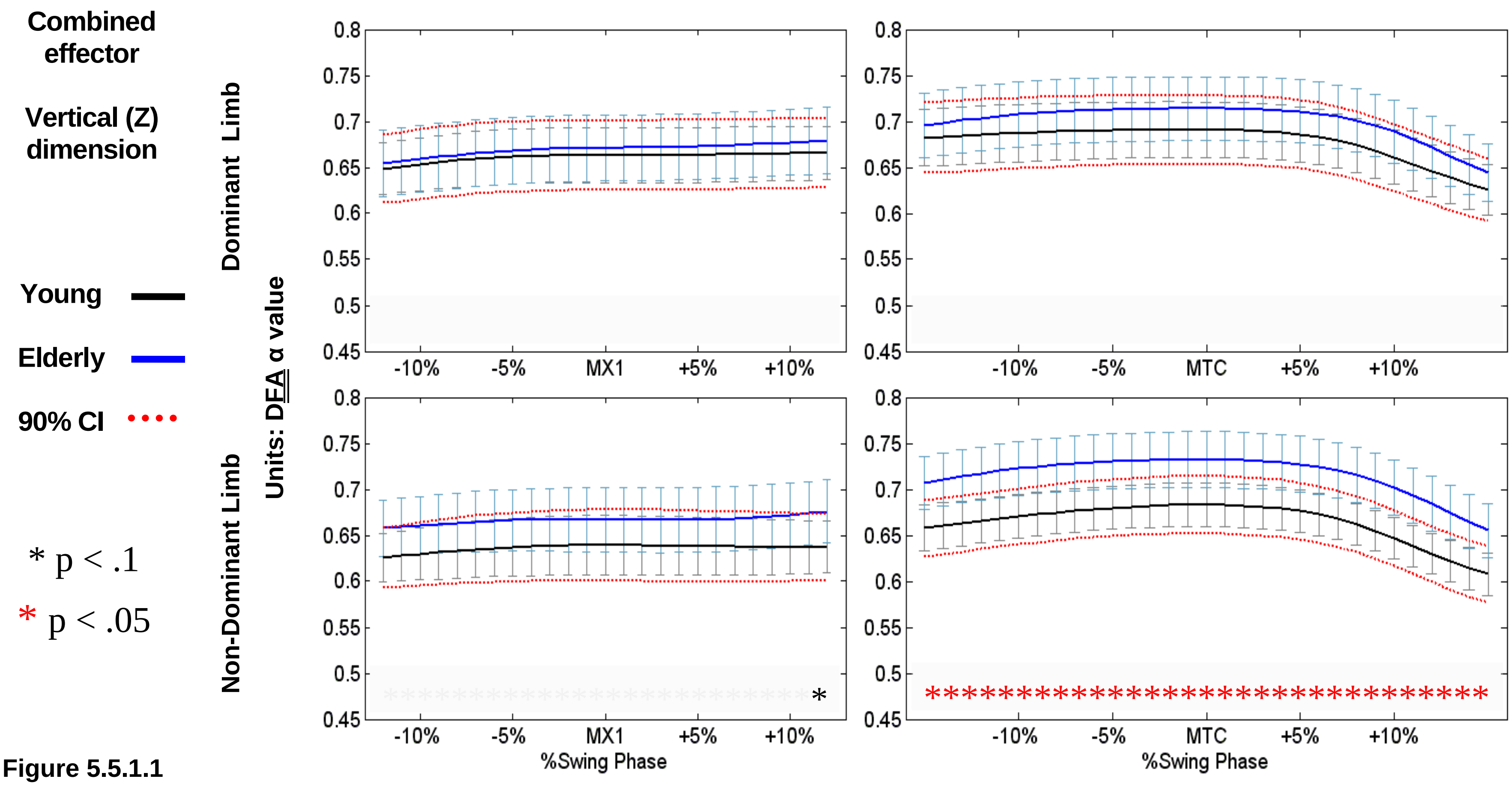


Figure 5.5.1.1

DFA profiles of the young and elderly groups for the combined effector systems in the vertical dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

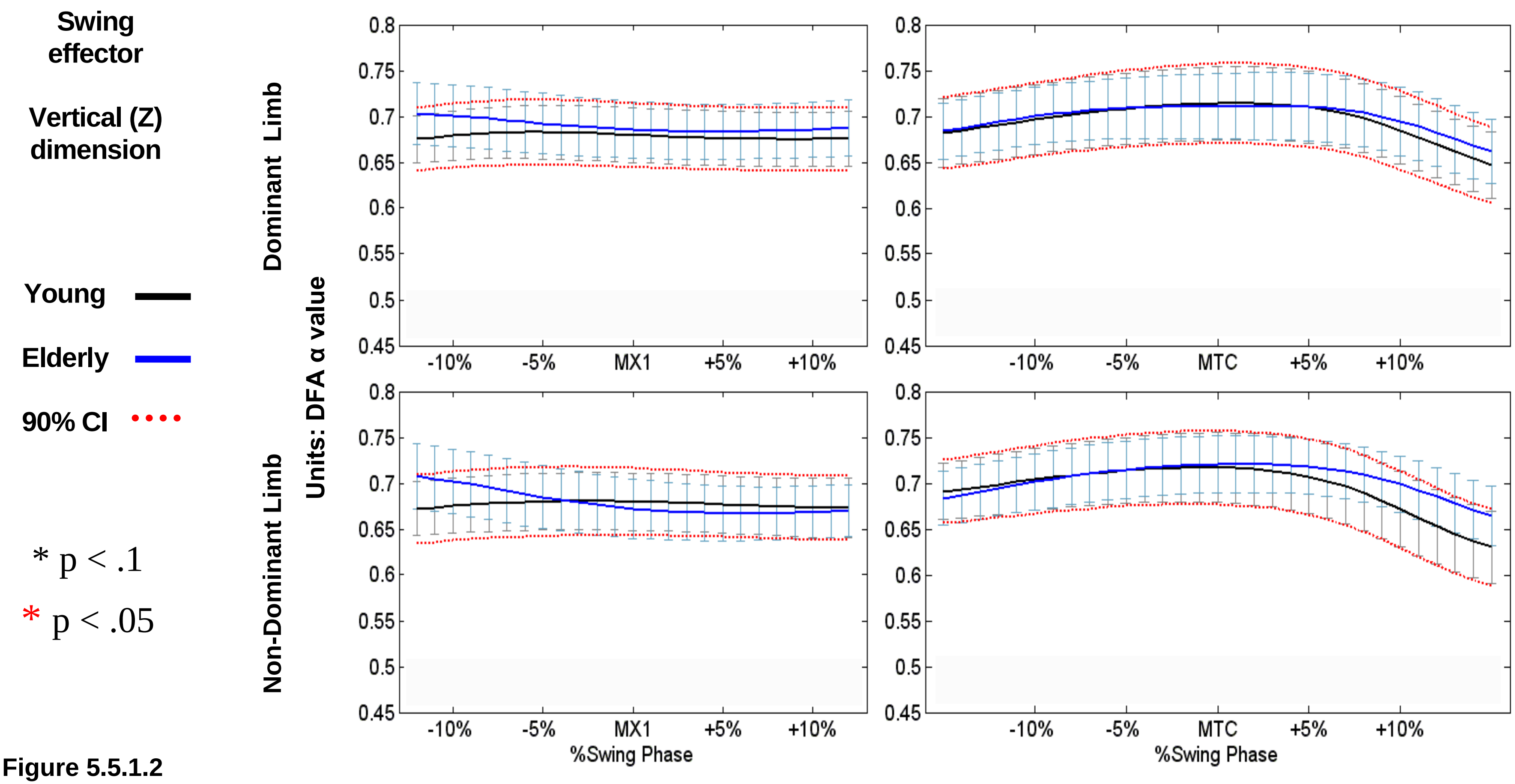


Figure 5.5.1.2

DFA profiles of the young and elderly groups for the swing effector in the vertical dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

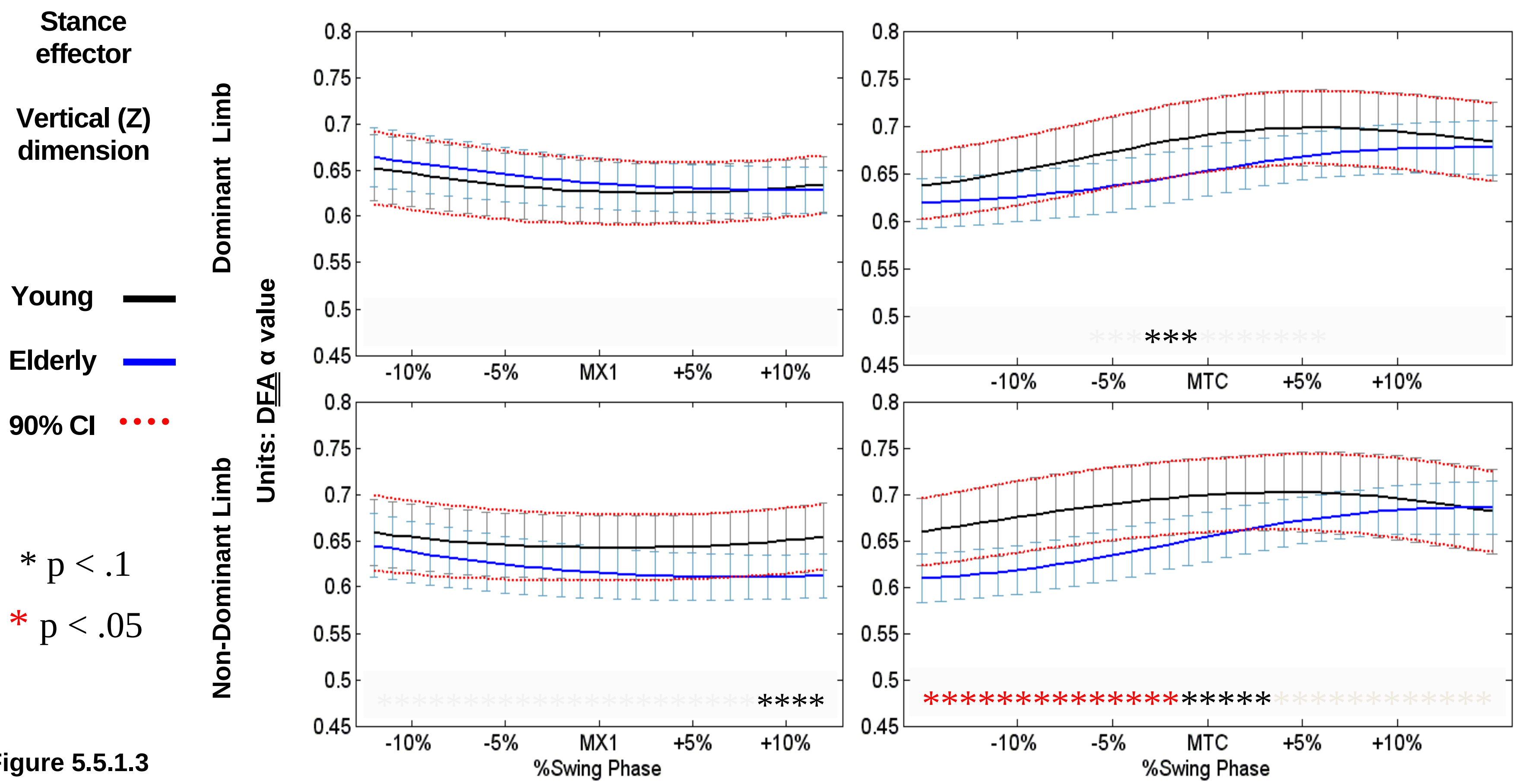


Figure 5.5.1.3

DFA profiles of the young and elderly groups for the stance effector system in the vertical dimension. The abscissa represents the normalised increments of swing time relative to reference ‘states’ MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

5.5.2 Anterior-Posterior dimension

Figures 5.5.2.1-5.5.2.3 display the anterior-posterior dimension of persistence for the combined, swing and stance effectors respectively. There are no significant differences between the groups for the combined effector, or the swing effector (Figures 5.5.2.1 and 5.5.2.2). Relative to the amplitude of the DFA signal in this dimension, the within group inter-participant variability is quite large for both groups. Neither group showed any differences between the dominant and non-dominant limbs. However, Figure 5.5.2.3 did reveal significantly more persistence for state-relevant time-slices in the MX1 region for both the dominant and non-dominant limbs.

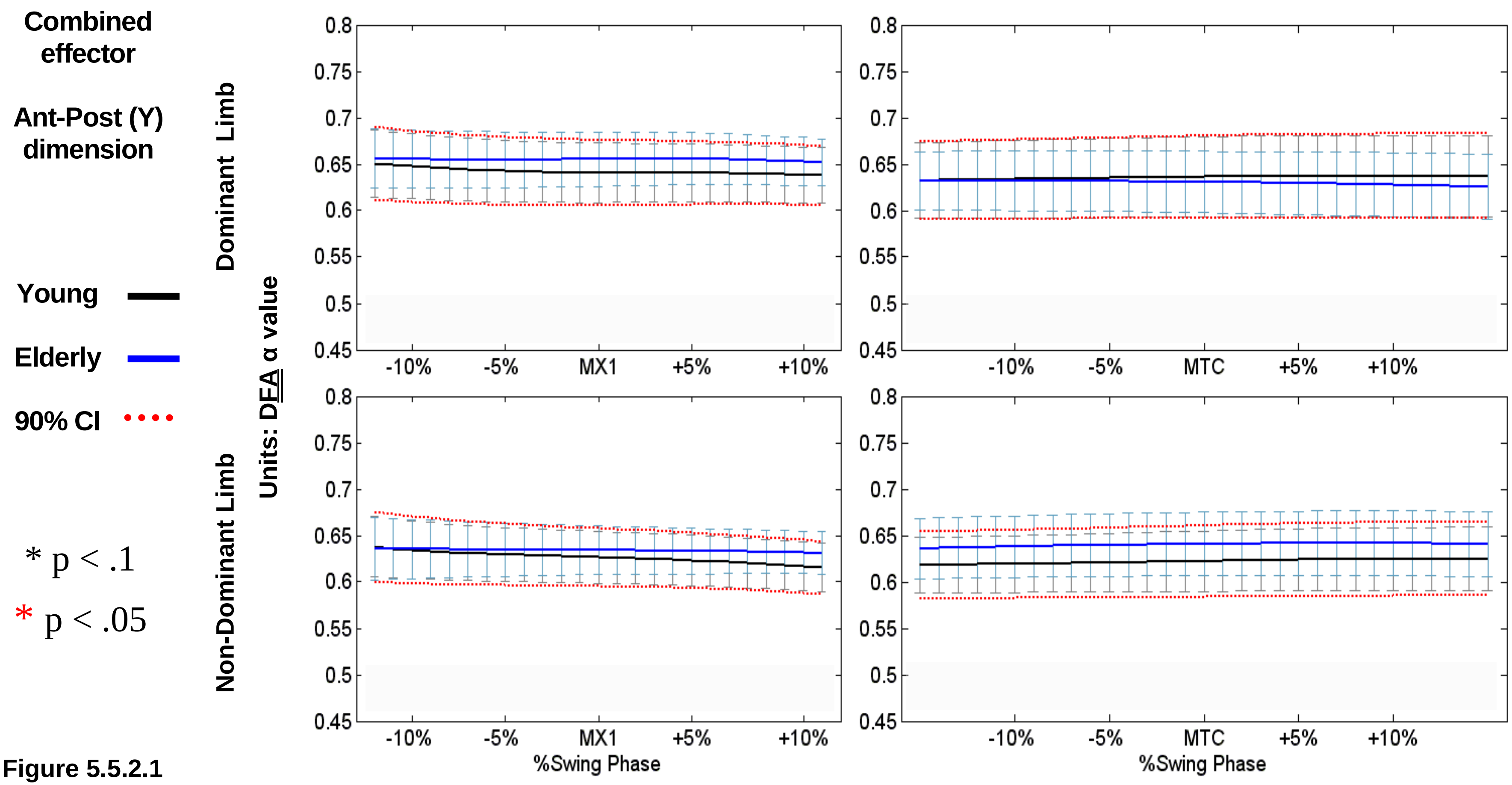


Figure 5.5.2.1

DFA profiles of the young and elderly groups for the combined effector system in the anterior-posterior dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

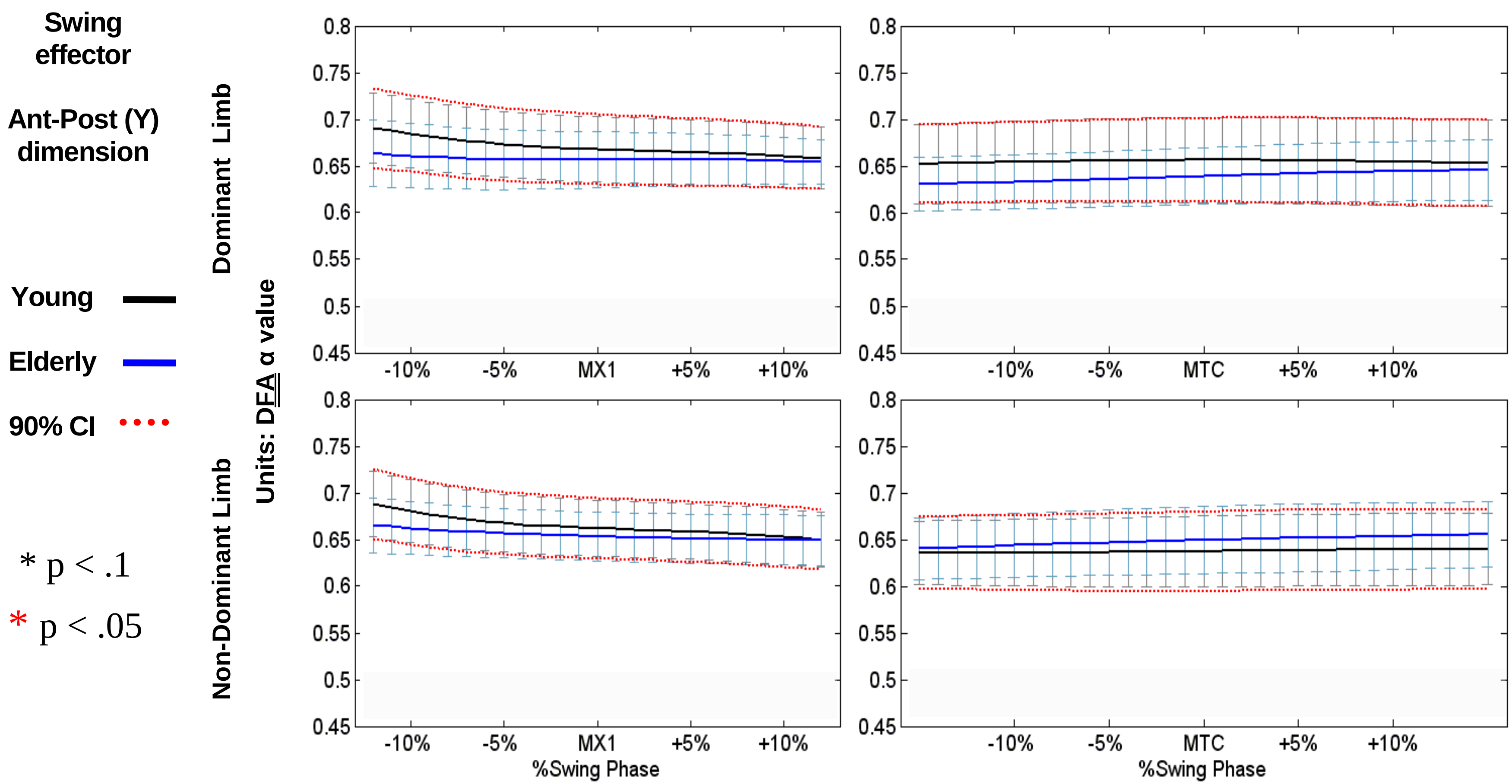


Figure 5.5.2.2

DFA profiles of the young and elderly groups for the swing effector system in the anterior-posterior dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

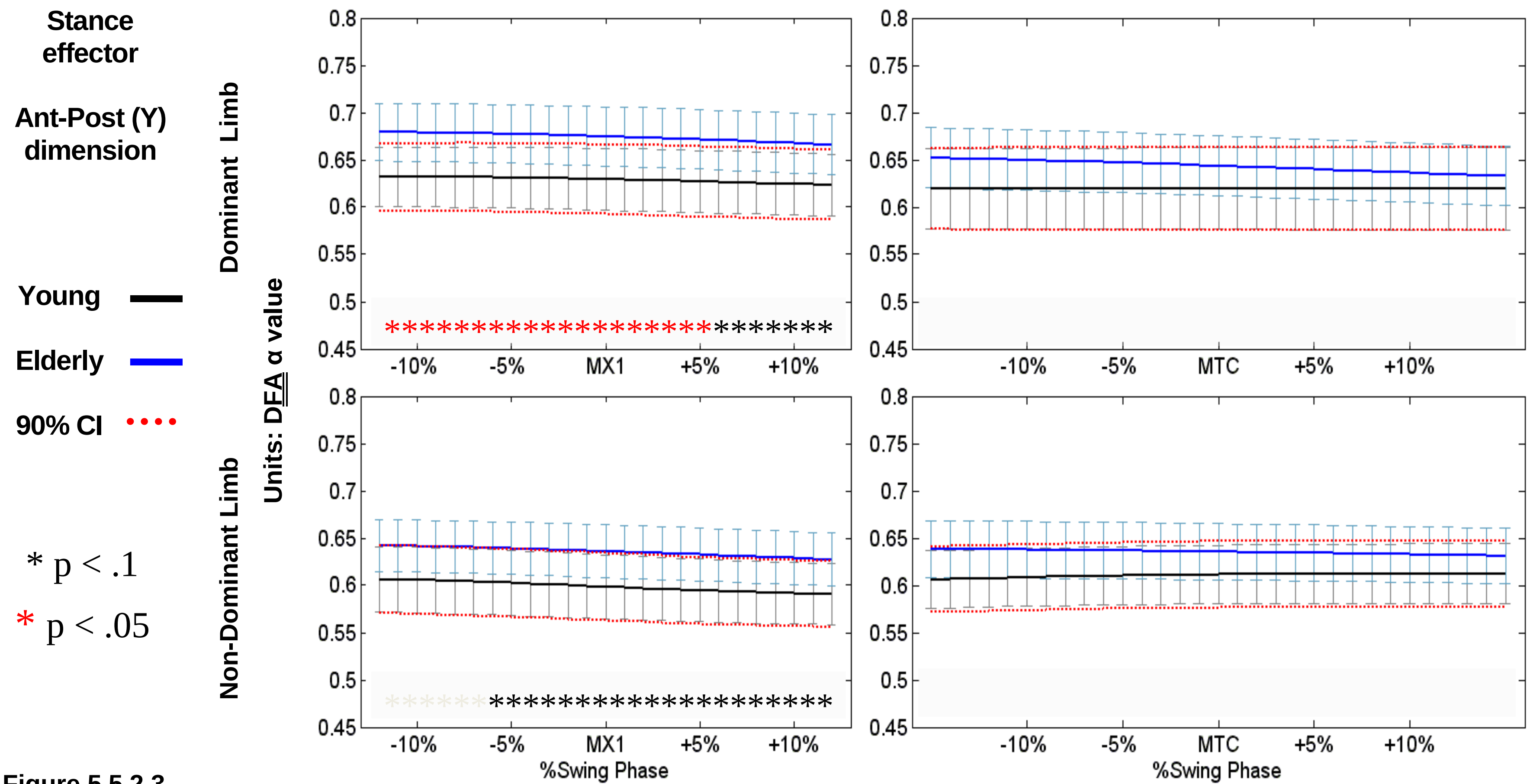


Figure 5.5.2.3

DFA profiles of the young and elderly groups for the stance effector system in the anterior-posterior dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

5.5.3 Medio-lateral dimension

Figures 5.5.3.1-5.5.3.3 display persistence in the medio-lateral dimension for the combined, swing and stance effectors, respectively.

Figure 5.5.3.1 shows the shape of the curve of the combined effector is different between the groups midway between the MX1 and MTC region (for example, the non-dominant limb at $MX1_{+10\%}$), although differences are not quite significant [M_Y = 0.65(0.09), M_E = 0.61(0.06), $t(54) = 1.6$, $p=.12$]. While the trend is similar for the dominant limb, there is not a significant difference ($p = .3$).

Qualitatively, the elderly combined and swing effectors of both the dominant and non-dominant limbs show a reduction in persistence during the MX1 region, in comparison to the young group (Figures 5.5.3.1 and 5.5.3.2). However, this difference between the groups is not quite significant ($p>.1$). As the swing-cycle progresses to the MTC region, the swing effector (Figure 5.5.3.2) shows a significant increase in persistence for the elderly group at MTC and beyond for both the dominant and non-dominant limbs. The between-group difference in the level of 'persistence' found in the medio-lateral dimension is evident only in the swing effector. Figure 5.5.3.3 shows the stance effector in the medio-lateral dimension has no between-group differences.

Within both groups, there were no significant differences in the medio-lateral dimension of the effectors between the dominant and non-dominant limbs, when comparing persistence measured by the DFA method.

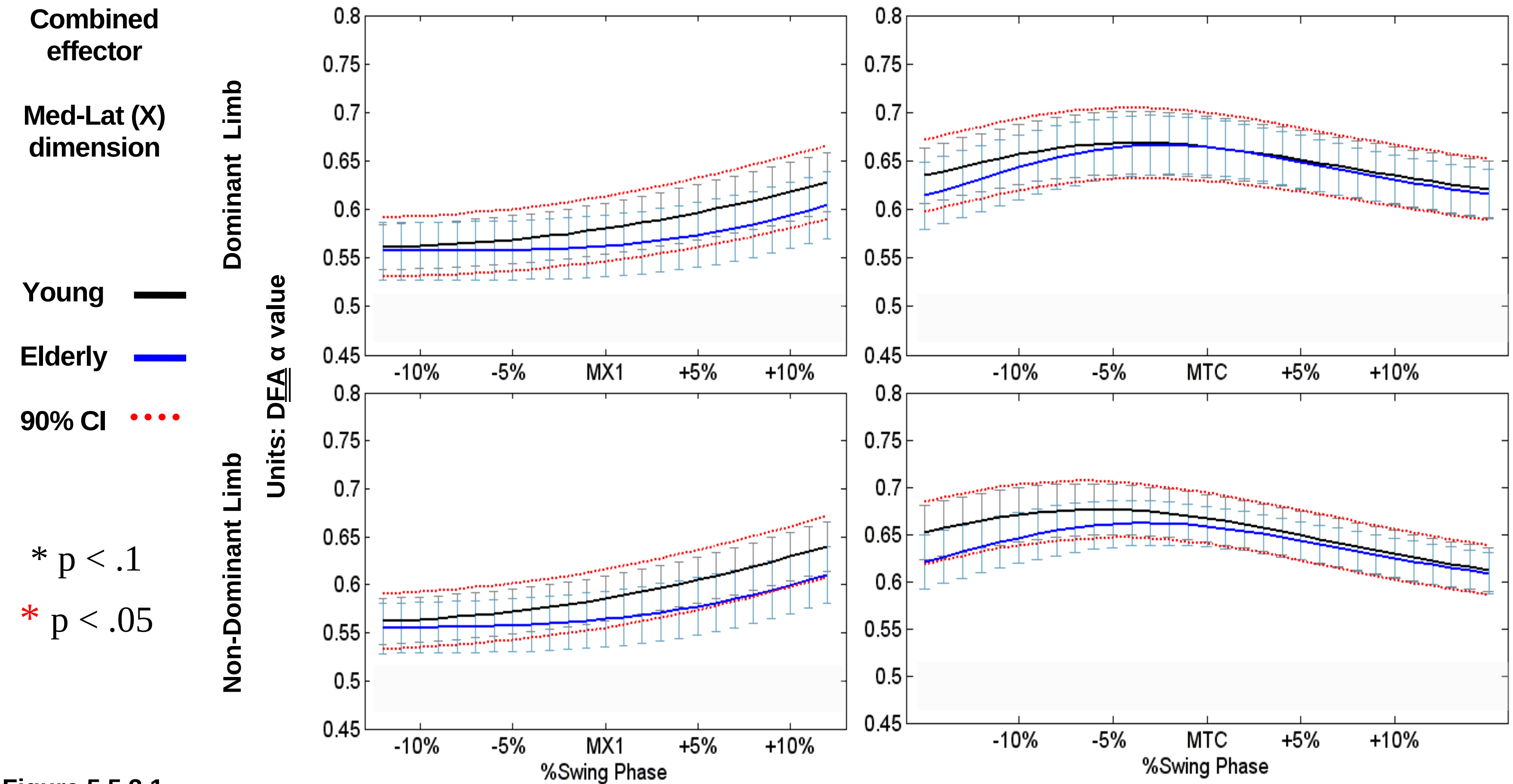


Figure 5.5.3.1

DFA profiles of the young and elderly groups for the combined effector system in the medio-lateral dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

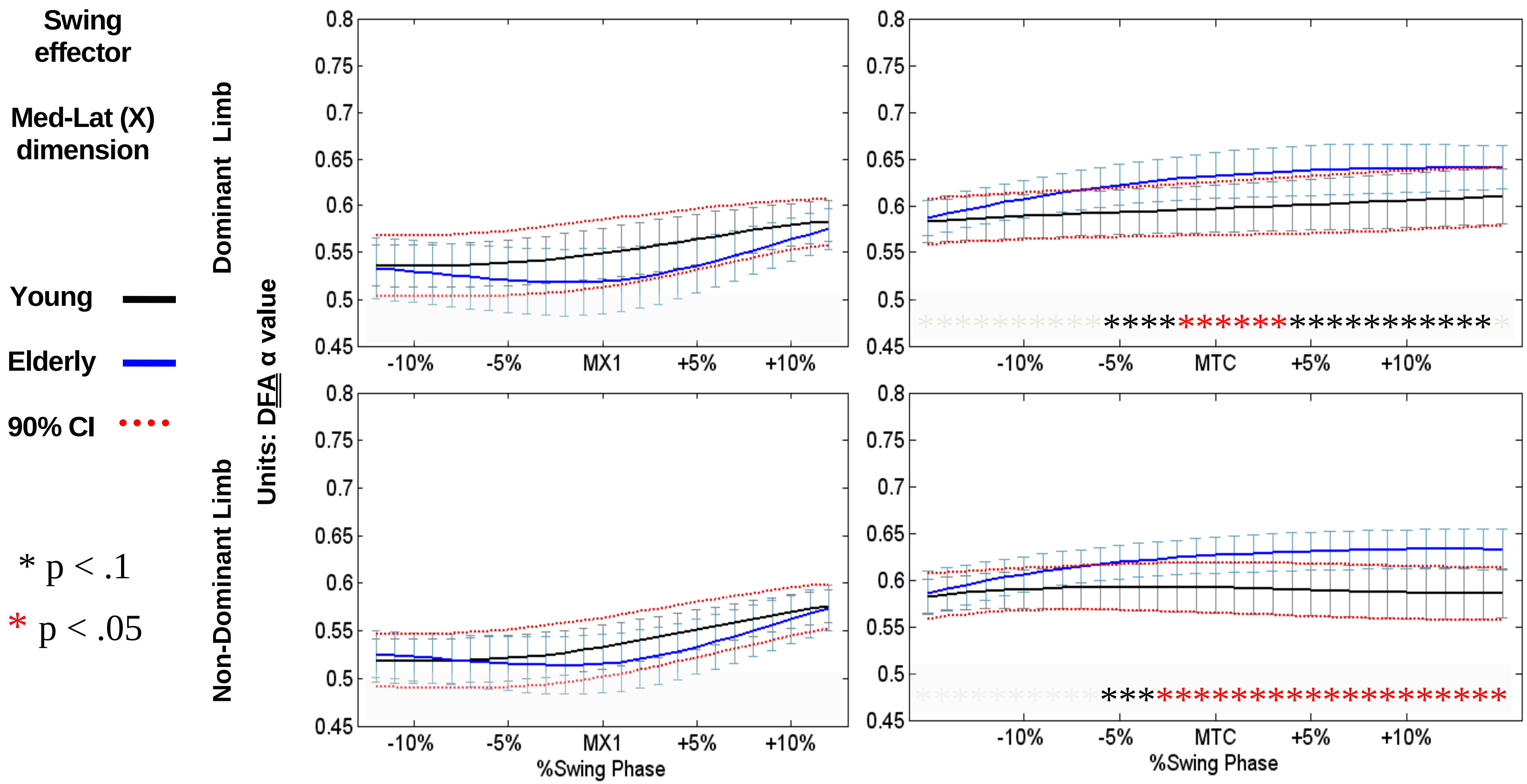


Figure 5.5.3.2

DFA profiles of the young and elderly groups for the swing effector system in the medio-lateral dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

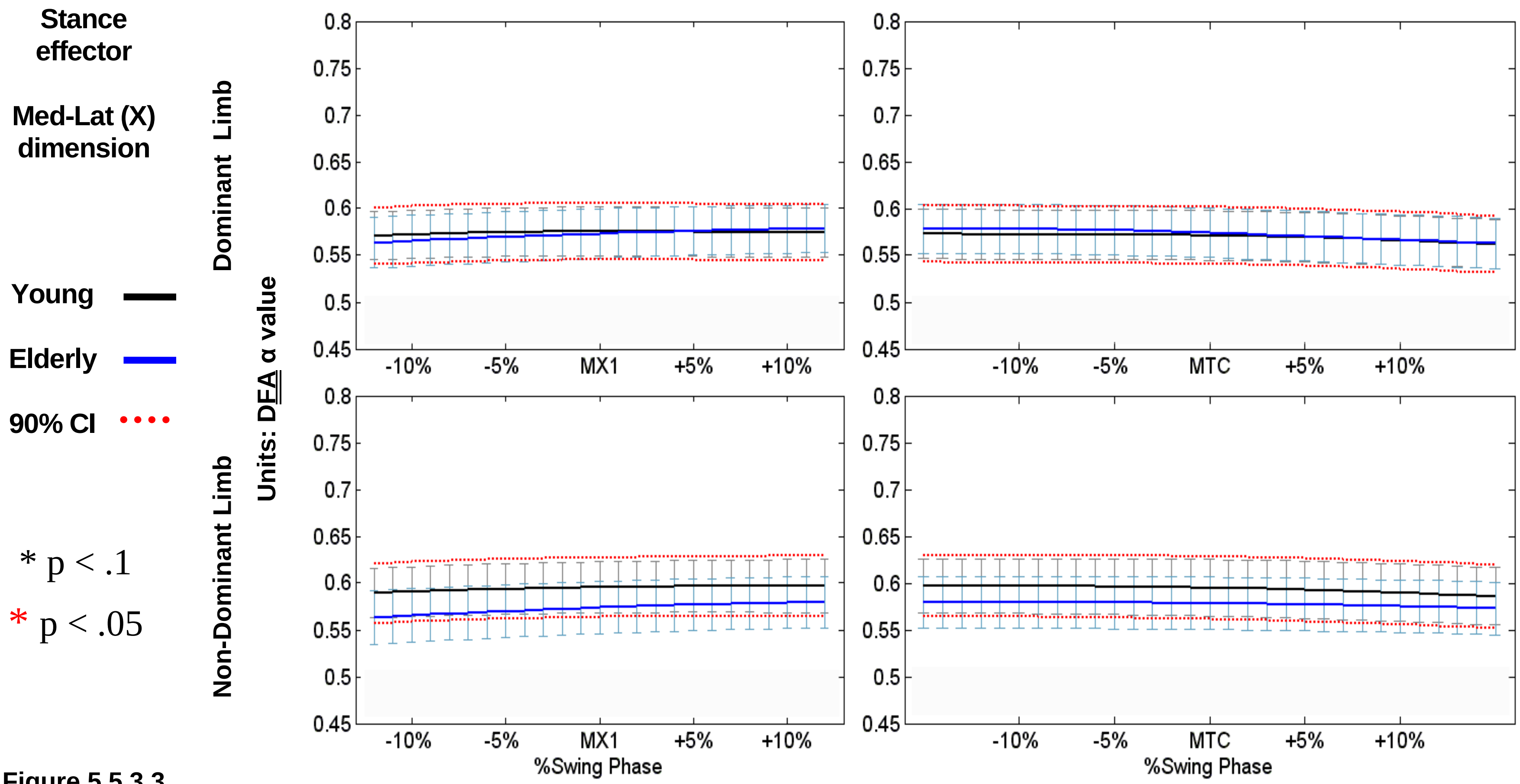


Figure 5.5.3.3

DFA profiles of the young and elderly groups for the stance effector system in the medio-lateral dimension. The abscissa represents the normalised increments of swing time relative to reference 'states' MX1 and MTC. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

5.5.3.1 Assessment of the auto-correlation function in the medio-lateral dimension

For the medio-lateral dimension, the DFA results demonstrate qualitatively, that there is a lack of persistence in the time-series for both groups, but particularly the elderly group during the MX1 region. Therefore, the auto-correlation ACF_{lag1} results were explored to get an indication of short-term anti-persistence. The results are displayed in Figure 5.5.3.1.1.

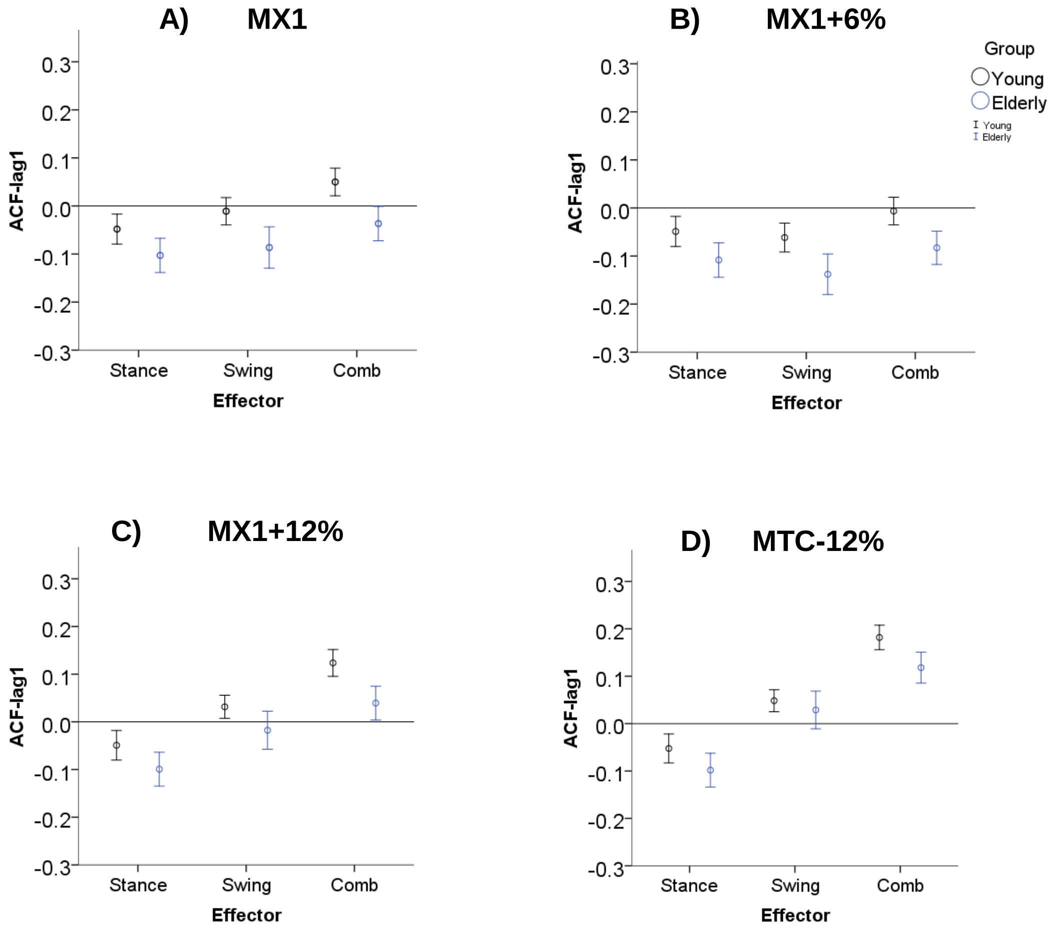


Figure 5.5.3.1.1

Effector plots of the auto-correlation function for time-lag 1 (ACF_{lag-1}) in the medio-lateral dimension for the young and elderly groups. Plots A, B, C, and D represent specific state-relevant time of the swing cycle, MX1, MX1+6%, MX1+12%, MTC-12%. Each error-bar represents the 95% confidence interval about the group mean. The group mean and error-bar has been obtained by combining the dominant and non-dominant limb cases into one group. Values below the zero reference represent anti-correlations. This is the likelihood that a large value will be followed by a small value (and vice-versa), i.e. opposite to persistence.

The ACF_{lag-1} results reveal significantly greater anti-correlations for all three effectors in the medio-lateral direction at different state-relevant times of the swing-cycle. In the combined effector, the elderly have significantly ($p<.005$) more anti-correlations at MX1, MX1+6%, MX1+12%, and MTC-12%. In the swing and stance effectors, the elderly have significantly ($p<.05$) more anti-correlations at MX1, MX1+6%, and MX1+12%. While MTC-12% is also significant for the stance effector at $p<.1$.

5.6 Between-limb differences in persistence

From all the time-slices where between dominant and non-dominant differences can be compared, there was only one finding of a significant difference between the dominant and non-dominant limbs. This was found for the elderly group in the anterior-posterior dimension of the stance effector. The dominant limb demonstrated significantly higher persistence in comparison to the non-dominant limb ($p<.1$), but this was not significant at $p<.05$.

5.7 Differences in persistence between effector and state

Figure 5.7.1 displays effector persistence differences between MX1 and MTC states in the vertical dimension. The graphs are separate for age groups and dominant limb groups. Displayed on the graphs are asterisk to indicate significant differences ($p<.05$) that relate to hypotheses 5.2 and 5.3, as listed in section 5.2.1.1.

Within-subject paired t-tests revealed between-state (MX1 and MTC) significant differences (Figure 5.7.1). For the stance effector, the young group shows a significant increase in persistence for both limbs between MX1 and MTC ($p<.005$). In contrast, the elderly group only display a significant increase in stance effector persistence for the non-dominant limb ($p<.05$). The elderly demonstrate a significant increase in the combined effector between MX1 and MTC for both limbs ($p<.005$). In contrast, the young group only displays a significant increase for the non-dominant limb ($p<.05$). Persistence in the swing effector increases significantly ($p<.05$) between MX1 and MTC, but only in the non-dominant limb of both groups of subjects.

A one-way ANOVA revealed between-effector significant differences within-states MX1 and MTC ($p<.05$). The young group showed only a significant between-effector difference for the stance-swing effectors at MX1 in the dominant limb ($p<.05$). The elderly at MX1 also demonstrated a significant difference between the stance and swing effectors, but only in the non-dominant limb ($p<.05$). At MTC, the elderly demonstrated in both limbs, that the stance effector has significantly less persistence-likelihood at MTC compared to persistence-likelihood in both the combined and swing effectors (p<.05).

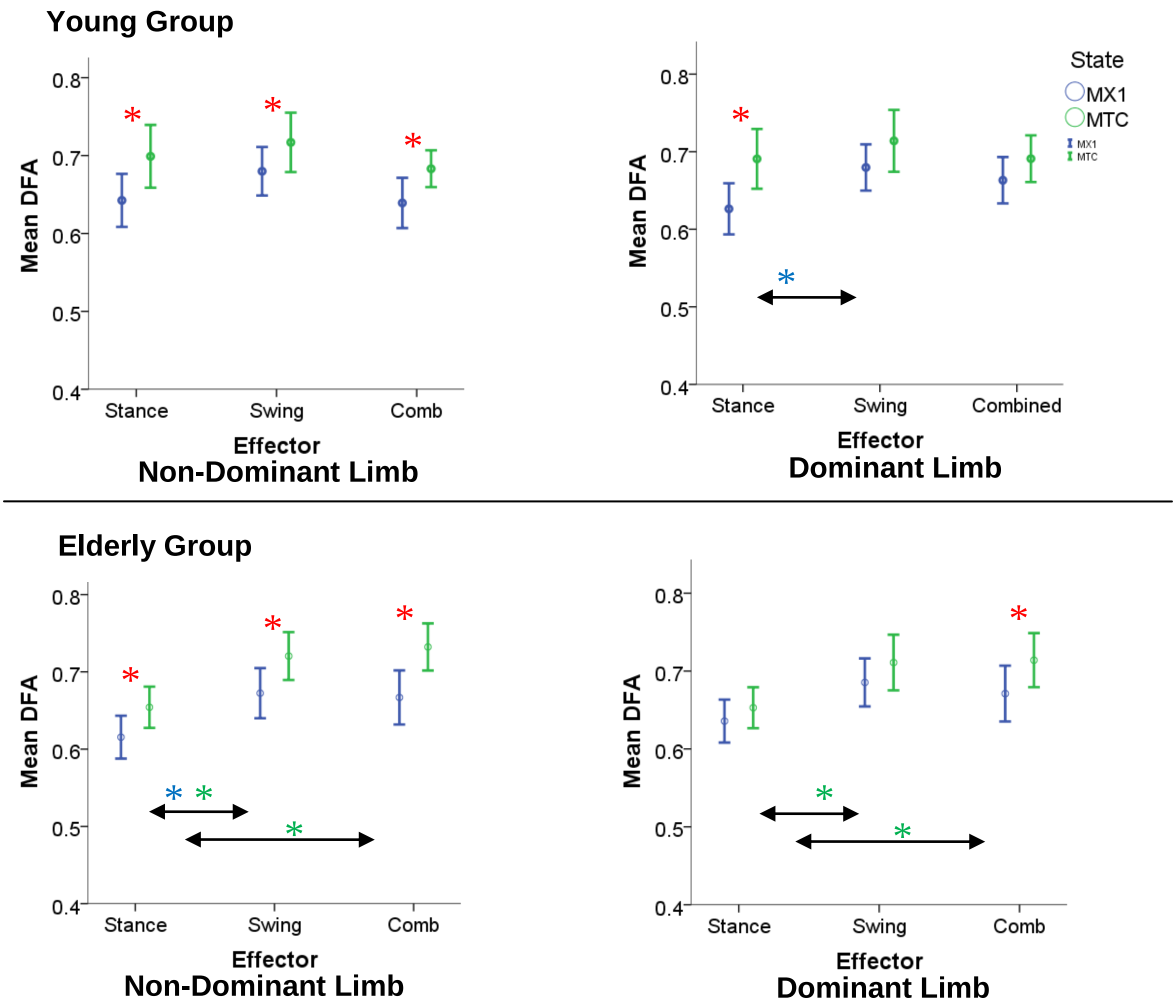


Figure 5.7.1

Group means and 95% confidence intervals (error-bars) for effector and state persistence. Top and bottom rows are the young and elderly groups respectively. Significant differences are for p<.05. One-way ANOVA and Bonferroni post-hoc test provides between-effector significant differences for within-states and are displayed as blue (within MX1 state) and green (within MTC state) asterisk. Paired t-test provide significant differences between MX1 and MTC states within-effector, and are displayed as red asterisk.

5.8 Interaction effect of age, effector and state-time on persistence of the effectors in the vertical dimension

This section explores the effect of interactions among independent variables 'effector', 'group' and 'state-relevant time' on the dependent variable of persistence. Sections 5.8.1.1-3 will investigate the null hypotheses 5.3 and 5.4 from section 5.2.1.1. Four discrete state-relevant time-slices are selected to represent the time-course of persistence change made within three regions: MX1 (MX1-12%, MX1-6%, MX1, MX1+6%), MX1-MTC (MX1+6%, MX1+12%, MTC-12%, MTC-6%) and MTC (MTC-6%, MTC, MTC+6%, MTC+12%). Except for the time scale difference between MX1+12% to MTC-12%, each time-slice is separated by a proportionate period of 6% of the entire swing phase. All significant differences are at $p<.05$. The dominant and non-dominant limbs have been pooled, in effect doubling the sample size from N = 28, to N = 56, within each group. This makes the assumption that the between-limb behaviour of the effectors within a person involves independent function within the loco-sensorimotor control system. There have been many diverse studies that have proposed independent function between the lower limbs during walking (Sadeghi, Allard, Prince and Labelle, 2000); therefore, there can exist some support for assuming that the left and right limb data can be considered as independent cases in a statistical analysis.

5.8.1 Time region MX1

There is a main effect for a significant difference between effectors $F(2, 220) = 17.07$, $p<.001$, $\eta^2 = .13$). Contrasts show that effector persistence is significantly higher for the swing effector when compared to the stance effector $F(1, 110) = 50.16$, $p<.001$, $\eta^2 = .31$), and significantly higher for the swing effector when compared against the combined effector $F(1, 110) = 14.43$, $p<.001$, $\eta^2 = .12$). Figure 5.8.1.1C illustrates the 'effector' x 'group' interaction, when persistence is averaged across state-relevant time, and statistically there was no significant interaction ($p>.2$). This tells us that when effector persistence is averaged across the MX1 region, both groups are not significantly different when comparing persistence in the stance- and combined-effectors with reference to swing-effector persistence. However, the effect of 'moment-to-moment' changes in effector persistence is observed to be different between the groups. Figures 5.8.1.1 A and B plots the interaction separately within each group.

To assess whether there were any moment-to-moment effects between 'effector' and 'group' in this region, the above interactions of 'effector' x 'group' were compared between 'state-relevant time-slices'. Statistically, there was a significant difference for interaction of 'effector' × 'state-time' when comparing between-groups ($F(6, 660) = 4.86$, $p<.01$, $\eta^2 = .04$). This finding makes the above finding of 'no-difference' between groups when comparing between effector persistence in the MX1 region to be a little misleading. Between each consecutive state-relevant time-slice, planned contrasts (simple effects) compared between-effectors on the persistence parameter at two levels: stance-swing; and swing-combined.

Comparing between the stance and swing effector showed that the young and elderly behave significantly different when these effectors are compared between time-slices MX1-12% and MX1-6%, and also between MX1 and MX1+6% ($F(1, 110) = 11.71$, $p<.001$, $\eta^2 = .10$; $F(1, 110) = 4.08$, $p<.05$, $\eta^2 = .04$, respectively). This tells us that the stance effector persistence relative to the combined effector persistence is significantly decreasing in the elderly group, compared to the young group, when comparing between levels of 'state-relevant time' prior to MX1 and following MX1. Planned contrasts comparing between the swing and combined effector showed that the young and elderly behave differently between the 'state-relevant times' MX1-12% and MX1-6%, and also MX1-6% and MX1 ($F(1, 110) = 13.70$, $p<.001$, $\eta^2 = .11$; $F(1, 110) = 8.95$, $p<.005$, $\eta^2 = .08$, respectively). This tells us that the combined effector persistence relative to the swing effector persistence is increasing significantly in the elderly group, compared to the young group, during state-time prior to MX1.

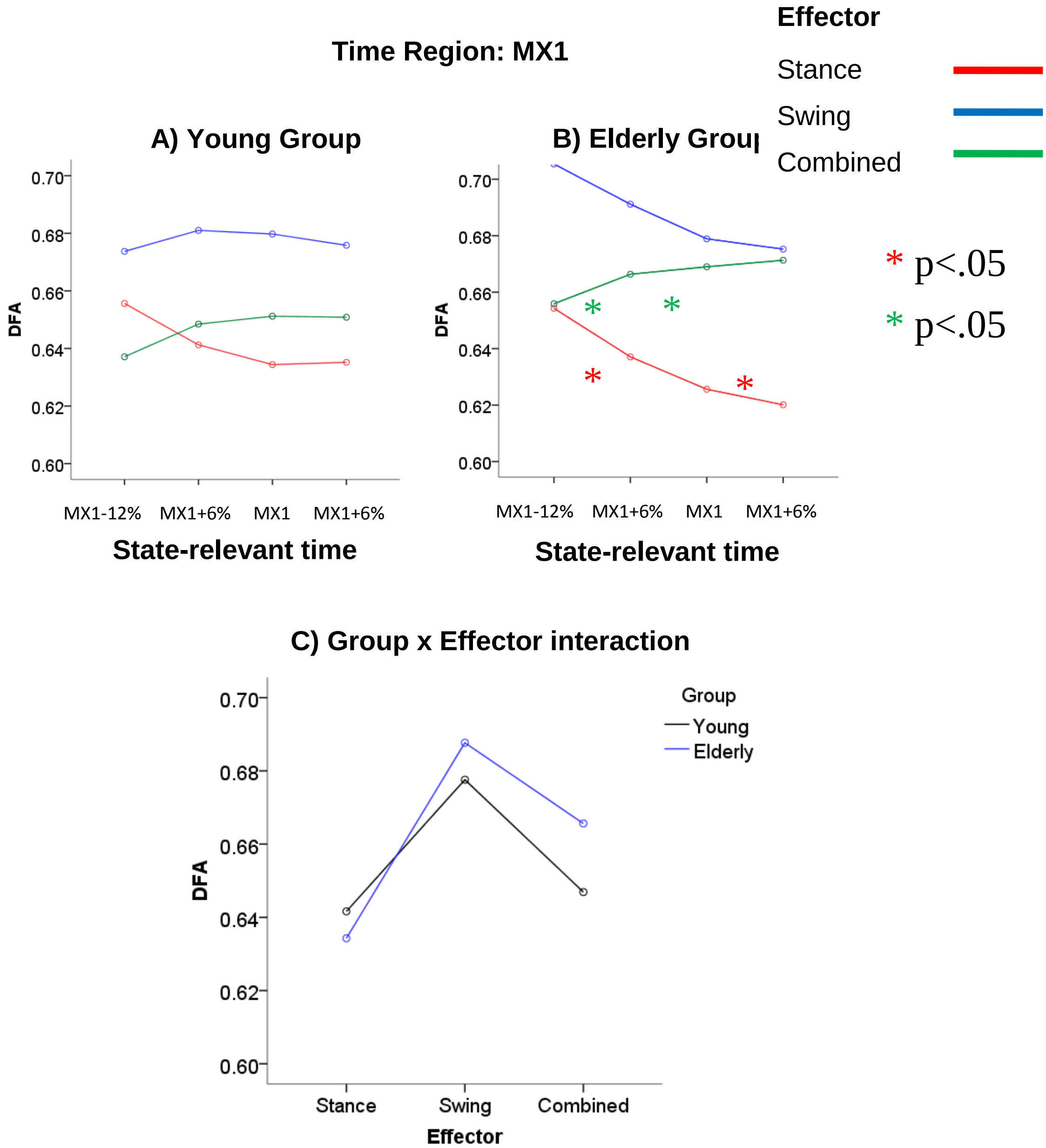


Figure 5.8.1.1.

Graphs of the vertical dimension of the mean DFA, α, (i.e. persistence-likelihood parameter) about the MX1 region. Graph A) demonstrates the interactions of effector and time state for the young group. Graph B) represents the elderly group interactions. Simple contrast interactions within the three-way ‘group’ x ‘effector’ x ‘state-time’ interaction are indicated by red and green asterisk, for where significant contrast interactions are found between stance and swing (Level 1, red), and swing and combined (Level 2, green), respectively. The asterisk are displayed on the elderly plot (B). In plot C) significant ‘group’ x ‘effector’ contrasts are indicated by a black asterisk (p<.05).

5.8.2 Time MX1-MTC

The state-relevant time-slices selected for examination in this section of the swing-cycle were MX1-12%, MX1-6%, MTC+6%, MTC+12%. This region of the swing-cycle defines the effector trajectories transitioning from states MX1 to MTC.

There was a significant main effect for effector ($F(2, 220) = 27.45$, $p<.001$, $\eta^2 = .20$). Contrasts show a significantly higher persistence for the swing effector compared to the stance effector ($F(1, 110) = 56.47$, $p<.001$, $\eta^2 = .34$).

The analysis of effector x group interaction on effector persistence contributes to null hypothesis 5.3, and also extends upon results from section 5.7, here by investigating the effect of age. When collapsing across time-slices for region MX1-MTC, the interaction effect of 'effector' × 'group' on average effector persistence is examined. Figure 5.8.2.1C illustrates the interaction, and this is significant ($F(2, 220) = 8.60$, $p<.005$, $\eta^2 = .07$). Planned contrasts find that persistence between the stance and swing effectors is significantly different ($F(1, 110) = 5.68$, $p<.05$, $\eta^2 = .05$). Figure 5.8.2.1C shows the stance effector persistence is significantly more decreased in the elderly, compared to the persistence decrease shown by the stance effector of the young group.

Also, a contrast shows that the persistence difference between the swing and combined effectors is significantly different between the groups ($F(1, 110) = 4.99$, $p<.05$, $\eta^2 = .04$). Figure 5.8.2.1C illustrates that the combined effector persistence is higher relative to the persistence in the swing effector for the elderly group, and this is significant when compared to the young who demonstrate a relatively lower persistence of the combined effector.

The hypothesis in this thesis is that the DFA represents the persistence-likelihood, which in-turn represents the degree of frequent and effective controller intervention to change effector states. Therefore, the results above tell us that the intervention frequency applied by the loco-sensorimotor controller that cause change to effector states, with respect to the combined effector and the stance effector, is significantly different between the groups during the transition period between MX1 and MTC. When considering the swing effector as a reference for both groups, compared to the young group, the elderly display significantly more frequent interventions that cause change in the stance effector states; while displaying significantly less frequent interventions that cause change in combined effector states.

