CHARACTERISING INTRA- AND INTER-LIMB GAIT STABILITY USING MINIMUM FOOT CLEARANCE

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A Master's Thesis

Submitted in Fulfillment of the Requirements for the award of Master of
Applied Science – Biomechanics
Victoria University
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Melbourne, Australia
November, 2002

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DEFINITION AND EXPLANATION OF TERMS

Asymmetry	an inequality of the inherent characteristics between the left and right limbs, or link systems
Contra-lateral	opposite limb or link system
Control	the desired state of a system
Coordination	refers to a system becoming coordinated
Coupling	two limbs operating together
Dynamic Pattern	time history changes of continuous or discrete cyclical time ordered data points
Dynamics	a maths-science field that describes the changes in systems by expressing its existing and evolving states
Entrainment	the spatial tendency of one limb to follow the contra-lateral limbs pattern
Fall	implies an unintentional change from an upright body position to a lie on the ground
Fractal	a pattern that is self-similar at various magnifications
Inter-limb coordination	the step-to-step process happening between limbs or link systems
Intra-limb coordination	the stride-to-stride process happening within a limb or link system
Limb dominance	also referred to as laterality, it represents the locomotor system's preferred limb or link system to carry out a complex task because it has greater functional ability

Link System	the chain of body segments from the distal end of one foot to the distal end of the contra-lateral foot.
Long-term correlations	specific structure to time ordered data points, composed of fractals
Multi scaling	refers to multiple neural inputs that integrate in a stochastic manner at different frequencies to produce fractal structure of a biological output
Minimum Foot Clearance	the minimum vertical distance between a point on the shoe out-sole surface and the ground during the swing phase
Poincaré plot	a map describing the cyclic positioning of data points, such that one point on the map is linked to the next data point and so on.
Stability	minimizing fluctuations from one moment to the next
Symmetry index	a value that describes the percentage difference between the left and right limbs
Trip	contact of the foot with an obstacle, ground surface or an obstruction that prevents the foot's trajectory from traveling freely during the swing phase of gait
Unsteadiness	measure of fluctuating changes from one moment to the next

ABBREVIATIONS AND SYMBOLS

CV	coefficient of variation
D1	short term variability quantified from the Poincaré plot
D2	long term variability quantified from the Poincaré plot
D1/D2	relativity ratio, also termed μ
DFA	detrended fluctuation analysis, also termed α
ES	effect size
IV	inconsistency of the variance
LLS	left link system, defined by the left limb mobilizing, right limb supporting
MFC	minimum foot clearance
NSI	non stationary index
r	correlation coefficient
RLS	right link system, defined by the right limb mobilizing, left limb supporting
S	skewness
SD	standard deviation
SI	symmetry index
K	kurtosis

ACKNOWLEDGEMENT

Thanks are due to the school of Human Movement, Recreation and Performance, Victoria University for providing the author with an opportunity to further my study. Extended thanks to all the staff members of the Biomechanics Unit and CRESS for providing excellent facilities, assistance, knowledge, guidance and support throughout the duration of the project.

Specific thanks goes to my supervisor, Dr. Rezaul Begg, for his teachings, support, patience, and generous sharing of his time. I would also like to thank: my co supervisor, Dr Russell Best, for his guidance, recommendations and support; my sister Samantha, for her wisdom, guidance and encouragement; the lab technician, Rob Stokes; and my colleagues from the biomechanics unit; Tuire, Kevin, Jimmi, Dean, Tim, Pat and Steve.

Most of all, I'd like to acknowledge the support from my family and friends.

STATEMENT OF RESPONSIBILITY

I hereby certify that I am responsible for the work submitted in this thesis, that the original is my own except as specified in acknowledgements and that neither the thesis or the original work contained therein has been submitted to this or any other institution for a higher degree.

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ABSTRACT

This research examined aging effects by characterising one parameter of the gait cycle from treadmill walking. The minimum foot clearance (MFC) during the swing phase of gait is an essential event that reveals two things: 1) how the locomotor system achieves control over many degrees of freedom when performing this spatial end point trajectory task while at the same time being challenged by postural stability; 2) the risk associated with tripping over an unseen obstacle due to the performance of this task. Poor gait stability and tripping induced falls are contemporary issues related to elderly populations, where characterising MFC variability from one stride to the next can provide important links between these issues. Two healthy adult groups were compared: (i) young (n = 10; age = 33.3 ± 2.9 yrs) and (ii) elderly (n = 8; age = 72.75 ± 3.6 yrs). The experimental set up and data collection process was conducted at the Victoria University Biomechanics Laboratory (Flinders St. Campus). Subjects completed one trial of at least 20-minutes of treadmill walking at their self-selected gait velocity. Left and right MFC data was collected simultaneously from two bilateral cameras (50Hz) perpendicular to the motion of the treadmill belt. MFC data was extracted from a geometric model applied to the spatial coordinates of two markers captured by the Peak MOTUS digitising system. Dynamic analyses methods (detrended fluctuation analysis and Poincaré plot statistics) demonstrated aging affects for the leg with the low mean MFC. A symmetry index described aging differences for both descriptive- and dynamic- MFC parameters. MFC data is typically non-normal with positive skewness and kurtosis. The results reveal that the elderly have poor control over the limb which travels closer to the ground, as opposed to the young group who invariably display greater control over the low traveling limb at MFC. A combination of poor limb control and low ground clearance in the elderly indicates a higher likelihood of tripping.

Chapter 1

INTRODUCTION

Falls among older Australians is a serious public health problem. In Australia it has been found that one in three community dwelling people aged 65 years and over fall at least once per year (Blake, Morgan, Bendall, Dallosso, Ebrahim, Arie, Fentem and Bassey, 1988). Serious injury affects 10% of these cases, while 30% require medical attention. In Australia the community cost for managing injurious falls is estimated at $AUD 2,369 million annually (NIPAC, 2000). The consequences of less injurious falls does not directly fall into the annual cost account, however, it has a significant impact on the function and quality of life in older persons. The act of a fall for an individual can cause a person to fear falling and thus lead to immobilization, reducing activity levels and further increasing their risk of subsequent falls. By 2051, the Australian Bureau of Statistics (1998) projects that 26% of the Australian population will be aged 65 years or more, over twice the current percentage. These issues emphasise the need to clarify the aetiology of falls in the community and how to prevent them.

In 1999, a report by the Commonwealth Department of Health and Aged Care (CDHAC) identified intrinsic risk factors behind falling in older people (NIPAC, 1999a; 1999b). CDHAC lists age-related decline and disturbances of balance and gait as modifiable risk factors based upon scientific evidence. The report detailed the need for addressing research gaps within a falls prevention framework. In a further report by the CDHAC (NIPAC, 2000) listed a need for further research into developing assessment strategies that provide early identification of at-risk individuals. Of the falls statistics listed, more than 60% occur during walking and transferring from one place to the next, and more than 50% of falls result from tripping (Campbell, Borrie and Spears, 1989).

Biomechanical analysis of gait has revealed an association between kinematic variability in gait parameters and elderly fallers (Hausdorff, Edelberg, Mitchell and Goldberger, 1997b; Maki, 1997). However, because kinematic variability involves averaging, it does not fully describe the subtle perturbations within the spatial or temporal gait parameters occurring from one moment to the next. Movement control is measured by variability (Schmidt and Lee, 1999), and the research literature is limited with information on how walking is controlled from one stride to the next.

Throughout the better part of ones lifespan, walking for most people is a simple task that is taken for granted. Most people would say that walking from here to there requires minimal effort and attention. This is an amazing accomplishment, given that there are billions of neurons, hundreds of muscles, and more than 100 moveable joints in the body which require organizing when walking. The Russian physiologist, Nikolai Bernstein explained how the body solves this problem of coordinating rhythmic walking by reducing a complex system to a simple system that can be regulated without conscious effort (Bernstein, 1967). Once the problem of coordinating the complexity of rhythmic walking is developed during infancy, able-bodied gait is maintained by self-organising processes. These processes are without voluntary control and characterise the spatial and temporal patterns in gait (Clarke, 1995). Dynamic patterns arise from the moment-to-moment fluctuations in those sub systems that have been assigned the task of reducing the complexity of coordinating gait. When these subtle perturbations are analysed, the pattern dynamics can distinguish between a highly functional system and a system that is showing signs of aging or disease associated with the motor control system (Hausdorff, Michell, Firtion, Peng, Cudkowicz, Wei and Goldberger, 1997a).

Motor control theory from a dynamic systems approach has demonstrated some promise on exploring the issue of how gait is controlled from one stride to the next (Kelso, 1995). The approach requires analysis of cyclic repetitions of a certain event in order to quantify the moment-to-moment changes occurring. A dynamic systems approach can be related to the nature of gait, because gait requires moment-to-moment maintenance of stability between the base of support and the body centre of mass (Dingwell, Cusumano, Cavanagh and Sternad 2001). Measuring the autonomous control of walking rhythm from one moment to the next, or one gait cycle to the next, therefore, may provide a better assessment of gait stability.

The minimum foot clearance (MFC) is an event during walking that is related to maintaining dynamic stability. The foot follows a characteristic trajectory in the swing phase and is assigned the goal of clearing the ground during mid swing. Specifically, MFC is the vertical distance between the shoe out-sole and the ground surface. Winter (1992) reported MFC distance to be 1.29 cm. This small distance indicates the precision required by the motor control system to perform this task. During MFC, one limb supports the body while the other limb is mobilising to a new forward position. This task requires the control over a body link system that encloses the distal end of the mobilising limb to the supporting foot of the contra-lateral limb. Throughout this link system there are six joints and 12 major degrees of freedom (Winter, 1992). Investigating the bilateral task of coordinating consecutive MFC events within and between the limbs is a new application to gait analysis and may shed important insights into gait and falls research.

Historically, gait analysis has assumed that the left and right lower extremities function in a symmetric manner (Sadeghi, Allard, Prince and Labelle, 2000). Many research studies have collected data using unilateral methods, and assumed that the results can generalise the full nature of able-bodied gait. This is surprising given that limb preference and lateral structure bias has been long reported in research (Peters, 1988). The presence of lateral bias would suggest different patterns between the lower limbs. Studies that have examined the kinematic and kinetic aspects of gait from a bilateral perspective report mixed results as to whether symmetry can be assumed, or whether symmetry can be attributed to lateral bias (Sadeghi et al., 2000). Motor control experiments modeling inter-limb coordination dynamics have demonstrated that perfect symmetry is difficult to achieve from one moment to the next when coupled limbs are attempting 'in-phase' or 'anti-phase' symmetry (Turvey, Schmidt and Beek, 1993). These experiments demonstrate that spatial symmetry of the lower limbs cannot be assumed when walking. There are no current studies that have investigated MFC descriptive characteristics, or MFC dynamics, from a bilateral perspective.