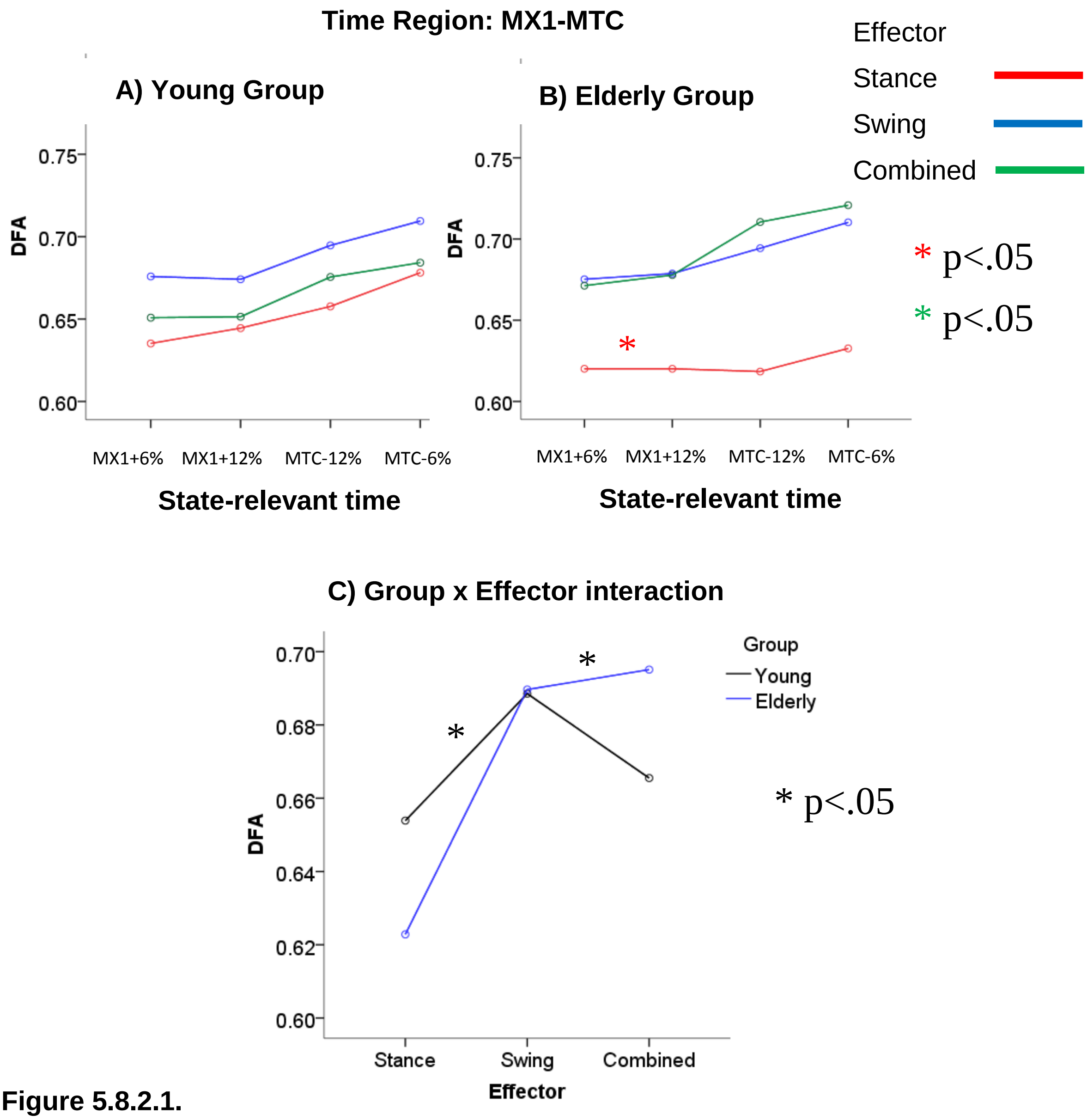


Figure 5.8.2.1.

Graphs of the vertical dimension of the mean DFA scaling exponent about the transition region between MX1-MTC. Graph A) demonstrates the interactions of effector and time state for the young group. Graph B) represents the elderly group. Graph C) illustrates the group x effector interaction when collapsing for state-relevant time. Simple contrast interactions within the three-way 'group' x 'effector' x 'state-time' interaction are indicated by red and green asterisk, for where significant contrast interactions are found between stance and swing (Level 1, red), and swing and combined (Level 2, green), respectively. The asterisk are displayed on the elderly plot (B). In plot C) significant 'group' x 'effector' contrasts are indicated by a black asterisk (p<.05).

Figure 5.8.2.1 A and B illustrates the ‘moment-to-moment’ contrasts of ‘effector’ × ‘time-state’ × ‘age’ on persistence. Statistically, there was a significant interaction ($F(6, 660) = 3.26$, $p<.05$, $\eta^2 = .03$). Planned contrasts reveal where the groups are significantly different. When comparing between age-groups, persistence is significantly different between the stance and swing effectors between the time-states MX1+6%-MX1+12% ($F(1, 110) = 9.31$, $p<.005$, $\eta^2 = .08$). This region indicates where the young group are significantly different in the way persistence increases in the stance effector, compared to the elderly group. For the young group with respect to the elderly group, this can be likened to a significant decrease in the intervention frequency of the loco-sensorimotor controller to cause a change in stance effector state.

When comparing between the swing and combined effectors on persistence, and when comparing between consecutive time slices, there were no significant differences. Therefore, while there is a significant interaction difference of ‘age’ x ‘combined- and swing- effector’ when persistence of the effectors is averaged across time slices of the MX1-MTC region, there is no significant difference when comparing for interactions between smaller ‘moment-to-moment’ time intervals within this region.

5.8.3 Time region MTC

This region of the swing-cycle defines the effector trajectories through the MTC region. The state-relevant time-slices selected for examination in this section of the swing-cycle were MTC-12%, MTC-6%, MTC, MTC+6%.

There was a main effect for effector ($F(2, 220) = 5.55$, $p<.005$, $\eta^2 = .048$). For all participants, contrasts revealed a significantly higher swing effector persistence compared to stance effector persistence ($F(1, 110) = 12.82$, $p<.005$, $\eta^2 = .10$).

The interaction of ‘effector’ x ‘group’ contributes to null hypothesis 5.3 and extends upon results from section 5.7 by investigating the effect of age on effector differences. Collapsing across time-slices for region MTC, the average effector persistence is compared between the three effectors in the effector × group interaction. Figure 5.8.3.1C illustrates the interaction which is significant ($F(2, 220) = 9.59$, $p<.001$, $\eta^2 = .08$). Planned contrasts were performed on the stance-swing, and the swing-combined pairings, to determine where the groups were significantly different. Results found that persistence between the stance and swing effectors is significantly different when comparing between groups ($F(1, 110) = 7.08$, $p<.01$, $\eta^2 = .06$). Likewise, persistence in the swing effector when compared to the combined effector, was significantly different when compared between groups ($F(1, 110) = 4.52$, $p<.05$, $\eta^2 = .04$). These significant contrasts were similar results that were found in the MX1-MTC transition region, and therefore a similar interpretation can be made for this MTC region.

To observe where the above simple interactions of effector x group may exist when comparing between smaller consecutive state-relevant time slices, the three-way interaction of 'effector' x 'state' x 'age' is not significant.

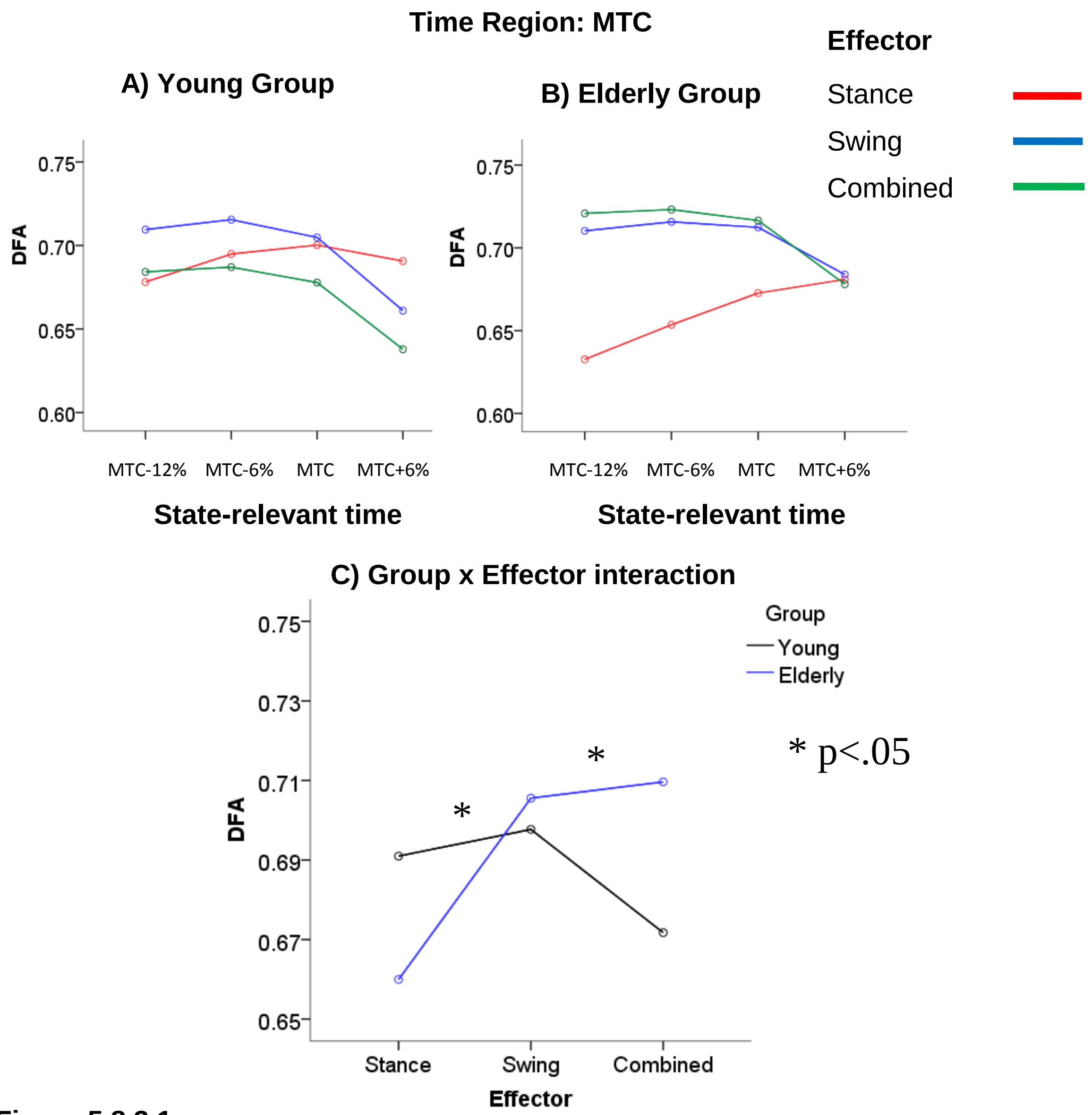


Figure 5.8.3.1.

Graphs of the vertical dimension of the mean DFA scaling exponent at the MTC region. Graph A) demonstrates the interactions of effector and time state for the young group. Graph B) represents the elderly group. Graph C) illustrates group x effector interaction. Interactions effect of 'group' x 'effector' x 'state-time' was not significant. In plot C) significant 'group' x 'effector' contrasts are indicated by a black asterisk (p<.05).

5.9 Inter-variable correlations between DFA, variance and skewness

So far, the results presented have investigated between-group differences for each separate dependent variable. This section investigates how the dependent variables are related. Specifically, to determine how skewness, variance and persistence are related with respect to controlling the effector systems that define the toe clearance task.

5.9.1 Background

Statistically, skewness is a parameter describing the distribution characteristic of the average tendency. Because skewness is a measure of the distribution shape, it has been hypothesised to be representing a control policy behind the performance of the toe clearance task (Begg et al., 2007). Anti-persistence of gait parameters has also been linked to a control policy of walking (Dingwell and Cusumano, 2010). Currently, there has been no research which has investigated a link between skewness and persistence of gait parameters. In terms of persistence and average measures of variance, previous studies of time-distance gait parameters have found that the magnitude of the stride-to-stride fluctuations (variance) is independent of the fluctuation dynamics (persistence) (Hausdorff et al., 2000; Gates and Dingwell, 2007). Theoretically, correlations among scores of a time-series are independent of the average size of the fluctuations, because the structure of the cycle-to-cycle fluctuations are not based upon the average variance. One research group has proposed that there exists an inverse relationship between variance and persistence based upon the changes observed across different walking speeds (Jordan et al., 2007).

The research question of whether the DFA scaling is adequately describing a control policy can obtain general support from the parameter of skewness. The dynamic change to the skewness parameter across the swing phase time slices can provide further evidence of a control policy. If a controller views the time course of a movement trajectory as a risk by associating 'cost' penalties then this can potentially be reflected in the evolution of the skewness parameter. Kurtosis is a secondary parameter that can describe the shape of a distribution. Kurtosis, representing the 'peakedness' of a distribution, can also be hypothesised to represent a task-relevant control policy. A distribution which has minor skewness can still possess a highly leptokurtic shape which is why the correlation of kurtosis and skewness between participants will not always be 'perfect'. It has been found previously, that with respect to toe trajectory data in walking, skewness and kurtosis share a strong positive correlation between participants (Begg et al., 2007). This is because, some 'participants' are likely to adopt a type of control policy that considers errors on both sides of a toe-clearance distribution as being task-relevant. This type of control can also be considered as risk-averse. A risk-averse policy can be associated with positive skewness and/or high kurtosis. Therefore, a risk-averse policy can be expressed by some combination of two parameters of a distribution. Alternatively, a risk-neutral policy can be expected to be associated with a combined low skewness and low kurtosis. Therefore, within any group of participants, three different forms of a control policy could be expressed in the shape of a distribution. Although it is likely that the explicit control policy is masked in parameters that represent average distribution shape, it is possible that a control policy can be revealed by investigating relationships between skewness, standard deviation (as a form of kurtosis) and the persistence-likelihood parameter, 'α'.

5.9.2 Correlation table

The correlation coefficient, 'r', represents the degree of co-variance between two variables and for values that approach 1, there is a strong association (co-variance) between variables. Correlation coefficients were computed between statistics of DFA (persistence-likelihood), skewness and variance (standard deviation) of the effector systems and their dimensions at the MTC event. The results are from combining both dominant and non-dominant limbs together within the young and elderly groups, and the correlation coefficients, 'r values', are presented in Table 5.9.2.1 and 5.9.2.2, respectively.

The tables are displaying only when $r>.2$, therefore, empty 'cells' represent $r<.2$. Significant correlations of $p<.05$ and $p<.01$, are associated with $r>.25$ and $r>.35$, respectively. The table of r values is ordered within three general levels. The first level is the control statistic: DFA, SD, and skewness. The second level is the effector: stance

(St), swing (Sw) and combined (Co). The third level is the dimension: medio-lateral (X), anterior-posterior (Y), and vertical (Z). For easier interpretation, shaded areas and colours of the table represents correlations between different statistics (Blue represents SD-DFA; Green represents Skew-DFA; Red represents Skew-SD). Non-shaded areas represent within-statistic correlations.

Table 5.9.2.1.

Correlation coefficients between DFA, skewness and standard deviation at the MTC event for young participants (dominant and non-dominant limbs are combined). Correlations of r>.35 and r>.25 are significant at p < .01 and p < .05 respectively (for two-tailed test).

YOUNG		DFA									STANDARD DEVIATION									SKEWNESS								
		Stance			Swing			Combined			Stance			Swing			Combined			Stance			Swing			Combined		
		X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z
DFA	St_x	1		.36			.38						-.28								.34							
	St_y		1	.40	.31	.73		.21	.84		-.32	-.29		-.34		-.27			-.28									
	St_z			1		.45	.68	.30	.31		-.24		.60						-.24		.21				-.23			
	Sw_x				1			.37			-.35	-.27		-.40		-.20									-.20			
	Sw_y					1			.94			-.23	.27															
	Sw_z						1	.23		.42	-.26		.57			.42				.23								
	Co_x							1									-.21		-.28			.22						
	Co_y								1			-.23		-.23														
	Co_z									1						.32			.39						.22			.23
SD	St_x										1	.38		.70		.29	.47	.20	.35									
	St_y											1	.39	.52	.82	.55	.37	.89	.51					-.29			-.22	
	St_z												1	.35	.32	.63		.28										
	Sw_x													1	.31	.55	.59	.33	.45									
	Sw_y														1	.30	.29	.98	.30		-.21		.23	-.44			-.34	
	Sw_z															1	.48	.32	.77						.20			
	Co_x																1	.27	.56		-.32						-.21	-.22
	Co_y																	1	.34		-.23		.21	-.44			-.34	
	Co_z																		1		-.22	.24			.28			.24
SKEW	St_x																			1			-.22					
	St_y																				1			.69			.85	
	St_z																					1	-.32	.22		-.34	.22	.31
	Sw_x																						1			.43		-.27
	Sw_y																							1			.94	
	Sw_z																								1			.90
	Co_x																									1		
	Co_y																										1	
	Co_z																											1

Table 5.9.2.2.

Correlation coefficients between DFA, skewness and standard deviation at the MTC event for elderly participants (dominant and non-dominant limbs are combined). Correlations of r>.35 and r>.25 are significant at p < .01 and p < .05 respectively (for two-tailed test).

ELDERLY		DFA									STANDARD DEVIATION									SKEWNESS								
		Stance			Swing			Combined			Stance			Swing			Combined			Stance			Swing			Combined		
		X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z	X	Y	Z
DFA	St_x	1	.21			.34		.48							.22							-.20		-.32	.25			.21
	St_y		1			.61	.22		.83	.34	.24	.32	.21	.35	.23	.35	.26	.29	.23		.21							.36
	St_z			1	.21		.27			.27	-.23	-.25	-.49			-.26						.23						
	Sw_x				1												.31		-.20									
	Sw_y					1	.25	.32	.86	.45	.28	.38	.36	.38	.41	.33	.29	.41	.21	-.21					.30			.54
	Sw_z						1		.30	.72		.29			.21	.37		.28	.38				-.28			-.28		
	Co_x							1							.26			.24										
	Co_y								1	.44	.35	.37	.28	.38	.33	.27	.26	.36							.21			.50
	Co_z									1	.23	.38			.22	.43		.29	.33				-.23			-.28		
SD	St_x										1	.35	.46	.60			.48	.20										.37
	St_y											1	.69	.52	.72	.66		.84	.60		.26							.48
	St_z												1	.60	.50	.56	.36	.51	.36		.28	-.22			.33			.55
	Sw_x													1	.36	.42	.43	.38	.30									.53
	Sw_y														1	.54		.96	.58						.24			.46
	Sw_z															1	.28	.54	.87						.35			.39
	Co_x																1		.21									.31
	Co_y																	1	.57		.25							.45
	Co_z																		1									.32
SKEW	St_x																			1		.37	-.50		-.45	.57		-.22
	St_y																				1			.27			.59	
	St_z																					1	-.65		-.86		.26	
	Sw_x																						1		.72		-.26	
	Sw_y																							1	-.22		.87	
	Sw_z																								1		-.31	.50
	Co_x																									1		
	Co_y																										1	
	Co_z																											1

There are many significant correlations found in the elderly that are not observed in the young, and vice-versa. In comparison to the young group, the elderly have more significant negative correlations between skewness parameters associating the stance and swing effectors. For example, skewness in the vertical dimension (r=-.86) and in the medio-lateral dimension (r=-.5). The elderly group also have a significant negative correlation between the vertical swing and medio-lateral stance effector (r=-.45), while the young group shows a significant negative correlation between the vertical stance and the medio-lateral combined effector (r=-.34). Both groups show a significant negative relationship between the vertical stance and medio-lateral swing effector (r_Y =-.32, r_E =-.65).

The young group show that in the anterior-posterior dimension most of the significant correlations between skewness and standard deviation are negatively correlated. Increased variance is related to decreased skewness in this dimension for the young group. In contrast, the elderly don't exhibit this behaviour, as all correlations are positive. For the elderly, the link between skewness and standard deviation is significant for most variables, as most variables correlate positively with the vertical dimension of the combined effector. The indication is that the larger the general variance in the effector system of the elderly, the larger the skewness found in the combined effector. The results imply that the elderly apply a control policy that penalises the small values if they exhibit variable output from the effector system.

When observing correlations within the standard deviation (SD) statistic, for both groups, SD is significantly correlated with component states in all three-dimensions of the effector systems. The variability (SD) within the anterior-posterior dimension shows a significant and high correlation value among the three effector systems. For example, the elderly group indicates a significant correlation for standard deviation between the stance-swing (r=.72), stance-combined (r=.84) and swing combined (r=.96). The young group also show significant correlations of similar correlation values. This suggests that if there exists variability (SD) of the anterior-posterior dimension of effector states at MTC, it propagates 'un-damped' in all effectors in the same dimension. When considering the more relevant vertical dimension of variability (SD) in the combined effector, there is a significant correlation with standard deviation in all other dimensions of the system, apart from two exceptions. First exception, within the young group, SD (variance) of the vertical combined effector is not significantly correlated with SD in the vertical stance effector. Second exception, the elderly medio-lateral SD of the stance effector is not correlated with the vertical combined effector. In the vertical dimension, the elderly have a significant correlation between swing effector variance and combined effector variance (r=.87); while stance effector variance is also significantly correlated with combined effector (r=.36). The elderly group has significant correlations between the anterior-posterior SD of the three effectors with the vertical combined effector ($r_{St\text{-}C}$ =.60, $r_{Sw\text{-}C}$=.58, $r_{C\text{-}C}$=.57). The young also have correlations, that aren't significantly different to the elderly ($r_{St\text{-}C}$=.51, $r_{Sw\text{-}C}$=.30, $r_{C\text{-}C}$=.34).

The correlations between DFA and the standard deviation and skewness can provide rich information about the locomotor system and the differences between the two groups. The elderly group show a significant correlation when associating skewness in the vertical dimension of the combined effector with the DFA of all three effectors (stance, swing and combined) in the anterior-posterior dimension ($r_{c\text{-}st}$ = .36, $r_{c\text{-}sw}$=.54, $r_{c\text{-}c}$=.50). This indicates that the penalty for low clearance is related to persistence of the effectors in the anterior-posterior direction. The young group demonstrate no such relationships.

The DFA in the anterior-posterior direction of the elderly is significantly correlated in 26 of the 27 SD statistics, all of which are positive r values. While this result doesn't explicitly identify that persistence-likelihood in the anterior-posterior effector components cause variability in the entire effector system, or visa-versa. This is an interesting result, because when contrasted with the young group, whom demonstrate significant correlations in 8 of the 27 SD statistics, 7 of which are negative correlations. When comparing between the groups on correlation coefficients (significant) of like co-variant variables, the between-group differences are significant for all (all p values<.05), except for the between-group comparison on $\text{swing}_{Y(A\text{-}P)}$ persistence with the $\text{stance}_{Z(Vertical)}$ SD. [Comparing independent r values was performed by taking the difference of their log normalised r scores and dividing by the combined standard error, e.g. (Field, 2009)]. If the elderly person has a variable system, they demonstrate an association with higher persistence of the three effectors anterior-posterior components at MTC. The young group in contrast demonstrate that higher persistence in the anterior-posterior effector components less likely to be associated with high effector variability. The highest correlation between the DFA and SD is found in the vertical component of the stance effector for both groups and this is a significant between-group difference (r_Y=.60, r_E=-.49, p<.001). The negative correlation of the elderly indicates that reduced persistence-likelihood (DFA) in the stance effector is associated with higher variability of the vertical swing hip position.

Finally, the persistence-likelihood found in the anterior-posterior direction of the three effectors shows significant correlations (r>.61) than the other two dimensions of medio-lateral and vertical. For the vertical direction, the stance and swing effectors have a significant correlation for the young of r_Y=.68, and for the elderly of r_E=.27. This between-group difference is significant, p<.05. This indicates that the young group have a significantly higher correlation for persistence-likelihood between the stance and swing effectors (for vertical dimension) at MTC.

These above results will be interpreted in greater depth in section 8.6.

5.10 Chapter 5 results summary

Below are the results related to the null hypothesis listed at the beginning of the chapter.

5.10.1 Hypotheses of persistence and control policies of effector trajectories

- Null hypothesis 5.1 There will not be a significant difference in persistence (DV) between age groups (IV) when comparing within effector (stance, swing, combined), dimension (medio-lateral, anterior-posterior, vertical), and state-relevant time (in sub-phase regions MX1 and MTC).
    - The null hypothesis was rejected for persistence in the vertical dimension of the combined-effector during MTC, and stance-effector preceding MTC. The elderly have significantly higher persistence in the combined-effector for the non-dominant limb ($p<.05$). The elderly have significantly reduced persistence in the stance-effector non-dominant limb ($p<.05$) and dominant limb ($p<.1$).
    - The null hypothesis was rejected for persistence in the anterior-posterior direction. The elderly have significantly increased persistence during MX1 of the stance-effector in the non-dominant limb ($p<.05$) and dominant limb ($p<.1$).
    - The null hypothesis was rejected for persistence in the medio-lateral direction. The elderly have significantly increased persistence in the swing-effector at MTC ($p<.05$).
- Null hypothesis 5.2 Effector persistence (DV) will not be significantly different within the three effectors (IV-1) between swing phase states MX1 and MTC (IV-2; repeat measures) within age groups (IV-3).
    - The null hypothesis was rejected for all within effector comparisons of the non-dominant limb in both groups. Effector persistence increased significantly between MX1 and MTC ($p<.05$).

- The null hypothesis was rejected for the dominant limb in the young group for the stance effector ($p<.05$). There was significantly increased persistence between MX1 and MTC.
- The null hypothesis was rejected for the dominant limb in the elderly group for combined effector. There was significantly increased persistence between MX1 and MTC.

♦ Hypothesis 5.3 First, effector persistence (DV) will not be significantly different between the three effectors (IV-1) when comparing within swing phase states MX1 and MTC (IV-2). Second, and in association with previous null-hypothesis, there will not be a significant interaction of 'effector' (IV-1) and 'group' (IV-3) on persistence (DV) when averaged across 'state-relevant times' within separate swing-phase regions of MX1, MX1-MTC and MTC.

- First, the null hypothesis was rejected when comparing between effectors within groups, at MX1 and MTC states. The null hypothesis of no significant between-group difference was rejected. When compared to the stance effector, the elderly group demonstrated significantly higher persistence in the swing effector and combined effector when compared at MTC, within both the dominant and non-dominant limbs. Both groups demonstrated significantly higher swing effector persistence, relative to stance effector persistence, when compared at MX1.
- Second, the null hypothesis was rejected for between effector persistence on 'averaged' effector persistence in two regions, MX1-MTC and MTC. The trends were the same in both regions. The elderly group stance-effector has significantly reduced persistence when compared to the combined-effector ($p<.05$). Compared to the young group, the elderly group has significantly increased persistence in the combined effector, when compared to the swing effector ($p<.05$).

- Null hypothesis 5.4 There will be no significant difference between young and older adults (IV-1) when comparing persistence (DV) changes between effectors (IV-2) between ‘moment-to-moment’ state-relevant times (IV-3; repeated measures).
    - The null hypothesis was rejected for ‘moment-to-moment’ effects on persistence between states of the MX1 region. The groups are significantly different when comparing between two levels of effector comparisons: stance-swing (Level 1); and combined-swing (Level 2). The stance effector persistence was lower compared to the swing effector, and persistence was significantly reducing for the elderly group between ‘state-relevant times’ MX1-12% to MX1-6%, and also between MX1 to MX1+6%. The combined effector had significantly increased relative to the swing effector persistence for MX1-12% and MX1-6%, and also MX1-6% and MX1, when comparing the elderly group to the young group.
- Null hypothesis 5.5 There will not be significant correlations between effector systems and their dimensions on ‘control policy’ statistics of DFA (persistence), SD (standard deviation, variance), and skewness. There will not be significant differences between young and older adults when comparing significant correlations.
    - The null hypothesis was rejected for many co-variant statistics.
    - The null hypothesis of no between group differences was rejected for many co-variant statistics.
    - The list of these rejected null-hypotheses is complex, and the reader is referred to section 5.9 for details.

Chapter 6

Coordination within the stance and swing effectors

6.1 Background

Statistics of persistence, variance and skewness provided evidence of control policies applied to the states of the effector trajectories at task-relevant points along the swing-cycle. To obtain further insight into the management of the effector trajectories, the co-variance of the within-effector segments that lead to stable effector trajectories will be examined in this chapter. If an effector trajectory is an important parameter to be controlled specific to a task, such as achieving a nominal toe-clearance state, then the segment components within the effector will need to be managed as a group by the loco-sensorimotor controller. Sharing patterns of redundant within-effector segments relevant to achieving a nominal end-effector state introduces the concept of synergies.

A synergy can be defined as a neural organisation of a set of elemental variables which have the purpose of stabilising (or destabilising) a hypothesised control variable (Latash et al., 2008). There are three characteristics that define a synergy: a redundant set of elements that share in the contribution to a task outcome; high stability of the control variable due to strong co-variation of elements; and flexibility to change from one stable co-variation pattern to an alternate pattern with equally stable outcome

(Latash et al., 2008). In this thesis, the term synergy applies to the set of segment positions and orientations that make up the configuration of the stance and swing effectors to stabilise the end-effector position, e.g. vertical position of the swing hip and toe respectively.

The Uncontrolled Manifold (UCM) hypothesis quantifies synergies from non-synergies. The hypothesis expects that trial-to-trial fluctuations of within-effector co-varying elements (i.e. elemental variables), considered to be relevant to the effector's task goal, are 'managed' by a central nervous system control policy. This is to ensure that the important task-goal of the effector remains relatively un-affected, i.e. improving task-goal stability. Similarly, the hypothesis expects that fluctuations of co-varying elemental variables of the effector that have limited relevance to the task goal (i.e. task-irrelevant) can be afforded a control policy of minimal intervention. Therefore, in the space of co-varying elemental variables there is a hypothesised observable structure to the fluctuations that resembles points clustering about a manifold. This manifold in element-space reflects how the control system manages the task variable. The task-irrelevant space of co-varying elements lies along the manifold plane and reflects the redundancy in the task variable (i.e. the uncontrolled manifold). The co-variance of fluctuations occurring in this space is called goal-equivalent variance (GE_V). The outcome is a stable task variable with a high tolerance for internal perturbations. Alternatively, the points orthogonal to the manifold plane reflect task-relevant fluctuations of the co-varying elements. The outcome is a less stable task variable. The covariance of fluctuations in this space is called non-goal equivalent variance (NGE_V).

The structuring of the covariance between the segments that are within the stance and swing effectors were investigated at state-relevant time slices within the MX1 and MTC regions. The UnControlled Manifold (UCM) hypothesis proposes that if the effector is performance-relevant and controlled, the GE_V will be 'significantly' greater than the NGE_V. The ratio between the GE_V and the NGE_V is termed a synergy parameter, such that when the ratio is significantly greater than 1, element co-variance is hypothesised to reflect a synergy (i.e. $UCM_{ratio}>1$, $p <.05$). With respect to the stance and swing effectors controlling the task of toe-clearance, the elements reflect the limb segments. Therefore, for a synergy to occur in the stance or swing effectors, segment co-variance will be spread mostly along the UCM (i.e. goal-equivalent variance GE_V), relative to the spread of segment co-variance distributed orthogonal to the UCM (i.e. non-goal equivalent variance, NGE_V). The further away UCM_{ratio} is from 1 demonstrates that there is a stronger synergy present to stabilise the task variable.

6.2 Method

The vertical task of the stance and swing effectors were investigated because they have relevance to a performance-variable in the vertical state of the combined effector, i.e. toe-to-ground clearance. The other effector task variables investigated were the medio-lateral task of the stance effector and the anterior-posterior task of the swing effector. These two additional task variables provided a contrast with the performance-relevant vertical task-variables. This allows a insight between the synergies formed to control medio-lateral stability and vertical stability of the swing-hip position during the swing-cycle. The decision to investigate the anterior-posterior component of the swing leg was because this task variable has relevance to the commonly investigated gait parameter of step-length.

The UCM hypothesis does not require temporal order of cycles but it does presume that from trial-to-trial the effector repeatedly performs the same task goal from the same initial conditions. Therefore, the stance and swing effectors each needed separate time-series relevant to their effector configuration at initial conditions and similar response behaviours. Specifically, this was the respective effector's vertical state at MX1 and the response in the effector's vertical displacement between MX1 and MTC. For n swing-cycles occurring within each limb, the swing-cycle trials were rank ordered according to an initial condition defined by the effector's (stance or swing) vertical displacement at MX1. This forms a newly ordered time series of swing-cycle trials which underwent a sub-division into three groups representing a (i) low-tercile (sub-average initial conditions), (ii) mid-tercile (average initial conditions), and (iii) upper-tercile (higher than average initial conditions). Within each of the three sub-groups, a secondary condition provided a new ranking order and this was based upon effector response to the initial condition it was faced with at MX1. This response was defined by the vertical displacement made by the effector (stance or swing) between MX1 and MTC. So, within each of the three initial condition 'terciles', there was a new 'within-tercile' ordering based upon the response taken. This new ordering then underwent a sub-division into groups of (i) risky response (lower-tercile), (ii) average response (mid-tercile), (iii) safe response (upper-tercile).

Each of the nine subgroups within each analysis process provided approximately sixty cycles for computing the UCM components of variance. The results from the analyses form the body of work in this chapter.

6.2.1 Dependent variables

The dependent variables used for testing hypotheses were the UCM components of variance: the degrees of freedom normalised (see section 3.6.2.4) goal equivalent variance (GEv); the degrees of freedom normalised non-goal equivalent variance (NGEv). The third dependent variable was the synergy index, ‘UCM_{ratio}’, calculated by GE_V/NGE_V.

6.2.2 Independent variables

The first independent variables were the nine condition-response groups: strong, moderate or weak within the MX1 initial condition of low, average, or high. These were explicitly: low-weak, low-moderate, low-strong, average-weak, average-moderate, average-strong, high-weak, high-moderate, high-strong. Another independent variable related to these groups were derived from concatenated trials of like-responses into one group. This doesn’t assume the same initial condition, but it assumes similar responses of strong, moderate and weak type. This process was done for analysis simplicity. This sub-division could arguably be based upon the initial condition or the response from the initial condition. The selecting of response type versus the MX1 initial condition is that those trials assigned to an initial MX1 condition can potentially diverge as the trajectory evolves. Grouping the trials with respect to the MX1-to-MTC response will incur trials with different starting points, however, their trajectory states will be more similar between MX1 and MTC. So, the response type classification is selected because it is most likely to represent trials which have similar trajectories.

The second independent variable was the state-relevant time-slices of the swing-cycle: MX1 (MX1-12%, MX1-6%, MX1, MX1+6%); MX1-MTC (MX1+6%, MX1+12%, MTC-12%, MTC-6%); and MTC (MTC-6%, MTC, MTC+6%, MTC+12%).

6.2.3 Statistical procedure

The specific statistical designs are listed under the null hypotheses in the next section. Most of the statistical analyses involve a three-way mixed-design ANOVA (with repeated measures on time-slice). This allowed a 'dynamic' exploration of the interactions of response (3) and time-slice (4) on the dependent variable of 'effector synergy', or UCM_{ratio}. Significance was interpreted with respect to the assumption of homogeneity which was presumed if it passed Levene's Equality of Variance significance test.

In many dependent variable cases, due to the re-sampling of the 'effector response' variable, there were positively skewed data distributions. Data transformations were performed for positively skewed data by taking the natural log of the raw score plus 1, i.e. $Z_{trans} = Ln(x_i)$, where x_i represents the participant 'raw' score. Likewise, for negatively skewed groups of data, the data scores were transformed by taking the square root of the 'raw' score. For making interpretation meaningful, figures are based upon the pre-transformed scores, but the transformed scores were used for computing the significance and effect sizes. If Mauchley's test of sphericity failed, significant interactions ($p < .05$) were assessed using the Greenhouse-Geisser and the Huynh-Feldt corrections where appropriate (Field, 2009).

Simple effects were carried out by planned comparisons (planned contrasts) on variables that provided information believed to be of the most value. Two levels of planned comparisons for response-type were made by comparing each-response against the moderate-response; i.e. comparing with the weak-response (level 1) and strong-response (level 2). Three levels of planned comparisons for time-slice were based upon consecutive time-slice comparisons; i.e. for the MX1 region level 1 is MX1-12% compared with MX1-6%, level 2 is MX1-6% compared with MX1, and level 3 is MX1 compared with MX1+6%.

Like previous statistical designs from Chapters 4 and 5, some analyses took a degree of liberty by considering the dominant and non-dominant limbs as independent cases (i.e. separate participants).

6.2.4 Hypotheses and statistical design

This chapter investigates the following null-hypotheses:

- Hypothesis 6.1 The UCM, UCM_{ratio} (DV) will not be significantly different from the value of 1 when compared at each state-relevant time-slice of the swing-cycle.
    - Tests were made within the four effector task variables (SW_V, SW_{A-P}, ST_V, ST_{M-L}; IV), limb (dominant and non-dominant; IV), age (young, older; IV).
    - T-test on UCM_{ratio} for significance of mean ± 95% group confidence interval > 1.
- Hypothesis 6.2 There will not be a significant difference in age (IV) on the effector synergy parameter (UCM_{ratio}; DV) across the investigated swing-cycle region.
    - Statistical analysis using multiple 'independent-sample' t-tests at each time-slice, within each limb type (dominant and non-dominant). Significance is set at $p<.1$.
    - Main effect of age from three-way mixed-design ANOVA. Significance is set at $p<.05$.
- Hypothesis 6.3 The synergy parameter, UCM_{ratio} (DV), will not be significantly different between the effector response types (IV; weak, moderate, strong). There will be no significant interaction of state-time and age on UCM_{ratio}.
    - The dependent variable is the UCM_{ratio} within the four task variables (SW_V, SW_{A-P}, ST_V, ST_{M-L}; IV)
    - The independent variables, 'state-time' will be related to the three regions of investigation [MX1 (MX1-12%, MX1-6%, MX1, MX1+6%); MX1-MTC (MX1+6%, MX1+12%, MTC-12%, MTC-6%); and MTC (MTC-6%, MTC, MTC+6%, MTC+12%]

- Statistical analysis using a three-way mixed design ANOVA. Main effect of 'state-time' and an interaction effect of 'state-time' and age with planned contrasts (simple interaction effects).
- Separate tests were run for the three time regions of the swing-cycle: MX1, MX1-MTC, and MTC.
- Separate tests were run for the four effector task variables.
- Limb data was pooled.

♦ Hypothesis 6.4 The synergy parameter, UCM_{ratio} (DV), will not be significantly different when comparing between state-relevant time-slices (IV). There will be no significant interaction of 'response' and age on UCM_{ratio}.

- The dependent variable is the UCM_{ratio} within the four task variables (SW_V, SW_{A-P}, ST_V, ST_{M-L}; IV)
- The independent variables, 'state-time' will be related to the three regions of investigation [MX1 (MX1-12%, MX1-6%, MX1, MX1+6%); MX1-MTC (MX1+6%, MX1+12%, MTC-12%, MTC-6%); and MTC (MTC-6%, MTC, MTC+6%, MTC+12%]
- Statistical analysis using a three-way mixed design ANOVA. Main effect of 'state-time' and an interaction effect of 'state-time' and age with planned contrasts (simple interaction effects).
- Separate tests were run for the three time regions of the swing-cycle: MX1, MX1-MTC, and MTC.
- Separate tests were run for the four effector task variables.
- Limb data was pooled.

♦ Hypothesis 6.5 The synergy parameter, UCM_{ratio} of the task variable (DV), when compared between effector responses (IV-1), will not be significantly different when compared between state-dependent time-slices (IV-2). The older adult group will not be significantly different to the young group (IV-3) when comparing the interaction effect of response (IV-1) and state-time (IV-2) on the UCM_{ratio} of the effector task variable (DV).

- Statistical analysis using a three-way mixed design ANOVA. Interaction effect of ‘response’ x ‘state-time’ and an interaction effect of ‘response’ x ‘state-time’ x ‘age’. Planned contrasts will explore simple interaction effects.
- Separate tests were run for the three time regions of the swing-cycle: MX1, MX1-MTC, and MTC.
- Separate tests were run for the four effector task variables.
- Limb data was pooled.

6.3 Inspection of the UCM components of variance in the stance and swing effector systems

Figures 6.3.1-3 below illustrates the mean curves of the three UCM statistics (GE_V, NGE_V and UCM_{ratio}) displayed across time-slices of the swing-cycle. The UCM variance was obtained for each effector task variable from the same swing-cycle trials. This data was derived from the sub-grouping of trials derived from the details of the combined effector states at MX1, and displacements between MX1 and MTC. This method provides no explicit control over the trial-to-trial consistency expressed by either the stance or swing effectors on the basis of their independent initial MX1 condition and their MX1-MTC response. However, Figures 6.3.1-3 reflect features between effector task variables across the swing-cycle that are generally revealed when applying the alternate trial grouping method based upon stance and swing effector states. The 'alternate' trial grouping method outlined in section 6.4 will be applied in the subsequent sections where statistical analysis tests are applied.

Figure 6.3.1 displays the curves relevant to the moderate response-type. Figures 6.3.2 and 6.3.3 display the curves for the weak and strong response-types, respectively. Each figure is grouped into two row panels and three column panels, according to limb type (row panels), and the initial state of the combined effector at MX1 (column panels).

The differences between the three figures (comparing between MX1-MTC response) and between the column panels (comparing between initial MX1 state) are very subtle. The differences between effectors are more obvious and consistent. For example, Figure 6.3.1 displays the moderate MX1-MTC response for three groups of initial MX1 states (low, average, high). For the third row of the top row-panel (UCM_{ratio} of dominant limb) and the third row of the bottom row-panel (UCM_{ratio} of non-dominant limb), there is a higher swing effector task variable of the vertical direction (blue curve) compared to the swing effector task variable in the anterior-posterior direction (green curve). There are some subtle differences between MX1 and MTC regions. There are some subtle differences between the young (solid line) and elderly (dashed line). A general observation is that the vertical swing effector (blue curve) reaches a maximum (where GE_V/NGE_V, or $UCM_{ratio} < 2$) near the time-slices of MX1+12% and MTC-12%; i.e. the transition phase between MX1 and MTC regions. In contrast, the anterior-posterior swing effector (green curve) reaches a minimum (where GE_V/NGE_V, or $UCM_{ratio} < 1$).

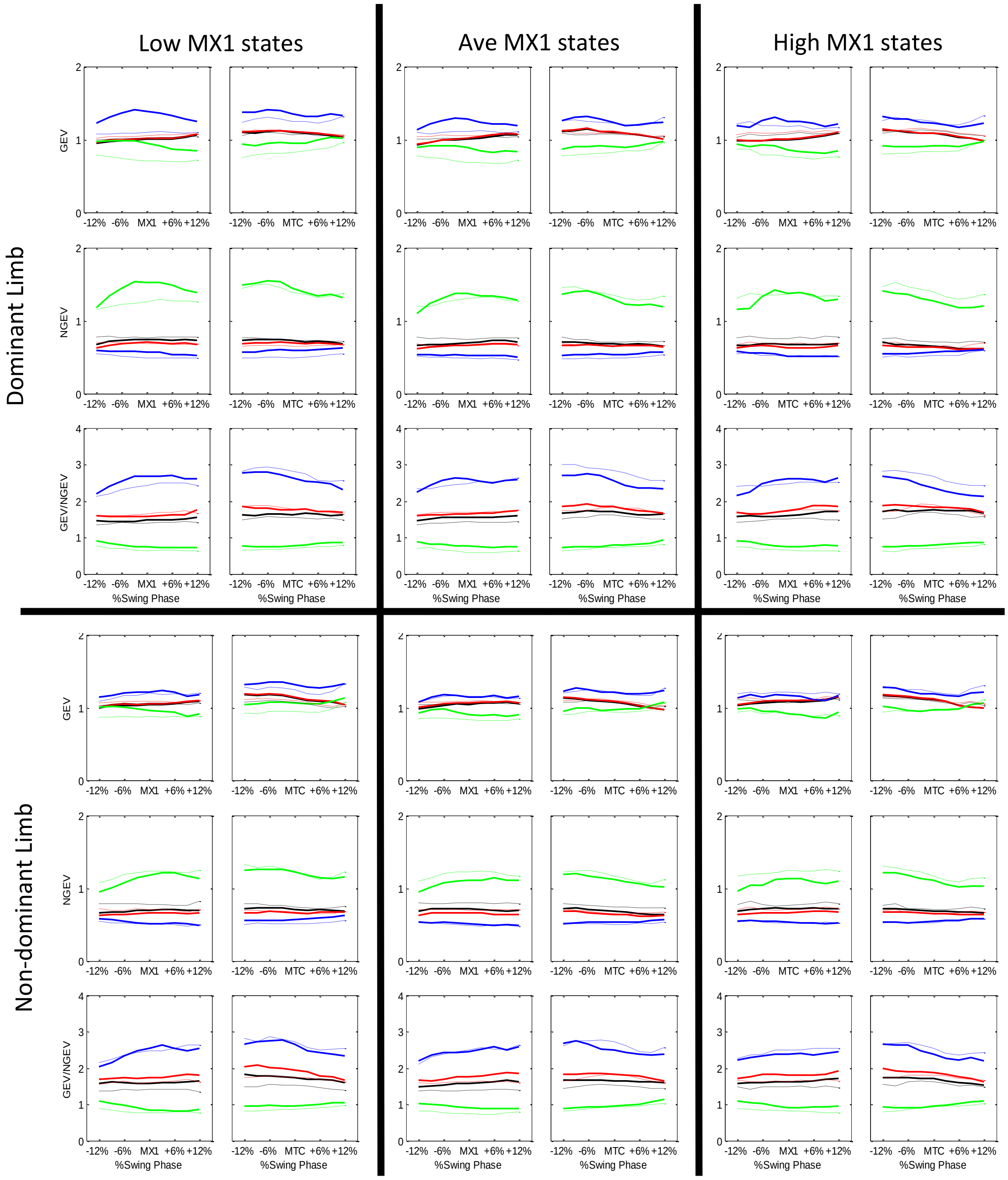


Figure 6.3.1.

Variables of the UnControlled Manifold (UCM) associated with trials grouped by the moderate combined effector response between MX1 and MTC, that are also associated with initial vertical states at MX1 of lowMX1, averageMX1 and highMX1. Young group (solid lines), elderly group (dashed lines) for effector variables $Stance_{M\text{-}L}$ (black), $Stance_{V}$ (red), $Swing_{A\text{-}P}$ (green), $Swing_{V}$ (blue). The upper three row panels relate to the preferred limb, the bottom three rows relate to the non-preferred limb. The rows of each respective limb, represent from top to bottom the V_{GE}, V_{NGE} and GE_{V}/NGE_{V} (i.e. UCMratio). The horizontal axis is the time-slice of the swing-cycle region MX1 and MTC. Column-panels represent initial MX1 states of low, average and high.

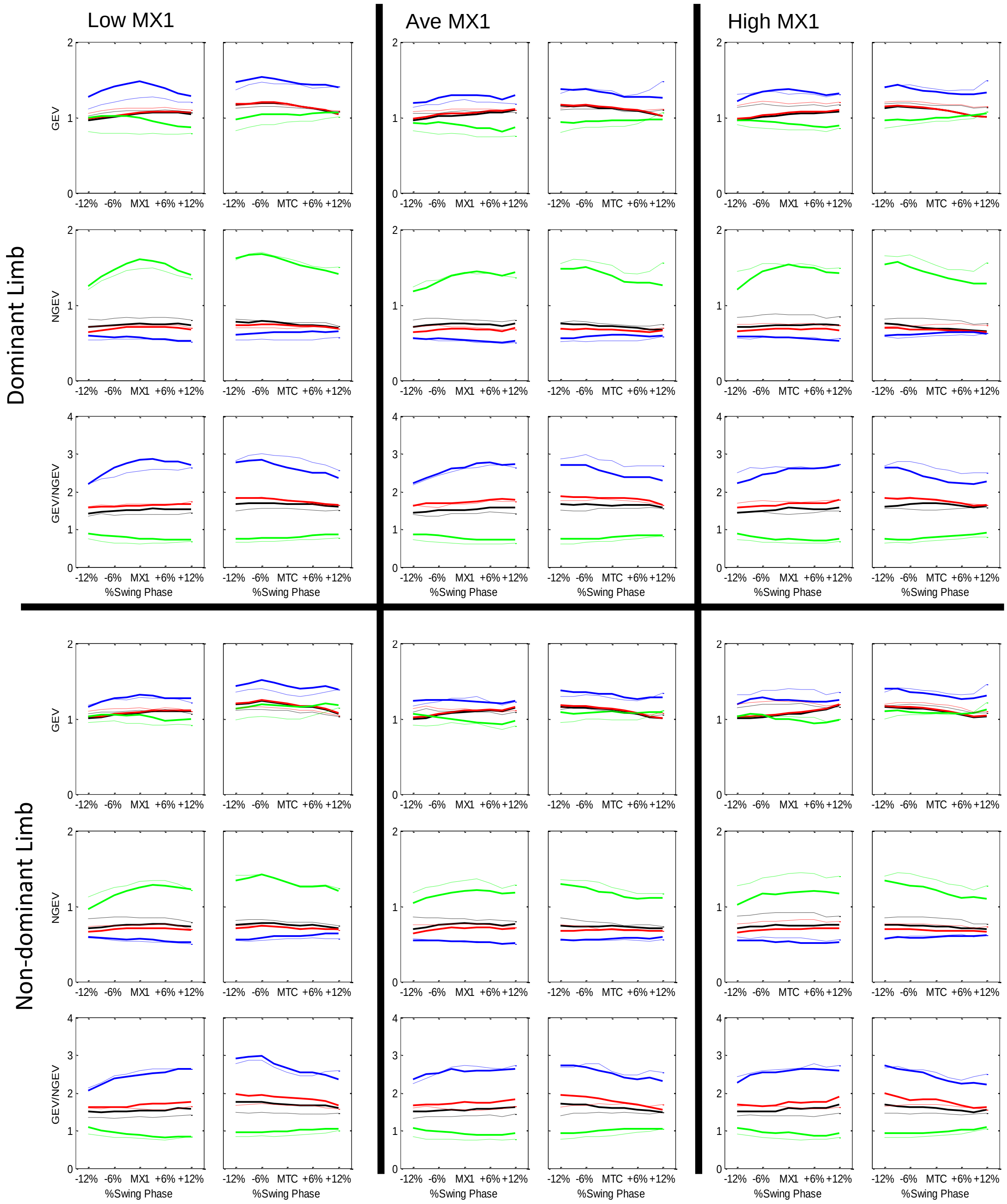


Figure 6.3.2.

Variables of the UnControlled Manifold (UCM) associated with trials grouped by the weak combined effector response between MX1 and MTC, that are also associated with initial vertical states at MX1 of lowMX1, averageMX1 and highMX1. Young group (solid lines), elderly group (dashed lines) for effector variables $Stance_{M\text{-}L}$ (black), $Stance_V$ (red), $Swing_{A\text{-}P}$ (green), $Swing_V$ (blue). The upper three row panels relate to the preferred limb, the bottom three rows relate to the non-preferred limb. The rows of each respective limb, represent from top to bottom the V_{GE}, V_{NGE} and GE_V/NGE_V (i.e. UCMratio). The horizontal axis is the time-slice of the swing-cycle region MX1 and MTC. Column-panels represent initial MX1 states of low, average and high.

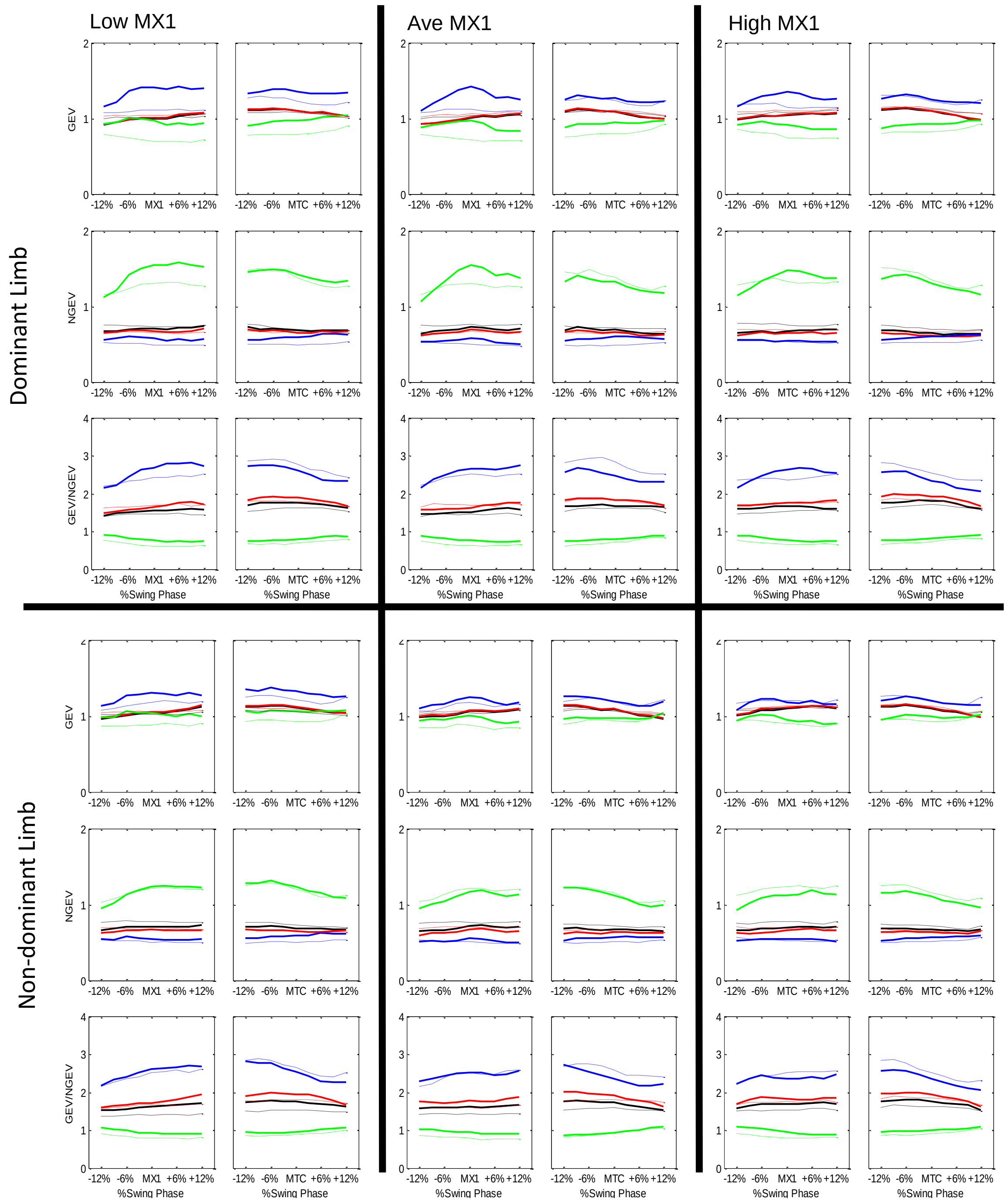


Figure 6.3.3.

Variables of the UnControlled Manifold (UCM) associated with trials grouped by the strong combined effector response between MX1 and MTC, that are also associated with initial vertical states at MX1 of lowMX1, averageMX1 and highMX1. Young group (solid lines), elderly group (dashed lines) for effector variables Stance$_{M-L}$ (black), Stance$_{V}$ (red), Swing$_{A-P}$ (green), Swing$_{V}$ (blue). The upper three row panels relate to the preferred limb, the bottom three rows relate to the non-preferred limb. The rows of each respective limb, represent from top to bottom the V$_{GE}$, V$_{NGE}$ and GE$_{V}$/NGE$_{V}$ (i.e. UCMratio). The horizontal axis is the time-slice of the swing-cycle region MX1 and MTC. Column-panels represent initial MX1 states of low, average and high.

The following observations relate to comparing Figure 6.3.3 (strong MX1-MTC response) with Figures 6.3.1-2 (moderate and weak MX1-MTC response), with focus on the top rows of the row-panels (i.e. GE_V for both dominant and non-dominant limbs). For the strong MX1-MTC response in Figure 6.3.3, at MX1 regions the GE_V of the vertical swing effector (blue curve) is higher than the elderly for low, average and high MX1 states. This discrepancy between groups is less evident for the non-dominant limb, and the discrepancy is also less obvious when comparing against the dominant limb of Figures 6.3.1 (moderate response) and 6.3.2 (weak response). These same observations are generally evident when observing the anterior-posterior swing effector (green curve) middle row within the row-panels (i.e. the NGE_V).

The above observations will be explored by grouping trials on the basis of stance and swing effector states between MX1 and MTC, rather than the states presented above from the combined effector. This method will provide more accurate details of the behaviour of the effector task variables. The statistical analyses will now follow.

6.4 Evidence of 'synergies' in the stance and swing effector systems

Figures 6.4.1A-D shows that the UCM_{ratio} is significantly greater than 1 (UCM_{ratio} > 1, t(55) > 10, p < .001) for all task variables except the swing-effector in the anterior-posterior direction. Figures 6.4.1A-C can all be considered controlled task-variables under the UCM hypothesis. The vertical task of the swing effector has the largest UCM_{ratio} and can therefore be implied to have the stronger synergy for stabilising the task variable of vertical effector length. The medio-lateral task of the stance effector has a lower UCM_{ratio} in comparison to the vertical task of the stance effector. For the anterior-posterior task of the swing effector (Figure 6.4.1D), the UCM hypothesis suggests that it is not a controlled task variable, because UCM_{ratio} is significantly less than 1 (i.e. mean difference is below 1), particularly between MX1-6% and MTC+6% (t(55) > 10, p < .001).

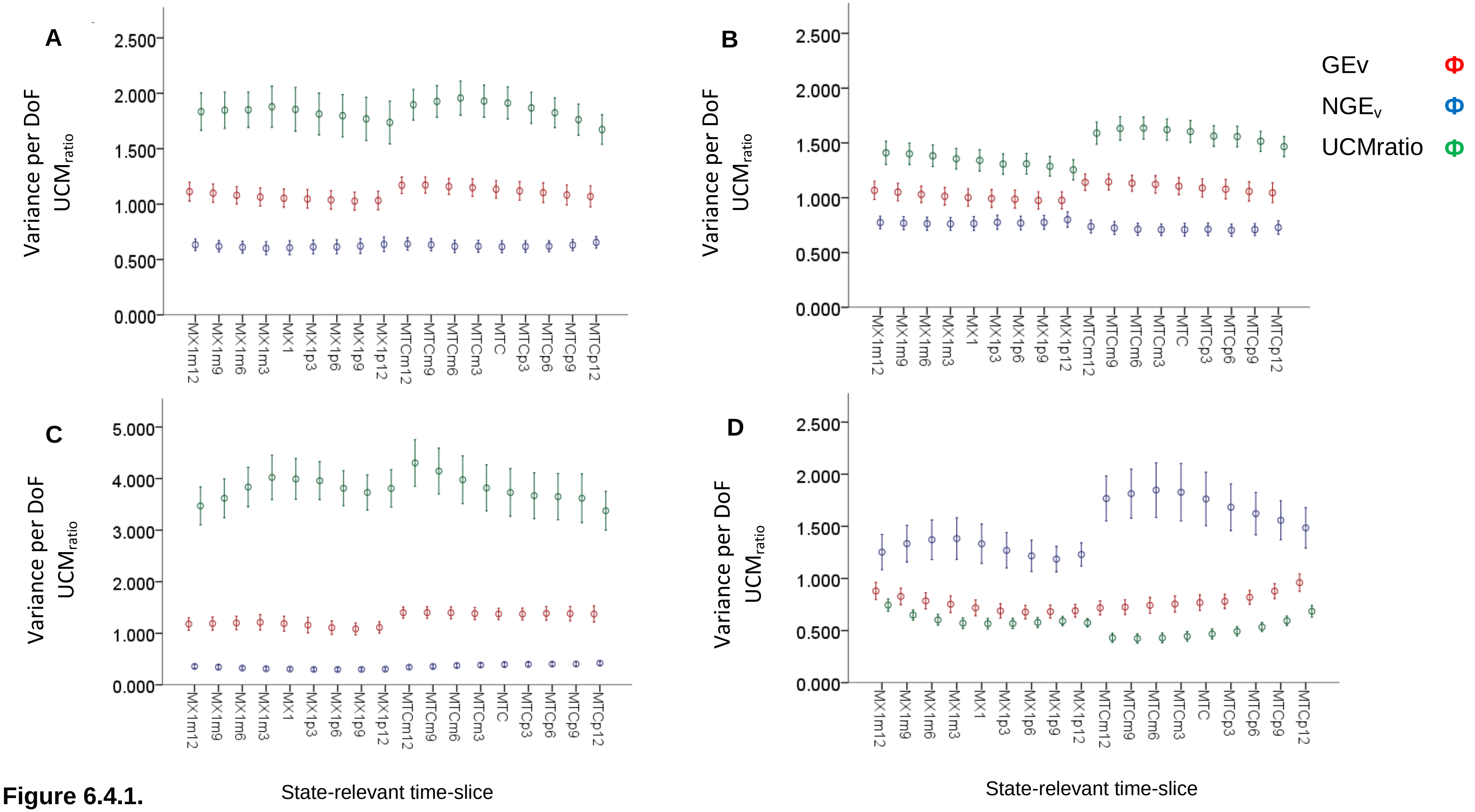


Figure 6.4.1.

An example of data taken from the elderly group demonstrating the different UCM components of variance for the effector performance variables: **A**) stance effector in the vertical direction; **B**) stance effector in the medio-lateral direction; **C**) swing effector in the vertical direction; **D**) swing effector in the anterior-posterior direction. Error-bars represent group mean and 95% confidence intervals. There is an observable time lag between UCM statistics for graphical display purpose. All three UCM statistics were obtained from the same time-slice.

6.5 Evidence of ageing differences in the 'synergies' of the stance and swing effectors

Figures 6.5.1 – 6.5.4 show the 'across swing phase' profile of the dependent variable, UCM_{ratio}, for the condition of 'moderate response to vertical swing limb length change (or swing hip height change) between MX1 and MTC given the initial condition of average length (or swing hip height) at MX1. The moderate response from average initial MX1 state was selected for comparison because this represents the participants 'nominal' effector behaviour.

The vertical swing effector (Figure 6.5.1) reveals a significantly higher UCM_{ratio} in the elderly group at MTC-12% for the dominant limb, and for the non-dominant limb at MTC-12% and MTC-9%. Based upon the lowest limit of the 95% error-bars, both groups have UCM_{ratios} significantly ($p<.05$) higher than 1 across all time-slices of the swing-cycle. In contrast, Figure 6.5.2 shows the UCM_{ratio} in the swing effector task-variable of the anterior-posterior direction is significantly less than 1 for the dominant limb of both groups across all time-slices of the swing-cycle. Except for the extreme time-slices of the MX1-MTC region, the non-dominant limb also shows a significantly lower UCM_{ratio} difference from 1 in both groups. Figure 6.5.2 reveals a significantly ($p<.1$) lower UCM_{ratio} in the elderly compared to the young at time-slices prior to the MX1 state of the non-dominant limb, and time-slices prior to the MTC state for both the dominant and non-dominant limbs.

These results tell us that the shared co-variance pattern in the swing effector segments that control the swing effector states in the anterior-posterior direction, is significantly ($p<.05$) more unstable in the elderly prior to MTC. Stated another way, variations in the shared co-variance pattern will have a bigger affect on the end-effector position in the anterior-posterior direction. However, the toe-clearance relevant vertical dimension of the swing effector trajectory indicates that elderly control of the swing effector trajectory is significantly ($p<.1$) more stable prior to MTC. This indicates that the shared co-variance pattern of the swing effector segments prior to MTC are managed/structured significantly different in the elderly.

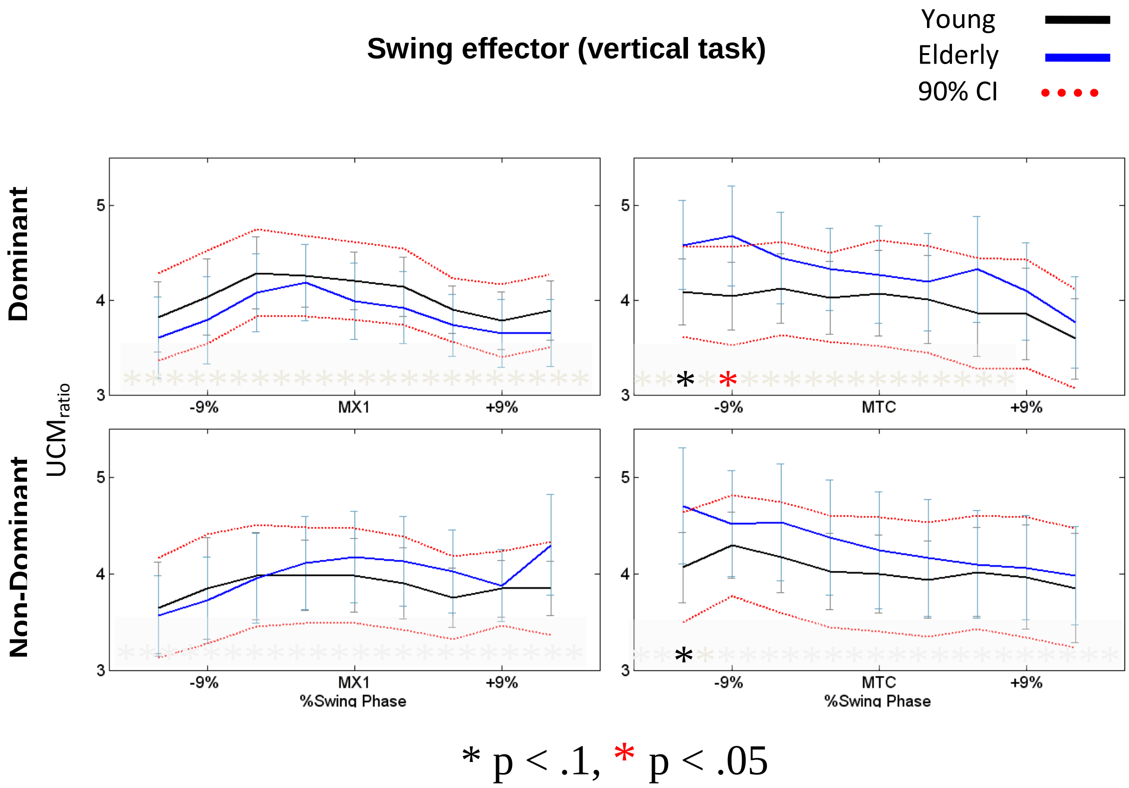


Figure 6.5.1.

UCM_{ratio} presented on the ordinate is for the swing effector in the vertical direction for both the dominant and non-dominant limbs. The comparison is made by the swing effector for the moderate response to vertical swing limb length change between MX1 and MTC given the initial condition of average vertical swing limb length at MX1. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

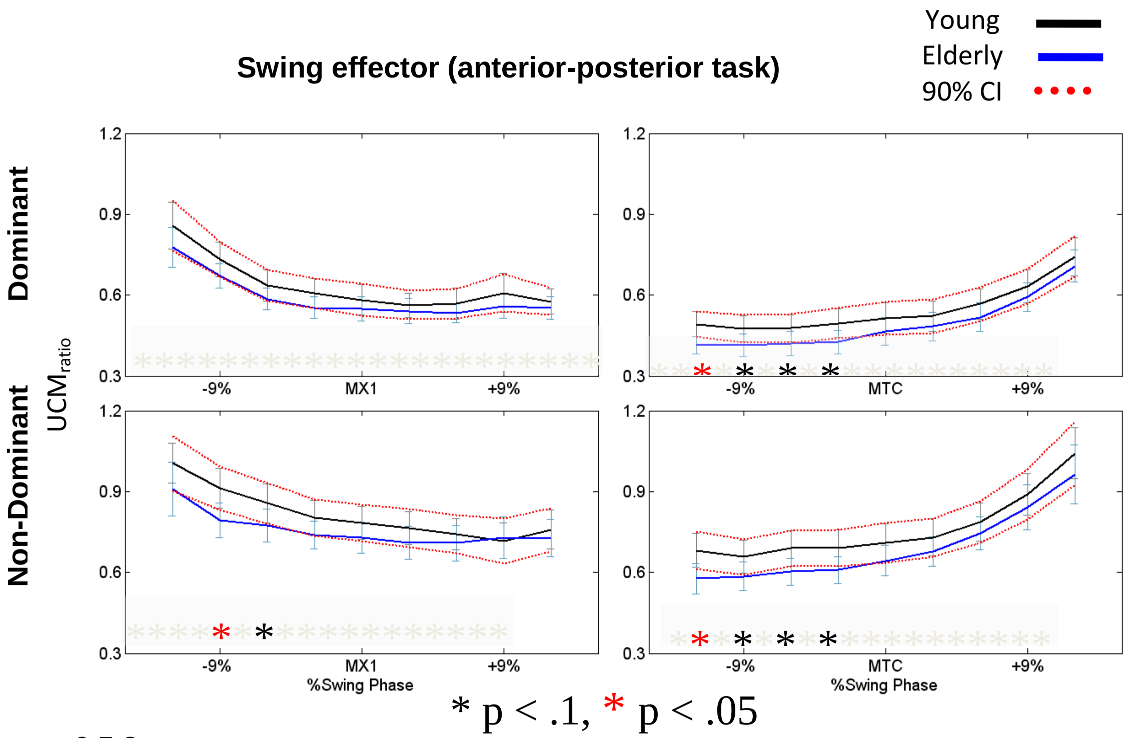


Figure 6.5.2.

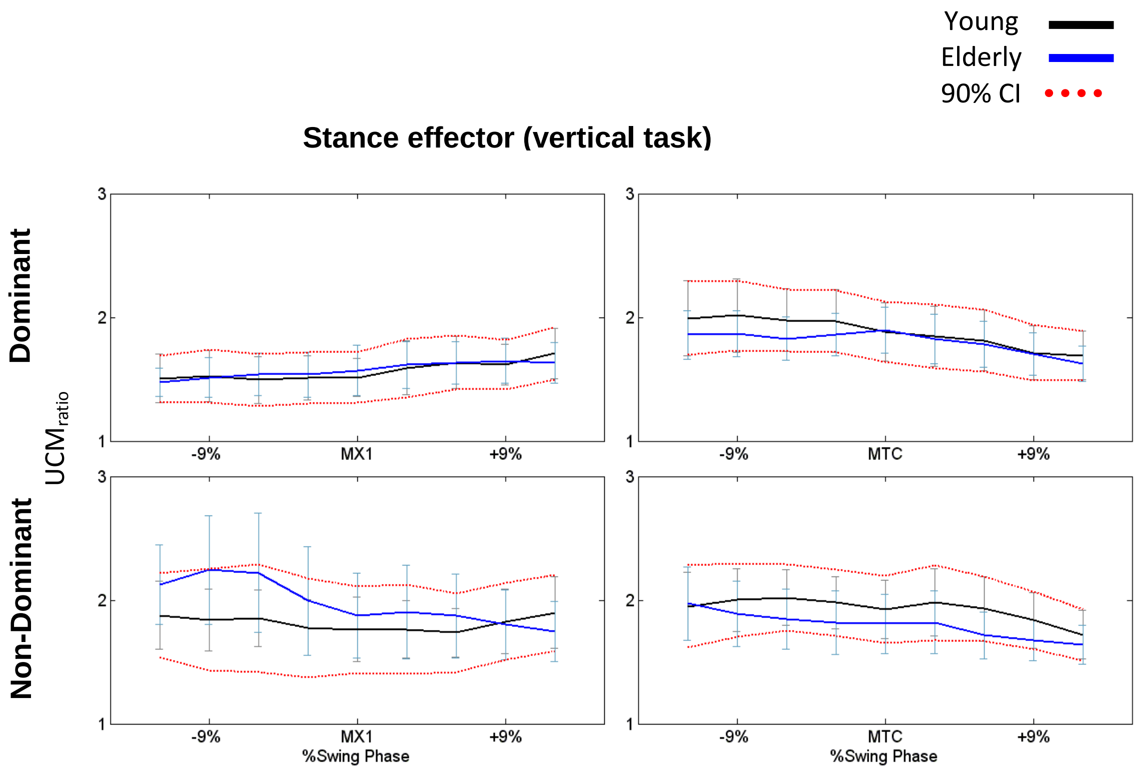
UCM_{ratio} presented on the ordinate is for the swing effector in the anterior-posterior direction for both the dominant and non-dominant limbs. The comparison is made by the swing effector for the moderate response to vertical swing limb length change between MX1 and MTC given the initial condition of average vertical swing limb length at MX1. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

The stance effector task variable in the vertical dimension is displayed in Figure 6.5.3. It shows a large inter-participant variability for the elderly non-dominant limb at the beginning of the MX1 region. Also, relative to the dominant limb, the young group appears to have larger inter-participant variability. There were no significant between-group differences in the stance effector task variable in the vertical direction.

Compared to the young, Figure 6.5.4 shows the elderly have a significantly ($p<.1$) lower UCM_{ratio} in the stance effector task variable in the medio-lateral direction. This is significant at the $p<.05$ level in comparisons made at most time-slices of the non-dominant limb, and at $p<.1$ for comparisons at time-slices MX1-3% and MTC-3% within the dominant limb. This indicates that the elderly have a significantly different UCM_{ratio}

in the stance effector medio-lateral direction. This tells us that the elderly's loco-sensorimotor controller's management strategy to structure effector variance, will be less able to stabilise the medio-lateral swing hip position if perturbations occur in the stance effector.

These areas will be further explored more thoroughly using a mixed-design ANOVA in the subsequent sections.



* p < .1, * p < .05

Figure 6.5.3.

UCM_{ratio} presented on the ordinate is for the stance effector in the vertical direction for both the dominant and non-dominant limbs. The comparison is made by the stance effector for the moderate response to vertical swing hip positional change between MX1 and MTC given the initial condition of average vertical swing hip height at MX1. Error bars represent the 95% confidence interval of the group mean (t(27)=2.052). A significant difference [t(54)>1.67, p<.1] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of p<.05 and p<.1.

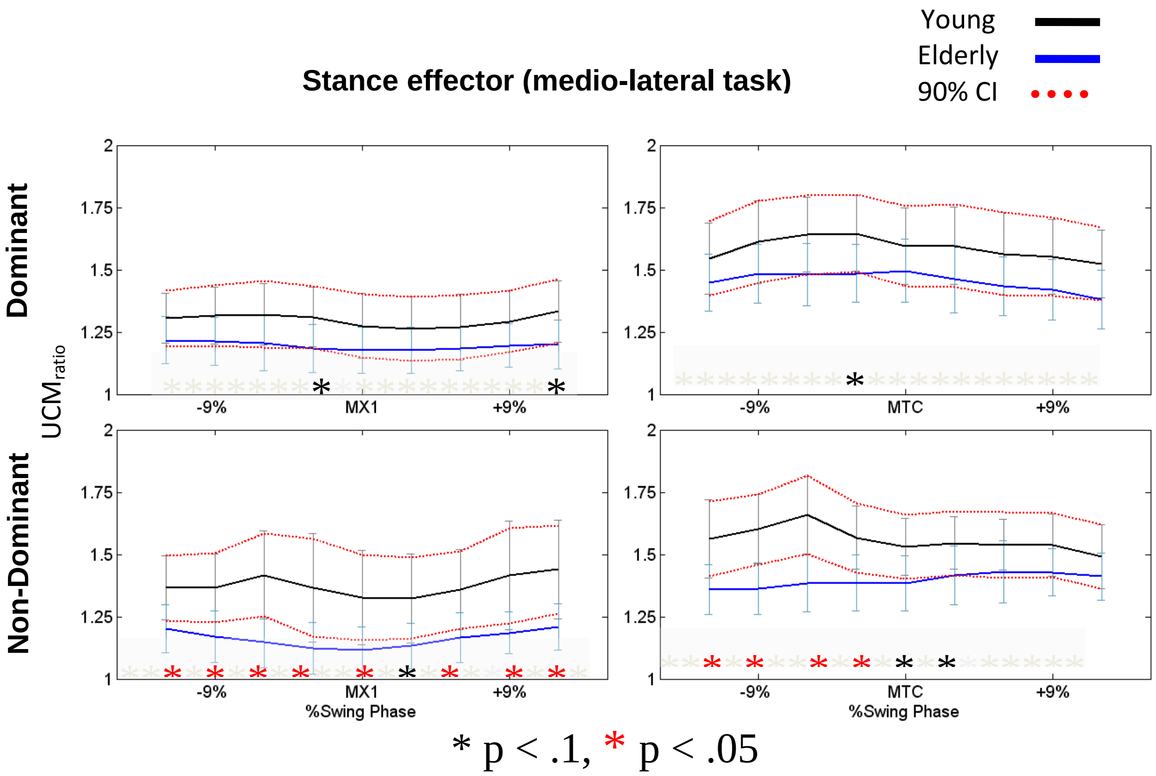


Figure 6.5.4.

UCM$_{ratio}$ presented on the ordinate is for the stance effector in the medio-lateral direction for both the dominant and non-dominant limbs. The comparison is made by the stance effector for the moderate response to swing hip positional change between MX1 and MTC given the initial condition of average swing hip height at MX1. Error bars represent the 95% confidence interval of the group mean ($t(27)=2.052$). A significant difference [$t(54)>1.67$, $p<.1$] is presumed when the elderly mean (blue curve) exceeds the ±90% confidence interval (red curve). The grey band with the red and black asterisk emphasises where time-states show between-group significant differences of $p<.05$ and $p<.1$.

6.6 Synergy effects due to state, response and age in the vertical direction of the swing effector

Figure 6.6.1 shows the effect of the different response types across the swing phase for the three UCM components of variance, which are contrasted between the two age groups. Line graphs are chosen for demonstrating the interaction effect of the dependent variable UCM_{ratio} on the independent variables of response, state-relevant time, and age group. A description of the graph features will precede the statistical analysis.

The graphs show that the 'weak' MX1-MTC response by the swing effector states is associated with a generally higher GE_V for both young and older adults. Also, in both groups the weak MX1-MTC response is associated with a generally higher response for NGE_V. The parameter of interest is the UCM_{ratio}, or GE_V/NGE_V (Figure 6.6.1C). This variable demonstrates that the moderate response in both the young and elderly groups have a higher UCM_{ratio} during the MTC region. During the same region, the UCM_{ratio} of the trials associated with a moderate response between MX1-MTC states contrasts with the lower UCM_{ratio} for strong effector response. This is also apparent in both groups. There is however a different profile for the elderly compared to the young for UCM_{ratio}.

Qualitative inspection of Figure 6.6.1C also shows a minima in the UCM_{ratio} that occurs at approximately MX1+9%, indicating that an increase in UCM_{ratio} of the stance effector vertical-task is occurring before the completion of the MX1 region. The timing of a maximum UCM_{ratio} occurs most notably at MX1-3% for both groups. A second UCM_{ratio} maximum occurs at approximately MX1+12% for the young group weak and strong responses. In contrast, the UCM_{ratio} associated with the moderate response trials of the young group peaks at approximately MTC-9%. For the elderly, the second maximum peak in UCM_{ratio} occurs at MTC-12% and MTC-9% for trials associated with the strong, moderate and weak responses.

In the next section, the general observations above will be explored with specific support from a three-way mixed design ANOVA.

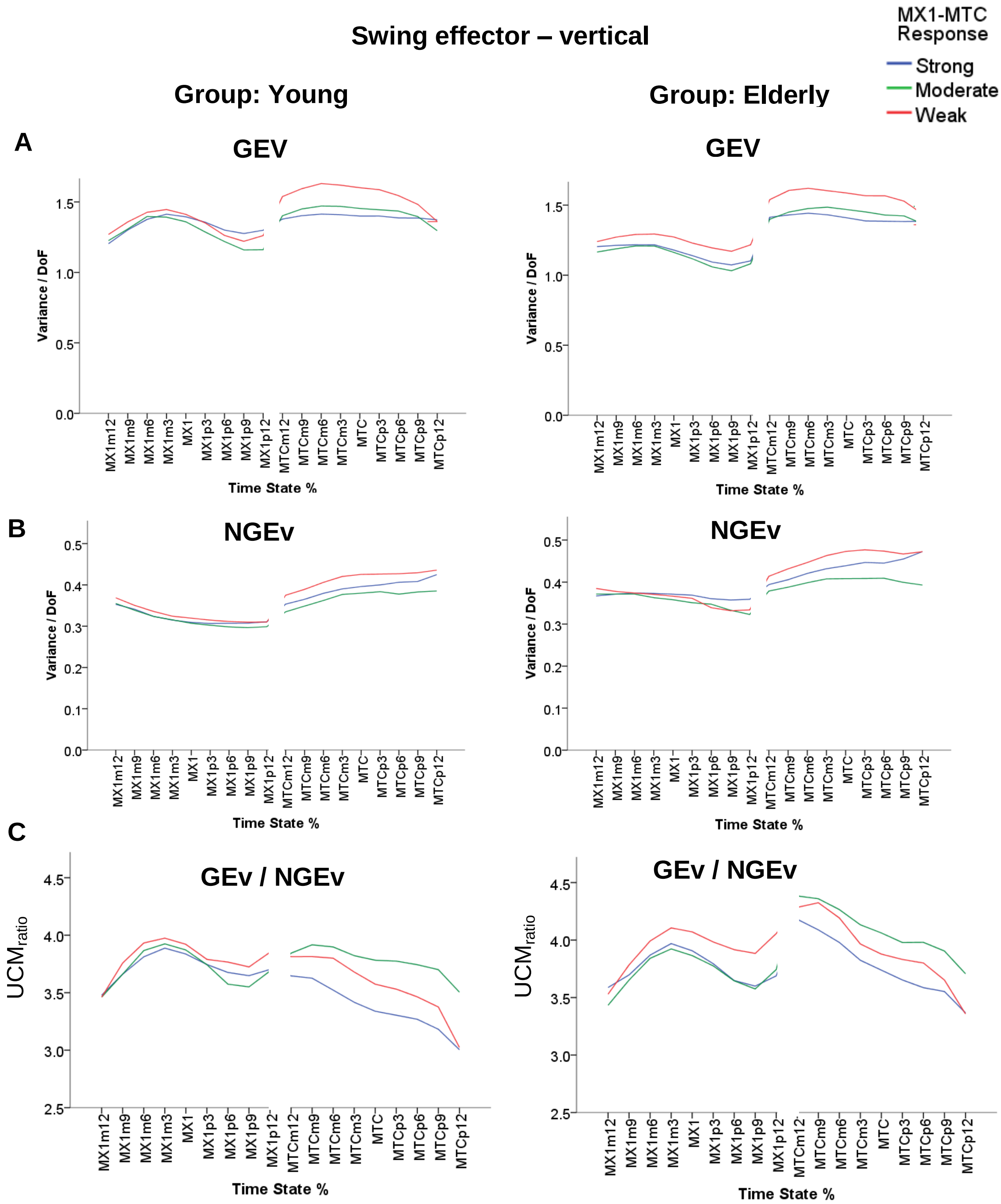


Figure 6.6.1.

UCM components of variance for the swing effector task variable of the vertical direction for both the young and elderly groups. Limb groups are collapsed into one group. This has a relatively small reduction effect on the 95% confidence error of the mean due to the increase in statistical degrees of freedom (i.e. *df* = N x 2). Comparisons are between weak, moderate and strong MX1-MTC responses across time-slices. The gap in the line graphs represents the change between MX1 and MTC regions. Except for this gap, the horizontal scale represents time-slices in 6% increments (of swing phase duration), where MX1+6% represents +6% of the swing phase post-MX1.

6.6.1 Synergy strength comparisons within the MX1-to-MTC transition state

This section will be set-out by investigating two areas related to the effect of independent variables (group, response-type and state-time) on the synergy parameter (UCM_{ratio}) found in the vertical task of the swing effector. First, determining the effect of 'effector MX1-MTC response-type' on the UCM_{ratio}. Second, determining the effect of 'state-relevant time' on the UCM_{ratio}. In following, there will be an investigation of whether these effects differ between-groups. This process was carried out by a three-way mixed design ANOVA (see section 6.2). The analysis collapsed the limbs into one group (i.e. doubling within group 'participants' and presuming independence). Analysis of within dominant and non-dominant limbs will be stated otherwise. All effects are reported as significant at the $p < .05$ level.

There was a significant main effect for effector response ($F(2, 220) = 29.4$, $p < .001$, $\eta^2 = .21$), indicating that synergy strength is associated with response type (e.g. Figure 6.6.1.1B). Contrasts revealed that the weak response is significantly higher than the moderate response [$F(1, 110) = 21.4$, $p < .001$, $\eta^2 = .16$] and the strong response was significantly lower than the moderate response [$F(1, 110) = 13.2$, $p < .001$, $\eta^2 = .11$]. Therefore, null hypothesis 6.3 is rejected because there is a significant difference ($p<.05$) when comparing between response types. The results indicate that the structuring of variance in the swing effector, for the vertical-task, is associated with the size of the vertical change in the swing effector state from MX1 to MTC. In comparison to synergies associated with moderate MX1-MTC responses, the weak MX1-MTC response has a stronger effector synergy, meaning that perturbations in the effector elements are less likely to cause a change to the vertical states of the swing effector trajectories. In contrast, a strong response has a weaker synergy with respect to synergies associated with moderate MX1-MTC response.

Figure 6.6.1.1B illustrates that the above contrasts between response-types appear to be consistent for both groups, and this is supported by the finding of no significant interaction effect for age and response ($F(2, 220)$, $p>.05$). This was also supported when analysed within the dominant and non-dominant limbs [i.e. $F(1, 54) < 1$, and $p > .05$]. Therefore, the above findings indicate that synergy 'strength' (i.e. UCM_{ratio} values) is 'response-related' during the MX1-MTC transition region, and this is consistent for all participants. However, there is a significant main effect for age ($F(1, 110) = 3.46$, $p < .05$, $\eta^2 = .034$). Therefore, when averaging UCM_{ratio} across response types, and across the state-relevant time slices of the MX1-MTC transition region, there is a significantly higher 'synergy strength' (i.e. higher UCM_{ratio} values) found in older adults.

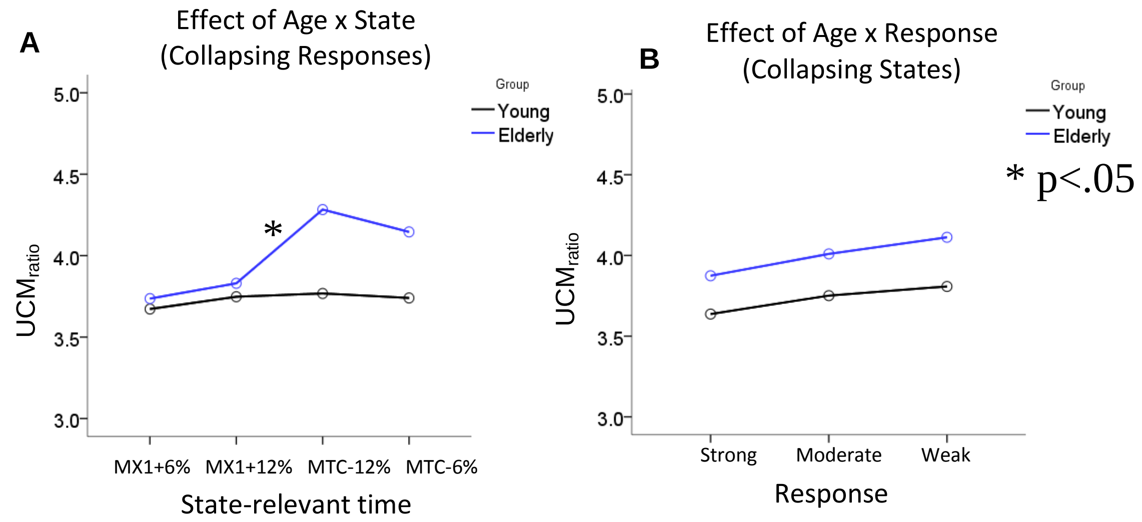


Figure 6.6.1.1.

Interactions on the dependent variable, UCM_{ratio} (GE_V/NGE_V), of the swing effector vertical task. Limbs have been collapsed into one group. **A**) Comparing between groups, for comparisons between time, on UCM_{ratio}. The significant interaction effect of 'group' x 'state-time' is indicated by a black asterisk. **B**) Comparing between groups, for comparisons between response-type, on UCM_{ratio}.

When averaged across all participants, there was a significant main effect of state-relevant time on the UCM_{ratio} value, $F(3, 330) = 8.8$, $p < .005$. Contrasts revealed that there was a significant increase in the UCM_{ratio} when comparing between the first two successive time states, i.e. between MX1+6%-MX1+12% [$F(1, 110) > 4.9$, $p < .05$, $\eta^2 = .04$]; MX1+12%-MTC-12% [$F(1, 110) > 15.3$, $p < .001$, $\eta^2 = .12$], and there was a significant decrease in the UCM_{ratio} between MTC-12%-MTC-6% [$F(1, 110) > 13.1$, $p < .001$, $\eta^2 = .11$]. However, when observing Figure 6.6.1.1A, there is a between group discrepancy, as there is a large increase in UCM_{ratio} between MX1+12% to MTC-12% for the elderly group.

To investigate how ageing may affect the synergy parameter strength in the swing effector (vertical task) from ‘moment-to-moment’ of the MX1-MTC region, an interaction analysis was performed on ‘group’ and ‘state-relevant time’, on UCM_{ratio}. There was a significant interaction effect between ‘group’ and ‘state-time’, $F(3, 330) = 5.8$, $p < .05$, $\eta^2 = .05$. Simple interaction effects from planned contrasts compared between the groups at each successive ‘state-time’ comparison (Figure 6.6.1.1A). A simple interaction of ‘group’ and ‘state-time’ was significant when comparing between MX1+12% and MTC-12% [$F(1, 110) = 13.9$, $p < .001$, $\eta^2 = .11$]. This result was also found to be significant when analyses were performed separately within the dominant and non-dominant limbs. The significant interaction result indicates that when compared to the young group, the older adults significantly increase the synergy strength associated with the vertical task of the swing effector from MX1+12% to MTC-12%. This tells us that during this period of the swing-cycle, the older adults structure the variability of segment covariations in such a way, that perturbations among swing effector segments are less likely to change the vertical state of the swing-effector trajectory.

Figure 6.6.1.2A-B displays the interaction of ‘response-type’ x ‘state-time’ within each age group. From the graphs displayed in Figure 6.6.1.2A-B, there are noticeable differences between the interactions of ‘response-type’ x ‘state-time’ in the young (Figure 6.6.1.2A) compared against the interactions of ‘response-type’ x ‘state-time’ in the elderly (Figure 6.6.1.2B). For example, when comparing between the moderate response and the strong response at state-relevant time-slice MX1+12%, both groups show similar UCM_{ratio} values. When then comparing between the next state-time of MTC-12%, the UCM_{ratio} associated with the moderate and strong responses in the elderly group have both increased. In contrast, when comparing this with the young group, only the UCM_{ratio} associated with the moderate response increased. This suggests a simple interaction of response and age during this state-relevant time comparison. However, the interaction effect of ‘state-time’ × ‘response-type’ × ‘group’ on UCM_{ratio} fails to reach significance [$F(6, 660) = 2.06$, $p = .06$, $\eta^2 = .02$].

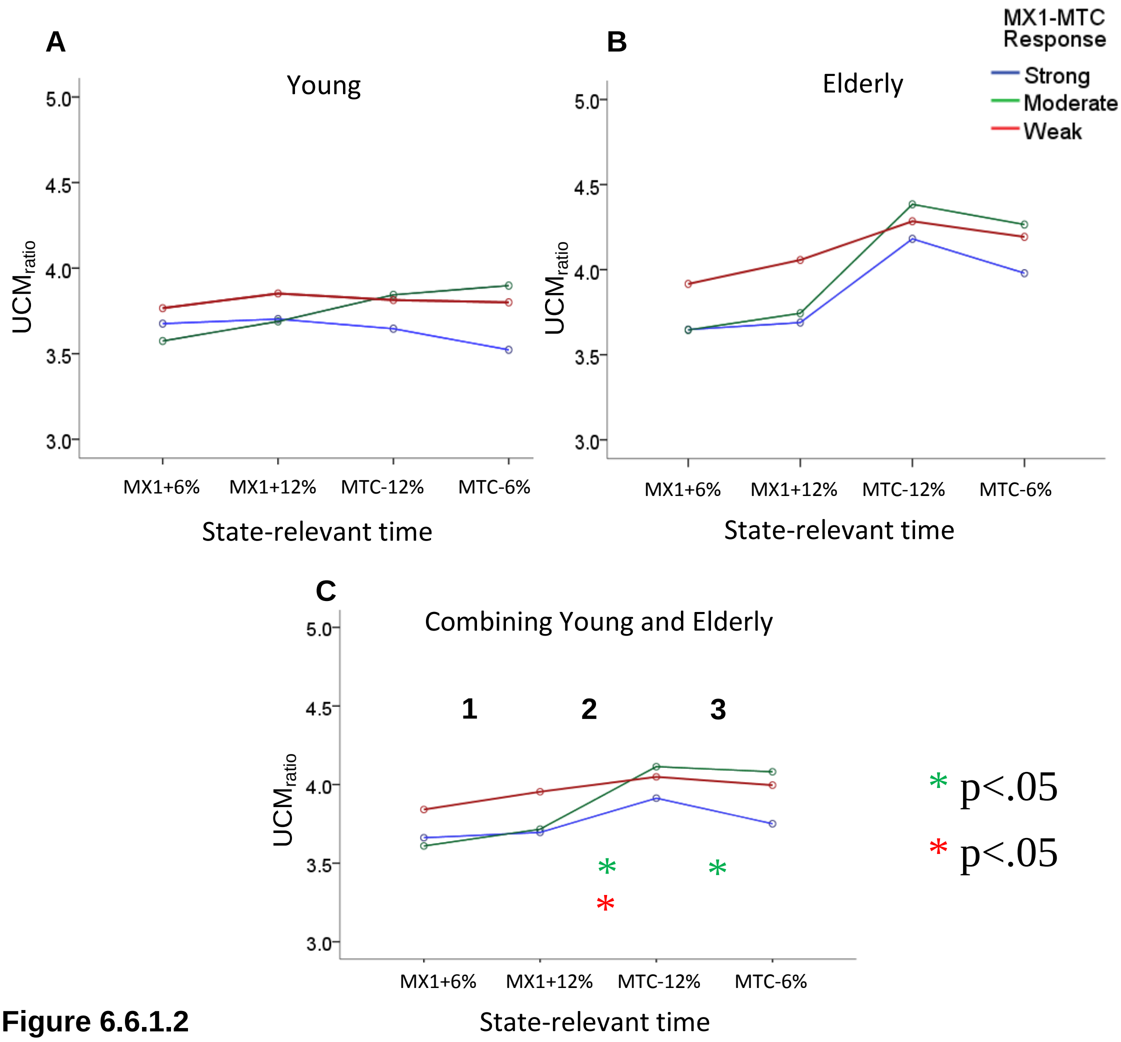


Figure 6.6.1.2

The ratio for component variance, UCM_{ratio}, displayed for the vertical task of the swing effector when comparing state-time and MX1-MTC response-type. The horizontal axis represents the sequence of state-relevant time-slices during the MX1-MTC transition region. Note: between MX1+12% and MTC-12% the scale is likely to be slightly greater than 6% proportion of swing phase time by about 3-5%. A) and B) represents the average response at each time-slice for the young and elderly groups respectively. C) when the groups are combined, the average response-type across time-slices is displayed. The numbers represent three levels of comparison between 'state-relevant time'. Between 'state-relevant times' there are red and blue asterisk to symbolise significant interactions of 'state-relevant time' x 'response-type' for respective levels of strong vs. moderate, and weak vs. moderate.

When performing the same three-way mixed design for the dominant and non-dominant limbs separately, both analyses also indicate no significance (p>.2). To determine whether the state-relevant time-slices selected in the three-way ANOVA are not capturing the three-way interactions during the MX1-MTC transition region, a modified ANOVA was applied. This now increased the state-relevant time slices between MX1 and MTC, from four to eight (e.g. MX1+3%, MX1+6%, MX1+9%, MX1+12%, MTC-12%, MTC-9%, MTC-6%, MTC-3%). The result was similar, there was no significant interaction indicated by the 'sphericity adjusted' Huynh-Feldt statistic ($F(6.75, 743) = 1.91$, $p=.07$, $\eta^2 = .017$). The above results tell us, that when comparing the groups along state-relevant times between MX1 and MTC, and where a synergy associated with 'one' effector response is compared with the synergy associated with 'another' effector response, there is no significant difference. Therefore, it cannot be accepted that one group has response-related synergies that are different to the other group, depending upon the response made to adjust the vertical state of the swing effector trajectory between MX1 and MTC.

Proceeding to investigate the interaction 'response-type' x 'state-time' on UCM_{ratio} reveals when combining all participants, for determining a general synergy behaviour associated with the vertical task of the swing effector (e.g. Figure 6.6.1.2C). There was a significant interaction between response-type and state-relevant time ($F(6, 660) = 14.2$, $p < .001$, $\eta^2 = .12$).

Six contrasts investigated simple interaction effects at several levels of comparisons. The first comparison level of response-type was between the strong and moderate responses, and the second level was between weak and moderate response. There were three comparison levels of 'state-time' (indicated as 1, 2, 3, on Figure 6.6.1.2C), where the first was comparing between MX1+6% and MX1+12%, the second was between MX1+12% and MTC-12%, with the third state-time comparison between MTC-12% and MTC-6%. Therefore, six simple interaction effects (contrasts) of 'state-time' x 'response-type' x 'group' were created by comparing among two levels of 'response-type' and three levels of 'state-time'.

The first two contrasts show that there isn't a significant interaction when comparing the moderate to the strong response when the state was at MX1+6% compared to when it was at MX1+12% [$F(1, 110) = 2.7$, $p = .1$], and when comparing the moderate and weak response with the states of MX1+6% and MX1+12%. This indicates that as the state progresses between MX1+6% and MX1+12%, there is a general increase in the UCM_{ratio} which is similar for all the response types.

The third and fourth contrasts compared the response levels with the MX1+12% and MTC-12% states. For all participants, the strong response affects the UCM_{ratio} significantly less compared to the moderate response [$F(1, 110) = 10.7$, $p < .001$, $\eta^2 = .09$]. Also between these same two states, the weak response affects the UCM_{ratio} significantly less than the moderate response [$F(1, 110) = 25.3$, $p < .001$, $\eta^2 = .19$].

The fifth and sixth contrasts examine the responses of moderate and strong, and moderate and weak, at the next successive state, MTC-12% and MTC-6%. The contrast between the moderate and strong response compared between state MTC-12% and MTC-6% is significant [$F(1, 110) = 6.6$, $p < .05$, $\eta^2 = .06$], while contrast six was not significant [$F(1, 110) = 0.1$]. This tells us that the strong response 'effort' to change the swing limb length decreases the UCM_{ratio} significantly more as the swing phase approaches closer to MTC compared to when the UCM_{ratio} is affected by moderate and weak 'efforts' to change the vertical extent of the swing limb.

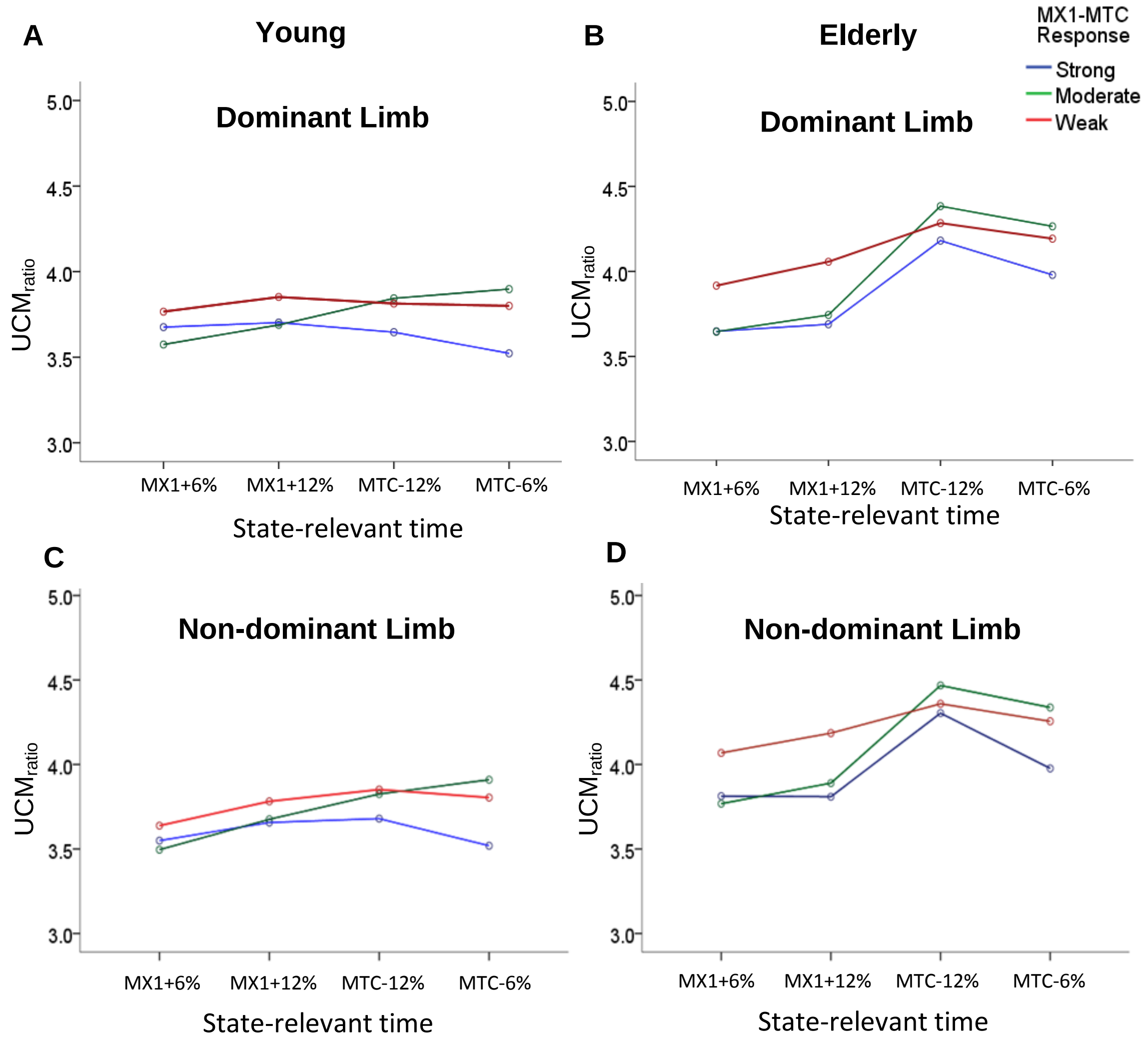


Figure 6.6.1.3

The ratio for component variance, UCM_{ratio}, displayed for the vertical task of the swing effector when comparing state-time and MX1-MTC response-type. These graphs separate dominant and non-dominant limbs. The horizontal axis represents the sequence of state-relevant time-slices during the MX1-MTC transition region. A) and B) represents the average response at each time-slice for the young and elderly dominant limb respectively. C) and D) represents the average response at each time-slice for the young and elderly non-dominant limb respectively.

6.6.2 Synergy strength comparisons at the MX1 region

This section will follow the same form of analysis in the previous section (6.6.1) by investigating the effects of the independent variables (response, state-relevant time, age) on the UCM_{ratio} associated with the vertical task of the swing effector. However, this section will address the early swing phase region of MX1.

Figure 6.6.2.1A-B shows the comparison between the young and elderly for the UCM_{ratio} across the MX1 region state-relevant times MX1-12%m MX1-6%, MX1 and MX1+6%. When collapsing the participants, Figure 6.6.2.1C shows that the UCM_{ratio} is a maximum near MX1. Figure 6.6.2.1A-B illustrates the state and response comparisons between the groups.

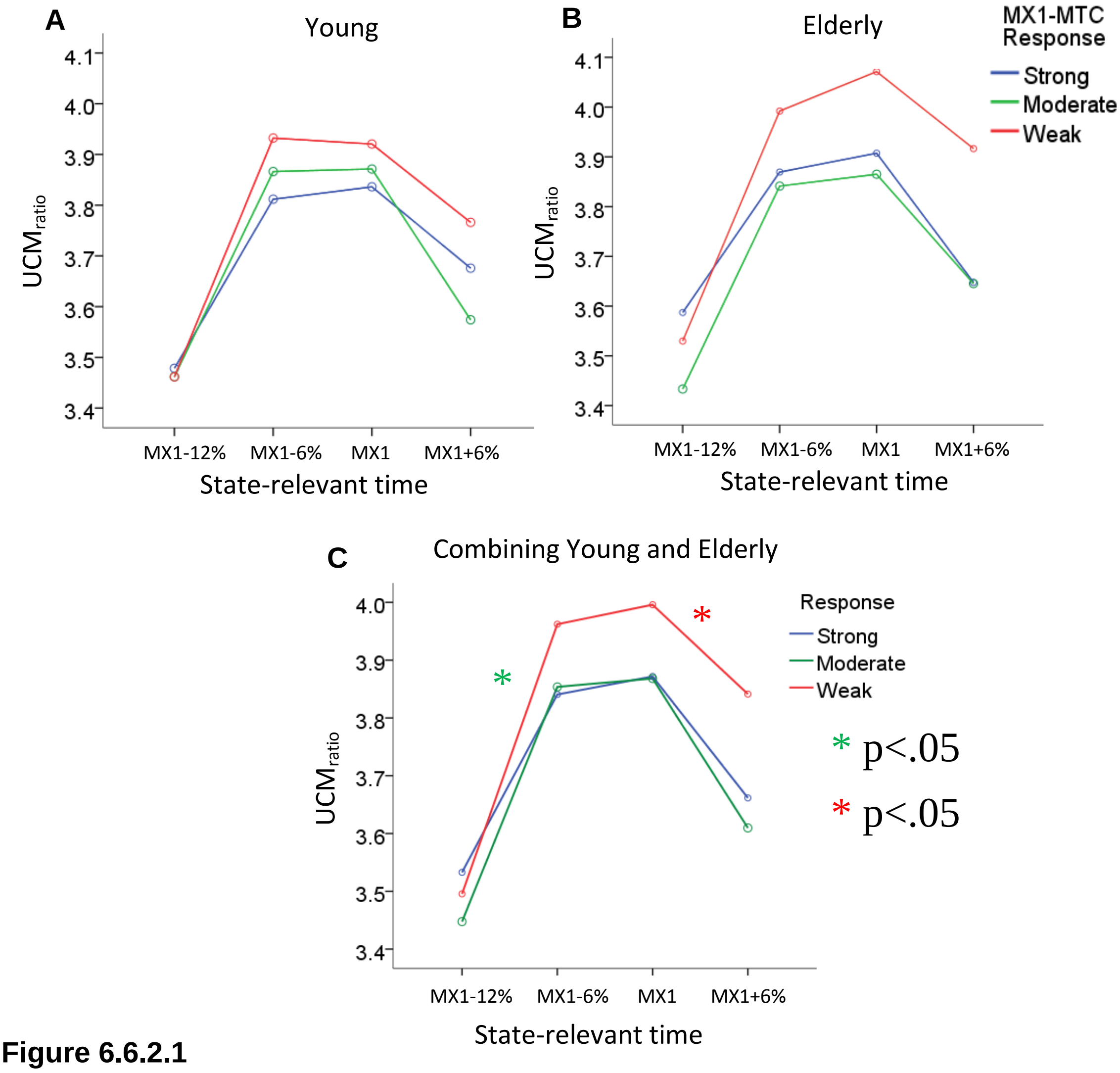


Figure 6.6.2.1

The ratio for component variance, UCM_{ratio}, displayed for the vertical task of the swing effector when comparing MX1 region of state-relevant time-slices and response-type. The horizontal axis represents the sequence of state-relevant time-slices during the MX1 region. A) and B) represents the average response at each time-slice for the young and elderly groups respectively. C) when the groups are combined, the average response-type across time-slices is displayed. The numbers represent three levels of comparison between state-relevant time. Between 'state-relevant times' there are red and blue asterisk to symbolise significant interactions of 'state-relevant time' x 'response-type' for respective levels of strong vs. moderate, and weak vs. moderate.

There was a significant main effect for 'response-type' [$F(2, 220) = 11.29$, $p < .001$, $\eta^2 = .09$] on UCM_{ratio}. Contrasts revealed that this was significant between the weak and moderate categories [$F(1, 110) = 30.9$, $p < .001$, $\eta^2 = .22$]. Alternately, contrasts comparing the strong and moderate response-types were not significant [$F(1, 110) = 1.5$, $p = .22$).

Figure 6.6.2.2B illustrates interactions of ‘group’ x ‘response-type’. However, there was no significant interaction effect of ‘group’ and ‘response-type’ ($F(2, 220) = 1.71$, $p = .18$), indicating that for this region of the swing-cycle, synergies associated with effector responses were not significantly different between groups.

When averaging the UCM_{ratio} across ‘state-relevant time’ and ‘response-type’, there was no main effect for ‘group’ ($p=.7$).

When investigating differences in synergy strength between ‘state-relevant time’ of the MX1 region, there was a significant main effect for ‘state-time’ [$F(2, 220) = 18.4$, $p < .001$, $\eta^2 = .14$]. This was significant with very large effect sizes when comparing between MX1-12% and MX1-6% [$F(1, 110) = 72.5$, $p < .001$, $\eta^2 = .40$], and when comparing MX1 with MX1+6% [$F(1, 110) = 4.8$, $p < .001$, $\eta^2 = .23$]. This tells us that the average synergy strength for all participants, when averaged across ‘response-type’, increases significantly prior to MX1-3% and then decreases significantly in the time region immediately following MX1.

There was not a significant interaction effect between ‘state-relevant time’ x ‘group’ [$F(2, 220) = 0.4$, $p=.9$]. Therefore, the ‘state-time’ dependent synergy behaviour reported above is not significantly different between groups.

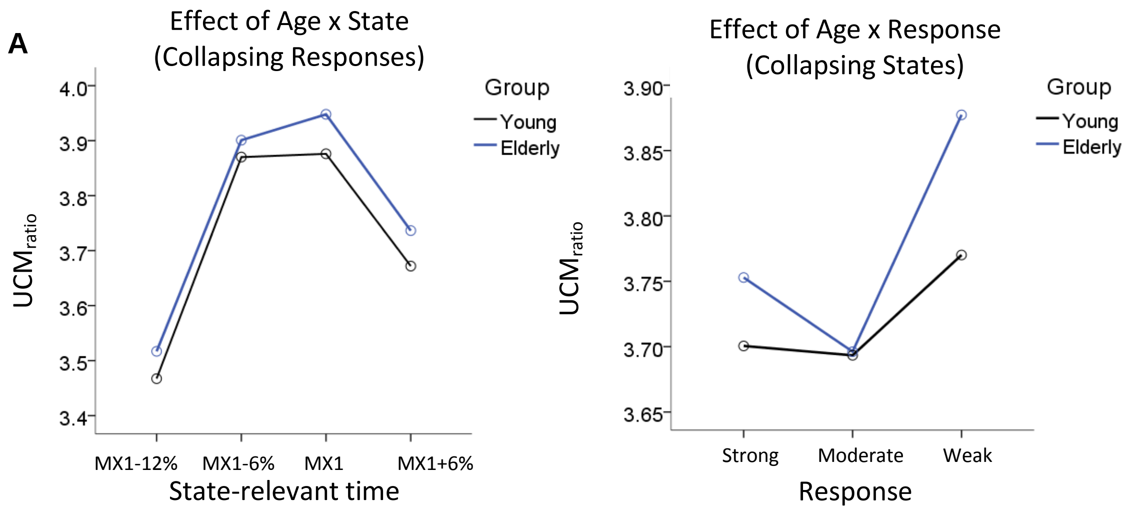


Figure 6.6.2.2.

MX1 region interactions on the dependent variable, UCM_{ratio} (GE_V/NGE_V), of the swing effector vertical task. Limbs have been collapsed into one group. **A**) Comparing between groups, for comparisons between time, on UCM_{ratio}. **B**) Comparing between groups, for comparisons between response-type, on UCM_{ratio}.

Figure 6.6.2.1C presented previously illustrates the interaction effect between state and response. This was significant [$F(4, 440) = 5.1$, $p< .001$, $\eta^2 = .04$]. The six simple interaction effects (contrasts) reveal significant comparisons between the moderate and strong responses when comparing between MX1-12% and MX1-6% [$F(1, 110) = 4.3$, $p < .05$, $\eta^2 = .05$]. The contrasts between moderate and weak responses were significant when compared between MX1 and MX1+6% [$F(1, 110) = 7.6$, $p < .01$, $\eta^2 = .065$].

When graphically comparing between groups on the 'response' x 'state-time' interaction between Figures 6.6.2.1A and B, the elderly group shows an increase in UCM_{ratio} associated with the weak response in comparison to the moderate response, and this is different to the young group. However, statistically, this is not significantly different [$F(6, 660) = 1.9$, $p = .12$, $\eta^2 = .02$].

The above analysis of 'state-relevant time' x 'response-type' interactions on synergy strength of the swing effector reveals the following. For trials associated with a strong MX1-MTC response by the swing effector, there is less increase in the synergy when comparing between time prior to MX1 (i.e. MX1-12% to MX1-6%), compared to synergies associated with trials of a moderate MX1-MTC response. This is not found to be significantly different between groups. For trials associated with a weak MX1-MTC response by the swing effector, there is less decline in synergy strength when comparing between a small time period following MX1, compared to synergies associated with moderate response trials. In section 6.6.1 it was found that the swing-cycle trials of the three categories of MX1-MTC response were associated with significantly different synergies. However, from the current analysis, it cannot be determined whether synergies of weak response trials existed in a significantly different form compared to synergies of moderate response trials, when compared prior to MX1.

6.6.3 Synergy strength comparisons through the MTC region

A three-way mixed design was applied as per the previous sections, however here only three levels were applied for state: MTC-6%, MTC and MTC+6%.

There was no main effect for 'group' [$F(1, 110) = 2.63$, $p < .011$, $\eta^2 = .023$]. This indicates that when averaging for 'response-type' and 'state-time', the synergy parameter was not significantly different between groups through MTC.

There was a significant main effect for 'state-time' [$F(2, 220) = 52.6$, $p < .001$, $\eta^2 = .32$] (Figure 6.6.3.1 A). This was significant when comparing between MTC-6% and MTC [$F(1, 110) = 81.0$, $p < .001$, $\eta^2 = .42$], and when comparing between MTC and MTC+6% [$F(1, 110) = 16.3$, $p < .001$, $\eta^2 = .13$]. The interaction between 'state-relevant time' and 'group' was not significant (Figure 6.6.3.1A). This indicates that the significant decrease in the synergy parameter, as it passes through the MTC region, is not significantly different between groups.

There was a significant main effect for 'response-type' [$F(2, 220) = 53.2$, $p < .001$, $\eta^2 = .33$] (Figure 6.6.3.1B). Contrasts revealed that this was significant between the strong and moderate categories with extremely large effect size [$F(1, 110) = 116.2$, $p < .001$, $\eta^2 = .51$]. Contrasts comparing the weak and moderate response categories were also significant with very large effect [$F(1, 110) = 24.6$, $p < .001$, $\eta^2 = .18$]. The interaction between 'response-type' and 'group' was not significant [$F(2, 220) = 1.45$, $p = .24$] (Figure 6.6.3.1B). This tells us that the synergy parameter found during the MTC region is dependent upon 'response-type', or more elaborately, the change in vertical state of the swing effector between MX1 and MTC.