Those aforementioned issues of gait stability and symmetry can be investigated by applying dynamic analysis methods to describe the MFC fluctuating patterns across both limbs simultaneously. Detrended fluctuation analysis (DFA) is now an established method for describing the dynamic fluctuations of gait, and has shown that a healthy gait control system reveals long-range correlations of consecutive strides over hundreds of gait cycles (Hausdorff, Peng, Ladin, Wei and Goldberger, 1995). DFA is used for quantifying gait instability (Hausdorff, Lertratanakul, Cudkowicz, Peterson, Kaliton and Goldberger, 2000), but has not been applied to spatial data series such as MFC. The second method for quantifying the moment-to-moment characteristics of consecutive MFC events is an original application for characterising dynamic systems described by the late nineteenth century French mathematician, Henry Poincaré. This approach requires the capturing of a dynamic behaviour at an instant (e.g. MFC) and mapping the captured value over repeated cycles in time (Clark, 1995). The map is called a Poincaré plot and there now exists descriptive statistics that characterize the form of the plot (Brennan, Palaniswami and Kamen, 2000). The Poincaré plot has not yet been applied to gait, however its concept has formed many high dimensional mathematical rationales that model various dynamic systems (Kelso, 1995).

Aging affects on the kinematic and kinetic variables of gait are commonly reported (e.g. Maki, 1997), and there is new evidence supporting the potential of dynamic systems applications for examining gait unsteadiness (Dingwell et al., 2001). Using dynamic systems concepts, this research addresses aging affects associated with an unsteady gait that is exposed in the MFC. The methodology of the research analyses walking on a treadmill for at least twenty minutes using a bilateral video set up. Gait is bilateral and the stability of gait is dynamically changing between limbs during forward progression. Simultaneous collection of data derived from cyclic repetitions of the MFC event within both limbs is new to gait analysis. Each successive MFC data point forms a collective time series order. The nature of the MFC is hypothesized as a challenging task for maintaining postural steadiness from one stride to the next during gait and is assumed to be susceptible to the effects of aging (Lord, Sherrington and Menz, 2001).

1.1 STATEMENT OF THE PROBLEM

Gait research has commonly assumed a symmetric pattern between the limbs and has therefore made conclusions from unilateral analysis approaches. There are many unsolved issues in gait research, such as coordination of the body centre of mass over the base of support from one moment to the next. This requires a comprehensive bilateral analysis of a dynamic parameter that is susceptible to posture stability when walking. The understanding of gait is needed for clarifying the risk factors characteristic of elderly persons who are predisposed to falls. The rise in the elderly population, the current rate of hospitalizations in this group due to falls, and the resource demands placed on the national health care system emphasises the importance of this type of investigation.

1.2 SIGNIFICANCE OF THE STUDY

The cost to the community due to falls is approximately $AUD 2.5 Billion (NIPAC, 2000). There is an abundance of gait and falls research circulating that are improving the understanding of the mechanisms behind the aetiology of falls in the elderly. However, there exists a void within this extensive body of literature that this investigation can begin to address. Little is known about the coordination and control of gait from a dynamic perspective. Little is known about the relationship between the left and right limbs and the strategy used by the locomotor system to negotiate the terrain, which can be determined from MFC characteristics.

The research design used in this dissertation utilizes the following advantages. Firstly, by adopting a bilateral approach to data analysis, an investigation into gait symmetry and between limb coordination can be undertaken. Secondly, new methods from dynamic pattern analysis will demonstrate a new insight into gait control. Thirdly, the MFC parameter investigated is a precise motor control task that is also linked with falls.

1.3 AIMS

General

The aim of the research is to examine the differences between healthy young and healthy elderly population groups on the minimum foot clearance event during treadmill walking.

Specific

The specific aims can be divided into two areas:

(1) To determine the presence of gait symmetry from MFC characteristics and to investigate associated ageing effects

(2) To determine whether MFC control is diminished due to aging effects

1.4 HYPOTHESES

The following null hypotheses were tested at the .05 significance level:

1. There is no significant difference of the MFC data distribution from a normal distribution
2. There is no significant difference between the elderly group and the young adult group for MFC Mean, CV, SD, S and K
3. There is no significant difference in the long-range correlations from random correlations in MFC data for both the elderly and young groups
4. There is no significant difference in DFA scaling between the elderly and young groups

5. There is no significant effect for measures of intra- or inter-limb unsteadiness between the elderly and young groups
6. There is no significant effect for measures of symmetry between groups
7. There is no significant difference of unsteadiness measures between inter- and intra-limb MFC events for both young and elderly groups
8. There is no significant effect for treadmill walking speed on unsteadiness parameters for both the young and elderly groups
9. There is no significant effect among the relationship of unsteadiness parameters for both the young and elderly groups

1.5 DELIMITATIONS

The study was delimited to:

1. eight healthy elderly subjects
2. ten healthy young subjects
3. 20 minutes of treadmill walking
4. bilateral data collection
5. digitizing of the shoe out-sole position

These delimitations will be discussed throughout the thesis.

1.6 LIMITATIONS

The limitations of the study:

1. 2-d data collection
2. Geometric model of the shoe out-sole position
3. 18 subjects

4. Inter-limb coupling methods

5. Treadmill walking

6. No limb dominance testing

These limitations will be discussed throughout the thesis.

Chapter 2

LITERATURE REVIEW

The review of literature encompassed the aetiology of falls; theoretical concepts underlying the control and coordination of gait; and the evolution of practical applications being performed that serve to analyse gait performance. The review begins with a background into the nature of the problem of falls in the elderly. The link with gait and locomotor tasks is identification within this field of literature. Following this section, a review of the basic features of gait, and the biomechanical parameters commonly reported in the literature is addressed. These sections are followed by the locomotor system, foot trajectory, inter-limb coordination and dynamic methods of analysis.

2.1 FALLS IN THE ELDERLY

2.1.1 The problem

Elderly falls is a serious national and international health concern. In Australia at least a third of people aged 65 years and over are falling at least once each year (Lord, Sambrook, Gilbert, Kelly, Nguyen, Webster and Eisman, 1994) and this is believed to be an underestimate due to non-reporting of non-injurious falls. This figure is commonly reflected in studies for both rural and urban areas (Luukinen, Koski, Hiltunen and Kivela, 1994; Lord, Ward, Williams and Anstey, 1993) and throughout other parts of the world (Campbell et al., 1989). Studies have reported slight differences in injury consequences of falls where between 22% and 60% of fallers suffer injury (Lord et al., 2001). In the US a report to congress revealed that persons aged over 70 years will spend 70% of their total health care costs on injury related falls, and the average cost per injured person is nearly double that for injured persons of other ages (Rice and McKenzie, 1989).

The annual incidence of falls appears slightly more prevalent in female groups within community settings (Campbell et al., 1989; Lord et al., 1993). Regardless of sex, institutional settings such as nursing homes report even higher fall rates, nearly twice the rate of community dwelling populations (Yip and Cumming, 1994). Nearly 80% of falls in the community occur from gait related activities and only a smaller proportion of falls result from other activities such as stair climbing, and negotiating within the home (Lord et al., 1993). In a prospective study by Campbell, Borrie, Spears, Jackson, Brown and Fitzgerald (1990) most falls that occurred in public places involved tripping on uneven ground.

Commonly, a subnormal walking pattern has been linked to falls (Wolfson, Whipple, Amerman and Tobin, 1990). From retrospective studies conducted on community dwelling elderly persons, more than 50% of falls occur during some type of locomotion (Lord, 1996; Tinetti, Speechley and Ginter, 1988). Within the community, Lord et al. (2001) found that 40% of fallers report that their fall was the result from a trip, while 21% from poor balance (see Figure 2.1.1.1).

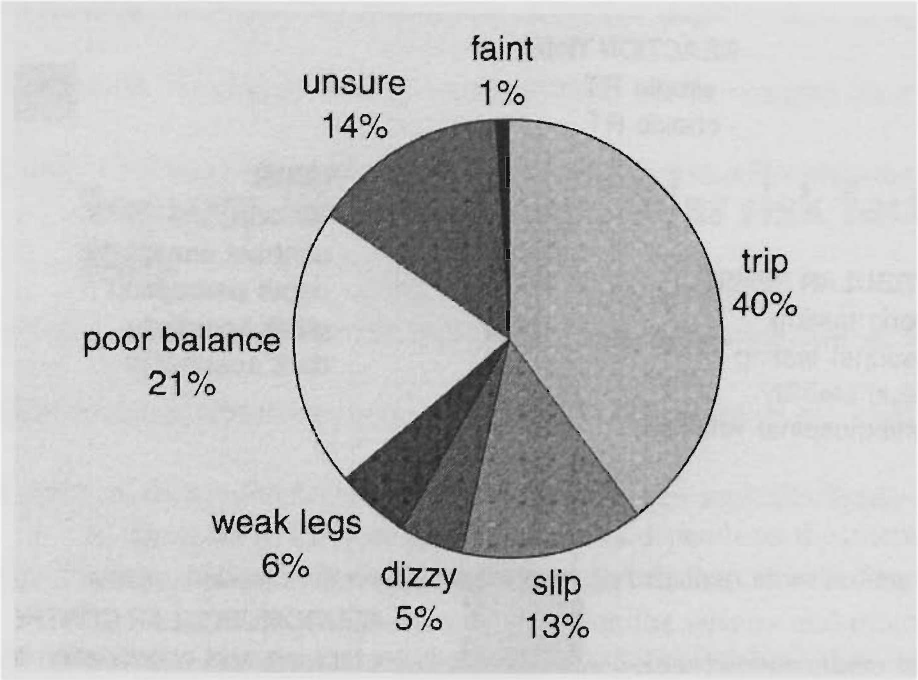

Figure 2.1.1.1 Causes of falls. (Source: Lord et al., 2001).

Another investigation reported that tripping is responsible for 24% of falls and slipping 17% (Tinetti, et al. 1988). Blake et al. (1988) have found that tripping is the cause of at least 50% of the falls occurring within a community setting. A fall occurring from ambulating has been defined by the Kellogg international working group as "unintentionally coming to ground or some lower level and other than as a consequence of sustaining an [external] blow loss of consciousness, sudden onset of paralysis as in stroke or epileptic seizure" (Gibson, Andres, Isaacs, Radebaugh, and Worm-Petersen, 1989).

The incidence of fall-related hospital admissions is found to increase exponentially after 60 years of age (Lord, 1990). This rate has been relatively stable over the last ten years. Injurious falls result in a loss of mobility and a decline in their walking ability. A proportion of this group will require walking aids. Between 2 and 6% of older people that suffer from a fall will incur a bone fracture and 1% will require a hip replacement (Speechly and Tinetti, 1991; Lord, 1990). Hip replacement surgery results in long-term bed rest and a rehabilitation program for regaining mobility. Bed rest results in further degeneration of muscle strength. In a study by Marottoli, Berkman and Cooney (1992) of the people recovering from hip replacement, only 15% were able to walk independently at 6 months post-surgery.

Further fall-related consequences to the individual are a developed fear of falling, morbidity and loss of independence affecting their quality of life (Marattoli et al., 1992). Fear of falling accounts for 48% of those people who have suffered a fall and this leads to a reduction in daily activities (Tinetti, Mendes de Leon, Doucette and Baker, 1994). This doesn't account for non-fallers who feel unsteady and consequently have similar fears that will affect walking function, which may place a person at further risk of subsequent falls (Maki, 1997). A certain percentage of persons who suffer from a fall will be referred to an institutional setting (Lord, 1994). Such are the changes in an elderly person's life due to a fall.