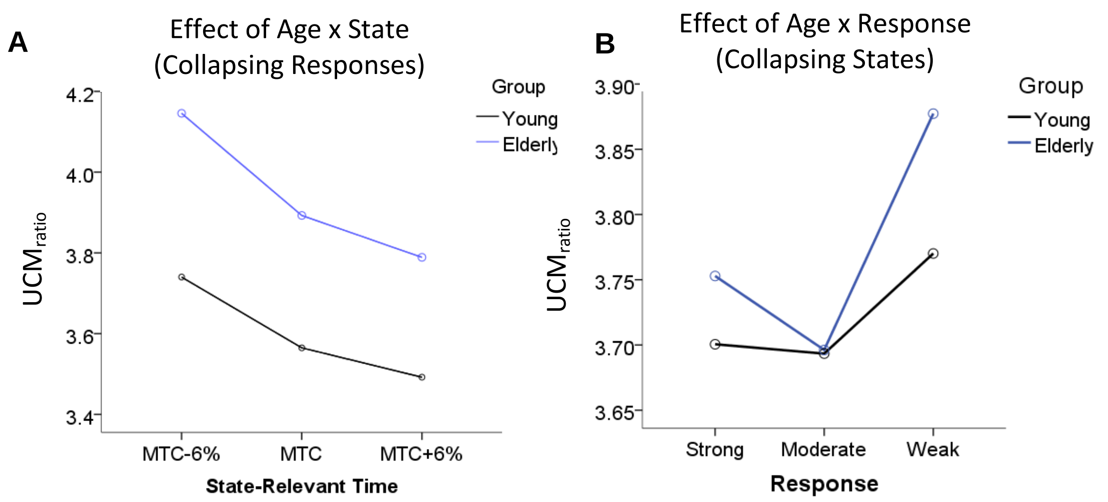


Figure 6.6.3.1.

MTC region interactions on the dependent variable, UCM_{ratio} (GE_V/NGE_V), of the swing effector vertical task. Limbs have been collapsed into one group. **A**) Comparing between groups, for comparisons between time, on UCM_{ratio}. **B**) Comparing between groups, for comparisons between response-type, on UCM_{ratio}.

Figure 6.6.3.2A-B illustrates 'state-time' and 'response-type' plots for UCM_{ratio} between groups. When collapsing groups, there was a significant interaction effect between 'state-time' x 'response-type' [$F(4, 440) = 3.62$, $p < .01$, $\eta^2 = .032$]. Contrasts show that this was significant between the moderate and weak response between MTC-6% and MTC [$F(1, 110) = 10.6$, $p < .005$, $\eta^2 = .088$]. This indicates that the strength of the synergy decreased significantly more in trials of the weak response, in comparison to the trials of moderate response. There was no significant interaction between 'state-time' x 'response-type' x 'group' (Figure 6.6.3.2A-B).

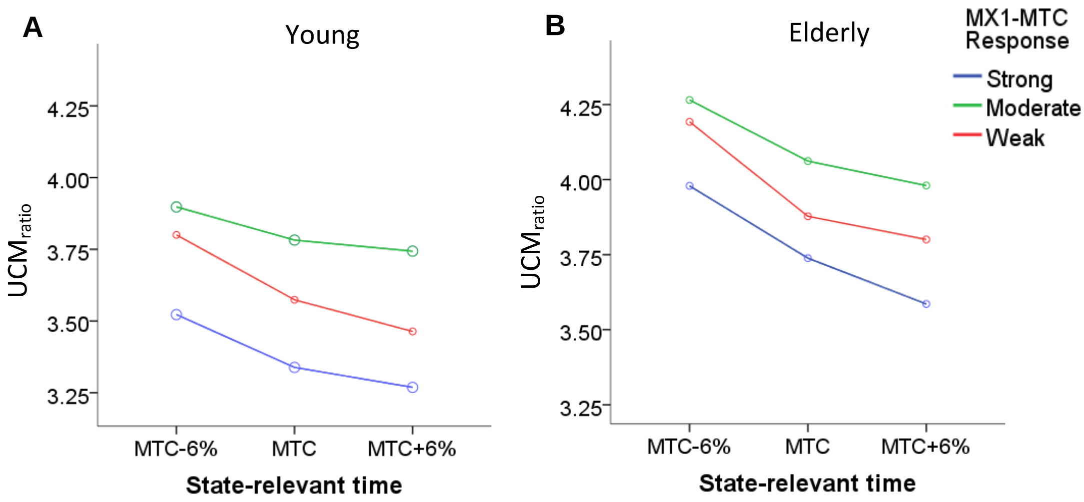


Figure 6.6.3.2

The ratio for component variance, UCM_{ratio}, displayed for the vertical task of the swing effector when comparing MTC region of 'state-relevant time-slices' and 'response-type'. The horizontal axis represents the sequence of state-relevant time-slices during the MTC region. A) and B) represents the average response at each time-slice for the young and elderly groups respectively.

6.7 Synergy effects due to state, response and age in the vertical direction of the stance effector

Figure 6.7.1 shows the effect upon the UCM components of variance of the stance effector in the vertical direction relative to the different response types across the swing phase states. The graphs show a slightly higher GEv and NGEv in the elderly graphs. A three-way mixed design ANOVA was performed on the dependent variables of NGE_V and GE_V, separately. There was an adjustment to the number of levels for independent variable of ‘state-time’ from previous analyses, by here increasing to five levels, for comparisons between six successive ‘state-times’ of MX1-9%, MX1, MX1+9%, MTC-9%, MTC, and MTC+9%. The other two independent levels of ‘response-type’ and ‘group’ remained as per previous sections. The results did not reveal a main effect for age [$F_{GEv}(1, 110) = 3.19$, $p = .077$; $F_{NGEv}(1, 110) = 3.17$, $p = .078$].

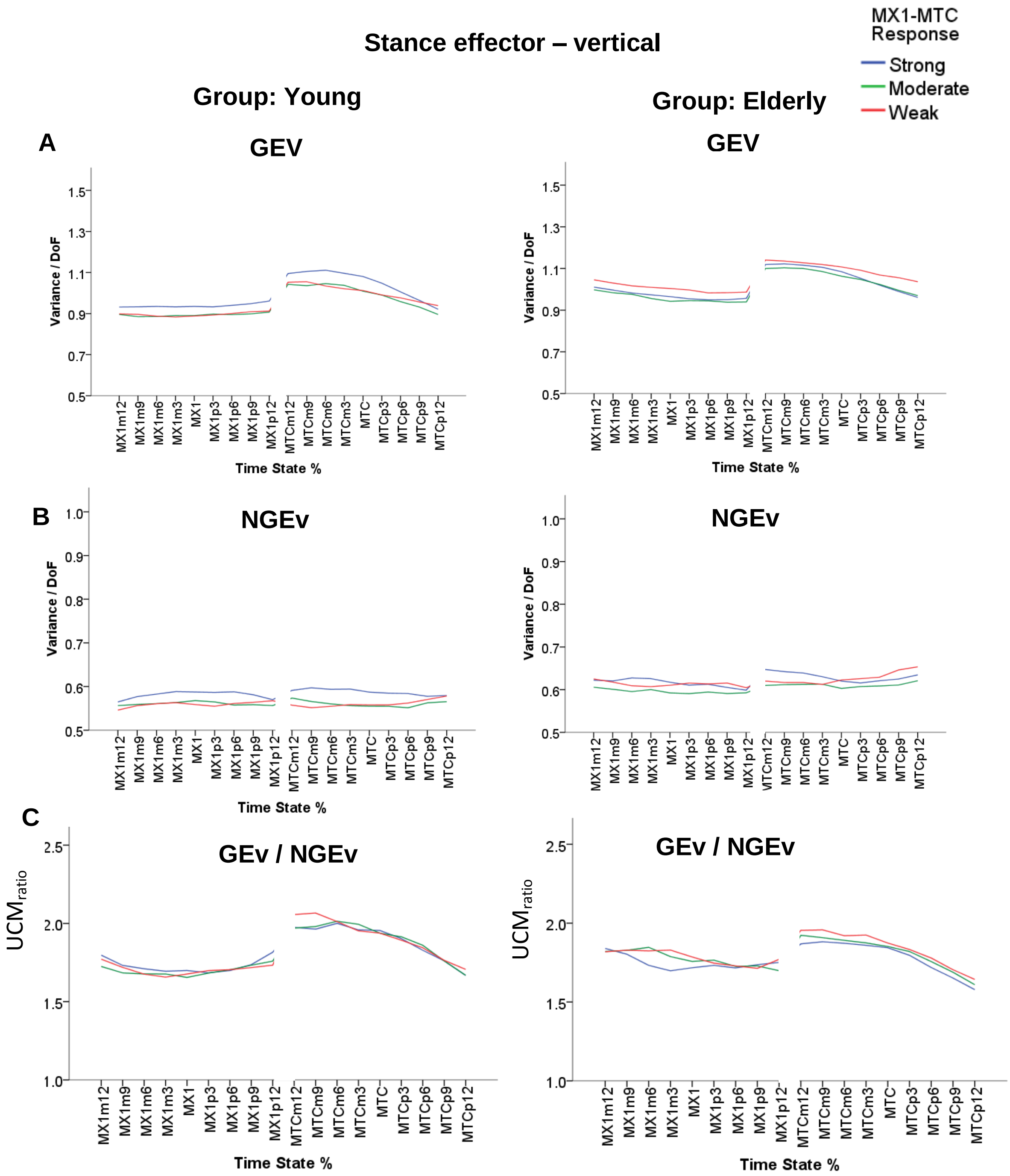


Figure 6.7.1.

UCM components of variance for the stance effector task variable of the vertical direction for both the young and elderly groups. Limb groups are collapsed into one group. This has a relatively small reduction effect on the 95% confidence error of the mean due to the increase in statistical degrees of freedom (i.e. *df* = N x 2). Comparisons are between weak, moderate and strong MX1-MTC responses across time-slices. The gap in the line graphs represents the change between MX1 and MTC regions. Except for this gap, the horizontal scale represents time-slices in 6% increments (of swing phase duration), where MX1+6% represents +6% of the swing phase post-MX1.

For the main analysis of this section, the statistical analysis method applying the three-way mixed design with repeated measures will be conducted on the dependent variable of UCM_{ratio}. For the analysis involving only the transition region between MX1 and MTC, four states will be included: MX1+6%; MX1+12%; MTC-12%; and MTC-6%. The other repeated measures variable of response category will include the categories of strong, moderate, and weak. Response-type can be described as the vertical change in stance effector state between MX1 to MTC, such that weak response indicates a small change between MX1 and MTC. Age group was the fixed independent factor.

For the MX1 region, there were no main effects for 'group', 'response-type' or 'state-time' (all $p>.14$). There were no interaction effects among the combinations of 'response-type' x 'group', 'state-time' x 'group', or 'response-type' x 'state-time' (all $p>.32$).

For the MX1-MTC region, Figure 6.7.2A-B plots the 'response-type' x 'state-time' interactions for each group separately. There is a significant main effect for state [$F(3, 330) = 56.1$, $p < .001$, $\eta^2 = .34$]. Contrasts revealed a significant difference between the successive states of MX1+6%-MX1+12% [$F(1, 110) = 5.5$, $p < .05$, $\eta^2 = .048$], and MX1+12%-MTC-12% [$F(1, 110) = 116.6$, $p < .001$, $\eta^2 = .52$]. This tells us that the strength of the average synergy (i.e. UCM_{ratio}) of the stance effector vertical task is significantly increasing on approach to MTC-12%. There was no interaction effect between 'group' and 'state-time'. Therefore, the synergy increase between these state-time slices is not different between groups.

The main effect for 'response' was not found to be significant [$F(2, 220) = 1.71$, $p = .185$]. There was no significant interaction effect for 'group' and 'response-type'.

There was a significant interaction effect for 'state' and 'response' with a small effect size [$F(6, 660) = 2.51$, $p < .05$, $\eta^2 = .02$]. The first contrast comparing between MX1+6% and MX1+12% with the comparison of the moderate and strong response shows a significant interaction [$F(1, 110) = 6.71$, $p < .05$, $\eta^2 = .06$]. The contrast of the next successive state-relevant time slice of MX1+12% to MTC-12% compared with the moderate and strong response was also significant [$F(1, 110) = 3.62$, $p < .05$, $\eta^2 = .03$]. These contrasts indicate the strength of the synergy associated with the strong response increases earlier compared to the moderate response, but then the moderate response shows an increase in the synergy strength at the next successive state which is at a rate comparatively higher compared to the change made by the strong response (Figure 6.7.2). These interactions of state and response were not found to be significantly different between the young and the older group.

The results indicate that the strength of the stance effector synergy for stabilising the swing hip position is not significantly different between groups during the MX1-MTC region of the swing phase.

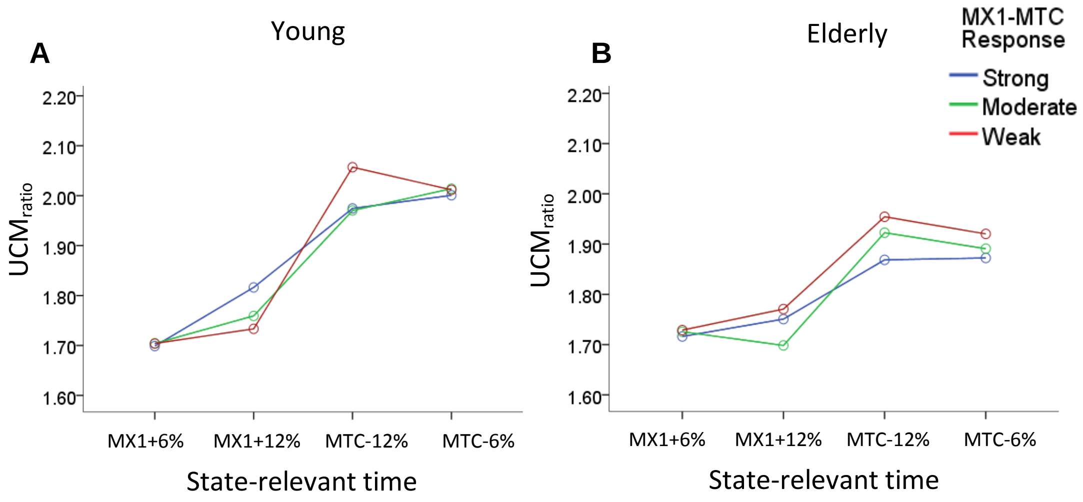


Figure 6.7.2

Data presented is for the stance effector in the medio-lateral direction for both the **A**) young group and **B**) elderly group. The limb preference has been collapsed into one group. The comparison is made for the average UCM_{ratio} across states MX1-12%(1), MX1-6%(2), MX1(3), MX1+6%(4). The comparison is made for the average UCM_{ratio} within the three response types of strong, moderate, and weak.

6.8 Synergy effects due to state, response and age in the medio-lateral direction of the stance effector

Figure 6.8.1 shows the effect upon the UCM variance variables of the medio-lateral task of the stance effector, relative to the different 'response-types' across the swing phase 'state-relevant time slices'. Response-type can be described as the vertical change in stance effector state between MX1 to MTC, such that weak response indicates a small change between MX1 and MTC. Graphically, the mean shows a slightly higher GEv and NGEv for the elderly. This was examined statistically from a three-way mixed design ANOVA as per previous section 6.7 (using five comparison levels of repeated measure variable 'state-time': MX1-9%, MX1, MX1+9%, MTC-9%, MTC, and MTC+9%. The other independent variables were as per section 6.7. The dependent variables were the GEv and NGEv. With respect to GEv, the results did not reveal a main effect for age [GE_v: $F(1, 110) = 1.3$, $p = .25$], however, with respect to NGEv there was a significant main effect for age [NGE_v: $F(1, 110) = 16.5$, $p < .001$, $\eta^2 = .13$]. This tells us that when averaging the 'state-relevant time-slices' of the swing phase and averaging across the 'response-type', the co-variance patterns of the elderly's stance effector has higher variability that causes a change to the task variable (i.e. higher NGE_v) compared to the young group. With respect to variability that does not cause a change to the task variable (i.e. task-irrelevant variability, GEv) there is no significant difference between the groups.

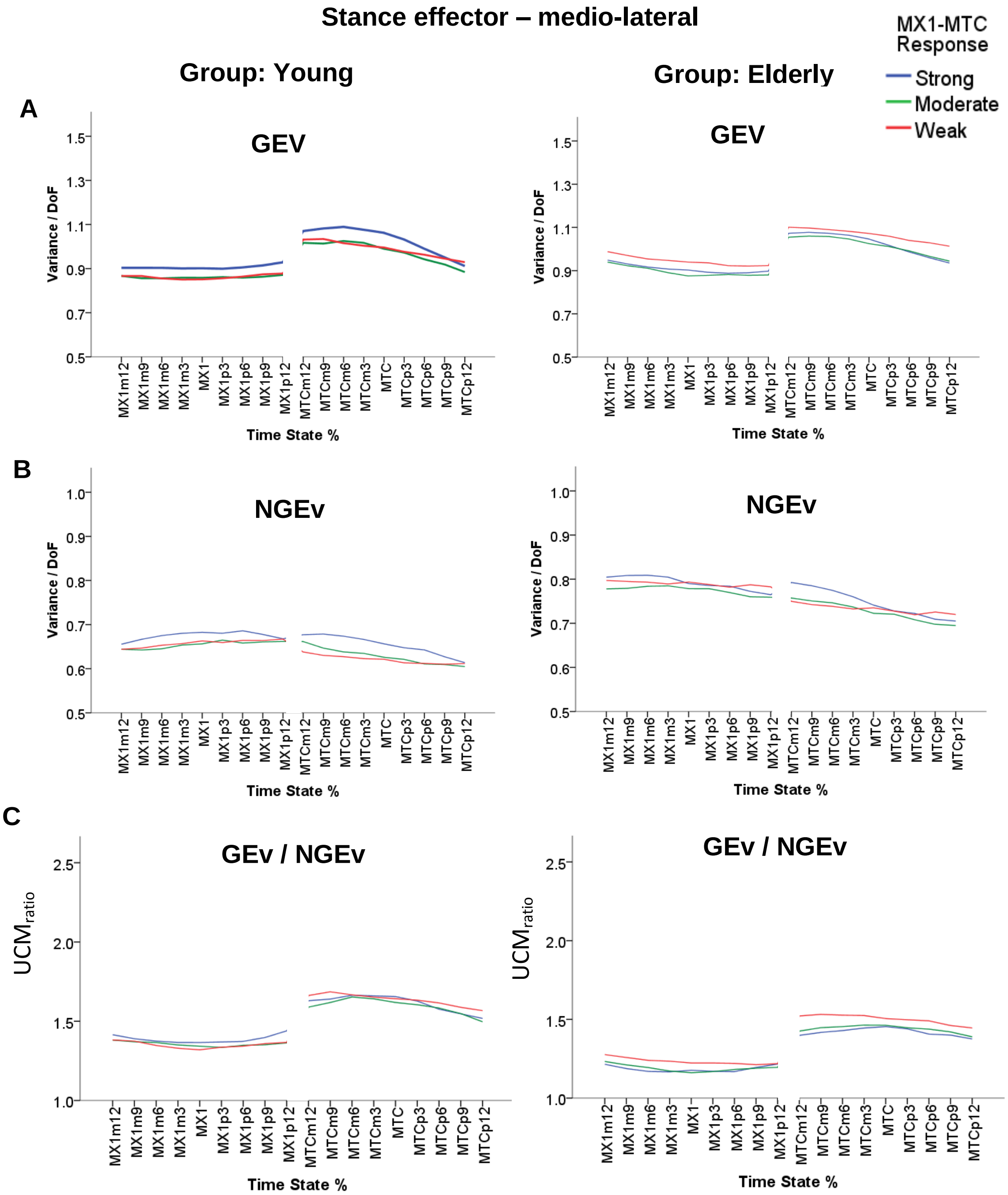


Figure 6.8.1.

UCM components of variance for the stance effector task variable of the medio-lateral direction for both the young and elderly groups. Limb groups are collapsed into one group. This has a relatively small reduction effect on the 95% confidence error of the mean due to the increase in statistical degrees of freedom (i.e. *df* = N x 2). Comparisons are between weak, moderate and strong MX1-MTC responses across time-slices. The gap in the line graphs represents the change between MX1 and MTC regions. Except for this gap, the horizontal scale represents time-slices in 6% increments (of swing phase duration), where MX1+6% represents +6% of the swing phase post-MX1.

For the main analysis of this section, the statistical analysis method applying three-way mixed design ANOVA (with repeated measures on 'state-relevant time') will be conducted on the dependent variable of UCM_{ratio}. For the analysis involving the initial swing phase leading up to and passing through MX1, there will be three levels of 'state-time' comparisons: MX1-12%-MX1-6%; MX1-6%-MX1; and MX1-MX1+6%. For the transition region between MX1 and MTC, four states will be included: MX1+6%; MX1+12%; MTC-12%; and MTC-6%. For the region passing through MTC, there will be two levels of 'state-time' comparisons: MTC-6%-MTC; MTC-MTC+6%. The other independent variable of response category will include the categories of strong, moderate, and weak. Age group was the fixed independent factor.

There was a main effect for 'state-time' [$F(1, 110) = 14.1$, $p < .001$, $\eta^2 = .11$]. Contrasts comparing between successive 'state-relevant time-slices' indicate that the strength of the synergy is decreasing as the swing phase approaches MX1; this is found for the 'state-relevant time slice' comparison between MX1-12% - MX1-6%, $F(1, 110) = 28.3$, $p < .001$, $\eta^2 = .2$, and for the state comparison of MX1-6% - MX1, $F(1, 110) = 10.2$, $p < .005$, $\eta^2 = .09$. However, there was no significant contrast between 'state-relevant times' of MX1-MX1+6% [$F(1, 110) = 1.8$, $p = .18$].

In the MX1 region, there was a main effect for 'group' [$F(1, 110) = 9.3$, $p < .005$, $\eta^2 = .078$]. The interaction between 'state-relevant time' x 'group' was not significant. This indicates that the general behaviour of participants is to decrease the synergy strength for the medio-lateral task of the stance effector during the swing-phase leading up to MX1.

There was no main effect for 'response-type' [$F(2, 220) = 1.96$, $p = .14$]. When averaging across 'state-time' there was a significant interaction effect for 'response-type' and 'group' [$F(2, 220) = 7.0$, $p < .005$, $\eta^2 = .06$]. Contrasts showed that both groups are not different when comparing between the moderate and the strong responses [$F(1, 110) = 1.8$, $p = .19$], but there was a significant difference when comparing groups between the moderate and the weak responses [$F(1, 110) = 8.8$, $p < .005$, $\eta^2 = .08$]. This indicates that, within the elderly group, the medio-lateral task of the stance effector has a significantly higher synergy parameter associated with the weak stance effector response (i.e. small increase of the swing hip height from MX1 to MTC) compared with the moderate response. This is not apparent for the young group.

The three-way interaction of ‘state-relevant time’ x ‘response-type’ x ‘group’ is not significant [$F(6, 660) = 0.4$, $p = .8$]. This indicates that the stronger synergy related to the weak response in the elderly can be considered significant to the young group at each moment of the swing phase, and this trend continues as it passes through the MX1 state (Figure 6.8.2A-B).

For the elderly, there is a strong medio-lateral synergy prior to MX1 associated with the particular trials where the stance effector trajectories have a small increase in displacement between MX1 and MTC (i.e. weak response). It is unknown here, whether relatively strong synergies that are associated with the weak response trials, are actually having an effect on the magnitude of effector displacement between MX1 and MTC.

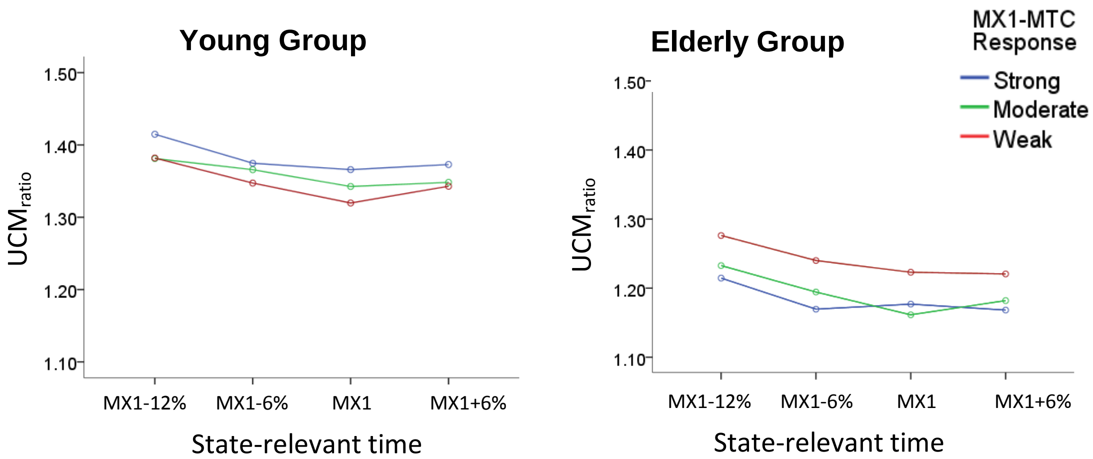


Figure 6.8.2

Data presented is for the stance effector in the medio-lateral direction for both the **A**) young group and **B**) elderly group. The limb preference has been collapsed into one group. The comparison is made for the average UCM_{ratio} across states MX1-12%(1), MX1-6%(2), MX1(3), MX1+6%(4). The comparison is made for the average UCM_{ratio} within the three response types of strong, moderate, and weak.

For the MX1-MTC region, there was a very large effect size for the main effect of ‘state-time’ [$F(3, 330) = 254.1$, $p < .001$, $\eta^2 = .7$]. Contrasts revealed there were significant increases in UCM_{ratio} for all four successive ‘state-times’ [$F(1, 110) > 8.1$, $p <$

.005, $\eta^2 = .07$], and the largest effect size was between MX1+12%-MTC-12% [$F(1, 110) = 381.6$, $p < .001$, $\eta^2 = .78$]. There was no significant interaction effect between 'group' and 'state-relevant time'; however, there was a significant main effect for 'group' [$F(1, 110) = 11.1$, $p < .001$, $\eta^2 = .09$]. This indicates that there is a significantly lower synergy in the elderly group during the MX1-MTC region, when the UCM_{ratio} is averaged across 'response-type' and 'state-relevant times'.

When examining differences between 'response-type', there was a significant main effect [$F(2, 220) = 7.1$, $p < .005$, $\eta^2 = .06$]. Contrasts revealed significant difference between moderate and weak [$F(1, 110) = 15.1$, $p < .001$, $\eta^2 = .12$], but not between moderate and strong [$F(1, 110) = 1.4$, $p < .23$].

There was a significant interaction effect for 'group' x 'response-type' [$F(2, 220) = 6.8$, $p < .005$, $\eta^2 = .06$]. Contrasts revealed a significant difference when comparing between groups, and between the weak and moderate response [$F(1, 110) = 6.7$, $p < .05$, $\eta^2 = .06$], but not for the strong and moderate [$F(1, 110) = 2.6$, $p = .11$] (Figure 6.8.3).

To examine where significant interactions between 'response-type' occurs during this MX1-MTC transition region, there was a significant interaction effect for 'state-time' x 'response-type' [$F(6, 660) = 7.51$, $p < .001$, $\eta^2 = .06$]. The contrast comparing the change between the first successive 'state-relevant time slices' of MX1+6% and MX1+12% shows an interaction when comparing between the moderate response with the strong response [$F(1, 110) = 8.2$, $p < .01$, $\eta^2 = .07$], but there was no significant difference for the moderate and weak response ($p = .77$). For the next contrast comparing between successive state-relevant time slices of MX1+12% and MTC-12%, there was a significant interaction when comparing the moderate response with the strong response [$F(1, 110) = 4.5$, $p < .05$, $\eta^2 = .04$] and also with the interaction of these states when comparing the moderate with weak [$F(1, 110) = 9.1$, $p < .005$, $\eta^2 = .08$]. For the contrasts comparing the next successive state-relevant time slices of MTC-12% and MTC-6%, there wasn't a significant interaction when comparing the moderate response with the strong response ($p = .32$) but the contrast of these 'time slices' with the comparison of the moderate with weak response was significant [$F(1, 110) = 5.3$, $p < .05$, $\eta^2 = .05$].

These contrasts indicate the strength of the UCM_{ratio} associated with the strong response increases earlier compared to the moderate response, but then the moderate response significantly increases the synergy strength (i.e. higher UCM_{ratio}) at the next successive 'state-relevant time slice' in comparison to the strong response (Figure 6.8.3). And during this same period, the weak response increases significantly compared to the moderate response. As the swing phase further approaches the MTC event, the synergy associated with the weak response does not maintain the same level of strengthening as the moderate response.

The interactions of 'state-relevant time' x 'response-type' were not found to be significantly different between the young and the older group [$F(1, 110) = 0.34$, $p = .85$]. This indicates that the strength of the stance effector synergy for stabilising the medio-lateral swing hip position, as described above, behaves in a similar way across all participants during the MX1-MTC region of the swing phase (Figure 6.8.3).

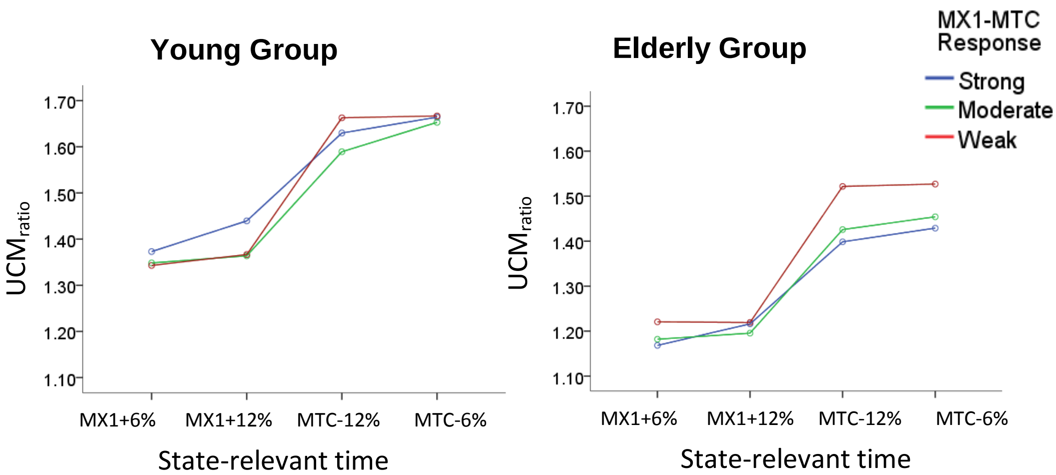


Figure 6.8.3

Data presented is for the stance effector in the medio-lateral direction for both the **A**) young group and **B**) elderly group. The limb preference has been collapsed into one group. The comparison is made for the average UCM_{ratio} across states MX1+6%(1), MX1+12%(2), MTC-12%(3), MTC-6%(4). The comparison is made for the average UCM_{ratio} within the three response types of strong, moderate, and weak.

6.9 Chapter 6 results summary

Below are the results related to the null hypothesis listed at the beginning of this chapter.

6.9.1 Hypotheses of synergies in effector systems

- Null hypothesis 6.1 The UCM, V_{ratio} (DV) will not be significantly different from the value of 1 when compared at each state-relevant time-slice of the swing-cycle.
  - The null hypothesis was rejected for all four effector task variables on the synergy ratio, V_{ratio} for all time slices of the swing-cycle. The value of V_{ratio} for the swing effector vertical task, the stance effector vertical task and the stance effector medio-lateral task was significantly greater than 1 ($p<.05$). The swing effector anterior-posterior task was significantly lower than 1 ($p<.05$).
- Null hypothesis 6.2 There will not be a significant difference in 'group' (IV) on the effector synergy (UCM_{ratio}; DV) across the investigated swing-cycle region.
  - The null hypothesis was rejected for the swing effector vertical and anterior-posterior tasks, and for the stance effector medio-lateral task.
  - There was a significantly lower UCM_{ratio} for the stance effector medio-lateral task in the elderly during both MX1 and MTC regions. There was a significantly higher UCM_{ratio} in the elderly swing effector vertical task at early state-relevant time slices of the MTC region. There was a significantly lower UCM_{ratio} for the elderly swing effector anterior-posterior task at early state-relevant time slices of the MTC region.

- Null hypothesis 6.3 The synergy parameter, UCM_{ratio} (DV), will not be significantly different between the effector response types (IV; weak, moderate, strong). There will be no significant interaction of 'response-type' and 'group' on UCM_{ratio}.
    - The null hypothesis was rejected for the MX1-MTC region. There is a significant effect of response type on synergy parameter, UCM_{ratio}. Across participants, the 'weak' response-type is significantly higher than the 'moderate' response-type. The 'moderate' response is significantly higher than the 'strong' response-type.
    - The null hypothesis of a significant response x group interaction was accepted.
    - When averaging across response type in the MX1-MTC region and comparing between groups, there is a significantly higher UCM_{ratio} in older adults for the vertical task of the swing effector.
    - For the medio-lateral task of the stance effector, the null hypothesis was rejected for the MX1-MTC region. When comparing between groups, the weak response was significantly higher relative to the moderate response in the elderly group.
- Null hypothesis 6.4 When the synergy parameter, UCM_{ratio} (DV), is averaged across 'response-type', there will not be a significant difference when comparing between 'state-relevant time-slices' (IV). There will be no significant interaction of 'state-relevant time' and 'group' on UCM_{ratio}.
    - For the vertical task of the swing effector, the null hypothesis was rejected for the interaction of 'group' x 'state-time'. There is a significant increase in the UCM_{ratio} between MX1+12% and MTC-12% for the elderly group when compared to the young group.
    - For the vertical task of the stance effector, the null hypothesis was accepted.

- For the medio-lateral task of the stance effector, the null hypothesis was rejected for the interaction of ‘group’ x ‘state-time’. There is a significant decrease in the UCM_{ratio} between MX1+12% and MTC-12% for the elderly group when compared to the young group.

♦ Null hypothesis 6.5 The synergy ratio, UCM_{ratio} of the task variable (DV), when compared between effector responses (IV-1), will not be significantly different when compared between state-dependent time-slices (IV-2). The older adult group will not be significantly different to the young group (IV-3) when comparing the interaction effect of response (IV-1) and state-time (IV-2) on the UCM_{ratio} of the effector task variable (DV).

- This null hypothesis was almost rejected (p=.06) for states between MX1-MTC for the swing effector vertical task. The elderly show a significantly higher synergy. For the MX1 and MTC regions, the null hypothesis was accepted for interactions on state-time x response x group. The synergy strength was not behaving significantly different between the groups when comparing between response types and state-time.
- Because the synergies associated with response-types are behaving in a similar way, this interaction hypothesis is going to be less sensitive to age differences. Null hypothesis 6.4 will be more reflective of age differences.

Chapter 7

Coordination between the stance and swing effectors

7.1 Background

This chapter investigates how the stance and swing effectors are coordinated. Coordination between the stance and swing effectors implies that they are in essence acting to satisfy a mutual task goal. This can be framed through the general concept of Nash equilibria (Nash, 1950). Although Nash equilibria was founded on inter-participant coordination, its concepts have recently been applied to explain sensorimotor coordination (Braun, Ortega and Wolpert, 2009).

Nash equilibrium is a theory of collaboration amongst variable elements, or actors, of a system. The theory suggests that there is an optimal solution between two actors. In satisfying a common goal like toe clearance, the configuration of the stance effector contributes either positively, or negatively, towards the concurrent swing effector task goal of configuring the swing limb length. In satisfying either of the three types of goal scenarios above, the stance and swing effector will need to act out a strategy of mutual cooperation. The problem for a system of two effectors with different goals is the selecting of a strategy which best achieves optimal coordination, so that Nash equilibria can be attained. A Nash equilibria can be attained by either a pure or mixed coordinative strategy (Braun, Ortega and Wolpert, 2011). For example, in a ‘mixed’ Nash equilibria, two actors behave stochastically, such that the ability of one actor to respond following the decision by the alternate actor provides less certainty in the outcome and is likely to require a third party mediator. The one benefit is that both actors have more freedom to select independent strategies. To help solve this coordination dilemma, an impartial third party mediator correlates the decision of both actors to attain the best outcome for both parties, i.e. fairness of assigning responsibility over the long term. Ideally, the third party mediator prefers not to intervene, because this comes at some expense, so the system is rewarded by actor A and actor B learning an autonomous correlated policy. The alternative solution, a ‘pure’ Nash equilibria, is defined by both actors selecting more deterministic strategies and therefore both actors have greater predictability. The cost of this deterministic strategy is that the system has less flexibility.

Irrespective of the three types of goal strategies adopted, there are four general response types to the length change combinations of the stance and swing effectors. This is defined by an effort to effect a vertical displacement between MX1 and MTC. Given that the toe point is low at MX1 a length change response by the two actors [swing effector, stance effector] could potentially be [up, stay], [stay, up], [stay, stay] or [up, up]. The terms ‘up’ and ‘stay’ could theoretically be replaced with terms which relate to the intervention strength by the controller, i.e. ‘strong’ or ‘weak’. Therefore, a control effort response within the stance and swing effectors can take the form of: [strong, weak], [weak, strong], [strong, strong] or [weak, weak]. In terms of the initial condition of the toe at MX1, the response of either effector can take on some degree of strength as the limb approaches MTC.

Braun, Ortega and Wolpert (2011) propose that there is a general criteria for successfully satisfying coordination: favourable initial conditions; stereotypical coordination pattern; a time difference of early convergence to a final position by one actor relative to the other. To a certain degree, these conditions are reflected in the design of the between-effector coordination analysis. The analysis considers: 1) the MX1 event as an important initial condition; 2) coupling strength pattern between the stance and swing effectors; 3) the comparative correction gain of each effector for influencing the final toe position at MTC.

7.2 Method and analysis design

7.2.1 Correlation analysis

To evaluate the coordination between the effectors a correlation analysis was applied. The correlation coefficients, ‘r’, of the covariance between three effector pairings were assessed: swing and stance; stance and combined; and swing and combined. The correlation value, is in some respects, an effect size describing the association between two variables, and ‘r’ is often recommended in statistics to represent effect size. An ‘r’ value of .5 describes approximately 25% of the covariance between variables. For ‘r’ values between .3 to .5, Cohen (1988) suggests that this represents a moderate to large effect, where r values greater than .5 represent a large effect. However, the correlation co-efficient, ‘r’, along with its statistical significance value, ‘p’, was computed for each of the three pairings.

Correlations were made at each state-relevant time-slice of the two sub-phases of the swing-cycle. To control for the possible effect of initial conditions, and a subsequent response behaviour, the walking trial was sub-divided into nine groups (as per Section 6.2). For this correlation analysis, only the moderate response data was used. Therefore, the input data represented approximately one third of the total swing cycle trials.

7.2.2 The null hypotheses

- ♦ Hypothesis 7.1 There will be no significant difference between groups (IV) in the effector correlations (DV) within each state-relevant time-slice of the swing phase;
    - o within effector pairing, within limb, independent sample t-tests at each time-slice
- ♦ Hypothesis 7.2 There will be no significant difference between groups when comparing between swing-combined and the stance-combined correlations at states MX1 and MTC;

- within group one-way ANOVA at MX1 and MTC and post-hoc test using Bonferroni tests

7.2.3 Correlations between the effectors state-response between MX1 and MTC

This section investigates the between-effector correlations when changing their state (vertical displacement) from MX1 and MTC. One time-series of vertical displacement data derived from the change in the vertical dimension of the effector state between MX1 and MTC was obtained for each effector (e.g. 'dSt' represents the change in the stance effector between MX1 and MTC), from all swing-cycle trials. From the three time-series displacement data, three correlations were performed: 'dSt-dCo'; 'dSw-dCo' and 'dSt-dSw'.

- ♦ Null hypothesis 7.3 There will not be a significant difference between the young and elderly when comparing between the three paired effect couplings.
  - Independent sample t-tests with significance set at ($p<.1$) for each between-group comparison

7.2.4 Persistence: effector trajectory change between MX1 and MTC

Using the same time-series of effector vertical displacements, the persistence of these three time-series was measured to provide a different insight into the way the effectors are controlled between MX1 and MTC.

- ♦ Null hypothesis 7.4 There will not be a significant difference when comparing between the young and elderly on the persistence associated with changes to effector states between MX1 and MTC;
  - Independent sample t-tests with significance set at ($p<.1$) for each between-group comparison

7.2.5 Correction gains between the effectors

This section explores which of the stance or swing effectors takes most responsibility for correcting the combined effector trajectory. The stance and swing effector displacement (representing correction) was correlated separately against the combined effector vertical displacement state at MX1. The 'correction gain' of both effectors was derived from the regression coefficient fitted to a bivariate plot. A correction 'gain symmetry' was also computed, which describes the relative coefficients between the stance and swing effectors, where the 'gain symmetry' = $(c_{swing})/[Abs(c_{stance})+Abs(c_{swing})]$. The absolute of the coefficient was taken because there existed some stance effector coefficients which were positive and not a consistent direction with the swing effector coefficient (see Figure 7.3.4.1.). Computed values greater than 0.5 indicate the relatively larger contribution by the swing effector for the initial 'error' position of the combined effector at MX1 (e.g. White and Diedrichsen, 2010).

- Null hypothesis 7.5 There will not be a significant difference between the correction gains of the stance effector and swing effector on the combined effector state.
    - Paired sample t-tests
- Null hypothesis 7.6 There will be no significant difference between the groups when comparing on the dependent variable of effector correction gain.
    - Independent sample t-tests

7.3 Results

7.3.1 Correlations within participants at each time state of the swing phase

This section explores the cooperation between effectors by correlating their trajectory states (of the vertical dimension) at common state-relevant time-slices in the subphase regions MX1 and MTC.

Figure 7.3.1.1 describes the results of the correlation analysis within each group for the MX1 region. Observations between the effector couplings indicates a large significant difference when comparing covariance between the stance and swing effector coupling, compared to the covariance of the stance-combined and swing-combined couplings. A strong positive correlation exists for effector couplings between the stance and combined, and the swing and combined effectors during the MX1 phase; these correlations are significant throughout the MX1 region ($p<.01$). In contrast, the stance-swing covariance reduces from a significant positive correlation at MX1-12%, to then become not significant near MX1, until becoming a significant negative correlation during the MTC region.

When comparing the stance-combined coupling and the swing-combined coupling within both the groups, there are differences when considering the 95% confidence intervals of the group 'r' means in Figure 7.3.1.1 and 7.3.1.2. At MX1+12% the young group have a correlation of the stance-combined effector coupling of $r_{St\text{-}Co} = .45$, and this is not significantly higher ($p=.95$) than the correlation of the swing-combined effector coupling $r_{Sw\text{-}Co} = .47$. In comparison, when comparing within the elderly group, there is a significantly higher ($p<.001$) stance-combined correlation $r_{St\text{-}Co} = .53$ compared to the swing-combined correlation $r_{Sw\text{-}Co} = .40$.

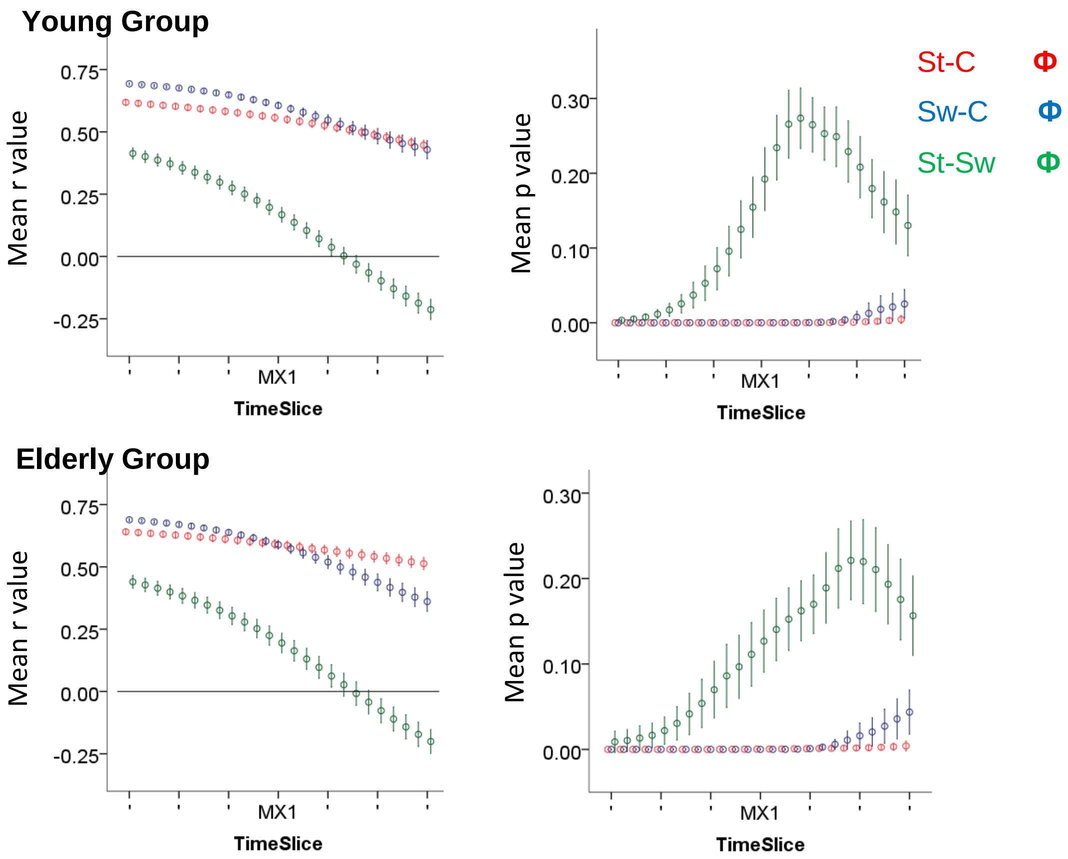


Figure 7.3.1.1

The coupling strength between paired effectors in the vertical dimension across MX1 region. The pairing code St-C, Sw-C, and St-Sw represents stance-combined, swing-combined and stance-swing effector couplings. The column of graphs on the left represents the group mean correlation coefficient (r). The column of graphs on the right represents the associated significance of the correlation (group mean of the p-values). The horizontal axis represents the time-slice of the MTC region in units of % swing phase. Each plotted group mean and 95% confidence interval (error bar) represents one time-slice of the region, spanning MX1-12% to MX1+12%. The data is taken from the trials representing the moderate response of trajectory change between MX1 and MTC.

Figure 7.3.1.2 describes the results of the correlation analysis within each group for the MTC region. Observations between the stance-combined and swing-combined effector couplings indicates a gradual increase in the proportion of covariance, as the swing phase progresses across through the MTC region.

Between MTC-5% to MTC, most correlation coefficients appear to reach a minimum. For the stance-swing effector covariance, this minimum indicates an increase in negative correlations. For the stance-combined and swing-combined correlation, this indicates a decrease in positive correlation.

At MTC-5%, the young group displays mean ‘r’ and ‘p’ values: stance-combined, $r_{St\text{-}Co}$= .36, $p_{St\text{-}Co}$=.022; swing-combined $r_{Sw\text{-}Co}$= .45, $p_{Sw\text{-}Co}$=.018; stance-swing $r_{St\text{-}Sw}$= -.34, $p_{St\text{-}Sw}$=.063. The elderly group in contrast, at MTC-5%, mean ‘r’ and ‘p’ values are: stance-combined $r_{St\text{-}Co}$= .45, $p_{St\text{-}Co}$=.017; swing-combined $r_{Sw\text{-}Co}$= .28, $p_{Sw\text{-}Co}$= .082; and stance-swing $r_{St\text{-}Sw}$= -.35, $p_{St\text{-}Sw}$=.050. All the mean p values listed above are below p<.1. Both groups demonstrate a maximum negative correlation in the stance-swing coupling near MTC-5%.

At MTC, the young group displays mean ‘r’ and ‘p’ values: stance-combined, $r_{St\text{-}Co}$= .36, $p_{St\text{-}Co}$=.022; swing-combined $r_{Sw\text{-}Co}$= .43, $p_{Sw\text{-}Co}$=.009; stance-swing $r_{St\text{-}Sw}$= -.31, $p_{St\text{-}Sw}$=.080. The elderly group in contrast, at MTC, mean ‘r’ and ‘p’ values are: stance-combined $r_{St\text{-}Co}$= .45, $p_{St\text{-}Co}$=.013; swing-combined $r_{Sw\text{-}Co}$= .32, $p_{Sw\text{-}Co}$= .056; and stance-swing $r_{St\text{-}Sw}$= -.32, $p_{St\text{-}Sw}$=.062.

There are also differences when comparing the stance-combined coupling with the swing-combined coupling within both groups. At MTC the young group demonstrate the correlation coefficient of the swing-combined effector coupling is significantly higher than the correlation coefficient of the stance-combined effector coupling (p<.05). In comparison, when comparing within the elderly group, the correlation coefficient is significantly lower in the swing-combined coupling compared to stance-combined coupling (p<.001). This tells us that the stance effector of the elderly group explains significantly more variance in the combined effector, compared to the swing effector. In contrast, at MTC the young group swing effector explains significantly more variance in the combined effector compared to the stance effector.

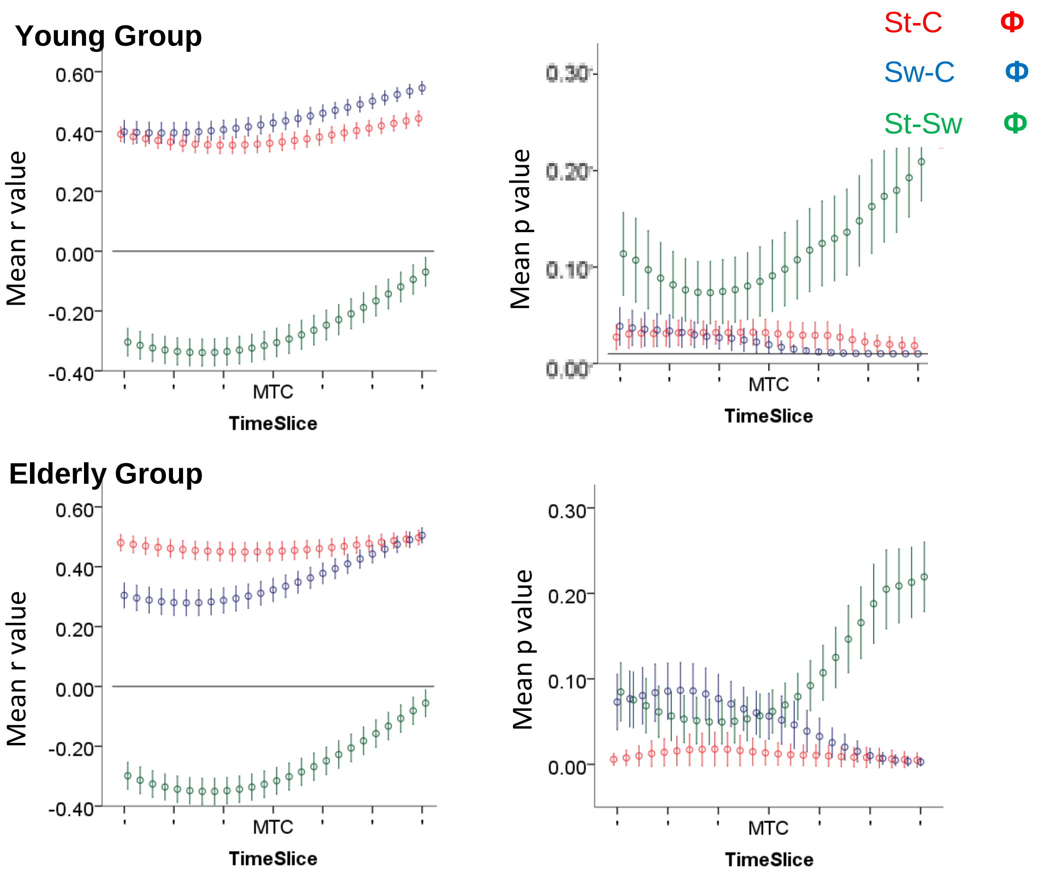


Figure 7.3.1.2

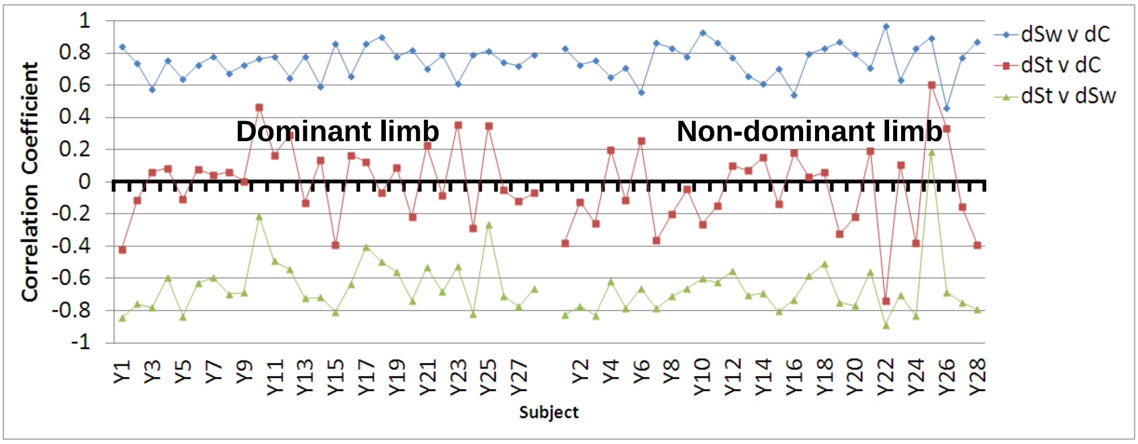
The coupling strength between paired effectors in the vertical dimension across the MTC region. The pairing code St-C, Sw-C, and St-Sw represents stance-combined, swing-combined and stance-swing effector couplings. The vertical axis represents the group mean correlation coefficient (r). The horizontal axis represents the time-slice of the MTC region in units of % swing phase. Each plotted group mean and 95% confidence interval (error bar) represents one time-slice of the region, spanning MTC-12% to MTC+12%. The data is taken from the trials representing the moderate response of trajectory change between MX1 and MTC.

In terms of a Nash equilibria, the above graphs indicate that the stance and swing effectors are not acting cooperatively during early swing i.e. MX1-10%), based upon the positive correlation. As the stance effector increases its vertical height, the swing effector also reduces its limb length (i.e. less negative displacement), and therefore both effectors are acting in the same way. This will create variability of the combined effector during this early phase of the MX1 region. In the MTC region, the stance-swing effector coupling begin to demonstrate cooperation, based upon the negative correlation which meets significance near MTC-5% ($p<.065$). These results imply that during the MTC region, as one of the stance or swing effectors acts to raise the toe height, the alternate effector acts to reduce the toe height. For Nash equilibria, this represents zero-sum behaviour because there has been no gain towards the common goal.

7.3.2 Correlation analysis between effectors when coordinating the change between MX1 and MTC

Figure 7.3.2.1 displays, for each participant and limb, the correlation coefficients between three effectors (St-stance, Sw-swing, C-combined) for the change in (‘d’) vertical displacement from MX1 to MTC. This plot indicates which effectors have a strong coupling relationship in their response from MX1 to MTC. Figure 7.3.2.1 demonstrates differences between subjects and between limbs. For example, there is a large difference in young case ‘Y25’ between dominant and non-dominant limbs, and when contrasting the non-dominant limb of case ‘Y25’ with non-dominant limb of case ‘Y26’. There are some extreme individual responses in the collaborative processes occurring between the stance and swing effectors when coordinating displacements between MX1 and MTC. In Figure 7.3.2.1, the non-dominant limb of the elderly and young both appear to demonstrate variable differences among participants within their group for covariance between ‘d-stance’ and ‘d-combined’. For example ‘Y22’ and ‘Y25’ in the young group, and ‘E6’ and ‘E19’ in the elderly group. This data from Figure 7.3.2.1 is summarised in Figure 7.3.2.2 and Table 7.3.2.1.

Young Group



Elderly Group

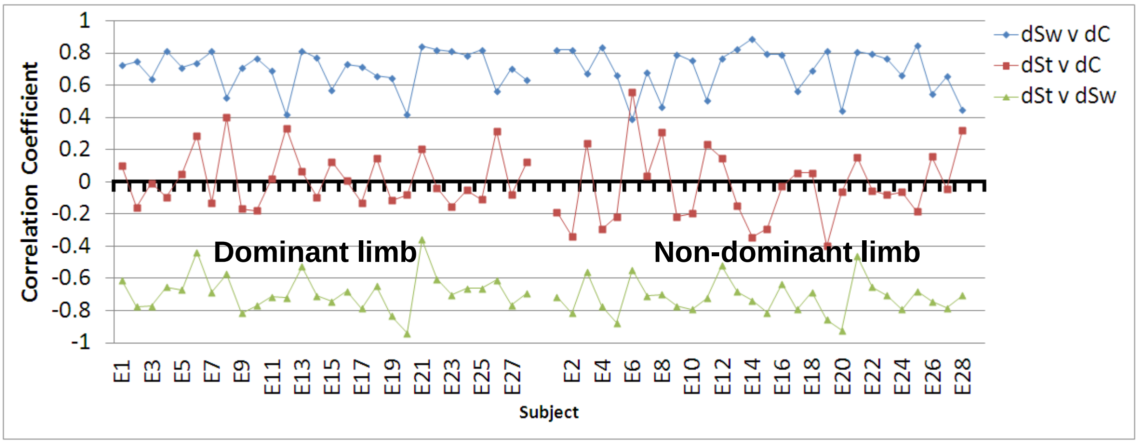


Figure 7.3.2.1

Graphs A and B show the young and elderly groups respectively, for correlation coefficient values when comparing between the three effectors time series generated from their vertical displacements between MX1 and MTC. The vertical axis is the correlation coefficient. Horizontal axis refers to subject case. Each subject's limb data is represented, the dominant limb (displaying case ID by even numbers) precedes the non-dominant limb.