The incidence rate of falls is expected to increase based upon population growth and current indications of hospital referred trends from falling. The aging trends in Australia for the 65 years and over population group has increased by 72% over the last 20 years (Australian Bureau of Statistics, 1998). This group currently represents 12% of the population. Projections by the Australian Beureau of Statistics (1998) estimate a growth of this group to represent 24-26% of the total population by 2051. For the old elderly (above 85 years) projections estimate that this group will increase from 1.2% to approximately 4.5%.

2.1.2 Falls prevention framework

Falls prevention programs over the last 10 years have made only slight gains judging by the rate of hospitalizations caused from falls in elderly groups. The Commonwealth Department of Health and Aged Care (CDHAC) has developed a falls prevention framework to help guide researchers. Within this framework, CDHAC has highlighted areas requiring further research and development. A need for predicting at-risk individuals and for assessing intervention programs that seek to reverse causal mechanisms is a priority (NIPAC, 1999b). To date, emerging cause and effect relationships have lacked supportive research strength. This is probably why the rate of hospitalizations due to falls has not steadied (Lord et al., 2001). Therefore, greater clarity between cause and effect relationships between gait characteristics and falls is required for the improvement of falls prevention programs.

2.1.3 Factors related to falling

Generally, the cause of falls may be categorized into two groups: intrinsic and extrinsic factors. Extrinsic factors have been shown to have less influence on falling in comparison to intrinsic factors. Intrinsic, or behavioural factors can be attributed to: prescribed medications; aging effects such as vision, musculoskeletal and neurophysiological systems that contribute to obstacle negotiation, balance and mobility; and a fear of falling. The extrinsic factors relate to environmental conditions, such as poor lighting, home settings (bathrooms and floor rugs), uneven and/or slippery footpath surfaces.

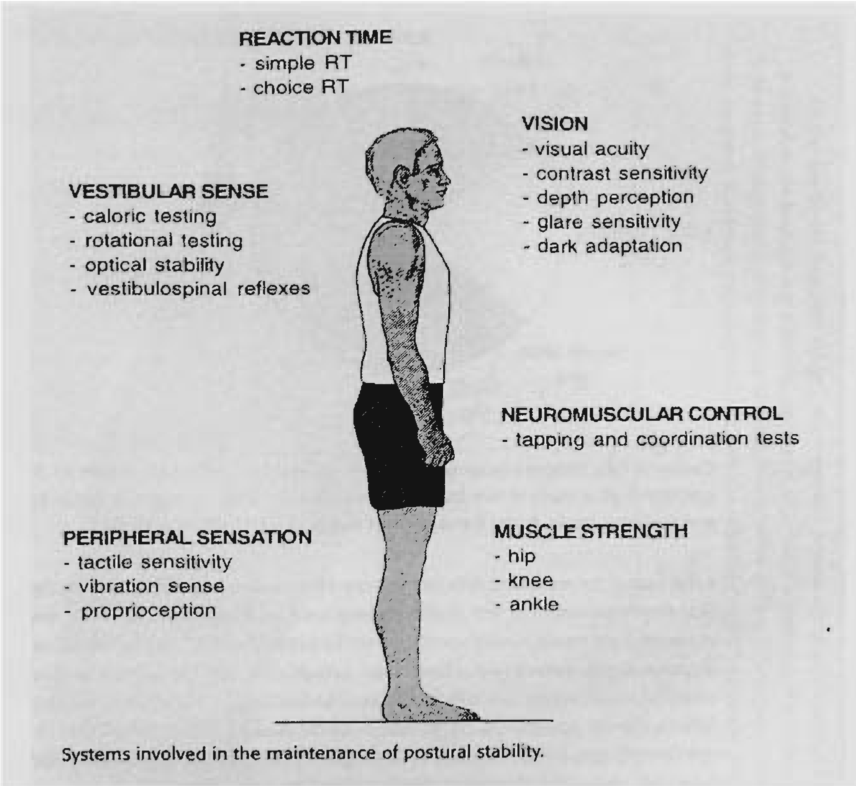

Figure 2.1.3.1. Systems involved in the maintenance of postural stability. (Source: Lord et al., 2001).

Age-related loss in neuromuscular and sensory function affects the ability to maintain postural stability during walking. It is during walking that we place challenging demands upon our body to control the centre of gravity within a stable boundary, requiring an ability to receive appropriate afferent information from the sensors. Lord et al. (2001) suggests that falling is related to age-related decline in vision, peripheral sensation, vestibular sense, muscle strength, reaction time, and neuromuscular control (Figure 2.1.3.1).

Studies report the following systems as an indication of falling risk factors. Of the sensory systems: contrast sensitivity (Ivers, Optom, Cumming, Mitchell and Attebo, 1998); and peripheral sensation demonstrate associations (Lord et al., 1994). Poor functioning of the neuromuscular system, such as muscle weakness (Lord et al., 1994) and reaction time (Lord, Clarke and Webster, 1991) are further risk factors linked to falls. The aging process is associated with significant changes to the nervous, muscular, and skeletal systems that influence physical capability and balance (Lord et al., 2001). Other risk factors cited are the number of medications used by a person, cognitive impairment or confusion, depression, age, health status, and decreased physical capability (Lord et al., 2001).

2.1.4 Fall prediction

Dysfunctional systems are linked to persons who have fallen, but, diagnosing the likelihood of a person having a fall based upon simple tests has proved elusive for researchers and clinicians. This is due to the ability of higher functional systems to complement systems that have regressed, such as vision accommodating for poor vestibular function. Lord et al. (2001) illustrates that a combination of dysfunctioning systems increases the likelihood of a person falling, however, these systems when tested individually may not show a below standard measure.

Multifactorial designs have investigated the interaction of a number of associated risk factors with falls. The risk factors that are commonly cited are those associated with health status; fear of falling; medications; gait; mobility; balance and posture. Prospective multifactorial studies by Berg, Alessio, Mills and Tong (1997) and Hill, Schwarz, Flicker and Carroll (1999) examined the interaction of a variety of variables for predicting fall behaviour. Berg et al. (1997) compared measures of vision, health, physical activity and performance, and fall history to predict the risk of recurrent falls in independent community dwelling older adults aged 60 - 88 years (n = 96). Stepwise logistic regression analysis found only vision and low systolic blood pressure to be strong predictors of recurrent fallers.

Similarly, 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 subjects who showed decreased sensory and cognitive functioning. From multivariate logistic regression analysis, measures of gait performance were most strongly associated with prediction of multiple fallers. Other studies have focused on gait function only, and have found differences between fallers and non-fallers and concluded that not all but certain specific gait parameters can predict falling (Hausdorff et al., 1997b). The authors found that variability of stride time intervals was associated with fallers, whereas the average measures of stride time were not.

People who have experienced more than one fall show a strong predictor of subsequent falling (Maki, 1997). Hence, having a fall does not seem to change any behavioural characteristic, which helps to prevent the person from falling again. This is interesting since studies have shown that fear is associated with modifications to walking (Tinetti Richman and Powell, 1990; Maki, 1997). Fear of falling has also been linked to a higher risk of falling (Maki, 1997), thus the psychological changes made to improve gait stability do not appear to reduce subsequent falls.

2.1.5 Gait assessment for predicting falls

Gait assessment is commonly performed when examining elderly people and their risk of falling. Gait assessment is an encompassing goal for defining the global functioning of the locomotor system, whereas gait analysis usually refers to more specific and technical aspects of assessment (Whittle, 1996). Gait assessment cannot be viewed simply from a biomechanical or motor control perspective, as gait requires the integration of all the underlying systems of movement working in concert (McClay, 1995). From a biomechanical viewpoint, gait is dictated by the inherent anatomical structure and alignment, available joint range of motion underlying our movement patterns (kinematics), and overall muscle strength generating these patterns (kinetics). From a motor control viewpoint, gait is achieved from the successful interplay of central and peripheral nervous system domains (Leonard, 1995). The biomechanical and motor control patterns can be measured against normative data to describe whether dysfunction is present within a person. Empirical studies of gait analyses performed on groups that have experienced a history of falls may then provide a rationale for theoretical constructs as to why falls occur. The development of this information can then lead to assessments of a person's risk of falling.

2.2 GAIT

Most of the systems and mechanisms measured as having hypothesized links with falls are related to the locomotor system. This is of little surprise because falling is a consequence of the locomotor system being unable to respond adequately (Pavol, Owings, Foley and Grabiner, 2001). There are significant changes to the mechanisms within the nervous, muscular and skeletal systems that are affected by the aging process (Grabiner, 1997). A key task of the locomotor system is integrating and augmenting these mechanisms to initiate, generate and maintain gait. Common aging effects listed above are thus going to affect gait. A poor gait may not always be responsible for causing a fall, nevertheless, observing gait can reveal the functional status of the locomotor system. Aging effects are specific, being unique to the individual, such that each system or mechanism alters its functional capacity independently. However, measuring systems or mechanisms independently does not always provide a good representation of locomotor performance. Finding and measuring the appropriate gait parameters that may capture the global functioning of the locomotor system is the challenge facing researchers in falls prevention.

2.2.1 Gait definition

From this point on, human walking will be referred to as gait. Gait is unique due to its bipedal nature, placing considerably larger equilibrium demands compared to the gait of other primates. Gait is an achievement of movement that transfers the body's centre of gravity over a base of support, provided by the lower limbs (Figure 2.2.1.1). As the body moves forward, one limb serves as a mobile source of support while the other limb advances itself to a new support site (Perry, 1993). The other limb then repeats this role, reproducing a similar pattern of movement. Hence, gait requires inter-limb coordination to propel and then recapture the centre of gravity over a dynamically changing base of support. A fall is related to the inability of the lower limbs to recapture the centre of gravity. Maintaining stability (this is referring to the ongoing control over many gait cycles) of the centre of gravity appears to be the ultimate challenge when walking. Nevertheless, failure to adapt to major perturbations in the environment by the lower limbs will challenge even the most stable postures during gait.

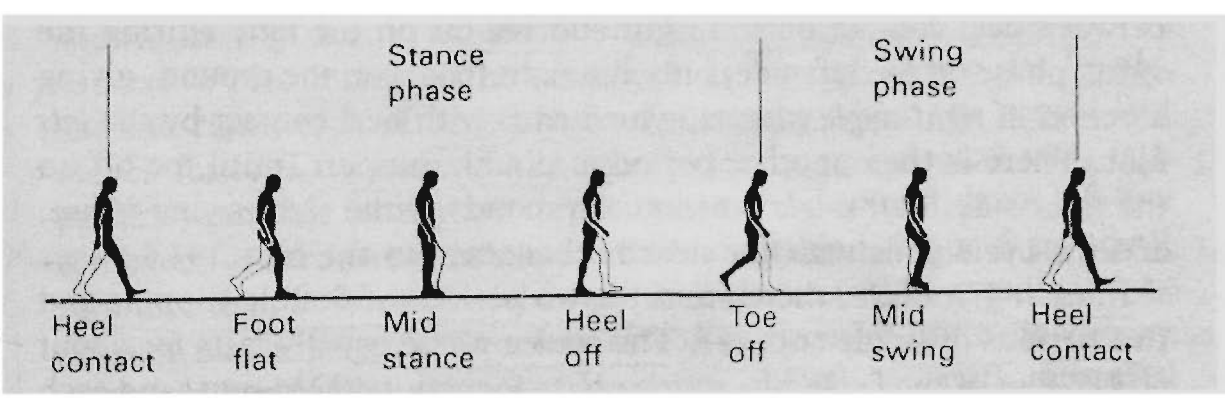

Figure 2.2.1.1. Dynamic changes of the base of support, from right heel contact to right heel contact. (Source: Whittle, 1993).

2.2.2 Walking cycle

The human gait cycle is defined as the interval between two successive occurrences of one of the repetitive events of walking (Whittle, 1993). Commonly, this interval is from heel contact to heel contact of the same foot (Figure 2.2.2.1). It is important to note here, that the left limb goes through exactly the same sequence of events as the right, but displaced in time by half a cycle. The swing phase contributes approximately 40 per cent of the gait cycle, concurrent with the single support phase of the contralateral limb. The stance phase lasts for approximately 60 per cent of the walking cycle, and is comprised of a single support phase and two double support phases. Double support contributes 20 per cent of the stance phase, 10 per cent prior to single support and 10 per cent following single support.

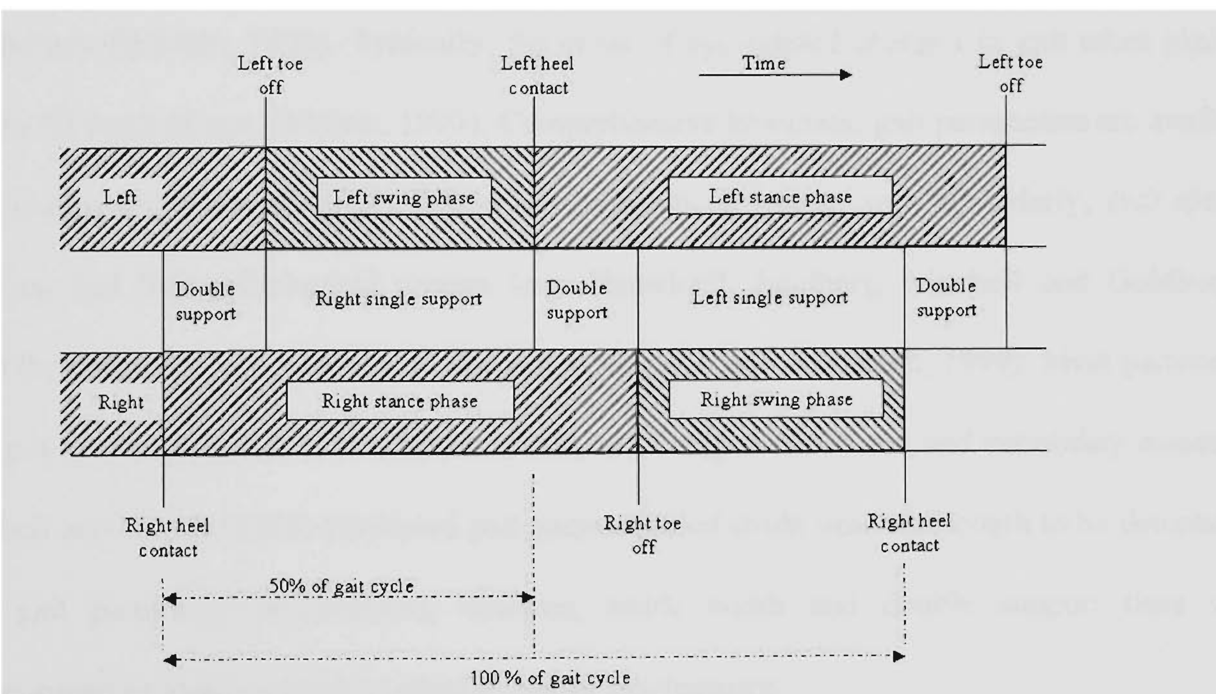

Figure 2.2.2.1. Timing of single and double support during a single gait cycle from right heel contact to right heel contact. Identical contra-lateral events are displaced in phase by half a walking cycle. (Source: Whittle, 1993).

2.2.3 Biomechanics of gait

Coordinating the gait cycle is dependent upon the biomechanical constraints inherent within the locomotor system. These constraints can be observed and analysed within and between two levels. At the micro level coordination requires control over the many inherent properties of joints, muscles and motor units, to achieve a movement solution (Holt, 1998). At the macro level, this solution may pertain to joint kinematics or kinetics between two segments or among multi segments (Holt, 1998). This review of literature as it relates to the biomechanics of gait will consider the macro level of coordination.

2.2.4 Gait measures

The elderly are subject to two influences – the effects of age itself, and the effects of pathology (Whittle, 1993). Typically, the onset of age-related changes in gait takes place at 60 to 70 years of age (Whittle, 1993). Comprehensive kinematic gait parameters are available for comparing biomechanical differences between (i) young and (ii) elderly, (iii) elderly fallers, and (iv) pathological groups (e.g. Hausdorff, Edelberg, Mitchell and Goldberger, 1997b; Dingwell, Ulbrecht, Boch, Becker, O'Gorman and Cavanagh, 1999). Most parameters of gait researched exist in two forms: mean, or averaged measures; and variability measures. Gabell and Nayak (1984) attributed gait parameters of stride time and length to be determined by gait patterning mechanisms, whereas, stride width and double support time were determined predominantly by balance control mechanisms.

Generally, aging affects on gait show that stride length is reduced, walking base of support is increased, and cadence (steps per minute) is reduced although variable (Winter, Palta, Frank and Walt, 1990; Prince, Corriveau, Hébert and Winter, 1997; Hageman and Blanke, 1986). Of the most noticeable biomechanical change observed in elderly gait is that their freely chosen, or preferred walking speed, is slower than that of young adults. This is expected as walking speed is related to cadence and stride length. Not all studies support the aforementioned results, for example Blanke and Hageman (1989) found no significant differences between younger and older males in walking speed, step width and length. There appears to be no obvious reasons why these studies have found conflicting results.

Slow walking can be due to a combination of increased double support time and prolonged single support time. Prolonged single support time challenges the stability of the body centre of mass (BCM), and stability of the BCM appears to be the primary constraint placed on the locomotor system of elderly people. Slow walking is associated with postural instability, and has differentiated between those subjects with a history of falls and those without, with fallers walking more slowly than non-fallers (Wolfson et al., 1990). However, Pavol, Owings, Foley and Grabiner (1999) found that increased walking speed is a strong predictor of falls resulting from an induced trip. Hausdorff et al. (1997b) compared gait speed of three groups: young, elderly non-fallers, and elderly fallers, and found a significant increase in the variability of the stride time intervals for elderly non-fallers when compared with young healthy adults. However, between non-fallers and fallers, gait speeds were similar. In one investigation, the association in the level of activity of an elder person's lifestyle could not be based upon walking speed (Grabiner, 1997), however, Prince et al. (1997) found that daily living function is related to walking speed.

Double support time is considered an indicator of postural stability (Whitall and Clarke, 1994). Elderly people generally display increased stance and double support time (Winter, 1991), adaptations generally attributed to a slower, safer and more stable gait. Hill et al. (1999) found double support time to be a strong predictor of fallers. However, in a retrospective and prospective investigation into kinematic gait parameters (stride-length, -time, -width, double support time, and gait symmetry), Feltner, MacRae and McNitt-Gray (1994) failed to find significant differences between fallers and non-fallers, or to predict subsequent falls (n = 17). The literature on gait speed and its association with falls incidence demonstrates that it is a contentious issue and one that is important for educators involved in falls prevention programs.

Many studies of gait are limited to unilateral methods, which assumes that one limb functions in a similar way to the contralateral limb. Furthermore, unilateral approaches overlook important phenomena of the coordinating relationship between left and right limbs. There is no research exploring the possibility that MFC is influenced by a 'coordinating mechanism' that is acting between the limbs. Section 2.6 reviews the research literature of inter-limb coordination in greater depth.

There is limited research exploring aging effects, coordinating foot trajectory symmetry and the association to falls. In a multifactorial investigation, Hill et al., (1999) found that the average difference between the left and right stance phase durations were a strong predictor of fallers. Begg and Sparrow (2000) compared bilateral parameters of stride length and stride duration for both young and elderly adult groups during unobstructed walking and when negotiating a raised surface. The results show that within the elderly group, the mean values obtained for each limb's stride length during unobstructed walking revealed a 5 cm difference, thus indicating spatial asymmetry.

Other investigations of gait asymmetry have identified differences of bilateral parameters in kinematics (Gunderson, Valle, Barr, Danoff, Stanhope and Snyder-Mackler, 1989), and kinetics (Crowe, Samson, Hoitsma and van Ginkel, 1996; Giakas and Baltzopoulos, 1997; Hamill, Bates and Knutzen, 1984). The presence of gait asymmetry has been confirmed (e.g. Crowe et al., 1996), however, reports of gait symmetry are more prevalent in the literature (Sadeghi et al., 2000).

2.3 GAIT UNSTEADINESS

Gait unsteadiness is observed in the inconsistency or arrhythmicity of stepping (Wolfson et al., 1990). Consistency in gait speed from one stride to the next is one parameter of a gait assessment battery that has been successful in determining falls (Wolfson et al., 1990). These qualitative measures of gait unsteadiness aim to assess the gait pattern from one stride to the next. Although this method is subjective, it does have some advantages over quantitative methods that describe gait changes as an average measure derived from several strides, which consequently overlook moment-to-moment inconsistencies. Characterising the moment-to-moment inconsistencies may allow a more comprehensive insight into how the locomotor control system responds to perturbations in the attempt to maintain gait steadiness.

Although Wolfson et al. (1990) have provided a clear definition of gait unsteadiness, there have continued to be different methods used for determining its presence in the research literature. The conclusions drawn from the various methods have contributed to a clouded perspective proposing cause and effect relationships linking gait and falls. Gait unsteadiness is commonly quantified by defining the variability of certain gait parameters. Currently, the literature on gait unsteadiness encompasses two measures of variability: (i) magnitude of fluctuations; and (ii) moment-to-moment fluctuations. It is noted here that both variability measures are not equated, but represent fundamentally different phenomena of the changes taking place. This is the root of the issue behind conflicting reports that have sought relationships between gait unsteadiness, locomotor functioning, gait speed and falls. The following two sections will address the two different measures of variability as they relate to gait unsteadiness.

2.3.1 Descriptive Variability

Measures of variability commonly reported in research based upon descriptive statistics are the standard deviation (SD) and coefficient of variation (CV). The SD and CV describe the average fluctuation magnitude and are both normalised with respect to the mean value (e.g. $CV=100\times SD/mean$).

Gabell and Nayak, 1984 did not find any significant differences between young (21-47 yrs, n = 32) and older adults (66-84 yrs, n = 32) for measures of coefficient of variation (CV) in double support time, stride width, stride time, and step length. Generally, the older group had a greater range of variability, which is understandable as this is related to the differences within aging processes between 66 and 84 years of age. One important reason why the study may have failed to find a significant difference is because the available length of walkway was only 3 meters, limiting successive strides to a likely maximum of two. Thus, in this study stride-to-stride variability information is dependent upon seemingly isolated trials.