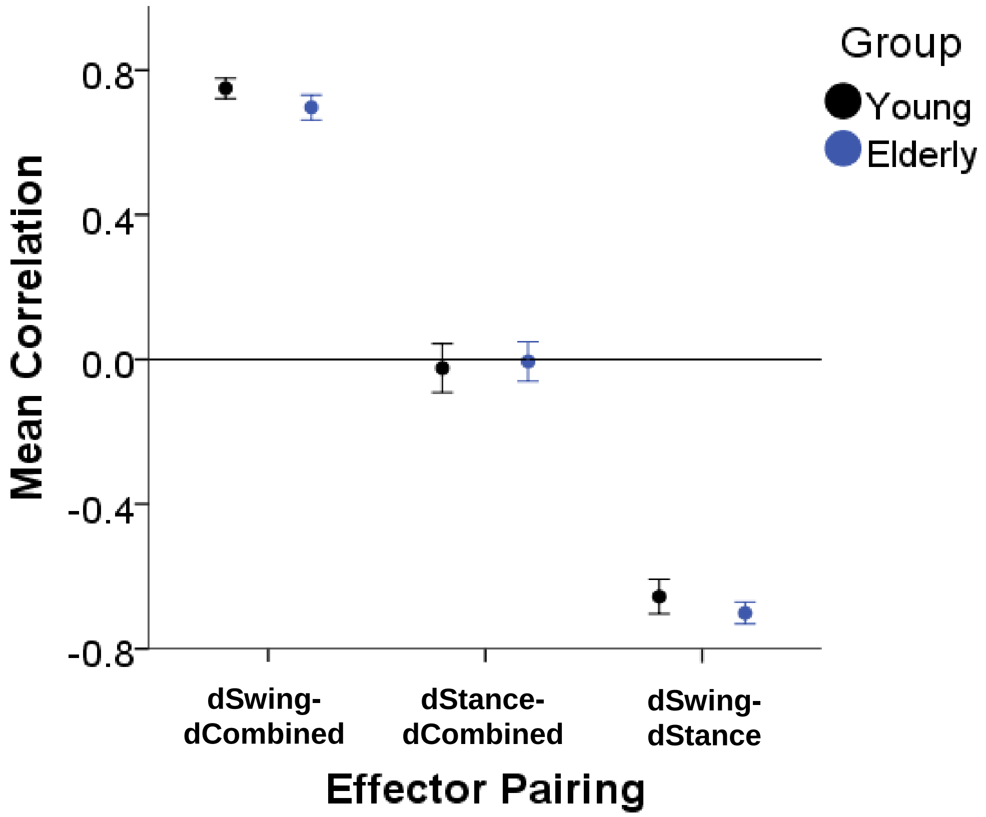


Figure 7.3.2.2

Between group differences for correlations between effector pairings for the variable 'effector vertical displacement between MX1 and MTC'. Limb data is pooled, and between limb data is presumed independent cases. Group means ± 95% confidence intervals are displayed. Between group significance is listed in Table 7.3.2.1 below.

Table 7.3.2.1 reports the significant differences for Figure 7.3.2.2. A paired t-test examined between-limb differences within the groups, however, there were no significant differences. For example, the difference between the dominant and non-dominant limb for 'dC v dSt' was not significant ($t(27)=1.39$, $p=.17$). There is nearly a significant difference when comparing the young and elderly within their dominant ($p= .102$) and non-dominant ($p = .095$) limbs for the correlation between combined effector displacement and swing effector displacement ('dC v dSw'). When pooling the limbs together, there was a significant difference between the groups when compared for the combined-swing coupling of the MX1-MTC effector response ('dC v dSw'). This tells us that in the young group, the swing effector change between MX1 and MTC explains significantly more of the variance in the combined effector change. This is consistent with results so far, such that the swing effector in the young group explains more variance in the combined effector when compared to the elderly group. However, when examining the change in the combined effector between MX1 and MTC, both groups demonstrate that the swing effector explains significantly more of the variance in the combined effector change, when compared to the stance effector change.

Table 7.3.2.1

Average regression coefficients for the vertical displacement changes between MX1 and MTC in the effectors of combined, stance and swing (listed below as dC, dSt, dSw respectively.

	Dominant Limb			Non-Dominant Limb			Pooled Limb Data		
Coupling	Young	Elderly	p	Young	Elderly	p	Young	Elderly	p
dC v dSt	.022	.021	.972	-.070	-.032	.581	-.024	-.006	.670
dC v dSw	.744	.699	.102	.756	.695	**.095***	.750	.697	**.019****
dSt v dSw	-.633	-.683	.190	-.679	-.720	.333	-.656	-.701	.110

* indicates significant difference at p<.01; ** indicates significance at p<.05

7.3.3 Serial correlations in the time series of the effector displacements between MX1 and MTC

The intention of the locomotor system to make a correction to the state of an effector trajectory, based upon an effector's initial state at MX1, and the need for intervention upon a particular effector trajectory to adjust course prior to MTC, was assessed by applying the DFA to effector's vertical displacements. Figure 7.3.2.1 shows the DFA values for the time series of relative displacements made by the three effectors between MX1 and MTC.

There were no significant differences found between the groups in either limb. For example, the between-group difference within the non-dominant limb combined effector was not significant (p = .15), and this was not significant when the limbs were pooled (t(1,110) = 1.63, p=.105).

When examining within group differences, from paired t-test, there are significant differences in the young group, that do not occur in the elderly group. For example, there is a significantly lower persistence for the change in combined effector when compared to the swing effector of the young group. This difference is not significant for the elderly group. Again, this supports results from Chapter 5, indicating that the persistence of the combined effector is significantly lower than the swing effector in the young group, whereas, this isn't the case for the elderly group.

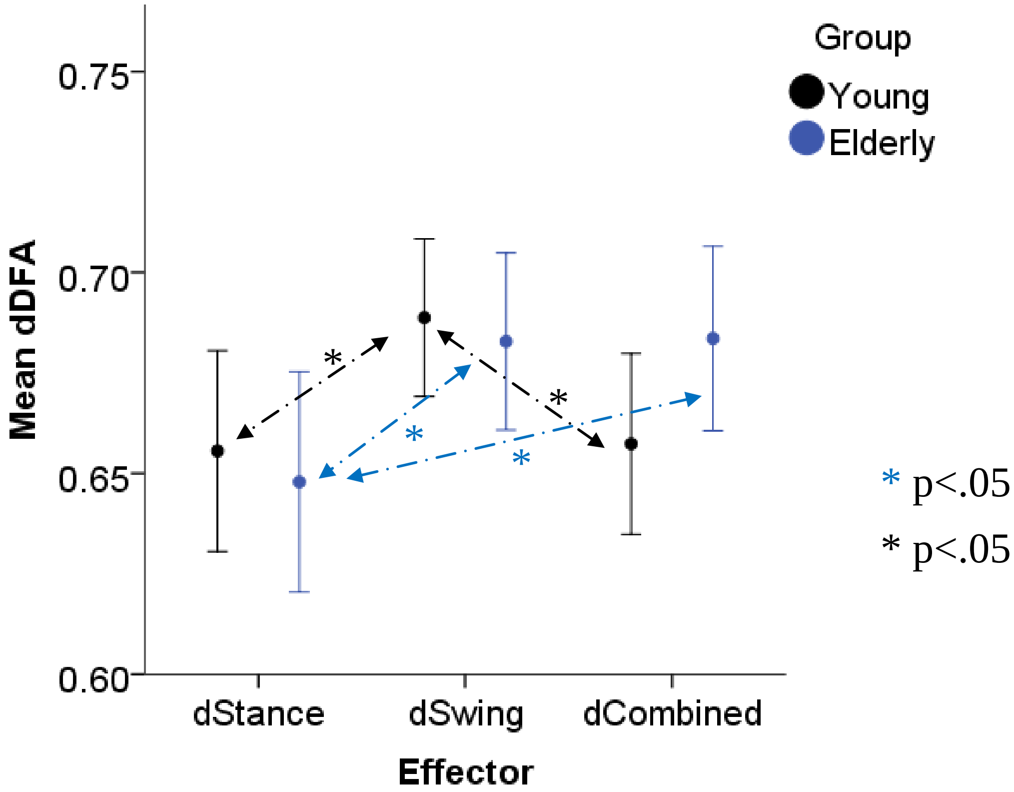


Figure 7.3.3.1

Between-group differences for correlations of the DFA, for the variable 'effector vertical displacement between MX1 and MTC'. Group means ± 95% confidence intervals are displayed. Asterisk represent significant within-group differences, p<.05.

7.3.4 Correction gain made between the stance and swing effectors to achieve nominal state at MTC based upon initial errors of the combined effector at MX1

The correction gain coefficients for each participant and limb, and for each effector, are displayed in Figure 7.3.4.1A-B. The correction gain made by the stance effector is indicated by red points, and the swing effector correction gain is indicated by blue points. For the young group, the graph from Figure 7.3.4.1A demonstrates that there are more points scattered below the vertical axis zero reference line. These 'negative correction gain correlations' tell us that when low MX1 values occur, there are large positive changes made by the effector. There are notable between subject variations within the young group. For example, in the non-dominant limb of the young group, both 'Y22' and 'Y10' display large correction gains by both effectors, whereas 'Y14' displays the opposite trend. This indicates that for subjects 'Y10' and 'Y22', the large gains made by the swing effector are 'off-set' by the decrease in gain made by the stance effector. This could be interpreted as a 'lazy' stance effector and the locomotor controller relies upon the swing effector. This contrasts with subjects who demonstrate relatively equal correction gains between the effectors. For case 'Y14', the correction gain in the combined effector between MX1 and MTC is due to the stance effector assignment (r_{St}=-.40), compared to the swing effector (r_{Sw}=.05). For 'Y14', the loco-sensorimotor controller awards responsibility to the stance effector to make the correction.

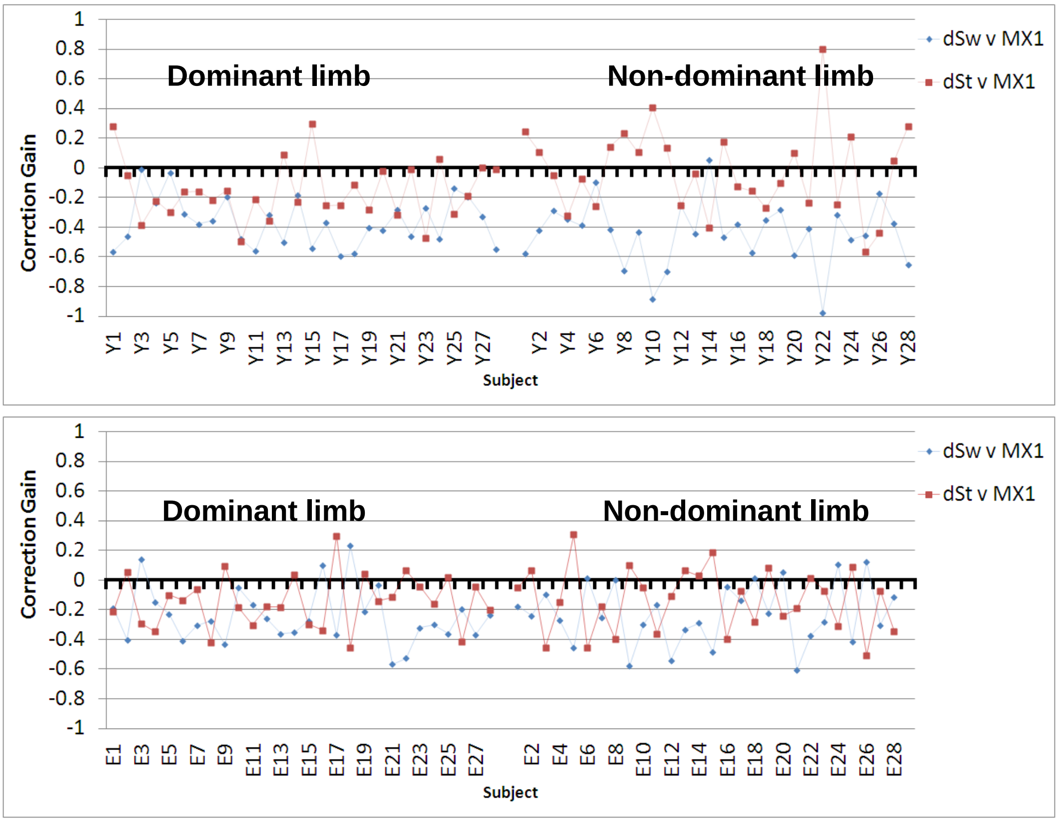


Figure 7.3.4.1

The correction gain plotted for the stance and swing effectors for each subject and limb. The correction gains for the swing effector and stance effector are plotted in blue and red respectively.

The swing effector has a correction gain coefficient significantly larger than the stance effector correction gain for both groups, when the limbs are pooled. The correction gain for 'dSw v MX1' and 'dSt v MX1' in the elderly group is r_{dSw}=-.24 and r_{dSt}=-.14 (p=.046). For the young group, the correction gains are significantly different, r_{dSw}=-.40 and r_{dSt}=-.09 (p<.001). When the limbs are separated into dominant and non-dominant, the young group demonstrates correction gain differences that are still significantly different between the effectors (i.e. p_{Dom}=.002, p_{NonDom}<.001). In contrast, the elderly group demonstrate no significant differences for correction gain between the effectors in either within-limb comparison (i.e. p_{Dom}= .115, p_{NonDom} = .215).

Table 7.3.4.1 summarises the data from Figure 7.3.4.1 of the correction gain coefficients. Between-group differences are indicated by p-values for within limb comparisons, and for pooled limb comparisons. The young group demonstrates a significantly ($p<.05$) larger correction gain for the swing effector when compared to the elderly group within the dominant and non-dominant limbs. The elderly almost demonstrate a significantly larger correction gain by the stance effector for the non-dominant limb ($p=.104$). However, the correction gain symmetry is significant for the non-dominant limb ($p<.1$), and when limbs are combined ($p<.05$). This result indicates that when altering the combined effector trajectory between MX1 and MTC, when compared to the young group, the elderly swing effector does not have the same effect on the outcome.

Table 7.3.4.1

Average regression coefficients for the vertical displacement changes of the stance and swing effectors predicted by MX1 position (abbreviations: Y is young group, E is elderly group).

	Dominant Limb			Non-Dominant Limb			Pooled Limb Data		
	Y	E	p	Y	E	p	Y	E	p
MX1 – dSt	-.159	-.144	.764	-.021	-.134	.104	-.090	-.139	.257
MX1 – dSw	-.364	-.246	**.015****	-.441	-.228	**.000****	-.403	-.237	**.000****
Gain Symmetry	-.626	-.536	.292	-.653	-.503	**.087***	-.640	-.519	**.048****

* indicates significant difference at $p<.1$; ** indicates significance at $p<.05$

Figure 7.3.4.2 displays the relationship between the combined effector's MX1-MTC correction gain made by the stance effector ('MX1 v dSt') and the coupling strength between the stance and swing effector ('dST v dSw'). There is a negative correlation for both the young and elderly groups. The best fit line representing the elderly group is $r=-.2$, and for the young group it is $r=-.42$. For the young group, the coupling between the stance and swing effector explains 18% ($r^2=.18$) of the variance in the correction gain made by the stance effector. The negative regression line indicates that when the correction gain for the stance effector increases, it is associated with increased cooperation between the stance and swing effectors (i.e. negative 'dSt v dSw' correlation). This means that the stance effector (or swing effector) will have limited correction gain when stance-swing effector cooperation is strong.

A strong cooperation between the stance and swing effectors represents a 'zero-sum gain' in the combined effector. Proportionately large displacements made by the stance effector are associated with proportionately small displacements made by the swing effector, and vice-versa. This indicates Nash-equilibria, where mutual cooperation exists between the stance and swing effectors, and the gain on the combined effector by the stance or swing effectors is minimal when cooperation is strongest. This is also apparent in Figure 7.3.1.2 when observing the effector coupling correlation, i.e. near MTC-5% the stance-combined minima occurs almost simultaneously with the stance-swing 'coupling' maxima. The young group demonstrate that strong cooperation between the stance and swing effectors is associated with a reduced stance-effector correction gain. However, Figure 7.3.4.2B shows that the elderly correction gain does not have a clear association with the coupling strength. The difference between regression lines of Figures 7.3.4.2A-B is almost significant, (z_{diff}=1.26, p=.104).

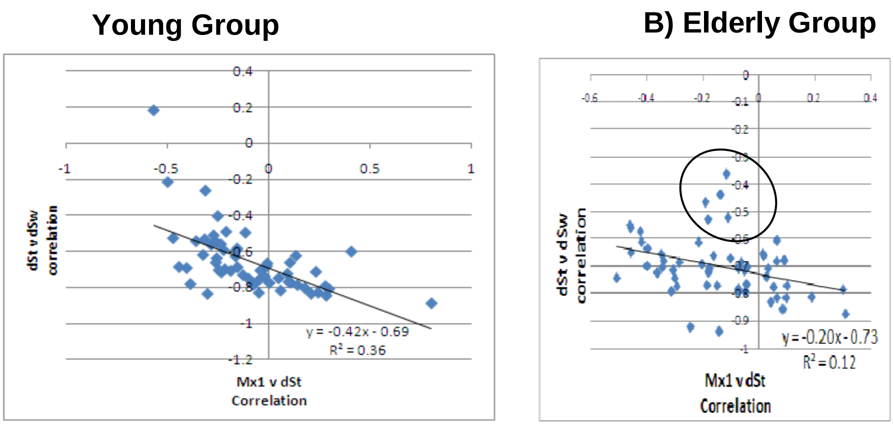


Figure 7.3.4.2

Showing the relationship between correction gain of stance and the coupling strength. Data is for the young group on the left and the elderly group on the right. An increase in stance correction gain is represented by negative values on the horizontal axis. An increase in 'dSt-dSw' coupling is indicated by negative values on the vertical axis.

The benefit of an effector system based upon mutual cooperation prior to MTC ensures a policy of stability in the combined effector. The negative consequences can arise when the state of the combined effector trajectory needs to change. If the cooperation between the stance and swing effectors is hard-wired in the neural circuitry, then at this state-relevant moment of the swing-cycle, the effect on stability will be optimal, however, the effect on flexible change in trajectory state will be compromised. Table 7.3.2.1 showed a stronger coupling between the stance and swing effectors (at p=.11), compared to the young, indicating a strong mutual cooperation that will lead to improved combined effector stability, but less adaptability.

7.4 Chapter 7 results summary

Below are the results related to the null hypotheses listed at the beginning of the chapter.

7.4.1 Hypotheses of synergies between effector systems

- Null hypothesis 7.1 There will be no significant difference between the young and elderly (IV) in the effector correlations (DV) within each state-relevant time-slice of the swing phase.
    - The null hypothesis was rejected. The coupling between the stance-combined and the swing-combined were significantly different between the groups during the MX1 and MTC regions.
    - The null hypothesis was accepted for the stance-swing coupling.
- Hypothesis 7.2 There will be no significant difference between groups when comparing between swing-combined and the stance-combined correlations at states MX1 and MTC;
    - The null hypothesis was rejected. There was a significant difference between the groups when comparing the difference between the stance-combined and the swing-combined correlations.
- Null hypothesis 7.3 There will not be a significant difference between the young and elderly when comparing between the three paired effect couplings.
    - The null hypothesis was rejected. There was a significant difference between the groups in between-effector coupling (covariance) on the change in effector states from MX1 to MTC.
- Null hypothesis 7.4 There will not be a significant difference when comparing between the young and elderly on the persistence associated with changes to effector states between MX1 and MTC;

    - The null hypothesis was rejected. There were significantly different values of persistence between the groups.

- Null hypothesis 7.5 There will not be a significant difference between the groups for the correction gain of the stance effector and swing effector on the combined effector state.
    - The null hypothesis was rejected. The young group applies a significantly different correction gain.

- Null hypothesis 7.6 There will be no significant difference between the groups when comparing on the dependent variable of effector correction gain.
    - The null hypothesis was accepted within the dominant and non-dominant limb.
    - The null hypothesis was rejected when limb data was pooled. There was a significantly different correction gain asymmetry between groups.

Chapter 8

Discussion

8.1 Introduction

Information about how the loco-sensorimotor control system manages the lower limbs to control the toe clearance task during walking is important for falls prevention because of the association toe clearance has with tripping risk. The general aim of this thesis was two-fold: first, to investigate how the toe clearance task is controlled during walking; second, to determine if natural ageing processes in females are associated with changes made to this trip-related walking task. The analysis objective was to measure parameters of lower limb coordination and control, which determines the performance of the toe clearance task. A combined kinematic chain (effector) described the collective segmental kinematics of the swing and stance limb during the swing phase of the gait cycle. Studies of gait control argue with evidence that the body plans and executes movements based upon information obtained about collective details of a system, rather than component parts of a system (Lacquaniti, Ivanenko and Zago, 2002; Ivanenko et al., 2007). Congruent with these findings, the current study analysed kinematic coordination and control of three effector systems representing the collective function of the limbs: 1) stance effector; 2) swing effector; and 3) the combined effector representing the collective elements of both the stance and swing effector.

In attempting to investigate the task of controlling toe clearance during walking, this thesis has incorporated general ideas from various research fields that investigate movement control. The first important idea is that the loco-sensorimotor system is dynamic, such that both passive dynamics and control dynamics are intricately integrated by the system during the management process that ultimately shapes the outcome of movement trajectories (Grillner, 2003; Collins et al., 2005; Hausdorff, Yogev, Springer, Simon and Giladi, 2005; Takakusaki and Okumura, 2008). The second important idea is that the nature of variability arising from systems generating repetitive movement trajectories has structure (Hausdorff et al., 1996) and this appears to be task-relevant (Scholz and Schoner, 1999; Dingwell et al., 2010). The third important idea is that trajectory formations of goal-directed movements (i.e. toe trajectories) emerge from elaborate integration of state-estimation (perception) and motor command (action) selection. To optimise task performance, this dual process of perception-action is mediated by a central nervous system controller that seeks to exploit a repertoire of solutions in its embodied multi-dimensional workspace so that ‘unwanted’ body-state behaviours are minimised (Todorov and Jordan, 2002; Kording and Wolpert, 2004). The mediation of this process is proposed to be reflected in the persistence-likelihood (DFA; Chapter 5 and section 7.33) statistic when comparing between the three effector systems investigated in this thesis.

The outcomes of all results indicate behaviour of the loco-sensorimotor system that is actively managing the toe clearance task. The toe clearance task has dynamic context, and the loco-sensorimotor controller will need to attune its control ‘utilities’– predictive control (management by sensory inference), reactive control (management by task-specific reflexes) and biomechanical control (management by exploiting musculo-skeletal properties) – to passively and actively manage the task (Wolpert, Diedrichsen and Flanagan, 2011). The integration of these three control ‘utilities’ is likely to be expressed in the statistics of control investigated in this thesis. Although, the interpretation of the control process will be broadly related to ‘passive dynamics’ and ‘control dynamics’ because the interaction of the three types of control ‘utilities’ is too complex to draw explicit links with the results obtained in this thesis. For older adults performing the toe clearance task, the gait controller needs to consider optimising risk-avoidance by applying a cost policy that penalises low toe trajectories. This appears to be captured by the skewness and variance statistic (Chapter 4 and section 5.9). An older adult also requires stable gait trajectories, and yet have the ability to flexibly transition, with efficiency, between repertoires of alternately stable trajectories for enhancing task solutions. This ability is proposed to be captured by the statistics of persistence (DFA; Chapter5) and effector covariance (UCM; Chapter 6).

At mid-swing of walking, there is a dual control problem faced by the gait controller, which is to optimise upright postural stability and provide 'risk-free' toe trajectory. As opposed to sequential ordering of task-goals, when these two task-goals appear simultaneously there is a cooperation dilemma between the stance effector and the swing effector. The stance effector is ultimately responsible for posture stability, and the swing effector contributes negligibly to this task-goal near mid-swing. Alternatively, the stance effector can contribute to the task-goal of toe clearance, and cooperate with the swing effector to achieve the goal. However, this can compromise the stance effector influence on upright posture. The management of this process was captured by the correlation analysis between effector systems in Chapter 7.

So far, the literature is limited with investigations about how the gait controller manages the toe clearance task within the context nature of walking sub-task dynamics. From any stride-to-stride transition, a gait controller might fluctuate between multiple policies of control that relate to optimising solutions for disparate tasks, or where one task involves multiple variables with competing costs that require minimising. For example, disparate tasks at mid-swing are the simultaneous goals of posture control and toe clearance. For toe clearance, there can be competing variables with costs that need to be minimised, such as minimising ground contact risk on one hand, and on the other hand reducing the cost of energy and movement error. In this context, energy efficient and precise movement patterns are not necessarily risk-free, such that the controller faces the dilemma of compromising energy and precision for low-risk. However, low-risk is potentially associated with effort and movement error.

While there are no known previous studies of human walking which have measured the toe clearance task in the dynamic context described above, there are science advances in other fields developing theories of dynamic control of human gait. Theories from hybrid limit cycles and stochastic optimal feedback control are useful for understanding how the loco-sensorimotor controller performs dynamically. These theories consider the embodied neuro-anatomical redundancy of the loco-sensorimotor control system and are successful at simulating the effortless, agile and accurate properties of human walking (e.g. dynamic walking organisation). Recently, implementing these theories and physically realising the algorithms to produce agile, efficient and accurate robotic systems have shown to be successful (e.g. Todorov et al. (2011)). Studies in this framework have also investigated how risk is optimised in movement systems (Nagengast et al., 2010). The current study approached the interpretation of the control and coordination results relative to the toe clearance task in congruence with the general ideas of stochastic optimal feedback control and hybrid limit cycles. An attempt is made in this discussion section to also appraise other relevant gait studies with empirical findings through this theoretical framework of optimal feedback control and dynamic systems.

This study undertook the approach to investigating toe clearance performance by considering dual tasks and competing cost policies as described above. This is a new approach to investigating toe clearance performance. The applications to data collection and kinematic representation are standard biomechanical methods. The methods for describing effector control and coordination are new to the description of toe clearance performance. First, the reconstruction of the biomechanical model into co-adaptive effector systems is a new recommendation in gait and posture analysis because the motor system has shown to respond to inferred task-relevant variables rather than anatomical variables (Ivanenko et al., 2007; Lockhart and Ting, 2007). Patterns of covariance within the hind-limb of cats show organisation based upon feedback from limb length and orientation, which is coded in the neural structures of the dorsal cerebellar spinal tract (Bosco et al., 2006). Therefore, this study re-organised the biomechanical model of the lower limbs into three effector systems to analyse the way a gait controller organises the task performance with respect to a collective of anatomical elements making up a ‘limb’ (Todorov, 2004; Schoner and Scholz, 2007; Wu and Popovic, 2010). Each effector was represented by a three-dimensional vector, and its point of application and direction were determined by the end-points of the effector system. The stance effector was composed of the shank, thigh and pelvis segments within the vector span of the stance foot centre of mass and the swing hip. The swing-limb effector system was defined as a vector directed from the swing-hip position to the swing-toe position (i.e. spanning the thigh, shank and foot segments). The combined effector system was defined as the collection of the segments within the span of the two effectors. A three-dimensional description of each of these three systems was constructed from a six-degrees-of-freedom biomechanical model. The three effectors were investigated both independently and inter-dependently in three-dimensional space. The segment elements within the effector systems were also investigated independently and inter-dependently in three-dimensional space. Second, this study overcomes the dilution of state-to-state information in cycle-to-cycle gait trajectories, which are common when gait cycles are time normalised to one or two repetitive events of the gait cycle. This thesis ensured that time points along the gait trajectory reflect between-cycle states relevant to the toe clearance task. Within each swing-cycle commonly occurring reference states with task-relevance were identified and used for creating time slices at two regions, one set of time slices during early swing (maximum toe clearance, MX1) and the second region at mid swing (minimum toe clearance, MTC). These two separate regions of time were constructed to distinguish a sequence of postural states in time offsets relevant to task-relevant states MX1 and MTC. Each region centred upon the respective reference state and time slices of the region were based upon a percentage proportion of that particular trial's swing phase time. This captured the state-dependent repeatability between trials for analysing the between cycle varying details of the task. To account for different embodied neuro-muscular-skeletal features of the left and right combined effector systems, this investigation separated between dominant and non-dominant sides. This is a new approach to investigating toe clearance.

Research studies show that the locomotor control system represents the stance and swing effector vectors in terms of task-relevant dimensions of limb orientation and limb length (Lacquaniti et al., 2002; Ivanenko et al., 2007) and this closely relates to the anterior-posterior and vertical components of the effectors in the current study. The stance-effector in the current study represents the lower limb contribution to body centre-of-mass control. There has been limited attention given to the representation of lower limb effector systems, and the interactions between and within the effector systems, as a way of comparing ageing effects in gait patterns. Currently, studies have found in the elderly that minimum toe clearance at mid-swing (MTC) is more variable but it displays the same mean tendency (Begg et al., 2007; Mills et al., 2007). One study group has attempted to investigate the association between specific details of toe clearance and joint kinematics (Mills and Barrett, 2001; Mills et al., 2007). While studies generally show that the joint angles in the elderly show a different configuration (DeVita and Hortobagyi, 2000; Mills et al., 2007), the average measures of joint variability are less discriminating between young and elderly (Mills and Barrett, 2001; Kang and Dingwell, 2007; Mills et al., 2007) even when accounting for walking speed (Kang and Dingwell, 2007). This is an example of the gait controller only paying attention to task-relevant details and emphasises two related points, which need to be addressed for making a better comparison about how an effector system controls a task-relevant end-point position. First, investigations should consider how the measured details in a redundant system are relevant to task performance. Second, investigating co-variance between joint angles and limb endpoint position should be performed in mutual space so the mapping between end-point position and joint angle variations is congruent. The current study considered these points by applying the method of the Uncontrolled Manifold hypothesis (UCM) to both the posture controlling and the movement generating effectors of the stance and swing limb. So, rather than treating the limb as a set of independent components, this study investigated redundancy by the computing task-relevant and task-irrelevant details through the UCM.

Quantifying the performance of task goals during walking has typically been computed from average measures, and there is research based upon these measures which have made strong contributions to our understanding of the gait tasks (Ivanenko et al., 2004; Neptune et al., 2009a). However, it has been nearly 15 years since it was stated that stride-to-stride events of the walking cycle demonstrate fractal-like long-term dependence, and the fact that leading research still adopts average measures to investigate the locomotor control system demonstrates the challenge of knowing how this feature integrates into the biomechanics and motor control of gait. The challenge for gait studies is to adequately understand what serial dependence represents because the fact that it exists in different levels between different populations demonstrates its significance (Hausdorff, 2007; Hausdorff, 2009). If it is agreed in the gait research community that representing the serial dependence holds important information of gait control, the current challenge is then to sufficiently capture these details in pathological gait with greater efficiency. Currently, walking trials need to be of an extended duration to adequately assess persistence in the serial correlations. To overcome this limitation, there needs to be further understanding of what serial dependence represents for gait. Understanding serial dependence will also give models of pathological gait some comparative validity. The current study has contributed to this issue of understanding serial dependence in gait.

Lyapunov stability of gait is a concept related to walking pattern trajectories, which describe the system's ability to dissipate the effect of perturbations during a gait cycle or across cycles (e.g. Dingwell, Cusumano, Cavanagh and Sternad, 2001; Granata and Lockhart, 2007). Stability in this context does not refer to state-dependent task goals within the cycle. To measure state-dependent stability within a cycle, a method is needed to be applied at discrete time states. This current study did this by applying the DFA method at each time slice of the MX1 and MTC swing-phase regions. Lyapunov stability can provide a general appraisal of how effective the passive control dynamics are at dissipating the effect of small perturbations in aperiodic gait cycles across a period of time (Su and Dingwell, 2007). While the passive dynamic features of aperiodic walking are impressive at maintaining long-term stability, without active control the system can be guaranteed to eventually exit its region of stability and fall down. This emphasises the contributions of active and passive control dynamics to maintain the system in a desirable limit-cycle neighbourhood. The concept of passive and active control dynamics has not typically been directly implicated with the DFA method, apart from one known exception which suggests that increased serial persistence (i.e. large DFA exponent) is associated with reduced local stability (i.e. large Lyapunov exponent) (Jordan et al., 2009). There is a recent view of DFA, based upon developing theories from research of empirical and modelling evidence (Scafetta et al., 2009), suggesting that the DFA captures the neural stress or the controllers effort to find a task solution. Another recent view of the DFA is related to its ability to represent task relevant and task-irrelevant details of gait (Dingwell et al., 2010). Therefore, the DFA method can be hypothesised to be an indicator of the control effort displayed by the gait controller to actively regulate the passive dynamics expressed by the limit-cycle behaviour.

The DFA has a popular association with characterising 1/*f*-type noise, which is a ubiquitous property in biological systems (Peng, Havlin, Hausdorff, Mietus, Stanley and Goldberger, 1995; Hausdorff et al., 1996; Goldberger et al., 2002; Wagenmakers et al., 2005; Farrell, Wagenmakers and Ratcliff, 2006; Torre and Wagenmakers, 2009). Original views of the DFA are steeped in dynamic systems theory, which posits that 1/*f*-type noise emerges from a non-hierarchical self-organising system driven by non-linear coupling interactions of oscillatory 'agents' (Kelso, 1995). This is in-line with the self-organising view of passive limit-cycle behaviour described in gait (Kuo and Donelan, 2010) and the dynamic systems view of coordination (Kelso, 1995; Jirsa and Kelso, 2004). More recently, concepts of hybrid limit-cycle systems have come from research studies simulating aperiodic human walking and the interactions of the sensorimotor system in the attempt to design energy efficient, robust and agile robots (Manchester et al., 2011). These studies take inspiration from complex tasks performed by animals and humans with limited energy expense and incredible agility (Tedrake, 2009). In nature, it is firmly believed that animals and humans take full advantage of the passive dynamic attributes of their systems and this is considered an important feature to integrate in model designs of human gait (Collins et al., 2005; Byl and Tedrake, 2009; Iida and Tedrake, 2010; Tassa et al., 2011). In contrast to passive dynamics, the philosophical origins of stochastic optimal feedback control theory suggested that coordination emerges from an optimisation solution according to a hierarchical control policy and a given workspace (Todorov and Jordan, 2002). While the two fields of optimal control and dynamics systems may have appeared to be somewhat at odds, recent progress is demonstrating that both areas have actually been converging to improve understandings of dynamic control of walking (e.g. www.dynamicwalking.org).

A recently published theoretical model of stochastic optimal feedback control acting upon a limit-cycle attractor suggested that there is a control cost incurred when deciding to divert from a desirable passive dynamic attractor (Todorov, 2009). The model also incorporated the optimisation of state estimates and motor commands and, therefore, this theoretical model was congruent with the development of ideas being pursued in this thesis. For example, the gait controller respects the minimum intervention principle, the emergent gait patterns arise from optimisation, and limited active control is needed when the biomechanics of gait can take advantage of physical embodied attributes in the effector systems to form a relatively stable limit-cycle attractor. From empirical evidence, Jordan et al., (2009) suggested that a larger DFA may be representing the controller's efforts to regulate the passive limit-cycle trajectories of gait. However, their view was that a larger DFA scaling exponent was representative of increased effort by the central nervous system to control stability. This thesis has the alternative view, such that an increase in DFA (persistence) reflects less effort to control the limit-cycle trajectories, such that limit-cycle trajectories are free to persist across periods of strides. Diverting from the desirable passive dynamic attractor is costly (Todorov, 2009), and the DFA will demonstrate the willingness of the gait controller to alter the passive limit-cycle trajectory. This is not an unconventional theory, as there have been elements of this minimalistic view of control which relates to increased persistence that has been suggested in other gait studies (e.g., Scafetta et al., 2009; Dingwell and Cusumano, 2010; Dingwell et al., 2010).

Due to complexity in the human biological system the link between control and coordination is a challenging problem but the dual concept of passive and control dynamics provides a suitable context for understanding how they relate. Some questions about the relationship between control and coordination are raised in the following. In the loco-sensorimotor control system, does coordination of the redundant inter-segment linkage require controlling, or, does coordination of the inter-segment linkage emerge 'passively' to provide superior control over important performance task variables? In this latter case, at what stage will a control intervention act intending to improve coordination dynamics, actually hinder the task-outcome performance? What is the best way for a controller to coordinate a collective system to achieve two independent task goals relatively simultaneously; whereby, one goal requires the other system's components to be part of a shared cooperation (e.g. posture control and toe clearance control)? The optimal feedback control theory posits that coordination arises from an optimised solution (Todorov, 2004). In addition, the level of control required to find a solution will ultimately affect the expression of coordination stability and flexibility, because movement solutions are limited when there has been a reduction to the workspace dimensionality (Guigon, 2010). This study hypothesises that the ageing process naturally reduces the workspace dimensionality and the coupling of neural effort to control limit-cycle trajectories will in-turn affect the coordination details expressed by the effector systems. The view of the UCM hypothesis is subsequently that it expresses the outcome of the workspace dimensionality, which is determined by the control effort and the embodied attributes of the system.

The mechanistic hypothesis taken in this thesis is that walking behaves like a stable hybrid limit-cycle that is attracted to the passive dynamic properties of the system. The musculo-skeletal structures which act as inertial and stiffness constraints set up the 'un-controlled' passive dynamics. For understanding human walking, there is strong reason to respect the passive dynamics of the system and this is why the model of walking has involved models incorporating limit-cycle attractors (Iida and Tedrake, 2010). Certainly, the dynamics are not purely passive because of external and internal perturbations, particularly those cyclic perturbations caused by the heel-to-ground collisions. These perturbations are the basis for the need to actively control posture during walking. There is an energy cost for diverging away from the passive dynamics. Tassa and Todorov (2011) extend upon earlier work by Todorov (Todorov, 2009) to show that an optimal feedback control model can simulate walking by accounting for the passive dynamics and the stochastic control dynamics. From a physiological perspective control effort implies an increase in energy due to the active commands required by the gait controller (Todorov, 2009). Therefore, the system is encouraged to learn the boundaries established by the passive dynamics and to develop a repertoire of patterns that allow the system to remain within these attractor boundaries of stability. Finding passively stable states, which minimise the potential need for divergence, and in-turn minimise control interventions, is therefore the goal of the loco-sensorimotor system. The larger the divergence required from the desirable passive dynamic will in-turn result in a reduction to the dimensionality available in the solution space. This affects the searching capacity for an optimal solution and reduces redundancy and therefore flexibility.

The physiological hypothesis taken in this thesis is that synergies are outcomes of an optimisation problem, which translates as a process of expanding low dimensional information into a higher dimensional solution space for enhancing the optimisation of a solution. The philosophy of optimal feedback control is at odds with the observed dimensionality reduction believed to be the basis of forming synergies (d'Avella, Saltiel and Bizzi, 2003; Ting and Macpherson, 2005; Ivanenko et al., 2006; Lockhart and Ting, 2007). From a physiological perspective, how does the system increase dimensionality when the physiological evidence suggests the alternate need to reduce dimensionality (i.e. degrees of freedom) by selectively modulating various muscle synergies? In OFCT a muscle synergy can be explained as an outcome of optimisation and this requires the expansion of dimensionality. The low-dimensional task-related muscle groupings called synergies can be explained in OFCT through the duality relationship linking the optimal/Bayesian inference and the optimal control law (Todorov, 2008). To act in the best way possible to meet a task goal the state estimates and the control law are mapped to each other in the same process (Todorov, 2009; Wolpert and Flanagan, 2010). A neuro-physiological explanation for this mechanistic process has a plausible biological substrate in the cerebellum (Scott, 2004). Therefore, at the very least, the cerebellum maintains an active interest in processing body state information during walking. From the above discussion, it is reasonable to consider that walking is also the outcome of an optimal feedback controller. This hypothesis will form a basis to interpreting the results of this thesis.

To reduce the risk of trip-related falls in older adults, it is critical to understand if a change made to the kinematic configuration of the limbs is adopted as a necessary design to improve performance. This chapter will explore this issue. Already it has been suggested that the toe clearance task is a constrained optimisation problem involving minimising the probability of ground contacts and minimising energy expenditure (Begg et al., 2007). This thesis will propose that the toe clearance task be represented in the context of a concurrent goal, which is stabilising the body centre of mass; and to therefore emphasise the necessity for defining separate goals of the stance and swing effector systems. The toe clearance optimisation problem, therefore, should include the regularisation term of postural stability as results indicate that it has a strong influence on the task performance criteria. Maintaining movement trajectories, which are close to the passive dynamics of the system, are conducive to minimising energy expenditure and movement errors. The minimisation of toe-to-ground contact probability will incur a control dynamic cost when divergence from the passively desirable movement trajectory is required. It is most likely that this control dynamic cost will be larger for elderly persons because of sensory and motor noise effects (Christou and Tracy, 2006; Faisal et al., 2008), and also because of age-associated neural degeneration which can lead to reduced dimensionality of the neuro-muscular workspace (Hausdorff et al., 1997b; Goldberger et al., 2002; Vaillancourt and Newell, 2002). The expected loss in complexity of the ageing loco-sensorimotor system results in a change to the dimensionality of the workspace. The other factor which changes the workspace dimensionality is when the system diverges from the passive dynamic limit-cycle. So, whether by default or design, the outcome of reduced dimensionality can lead to actions evolving from deterministic processes and the search for an optimal 'cost-to-go' solution in a low dimensional workspace becomes taxing to the system.

Generally, the elderly will be challenged to act optimally in the face of dynamic uncertainty and will have 'more at risk' by the outcome of their decisions. This is expected to lead to increased control effort and an adjustment to the control policy that determines the nature of the performance criterion associated with the toe clearance task. This policy will ultimately face the dilemma of whether to reduce constraints that limit workspace dimensionality, and therefore allow an expansion of movement solutions – expressed by movement trends from trial-to-trial evolving like a random-walk process as movement solutions from previous trials will decay slowly (Verstynen and Sabes, 2011) – which is less taxing in terms of control effort, but the cost of this policy incurs less certainty in the movement outcome. To cope with this uncertainty a greater penalty could be invoked in the control policy, or alternatively, it will be appropriate to develop a new cost function that results in a new biomechanical pattern for performing the task.

The optimal 'cost-to-go' solution can be defined as the cumulative cost from a given starting position and acting optimally thereafter to a goal state. The optimal 'cost-to-go' can also be defined as the dual optimal/Bayesian inference of the state estimate using the ‘prior’ and ‘posterior’ probabilities (Todorov, 2008) which emphasises the link between sensory feedback and movement generation (Wolpert and Flanagan, 2010). The movement act is considered optimal according to a performance criterion. This might be likened to the given MX1 state and acting optimally thereafter to get to the desired MTC state. To make the ‘cost-to-go’ small, an error tolerance might be considered by the controller to allow workspace dimensionality to expand (facilitating solution abundance) and therefore relax the solution constraints during the ‘cost-to-go’ search function. Applying this reasoning to the walking performance of older adults, suggests that if they can perform the tasks in a better way by expanding the task solutions then increasing the level of error-tolerance of the state estimates should be applied. This of course comes at a cost of larger variance in task-relevant states. However, the optimal controller realises there is a balance of error and effort, by considering that there will be an increase in variance due to error-tolerance and a simultaneous reduction in signal-dependent-noise variance because of a reduced need for motor correction (O'Sullivan et al., 2009).

To perform the toe clearance task in an adaptable way, the loco-sensorimotor system will need to rely upon a complex high-dimensional neuro-biomechanical workspace which integrates multi-scale sub-systems. Central and peripheral factors within the loco-sensorimotor control system are thought to be determinants of workspace dimensionality: 1) health status of the neural network apparatus governing the sensorimotor system (Hausdorff et al., 1997b; Hausdorff, Cudkowicz, Firtion, Wei and Goldberger, 1998; Hausdorff et al., 2000; Hausdorff et al., 2001; Manor, Costa, Hu, Newton, Starobinets, Kang, Peng, Novak and Lipsitz, 2010); 2) cognitive function and task goals (Hausdorff et al., 1996; Dingwell and Cusumano, 2010; Dingwell et al., 2010; Manor et al., 2010); 3) passive biomechanical attributes (Gates et al., 2007). This capacity of the neuro-musculo-skeletal workspace can be improved in two ways. First, the loco-sensorimotor controller should utilise the regions of stable limit-cycle behaviour, by respecting the passive dynamics inherent to the natural biomechanical attributes of the system (Collins et al., 2005). This utility minimises the need for large correction gains from online sensory feedback. Indirectly, there appears to be evidence that when there is limited external stress, or limited corrective control dynamics, the system remains close to the desirable passive dynamic state (Scafetta et al., 2009; Manor et al., 2010; Tassa et al., 2011). Theoretically, the system will operate from an optimal workspace dimensionality and display higher levels of complexity. This is important for optimal and flexible movement performance (Todorov, 2009). The idea that when planning gait trajectories the loco-sensorimotor controller takes advantage of the 'open-loop' passive system as a way to reduce the dependence upon the 'closed-loop' active feedback system has only just started to be developed in the form of hybrid limit-cycle models (Iida and Tedrake, 2010; Tassa et al., 2011). The biomechanical attributes of a system are a significant contributor to control through the passive dynamic characteristics of the musculo-skeletal system (Srinivasan and Ruina, 2006; Gomes and Ruina, 2011). The role of lower level controllers in the intra-spinal networks may also be conducive to maintain high-dimensional workspace capacity (Scafetta et al., 2009; Iida and Tedrake, 2010). However, maintaining movement trajectories close to the desired passive dynamic attractor will not always be conducive to meeting a task goal, and divergence from the passive limit-cycle attractor will need to occur. Additionally, in a noisy system the stability of the passive dynamic is sensitive to perturbations, thus indicating a potential need to regulate the passive dynamics. The need to intervene and divert from a stable limit-cycle attractor will come at some control cost and by intervening the dimensionality of the workspace will be reduced because there now exists a target state to move towards (Todorov, 2009; Tassa et al., 2011; Todorov, 2011). An optimal solution costs more effort when searching through a workspace with reduced dimensionality (Todorov, 2009). Now this introduces the second way a controller can increase the dimensionality of the workspace and subsequently help to reduce the control cost. By re-adjusting the control policy, the need to intervene can be reduced as long as the movement performance meets the intended task goal. This re-adjustment acknowledges a potential compromise to other regularisation terms, such as trading control effort for movement error (Nagengast et al., 2010). Less intervention by the controller in this case means that the error tolerance will be relaxed and cause the task performance to become more variable. This will have a different effect on dynamic variability (e.g. DFA) and to the average variations (e.g. SD).

This final chapter will interpret the results from the perspective that the loco-sensorimotor system acts like a stochastic optimal feedback controller that uses the desirable passive dynamics of the system to its advantage. Therefore, gait trajectories may be considered to act more like a hybrid limit-cycle (Iida and Tedrake, 2010). The emergence of coordination is expected to be the outcome of this dynamic process.

8.2 Ageing effects on the kinematic states of the stance and swing limbs

8.2.1 Ageing effects on time-distance

In Chapter 4 (section 4.4) the time-distance parameters of gait demonstrated that the elderly participants walked with significantly reduced step time (i.e. significantly higher cadence), however walking speed and average step length was not significantly different between groups. Variability of step time was also not significantly different between groups. Increased step time variability is reported to have a strong association with reduced functional status of many sensorimotor capacities, including cognitive processing, reaction time, measures of balance, and muscle strength (Hausdorff et al., 2001; Hausdorff et al., 2005; Brach et al., 2007a; Callisaya et al., 2010). Walking speed has been associated with muscle strength measures (Callisaya et al., 2009). The 'no significant difference' between the groups found in this study, for step time variability and walking speed, supports that the elderly sample were displaying relatively healthy functional abilities.

The result of average step length and step width distance was not significantly different between the groups. However, the step-to-step variability of these parameters was significantly higher in the elderly group. Interpreting this result in light of step timing variability results can be made based upon findings from other research. In terms of associating the results with functional capacities of the central nervous system, sensory system, or motor system, there has been some slightly different views of what the variability found in time-distance measures of gait is actually reflecting. In the research literature, increased step width variability is associated with reduced balance ability (Owings and Grabiner, 2004; Brach et al., 2007a; Callisaya et al., 2010). Recently, step length variability has also been associated with balance ability (Callisaya et al., 2010). Muscle strength is not associated with step width or step length variability (Brach et al., 2007a; Callisaya et al., 2010). Gabell and Nayak (Gabell and Nayak, 1984) proposed a hypothesis twenty years ago that increased variations in step timing can be attributed to central processing dysfunction, whereas increased step width variations can be attributed to balance dysfunction. Recent studies suggest that this is not an exclusive association. Brach et al. (2007a) who tested a large sample of elderly people (n=558), found that reduced sensory function was actually associated with reduced step width variability. Brach et al also supported the theory that stance time variability is associated with central nervous system function, however, sensory function was not associated with step-time variability. Other studies find that step-time variability is related to many reduced sensorimotor functions, as well as reduced central processing function. From a large sample of elderly people (n=412), it was found that step-time variability was associated with balance function, while being almost significant with proprioception (p=.07) (Callisaya et al., 2010). Furthermore, the sensory measure of proprioception was associated with double stance time variability, but not step-width or step-length variability (Callisaya et al., 2010). Therefore, there are certainly some anomalies found in the results of these population-based studies that have linked gait variability to sensorimotor function. It would therefore be misleading to propose that the significantly higher step width variability found in the elderly sample of this thesis, is exclusively reflecting sensory function. It is, however, most likely to be representing reduced balance ability during walking, which is the integration of many sensorimotor systems.

Revealing that step length is negatively skewed is a new finding in the gait literature, and whilst both groups demonstrated mean negative skewness for step length, the elderly displayed a significantly larger negative skewness. While the mean step length (normalised to limb length) was not different between groups, the significant larger negative skewness of the elderly indicates a control policy that applies a larger penalty for longer step errors, relative to the penalty applied for errors in short steps. Shorter steps have been associated with persons with balance instability while walking (Maki, 1997; Menz, Lord and Fitzpatrick, 2003). This link appears to infer that the elderly group were more unstable while walking.

The reduced persistence found in the step-width parameter (section 4.4) is another new finding in the gait literature. The contrast between persistence-likelihood measures, related to step-length and step-width, reinforces the theory that there is relatively greater control applied in the medio-lateral dimension of walking (step-width), compared to the anterior-posterior direction (step-length) (Bauby and Kuo, 2000; Kuo and Donelan, 2010). This also supports the theory that reduced persistence is reflecting processes of active control. When comparing persistence-likelihood between groups, there were no significant differences, suggesting that the type of control processes managing step-to-step performance was not applied differently between groups.

8.2.2 Average configuration of the stance, swing and combined effector states

There is a limit to the studies investigating three-dimensional trajectories of effector systems as defined in this thesis, except for studies investigating the vertical dimension of the combined effector (i.e. toe trajectories; (e.g. Osaki et al., 2007)) and swing effector (e.g. Ivanenko et al., 2002; Ivanenko et al., 2007). As such, the results describing the effector systems are relatively new to the gait literature.

8.2.2.1 Configurations in the direction of toe clearance

The combined effector was not significantly different between the groups for comparisons on displacements represented by limb-length normalised units and absolute metric units at the MTC event. This is consistent with other studies investigating vertical 'toe' states at MTC (Winter, 1992; Mills and Barrett, 2001; Begg et al., 2007; Mills et al., 2007). However, at MX1, when the limbs were pooled together, there was a significantly lower combined effector state in the elderly. Many studies that have investigated toe clearance had over-looked details at the MX1 event (e.g. Winter, 1991a; Begg et al., 2007; Mills et al., 2007; Osaki et al., 2007). The results also showed a significant difference when comparing between groups on the combined effector trajectory from MX1 to MTC (section 4.6.4). This suggests that the combined effector trajectory of the elderly follows a path between MX1 and MTC that has a significantly reduced decline in vertical position compared to the young group. In other words, when compared to the young group, the elderly combined effector trajectory started at MX1 in a significantly lower position and then finished at MTC by following a trajectory profile that was significantly 'flatter'. So, while there was no difference at MTC, there was a significant difference in the approach to MTC. In a study that investigated gait pattern changes comparing normal walking conditions with a condition of 'fore-warning' of tripping risk, the mean 'toe' trajectory found in trials associated with 'fore-warning' reflected a flatter approach to MTC, but this was mainly due to a significant increase of the 'toe' at MTC (Pijnappels, Bobbert and van Dieen, 2001). The authors did not investigate interactions between MX1 and MTC, and the 'toe' position was in fact representing the posterior foot position of the fifth metatarsal head. Had the 'toe' position of the study actually been representing the distal inferior position of the foot, the apparent 'toe' trajectory will have appeared differently and potentially provided different details between conditions about the states of the 'toe' trajectory at MTC.

The general configuration of the stance and swing effectors (displacements normalised to limb-length) was significantly different between groups. The non-dominant limb of the elderly displayed significantly 'longer' vertical displacement for both the swing and stance effector at MX1 when compared to the young group, and when the limb data was pooled together this trend remained significant (Figure 4.6.2.1 and 4.6.3.1).

Section 4.6.4 showed that the elderly group had a 'longer' stance effector in the vertical dimension at both MX1 and MTC (when pooling the limb data). The elderly also demonstrated a significantly 'longer' swing effector length in the vertical dimension at MX1. When effector states are averaged between MX1 and MTC, the stance effector was significantly 'longer' in the vertical direction; therefore, representing a significantly higher (scaled to limb lengths) swing hip position. The vertical velocity of the stance effector was increased significantly for both the dominant and non-dominant limbs in the elderly during early stance (MX1-10%; Figure 4.7.2.1). The swing effector of the elderly was also significantly 'longer' in the vertical dimension, indicating a significantly longer swing limb in the elderly. For the elderly, this suggests more work is being done by the stance musculature and less work by the swing musculature (Neptune et al., 2008). This would make the walking pattern adopted by the elderly more metabolically costly, because the work to raise the body centre-of-mass is significantly greater than muscle work to progress the swing limb (Neptune, Zajac and Kautz, 2004). Therefore, the configuration of the stance and swing effectors adopted by the elderly appears to compound upon the muscle work differential between stance and swing limbs. A strategy of raising the stance effector and lengthening the swing effector appears to go against a control policy of optimising energy efficiency. However, this could represent a functional compensation for lower extremity muscle weakness in the plantar-flexors. Simulation studies show that soleus activity during early stance, and gastrocnemius activity during late stance (pre-swing), transfer energy to provide upright support through the stance effector and forward progression to the swing effector, respectively. Studies on adaptations due to ageing suggest that the trunk and hip extensors increase power to compensate for muscle weakness found in the lower extremity of elderly groups (McGibbon and Krebs, 2001). Therefore, the states of the effector trajectories observed might be a combined effect of a biomechanical compensation due to muscle weakness and an updated control strategy that accommodates these changes.

8.2.2.2 Configurations in the direction of forward progression

In the anterior-posterior direction, the elderly demonstrated that the effectors were significantly more posteriorly oriented. For the combined effector and the swing effector, the elderly toe position was significantly more posteriorly oriented at MTC with respect to the stance foot (combined effector), and with respect to the swing hip (swing effector), and this was significant for both the dominant and non-dominant limbs. The stance effector was also found to be significantly more posteriorly orientated at MTC, but only for the non-dominant limb. With respect to the elderly group, the collective effect of a posterior orientation in the stance and combined effector represents that there is a greater proportion of lower-limb mass above the stance foot at MTC. However, the significant anterior tilt of the pelvis would suggest that the trunk position of the elderly is tilted forwards, that is unless they are displaying significantly greater trunk extension than the young group, which is unknown. Irrespective of the contribution made by the upper body mass, the collective configuration of the lower-limbs in the elderly suggests more of the body mass was being positioned with a significantly more vertical alignment above the support envelope area (i.e. stance foot) at MTC. This displays risk-averse behaviour in the non-dominant limb of the elderly because of the alternative scenario; if the body centre-of-mass is forward of the support envelope area, then the application of ground forces to cause an external moment that resurrects postural stability – providing angular acceleration to re-align the centre-of-mass above the support foot (i.e. centre-of-pressure) – will be a more demanding task. Also, if a trip occurs at MTC and the body centre-of-mass is positioned further forward of the stance foot, greater angular acceleration (via external moment) will be required to arrest a significantly greater forward angular momentum of the trunk. It is reported that older adults have less strength and muscular coordination to restrain forward angular momentum following a trip (Pijnappels et al., 2005).

8.2.2.3 Configurations in the direction of medio-lateral balance

The stance effector of the elderly showed significantly greater medial displacement relative to the stance foot during MX1 and MTC for the non-dominant limb. This can be explained by significant differences of frontal plane segment angles of the non-dominant limb compared to the young group. The elderly have significantly reduced pelvic obliquity and significantly greater vertical alignment of the thigh. This posture requires work by the hip abductors, and while this provides greater vertical position of the swing hip, it also effects an external force that cause an increase in medial acceleration (towards unsupported side) of the body centre of mass (Maki, McIlroy and Fernie, 2003; Pandy, Lin and Kim, 2010). It was found that the elderly have significantly larger velocity (normalised to limb-lengths(LL)/s)of the stance effector in the direction of the unsupported side (Figure 4.7.2.3). Therefore, under this postural configuration, there is a higher swing hip position, but the expected trade-off is medio-lateral instability. The finding that the velocity of the swing effector was increased significantly towards the direction of the unsupported side (Figure 4.7.3.3) is likely to be an 'angular momentum conservation' response to the stance effector (Hofmann et al., 2009). For the dominant limb, we observed a similar response when comparing between the stance effector velocity and swing effector velocity in Figures 4.7.2.3 and 4.7.3.3, respectively.

In summary, the elderly displayed a higher vertical position of the swing hip in the non-dominant limb, however, the trade-off is that there is an increase in medial motion (with respect to stance foot) towards the unsupported side during the MX1 region of the swing cycle. In response, the swing foot displayed behaviour that suggested a compensation role by undertaking trajectories with significantly increased lateral velocity.

8.2.3 Age-related changes in the configuration of the stance and swing limb segment angles

There were significantly different segment angles found when comparing between the young and elderly. These significant differences included the pelvis, thigh and shank for both the stance and swing limbs. This section will make inferences about the elderly walking patterns by using the young group as the normal/reference gait pattern. The assumption is that normal ageing process has caused deviations from the reference pattern. While the data is limited to kinematic details, more recently there has been a strong view that kinematic details can answer many questions about the walking pattern adopted (Baker, 2006; Baker, McGinley, Schwartz, Beynon, Rozumalski, Graham and Tirosh, 2009). Indeed, kinetic analyses offer extra detail, but the biggest advance in the area of gait analysis has been the developments of simulating muscle function to explain walking patterns (Anderson and Pandy, 2001). The interpretation of the effector and segment kinematics will be based upon combining recent works on muscle simulations with the kinematic patterns observed from the results in Chapter 4.

Figure 4.10.3.1 described the general configurations in the segment differences between the young and elderly. There were significant differences found in the states of the segment configurations at the MTC event. The elderly displayed a significant increase in anterior pelvic tilt. The elderly had significantly higher internal rotation of the thigh at MX1 and significantly less pelvic obliquity at MX1. These findings have been reported by another study investigating the thigh-pelvis kinematics between elderly and young males (Mills et al., 2007). This configuration at MX1 is typical of crouch gait postures. The elderly also had significant differences in the sagittal plane angles of the thigh and shank throughout the swing-cycle, for both the stance limb and swing limb. An important finding was that the swing-limb foot angle was not significantly different between groups at MTC. From the significant increase in forward rotation of the shank (of swing-limb) at MTC, this implies that ankle dorsi-flexion (of swing-limb) of the elderly has increased at MTC, when compared to the young. Pijnappels et al (2001) found that ankle dorsi-flexion increases significantly when there was a threat of tripping, whereas flexion of the hip and knee did not change significantly. A few studies have supported the theory that the foot orientation remains invariant under different walking conditions and is therefore suspected to be a control parameter of walking (Hurmuzlu, Basdogan and Carollo, 1994; Redfern and Schumann, 1994; Grasso, Ivanenko, Zago, Molinari, Scivoletto, Castellano, Macellari and Lacquaniti, 2004; Osaki et al., 2007).

8.2.3.1 The effect of crouch gait on walking mechanics

There is an apparent mechanical discrepancy when contrasting the results of the stance effector vector and the stance limb segment angles of the shank, thigh and pelvis. The larger vertical displacement of the stance effector in the elderly at MTC contrasts with the finding of the average segment angle configurations in the stance effector. At MX1 there was also a significantly longer stance effector and significantly different segment angles of the stance-limb, however this result can be explained by association with the significantly larger pelvic obliquity demonstrated by the young group at MX1. Winter (1992) found from a sensitivity analysis of the stance effector joint angles that the vertical end-point position of the combined effector was influenced by hip abduction, which was demonstrated to be nearly four times more effective at raising the end-point than stance knee flexion (Winter, 1992). Therefore, reduced pelvic obliquity might be considered the major reason why the elderly are still able to manage a significantly ‘longer’ stance effector length (i.e. higher swing hip position) even though they demonstrated a significant crouch gait posture. In contrast, at MTC the pelvic obliquity (frontal plane) was not significantly different between groups (when the limb data is pooled, p=.3), whereas thigh and shank alignment remained significantly different. Figure 4.6.2.1 demonstrates that for the non-dominant limb, the stance effector was significantly longer in the elderly at MTC, but not the dominant limb. Likewise, Figure 4.10.1.2 demonstrated that pelvic obliquity was significantly larger in the non-dominant for the young group at MTC, and there was no difference in the dominant limb. In summary, the only other explanation is the existence of individual differences when configuring the multiple degrees-of-freedom that contributes differently between individuals to contribute to a longer stance effector. For example, various combinations of reduced pelvic obliquity and reduced lateral tilt of the shank and thigh, as well as various vertically aligned sagittal plane configurations of the thigh and shank can all effect a longer stance effector. Various combinations of these may not be reflected in the group mean because of redundancies when they collectively effect the stance effector displacement.

The more vertically aligned thigh segment of the elderly during the swing phase might be directly caused by the significant anterior tilt of the pelvis. For the swing effector of the elderly to adopt the same average thigh configuration as the young group, the elderly would need to perform greater hip flexion work because of the anterior orientation of their pelvis. There are reports that the elderly have less 'pre-swing' energy supplied by the gastrocnemius to progress the swing limb forwards and they require increased reliance of the hip flexors (e.g. McGibbon and Krebs, 2001; Goldberg and Neptune, 2007). The significant increase in anterior pelvic tilt in the elderly is debated to be a postural symptom of ageing rather than an autonomous modification associated with the dynamic task of walking (Kerrigan, Lee, Collins, Riley and Lipsitz, 2001; Kerrigan, Riley, Lelas and Croce, 2001; Kerrigan, Xenopoulos-Oddsson, Sullivan, Lelas and Riley, 2003; Lee, Zavarei, Evans, Lelas, Riley and Kerrigan, 2005; Watt, Jackson, Franz, Dicharry, Evans and Kerrigan, 2011). One study proposed that reduced hip extension and increased anterior pelvic tilt are related to walking speed, such that the anterior pelvic tilt increases with walking speed, and not found to be due to age-related postural changes (Lee et al., 2005). However, Lee et al. (2005) did not measure trunk angle and therefore they could not rule out the contribution of increased forward trunk lean as a response to increasing speed, as the research also showed the young group demonstrated an increase in anterior tilt by a similar margin with fast walking. Therefore, it is not conclusive that anterior pelvic tilt is a dynamic strategy, but it is most likely due to structural changes in muscle-tendon stiffness (Kerrigan et al., 2003; Watt et al., 2011).

The stance effector velocity profiles appear to support the findings in the literature that result from a crouch gait posture. This study observed that the elderly continued to have upward vertical velocity at MTC compared to the young group who displayed near-zero velocity at MTC (Figure 4.7.2.1). In an experiment that asked unimpaired participants to walk under two conditions of trunk flexion, the resulting stance limb pattern was similar to a crouch gait posture (Saha, Gard and Fatone, 2008). The crouch gait pattern demonstrated in the study by Saha et al. (2008) showed a delay in peak vertical position of the centre of mass, occurring following mid-stance. This compares with the delayed peak in the velocity profile of the elderly group stance effector in Figure 4.7.2.1. In the only crouch gait simulation study exploring muscle contributions to the centre of mass, the hip extensors are required to contribute much more than in normal walking to propel the body forwards during early stance (Steele, Seth, Hicks, Schwartz and Delp, 2010). The soleus and gastrocnemius muscles need to contribute more to upward posture support compared to normal gait, and the passive contribution of the skeleton provides less upward support in crouch gait but it gives more acceleration to the body centre of mass in the forwards direction (Steele et al., 2010). This infers high forwards velocity of the stance effector in early stance. In this thesis, the elderly stance effector did show a higher mean forward velocity, however this was not significantly higher than the young group. From the simulation study by Steele et al. (2010), a crouch gait posture favours forwards progression by relying less upon the ankle plantar-flexors and more upon the skeletal structure and hip extensors.

The finding of an increased swing limb length in the elderly can be explained by a crouch gait posture. Referring again to the crouch gait simulation study of Steele et al. (2010), there was a finding of increased knee extensor moments during pre-swing. Empirical studies propose that crouch gait has a passive effect on stiff-knee walking (van der Krogt, Bregman, Wisse, Doorenbosch, Harlaar and Collins, 2010). Increased knee extensor activity during pre-swing is reported to be associated with reduced knee flexion velocity at toe off, and subsequent reduction in knee flexion during mid swing (Goldberg et al., 2003).

Due to the skeletal posture and the altered moment arms of the articulating muscles, a crouch gait posture limits the energy return contribution from passive musculo-skeletal structures (e.g. length-tension, length-velocity, muscle tendon unit) for maintaining centre of mass control during mid-swing (McGeer, 1993; Collins et al., 2005; Kuo, 2007). This in-turn will demand a more energetically active contribution from the muscles, and this will be costly for the elderly during stance (Steele et al., 2010). So, for the stance limb in the sagittal plane alone, there looks to be a potential requirement to increase neural commands to the muscles to provide sagittal plane support and forwards propulsion. The old theory of passive postural control during inverted pendulum gait is confronted by recent suggestions that postural control is an active optimisation process (Guigon, 2010).The frontal plane configuration of the elderly is in a comparably less-ideal position during stance because of the vertical pelvic alignment. The reduced range of motion in pelvic obliquity can potentially indicate two related causes: 1) reduced hip/pelvis strength; or 2) reduced adaptive trunk control. From the above discussion, the limb configurations of the elderly are showing a loss in their potential to create passive walking dynamics.

8.2.3.2 Effects on toe sensitivity due to swing effector configuration

The toe position at any instant of the swing-cycle is sensitive to the configuration of the swing and stance effector segments (Winter, 1992; Moosabhoy and Gard, 2006). The effect of the segment angular change on the toe position was investigated to appreciate the biomechanical state of the swing effector upon toe position from a sub-sample of the young (N_Y=14) and elderly (N_E=14) participants. A partial derivative application was based upon the same method outlined by Moosabhoy and Gard (2006). Stated more simply, when one joint angle was adjusted in the segment chain of the swing effector, the effect on the toe position was observed. At MTC, for one radian change in joint flexion at the ankle, knee or hip, the effect on the displacement of the vertical toe state in the sagittal plane is described in the following. The data found that toe sensitivity (m/rad) for the ankle and knee joint flexions were: ankle flexion Y_{Ankle} = 0.133 m/rad, E_{Ankle} = 0.139 m/rad (p>.1); knee flexion Y_{Knee} = 0.048 m/rad, E_{Knee} = 0.057 m/rad (p>.1). For toe sensitivity from one radian of hip flexion there was a significant difference, Y_{Hip} = 0.112 m/rad, E_{Hip} = 0.078 m/rad, p<.05. Shemmel et al. (2007) have reported that the joint angles do not act independently, but rather they co-vary as a functional unit. Therefore, when summing the sensitivities associated with each joint separately, the young group demonstrated a higher sensitivity of 29.3cm, compared to the elderly, 27.4cm, and this was significant at p<.1 (t(26)=1.71, p=.098). This ‘effector sensitivity’ suggests that the biomechanical state of the young group’s swing effector is more readily able to effect an increase (or decrease) in toe position at MTC. However, the elderly group have a swing effector that is more stable to variations occurring in the joints.

8.2.4 Summarising the biomechanical effect of the effector states

The biomechanical conditions of the stance and swing effector systems are relevant background to examining the control and coordination of the combined effector. The biomechanical contributions have only been investigated in a limited way because the main focus of this thesis was to explore changes to control and coordination in the effector systems of the elderly. The above discussion provides a link between biomechanics and control, and this will be evident in the remaining discussion on control and coordination of the toe clearance task.

8.3 Variability in the stance, swing and combined effector systems is task-relevant

In following from the finding that the elderly are more variable for step-length and step-width data, older adults are found to be significantly more variable for the three effectors in all three dimensions (section 4.8.1 - 4.8.3).

Gait variability has a strong association with falls in older adults (Hausdorff et al., 2001; Lord et al., 2007). The significant increase in variance found in the elderly group can be due to different contributing sources within the loco-sensorimotor control system. Theoretically, increased variability can represent a proliferation of inherent noise through the system (Christou and Tracy, 2006; Faisal et al., 2008). Increased variability can represent increased uncertainty when estimating the state of gait trajectories (e.g. Kording and Wolpert, 2004). It can represent the type of motor commands selected to perform the movement tasks by invoking significantly more signal dependent noise (e.g. Harris and Wolpert, 1998). A recent study proposes that gait variability is related to strength and range of motion rather than effect of alterations in walking speed (Kang and Dingwell, 2008b). Increased variability can represent a policy to tolerate movement error (e.g. Nagengast et al., 2010), reflect a poorly designed coordination policy (Scholz and Schoner, 1999), or represent a biomechanical disposition to increased sensitivity of angular variations within the effector segments (e.g. Moosabhoy and Gard, 2006). This study did not design a method to discriminate why the elderly display larger variance of gait trajectories. However, some of these issues will be discussed later in the Chapter. For now, increased variability can be a reflection of any combination of the above issues.

The significant increase in variability was most commonly associated with the stance effector, where the elderly displayed significantly higher variability at MX1 and MTC for both the dominant and non-dominant limbs, in all dimensions (Figure 4.8.2.1-3). Variations of the stance effector have a likely association with posture control during walking, and the increased variability of the stance effector is a likely demonstration that posture control is a more difficult task for the elderly compared to the young.

There was one instance where the elderly were actually displaying less variability, and this was for the dominant limb swing effector in the medio-lateral dimension. This contrasts with the combined effector in the same dimension, which the elderly displayed greater variability about the MX1 region. Of interest was the finding that the swing effector did not display significantly higher variability in the vertical dimension at MX1 or MTC, although the non-dominant limb was significantly higher prior to MTC (near MTC-15%; Figure 4.8.3.1).

The major interest of this thesis is the state of the effectors in the vertical dimension. The profiles of variance displayed in Figures 4.8.1.1, 4.8.2.1 and 4.8.3.1 display the variance along time slices of the swing-cycle. Both the stance and swing effectors display a minimum of variance near MTC. For the combined effector (Figure 4.8.1.1), there was actually an increase in variance found from MTC-15% to MTC, where the minimum variance appears to be near MTC-15%. This profile looks consistent between groups and between limbs.

Section 4.8.4 described how the groups change the state of effector variability in the vertical dimension at state-relevant times of the swing phase. Variance (in the vertical dimension) of the stance and swing effectors decreased significantly between MX1 and MTC, for both groups. However, for the combined effector, only the young group demonstrated a significant reduction in variability between MX1 and MTC (Figure 4.8.4.1), and there was not a significant difference in the variability of the combined effector between groups at MX1. There was a significant interaction effect that indicates the young group significantly reduce the variability of the combined effector compared to the elderly. For the elderly, the no change in variability in the combined effector between MX1 and MTC is an interesting finding, particularly because they do exert a significant variability reduction between MX1 to MTC in the stance and swing effectors.

The significant 'no-change' in variability of the combined effector between MX1 and MTC for the elderly could represent several different contributing sources as outlined above, however, it is appealing to consider the effect of a cost policy. The issue of a control policy to penalise control effort was raised in section 2.3.4. Does the finding of 'no-change' in variance between MX1 and MTC demonstrate that the elderly apply less control, and tolerate errors in the combined effector, so that they reduce the cost of control effort? If this is occurring, then the reward for this policy will be relevant to improving certainty in the movement control costs they accrue while walking (section 2.3.4). Alternatively, the finding here may occur in spite of a policy that predominantly aims to minimise error costs. If this is occurring, then the 'no-change' in variance can reflect many potential sources, such as poor coordination between the stance and swing effectors, or a saturation in accuracy for estimating the state of the toe as it nears the MTC event (section 2.3.2.1).

For both the young and elderly, there was significantly higher variance (vertical direction) of the swing effector compared to the combined effector, and when the swing effector was compared to the stance effector, at both MX1 and MTC. The stance effector variance (vertical direction) was significantly lower than the combined effector at MTC, but not at MX1, for both groups. For the elderly group, the stance effector was significantly higher than the combined effector at MX1. At MTC, the vertical state of the swing hip position was less variable than the vertical state of the toe. These results suggest that the task of controlling the vertical length of the swing limb is either more demanding, or less relevant, when compared to the task of controlling the vertical position of the swing hip. The control policy of the stance effector will be relevant to minimising upright posture costs (and potentially ground clearance risk), and the sensorimotor system is certainly well tuned to control this task (Quevedo, Stecina, Gosgnach and McCrea, 2005; Quevedo, Stecina and McCrea, 2005; Robert, Zatsiorsky and Latash, 2008). In comparison, the control policy associated with the swing effector length will have a cost function that includes a trade-off between minimising ground clearance risk and energy (Anderson et al., 2004; Begg et al., 2007). The significant difference in variance between the two effectors suggests that control of the swing effector is a more challenging task. For the swing effector at MX1 and MTC, there was no significant difference in variance between the groups.

It was reported earlier that the elderly demonstrated a significantly 'flatter' vertical trajectory of the combined effector between MX1 and MTC (section 4.6.4), however, variability of the combined effector did not significantly reduce in the elderly group. In contrast, the young group demonstrated a significant reduction in variability when comparing between MX1 and MTC.

8.4 Skewness in the stance, swing and combined effector systems is task-relevant

This thesis offers new information to the gait literature in the form of the skewness description of effector states at task-relevant time slices of the swing cycle (section 4.9). The hypothesis that gait trajectories are randomly distributed about a central tendency can be rejected based upon the skewness result being significantly different to zero for both young and elderly. This was observed from the group 95% confidence intervals for the combined effector in all dimensions, the stance effector in the vertical and anterior-posterior dimension, and the swing effector in all dimensions. Skewness is an interesting parameter because it represents asymmetric control, which could be due to a combination of two control factors. First, control penalties assigned to one side of the distribution are different when compared to the other part. Second, the biomechanical configuration at one side of the distribution has greater 'passive' control when compared to the other. Skewness is a parameter that has been overlooked in the gait literature, apart from some exceptions (Begg et al., 2007; Mills et al., 2007; Best and Begg, 2008). Begg et al. (2007) identified skewness in the distribution of toe states at MTC, and Best et al. (2008) demonstrated that skewness is required to provide an accurate probability model of the toe state at MTC. Without describing skewness details, Mills et al. (2007) reported that tests of normality were rejected for more than two thirds of the 900 gait kinematic distributions they investigated, which included distributions of toe states at MTC. Apart from these studies, there has not been any known use of the skewness parameter in the gait literature to describe gait kinematics.

The results from this thesis demonstrated that skewness in gait parameters is sensitive to ageing. First, there was the significant finding of increased negative skewness in the elderly step-length. There are other significant differences between the groups on measures of skewness relevant to the effector trajectories. In terms of contrasting skewness with variability, an increased gait variability represents the inconsistency of gait patterns, whereas increased skewness represents an inconsistency of control being applied to gait. The results of significant skewness support the theory that gait variability is not controlled equally, everywhere, but mostly where movements are relevant to task performance. Interpreting skewness will be made in the context of control policies because skewness lends itself to theories of asymmetric penalties of movement errors. Skewness is limited by its inability to determine the true nature of a control policy, but it can reflect some general insight into how control is applied.

The elderly demonstrated a significantly different control policy compared to the young group for the combined effector. The elderly loco-sensorimotor controller placed a significantly greater penalty on vertical toe states that get too low at MX1, while rewarding higher states. At MTC in the non-dominant limb, the elderly showed the same control policy (e.g. Figure 4.9.1.1, section 4.9). This can be interpreted as the elderly stance effector significantly changed its cost policy between MX1 to MTC, by assigning a greater penalty for swing hip positions that got too low, while rewarding higher states. In contrast, the young group penalised swing hip states that got too high. This can be observed in Figure 4.9.2.2 (from section 4.9.2), as there is significant negative skewness in the young group during the approach to MTC, for example negative skewness is significant at MTC-1% ($t(106)=2.33$, $p<.05$) but not quite significant at MTC ($t(106)=1.94$, $p=.056$). The controller of both groups penalised vertical states of the toe that got too low with respect to the hip (swing effector), and with respect to the ground (combined effector). For both groups, this policy has been expressed to a significantly greater extent in the combined effector when compared to the swing effector, and is likely to be indicating that the combined effector is the performance-relevant variable when compared to the swing effector. The interesting finding (see Figure 4.9.4.1, section 4.9) is that the significant change in the combined effector control policy expressed by the young group, occurs in spite of the stance effector policy to penalise high swing hip positions.

The elderly applied a cost policy that penalised low vertical states of the swing effector, stance effector and ultimately part of the policy to control the combined effector. In contrast, the young group showed an ability to apply the same policy to the combined effector at MTC, without applying a control policy that penalises low swing hip states. A likely explanation is that the young group was able to do this because of an effective inter-effector cooperation between the stance and swing effector systems.

The benefit of not penalising low swing hip states allows the pelvis to contribute to posture control. In contrast, penalising low swing hip positions will require positive muscle work from the stance limb extensors at the ankle, knee and hip, as well as the hip abductors. The elderly significantly demonstrated a crouch gait posture, even though they have a policy of penalising low swing hip states. This control policy is a likely explanation that there existed a significant crouch gait posture in the elderly group, while the stance effector is significantly 'longer'. Based upon the control policy differences between groups, it is likely that the elderly will be required to perform far more muscle work, compared to the young group, during the task of controlling toe clearance. The action of this muscle work for raising the swing hip position will likely result in larger postural instability (Pandy et al., 2010).

Skewness found in the anterior-posterior direction displayed discrete differences between MX1 and MTC regions when observing the three effectors (e.g. Figures 4.9.1.2, 4.9.2.2, and 4.9.3.2 in section 4.9). For example in Figure 4.9.1.2 (see section 4.9), with reference to the non-dominant limb, the elderly combined effector demonstrated positive skewness in the MX1 region (significantly greater than zero), whereas negative skewness in the MTC region (significantly less than zero). With reference to MX1 in Figure 4.9.1.2 (see section 4.9), the loco-sensorimotor controller of the elderly penalised toe states that were posteriorly displaced from the stance foot, relative to anterior toe states. This is significantly different to the young group, who showed no significant skewness at MX1. At this region, toe states of the elderly were penalised if they were low, and if lagging in the direction of walking progression. At MTC, the opposite situation arises. The loco-sensorimotor controller of the young group significantly penalised toe states if they were too far forward of the stance foot, whereas the elderly group were significantly different in that they did not apply such a strong penalty. The significantly greater posterior orientation of the combined effector in the elderly at MTC (Figure 4.6.1.3) suggests that there might be less need for penalising forward states.

The penalties applied to the combined effector, can be an outcome of the penalty applied to attain a postural state of the body, and subsequently determine swing hip position. There was a different policy for controlling the anterior-posterior state of the stance effector in the elderly (Figure 4.9.2.3), i.e. location of swing hip position relative to the support foot. The elderly stance effector was orientated significantly more posteriorly at MTC compared to the young group in the non-dominant limb (Figure 4.6.2.2). The elderly penalised posterior states of the swing hip position, which was significantly opposite to the young group because they penalised anterior states of the swing hip position. Therefore, while the stance effector of the elderly was posteriorly oriented, they penalised this orientation. Compared to the elderly, the negative skewness in the young group swing effector (Figure 4.9.3.3) also showed a significantly stronger penalty to toe positions that got too far forward of the swing hip.

In summary, there is a policy of control applied by the young group that is consistent between the stance and swing effectors, and ultimately the combined effector. This policy suggests that the controller does not allow the swing hip position, or the toe position, to move too far forward of the support foot. In contrast, the elderly controller is predominantly concerned about penalising posterior states of the swing hip position, even though it displays a significantly greater posterior orientation compared to the young group.

8.5 Persistence in the stance, swing, and combined effector systems is task-relevant

The interpretation of long-term correlations in gait variables has been challenging for researchers. There has been many different terms used between research groups that aim to capture the theorised phenomena that is occurring in the dynamic system studied. Frameworks for understanding persistence have generally been associated with two complimentary theoretical perspectives: the ‘mechanistic’ and the ‘nomothetic’ (Torre and Wagenmakers, 2009). Section 2.4.3 identified the mechanistic approaches of mathematical models designed to simulate different forms of limit-cycle stability of biological processes from which persistence emerges (Ashkenazy et al., 2002; Gates et al., 2007; Dingwell et al., 2010). Section 2.3.1.1 contrasted ‘nomothetic’ theories of hierarchical and self-organised control and the theoretical construct of dynamical systems (Kelso, 1995; Temprado, 2004; Heylighen, 2010). Section 2.3.2 reviewed the research demonstrating mechanistic models are successfully capturing how humans move with precise control for discrete movement tasks. The advancement of research simulating the control of continuous dynamic walking has been able to take advantage of these issues and develop new mechanistic models that integrate hierarchical ‘control’ with self-organised ‘passive’ dynamics (Manchester et al., 2011; Tassa et al., 2011). These models simulate the loco-sensorimotor systems workspace and demonstrate that optimal gait solutions emerge when a self-organised ‘passive’ dynamic process search for optimal movement solutions is regulated by a hierarchical ‘control’ dynamic (Manchester et al., 2011; Tassa et al., 2011). The optimisation problem demonstrates that the frequency of regulating the ‘passive’ search process by the ‘control’ dynamic comes at a cost (Todorov, 2009; Todorov, 2011). The interpretation of the persistence-likelihood parameter is based upon the frequency of regulating the ‘passive’ search by the ‘control’ dynamic. Under this view, persistence will represent less ‘control intervention’, or less ‘control effort’. In the discussion below, the term persistence-likelihood will be used to state results and the term control effort will be used when interpreting the meaning of the results.

There has not been any known studies that have profiled states of gait trajectories using the DFA method at discrete points along a task-relevant time-course of the swing-cycle. There has not been any known studies that have claimed to profile the control effort applied to the swing cycle of walking. Therefore, support for the interpretation will come from indirect studies.

8.5.1 'Effort' to control the vertical toe state is gradually reduced leading up to MTC

Within the groups, persistence-likelihood increased significantly between MX1 and MTC for all three effectors in the vertical dimension. This was found for the non-dominant limb. For the dominant limb, only the stance effector persistence increased significantly within the young group, and only the combined effector persistence increased significantly within the elderly group. Therefore, there was a general reduction in control effort as the swing-cycle approached the MTC event. This suggests a cost function that assigns a progressive increase in the penalty awarded to the effort to control the effector states. This also indicates that there is a potential benefit to task performance by reducing the control effort applied to the effector states.

There are two possibilities underlying control effort: 1) the loco-sensorimotor workspace is high dimensional with a luxury of solutions and therefore the controller can afford to penalise control effort; or 2) the loco-sensorimotor workspace is low-dimensional and candidate solutions are less abundant, but the control policy penalises control effort. This can be explained in the following. In a low-dimensional workspace, the continued search for a candidate solution will be more exhaustive, in comparison to a higher dimensional workspace where redundancy exists (Todorov, 2009; Tassa et al., 2011). In either system, the search effort applied can be identical, however, a low-dimensional workspace will need to consider the trade-off for tolerating 'second-best' candidate solutions. In a workspace with reduced redundancy, a controller with a cost on control effort will stop the exhaustive search and will need to settle for the next-best solution. Therefore, persistence-likelihood represents the effort to find a candidate solution, irrespective of whether the candidate solution results in a desirable state. Therefore, for a low-dimensional system, the reduction in control effort suggests something about the compromise in the cost function that reduces the penalty when arriving at sub-optimal candidate solutions. For a highly redundant system, the reduction in control effort suggests a luxury of optimal movement options. Therefore, for systems with different dimensionalities, the same outcome in persistence can represent markedly different underlying processes.

Alternatively, reduced persistence of a performance variable suggests that the controller has reduced the penalty cost to control effort. This might be due to two reasons. First, where a cost function assigns a relatively large penalty to small movement errors, and therefore the controller expects an exhaustive search for an optimal state. Second, if the cost function is designed for a challenging task goal, and the redundancy in the workspace is low dimensional, the control effort will not be penalised because it will need to be exhaustive for satisfying a cost function that requires meeting the task goal. Again, the same outcome in persistence can represent different underlying processes.

The suggestion that there is reduced control effort leading up to MTC contradicts the findings of Nagengast et al. (2010) who observed a gradual increase in control effort as the goal state approached. However, the experiment by Nagengast et al. (2010) was different to walking because the participants visually guided a 'virtual ball' with their upper limbs and this differs from the context of the toe clearance task in the type of sensory feedback and the accurate knowledge of results. In walking, the knowledge of results for the toe clearance task is uncertain and needs to be optimally inferred through proprioceptive feedback.

The other contradiction of control effort is related to the result that there is significant increases in skewness between MX1 and MTC. This result indicates that the controller penalises low states of the swing and combined effector leading up to MTC. It is difficult to reconcile that a controller applies a large penalty to movement outcomes, but also applies a large penalty for control effort (i.e. effort to control the movement solution). This raises an interesting issue related to the notion of a low-level controller (review, section 2.3.1.1.1. and 2.2.3.2.4), that is involved with a 'self-organised' passive dynamic search for movement solutions that have been trained to penalise low states.

There is some evidence from the results that does support the suggestion that control effort is reduced prior to MTC. This is the finding that the variance of the combined effector slightly increases as MTC is approached from MTC-15% (Figure 4.8.1.1). There can be two possibilities for the observed reduction in control effort found in the effectors leading up to MTC. First, the limb biomechanics are configured optimally for the task. Second, there is more state-estimate accuracy of the effector position by minimising outgoing motor commands or applying consistent motor commands from trial-to-trial (review, section 2.3.2.1). These two issues will be discussed below.

There may only be negligible errors in the task goal for a significant increase in control effort leading up to MTC. This relates to findings of the synergy ratios in the stance and swing effectors (results, Chapter 6), and the results of toe sensitivity (results, section 8.2.2.3). For example, the mechanical configuration of the swing effector lengthens near MTC and the changes in segment angles have less effect on the toe position. For the elderly, the total sensitivity of the swing effector was almost significantly ($p=.098$) reduced, compared to the young (see section 8.2.3.2). This configuration can place an average 'ceiling' on how bad errors can get when segments are biomechanically configured in this way. The controller would therefore consider the extra error cost negligible and apply greater penalty to control effort. More convincing support for the theory of 'negligible cost' by reducing control effort relates to the finding that the swing effector has a strong synergy for the vertical task of the swing effector (section 6.6). The results from Chapter 6 indicates that the controller has selected motor commands that seek to stabilise the vertical toe states prior to MTC, indicated by a synergy structure that was significantly reduced in task-relevant states between MTC-12% and MTC.

Reducing control effort is likely to reduce motor commands and the associated signal dependent noise effects. Optimising the performance of goal directed movements benefits from the reduction of signal dependent noise (Harris and Wolpert, 1998; Hamilton and Wolpert, 2002; Hamilton et al., 2004). In upper-limb obstacle avoidance tasks, performance is improved by increasing limb stiffness near the obstacle. While this might appear to actually contradict the signal dependent noise cost, the control system effectively distributes a large proportion of this noise in a goal-irrelevant dimension. Stated more explicitly, the increase in motor commands recruited to increase joint stiffness is associated with an increase in noise-effected movement errors, however, the motor commands selected are noisy in a task irrelevant direction, and ensuring that noise in the direction of the expected perturbation is kept to a minimum (Franklin, Liaw, Milner, Osu, Burdet and Kawato, 2007; Selen et al., 2009). This structure of signal-dependent-noise when co-contracting muscles demonstrates how the cost of movement variability can be weighted with the benefit to reduce task uncertainty through mechanical stability. This is considered to be a quick and flexible response available to the CNS (Franklin et al., 2008), and can therefore relate to the proposed idea of a low-level controller. The research studies of upper-limb avoidance tasks have also demonstrated increased skewness of limb endpoint states when negotiating around an obstacle, such that variance of the states were minimized in the proximity of the obstacle (e.g. Hamilton and Wolpert, 2002). This type of study has not been applied to walking, but the outcomes show consistency with the results of this thesis that relate to reducing signal dependent noise and increased skewness. This thesis indicates that the structuring of low-level controller driven motor commands creates stiffness in the vertical direction of the swing limb, and while there is an associated increase in signal dependent noise the type of motor commands selected distribute the variability into the anterior-posterior direction of the swing effector. This theory supports the finding in this thesis, whereby the effort to control the vertical toe trajectory is gradually reduced as the task goal approaches, and correspondingly the vertical skewness increases, while the anterior-posterior variability increases.

While tasks are sensitive to motor commands that create movement error, Nagengast et al. (2010) demonstrated that 'controllers' are also sensitive to task uncertainty. Therefore the controller's policy will weigh the costs of motor command signals on movement error, and also the certainty that the control policy will provide consistent costs (review, section 2.3.4). By reducing control effort, the 'self-organised' movement solution will provide outgoing motor commands that are less corrupted by signal dependent noise (Harris and Wolpert, 1998), and also be repetitively familiar for more accurate forward model (efferent copies), such that the combination of both will improve the certainty of state-estimate, and certainty of task performance. This form of acquiring greater certainty of the effector state-estimates is associated with the integrated probability distributions of ‘prior’ and ‘likelihood’ (section 2.3.4).

8.5.2 ‘Effort’ to control the vertical hip position is increased in elderly leading up to MTC

The significant differences between the groups on persistence-likelihood of the combined effector states at MTC and the stance effector states prior to MTC indicates that the elderly control the task of toe clearance significantly different to the young (section 5.5.1). The elderly demonstrate an increase in persistence of the combined effector and a reduction in persistence of the stance effector as the swing phase approaches MX1 and this trend continues through to the MTC region (Figure 5.8.1.1 and 5.8.2.1). This interaction between the combined effector and stance effector are new observations in gait literature. The full implications of this finding are discussed in the context of toe clearance control in section 8.9.

Keeping in mind the general assumption linking ‘persistence’ with control effort – and the neural stress theory (Scafetta et al., 2009) and the theory of redundancy loss (Dingwell et al., 2010) – the reduced persistence-likelihood in the stance effector is maintained for longer which can potentially cause a problem for co-adaptation between the stance and swing effectors when controlling the combined effector. An ageing effect can potentially be reflected by the theory proposed by Dingwell et al. (2010), indicating a likely reduction in redundancy. This raises the question of how the system can then partition equally between the component effectors the responsibility for correcting toe clearance errors (White and Diedrichsen, 2010).

8.5.3 ‘Effort’ to control the medio-lateral effector states is greater at MX1

The nature of short-term anti-corrections (anti-persistence) was also confirmed by lag 1 auto-correlation function (ACF_{lag1}), and from a close agreement between ACF_{lag1} and the DFA method. The presence of a correlation at lag-1 does not suggest that long-range correlations are a manifestation of short-term processes. A time series with long-term dependence will often display a strong correlation at a short-time scale and then very slowly decay, but the sum of the correlation values doesn't settle to a finite value. The ACF was not reported for time lags greater than 1 to determine the rate of decay found in our correlated data. A slow rate of decay for time lags approaching 20-50 could help to confirm the existence of long-term dependence. What we have been able to confirm is that the short-term auto-correlation applied indicates a significant level of anti-correlation, or over-correction, being applied to the stance and swing effectors in the medio-lateral direction and this remains relatively invariant throughout the swing phase. The 'persistence' revealed this as a random-like correlation ($\alpha = 0.55$). The only other relatively invariant profile of the auto-correlation was the stance effector in the anterior-posterior direction. However, the medio-lateral direction of the stance effector had significantly lower level of persistence compared to the anterior-posterior and vertical directions. In following on from the earlier section discussing the relative 'passive' stability associated with anterior-posterior and lateral step parameters, it is expected that lateral motion would be associated with increased anti-persistent behaviour. Earlier, we reported that step width did not show a difference in the amount of anti-persistent behaviour between the groups. Lateral control of leg swing has been suggested as a strategy to control lateral instabilities during walking (Kuo and Donelan, 2010). The ACF_{lag1} (Figure 5.5.3.1.1) and the 'persistence' results show that elderly participants are associated with significantly more anti-persistent behaviour during the MX1 region, for all three effectors. For both groups, the combined effector displays the greatest level of anti-persistence occurring at MX1. The significant group interaction shows that the elderly increase control effort as the swing leg passes through MX1 while the young group reduces control effort. This suggests that the elderly have less penalty assigned to search for effector movement solutions that satisfy medio-lateral states. While this has been found with the effector trajectories, there was no significant difference in persistence for step-width. This might appear contradictory, however, there is an advantage of tracking the trajectory through the swing-cycle time course for observing where the young and elderly become significantly different.

The young maintain a state of medio-lateral control of the swing effector passing through MTC whereas the elderly reduce the effort to control the medio-lateral direction of the swing limb. The young demonstrate that the effort to regulate the swing limb orientation in a medio-lateral direction is important for the toe clearance task goal and they prefer a more lateral orientation of the swing limb at MTC. In contrast it seems that the elderly are controlling for a more vertically aligned swing limb at MX1. The medio-lateral orientation of the combined effector or stance effector is not positioned differently between the groups. The control of the spatial orientation could also be a response to control the velocity of the effectors in this direction.

8.6 Reconciling the parameters of control: exploring a relationship between persistence, skewness and variability

This section discusses the results stated in section 5.9. Unlike mean and standard deviation, skewness is rarely used as a way of measuring the performance of a gait variable. There are a few good reasons for this, one of which is because not all gait data is skewed, and second, skewness is largely unknown in terms of how it relates to task performance. However, skewness is relevant to MTC data and, just like $1/f$-type noise is important to improve understanding of biological systems, so too is skewness important for understanding MTC control. It has been suggested that skewness is a likely outcome from a control strategy to optimise toe clearance performance (Begg et al., 2007). Begg et al. (2007) indicated that skewness could be a control policy to regulate MTC errors at the critical 'ground contact' side of a distribution while ignoring the 'variance' associated with the other 'energy efficiency' side of the distribution. Skewness has also been noted as important in motor tasks involving upper limb endpoint tracking around an obstacle (Hamilton and Wolpert, 2002). While there has been an attempt by Hamilton and Wolpert (2002) to explain the skewness with upper limb trajectories, in one MTC study the source of skewness has been associated with other statistics of MTC distribution (Begg et al., 2007), and there have been other studies measuring MTC but not reporting 'skewness' (Mills et al., 2007; Khandoker et al., 2008). Skewness could potentially be a key measure of MTC performance and it has been somewhat ignored in MTC studies probably because of its unknown relationship to loco-sensorimotor control. It is therefore important to further investigate its nature and explore where it can originate. One general hypothesis is that 'skewness' comes from a combination of either: 1) the outcome of a 'hard-wired' low-level controller; or, 2) it is a passively derived source from biomechanical origins (e.g., Moosabhoy and Gard, 2006; Gates et al., 2007). This section will investigate the correlated associations between three control variables of 'variance', 'skewness', and 'persistence-likelihood' (section 5.9).

For all participants, it is found that if the 'variance' exists in the anterior-posterior direction of the effectors, then it generally propagates to the medio-lateral and vertical dimension of the effectors. ‘Variance’ in the vertical dimension of the combined effector is related to ‘variance’ in all other dimensions of the effector system. This is the case for both groups, but with the notable exception for the young group’s stance effector in the vertical dimension. This is notable because in the young group a variable stance effector in the vertical direction is not associated with variations of the combined effector in the same direction (although somewhat weak r_E=.36, it contrasts with the young, r_Y<.20).

The anterior-posterior variability could be a good reflection of the natural variability in the loco-sensorimotor system because it appears to be permeating in an unregulated way irrespective of age and it correlates very highly with other effectors in the anterior-posterior direction. The very strong correlations existing between the anterior-posterior components of ‘variance’ among the effectors – particularly the combined effector – indirectly suggests that all participants loosely regulate the propagation of anterior-posterior ‘variance’. Further support comes from the ‘persistence’, which also indicates that where the stance effector persists in the anterior-posterior direction, persistence also exists in the anterior-posterior dimension of the swing and combined effectors. Importantly, the anterior-posterior ‘variance’ is found to also correlate strongly with variance in the vertical component of the combined effector. The correlations are not as strong in the young group compared to elderly for anterior-posterior variance in the stance, swing and combined effectors correlating with the vertical variance in the combined effector: r_E=.60,.58, .57; r_Y= .51, .30, .34. Further support comes from ‘persistence’, which shows that ‘persistence’ in the anterior-posterior component of the elderly effectors correlates with the ‘persistence’ in the vertical components. This isn’t found in the young group and it generally indicates that the young are better adept at regulating the effect of general system variance influencing the combined effector.

An intriguing finding is when contrasting differences between the groups in the way that anterior-posterior ‘variance’ in the effectors correlates with ‘skewness’. The young group show that in the anterior-posterior dimension most correlations between ‘skewness’ and standard deviation are negatively correlated. Correlations indicate that an increased standard deviation is related to increased negative ‘skewness’ at MTC for the young group. Although there are some weak correlations, the trend is telling. Increased 'variance' is related to decreased 'skewness' in this dimension for the young group. In contrast, the elderly don't exhibit any likeliness to this behaviour. Again, while assuming that 'skewness' is a control policy, this is suggestive that the young group regulate 'variance' in the anterior-posterior direction. In contrast, the elderly show that anterior-posterior variability is positively correlated to 'skewness' in the vertical combined effector. Also, the 'persistence' shows a slightly stronger correlation for the same relationships in the elderly, which is not evident for the young group. This is also supporting that when the elderly lack tight regulation of the anterior-posterior 'variance' it results in increased 'skewness' in the vertical component of the combined effector. So, for the elderly, this appears to be an indirect approach to regulating anterior-posterior 'variance'.

Contrasting the correlations between the two groups could underlie a different neural control policy, or some passive influence brought about by certain mechanical configurations adopted at MTC (Moosabhoy and Gard, 2006). In context of non-linear dynamic systems, 'skewness' could be capturing some very basic behavioural feature of trajectory convergence upon a fixed point attractor. The learned neural system might be behaving like a fixed attractor when initial conditions are close to the attractor compared to when the initial conditions are further away. The result might emerge as a skewed distribution. For the elderly, this possible behaviour of being drawn towards a fixed attractor is realised when the anterior-posterior 'variance' is regulated by low-level/passive control origins and the system is allowed to settle into an attractor state. Under this hypothesis, vertical 'skewness' might be manifested from passive control structures of a biomechanical origin and also some '*self-organised*' form of low level neural control. The toe clearance (vertical dimension of the combined effector) 'skewness' behaviour emerges due to unregulated anterior-posterior trajectories. In contrast, the young group can be hypothesised to regulate large forward variations in the orientation of the swing limb from a different control policy. Because the young appear to regulate anterior-posterior 'variance', the hypothesised passive relationship between anterior-posterior 'variance' and vertical 'skewness' might not get a chance to be realised. Skewness in the combined effector of the young is not correlated with effector 'variance' and therefore the existence of 'skewness' found in the combined effector of the young participants might be reflecting a higher-level control policy. A further intriguing finding is that 'skewness' in the elderly combined effector is correlated with 'variance' for all effectors and respective dimensions. The noisier the system in the anterior-posterior, the higher the level of positive 'skewness' can be found in the vertical component of the combined effector. Also, the noisier the vertical swing and stance effectors, the more 'skewness' in the vertical stance and swing effectors. This is intriguing because it is not found at all in the young group.

To provide further evidence to the issue of 'skewness' being derived from high-level, low-level or purely passive sources of control, the results of 'skewness' in the elderly further demonstrate a telling contrast when comparing the stance and swing effectors in the vertical and medio-lateral dimensions. In the elderly there are negative correlations between the two effectors, suggesting a strong between-effector coupling. Correlations show effector compensation occurring in the vertical and medio-lateral dimensions, and this coupling isn't indicated by the young group. This strong relationship of between effector coupling indicates a neural presence of control. Determining the hierarchical level of where the adaptations are being governed cannot be made from the analysis, as it could be due to a higher-level 'third party mediator' presence, or it could be due to a low-level sub-cortical co-adaptation response. When comparing the 'skewness' parameter between-effectors for the young group, there is a strong relationship between the component effectors which indicates the young group is penalising anterior-posterior direction errors of equal proportion in the stance and swing effectors. Additionally, the between-group correlation differences are significant, for the correlations between the stance and swing effectors in the vertical direction (r_Y=.68, r_E=.27, $p<.05$). The above associations between anterior-posterior 'skewness' and vertical 'persistence' found in the young group is not indicating a responsive co-adaptation strategy, but a general control policy being enforced equally between the effectors. This is not apparent in the elderly. Further evidence that the elderly are employing a responsive adaptation policy of control is demonstrated by the strong 'skewness' correlation (r=-.86) between the stance and swing effector in the vertical direction. This contrasts to the young who do not have a significant correlation ($p<.05$)

for the same relationship. A third control policy based upon between-effector co-adaptive response is found in the elderly for the 'skewness' correlation in the medio-lateral direction (r=-.50). This indicates that between-effector compensation is occurring to address errors. Therefore, a responsive co-adaptation is occurring between the effectors and this is a form of control policy that is somewhat exclusive to the elderly group. The responsive co-adaptations in the medio-lateral and vertical dimensions would be important for the two gait tasks of lateral stability and toe clearance. This provides some rationale for considering that the elderly are dependent upon a hard-wired low-level neural control policy between the two component effectors. This type of control policy has been found in specifically designed movement tasks involving separate task-goals between two upper limb effector systems (Diedrichsen, 2007; Diedrichsen and Dowling, 2009; Diedrichsen et al., 2010). Applying this area of bimanual coordination is quite new for gait analysis because gait is rarely investigated as two effector systems cooperating either inter-dependently or independently towards a shared task goal. This area will be discussed in more detail in section 8.9.

The correlations which the 'persistence' shares with the 'variance' and 'skewness' can also provide rich information about the loco-sensorimotor system and the differences between the two groups. The hypothesis that the persistence is measuring control effort does not imply that control effort is also representing an increase in signal dependent noise. This is because signal dependent noise relates to the output intensity of the motor commands rather than the 'neural stress' being applied from sensory afferents and descending commands upon the neural centres. Interestingly, the highest correlation between the 'persistence' and 'variance' is found in the vertical component of the stance effector for both groups, however they display significantly opposite associations (r_Y=.60, r_E=-.49). For the elderly, low 'persistence' (i.e. increased control effort) in the stance effector is related to a more 'variable' swing hip position. This is in contrast to the young group, which demonstrates the opposite, such that an increase in 'persistence' (i.e. reduced control effort) is associated with increased 'variance' in the swing hip position.

8.7 Within-effector coordination is task-relevant

Chapter 6 investigated the synergy associated with stabilising the stance and swing effector trajectories. Maximum stability of the stance and swing effectors in the vertical direction, and medio-lateral direction for stance, occurs approximately between 6-12% of proportionate swing time prior to MTC. The synergy for these effector tasks was significant, indicating that the loco-sensorimotor controller manages outgoing motor commands to maximise the stability of the effector endpoint. For the vertical task of the swing effector, there was a significant increase in the synergy strength during the MX1-MTC region of the swing phase. The maximum synergy indicates that the stability of effector states is important prior to MTC. For the elderly group swing effector, they demonstrate significantly stronger synergies compared to the young group (Figure 6.6.1.1A and 6.6.3.1A). This suggests that the loco-sensorimotor workspace of the older adult has been designed for motor solutions that reduce the variance of the swing limb length. Figure 4.8.4.1 demonstrated that the swing effector variance is not significantly lower for the elderly group. When considering Figure 6.6.1.1A and 4.8.4.1, the implication is that, the effect of noise permeating through the elderly system is not reduced significantly in comparison to the young, even though the elderly employ a stronger synergy of the swing effector.

Now with respect to the stance effector, the UCM shows that the degrees of freedom are organized to control vertical stability more than medio-lateral stability, but both tasks of controlling the vertical and medio-lateral swing hip position are considered important by the controller. The elderly appear to consider that the vertical stability is of higher relative importance compared to the medio-lateral stability when compared to the young. When observing Figure 6.7.1 and 6.8.1 the elderly appear to lose the ability to organize the degrees of freedom in the medio-lateral dimension, but when comparing against the young group, this does not appear to occur in the attempt to provide a benefit of stabilizing in the vertical dimension. The groups were not significantly different for the vertical stance effector. In contrast, the elderly were significantly less able to structure the variability in their workspace to stabilise the stance effector's medio-lateral task. It might be that the task of stance effector stability is considered important; however, the neural stress brought about by biomechanical-related constraints means that the stance effector workspace could prevent the vertical degrees of freedom from further evolving into a more organized covariant pattern.

Before proceeding, it must be said that not all synergies (i.e. high UCM_{ratios}) are optimal in the context of some task performance because, by their design, synergies represent stability against component changes. An increased stability of the toe endpoint against perturbations of swing-leg segment angles might come at a cost, particularly if it relates to a need for quick responsive actions like raising the toe-to-ground clearance height. For example, a strong synergy reduces the effect of variations on the task-relevant direction of the effector trajectory. Therefore, if a change to the state of the effector endpoint is required, the controller might need to consider finding a new synergy solution from the workspace.

8.7.1 Synergies associated with the swing effector

Toe-trajectory in the anterior-posterior direction does not appear to be subjected to control like the swing effector task in the vertical direction. Because the same degrees of freedom exist for both the anterior-posterior and vertical tasks, the strong alignment of the degrees of freedom for one task must be freed-up from another. Therefore, strong synergies cannot exist in all dimensions because there is a trade-off. Others have identified a trade-off where task-relevant muscle co-contractions achieve mechanical stability in one dimension at the expense of extra variance in the other dimension (Franklin et al., 2007; Selen et al., 2009). The elderly demonstrate that the degrees of freedom within the swing effector are released from the anterior-posterior task to ensure adequate performance in the vertical direction. This might also be taken to represent impedance control in the vertical dimension at the expense of added noise in the anterior-posterior direction. Therefore, errors in the anterior-posterior appear to be unregulated and can be free to persist. This type of behaviour, which is most prevalent in the elderly, relates to significant correlations in the elderly for anterior-posterior ‘persistence’ correlated with vertical ‘skewness’. So, when stiffness increases as a response to stabilise the vertical state, the degrees of freedom are released in the anterior-posterior direction, and this corresponds with increased ‘persistence’.

As discussed in section 8.2.3.2, the elderly swing effector is mechanically configured in a way that makes the toe state less sensitive to segment variations. The UCM_{ratio} indicates that the controller manages variance to ensure minimal effect on vertical toe state. The elderly also demonstrate a significant rise in synergy strength between MX1+12% and MTC-12%, compared to the young. This suggests significantly increased stiffness, or impedance control, by the elderly in the swing effector prior to MTC.

8.7.2 Synergies associated with the stance effector

The UCM_{ratio} was significantly greater than one for the stance effector in the vertical and medio-lateral tasks. The controller of the young group recognises the vertical swing hip position is important for toe clearance performance, but also medio-lateral position is also important. For the young group, the degrees of freedom trade-off between the medio-lateral and vertical directions is not as apparent when comparing the vertical and anterior-posterior tasks of the swing effector. The stance effector is likely to be controlling stability equally in all three directions (Latash, Scholz, Danion and Schoner, 2001; Robert et al., 2008). Both the young and older adults increase stance effector persistence and synergy strength near MTC. To determine more accurately the contribution to swing hip location versus whole body posture control there would need to be a more elaborate stance effector model that includes not only the body centre of mass position (Scholz and Schoner, 1999). This study was most focused on the kinematic contribution of the stance effector to the toe clearance task by providing a stable swing hip position, irrespective of the stance effector ability to satisfy posture control.

8.7.3 Within-effector coordination is dependent upon MX1 conditions

The synergy parameter (UCM_{ratio}) of the swing (and stance) effector might be scaled in response to the combined neural and mechanical 'pressure' placed upon the controller to change the vertical displacement of the effector, given an initial condition at MX1. These initial conditions were not manipulated by experimental design, because they occurred as a process of natural walking behaviour. The UCM component variance was computed for swing trials belonging to three categories of MX1 initial conditions, which were termed: weak response pressure, moderate response pressure and strong response pressure. To determine whether the synergies are due to these three initial conditions, the change made to alter the effector length across the early-to-mid swing phase can be associated with different synergy strengths (UCM_{ratio}).

There are no significant differences in the UCM_{ratio} behaviour between groups. When comparing the response-dependent UCM_{ratio} across the entire region, the synergies associated with strong effector changes between MX1 and MTC are lower compared to synergies associated with moderate changes in effector states. When considering the UCM component of variance that is in the task-irrelevant direction, the trials associated with weak responses had the task-irrelevant variance, but it also had the highest task-relevant variance. Given that this particular variance structuring between the task-irrelevant variance (i.e. errors don't affect the goal state GEV) and task-relevant variance (i.e. errors affect the goal state, NGEV) is representing a well designed synergy, it can be hypothesized that the neural controller has less pressure to search for appropriate 'synergy producing' motor commands from the workspace of available synergies when the initial conditions are optimal. This could give rise to a number of different synergies being employed to achieve the same outcome. In turn the task-relevant variance has increased and this is a luxury afforded by the controller because errors along the task-relevant dimension are of less consequence (cost of movement errors carry less penalty) when the initial conditions are 'optimal'. Both groups demonstrate that the 'moderate response pressure' has the lowest task-relevant variance and also a relatively low task-redundant variance, such that jointly the UCM_{ratio} is relatively high. This can be explained from the optimal feedback control theory perspective as a commonly recruited muscle synergy which optimally achieves the 'cost-to-go' solution of minimizing energy, movement error and risk, as the toe approaches MTC (Todorov, 2004). For this synergy response, both groups demonstrate the high rate of change in the UCM_{ratio} during the MX1-MTC region.

The 'weak' response-type is associated with significantly stronger synergies prior to the critical transition from MX1 to MTC. This relationship changes prior to MTC. This is a result supportive of the control effort hypothesis, such that redundancy is affected due to control effort. In general, the synergy associated with moderate response to change the state of the effectors from MX1 to MTC is highest when compared to the synergies associated with the strong response trials, or the weak response trials.

8.8 The swing effector is assigned responsibility for making corrections to the toe state between MX1 and MTC

Chapter 7 investigated how the toe clearance task is realised by the cooperation between the stance and swing effectors. The vertical states of the three effectors were correlated.

The correction gain analysis demonstrated that the change in swing effector displacement between MX1 and MTC has a significant negative correlation with the state of the toe at MX1 (section 7.3.4). This suggests that the swing effector makes some contribution to correcting the combined effector position by responding to its location at MX1. For the correlations between the toe state at MX1 and the stance effector displacement between MX1 and MTC, there was a weak prediction ability for the young group (r_Y=-.09). These correlated values were defined as the correction gain made by either the swing or the stance effector. The large correlation demonstrates that the effect of the toe (combined effector) state at MX1 is associated with a response by the effector to change its vertical state from MX1 and MTC. However, the elderly group make significantly less change in the swing effector state as a response to the combined effector at MX1 (r_Y=-.40, r_E=-.24, $p<.001$). To test this more explicitly, a correction gain asymmetry index was computed to contrast the two groups based upon the regression coefficients computed for the correction gains within the separate component effectors. The results demonstrated that given that the MX1 location is used for determining the corrections of the effector states, the elderly group did not use the swing effector significantly more than the stance effector to correct for initial MX1 position (Table 7.3.4.1, section 7.3.4). In contrast, the young demonstrate a significantly stronger ($p < .05$) response by using the swing effector to make correction changes from the state estimates at MX1. This finding contributes to the story of the thesis, such that the elderly have a different role for the stance effector during the toe clearance control task. The young group are more adept at making corrections with the swing effector. The correction gains made by the swing effector were significantly higher in the young group. This demonstrates that the young are likely to be more adept to using sensory information at MX1 to make ‘online’ corrections to the swing effector. The effort to control a trajectory between MX1 and MTC was investigated by computing the persistence for the time series of effector changes made between MX1 and MTC. The persistence suggested that the control effort to find a candidate combined effector trajectory is significantly greater for the combined effector when compared to the swing effector. The elderly group did not display a significant difference between the combined effector and the swing effector (Figure 7.3.3.1). For young healthy adults, this result indicates that the effort to control the vertical states of combined effector trajectories from MX1 to MTC is a significantly different process compared to the effort applied to swing effector trajectories. This is not apparent for the healthy elderly group of this study, and adds further support to previous findings that have suggested that the control effort applied to the combined effector states is reduced in elderly groups at risk of tripping (Khandoker et al., 2008). This opens the interesting possibility of workspace redundancy and the context of performance variables and task variables. For example, the workspace of the young subjects may be designed for redundant solutions, such that an effort to search for candidate solutions for the combined effector is potentially rewarding. For the elderly, if this redundancy does not exist, the effort to search will not provide the same rewards for the cost policy.

To determine whether the strength of the cooperation between the stance and swing effectors predicts the effect of the stance correction gain, a linear regression was fitted to the group data. A negative regression slope of r_Y=-.42 for the young group indicated that the increase in stance-swing coupling was associated with a decrease in the stance correction gain. This means that when there is weak cooperation between the stance-swing effectors, the correction gain on the combined effector will likely come from the stance effector. This indicates that if the MX1-MTC change in swing effector length remains invariant between trials, the stance effector will take responsibility by making corrections to errors in MX1 combined effector states. Figure 7.3.4.2 illustrated that this relationship is apparent but not as consistent for the elderly. There is a sub-group of elderly participants who demonstrate a weak stance-swing cooperation, but they do not show a strong stance correction gain.

8.8.1 Effector correlations referent to time state

The moment-to-moment coupling between the three effectors across the swing-cycle was investigated in section 7.3.1. This analysis did not refer to the MX1 as an initial condition, but rather considered that inter-effector interactions in the vertical direction are not related to some external task goal but depend upon the time state of the swing phase. When determining which effector has a stronger coupling relationship with the combined effector at a time between MX1 and MTC, e.g. MX1+12%, the two groups were significantly different. Figure 7.3.1.1 demonstrated that the swing effector of the young group is not coupled with the combined effector significantly stronger, when compared to the stance-combined coupling ($r_{c\text{-swing}} = .47$ $r_{c\text{-stance}} = .45$, $p=.95$). In contrast, the stance effector of the elderly has a significantly stronger coupling relationship with the combined effector ($r_{c\text{-swing}}=.4$, $r_{c\text{-stance}}=.53$, $p<.001$). The interpretation of these findings is that the swing effector in the elderly is producing a relatively invariant vertical length and therefore any changes to the combined effector are generated by changes in the stance effector. In contrast, the young group do not display a particular coupling bias at MX1+12%.