Maki (1997) conducted an investigation, which clarifies the issue of walking speed, fear of falling and stability. In a prospective study, falls data was collected over twelve months, from 75 self-care residents, aged from 62 to 96 years, following initial gait assessments. The walkway used by Maki is also limited to several successive trials, as it is only 8 meters long and the first 3 and last 3 steps were discarded. Averaged measures of stride length, double support time, and speed were associated with fear, however, not related to falling.

Conversely, measures of variability in stride length, speed, and double support were independently associated with falling but not fear of falling. Thus it appears that average measures of gait parameters may not differentiate between fallers and non-fallers. The variability of these parameters may suggest that variability measures of gait offer greater potential insight for differentiating between fallers and non-fallers.

Hausdorff et al. (1997b) measured for the first time, variability in stride time intervals for an extended walk (6 minutes) using standard deviation (SD) and coefficient of variation (CV). Variability as measured in the previous studies is limited in the number of consecutive strides. By investigating stride-to-stride time interval variability of the various phases (stance, stride, and swing) within a cycle, community dwelling elderly fallers were significantly higher for all phases when compared with young subjects. The coefficient of variation of the elderly fallers was nearly twice that of the non-fallers, and significantly higher on all measures. The difference is best observed from the time series graphs (Figure 2.3.1.1), showing that elderly fallers are unable to maintain a constant gait cycle time and an inability to regulate events within a gait cycle.

Due to the retrospective nature of the study design, Hausdorff et al. (1997b) acknowledged the limited conclusions that may be drawn from this study in relation to falls aetiology. Higher variability found in the fallers may be an adaptation to fear of falling, rather than predisposing characteristics causing falls. Interestingly, the mean gait speed was identical between fallers and non-fallers (1.13 m/sec). This indicates that gait unsteadiness measured by descriptive variability may not be due to gait speed.

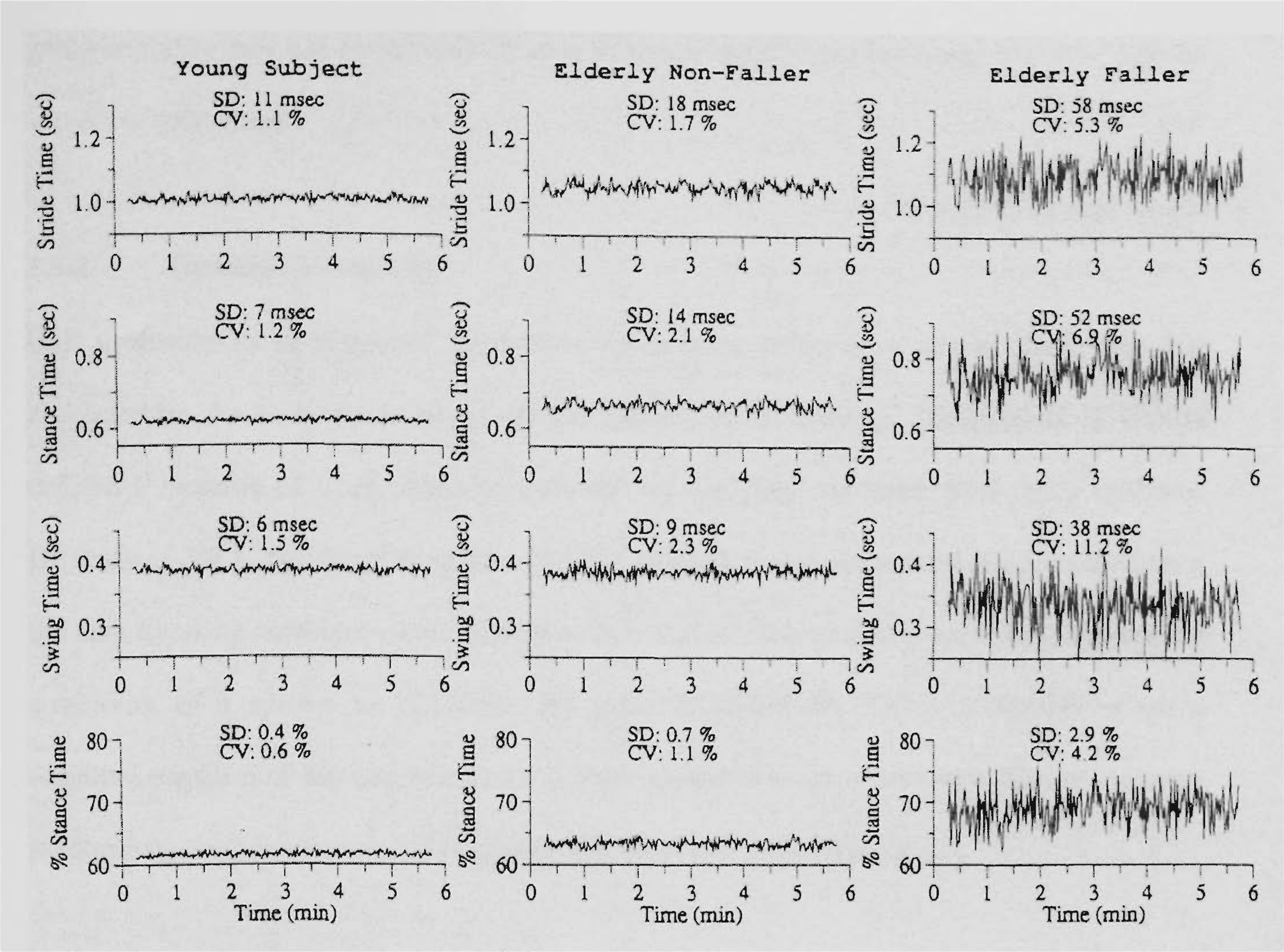

Figure 2.3.1.1 Time series of the temporal parameters of gait from 6 minutes of walking SD, standard deviation; CV coefficient of variation). The graphs demonstrate the increase in variability magnitude for an elderly faller, whereas only slight increases in variability compares an elderly 'non-faller' from the young subject. Adapted from Hausdorff et al. (1997b).

Maki (1997) suggests that increased stride-to-stride variability in elderly fallers is due to impaired motor control. Studies by Hausdorff et al. (1997a) investigated the relationship of impaired motor control and stride-to-stride variability by analyzing subjects with Huntington's disease. The nature of the disease affects the central nervous system, primarily the basal ganglia, which is a primary regulator of motor control. The results showed that subjects with Huntington's disease had significantly higher stride interval CV. There was no significant difference in mean stride interval. Of interest from Hausdorff et al. (1997a) was that the gait speed was significantly higher in the age matched control group, implying that gait variability can not be directly related to motor control performance, but may also be subject to gait speed.

2.3.2 Dynamic Variability

Gait unsteadiness as measured by dynamic variability refers to a quantitative value that characterizes the temporal order of the gait parameter fluctuations. Dingwell et al. (2001) defined a measure of 'local dynamic stability' by applying non-linear time-series methods. Different to DFA, the 'local dynamic stability' quantifies spatio-temporal fluctuations within the data by using state-space reconstruction techniques. This method proposes to measure the sensitivity of a system to infinitesimally small perturbations. These perturbations are a sensitive measure of the neuromuscular control system. Hence, where any dysfunction may predominate within a person, this measure should detect its occurrence.

Dingwell, Cusumano, Cavanagh and Sternad (2000) demonstrated that 'local dynamic stability' of the upper body is improved by slower walking speeds in neuropathic patients. To a lesser degree, limb joint movements showed trends of greater dynamic stability in comparison to the age-weight matched control group. Using the same group of subjects Dingwell and Cavanagh (2001) ??? (new ref) found that reduced walking speed rather than a reduction in motor-sensory function corresponded with increased 'descriptive variability'.
In a prospective multifactorial study design, Hausdorff, Nelson, Kaliton, Layne, Bernstein, Nuernberger and Faitarone-Singh (2001), measured gait unsteadiness and its potential reversibility in functionally impaired older adults using both aspects of variability: (i) descriptive variability (e.g. SD and CV) and (ii) dynamic variability. Hausdorff et al. (2001) described a form of dynamic variability as fluctuation dynamics. This was measured by the 'non-stationary index' (NSI), and the 'inconsistency of the variance' (IV). An alternative

method to the fluctuation magnitude represents a sliding window approach to understanding how the data changes dynamically. The method is applied by dividing the time series data into equal sized windows and calculating the mean and standard deviation within each window. The standard deviation of the summed (i) window means, and (ii) window standard deviations, represents NSI and the IV respectively. Time interval variability and the inconsistency of the variance formed a 'composite instability index'. The 'composite instability index' may be likened to the general term proposed earlier of gait unsteadiness.

In support of variability measures and its multifactorial aetiology, increased stride time SD was associated with decreased functional status, decreased health related quality of life, reduced neuropsychological status, decreased physiological capacity, lower physical activity levels and lower health status. Do people with these associated problems walk slower, which contributes to increased variability? A measure of habitual gait speed was not independently associated with stride variability after multivariate analysis, however, average stride time was significantly related. The measure of stride was higher in those subjects who had fallen within 12 months prior to baseline tests.

The 'composite instability index' demonstrated that gait unsteadiness is multifactorial. Importantly, gait unsteadiness as measured by the 'composite instability index' is reversible after an exercise intervention program (Hausdorff et al., 2002). This indicates the potential importance of dynamic stability measures for assessment of prevention programs, which seek to improve functional ability through exercise interventions. Again, it is noted that gait speed was not found to be a predictor of the 'composite instability index'.

In other studies into aging, Hausdorff et al. (1997a) utilized a method (detrended fluctuation analysis, DFA) to quantify the temporal structure of stride-to-stride intervals. The method has shown that aging and aging associated diseases demonstrate that there is a breakdown in the structuring of the correlations over time. Functional impairment is evident in uncorrelated data. Hausdorff et al. has demonstrated this theory among different functional groups, such as pathological (Parkinson's disease, Huntington's disease and amyotropic lateral sclerosis), young and elderly adults (Hausdorff, Purdon, Peng, Ladin, Wei, and Goldberger, 1996; Hausdorff et al., 1997a; Hausdorff et al., 2000). Although the method has been applied extensively to walking for many groups, it has not yet been applied to an elderly fallers group.

2.3.3 Variability in motorised treadmill walking

Treadmill walking and its affect on variability of gait parameters show conflicting reports in the literature. Matsas, Taylor and McBurney (2000) found that after 6 minutes of treadmill walking, kinematic gait parameters resembled patterns similar to overground walking. The authors did not focus on variability differences between walking mediums, however, the graphed results seem to imply that knee joint angle standard deviation appears similar to overground walking standard deviation. In contrast, other reports indicate that treadmill effects on the walking pattern minimise the moment-to-moment variability of certain kinematic gait parameters (Dingwell et al., 2001). Alton, Baldey, Caplan and Morrissey (1998) noted a trend in greater hip range of motion, such that hip flexion and extension is higher in treadmill walking. A significant difference was found for increased cadence and decreased stance time in treadmill walking. Interestingly, the authors did not find a difference in stride length. These findings on the effects of treadmill walking suggests that walking on a treadmill requires familiarity, which after several minutes the means and variability of gait parameters begins to represent normal overground walking patterns. However, the effect on moment-to-moment variability of gait parameters suggests a certain constraint placed on the locomotor control system (Dingwell et al., 2001).