8.9 Increasing redundancy of between-effector coordination leads to adaptive control of the toe state

An aim of this thesis was to investigate the inter-effector coordination and control effect with the toe clearance task. The stance and swing effectors demonstrate coupling behaviour, which is not stereotypical among participants because it appears that participants vary in the way the task is represented in the neural control hierarchy. The toe position at MX1 can be used as a reference to correct for errors by using either the swing or the stance effectors. Young participants use the swing effector when responding to the MX1 position. In contrast, the elderly use the swing effector to less effect when making corrective changes to the combined effector. This could be due to the way the task is represented in the neural hierarchy. A large correction gain can suggest that the state of the combined effector endpoint, toe, is being monitored and is represented by a state estimate, and the collective effector behaviour responds to this global state-estimate.

Young participants apply control effort to the combined effector relative to the stance and swing effectors. This represents workspace redundancy that can only be achieved through a collective cooperation between the stance and swing effectors. This raises the question of goal sharing between the effectors versus independent goals between the effectors. From the evidence, it appears that the elderly represent the task of walking as goals pursued separately by the stance and swing effectors. The question here is, which response is better for performing a multi-effector task with separate task goals which are likely to appear sequentially as the swing phase evolves? Is low-level adaptation a better choice where one effector is tuned into the state of the alternate effector and responds accordingly? Or, is a third party mediator going to provide more flexible coordination by co-adaptation responses based upon a collective state estimate of a mutual task goal. There has not been any known studies in gait that have addressed this issue. There has been novel control tasks that have only recently begun to look at the benefit of sharing a goal task between two effector systems (Diedrichsen, 2007; Braun et al., 2011). Both studies indicate that there is benefit when effectors share the task goal, and there is benefit when effectors converge upon the same task goal with sufficient time prior to the task goal appearing (Braun et al., 2011).

Mechanically, the swing and stance effectors can act on the same task variable (toe clearance) at the MTC event, irrespective of whether the effectors behave mutually or independently. From an independent task goal perspective, the UCM hypothesis demonstrates that MTC is a task-relevant event for both effectors to improve their end-point stability (UCM_{ratio}>1), whereas the MX1 event is task-relevant only to the swing effector. Therefore, MTC is a task-relevant goal for both effectors, but how does the loco-sensorimotor controller manage the effectors to cooperate together? The coupling of the effectors is 'miscoordinated' at MX1 because increases in stance effector lengths are correlated with decreasing swing effector lengths. As the MTC region approaches, the stance-swing coupling becomes significantly 'coordinated', and is maximally coordinated just prior to MTC. Therefore, as the stance effector lengthens, so too does the swing effector, and the sum of the gain on the combined effector suggests a zero-sum situation. Also, the coordination between the effectors when making change to a trajectory state from MX1 and MTC show a stronger negative correlation (Table 7.3.2.1) compared to observed effector couplings across cycles. This demonstrates the strong within-cycle cooperation, which is almost significantly stronger in the elderly (p=.110). Strong coordination of the stance and swing effector, however, may not be optimal for toe clearance performance. There needs to be consideration of the context in the way the task goals are represented by the loco-sensorimotor controller. If the loco-sensorimotor controller sets up the toe clearance task goal as a mutual goal between the effectors, then there needs to be observations of two criteria. First, the identification of whether the effectors are able to converge on a mutual goal of toe clearance in sufficient time prior to MTC. Second, identification of whether the goal is represented at task space in a low-level controller, or whether the goal is represented at in mutual task space by a higher-level controller. This is important because the coordination between the stance and swing effectors could belong to a more natural hard-wired low-level constraint (i.e. intra-spinal or sub-cortical level), or the coupling could be governed by a hierarchical influence.

If the stance and swing effector were part of separate goals, then they would act in accordance of optimising an independent cost policy in the process of satisfying the goal (Diedrichsen, 2007; Diedrichsen and Dowling, 2009). Assuming the elderly pursue independent task goals, the evidence indicates that the goals are awarded different cost policies. The stance effector cost policy demonstrates that it incurs less penalty for control effort. The cost policy for the swing effector has more control effort penalty for reaching its goal. This might represent the outcome importance of each goal. The swing effector task goal is most likely to be reached with limited control effort. In contrast, the stance effector goal is valued because the penalty cost for control effort is decreased. The control policy for the stance effector might require minimising many costs to satisfy multiple goals, such as forward progression, upright posture and controlling the swing hip state. In this case, the cost of control effort will be relaxed in the attempt to search the workspace for desired solutions. The swing effector in contrast reduces control effort by reducing its search for candidate solutions. It is able to do this when the candidate solution is consistently selected. In this case, the task goal for the swing effector is less complex. However, in the attempt for the stance effector to meet its control policy, the elderly controller adopts a pelvis with less obliquity, and in-turn incurs a medio-lateral instability during MX1. At MX1 the elderly apply significantly more control effort for the swing effector to search for a solution to lateral stability during MX1. The swing effector will need to be 'responsive' to the state-estimates of the stance-effector/upper-body. Under this control policy arrangement, this poses a problem if the stance effector needs to solve the task of toe clearance and upright posture control simultaneously.

For the young subjects, a hypothetical control policy that demonstrates mutual goal sharing between the effectors at MTC can be described by the following. The swing hip position can reduce height, as the stance effector allows pelvic obliquity to occur. The swing effector is tuned into the higher controller and the state-estimate of the endpoint of the combined effector. The stance effector has attained a goal of satisfying the condition for upright posture/balance prior to the MTC region and can share in the task goal of toe clearance. The stance effector and the swing effector both provide sensory feedback information that allows a state-estimate of the combined effector toe position to be optimised. As such, they are in a position to respond optimally to the state of the toe represented in external space, through a collective cost policy. The combined effector reflects this policy in the level of control effort because it now becomes the performance variable.

Based upon these two hypothetical scenarios, that do seem to have plausible supportive evidence from the data, the elderly manage the toe clearance task in a different manner to the young group. It is difficult to know which of the two strategies is more costly for the loco-sensorimotor controller. Most likely, the representation of the toe clearance task by a mutual higher-level controller will require greater central nervous system resources. One issue might be the ability to switch between mutual and independent task goals if the stance effector has satisfied it’s task prior to MTC.

There is a hypothesis of bimanual coordination between two effectors that relates to the state-estimate dependency in which coupling or co-adaptation of the effectors emerges. Figure 8.9.1 illustrates the skew features of this hypothesis. In one situation, the control task is derived by independent state-estimates of each component effector, i.e. details of lengths and orientations, or loading forces (Figure 8.9.1 A). An alternative situation is when the loco-sensorimotor control system computes a state-estimate for the endpoint of the combined effector, i.e. the toe position in external space (Figure 8.9.1 B). The latter situation leads to shared bimanual corrections and optimal performance (Diedrichsen et al., 2010). In this case, coordination of the combined effector can be maintained even when the toe position is not visually observed because there exists a common integration of sensory feedback and a forward model information to optimise the state-estimate of the inferred toe position. Under this state-estimate process, the motor commands sent to each effector are highly coordinated for the task goal. Coordination and control of the effectors will be optimal for flexible toe clearance performance. This model appears to reflect how the young group manage toe clearance performance.

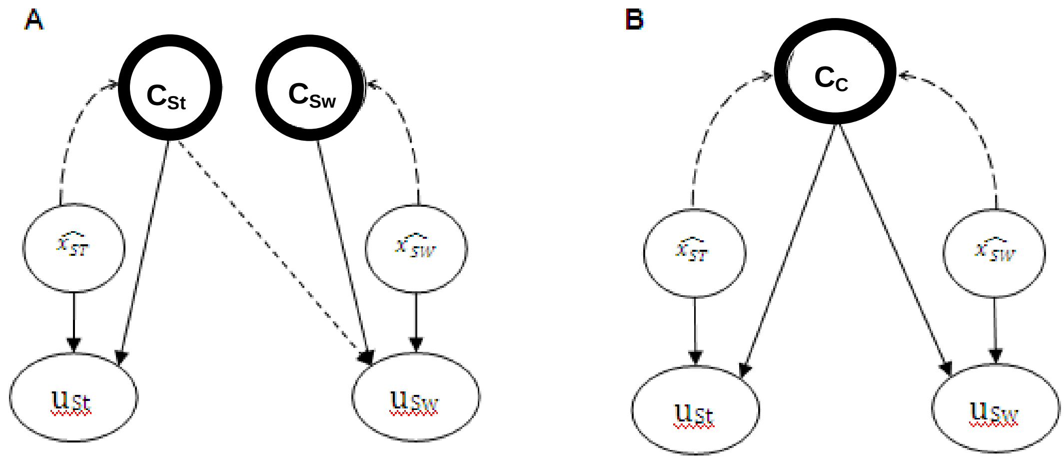


Figure 8.9.1

Coordination between effectors is dependent upon collective or independent state estimates of the performance variable (**C,** within bold circles). The motor commands, *u*, to the effectors (stance and swing effectors, $\hat{x}_{St}$, $\hat{x}_{Sw}$ respectively) are based upon the state estimates of the performance variable, $\mathbf{C_{St}}$**,** $\mathbf{C_{Sw}}$**,** or $\mathbf{C_C}$. The higher level state-estimates ($\mathbf{C_{St}}$) is computed from a forward model (dashed line) of the effectors outgoing motor commands, ***u***. A) for the independent task goals of the stance and swing effectors, the performance variable's state-estimate is updated independently. Alternatively, the swing effector motor commands U_{Sw} can come from the stance effector performance variable and it's state estimate (dashed line). B) for mutual task goal with state-estimates derived from two forward models of the stance and swing effectors to update a common performance variable ($\mathbf{C_C}$). The subsequent motor commands are sent from the common state-estimate of the performance variable ($\mathbf{C_C}$). Figure has been adapted and modified from Diedrichsen et al. (2010).

The discussion in the previous section indicated that the swing effector is related more strongly to the stance effector than the position of the combined effector at MX1. This was found for all participants, however, this was found to be accentuated in the elderly. The strong negative correlations between the respective swing and stance effector vertical displacements from MX1 to MTC, points towards a low-level controller. The results indicate that the swing effector is controlling its actions based upon subserving the state estimates of the stance effector. Walking has been described to behave more like a catching task than a rhythmic tapping task (Hausdorff et al., 2005), and this implies a hierarchical representation of tracking a state estimate in external space. The concept of a 'third party mediator', or a hierarchical state estimate, is a better way for flexible control (Li et al., 2005) for two reasons. First, it is expected that this combined approach allows a more accurate state estimate of the collective variable (i.e. toe clearance of the combined effector) by combining the state estimates of each effector (Diedrichsen and Dowling, 2009). Second, it is expected that the actions between the stance and swing effectors will be more coordinated (Diedrichsen and Dowling, 2009).

Figure 8.9.2 summarises the coordination between the stance and swing effectors for a low-level ('hard-wired') controller. Figure 8.9.2A illustrates the limit-cycle attractor from which coordination between the effectors is stable. Figure 8.9.2B illustrates along the green line where the stable basin of attraction exists. The green area represents coordination, because this is where the actions of the stance and swing effector create a zero-sum solution. There will be a cost when moving away from the green area of the passively stable limit-cycle into the red region of instability.

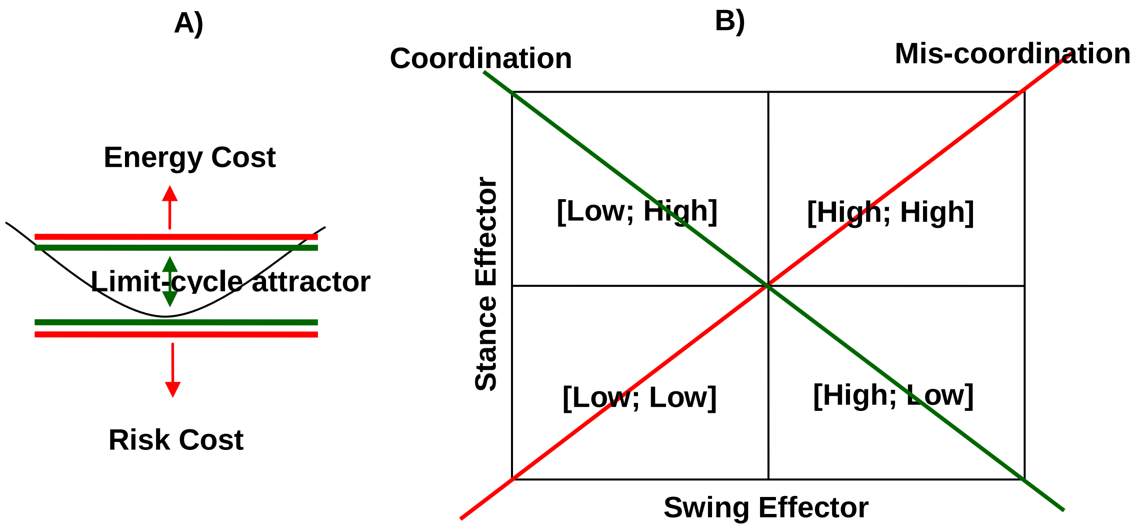


Figure 8.9.2

Coordination between the two effectors. A) regions of stability associated with cost. B) coordination is a stable point attractor, which evolves from a pure Nash equilibria solution and therefore makes the probability of coordination much more likely. Mis-coordination can invoke two costs, one is a kinematic-associated cost of ground contact risk, the other is a kinetic-associated cost of energy inefficiency.

The swing effector demonstrates persistence for both the young and elderly groups. This is likely to be due to different underlying cost policies and control strategies. The swing effector of the young group has the luxury of searching a high-dimensional workspace with abundant candidate solutions. The swing effector of the elderly group may or may not have a high-dimensional workspace, but the role of the swing effector may be utilising less of this capacity in the quest for solution consistency. The control effort is low because the set of solutions is tolerated by the controller. The biomechanical ability of the stance effector in the elderly appears to be dictating the type of cost policy adopted. In the vertical direction of the elderly stance effector, there is a significantly extended period of low 'persistence' as the MTC task approaches (Figure 5.8.2.1). This potentially signals the inability to shift from an independent task goal to a mutually shared task goal with the swing effector. If the stance effector is executing with less certainty of the swing hip location, then the swing effector will need to adjust its control policy if it is to maintain the same level of control effort. Minimising control effort resources in the elderly is potentially a high cost because of their reduced processing capacities (review, section 2.2.3.2.2). The change in policy for the swing effector may reduce the combined penalties for increased energy cost and increased movement error. The outcome of will be energy inefficient, and it will result in a higher toe state (reduce uncertainty of ground contact risk) with more variability. Figure 4.8.4.1 illustrated the significant 'no-change' in the combined effector variability between MX1 and MTC. This result is potentially signalling a policy that is tolerating error and indicating independent goals between the swing and stance effector.

If a switch to a higher level controller takes place, such that the end-effector position is used rather than the stance effector state estimate, then the stance and swing effectors respond mutually for controlling the combined effector. If the system has a degree of redundancy the mutual response should demonstrate persistence for both the stance and swing effectors, while the controlled end-effector position demonstrates anti-persistence. If the system has reduced redundancy, the stance and swing effectors will demonstrate more anti-persistence in the attempt to control the collective end-point effector. Lastly, if the system is unable to converge upon a mutual task goal then the effectors remain adaptive to each other and the collective end-point is not represented in external space. The outcome is an end-point that is free to persist under the interactions of local state estimates of the stance and swing effector systems. This points towards a sub-optimal state of performance for governing the final trajectory of the toe as it approaches the critical MTC event. Persistence in the combined effector is higher than the swing effector in the elderly, however, this was not the case in the young participants. The elderly will find it difficult to converge at a mutual task goal because the stance effector goal has not been established early enough. It is hypothesised that the stance and swing effectors of the young participants are able to converge earlier, and represent the toe clearance task mutually in external space, therefore doing a better job of controlling the collective end-effector. This finding was most prominent in the non-dominant limb comparisons of young and elderly.

The above discussion indicated the need for the swing effector to pay close attention to the stance effector position because the stance effector is pre-occupied with an alternate task goal. Provided the swing effector anticipates that current position of the swing hip at MX1 and the change to be made by the swing hip between MX1 and MTC, the swing effector can execute a co-adapting strategy. Irrespective of the initial MX1 position, the vertical displacements made by the stance and swing effector are strongly negatively correlated. This is consistent with the co-adaption coupling theme. This correlation is stronger for the elderly participants, which demonstrate that the stance effector change predicts the swing effector change. This is a stronger relationship than the swing effector displacement change based upon the MX1 initial position. This raises the interesting issue of how the loco-sensorimotor control system encodes a reference system for determining changes, and research has shown that limb length and orientation kinematics, as well as stance limb kinetics is related to dorsal spinal-cerebellum tract activity (Bosco and Poppele, 2000; Bosco, Poppele and Eian, 2000; Bosco et al., 2006). For the toe clearance task, making corrections in the swing effector is not based predominantly upon an external reference like the toe position at MX1, but rather it is an internal – possibly hard-wired coordination – coding relevant to the dynamics of the stance effector. A more thorough multi-regression analysis will demonstrate a more precise conclusion, but from these initial results it appears the swing effector acting out commands based from 'intrinsic' low-level state estimates at the joint level (and limb length/orientation) associated with the stance effector, rather than responding to the collective effector task goal represented in higher-level state estimates. This finding points towards a learned state response, which is likely to be processed through the lower level CPG circuitry. Some motor control studies have demonstrated that when familiarity of the task is gained and task complexity is reduced, the state estimate reference frame that represents the task will gradually move from a body reference to an external reference (Ahmed, Wolpert and Flanagan, 2008). The more complex tasks are represented in body level states and familiarity allows higher level encoding. This contradicts the CPG hypothesis for automated walking coordination patterns and reducing higher level control intervention and other findings into limb based reference frames in walking (e.g. Bosco et al., 2006), while other studies have also shown a contrast (Shadmehr and Moussavi, 2000). This is also an effect of ageing, because the low-level coordination coupling is a behaviour appearing to be adopted more strongly by the elderly.

Diedrichsen et al. (2010) suggests that human coordination has evolved to achieve single goals flexibly, rather than achieving multiple goals simultaneously. This means that the human body cannot optimally achieve simultaneously the toe clearance task and the task of body posture control, however, if task goals appear sequentially, the system is encompassed with the flexibility to re-assign effectors to achieve both task goals well. This emphasizes the importance for establishing posture control prior to the MTC event occurring for allowing maximal flexibility of coordination (redundancy). The finding that the swing effector and the stance effector are strongly correlated could represent some partial effect from a hard-wired sub-cortical gating mechanism (see review by Diedrichsen et al., 2010) which creates a symmetric response between effectors when independent task goals are presented simultaneously. This suggests that if a rise in the swing hip is required at a similar time when a rise in the swing effector is required to achieve adequate toe clearance, there is the gating constraint that could manifest as a coupling and correlation between the effectors. The finding that the control effort in the stance effector continues for longer into the swing phase and that there remains an upward velocity of the stance effector at MTC, suggests a prolonged extensor effort by the stance limb. Ideally, both effectors are acting on the same task variable.

8.10 Study limitations

8.10.1 Subject sampling

Both elderly males and females are at risk of falls. However, the results from this thesis cannot be extrapolated to the male population of healthy older adults. Single gender was selected because it was deemed to provide greater homogeneity of the elderly subject group. This is because a difference exists between males and females with the ageing process and this is exemplified by falls-related hospital admissions in the age group of females that matches those of males whom are ten years their senior.

Within group sub-samples. Comment here on the issue that there are likely to be groups of subjects that act similarly, this may be predisposed to a biomechanical inclination.

8.10.2 Screening of subjects and inferring age-related changes are due to the natural ageing process

The subject screening procedure probably indicates that the classification of the elderly subjects was a delimitation rather than a limitation in this study. However, it is quite possible that the subjects included did have some type of pathology that was affecting their gait. This is a problem for all research studies investigating elderly and the screening problems associated with detecting co-morbidities and functional impairments.

The premise of the thesis studies was to find how the natural ageing process would affect toe trajectory control. This required selection of the elderly so that pathological ageing could be confidently ‘ruled out’ as a cause of any differences found between young and elderly. It is a limitation of this study, and most studies investigating the effects of healthy ageing, such that the sample may not represent healthy ageing. The procedure that was administered to ensure that the participants were healthy was quite thorough. The steps of screening ensured that the subjects were at least average for physical performance. The scores for the timed up and go test were all below 11 seconds, and it was found in a sample of well screened healthy elderly females that the group average TUG test was 9.7 seconds. This further demonstrates that the subjects were of sufficient physical ability. The TUG test is also found to be a strong predictor for impaired executive function. Given the process for testing, the methods undertaken give confidence for ensuring the subjects were of sufficient physical and cognitive health.

8.10.3 Issues of encumbrance when walking

It is interesting to find that there have been no studies found in the Journal of Biomechanics or the Journal of Gait and Posture, which have investigated the effect of marker set-up on gait pattern. The type of set up applied in this study is unique, and was probably the first study in the world to use such an elaborate set up, with active markers and marker clusters. This is a limitation of the experiment, because the degree of encumbrance may have implicitly affected the walking patterns. However, 60+ people providing subjective anecdotal feedback about the encumbrance of the set up was that: “they didn’t notice the equipment, and it did not feel like it restricted their normal walking, and there was no sense of constraint, hindrance or discomfort during testing. There are no studies that have investigated changes to walking patterns based on the fitting of equipment to body parts of subjects. Although the test set up involved elaborate equipment, the participants subjective report of walking encumbrance was. The neoprene wrap are a standard method for securing marker clusters. A wrap-around pelvic belt was positioned around the ASIS and PSIS bony prominences to minimise discomfort. Marker clusters provide greater 6-DoF accuracy for body motion, and while the degree of encumbrance increases with any marker set-up, there is greater confidence of reporting accurate data. The equipment was wired, but the wires were placed posterior to the lower limbs. The wires connected between the markers and a backpack were secured ‘out-of-sight’ posterior aspect of subject’s lower limbs and trunk, and were of sufficient length to allow a full range of joint motion. The subjects performed a standing squat prior to walking to ensure the freedom to fully flex the hip, knee and ankle joints in the sagittal plane. Subjects were able to swing their arms freely during walking. The lower limb range of motion was not constrained. The backpack was padded and compact. No participants reported discomfort from the testing set up. The spring loaded harness provided ‘un-weighting’ of the test backpack, so the weight of this device had no effect on backpack weight. Particular care was provided during the marker set up to ensure that there was no change to the range of motion of the joints. There was a treadmill familiarity test, and during this time the subjects were frequently asked to report any presence of discomfort or hindrance as they walked. There were certain problems on the odd occasion, however, the issues were modifiable. They generally related to tightness of taping or a loose wire. The set-up was consistent between all participants of the study. The weight of the harness or backpack was never reported as a concern and any change to walking pattern during the test process would not likely to be due to this issue.

8.11 Conclusion

This study has provided both theoretical and practical outcomes to help address the problem of trip-related falls in the community by examining the way the toe clearance task is performed, with particular emphasis on factors that might theoretically be associated with tripping.

There is a difference in the way older adults walk when compared to their younger counterparts. There are many changes made to walking patterns because of ageing and these have been well documented. The particular changes that are made to walking that can place a person at increased tripping risk have not been extensively studied, for example, performing the toe clearance task while walking. With relevance to the toe clearance task, this study demonstrated how the loco-sensorimotor system changes biomechanically with age and how it adapts internally by designing a new task-management plan. The new management plan adopted has implications for the way an older person will plan their toe trajectories during walking. When young people perform the task of toe clearance, they do it by collectively involving the stance and swing limb, such that the task solution emerges from the mutual information sharing between the systems belonging to the stance and swing limb. In contrast, the elderly perform the same task by partitioning their system into independent components, whereby the support limb undertakes responsibility for maintaining posture control and toe clearance. For the elderly, the swinging limb has been given a minor role in this task management policy. The effect of adopting this management policy results in a sub-optimal solution, and the state of the toe will be represented with respect to the ground surface in a less precise way. The reason why the older adults don't apply the same management process as young people do, is because it costs more central nervous system effort to search for movement solutions. The older adults ageing process has reduced the abundant reserves of high-dimensional movement solutions into a low dimensional replica. The search through low-dimensional 'work'-space can become exhaustive when there is a goal state to be found. This theory is consistently supported by the results from this thesis.

The walking pattern observed in the elderly participants of this study display a biomechanical configuration of the stance limb, which suggests that more effort will be required to allow it to perform the basic tasks of forward progression and upright support. The result of needing more expensive motor demands indicates that they have got themselves into a walking position that cannot take 'passive' advantage of the musculo-skeletal system. The stance limb therefore has a bigger task to carry the trunk and send it forwards. It is likely that the demands of this task, prevents the 'manager', (i.e., the loco-sensorimotor controller), from performing the task of toe clearance in an optimal way. This thesis shows that the control effort associated with the support limb extends much longer into the swing phase for the elderly group. As a consequence, the task of toe clearance cannot be represented as a goal formed by collaboration between both lower limbs. Instead, the toe clearance task for the elderly demonstrates independent roles between limbs. The swinging limb acts in response to the stance limb, but mainly the swinging limb is concerned with reproducing a consistent limb length. Therefore, when the swinging toe approaches a state where it will travel at approximately 1cm from the ground, the stance limb takes responsibility for a new task goal, and that is, to ensure the toe clears the ground surface. This independent assignment of the lower limbs is a suboptimal solution and one that the younger people don't apply. Further work will need to explore how the elderly can regain the mutual assignment between the swinging limb and stance limb, and how, and when and why this change comes about as ageing progresses.

The ageing process affects the loco-sensorimotor system in various ways that are not consistent between individuals. The results of this study emerged from general statistics and considered both groups homogenous samples of their population. For studies into elderly groups, this outcome can be hidden because of the relatively larger diversity within the population. There were some elderly participants in the sample who demonstrated similar behaviours to the average of the young group. Likewise, there were young participants that demonstrated walking behaviours closer to the elderly group. Nevertheless, the results obtained within this thesis maintained a common thread of evidence when comparing between the groups. Further work will explore sub-samples within the groups (e.g. via cluster analysis) to identify the association of these walking policies with particular group characteristics, using groupings other than the common ‘young’ versus ‘elderly’ comparisons.

This study has brought about new knowledge to the field of falls prevention and walking control. It has applied existing tools in new ways and applied new research theories to explain complex phenomena.

8.12 Clinical implications of the results

The implication of the outcomes from the four studies have clinical implications. A basic message from this study suggests that posture control training could potentially improve the ability to negotiate and clear obstacles by the swing limb. The finding that Tai Chi has positive benefits for minimising falls risk might be due to its effective outcome on posture control. This in turn can effect toe trajectory control. Further studies should investigate the effect of Tai Chi on the control of the swing hip position earlier in single leg stance phase, and hence allow the loco-sensorimotor control system to facilitate cooperation between the stance and swing limbs for negotiating an external ground reference. This ensures redundancy and flexibility to find an adequate movement solution.

The study demonstrated that toe clearance is controlled better when posture control is established earlier in the swing phase. This sub-task of walking might correspond to certain exercise intervention programs that administer exercises that parallel the facilitation of this movement task.

The swing limb in the elderly appears to be less flexible to alter its position and there should be exercise intervention to help facilitate more adaptable solutions. It would be a first step to investigate whether the re-establishment of posture control in elderly will in-turn facilitate responsive swing limb solutions.

A prospective study that includes these measures with posture control exercises, like Tai Chi, is worthy of being investigated. The effect that Tai Chi can have for the establishing redundant and flexible solutions that optimise the management of safe toe trajectories can be determined. In addition, a follow up study of prospective falls will help to reveal the ultimate effect these issues have for reducing trip-related falls.

8.13 Proposal for future research

- Cluster analysis of sub-groups. A limitation of this thesis is that there could be elderly participants acting in a similar way to young participants and vice versa.
- Validating persistence as a measure of control effort and investigating other potential parameters that can capture control effort
- Investigating control effort behind realising synergies
- Incorporating a more complete effector model by introducing the body centre of mass. The limitation of the current study is the assumption of swing hip representing posture control. A more complete understanding of MTC control can be determined by delineating between swing hip position control and posture control.
- Probing the interactions of the stance and swing effector dependency. Independent versus mutual effector goals. The effect of coupling between the stance and swing effectors could be due to task goal representation. The effect of re-learning a mutual task goal might de-couple this relationship.
- This can be investigated through experimental design incorporating:
    - Effect of altering the performance goals:
        - posture control target
        - toe clearance height target
    - Effect of altering the sensory feedback:
        - E.g. virtual reality effect on optic flow
    - Effect of noise in the movement system:
        - Noisy force field acting on the foot during leg swing
        - Noisy forces acting on posture control during leg swing
    - Effect of biomechanics:

        - E.g. re-learning effector configurations
    - Forward dynamics investigation of each subject to determine their muscle coordination strategy
    - Overground and treadmill walking
    - Walking speeds
    - Investigate the effect of different control policies on tripping likelihood
    - Follow up study with the elderly participants to determine inter-effector gait changes and falls history
    - Controlled tripping studies on young and elderly subjects

References

Ahmed, A. A., Wolpert, D. M., Flanagan, J. R. 2008. Flexible representations of dynamics are used in object manipulation. Current Biology 18, 763-768.

Anderson, F. C., Goldberg, S. R., Pandy, M. G., Delp, S. L. 2004. Contributions of muscle forces and toe-off kinematics to peak knee flexion during the swing phase of normal gait: an induced position analysis. Journal of Biomechanics 37, 731-737.

Anderson, F. C., Pandy, M. G. 2001. Dynamic optimization of human walking. Journal of Biomechanical Engineering 123, 381-390.

Anderson, F. C., Pandy, M. G. 2003. Individual muscle contributions to support in normal walking. Gait and Posture 17, 159-169.

Anstey, K. J., Wood, J., Kerr, G., Caldwell, H., Lord, S. R. 2009. Different cognitive profiles for single compared with recurrent fallers without dementia. Neuropsychology 23, 500-508.

Ashkenazy, Y., M. Hausdorff, J., Ch. Ivanov, P., Eugene Stanley, H. 2002. A stochastic model of human gait dynamics. Physica A: Statistical Mechanics and its Applications 316, 662-670.

Australian Bureau of Statistics. 2009. *Population Projections Australia, 2006-2101*. Catalog number 3222.0., ABS, Canberra.

Baker, R. 2001. Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait and Posture 13, 1-6.

Baker, R. 2006. Gait analysis methods in rehabilitation. Journal of Neuroengineering and Rehabilitation 3, 4.

Baker, R., McGinley, J. L., Schwartz, M. H., Beynon, S., Rozumalski, A., Graham, H. K., Tirosh, O. 2009. The gait profile score and movement analysis profile. Gait and Posture 30, 265-269.

Barrett, R. S., Mills, P. M., Begg, R. K. 2010. A systematic review of the effect of ageing and falls history on minimum foot clearance characteristics during level walking. Gait and Posture 32, 429-435.

Bashan, A., Bartsch, R., Kantelhardt, J. W., Havlin, S. 2008. Comparison of detrending methods for fluctuation analysis. Physica A: Statistical Mechanics and its Applications 387, 5080-5090.

Bauby, C. E., Kuo, A. D. 2000. Active control of lateral balance in human walking. Journal of Biomechanics 33, 1433-1440.

Bays, P. M., Wolpert, D. M. 2007. Computational principles of sensorimotor control that minimize uncertainty and variability. Journal of Physiology 578, 387-396.

Begg, R., Best, R., Dell'Oro, L., Taylor, S. 2007. Minimum foot clearance during walking: Strategies for the minimisation of trip-related falls. Gait and Posture 25, 191-198.

Begg, R. K., Sparrow, W. A. 2000. Gait characteristics of young and older individuals negotiating a raised surface: implications for the prevention of falls. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 55, 147-154.

Benedetti, M. G., Catani, F., Leardini, A., Pignotti, E., Giannini, S. 1998. Data management in gait analysis for clinical applications. Clinical Biomechanics 13, 204-215.

Berg, W. P., Alessio, H. M., Mills, E. M., Tong, C. 1997. Circumstances and consequences of falls in independent community-dwelling older adults. Age and Ageing 26, 261-268.

Bernstein, N. 1967. The coordination and regulation of movements. Oxford: Pergamon Press

Best, R., Begg, R. 2008. A method for calculating the probability of tripping while walking. Journal of Biomechanics 41, 1147-1151.

Bianchi, L., Angelini, D., Orani, G. P., Lacquaniti, F. 1998. Kinematic coordination in human gait: relation to mechanical energy cost. Journal of Neurophysiology 79, 2155-2170.

Bizzi, E., Cheung, V. C., d'Avella, A., Saltiel, P., Tresch, M. 2008. Combining modules for movement. Brain research reviews 57, 125-133.

Black, D. P., Smith, B. A., Wu, J., Ulrich, B. D. 2007. Uncontrolled manifold analysis of segmental angle variability during walking: preadolescents with and without Down syndrome. Experimental Brain Research 183, 511-521.

Blake, A. J., Morgan, K., Bendall, M. J., Dallosso, H., Ebrahim, S. B., Arie, T. H., Fentem, P. H., Bassey, E. J. 1988. Falls by elderly people at home: prevalence and associated factors. Age and Ageing 17, 365-372.

Borghese, N. A., Bianchi, L., Lacquaniti, F. 1996. Kinematic determinants of human locomotion. Journal of Physiology 494 (Pt 3), 863-879.

Bosco, G., Eian, J., Poppele, R. E. 2005. Kinematic and non-kinematic signals transmitted to the cat cerebellum during passive treadmill stepping. Experimental Brain Research 167, 394-403.

Bosco, G., Eian, J., Poppele, R. E. 2006. Phase-specific sensory representations in spinocerebellar activity during stepping: evidence for a hybrid kinematic/kinetic framework. Experimental Brain Research 175, 83-96.

Bosco, G., Poppele, R. E. 2000. Reference frames for spinal proprioception: kinematics based or kinetics based? Journal of Neurophysiology 83, 2946-2955.

Bosco, G., Poppele, R. E., Eian, J. 2000. Reference frames for spinal proprioception: limb endpoint based or joint-level based? Journal of Neurophysiology 83, 2931-2945.

Brach, J. S., Berlin, J. E., VanSwearingen, J. M., Newman, A. B., Studenski, S. A. 2005. Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. Journal of Neuroengineering and Rehabilitation 2, 21-21.

Brach, J. S., Berthold, R., Craik, R., VanSwearingen, J. M., Newman, A. B. 2001. Gait variability in community-dwelling older adults. Journal of the American Geriatrics Society 49, 1646-1650.

Brach, J. S., Studenski, S., Perera, S., Vanswearingen, J. M., Newman, A. B. 2007a. Stance time and step width variability have unique contributing impairments in older persons. Gait and Posture

Brach, J. S., Studenski, S. A., Perera, S., VanSwearingen, J. M., Newman, A. B. 2007b. Gait variability and the risk of incident mobility disability in community-dwelling older adults. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 62, 983-988.

Braun, D. A., Ortega, P. A., Wolpert, D. M. 2009. Nash equilibria in multi-agent motor interactions. PLoS Computational Biology 5, e1000468.

Braun, D. A., Ortega, P. A., Wolpert, D. M. 2011. Erratum to: Motor coordination: when two have to act as one. Experimental Brain Research 212, 645-646.

Bruijn, S. M., Meijer, O. G., van Dieen, J. H., Kingma, I., Lamoth, C. J. C. 2007. Coordination of leg swing, thorax rotations, and pelvis rotations during gait: The organisation of total body angular momentum. Gait and Posture

Buczek, F. L., Rainbow, M. J., Cooney, K. M., Walker, M. R., Sanders, J. O. 2010. Implications of using hierarchical and six degree-of-freedom models for normal gait analyses. Gait and Posture 31, 57-63.

Byl, K., Tedrake, R. 2009. Metastable Walking Machines. International Journal of Robotics Research 28, 1040-1064.

Callisaya, M. L., Blizzard, L., McGinley, J. L., Schmidt, M. D., Srikanth, V. K. 2010. Sensorimotor factors affecting gait variability in older people--a population-based study. The journals of gerontology. Series A, Biological sciences and medical sciences 65, 386-392.

Callisaya, M. L., Blizzard, L., Schmidt, M. D., McGinley, J. L., Lord, S. R., Srikanth, V. K. 2009. A population-based study of sensorimotor factors affecting gait in older people. Age and Ageing 38, 290-295.

Camomilla, V., Cereatti, A., Vannozzi, G., Cappozzo, A. 2006. An optimized protocol for hip joint centre determination using the functional method. Journal of Biomechanics 39, 1096-1106.

Campbell, A. J., Borrie, M. J., Spears, G. F. 1989. Risk factors for falls in a community-based prospective study of people 70 years and older. Journal of Gerontology 44, 112-117.

Campbell, A. J., Borrie, M. J., Spears, G. F., Jackson, S. L., Brown, J. S., Fitzgerald, J. L. 1990. Circumstances and consequences of falls experienced by a community population 70 years and over during a prospective study. Age and Ageing 19, 136-141.

Cappozzo, A., Cappello, A., Della Croce, U., Pensalfini, F. 1997. Surface-marker cluster design criteria for 3-D bone movement reconstruction. IEEE Transactions on Biomedical Engineering 44, 1165-1174.

Cappozzo, A., Catani, F., Croce, U. D., Leardini, A. 1995. Position and orientation in space of bones during movement: anatomical frame definition and determination. Clinical Biomechanics 10, 171-178.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., Croce, U. D. 1996. Position and orientation in space of bones during movement: experimental artefacts. Clinical Biomechanics 11, 90-100.

Cappozzo, A., Della Croce, U., Leardini, A., Chiari, L. 2005. Human movement analysis using stereophotogrammetry: Part 1: theoretical background. Gait and Posture 21, 186-196.

Cavagna, G., Heglund, N. C., Taylor, C. R. 1977. Mechanical work in terrestrial locomotion: two basic mechanisms for minimising energy expenditure. American journal of physiology. 233, R243-R261.

Chapman, G. J., Hollands, M. A. 2007. Evidence that older adult fallers prioritise the planning of future stepping actions over the accurate execution of ongoing steps during complex locomotor tasks. Gait and Posture 26, 59-67.

Charlton, I. W., Tate, P., Smyth, P., Roren, L. 2004. Repeatability of an optimised lower body model. Gait and Posture 20, 213-221.

Chen, J. S., Sambrook, P. N., Simpson, J. M., Cameron, I. D., Cumming, R. G., Seibel, M. J., Lord, S. R., March, L. M. 2009. Risk factors for hip fracture among institutionalised older people. Age and Ageing 38, 429-434.

Chen, J. S., Simpson, J. M., March, L. M., Cameron, I. D., Cumming, R. G., Lord, S. R., Seibel, M. J., Sambrook, P. N. 2008a. Fracture risk assessment in frail older people using clinical risk factors. Age and Ageing 37, 536-541.

Chen, J. S., Simpson, J. M., March, L. M., Cameron, I. D., Cumming, R. G., Lord, S. R., Seibel, M. J., Sambrook, P. N. 2008b. Risk factors for fracture following a fall among older people in residential care facilities in Australia. Journal of the American Geriatrics Society 56, 2020-2026.

Choi, J. T., Bastian, A. J. 2007. Adaptation reveals independent control networks for human walking. Nature Neuroscience 10, 1055-1062.

Chou, L.-S., Draganich, L. F. 1998. Placing the trailing foot closer to an obstacle reduces flexion of the hip, knee, and ankle to increase the risk of tripping. Journal of Biomechanics 31, 685-691.

Chou, L. S., Song, S. M., Draganich, L. F. 1995. Predicting the kinematics and kinetics of gait based on the optimum trajectory of the swing limb. Journal of Biomechanics 28, 377-385.

Christou, E. A., Enoka, R. M. 2011. Aging and movement errors when lifting and lowering light loads. Age 33, 393-407.

Christou, E. A., Shinohara, M., Enoka, R. M. 2003. Fluctuations in acceleration during voluntary contractions lead to greater impairment of movement accuracy in old adults. Journal of Applied Physiology 95, 373-384.

Christou, E. A., Tracy, B. L. 2006. Aging and variability in motor output. In: Davids, K., Bennett, S., Newell, K. M. (Eds.), Movement System Variability. Human Kinetics, Auckland, pp. 199-215.

Cohen, J. 1988. Statistical power analysis for the behavioural sciences. Laurence Earlbaum Associates

Cole, G. K., Nigg, B. M., Ronsky, J. L., Yeadon, M. R. 1993. Application of the joint coordinate system to three-dimensional joint attitude and movement representation: a standardization proposal. Journal of Biomechanical Engineering 115, 344-349.

Collins, S., Ruina, A., Tedrake, R., Wisse, M. 2005. Efficient bipedal robots based on passive-dynamic walkers. Science 307, 1082-1085.

Cripps, R., Carman, J. 2001. Falls by the Elderly in Australia: trends and data for 1998. Injury Research and Statistics Series. In: AIHW cat. no. INJCAT 35. Australian Institute of Health and Welfare, Adelaide

Cusumano, J. P., Cesari, P. 2006. Body-goal variability mapping in an aiming task. Biological Cybernetics 94, 367-379.

d'Avella, A., Saltiel, P., Bizzi, E. 2003. Combinations of muscle synergies in the construction of a natural motor behavior. Nature Neuroscience 6, 300-308.

Daffertshofer, A., Lamoth, C. J. C., Meijer, O. G., Beek, P. J. 2004. PCA in studying coordination and variability: a tutorial. Clinical Biomechanics 19, 415-428.

Delignieres, D., Torre, K. 2009. Fractal dynamics of human gait: a reassessment of the 1996 data of Hausdorff et al. Journal of Applied Physiology 106, 1272-1279.

Della Croce, U., Leardini, A., Chiari, L., Cappozzo, A. 2005. Human movement analysis using stereophotogrammetry: Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. Gait and Posture 21, 226-237.

DeVita, P., Hortobagyi, T. 2000. Age causes a redistribution of joint torques and powers during gait. Journal of Applied Physiology 88, 1804-1811.

Diedrichsen, J. 2007. Optimal task-dependent changes of bimanual feedback control and adaptation. Current Biology 17, 1675-1679.

Diedrichsen, J., Dowling, N. 2009. Bimanual coordination as task-dependent linear control policies. Human Movement Science 28, 334-347.

Diedrichsen, J., Shadmehr, R., Ivry, R. B. 2010. The coordination of movement: optimal feedback control and beyond. Trends in Cognitive Sciences 14, 31-39.

Dimitrijevic, M. R., Gerasimenko, Y., Pinter, M. M. 1998. Evidence for a spinal central pattern generator in humans. Annals of the New York Academy of Sciences 860, 360-376.

Dingwell, J. B., Cusumano, J. P. 2010. Re-interpreting detrended fluctuation analyses of stride-to-stride variability in human walking. Gait and Posture 32, 348-353.

Dingwell, J. B., Cusumano, J. P., Cavanagh, P. R., Sternad, D. 2001. Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. Journal of Biomechanical Engineering 123, 27-32.

Dingwell, J. B., Cusumano, J. P., Sternad, D., Cavanagh, P. R. 2000. Slower speeds in patients with diabetic neuropathy lead to improved local dynamic stability of continuous overground walking. Journal of Biomechanics 33, 1269-1277.

Dingwell, J. B., John, J., Cusumano, J. P. 2010. Do humans optimally exploit redundancy to control step variability in walking? PLoS Computational Biology 6, e1000856.

Dingwell, J. B., Kang, H. G. 2007. Differences between local and orbital dynamic stability during human walking. Journal of Biomechanical Engineering 129, 586-586.

Dingwell, J. B., Marin, L. C. 2006. Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. Journal of Biomechanics 39, 444-452.

Dingwell, J. B., Robb, R. T., Troy, K. L., Grabiner, M. D. 2008. Effects of an attention demanding task on dynamic stability during treadmill walking. Journal of Neuroengineering and Rehabilitation 5, 12.

Dingwell, J. B., Ulbrecht, J. S., Boch, J., Becker, M. B., O'Gorman, J. T., Cavanagh, P. R. 1999. Neuropathic gait shows only trends towards increased variability of sagittal plane kinematics during treadmill locomotion. Gait and Posture 10, 21-29.

Donelan, J. M., Kram, R., Kuo, A. D. 2002. Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. Journal of Experimental Biology 205, 3717-3727.

Donker, S. F., Daffertshofer, A., Beek, P. J. 2005. Effects of velocity and limb loading on the coordination between limb movements during walking. J Mot Behav 37, 217-230.

Doya, K. 2009. How can we learn efficiently to act optimally and flexibly? Proceedings of the National Academy of Sciences of the United States of America 106, 11429-11430.

Durlach, N. I., Mavor, A. S. (Eds.) 1995. Virtual Reality: scientific and technological challenges. National Academy Press, Washington, DC.

Duysens, J., Clarac, F., Cruse, H. 2000. Load-regulating mechanisms in gait and posture: comparative aspects. Physiological Reviews 80, 83-8133.

Ehrig, R. M., Taylor, W. R., Duda, G. N., Heller, M. O. 2006. A survey of formal methods for determining the centre of rotation of ball joints. Journal of Biomechanics 39, 2798-2809.

Elble, R. J., Thomas, S. S., Higgins, C., Colliver, J. 1991. Stride-dependent changes in gait of older people. Journal of Neurology 238, 1-5.

Emken, J. L., Benitez, R., Sideris, A., Bobrow, J. E., Reinkensmeyer, D. J. 2007. Motor adaptation as a greedy optimization of error and effort. Journal of Neurophysiology 97, 3997-4006.

England, S. A., Granata, K. P. 2007. The influence of gait speed on local dynamic stability of walking. Gait and Posture 25, 172-178.

Englander, F., Hodson, T. J., Terregrossa, R. A. 1996. Economic dimensions of slip and fall injuries. Journal of Forensic Sciences 41, 733-746.

Enoka, R. M., Christou, E. A., Hunter, S. K., Kornatz, K. W., Semmler, J. G., Taylor, A. M., Tracy, B. L. 2003. Mechanisms that contribute to differences in motor performance between young and old adults. Journal of Electromyography and Kinesiology 13, 1-12.

Faisal, A. A., Selen, L. P., Wolpert, D. M. 2008. Noise in the nervous system. Nature Reviews. Neuroscience 9, 292-303.

Farrell, S., Wagenmakers, E. J., Ratcliff, R. 2006. 1/f noise in human cognition: is it ubiquitous, and what does it mean? Psychonomic Bulletin and Review 13, 737-741.

Field, A. 2009. Discovering statistics using SPSS. SAGE Publications Ltd, London.

Forner-Cordero, A., Koopman, H. J., van der Helm, F. C. 2006. Describing gait as a sequence of states. Journal of Biomechanics 39, 948-957.

Franklin, D. W., Burdet, E., Tee, K. P., Osu, R., Chew, C. M., Milner, T. E., Kawato, M. 2008. CNS learns stable, accurate, and efficient movements using a simple algorithm. The Journal of neuroscience : the official journal of the Society for Neuroscience 28, 11165-11173.

Franklin, D. W., Liaw, G., Milner, T. E., Osu, R., Burdet, E., Kawato, M. 2007. Endpoint stiffness of the arm is directionally tuned to instability in the environment. The Journal of neuroscience : the official journal of the Society for Neuroscience 27, 7705-7716.

Friedman, S. M., Munoz, B., West, S. K., Rubin, G. S., Fried, L. P. 2002. Falls and fear of falling: which comes first? A longitudinal prediction model suggests strategies for primary and secondary prevention. Journal of the American Geriatrics Society 50, 1329-1335.

Fuller, J., Liu, L. J., Murphy, M. C., Mann, R. W. 1997. A comparison of lower-extremity skeletal kinematics measured using skin- and pin-mounted markers. Human Movement Science 16, 219-242.

Gabell, A., Nayak, U. S. 1984. The effect of age on variability in gait. Journal of Gerontology 39, 662-666.

Garcia, M., Chatterjee, A., Ruina, A., Coleman, M. 1998. The simplest walking model: stability, complexity, and scaling. Journal of Biomechanical Engineering 120, 281-288.

Gates, D. H., Dingwell, J. B. 2007. Peripheral neuropathy does not alter the fractal dynamics of stride intervals of gait. Journal of Applied Physiology 102, 965-971.

Gates, D. H., Su, J. L., Dingwell, J. B. 2007. Possible biomechanical origins of the long-range correlations in stride intervals of walking. Physica A: Statistical Mechanics and its Applications 380, 259-270.

Gera, G., Freitas, S., Latash, M., Monahan, K., Schoner, G., Scholz, J. 2010. Motor abundance contributes to resolving multiple kinematic task constraints. Motor Control 14, 83-115.

Gillespie, L. D., Robertson, M. C., Gillespie, W. J., Lamb, S. E., Gates, S., Cumming, R. G., Rowe, B. H. 2009. Interventions for preventing falls in older people living in the community. Cochrane Database of Systematic Reviews, CD007146.

Goldberg, E. J., Neptune, R. R. 2007. Compensatory strategies during normal walking in response to muscle weakness and increased hip joint stiffness. Gait and Posture 25, 360-367.

Goldberg, S. R., Ounpuu, S., Delp, S. L. 2003. The importance of swing-phase initial conditions in stiff-knee gait. Journal of Biomechanics 36, 1111-1116.

Goldberger, A. L., Amaral, L. A. N., Hausdorff, J. M., Ivanov, P. C., Peng, C. K., Stanley, H. E. 2002. Fractal dynamics in physiology: alterations with disease and aging. Proceedings of the National Academy of Sciences of the United States of America 99 Suppl 1, 2466-2472.

Gomes, M., Ruina, A. 2011. Walking model with no energy cost. Physical Review. E, Statistical, Nonlinear, and Soft Matter Physics 83, 032901.

Gorniak, S. L., Zatsiorsky, V. M., Latash, M. L. 2007. Hierarchies of synergies: an example of two-hand, multi-finger tasks. Experimental Brain Research 179, 167-180.

Graci, V., Elliott, D. B., Buckley, J. G. 2009. Peripheral visual cues affect minimum-foot-clearance during overground locomotion. Gait and Posture 30, 370-374.

Granata, K. P., Lockhart, T. E. 2007. Dynamic stability differences in fall-prone and healthy adults. Journal of Electromyography and Kinesiology

Grasso, R., Ivanenko, Y. P., Zago, M., Molinari, M., Scivoletto, G., Castellano, V., Macellari, V., Lacquaniti, F. 2004. Distributed plasticity of locomotor pattern generators in spinal cord injured patients. Brain 127, 1019-1034.

Grillner, S. 2003. The motor infrastructure: from ion channels to neuronal networks. Nature Reviews. Neuroscience 4, 573-586.

Grimmer, S., Ernst, M., Gunther, M., Blickhan, R. 2008. Running on uneven ground: leg adjustment to vertical steps and self-stability. The Journal of experimental biology 211, 2989-3000.

Grood, E. S., Suntay, W. J. 1983. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. Journal of Biomechanical Engineering 105, 136-144.

Guigon, E. 2010. Active control of bias for the control of posture and movement. Journal of Neurophysiology 104, 1090-1102.

Haken, H., Kelso, J. A., Bunz, H. 1985. A theoretical model of phase transitions in human hand movements. Biological Cybernetics 51, 347-356.

Hamilton, A. F. d. C., Jones, K. E., Wolpert, D. M. 2004. The scaling of motor noise with muscle strength and motor unit number in humans. Experimental Brain Research 157, 417-430.

Hamilton, A. F. d. C., Wolpert, D. M. 2002. Controlling the statistics of action: obstacle avoidance. Journal of Neurophysiology 87, 2434-2440.

Harris, C. M., Wolpert, D. M. 1998. Signal-dependent noise determines motor planning. Nature 394, 780-784.

Hart, S., Gabbard, C. 1998. Examining the mobilizing feature of footedness. Perceptual and Motor Skills 86, 1339-1342.

Haruno, M., Wolpert, D. M. 2005. Optimal control of redundant muscles in step-tracking wrist movements. Journal of Neurophysiology 94, 4244-4255.

Hausdorff, J. M. 2005. Gait variability: methods, modeling and meaning. Journal of Neuroengineering and Rehabilitation 2, 19-19.

Hausdorff, J. M. 2007. Gait dynamics, fractals and falls: Finding meaning in the stride-to-stride fluctuations of human walking. Human Movement Science

Hausdorff, J. M. 2009. Gait dynamics in Parkinson's disease: common and distinct behavior among stride length, gait variability, and fractal-like scaling. Chaos 19, 026113.

Hausdorff, J. M., Cudkowicz, M. E., Firtion, R., Wei, J. Y., Goldberger, A. L. 1998. Gait variability and basal ganglia disorders: stride-to-stride variations of gait cycle timing in Parkinson's disease and Huntington's disease. Movement Disorders 13, 428-437.

Hausdorff, J. M., Edelberg, H. K., Mitchell, S. L., Goldberger, A. L., Wei, J. Y. 1997a. Increased gait unsteadiness in community-dwelling elderly fallers. Archives of Physical Medicine and Rehabilitation 78, 278-283.

Hausdorff, J. M., Lertratanakul, A., Cudkowicz, M. E., Peterson, A. L., Kaliton, D., Goldberger, A. L. 2000. Dynamic markers of altered gait rhythm in amyotrophic lateral sclerosis. Journal of Applied Physiology 88, 2045-2053.

Hausdorff, J. M., Mitchell, S. L., Firtion, R., Peng, C. K., Cudkowicz, M. E., Wei, J. Y., Goldberger, A. L. 1997b. Altered fractal dynamics of gait: reduced stride-interval correlations with aging and Huntington's disease. Journal of Applied Physiology 82, 262-269.

Hausdorff, J. M., Peng, C. K., Ladin, Z., Wei, J. Y., Goldberger, A. L. 1995. Is walking a random walk? Evidence for long-range correlations in stride interval of human gait. Journal of Applied Physiology 78, 349-358.

Hausdorff, J. M., Purdon, P. L., Peng, C. K., Ladin, Z., Wei, J. Y., Goldberger, A. L. 1996. Fractal dynamics of human gait: stability of long-range correlations in stride interval fluctuations. Journal of Applied Physiology 80, 1448-1457.

Hausdorff, J. M., Rios, D. A., Edelberg, H. K. 2001. Gait variability and fall risk in community-living older adults: A 1-year prospective study. Archives of Physical Medicine and Rehabilitation 82, 1050-1056.

Hausdorff, J. M., Yogev, G., Springer, S., Simon, E. S., Giladi, N. 2005. Walking is more like catching than tapping: gait in the elderly as a complex cognitive task. Experimental Brain Research 164, 541-548.

Hausdorff, J. M., Zemany, L., Peng, C., Goldberger, A. L. 1999. Maturation of gait dynamics: stride-to-stride variability and its temporal organization in children. Journal of Applied Physiology 86, 1040-1047.

Herman, T., Giladi, N., Gurevich, T., Hausdorff, J. M. 2005. Gait instability and fractal dynamics of older adults with a "cautious" gait: why do certain older adults walk fearfully? Gait and Posture 21, 178-185.

Herman, T., Mirelman, A., Giladi, N., Schweiger, A., Hausdorff, J. M. 2010. Executive control deficits as a prodrome to falls in healthy older adults: a prospective study linking thinking, walking, and falling. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 65, 1086-1092.

Herr, H., Popovic, M. 2008. Angular momentum in human walking. Journal of Experimental Biology 211, 467-481.

Heylighen, F. (Ed.) 2008. Complexity and Self-Organization. Taylor & Francis

Heylighen, F. 2010. The Self-Organization of Time and Causality: Steps Towards Understanding the Ultimate Origin. Foundations of Science 15, 345-356.

Hill, K., Schwarz, J., Flicker, L., Carroll, S. 1999. Falls among healthy, community-dwelling, older women: a prospective study of frequency, circumstances, consequences and prediction accuracy. Australian and New Zealand Journal of Public Health 23, 41-48.

Hill, K. D., Schwarz, J. A., Kalogeropoulos, A. J., Gibson, S. J. 1996. Fear of falling revisited. Archives of Physical Medicine and Rehabilitation 77, 1025-1029.

Hodkinson, H. M. 1972. Evaluation of a mental test score for assessment of mental impairment in the elderly. Age and Ageing 1, 233-238.

Hof, A. L. 2007. The equations of motion for a standing human reveal three mechanisms for balance. Journal of Biomechanics 40, 451-457.

Hofmann, A., Popovic, M. B., Herr, H. 2009. Exploiting Angular Momentum to Enhance Bipedal Centre-of-Mass Control. In: IEEE International Conference on Robotics and Automation, Kobe, Japan

Holden, J. P., Orsini, J. A., Siegel, K. L., Kepple, T. M., Gerber, L. H., Stanhope, S. J. 1997. Surface movement errors in shank kinematics and knee kinetics during gait. Gait and Posture 5, 217-227.

Houck, J., Yack, H. J., Cuddeford, T. 2004. Validity and comparisons of tibiofemoral orientations and displacement using a femoral tracking device during early to mid stance of walking. Gait and Posture 19, 76-84.

Howard, I. S., Ingram, J. N., Wolpert, D. M. 2008. Composition and decomposition in bimanual dynamic learning. The Journal of neuroscience : the official journal of the Society for Neuroscience 28, 10531-10540.

Hsu, W. L., Scholz, J. P., Schoner, G., Jeka, J. J., Kiemel, T. 2007. Control and estimation of posture during quiet stance depends on multijoint coordination. Journal of Neurophysiology 97, 3024-3035.

Hurmuzlu, Y., Basdogan, C., Carollo, J. J. 1994. Presenting joint kinematics of human locomotion using phase plane portraits and Poincare maps. Journal of Biomechanics 27, 1495-1499.

Hurst, H. E. 1951. Long Term Storage Capacity of Reservoirs. Transactions of the American Society of Civil Engineers 116, 770-799.

Iida, F., Tedrake, R. 2010. Minimalistic control of biped walking in rough terrain. Autonomous Robots 28, 355-368.

Ijspeert, A. J. 2008. Central pattern generators for locomotion control in animals and robots: a review. Neural networks : the official journal of the International Neural Network Society 21, 642-653.

Ivanenko, Y. P., Cappellini, G., Dominici, N., Poppele, R. E., Lacquaniti, F. 2005. Coordination of locomotion with voluntary movements in humans. Journal of Neuroscience 25, 7238-7253.