2.4 LOCOMOTOR SYSTEM

Dynamic stability of gait may be defined as maintaining control over unintended perturbations (Clarke, 1995; Dingwell et al., 2000). However, perturbations are an inherent part of the locomotor system. Clarke (1995) describes a highly functional locomotor system as one that is attracted towards a stable state, such that the locomotor system quickly responds to a perturbation. When walking, a highly functional locomotor system responds to externally influenced perturbations, such as a trip or when avoiding an obstacle by efficiently restabilising the walking pattern. A dysfunctional locomotor system is expected to take longer to return to a stable walking pattern because it will find it more difficult to resist these perturbations. The stability of the walking pattern is constrained by mechanisms operating upon and within the locomotor system.

This section explores the complex neurophysiological system of locomotion, investigating the augmenting of mechanisms that lead to gait unsteadiness. These mechanisms are illustrated in the following diagram (Figure 2.4.1.1). The diagram shows the locomotor system is largely made up of the central nervous system, and the neuromusculoskeletal system. The intended gait pattern is determined by interactions among these systems and the influence of postural control and joint range of motion. Stability of gait is subject to other mechanisms such as fear of falling, neural conduction and afferent information.

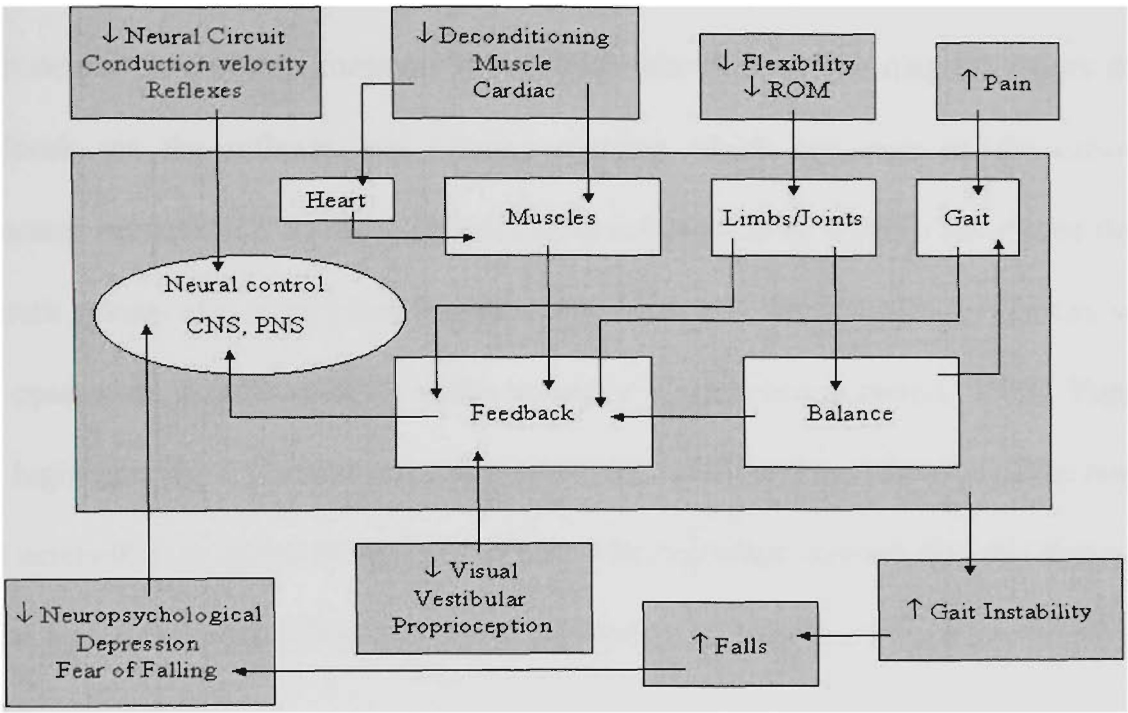

Figure 2.4.1.1. The block diagram summarises the factors associated with the physiological and neuropsychological factors that are associated with gait unsteadiness and falls. (Source: Hausdorff et al., 2001).

The central and peripheral nervous system (CNS) generates the gait pattern, appropriate propulsive forces, modulates changes in centre of gravity, coordinates multi-limb trajectories, adapts to changing conditions and changing joint positions, coordinates visual, auditory, vestibular and peripheral afferent information, and accounts for the viscoelastic properties of muscles. It completes this task within milliseconds and often in conjunction with a multitude of other bodily functions (Leonard, 1995).

Mechanical and neurological factors reduce complexity of the gait control process (Shea, Shebilske and Worchel, 1993). Once walking has been initiated, the swing phase of walking can be achieved without 'higher-level' control (Shea et al., 1993). That is the system can rely upon the mechanical properties of the leg, gravity, and neurological reflexes. This allows us to carry out walking without much physical or conscious effort. Figure 2.4.1.2 illustrates the neural mechanisms involved in the processes of locomotion. Figure 2.4.1.2 (A) illustrates the main neural components common to the locomotor system. The diagram shows the afferent feedback via the reflexes and sensory systems which converge on the central pattern generator network (CPG). The CPG uses this information to regulate locomotor rhythm. The afferent system also copies this feedback information to the higher brain centres, which does not operate on this information unless required to intervene (Leonard, 1995). Figure 2.4.1.2 (B) highlights the CPG network arranged at the spinal level and the role of the basal ganglia and cerebellum in motor control regulation. The important concept that this figure illustrates is the operation of the CPGs to provide necessary and situationally appropriate efferent drive to the motor units.

Gait rhythm is believed to be sustained by central pattern generators (CPGs) and the supra spinal system serves as the initiator and driver. The corticospinal tract (CST) may be responsible for integrating with the CPGs to fractionate the gait pattern and allow it to adapt to changing afferent conditions (Leonard, 1995). Afferent feedback, in the form of the kinesthetic and vestibular systems, detects perturbations in biomechanical information, such as in joint and muscle movements, which then provide fast corrective responses through mono and/or polysynaptic reflexes to supra spinal centres and also to corticospinal regions (CPGs). The CPGs drive the rhythmic nature of walking, however, continually responding to afferent input and collaboration with supraspinal input for shaping the final gait output. Hausdorff et al. (1995) suggested that the gait output rhythm is possibly being influenced by memory process, which is evident by the presence of long-range correlations. The mathematical explanation for this inherent memory is linked to the operational interaction of CPGs (Hausdorff et al., 1995). This operation is being constantly modified by afferent feedback acting as inputs on frequency modes within the CPG network (Schmidt and Lee, 1999). The CPGs may act like a filter to the influencing input information to produce power law, scale invariant fluctuations.

Pinpointing the neural pathways responsible for making instantaneous and ongoing corrections during spatial coordination tasks is a difficult challenge facing neuroscientists (Leonard, 1995). Diminished kinesthetic feedback, which can occur because of aging processes (Lord et al., 2001), will reduce spatial control over natural limb movements such as the foot trajectory during gait (Gandevia and Burke, 1992). This lack of control is characterised by infinitesimally small perturbations occurring from one stride to the next.

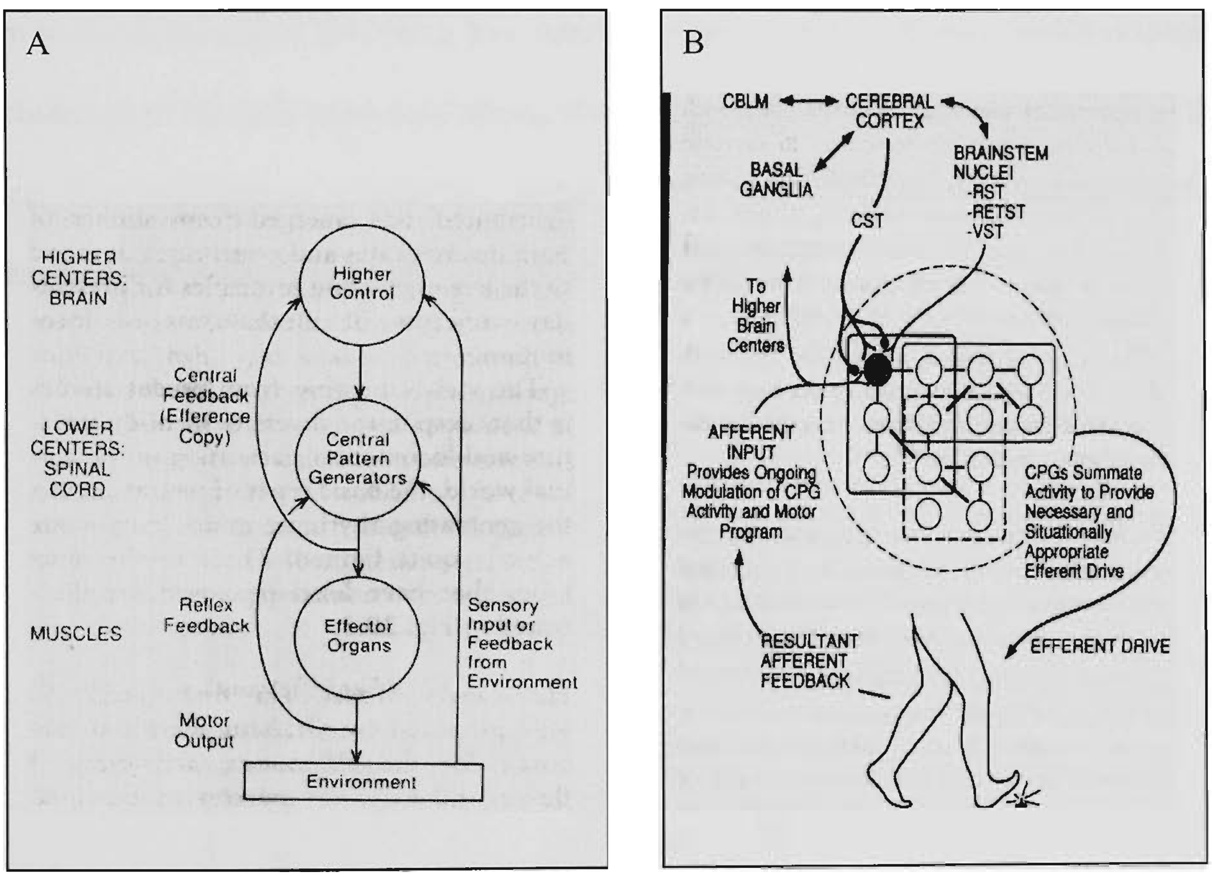

Figure 2.4.1.2. The process of feedback control during walking and the generation of walking rhythm. The diagram (A) provides a general view of the major systems. (Source: Sheppard, 1994).
Diagram (B) highlights the higher centres role in augmenting and controlling walking rhythm (Source: Leonard, 1995).

2.5 FOOT TRAJECTORY

2.5.1 Description of the minimum foot clearance (MFC) event

Foot trajectory refers to the flight of the foot during the swing phase, from toe-off until heel-contact, constituting approximately 40% of the gait cycle (see Figure 2.5.1.1).

Foot trajectory is task oriented, to reposition the foot in the direction of travel while avoiding ground contact prior to footfall. There are seven segments involved, constituting a link system with 12 major angular degrees of freedom (Winter, 1992). The control of foot trajectory begins at the foot during stance and proceeds up to the hip, across the pelvis, and down to the distal end of the swing foot segment (Figure 2.5.1.2). Winter (1992) emphasises the challenge of the task when considering the large number of muscles crossing those joints. Hence, foot trajectory is a sensitive output of the locomotor system, holding important information within the framework linked with aging and falls.