Ivanenko, Y. P., Cappellini, G., Dominici, N., Poppele, R. E., Lacquaniti, F. 2007. Modular control of limb movements during human locomotion. Journal of Neuroscience 27, 11149-11161.

Ivanenko, Y. P., Dominici, N., Cappellini, G., Lacquaniti, F. 2005. Kinematics in newly walking toddlers does not depend upon postural stability. Journal of Neurophysiology 94, 754-763.

Ivanenko, Y. P., Grasso, R., Macellari, V., Lacquaniti, F. 2002. Control of foot trajectory in human locomotion: role of ground contact forces in simulated reduced gravity. Journal of Neurophysiology 87, 3070-3089.

Ivanenko, Y. P., Grasso, R., Zago, M., Molinari, M., Scivoletto, G., Castellano, V., Macellari, V., Lacquaniti, F. 2003. Temporal components of the motor patterns expressed by the human spinal cord reflect foot kinematics. Journal of Neurophysiology 90, 3555-3565.

Ivanenko, Y. P., Poppele, R. E., Lacquaniti, F. 2004. Five basic muscle activation patterns account for muscle activity during human locomotion. Journal of Physiology 556, 267-282.

Ivanenko, Y. P., Poppele, R. E., Lacquaniti, F. 2006. Motor control programs and walking. Neuroscientist 12, 339-348.

Jirsa, V. K., Kelso, J. A. S. 2004. Coordination Dynamics: Issues and Trends. Springer-Verlag, Heidelberg.

Johansson, R. S., Flanagan, J. R. 2009. Coding and use of tactile signals from the fingertips in object manipulation tasks. Nature Reviews. Neuroscience 10, 345-359.

Jones, K. E., Hamilton, A. F., Wolpert, D. M. 2002. Sources of signal-dependent noise during isometric force production. Journal of Neurophysiology 88, 1533-1544.

Jordan, K., Challis, J. H., Cusumano, J. P., Newell, K. M. 2009. Stability and the time-dependent structure of gait variability in walking and running. Human Movement Science 28, 113-128.

Jordan, K., Challis, J. H., Newell, K. M. 2006. Long range correlations in the stride interval of running. Gait and Posture 24, 120-125.

Jordan, K., Challis, J. H., Newell, K. M. 2007. Walking speed influences on gait cycle variability. Gait and Posture 26, 128-134.

Kang, H. G., Dingwell, J. B. 2006. Intra-session reliability of local dynamic stability of walking. Gait and Posture 24, 386-390.

Kang, H. G., Dingwell, J. B. 2007. SPEED CONTROLLED COMPARISON OF GAIT VARIABILITY IN HEALTHY YOUNG AND OLDER ADULTS. Journal of Biomechanics 40, S246.

Kang, H. G., Dingwell, J. B. 2008a. Effects of walking speed, strength and range of motion on gait stability in healthy older adults. Journal of Biomechanics 41, 2899-2905.

Kang, H. G., Dingwell, J. B. 2008b. Separating the effects of age and walking speed on gait variability. Gait and Posture 27, 572-577.

Kantelhardt, J. W., Koscielny-Bunde, E., Rego, H. H. A., Havlin, S., Bunde, A. 2001. Detecting long-range correlations with detrended fluctuation analysis. Physica A: Statistical Mechanics and its Applications 295, 441-454.

Karst, G. M., Hageman, P. A., Jones, T. F., Bunner, S. H. 1999. Reliability of foot trajectory measures within and between testing sessions. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 54, 343-347.

Kaya, B. K., Krebs, D. E., Riley, P. O. 1998. Dynamic stability in elders: momentum control in locomotor ADL. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 53, M126-134.

Kelso, J. A. S. 1995. Dynamic Patterns: The Self Organisation of Brain and Behaviour. M.I.T. Press, Cambridge

Kelso, J. A. S. 2004. The Problem of Coordination. In: Jirsa, V. K., Kelso, J. A. S. (Eds.), Coordination Dynamics: Trends and Issues. Springer-Verlag, Heidelberg

Kepple, T. M., Siegel, K. L., Stanhope, S. J. 1997. Relative contributions of the lower extremity joint moments to forward progression and support during gait. Gait and Posture 6, 1-8.

Kerrigan, D. C., Lee, L. W., Collins, J. J., Riley, P. O., Lipsitz, L. A. 2001. Reduced hip extension during walking: Healthy elderly and fallers versus young adults. Archives of Physical Medicine and Rehabilitation 82, 26-30.

Kerrigan, D. C., Riley, P. O., Lelas, J. L., Croce, U. D. 2001. Quantification of pelvic rotation as a determinant of gait. Archives of Physical Medicine and Rehabilitation 82, 217-220.

Kerrigan, D. C., Xenopoulos-Oddsson, A., Sullivan, M. J., Lelas, J. J., Riley, P. O. 2003. Effect of a hip flexor[ndash]stretching program on gait in the elderly. Archives of Physical Medicine and Rehabilitation 84, 1-6.

Khandoker, A. H., Taylor, S. B., Karmakar, C. K., Begg, R. K., Palaniswami, M. 2008. Investigating Scale Invariant Dynamics in Minimum Toe Clearance Variability of the Young and Elderly During Treadmill Walking. IEEE Transactions on Neural Systems and Rehabilitation Engineering 16, 380-389.

Kording, K. P., Ku, S.-p., Wolpert, D. M. 2004. Bayesian integration in force estimation. Journal of Neurophysiology 92, 3161-3165.

Kording, K. P., Wolpert, D. M. 2004. Bayesian integration in sensorimotor learning. Nature 427, 244-247.

Kram, R., Domingo, A., Ferris, D. P. 1997. Effect of reduced gravity on the preferred walk-run transition speed. Journal of Experimental Biology 200, 821-826.

Kreisfeld, R., Harrison, J. E. 2010. Hospital separations due to injury and poisoning 2005-06. In: Injury research and statistics series no. 55. AIHW, Canberra

Kuo, A. D. 2002a. Energetics of actively powered locomotion using the simplest walking model. Journal of Biomechanical Engineering 124, 113-120.

Kuo, A. D. 2002b. The relative roles of feedforward and feedback in the control of rhythmic movements. Motor Control 6, 129-145.

Kuo, A. D. 2007. The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective. Human Movement Science

Kuo, A. D., Donelan, J. M. 2010. Dynamic principles of gait and their clinical implications. Physical Therapy 90, 157-174.

Kurtzer, I., Pruszynski, J. A., Scott, S. H. 2009. Long-latency responses during reaching account for the mechanical interaction between the shoulder and elbow joints. Journal of Neurophysiology 102, 3004-3015.

Kurtzer, I. L., Pruszynski, J. A., Scott, S. H. 2008. Long-latency reflexes of the human arm reflect an internal model of limb dynamics. Current Biology 18, 449-453.

Lacquaniti, F., Ivanenko, Y. P., Zago, M. 2002. Kinematic control of walking. Archives Italiennes de Biologie 140, 263-272.

Lafortune, M. A., Cavanagh, P. R., Sommer, H. J., Kalenak, A. 1992. Three-dimensional kinematics of the human knee during walking. Journal of Biomechanics 25, 347-357.

Lam, T., Dietz, V. 2004. Transfer of motor performance in an obstacle avoidance task to different walking conditions. Journal of Neurophysiology 92, 2010-2016.

Latash, M. L. 2008. Synergy. Oxford, New York.

Latash, M. L. 2010a. Motor synergies and the equilibrium-point hypothesis. Motor Control 14, 294-322.

Latash, M. L. 2010b. Two Archetypes of Motor Control Research. Motor Control 14, e41-e53.

Latash, M. L., Gorniak, S., Zatsiorsky, V. M. 2008. Hierarchies of Synergies in Human Movements. Kinesiology (Zagreb) 40, 29-38.

Latash, M. L., Scholz, J. F., Danion, F., Schoner, G. 2001. Structure of motor variability in marginally redundant multifinger force production tasks. Experimental Brain Research 141, 153-165.

Latt, M., Menz, H., Fung, V., Lord, S. 2007. Walking speed, cadence and step length are selected to optimize the stability of head and pelvis accelerations. Experimental Brain Research

Latt, M. D., Menz, H. B., Fung, V. S., Lord, S. R. 2008. Walking speed, cadence and step length are selected to optimize the stability of head and pelvis accelerations. Experimental Brain Research 184, 201-209.

Leardini, A., Cappozzo, A., Catani, F., Toksvig-Larsen, S., Petitto, A., Sforza, V., Cassanelli, G., Giannini, S. 1999. Validation of a functional method for the estimation of hip joint centre location. Journal of Biomechanics 32, 99-103.

Leardini, A., Chiari, L., Croce, U. D., Cappozzo, A. 2005. Human movement analysis using stereophotogrammetry: Part 3. Soft tissue artifact assessment and compensation. Gait and Posture 21, 212-225.

Lee, L. W., Zavarei, K., Evans, J., Lelas, J. J., Riley, P. O., Kerrigan, D. C. 2005. Reduced Hip Extension in the Elderly: Dynamic or Postural? Archives of Physical Medicine and Rehabilitation 86, 1851-1854.

Li, W., Todorov, E., Pan, X. 2004. Hierarchical optimal control of redundant biomechanical systems. Conference Proceedings, Annual International Conference of the IEEE Engineering in Medicine and Biology Society 6, 4618-4621.

Li, W., Todorov, E., Pan, X. 2005. Hierarchical Feedback and Learning for Multi-joint Arm Movement Control. Conference proceedings : ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Conference 4, 4400-4403.

Lin, Y. C., Walter, J. P., Banks, S. A., Pandy, M. G., Fregly, B. J. Simultaneous prediction of muscle and contact forces in the knee during gait. Journal of Biomechanics 43, 945-952.

Liu, D., Todorov, E. 2007. Evidence for the flexible sensorimotor strategies predicted by optimal feedback control. Journal of Neuroscience 27, 9354-9368.

Liu, D., Todorov, E. 2009. Hierarchical optimal control of a 7-DOF arm model. In: 2nd IEEE Symposium on Adaptive Dynamic Programming and Reinforcement Learning, pp. pp 50 - 57

Lockhart, D. B., Ting, L. H. 2007. Optimal sensorimotor transformations for balance. Nature Neuroscience 10, 1329-1336.

Loeb, G. E., Brown, I. E., Cheng, E. J. 1999. A hierarchical foundation for models of sensorimotor control. Experimental Brain Research 126, 1-18.

Lord, S. R. 1993. Hip fractures: changing patterns in hospital bed use in NSW between 1979 and 1990. Australian and New Zealand Journal of Surgery 63, 352-355.

Lord, S. R. 1994. Predictors of nursing home placement and mortality of residents in intermediate care. Age and Ageing 23, 499-504.

Lord, S. R., Sherrington, C., Menz, H., Close, J. C. T. 2007. Falls in Older People: Risk factors and strategies for prevention. Cambridge University Press, Cambridge.

Lord, S. R., Ward, J. A., Williams, P., Anstey, K. J. 1993. An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study. Australian Journal of Public Health 17, 240-245.

Lowrey, C. R., Watson, A., Vallis, L. A. 2007. Age-related changes in avoidance strategies when negotiating single and multiple obstacles. Experimental Brain Research 182, 289-299.

Lu, T. W., O'Connor, J. J. 1999. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. Journal of Biomechanics 32, 129-134.

Luukinen, H., Koski, K., Hiltunen, L., Kivela, S. L. 1994. Incidence rate of falls in an aged population in northern Finland. Journal of Clinical Epidemiology 47, 843-850.

MacLellan, M. J., Patla, A. E. 2006. Adaptations of walking pattern on a compliant surface to regulate dynamic stability. Experimental Brain Research 173, 521-530.

Magaziner, J., Hawkes, W., Hebel, J. R., Zimmerman, S. I., Fox, K. M., Dolan, M., Felsenthal, G., Kenzora, J. 2000. Recovery from hip fracture in eight areas of function. The journals of gerontology. Series A, Biological sciences and medical sciences 55, M498-507.

Maki, B. E. 1997. Gait changes in older adults: predictors of falls or indicators of fear. Journal of the American Geriatrics Society 45, 313-320.

Maki, B. E., McIlroy, W. E., Fernie, G. R. 2003. Change-in-support reactions for balance recovery. IEEE Engineering in Medicine and Biology Magazine 22, 20-26.

Manal, K., McClay, I., Richards, J., Galinat, B., Stanhope, S. 2002. Knee moment profiles during walking: errors due to soft tissue movement of the shank and the influence of the reference coordinate system. Gait and Posture 15, 10-17.

Manal, K., McClay, I., Stanhope, S., Richards, J., Galinat, B. 2000. Comparison of surface mounted markers and attachment methods in estimating tibial rotations during walking: an in vivo study. Gait and Posture 11, 38-45.

Manchester, I. R., Mettin, U., Iida, F., Tedrake, R. 2011. Stable dynamic walking over uneven terrain. International Journal of Robotics Research 30, 265-279.

Manor, B., Costa, M. D., Hu, K., Newton, E., Starobinets, O., Kang, H. G., Peng, C. K., Novak, V., Lipsitz, L. A. 2010. Physiological complexity and system adaptability: evidence from postural control dynamics of older adults. Journal of Applied Physiology 109, 1786-1791.

Marottoli, R. A., Berkman, L. F., Cooney, L. M. 1992. Decline in physical function following hip fracture. Journal of the American Geriatrics Society 40, 861-866.

McAndrew, P. M., Dingwell, J. B., Wilken, J. M. 2010. Walking variability during continuous pseudo-random oscillations of the support surface and visual field. Journal of Biomechanics 43, 1470-1475.

McGeer, T. 1990. Passive Dynamic Walking. International Journal of Robotics Research 9, 62-82.

McGeer, T. 1993. Dynamics and control of bipedal locomotion. Journal of Theoretical Biology 163, 277-314.

McGibbon, C. A., Krebs, D. E. 2001. Age-related changes in lower trunk coordination and energy transfer during gait. Journal of Neurophysiology 85, 1923-1931.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., Lord, S. R. 2008. Optimizing footwear for older people at risk of falls. Journal of Rehabilitation Research and Development 45, 1167-1181.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., Lord, S. R. 2009. Effects of walking surfaces and footwear on temporo-spatial gait parameters in young and older people. Gait and Posture 29, 392-397.

Menz, H. B., Lord, S. R., Fitzpatrick, R. C. 2003. Acceleration patterns of the head and pelvis when walking are associated with risk of falling in community-dwelling older people. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 58, M446-452.

Menz, H. B., Lord, S. R., Fitzpatrick, R. C. 2007. A structural equation model relating impaired sensorimotor function, fear of falling and gait patterns in older people. Gait and Posture 25, 243-249.

Menz, H. B., Morris, M. E., Lord, S. R. 2006. Footwear characteristics and risk of indoor and outdoor falls in older people. Gerontology 52, 174-180.

Miller, C. A., Feiveson, A. H., Bloomberg, J. J. 2009. Effects of speed and visual-target distance on toe trajectory during the swing phase of treadmill walking. Journal of Applied Biomechanics 25, 32-42.

Mills, P. M., Barrett, R. S. 2001. Swing phase mechanics of healthy young and elderly men. Human Movement Science 20, 427-446.

Mills, P. M., Barrett, R. S., Morrison, S. 2007. Toe clearance variability during walking in young and elderly men. Gait and Posture

Moe-Nilssen, R., Aaslund, M. K., Hodt-Billington, C., Helbostad, J. L. 2010. Gait variability measures may represent different constructs. Gait and Posture 32, 98-101.

Moe-Nilssen, R., Helbostad, J. L. 2005. Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. Gait and Posture 21, 164-170.

Moosabhoy, M. A., Gard, S. A. 2006. Methodology for determining the sensitivity of swing leg toe clearance and leg length to swing leg joint angles during gait. Gait and Posture 24, 493-501.

Muller, H., Sternad, D. 2004. Decomposition of variability in the execution of goal-oriented tasks: three components of skill improvement. Journal of Experimental Psychology: Human Perception and Performance 30, 212-233.

Nagano, H., Begg, R. K., Sparrow, W. A., Taylor, S. 2011. Ageing and limb dominance effects on foot-ground clearance during treadmill and overground walking. Clinical Biomechanics 26, 962-968.

Nagengast, A. J., Braun, D. A., Wolpert, D. M. 2010. Risk-sensitive optimal feedback control accounts for sensorimotor behavior under uncertainty. PLoS Computational Biology 6, e1000857.

Nash, J. F. 1950. Equilibrium Points in N-Person Games. . Proceedings of the National Academy of Sciences of the United States of America, 48-49.

Neptune, R. R., Clark, D. J., Kautz, S. A. 2009a. Modular control of human walking: a simulation study. Journal of Biomechanics 42, 1282-1287.

Neptune, R. R., McGowan, C. P. 2011. Muscle contributions to whole-body sagittal plane angular momentum during walking. Journal of Biomechanics 44, 6-12.

Neptune, R. R., McGowan, C. P., Kautz, S. A. 2009b. Forward dynamics simulations provide insight into muscle mechanical work during human locomotion. Exercise and Sport Sciences Reviews 37, 203-210.

Neptune, R. R., Sasaki, K., Kautz, S. A. 2008. The effect of walking speed on muscle function and mechanical energetics. Gait and Posture 28, 135-143.

Neptune, R. R., Zajac, F. E., Kautz, S. A. 2004. Muscle mechanical work requirements during normal walking: the energetic cost of raising the body's center-of-mass is significant. Journal of Biomechanics 37, 817-825.

Neptune, R. R., Zajac, F. E., Kautz, S. A. 2009c. Author's Response to Comment on "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking" (Neptune et al., 2001) and "Muscle mechanical work requirements during normal walking: The energetic cost of raising the body's center-of-mass is significant" (). Journal of Biomechanics 42, 1786-1789.

Newell, K. M., Deutsch, K. M., Sosnoff, J. J., Mayer-Kress, G. 2006. Variability in motor output as noise: A default and erroneous proposition? In: Davids, K., Bennett, S., Newell, K. M. (Eds.), Movement System Variability. Human Kinetics, pp. 3-22.

Newell, K. M., Mayer-Kress, G., Liu, Y. T. 2009. Aging, time scales, and sensorimotor variability. Psychology and Aging 24, 809-818.

Nikolaus, T., Bach, M. 2003. Preventing falls in community-dwelling frail older people using a home intervention team (HIT): results from the randomized Falls-HIT trial. Journal of the American Geriatrics Society 51, 300-305.

O'Connor, C. M., Thorpe, S. K., O'Malley, M. J., Vaughan, C. L. 2007. Automatic detection of gait events using kinematic data. Gait and Posture 25, 469-474.

O'Loughlin, J. L., Robitaille, Y., Boivin, J. F., Suissa, S. 1993. Incidence of and risk factors for falls and injurious falls among the community-dwelling elderly. American Journal of Epidemiology 137, 342-354.

O'Sullivan, I., Burdet, E., Diedrichsen, J. 2009. Dissociating variability and effort as determinants of coordination. PLoS Computational Biology 5, e1000345.

Orban, G., Wolpert, D. M. 2011. Representations of uncertainty in sensorimotor control. Current Opinion in Neurobiology 21, 629-635.

Osaki, Y., Kunin, M., Cohen, B., Raphan, T. 2007. Three-dimensional kinematics and dynamics of the foot during walking: a model of central control mechanisms. Experimental Brain Research 176, 476-496.

Owings, T. M., Grabiner, M. D. 2004. Variability of step kinematics in young and older adults. Gait and Posture 20, 26-29.

Pandy, M. G. 2001. Computer modeling and simulation of human movement. Annual Review of Biomedical Engineering 3, 245-273.

Pandy, M. G., Lin, Y. C., Kim, H. J. 2010. Muscle coordination of mediolateral balance in normal walking. Journal of Biomechanics 43, 2055-2064.

Patla, A. E. 2003. Strategies for dynamic stability during adaptive human locomotion. IEEE Engineering in Medicine and Biology Magazine 22, 48-52.

Pavol, M. J., Owings, T. M., Foley, K. T., Grabiner, M. D. 1999. Gait characteristics as risk factors for falling from trips induced in older adults. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 54, 583-590.

Pavol, M. J., Owings, T. M., Foley, K. T., Grabiner, M. D. 2001. Mechanisms leading to a fall from an induced trip in healthy older adults. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 56, 428-437.

Pavol, M. J., Pai, Y.-C. 2007. Deficient limb support is a major contributor to age differences in falling. Journal of Biomechanics 40, 1318-1325.

Peng, C. K., Buldyrev, S. V., Hausdorff, J. M., Havlin, S., Mietus, J. E., Simons, M., Stanley, H. E., Goldberger, A. L. 1994. Non-equilibrium dynamics as an indispensable characteristic of a healthy biological system. Integrative Physiological and Behavioral Science 29, 283-293.

Peng, C. K., Hausdorff, J. M., Havlin, S., Mietus, J. E., Stanley, H. E., Goldberger, A. L. 1998. Multiple-time scales analysis of physiological time series under neural control. Physica A: Statistical and Theoretical Physics 249, 491-500.

Peng, C. K., Havlin, S., Hausdorff, J. M., Mietus, J. E., Stanley, H. E., Goldberger, A. L. 1995. Fractal mechanisms and heart rate dynamics. Long-range correlations and their breakdown with disease. Journal of Electrocardiology 28 Suppl, 59-65.

Pfeifer, R., Lungarella, M., Fumiya, I. 2007. Self-Organization, Embodiment, and Biologiaclly Inspired Robotics. Science 318, 6.

Pham, Q. C., Hicheur, H., Arechavaleta, G., Laumond, J. P., Berthoz, A. 2007. The formation of trajectories during goal-oriented locomotion in humans. II. A maximum smoothness model. European Journal of Neuroscience 26, 2391-2403.

Piazza, S. J., Delp, S. L. 1996. The influence of muscles on knee flexion during the swing phase of gait. Journal of Biomechanics 29, 723-733.

Pijnappels, M., Bobbert, M. F., van Dieen, J. H. 2001. Changes in walking pattern caused by the possibility of a tripping reaction. Gait and Posture 14, 11-18.

Pijnappels, M., Bobbert, M. F., van Dieen, J. H. 2005. Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait and Posture 21, 388-394.

Pijnappels, M., Reeves, N. D., Maganaris, C. N., van Dieen, J. H. 2007. Tripping without falling; lower limb strength, a limitation for balance recovery and a target for training in the elderly. Journal of Electromyography and Kinesiology

Popovic, M. B., Hofmann, A., Herr, H. 2004. Angular Momentum Regulation during Human Walking: Biomechanics and Control. In: Proceedings of the IEEE International Conference on Robotics and Automation, New Orleans, LA, USA, pp. pp. 2405-2411.

Potter-Forbes, M., Aisbett, C. 2003. Injury Costs! A Valuation of the Burden of Injury in New South Wales in 1998-1999. In. NSW Injury Risk Management Research Centre, University of New South Wales, Sydney

Prince, F., Corriveau, H., Hebert, R., Winter, D. A. 1997. Gait in the elderly. Gait and Posture 5, 128-135.

Pruszynski, J. A., Kurtzer, I., Scott, S. H. 2008. Rapid motor responses are appropriately tuned to the metrics of a visuospatial task. Journal of Neurophysiology 100, 224-238.

Quevedo, J., Stecina, K., Gosgnach, S., McCrea, D. A. 2005. Stumbling corrective reaction during fictive locomotion in the cat. Journal of Neurophysiology 94, 2045-2052.

Quevedo, J., Stecina, K., McCrea, D. A. 2005. Intracellular analysis of reflex pathways underlying the stumbling corrective reaction during fictive locomotion in the cat. Journal of Neurophysiology 94, 2053-2062.

Raz, N., Lindenberger, U., Rodrigue, K. M., Kennedy, K. M., Head, D., Williamson, A., Dahle, C., Gerstorf, D., Acker, J. D. 2005. Regional brain changes in aging healthy adults: general trends, individual differences and modifiers. Cerebral Cortex 15, 1676-1689.

Redfern, M. S., Schumann, T. 1994. A model of foot placement during gait. Journal of Biomechanics 27, 1339-1346.

Reinbolt, J. A., Fox, M. D., Arnold, A. S., Ounpuu, S., Delp, S. L. 2008. Importance of preswing rectus femoris activity in stiff-knee gait. Journal of Biomechanics 41, 2362-2369.

Reinschmidt, C., van Den Bogert, A., Murphy, N., Lundberg, A., Nigg, B. 1997. Tibiocalcaneal motion during running, measured with external and bone markers. Clinical Biomechanics 12, 8-16.

Rice, D. P., Mackenzie, E. J. 1989. Cost of injury in the United States: A report to congress. In. Institute for Health and Aging, University of California, San Francisco

Riggs, B. L., Melton, L. J., 3rd, Robb, R. A., Camp, J. J., Atkinson, E. J., Oberg, A. L., Rouleau, P. A., McCollough, C. H., Khosla, S., Bouxsein, M. L. 2006. Population-based analysis of the relationship of whole bone strength indices and fall-related loads to age- and sex-specific patterns of hip and wrist fractures. Journal of bone and mineral research : the official journal of the American Society for Bone and Mineral Research 21, 315-323.

Robert, T., Zatsiorsky, V. M., Latash, M. L. 2008. Multi-muscle synergies in an unusual postural task: quick shear force production. Experimental Brain Research

Roos, M. R., Rice, C. L., Vandervoort, A. A. 1997. Age-related changes in motor unit function. Muscle and Nerve 20, 679-690.

Rosano, C., Aizenstein, H., Brach, J., Longenberger, A., Studenski, S., Newman, A. B. 2008. Special article: gait measures indicate underlying focal gray matter atrophy in the brain of older adults. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 63, 1380-1388.

Rossignol, S. 2000. Locomotion and its recovery after spinal injury. Current Opinion in Neurobiology 10, 708-716.

Rossignol, S., Dubuc, R., Gossard, J. P. 2006. Dynamic sensorimotor interactions in locomotion. Physiological Reviews 86, 89-154.

Roudsari, B. S., Ebel, B. E., Corso, P. S., Molinari, N.-A. M., Koepsell, T. D. 2005. The acute medical care costs of fall-related injuries among the U.S. older adults. Injury 36, 1316-1322.

Ruina, A., Bertram, J. E., Srinivasan, M. 2005. A collisional model of the energetic cost of support work qualitatively explains leg sequencing in walking and galloping, pseudo-elastic leg behavior in running and the walk-to-run transition. Journal of Theoretical Biology 237, 170-192.

Sadeghi, H., Allard, P., Prince, F., Labelle, H. 2000. Symmetry and limb dominance in able-bodied gait: a review. Gait and Posture 12, 34-45.

Saha, D., Gard, S., Fatone, S. 2008. The effect of trunk flexion on able-bodied gait. Gait and Posture 27, 653-660.

Sasaki, K., Neptune, R. R., Kautz, S. A. 2009. The relationships between muscle, external, internal and joint mechanical work during normal walking. Journal of Experimental Biology 212, 738-744.

Scafetta, N., Marchi, D., West, B. J. 2009. Understanding the complexity of human gait dynamics. Chaos 19, 026108.

Schaal, S., Mohajerian, P., Ijspeert, A. 2007. Dynamics systems vs. optimal control--a unifying view. Progress in Brain Research 165, 425-445.

Schablowski, M., Gerner, H. J. 2010. Comparison of two measures of dynamic stability during treadmill walking.

Scholz, J. P., Schoner, G. 1999. The uncontrolled manifold concept: identifying control variables for a functional task. Experimental Brain Research 126, 289-306.

Schoner, G. 1995. Recent developments and problems in human movement science and their conceptual implications. Ecol Psych 8, 291-314.

Schoner, G., Haken, H., Kelso, J. A. 1986. A stochastic theory of phase transitions in human hand movement. Biological Cybernetics 53, 247-257.

Schoner, G., Scholz, J. P. 2007. Analyzing variance in multi-degree-of-freedom movements: uncovering structure versus extracting correlations. Motor Control 11, 259-275.

Schulz, B. W., Lloyd, J. D., Lee, W. E., 3rd 2010. The effects of everyday concurrent tasks on overground minimum toe clearance and gait parameters. Gait and Posture 32, 18-22.

Schwartz, M. H., Rozumalski, A. 2005. A new method for estimating joint parameters from motion data. Journal of Biomechanics 38, 107-116.

Schwartz, M. H., Trost, J. P., Wervey, R. A. 2004. Measurement and management of errors in quantitative gait data. Gait and Posture 20, 196-203.

Scott, S. H. 2004. Optimal feedback control and the neural basis of volitional motor control. Nature Reviews. Neuroscience 5, 532-546.

Seay, J. F., Haddad, J. M., van Emmerik, R. E. A., Hamill, J. 2006. Coordination variability around the walk to run transition during human locomotion. Motor Control 10, 178-196.

Seidler, R. D., Bernard, J. A., Burutolu, T. B., Fling, B. W., Gordon, M. T., Gwin, J. T., Kwak, Y., Lipps, D. B. 2010. Motor control and aging: links to age-related brain structural, functional, and biochemical effects. Neuroscience and Biobehavioral Reviews 34, 721-733.

Selen, L. P., Franklin, D. W., Wolpert, D. M. 2009. Impedance control reduces instability that arises from motor noise. The Journal of neuroscience : the official journal of the Society for Neuroscience 29, 12606-12616.

Selles, R. W., Bussmann, J. B. J., Wagenaar, R. C., Stam, H. J. 2001. Comparing predictive validity of four ballistic swing phase models of human walking. Journal of Biomechanics 34, 1171-1177.

Shadmehr, R., Krakauer, J. W. 2008. A computational neuroanatomy for motor control. Experimental Brain Research 185, 359-381.

Shadmehr, R., Moussavi, Z. M. 2000. Spatial generalization from learning dynamics of reaching movements. The Journal of neuroscience : the official journal of the Society for Neuroscience 20, 7807-7815.

Shemmell, J., Johansson, J., Portra, V., Gottlieb, G. L., Thomas, J. S., Corcos, D. M. 2007. Control of interjoint coordination during the swing phase of normal gait at different speeds. Journal of Neuroengineering and Rehabilitation 4, 10-10.

Simoneau, G. G., Krebs, D. E. 2000. Whole body momentum during gait: a preliminary study of non-fallers and frequent fallers. . Journal of Applied Biomechanics 16, 1-13.

Sosnoff, J. J., Newell, K. M. 2007. Are visual feedback delays responsible for aging-related increases in force variability? Experimental Aging Research 33, 399-415.

Spirduso, W. W., Francis, K. L., MacRae, P. G. 2005. Physical Dimensions of Aging. Human Kinetics

Spoor, C. W., Veldpaus, F. E. 1980. Rigid body motion calculated from spatial co-ordinates of markers. Journal of Biomechanics 13, 391-393.

Srinivasan, M., Ruina, A. 2006. Computer optimization of a minimal biped model discovers walking and running. Nature 439, 72-75.

Stagni, R., Fantozzi, S., Cappello, A., Leardini, A. 2005. Quantification of soft tissue artefact in motion analysis by combining 3D fluoroscopy and stereophotogrammetry: a study on two subjects. Clinical Biomechanics 20, 320-329.

Stagni, R., Leardini, A., Cappozzo, A., Grazia Benedetti, M., Cappello, A. 2000. Effects of hip joint centre mislocation on gait analysis results. Journal of Biomechanics 33, 1479-1487.

Steele, K. M., Seth, A., Hicks, J. L., Schwartz, M. S., Delp, S. L. 2010. Muscle contributions to support and progression during single-limb stance in crouch gait. Journal of Biomechanics 43, 2099-2105.

Su, J. L., Dingwell, J. B. 2007. Dynamic stability of passive dynamic walking on an irregular surface. Journal of Biomechanical Engineering 129, 802-810.

Takakusaki, K., Okumura, T. 2008. Neurobiological Basis of Controlling Posture and Locomotion. Advanced Robotics 22, 1629-1663.

Tassa, Y., Erez, T., Todorov, E. 2011. Optimal limit-cycle control recast as Bayesian inference. In: World Congress of the Internationale Federation of Automatic Control

Tedrake, R. 2009. Underactuated robotics: learning, planning, and control for efficient and agile machines In: Course Notes 6.832. Massachusetts Institute of Technology

Temprado, J.-J. 2004. A Dynamical Approach to the Interplay of Attention and Bimanual Coordination. In: Jirsa, V. K., Kelso, J. A. S. (Eds.), Coordination Dynamics: Issues and Trends. Springer-Verlag, Heidelberg

Tinetti, M. E., Mendes de Leon, C. F., Doucette, J. T., Baker, D. I. 1994. Fear of falling and fall-related efficacy in relationship to functioning among community-living elders. Journal of Gerontology 49, 140-147.

Tinetti, M. E., Richman, D., Powell, L. 1990. Falls efficacy as a measure of fear of falling. Journal of Gerontology 45, 239-243.

Tinetti, M. E., Speechley, M., Ginter, S. F. 1988. Risk factors for falls among elderly persons living in the community. New England Journal of Medicine 319, 1701-1707.

Tinetti, M. E., Williams, C. S. 1997. Falls, injuries due to falls, and the risk of admission to a nursing home. New England Journal of Medicine 337, 1279-1284.

Ting, L. H., Macpherson, J. M. 2005. A limited set of muscle synergies for force control during a postural task. Journal of Neurophysiology 93, 609-613.

Todorov, E. 2004. Optimality principles in sensorimotor control. Nature Neuroscience 7, 907-915.

Todorov, E. 2005. Stochastic optimal control and estimation methods adapted to the noise characteristics of the sensorimotor system. Neural Computation 17, 1084-1108.

Todorov, E. 2008. General Duality between Optimal Control and Estimation. In: 47th IEEE Conference on Decision and Control, Cancun, Mexico, pp. 4286-4292.

Todorov, E. 2009. Efficient computation of optimal actions. Proceedings of the National Academy of Sciences of the United States of America 106, 11478-11483.

Todorov, E. 2011. Finding the most likely trajectories of optimally-controlled stochastic systems. In: World Congress of the International Federation of Automatic Control

Todorov, E., Jordan, M. I. 2002. Optimal feedback control as a theory of motor coordination. Nature Neuroscience 5, 1226-1235.

Torre, K., Delignieres, D. 2008. Unraveling the finding of 1/f beta noise in self-paced and synchronized tapping: a unifying mechanistic model. Biological Cybernetics 99, 159-170.

Torre, K., Wagenmakers, E. J. 2009. Theories and models for 1/f(beta) noise in human movement science. Human Movement Science 28, 297-318.

Tresch, M. C., Saltiel, P., d'Avella, A., Bizzi, E. 2002. Coordination and localization in spinal motor systems. Brain Research. Brain Research Reviews 40, 66-79.

Tseng, Y. W., Scholz, J. P., Valere, M. 2006. Effects of movement frequency and joint kinetics on the joint coordination underlying bimanual circle drawing. J Mot Behav 38, 383-404.

Turvey, M. T. 2007. Action and perception at the level of synergies. Human Movement Science

Turvey, M. T., Fonseca, S. 2009. Nature of motor control: perspectives and issues. Advances in Experimental Medicine and Biology 629, 93-123.

Vaillancourt, D. E., Newell, K. M. 2002. Changing complexity in human behavior and physiology through aging and disease. Neurobiology of Aging 23, 1-11.

Valero-Cuevas, F. J., Venkadesan, M., Todorov, E. 2009. Structured variability of muscle activations supports the minimal intervention principle of motor control. Journal of Neurophysiology 102, 59-68.

van Beers, R. J., Haggard, P., Wolpert, D. M. 2004. The role of execution noise in movement variability. Journal of Neurophysiology 91, 1050-1063.

van der Krogt, M. M., Bregman, D. J., Wisse, M., Doorenbosch, C. A., Harlaar, J., Collins, S. H. 2010. How crouch gait can dynamically induce stiff-knee gait. Annals of Biomedical Engineering 38, 1593-1606.

van Dieen, J. H., Pijnappels, M. 2007. Falls in older people. Journal of Electromyography and Kinesiology

van Dieen, J. H., Pijnappels, M., Bobbert, M. F. 2005. Age-related intrinsic limitations in preventing a trip and regaining balance after a trip. Safety Science 43, 437-453.

Van Emmerik, R. E. A., McDermott, W. J., Haddad, J. M., Van Wegen, E. E. H. 2005. Age-related changes in upper body adaptation to walking speed in human locomotion. Gait and Posture 22, 233-239.

Vaziri, S., Diedrichsen, J., Shadmehr, R. 2006. Why does the brain predict sensory consequences of oculomotor commands? Optimal integration of the predicted and the actual sensory feedback. The Journal of neuroscience : the official journal of the Society for Neuroscience 26, 4188-4197.

Veldpaus, F. E., Woltring, H. J., Dortmans, L. J. 1988. A least-squares algorithm for the equiform transformation from spatial marker co-ordinates. Journal of Biomechanics 21, 45-54.

Verbaken, J. H., Johnston, A. W. 1986. Population norms for edge contrast sensitivity. American Journal of Optometry and Physiological Optics 63, 724-732.

Verdaasdonk, B. W., Koopman, H. F., van der Helm, F. C. 2009. Energy efficient walking with central pattern generators: from passive dynamic walking to biologically inspired control. Biological Cybernetics 101, 49-61.

Verstynen, T., Sabes, P. N. 2011. How each movement changes the next: an experimental and theoretical study of fast adaptive priors in reaching. The Journal of neuroscience : the official journal of the Society for Neuroscience 31, 10050-10059.

Vogt, L., Portscher, M., Brettmann, K., Pfeifer, K., Banzer, W. 2003. Cross-validation of marker configurations to measure pelvic kinematics in gait. Gait and Posture 18, 178-184.

Wagenmakers, E. J., Farrell, S., Ratcliff, R. 2004. Estimation and interpretation of 1/falpha noise in human cognition. Psychonomic Bulletin and Review 11, 579-615.

Wagenmakers, E. J., Farrell, S., Ratcliff, R. 2005. Human cognition and a pile of sand: a discussion on serial correlations and self-organized criticality. Journal of Experimental Psychology: General 134, 108-116.

Watt, J. R., Franz, J. R., Jackson, K., Dicharry, J., Riley, P. O., Kerrigan, D. C. 2010. A three-dimensional kinematic and kinetic comparison of overground and treadmill walking in healthy elderly subjects. Clinical Biomechanics 25, 444-449.

Watt, J. R., Jackson, K., Franz, J. R., Dicharry, J., Evans, J., Kerrigan, D. C. 2011. Effect of a supervised hip flexor stretching program on gait in frail elderly patients. PM & R : the journal of injury, function, and rehabilitation 3, 330-335.

Weerdesteyn, V., Nienhuis, B., Geurts, A. C. H., Duysens, J. 2007. Age-related deficits in early response characteristics of obstacle avoidance under time pressure. Journals of Gerontology. Series A, Biological Sciences and Medical Sciences 62, 1042-1047.

West, B. J., Latka, M. 2005. Fractional Langevin model of gait variability. Journal of Neuroengineering and Rehabilitation 2, 24-24.

White, O., Diedrichsen, J. 2010. Responsibility assignment in redundant systems. Current Biology 20, 1290-1295.

Whitney, J. C., Lord, S. R., Close, J. C. T. 2005. Streamlining assessment and intervention in a falls clinic using the Timed Up and Go Test and Physiological Profile Assessments. Age and Ageing 34, 567-571.

Whittle, M. W., Levine, D. 1999. Three-dimensional relationships between the movements of the pelvis and lumbar spine during normal gait. Human Movement Science 18, 681-692.

Winter, D. A. 1991a. The Biomechanics and Motor Control of Human Gait: normal, elderly, and pathological. University of Waterloo Press, Waterloo.

Winter, D. A. 1991b. Changes in gait with aging. Canadian Journal Of Sport Sciences = Journal Canadien Des Sciences Du Sport 16, 165-167.

Winter, D. A. 1992. Foot trajectory in human gait: a precise and multifactorial motor control task. Physical Therapy 72, 45.

Winter, D. A. 2005. Biomechanics and Motor Control of Human Movement. John Wiley & Sons, New Jersey.

Wolpert, D. M. 2007. Probabilistic models in human sensorimotor control. Human Movement Science

Wolpert, D. M., Diedrichsen, J., Flanagan, J. R. 2011. Principles of sensorimotor learning. Nature Reviews. Neuroscience 12, 739-751.

Wolpert, D. M., Flanagan, J. R. 2009. Forward Models. In: Bayne, T., Cleemans, A., Wilken, P. (Eds.), The Oxford Companion to Consciousness, pp. pp294-296.

Wolpert, D. M., Flanagan, J. R. 2010. Motor learning. Current Biology 20, R467-472.

Wolpert, D. M., Ghahramani, Z., Jordan, M. I. 1995. An internal model for sensorimotor integration. Science 269, 1880-1882.

Wu, G., Cavanagh, P. R. 1995. ISB recommendations for standardization in the reporting of kinematic data. Journal of Biomechanics 28, 1257-1261.

Wu, J., Popovic, Z. 2010. Terrain-adaptive bipedal locomotion control. ACM Trans. Graph. 29, 1-10.

Wu, W., Meijer, O. G., Lamoth, C. J. C., Uegaki, K., van Dieen, J. H., Wuisman, P. I. J. M., de Vries, J. I. P., Beek, P. J. 2004. Gait coordination in pregnancy: transverse pelvic and thoracic rotations and their relative phase. Clinical Biomechanics 19, 480-488.

Zajac, F. E., Neptune, R. R., Kautz, S. A. 2002. Biomechanics and muscle coordination of human walking: Part I: Introduction to concepts, power transfer, dynamics and simulations. Gait and Posture 16, 215-232.

Zajac, F. E., Neptune, R. R., Kautz, S. A. 2003. Biomechanics and muscle coordination of human walking: Part II: Lessons from dynamical simulations and clinical implications. Gait and Posture 17, 1-17.

Appendix

Below is a copy of the forms that were sent out to the elderly subjects and their General Practitioner. These forms and selection processes indicate the steps taken to recruit healthy cohort of elderly population.

10.1 Appendix A

Victoria University

PO Box 14428
MELBOURNE CITY MC VIC 8001
Australia

Telephone:
(03) 9688 4000
Facsimile:
(03) 9919 1110

THE EFFECT OF AGEING ON THE CONTROL OF TOE CLEARANCE DURING WALKING

INFORMATION TO PARTICIPANTS:

Aims:

To examine how the ageing process affects biomechanical parameters of the locomotor system which coordinate and control toe clearance during walking.

Methods:

This research will consist of a minimum of 100 healthy, community-dwelling elderly (65+ years). To be an eligible participant for the research, you must be free from any diseases affecting gait, such as any neurological, musculoskeletal, cardiovascular, or respiratory disorders; rheumatoid arthritis; or diabetes, and will require a clearance and general health report from your general practitioner. All participants must be able to walk without the use of a gait aid and have no conditions that might impair normal walking (e.g. back pain, hip/knee replacements, severe osteoporosis, balance/vestibular disorders, foot problems such as ulcers or painful bunions). Also, participants must have not have had a fall within the last 10 years. Participants must also live in the community, perform regular outdoor activities and be generally independent in activities of daily living. Participants with vision impairment not correctable with lenses will be excluded from the study.

All participants will be asked to undergo a small battery of health tests on the day of testing. These include:

- Contrast sensitivity (Melbourne Edge Test);
- A timed up and go (TUG) test;

- An abbreviated mental test score (AMTS);
- Modified Falls Efficacy Scale (a ten item questionnaire where the participant rates their level of confidence in completing some everyday activities without falling);
- General health survey (a 20-item questionnaire where the participant self evaluates their health);
- Anthropometric measurements of the participants will be taken, including body height, body mass, limb length.

It is estimated that testing procedures will take approximately 2-2½ hours per participant. During this time a series of measurements will be carried out prior to the testing including measurement of body height; mass; leg length; ankle, knee and hip range of motion. In addition, 5 screening tests will be performed including vision, proprioception, body sway while standing, and reaction time tests. It is estimated that testing procedures will take approximately 2½ hours per participant. Motion capture (Optotrak Certus, Northern Digital™, Ontario, Canada) of the trunk, pelvis and lower limbs will be obtained by placing plastic shells on these segments of the body. A total of eight plastic shells will be attached to elastic neoprene bands wrapping around the skin of the thigh, leg and foot segments, the superior part of the pelvis and the middle back region (between the lumbar and thoracic vertebral area). Four small, lightweight 'infrared emitting diode' (IRED) markers (approximately 1cm in diameter) will be attached to each plastic shell and be wired to a 'control box' located on a lightweight backpack (~2 kg) positioned on the participant. Electromyography electrodes will record muscle activity and be attached to each lower limb and the lower back. This will involve shaving and using an alcoholic wipe to prepare the skin prior to electrode placement. These electrodes will also be connected to the 'control box'. Palpating bony landmarks is necessary and will be performed by the researcher for finding joint centers of the foot, ankle, knee, hip (greater trochanter), superior aspects of the pelvis (superior aspect of the iliac crest and anterior superior iliac spine) and vertebral processes. Wearing of a safety harness is compulsory for all participants, so as to support the trunk in the unlikely event of a fall caused by a loss of balance. During the testing procedure you are required to wear comfortable, flat soled, walking shoes, a tight fitting singlet or t-shirt and tight fitting above knee exercise shorts. The latter two items of clothing can be supplied for your convenience.

The walking test procedure will involve 3 parts:

1. walking overground for approximately 10 minutes.
2. 10 minute treadmill familiarity and speed adjustment test. During this period a steady increase in walking speed (0.5 km/hr increases after 40 second bouts) will be undertaken to determine your maximal voluntary walking speed and your preferred walking speed.
3. treadmill walking at your preferred walking speed for approximately 20 minutes.

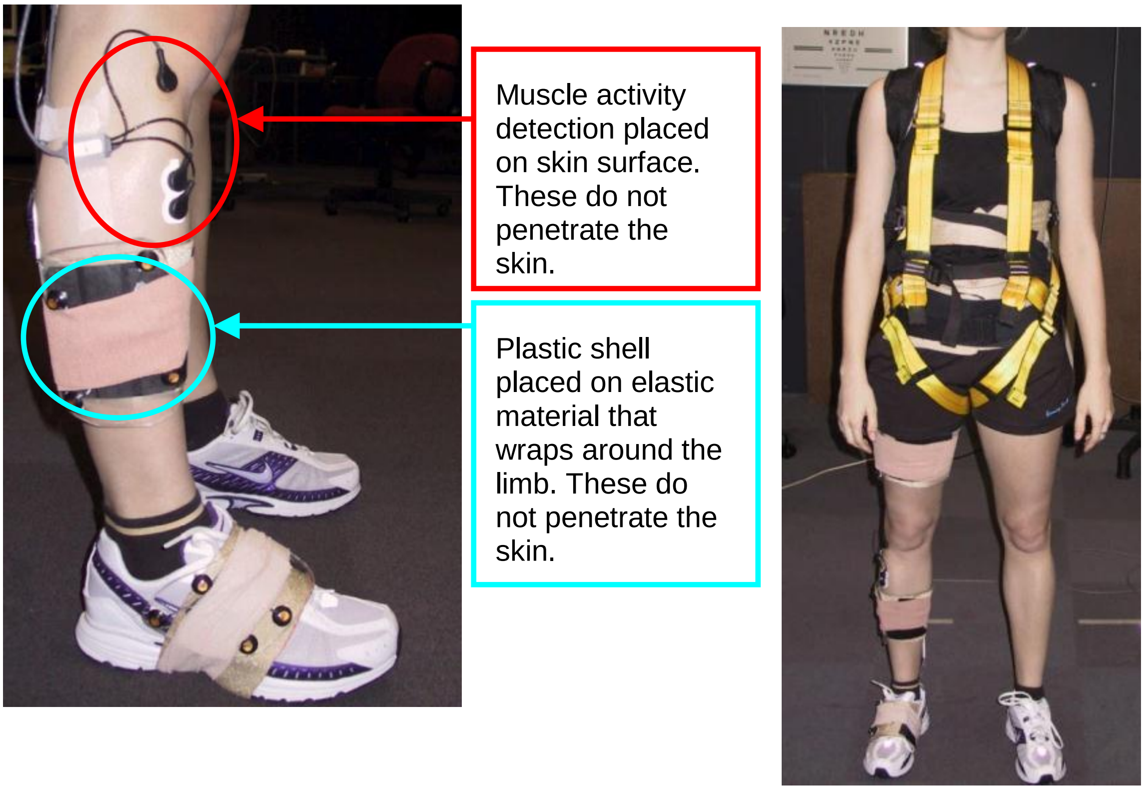
Risks and safeguards:

The study procedure has been approved by the Victoria University Human Research Ethics Committee. This testing is done while walking overground and on a motorized treadmill. If you feel any discomfort during testing, please stop walking. Testing will be constantly supervised and if any discomfort is present the researcher will ask you to stop walking and you will be given the option to withdraw from the testing procedure.

Access to the lab during testing is limited to the researchers and participants. The researcher supervising the testing has a first aid certificate and in the unlikely event of an injury the researcher will administer basic first aid. Telephones are located within the facility if further medical attention is needed.

Please refer to the pictures overleaf that describe the testing set-up prior to filling in the 'participant consent form'.

- ♦ Above knee lycra exercise shorts are supplied (freshly cleaned!), but you are welcome to provide your own
- ♦ Tight fitting t-shirt or exercise singlet are also supplied (freshly cleaned!), however you are welcome to provide your own
- ♦ Attachment of electrodes on the calf muscles (propulsion force and foot control) and the muscles of the lower back (trunk and posture control).
- ♦ Attachment of eight plastic shells on <u>both</u> the left and right limb segments, pelvis and mid back. Each plastic shell will be fastened to neoprene material which wraps around the skin of each limb segment.



Any queries about your participation in this project may be directed to the researchers (Name: Mr Simon Taylor, ph. 03-9919 1133; Dr Russell Best, ph. 03-9919 1118; Dr Rezaul Begg, ph 03-9919 1116). If you have any queries or complaints about the way you have been treated, you may contact the Secretary, University Human Research Ethics Committee, Victoria University of Technology, PO Box 14428 MC, Melbourne, 8001 (telephone no: 03-9688 4710).

10.2 Appendix B

Victoria University

PO Box 14428 Telephone:

MELBOURNE CITY MC VIC 8001 (03) 9688 4000

Australia Facsimile:

(03) 9919 1110

Date: 8 August 2006

Dear _________________,

First of all we would like to thank you for volunteering for this study into the ageing effects on locomotion. Here at Victoria University, our research focus is to understand why healthy elderly fall from trip related accidents. The purpose of this research is to investigate coordination and control over the foot motion during the swing phase of walking. Three testing protocols will be undertaken by the participants: 1) overground walking; 2) walking under changes in treadmill speed; and 3) a period of approximately 20 minutes of continuous treadmill walking. This investigation will test 2 healthy female groups – young and elderly. Your participation is important and will help lead us to design a model that will aim to reduce trip related falls in healthy elderly populations.

Inside this information pack you should find an 'Information to Participants' sheet explaining the purpose of the research, the way the research will be conducted, and the risks and safeguards that will be implemented. For your own safety, participation will not commence until the consent form has been completed.

After obtaining clearance from a medical GP, please call Simon on 99191128 to discuss an appointment time for testing your gait pattern. This session will be conducted at the Victoria University Biomechanics lab, located in the basement, 300

Flinders Street Campus. A map of Victoria University, City Flinders Campus is also provided for your convenience.

Thank you once again for your participation. If you have any questions or concerns, don’t hesitate to contact myself (Simon) or my supervisors Dr Rezaul Begg on 9919 1116 and Dr. Russell Best on 9919 1118.

Yours Sincerely,

Simon Taylor

Victoria University

PO Box 14428 | Telephone:
MELBOURNE CITY MC VIC 8001 | (03) 9688 4000
Australia | Facsimile:
(03) 9919 1110

Dear General Practitioner,

____________________ has agreed to volunteer in our study investigating healthy elderly (aged 65+) gait over a one-year period to determine whether the reported differences in gait post-fall occur in those elderly that fall within this period, or whether these reported different gait characteristics exist pre-fall. More specifically, it will the primary focus of this study to investigate if toe clearance changes following a fall, and to examine pre-fall toe clearance parameters of fallers to determine whether certain variables predispose elderly to falling. As a safety precaution, we request that participants obtain medical approval from their General Practitioner to ensure there are no underlying cardiorespiratory, or other medical conditions, which might present a health risk.

The study requires participants to walk continuously for 30 minutes at a self-selected comfortable walking pace on a treadmill. All participants will wear a safety harness and treadmill familiarization will be provided prior to this task. The 30-minute walking task alone has been conducted previously at the University with all elderly participants managing well. All methods have been approved by the Victoria University Human Research Ethics Committee.

Mobility and vision tests will be conducted in the Biomechanics Laboratory at Victoria University. Prior to conducting these screening tests and collecting data, it is essential to ensure all participants are 'healthy' and have no medical conditions (e.g. cardiac condition) that might compromise health and safety during the study. We would appreciate it if you would examine _______________ and complete the attached sheet. There is space for you to add any comments if you wish. ______________will return this sheet to us when they come in for testing. If you have any queries, please do not hesitate to contact us on 03-9919 1128 (Simon Taylor & Amanda Johns, PhD candidates) or 03-9919 1116 (Dr. Rezaul Begg, or Dr. Russell Best 03-9919 1118 (Supervisors)).

Please provide _____________ with receipt for this visit, as the university will reimburse costs of this visit.

Thank you for your time.

Regards,

Simon Taylor

Supervisors:

Dr. Rezaul Begg

Dr. Russell Best

INFORMATION ABOUT THE STUDY

Considering that falls are the leading cause of injury in the elderly, the most expensive of all accidental injuries in Australia, and that falls have a detrimental affect physiologically, psychologically, financially and socially to the individual, the factors that predispose an individual to falling needs to be identified. There is evidence to suggest that elderly fallers have different toe clearance characteristics during gait compared to their healthy elderly (non-faller) counterparts. This investigation will be the first longitudinal study in this research area to analyze the toe clearance parameters of elderly fallers over a one-year period with the aim of determining what makes these individuals more likely to experience a fall, and/or whether an elderly adapts to a 'safer' gait pattern after experiencing a fall in order to reduce the likelihood of additional falls. The second part of investigation is to determine the ageing effects on the biomechanical parameters that act to coordinate and control the toe clearance event. The findings of this research have the potential to answer key questions regarding falls in the elderly where previous research has not been able to provide conclusive evidence to what actually happens to the gait of fallers pre- and post-fall. The outcomes from this research will provide the basis for future research and, it is envisaged, will go some way in identifying the gait characteristics that may predispose elderly individuals to falling.

The study will be conducted at the Victoria University Biomechanics Laboratory located at 300 Flinders Street. Firstly, the participant will undertake a series of preliminary tests as mentioned below. Participants, wearing a safety harness, will then begin a familiarization period of treadmill walking. During this period a steady increase in walking speed (0.5 km/hr increases after 40 second bouts) will be undertaken to determine the participant's maximal voluntary walking speed and their preferred walking speed. After a rest period, the participant will walk continuously on the treadmill a comfortable, self-selected walking speed for approximately 30 minutes. Approximately 5 minutes of overground walking trials will also be collected and will act as a reference. Participants left and right feet, leg and thigh segments, also the pelvis and trunk will all have attached lightweight plastic shells. Each shell will have four wired markers (approximately 1cm in diameter). The motion of these markers will be measured using the Optotrak 3D measurement system. To determine muscular activity during the testing procedure, EMG recordings will be taken. Two EMG electrodes will be placed on the lateral head of the gastrocnemius and tibias anterior for the left and right legs. A ground electrode will be placed on the knee. Two more pairs of electrodes will be attached on the left and right erectors spinae. These are established and accepted methods used in biomechanics research.

Lower limb segment motion, muscle activity and joint force and moment data of the participants' gait will be collected. Data gathered by NDI® Optotrak (motion and ground reaction force) and Noraxon® (EMG) will be analysed in various ways. The main focus of the data analysis will look at the differences in gait, if any, that occur with the ageing process, and between pre- and post-fall in those elderly that fall within the one-year period.

All methods have been approved by the Victoria University Human Research Ethics Committee. Participants will receive training on the treadmill where necessary and are free to withdraw from the study at any time.

About the participants

All participants must be able to walk without the use of a gait aid and have no musculoskeletal/orthopaedic or other conditions that might impair normal walking (e.g. arthritis, back pain, hip/knee replacements, severe osteoporosis, balance/vestibular disorders, medications, foot problems such as ulcers or painful bunions). Participants must also live in the community, regularly activity outdoors and generally display independence with daily living activities. Participants with vision impairment not correctable with lenses will be excluded from the study results.

Screening tests to determine level of visual function, mobility, fear of falling and cognitive state, will be conducted at Victoria University. Participants scoring poorly will be excluded from the study. The screening tests include:

- Contrast sensitivity (Melbourne Edge Test);
- A timed up and go (TUG) test;
- An abbreviated mental test score (AMTS);

- Modified Falls Efficacy Scale (a ten item questionnaire where the participant rates their level of confidence in completing some everyday activities without falling);
- General health survey (SF-36, a 20-item questionnaire where the participant self evaluates their health);
- Anthropometric measurements of the participants will be taken, including body height, body mass, limb length.

These are all validated and routine tests used in similar studies, and within clinical practice, to test aspects of vision, physical performance, fear of falling and cognitive state.

If you have any queries regarding the study, please do not hesitate to contact us on 9919 1128 (Simon Taylor, PhD candidate), or 9919 1116 (Dr. Russell Best, Principal Supervisor).

Victoria University

PO Box 14428 Telephone:

MELBOURNE CITY MC VIC 8001 (03) 9688 4000

Australia Facsimile:

(03) 9919 1110

Participant Name: ______________________________

Address: ______________________________

____

GP Name:

Address:

Telephone:

In your opinion, is ______________________ of a sufficient health status to participate in the outlined study (please circle one)?

Yes No

Any comments?

Signed: **Date:**