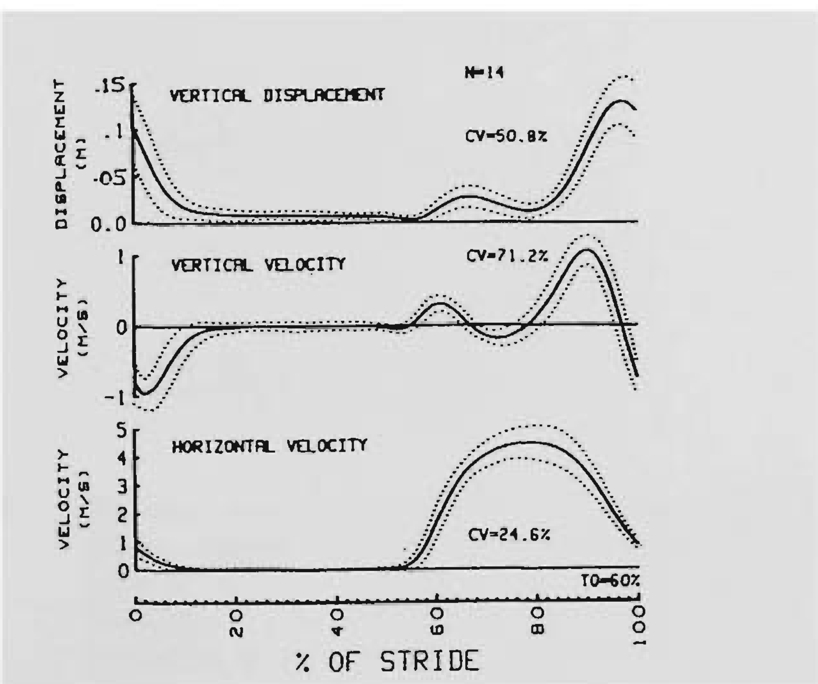

Figure 2.5.1.1. Kinematic diagram of the foot trajectory during the swing phase. The onset of the swing phase begins at 60% of the stride time. The dotted lines represent the amount of variability reported as a percentage of the total gait cycle. (Source: Winter, 1991).

Foot trajectory has become a contemporary focus of gait analysis (e.g. Winter et al., 1990; Karst, Hageman, Jones and Bunner, 1999; Best, Begg, Ball and James, 2000). Occurring at mid swing, the minimum foot clearance (MFC) event is defined by Winter (1991) as the relative vertical distance between two minimum points identified by the two lowest positions of the toe marker at separate events of the gait cycle: toe off and the mid swing. MFC typically occurs at approximately 75% of the gait cycle (Winter, 1990). An alternate method to describe MFC is to define a point located on the 'out-sole' surface of the shoe (Best, Begg and James, 1999; Starzell and Cavanagh, 1999). Dingwell et al. (1999) define MFC as the lowest position of the 5th metatarsal head at late swing (70-100% of stride) in relation to its lowest stance position (1-40% of stride).

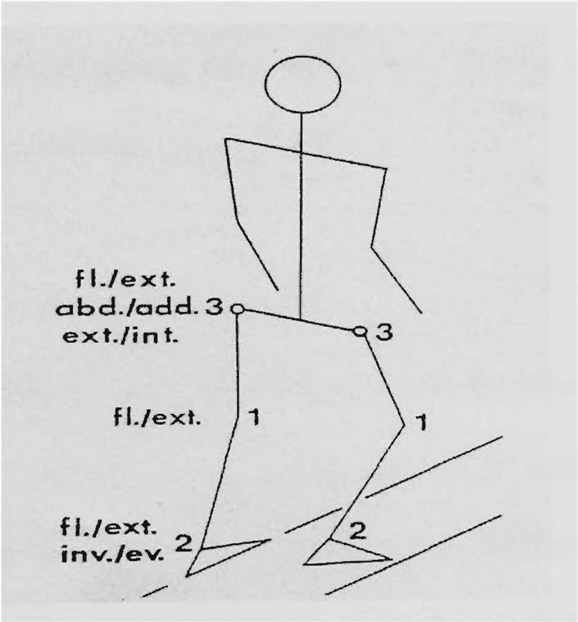

Figure 2.5.1.2 The link system chain during the swing phase of gait. At each joint a number represents the degrees of freedom allowing joint to rotation. The MFC is controlled equally by the interaction of all the joints in the seven segment chain. Definition of terms: fl./ext.: flexion and extension; abd./add.: abduction and adduction; ext./int.: external and internal rotation; inv./ev.: inversion and eversion. (Source: Winter, 1991).

Overground walking studies report small distances in the MFC for both healthy young and healthy elderly groups. By adopting similar methods, Winter (1992) (young group) and Karst et al. (1999) (elderly group) have recorded similar mean MFC values of 1.29cm, and in comparison, Whittle (1993) reported a value of 1.4cm for a young adult group. Karst et al. also demonstrated that while the mean value of the MFC decreased by 0.14 cm during faster walking, the standard deviation increased from ±0.68cm to ±0.73cm. In comparison, Winter (1992), reported a lower standard deviation (±0.4cm). Dingwell et al. (1999) studied kinematic profiles of the foot trajectory for three elderly groups on a treadmill. The elderly control group had a lower variability (SD±0.26cm) and a higher MFC value (0.93cm) when compared to age-weight matched diabetics without neuropathological signs (NNP = 0.90±0.29cm) and diabetics with significant neuropathy (NP = 0.82±0.32cm). Although no significance was reported, a clear trend existed between the all three groups, suggesting that MFC variability increases with sensory loss.

2.5.2 Postural stability during MFC

The MFC event appears to be important based upon the relationship with the body centre of mass and the base of support. During single support the body is unstable because the centre of mass travels along the medial border of the stance foot and not within the base of support (Prince et al., 1997). The hip musculature is needed to control the upper body to avoid tilting. Hence, the spatial trajectory of the foot is related to postural support control of the supporting limb.

Gehlsen and Whaley (1990) found that 66% of falls reported in their study was caused by either tripping or slipping. MFC and the heel contact velocity (HCV) are theoretically recognised for influencing falls (Winter, 1992). The diagrams in Figure 2.5.2.1 (A) and (B) describe the body centre of mass (BCM) in relationship to the swing limb. From the diagram 2.5.2.1 (A), the positioning of the BCM is forward of the stance foot at MFC. The swing foot at MFC is at its maximal horizontal velocity. If contact is made at this point there is a risk of a fall because the swing limb, having been interrupted, will be unable to recover and 'catch' the BCM. Similarly, diagram 2.5.2.1 (B) shows the BCM is behind the swing foot at heel contact. If poor surface friction between the shoe and the ground is added to a high forward velocity of the heel at heel contact, a risk of slipping and falling backwards is increased. In a comparison of two groups, young and old, Winter (1992) found for both groups that the heel contact velocity CV throughout foot trajectory was lowest during the last 10 per cent of the swing phase, suggesting a high level of control at this critical event (see Figure 2.5.1.1). Additionally, elderly populations have lower CV values, showing greater homogeneity in gait compared with younger adults (Winter, 1991). Issues of overuse are associated with 'grooved' motor patterns and may suggest this is not desirable. However, high variability during MFC is also undesirable and is likely to result in tripping.

This dissertation focuses on the MFC event because tripping, as compared to slipping, is a more prevalent cause of falling in the community. Also, the fine control required at the MFC event is sensitive to the locomotor systems functioning; and the interdependent nature of the MFC is under the multi-limb control of both the stance and swing limbs. In view of other findings from stride-to-stride variability measures of gait parameters, especially those indicated by Hausdorff et al. (1997b), the consequences associated with similarly occurring variability witnessed in MFC has strong implications for trip related falls.

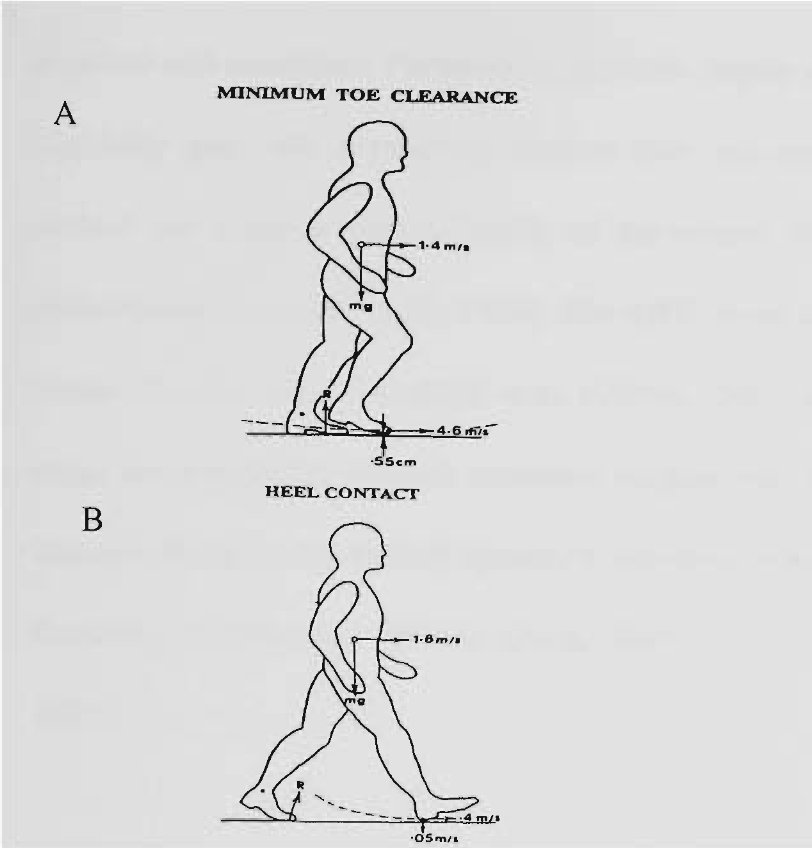

Figure 2.5.2.1 (A) the MFC, and (B) horizontal heel velocity at heel contact. (Source: Winter, 1992).

As indicated earlier, MFC is a 'many degrees of freedom' problem confronted by the locomotor system. The MFC phase of foot trajectory is sensitive to at least 6 muscle groups (Winter, 1992). Bernstein (1967) proposed that the body addresses the 'many degrees of freedom' problem by freezing the relevant body parts into functional units, termed synergies. Bernstein proposed that motor coordination and control, such as the minimum foot clearance event, is a product of a self-regulating 'system' of mechanisms that permits flexibility. According to Swinnen, Massion and Heuer (1994), learning new tasks that requires control of many degrees of freedom is affected by preexisting coordination modes. In gait, Hausdorff et al. (1995) proposes that there can be many different modes linked to central pattern generators, which randomly switch during rhythmic walking. Once a task, like gait, has been well learned (e.g. following early childhood), the degrees of freedom become controlled, invariant and consistent. Fortunately, however, highly developed and automated motor skills, especially gait, still manage to express both movement consistency and flexibility. This distinct but complementary feature of movement allows adaptability within a changing environment (Swinnen et al., 1994). The MFC event requires these features to negotiate the terrain. Results from Hausdorff et al. (1997a; 2000) demonstrating the breakdown of long-range correlations in the gait processes suggest that the elderly begin to lose the adaptive features of the motor control system. Long-range correlations are believed to represent the flexibility in biological systems (Peng, Mietus, Hausdorff, Havlin, Stanley and Goldberger, 1993).

2.6 INTER-LIMB COORDINATION

The study of the principles governing inter-limb coordination is an evolving discipline of behavioural science. It is a remarkable spatio-temporal feat when coordinating various limb and body parts - in spite of the large differences in inherent characteristics of the effectors and multiple degrees of freedom involved - to achieve smooth, effortless, multiple tasks simultaneously (Swinnen and Carson, 2002). Understanding the principles of coordination deserves attention when the objective is to maximise functional independence of disordered populations (Swinnen and Carson, 2002).

2.6.1 Gait symmetry

Historically, gait symmetry was assumed in the literature for the sake of simplicity in data collection and analysis (Sadeghi et al., 2000). The review paper by Sadeghi et al. (2000) challenges the concepts of gait symmetry by exploring research that confirms the presence of a dominant limb preferred by the body when carrying out certain tasks (e.g. Peters, 1988;

Hart and Gabbard, 1998). Therefore, Sadeghi et al. (2000) reasonably deduces that a dominant limb must be a constraint upon achieving gait symmetry.

The primary goal of walking can be divided into two tasks; stabilising and propelling the BCM. A further necessary task is negotiating the terrain with adequate clearance of the ground during forward progression of the swing foot. This task may be a combined subtask of the primary goals (Peters, 1988). In effect negotiating the terrain occurs during both stability and progression. The right limb leads forwards while the left limb stabilizes, which can be termed the right link system (RLS) for simplicity. The next sequence being, left limb leads while being supported by the right limb, alternatively this can be termed the left link system (LLS). The segment link system was explained in further detail in section 2.5.1. The function of these two alternate link systems (RLS and LLS) is phase shifted by half a gait cycle (see Figure 2.2.2.1), which forms the bilateral locomotor task of walking. The separateness of the two link systems may be conceptualised as independent synergies, each one responsible for organizing their respective left and right link systems. A further concept is the presence of a higher system that seeks to entrain the two link systems, so as they may collaborate towards a functional coordinating goal. Is this collaboration more inclined to be temporal, or can it exist in spatial features also?

Is stability between the link systems equal? Is propulsion between the link systems equal? If one link system is functionally greater, what does this mean for the global operation of walking? Can the minimum foot clearance parameter represent both stability and propulsion? Do they both negotiate the terrain with an equal performance? In general, are the left and right systems equal, in other words symmetric? How do we measure, define and describe their equality? Is it bad if they are not equal? The answers to these questions will be addressed throughout sections of this dissertation.

Stabilising perturbations appears to be a combined task carried out by the support limb during weight transitions and the mobilisation limb when performing its primary role of adapting to the necessary spatial orientation during foot trajectory. Combined, these actions will perform optimally for the dominant link system, however, the non-dominant link system's performance may not be of the same quality. How this affects bilateral symmetry of foot trajectory is unclear, however, given the different functioning roles of the limbs (Peters, 1988), it makes sense to assume that it would be asymmetric. Research is limited to unilateral methodology and no current literature can shed insight into bilateral performance of the MFC.

It is believed that force-coding proprioceptors in the stance limb organise synergetic reflex responses to the opposing swing limb (Cattaert, 1994). This reflex is another form of the many types of afferent information that interacts with and modifies the operation of the central pattern generator (see Figure 2.6.1.1). This demonstrates that inter-leg coordination of spatial trajectory of the foot is provided by the loading response in the stance limb. This has implications for the safe walking over uneven terrain. The following section explores the relationship between spatio-temporal patterns of foot trajectory being constrained at: (i) the inter-limb level by the coupled oscillator system, such that the left step influences the right forming a collective behaviour; given that (ii) at the intra-limb level - as previously discussed - is related to long-range correlations of stride time intervals (Hausdorff et al., 1995). The issue of how these constraints apply spatially is also explored in this dissertation.

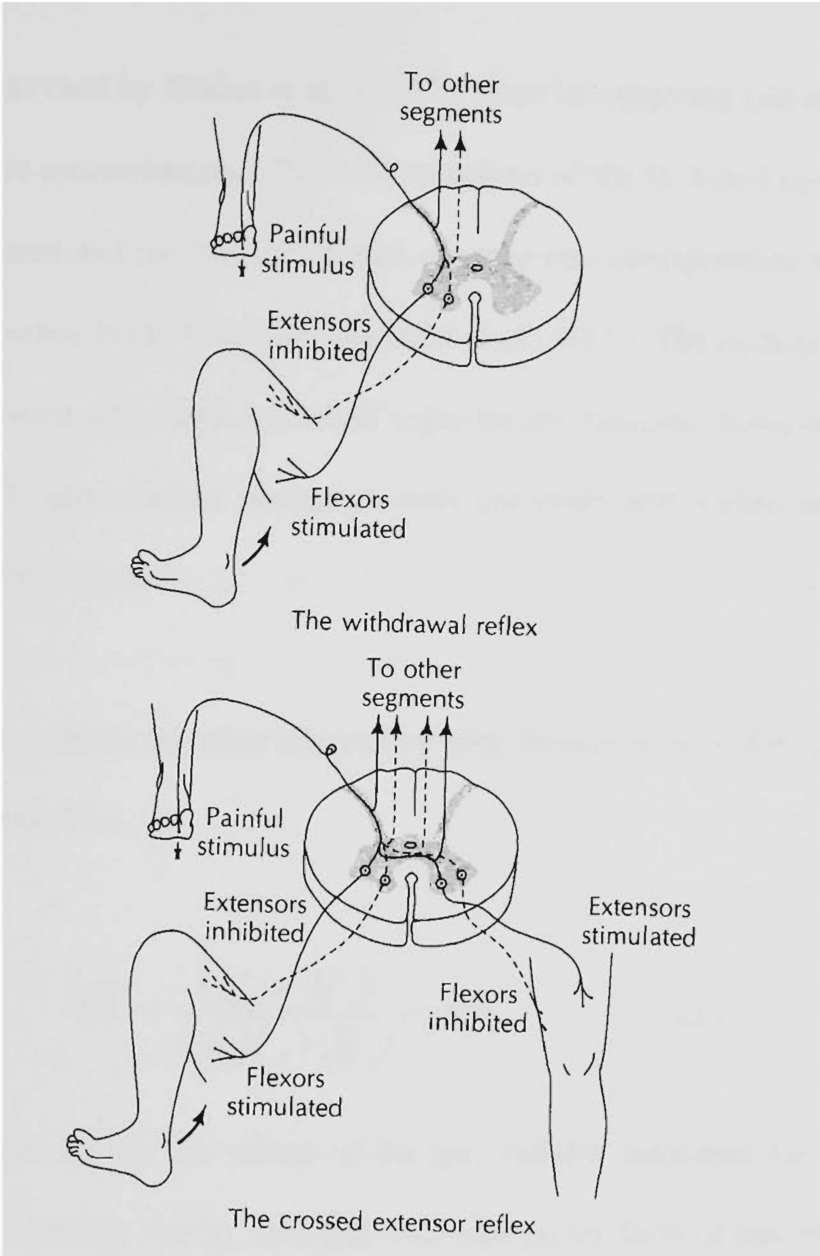

Figure 2.6.1.1 The withdrawal and crossed extensor reflex highlighting the interdependent neural exchange between the lower limbs. (Adapted from: Shea et al., 1993).

Different statistical equations and methods have been constructed to define the presence of asymmetry, such as t-tests (Allard, Lachance, Aissaoui and Duhaime, 1996), multi analysis of variance (MANOVA) designs (Gunderson et al., 1989), and symmetry index (SI) equations (Herzog, Nigg, Read and Olsson, 1989; Becker Rosenbaum, Kriese, Gerngrob and Claes, 1995; Giakas et al., 1997). The various measures of describing the symmetry has possibly led to mixed reports of its presence. A symmetry index (SI; see equation 2.1) value of greater than 10% was used by Giakas et al., (1997), when investigating gait asymmetry in the ground reaction force measurements. There are criticisms of the SI, based upon the size of the means being compared and the fact that it does not take into consideration of the standard deviation of the parameters being measured (Sadeghi et al., 2000). The authors of this critique suggest calculating symmetry from statistical significance methods, however, this may be fraught with 'type-1' errors when numerous tests are conducted within subjects and among the parameters investigated.

The symmetry index equation adopted by both Becker et al. (1995) and Giakis et al. (1997) can be expressed as:

$$\mathrm{SI} = \frac{(X_R - X_L)}{0.5(X_R + X_L)} \times 100\% \qquad (2.1)$$

where X_R and X_L are the values of the gait variable measured for the right and left limb respectively. The X_R and X_L variables may take on the form of any common parameter. A SI equal to zero indicates perfect symmetry (i.e. $X_R = X_L$). The limitation of this equation is due to the ratio of the difference between limbs relative to the average of the combined values. If the measured X_R and X_L values are large but show a relatively small difference, the SI will decrease towards zero and indicate symmetry. Also, the relative left and right values represent a summed parameter, which may be misleading for moment-to-moment dynamics. The benefit of the symmetry index is that it is a within subject comparison, such that it is a measure of how a person's motor control system adapts and responds relative to its own internal reference system which may provide useful insight rather than comparing the left and right limb parameters against a group sample.

2.6.2 Dynamic theory of inter-limb coordination

Using a dynamical systems approach, theoretical principles of interlimb coordination have been established from a diverse range of experiments in animal and human behaviour. The basis to the principles can be attributed to the work conducted by the German behavioural physiologist E. von Holst. His observations began with fin movements of the *Labrus* fish and are applied as a general theory for inter-limb coordination dynamics (Turvey et al., 1993). The theory states that inter-limb coordination involves two inter-active and competing bimanual control components: relative coordination (docility to be attracted out of a fixed phase, or toward a fixed phase) and absolute coordination (maintaining fixed phase) (Turvey et al., 1993; Kelso, 1995).

2.6.3 Principles of bimanual coordination

The concept of relative phase competition and cooperation dynamics has inspired the development of a mathematical model for describing inter-limb coordination dynamics (Haken, Kelso and Bunz; the H-K-B model; Kelso, 1995). The model was originally developed for investigating self-sustained oscillations in non-biological phenomena (Amazeen, Amazeen and Turvey, 1998). Modifications of the H-K-B model have allowed researchers to explore and derive principles underlying inter-limb dynamics. The principles developed from the H-K-B model are important for understanding the nature of stability when achieving symmetrical behaviour between two limbs.

The H-K-B model was developed from low-dimensional non-linear dynamics into a form that is now highly dimensional and allows for stochastic and deterministic processes commonly encountered by researchers investigating inter-limb coordination (Beek, Peper and Daffertshofer, 2002). The concepts of the H-K-B model have implications for the investigation of inter-limb coordination in gait (Kelso, 1995), such as the stochastic and deterministic parameters that form the components of the model. However, limitations of the H-K-B model and its current developments have been expressed in terms of its suitability to gait (Reik and Carson, 2001). The experimental conditions from which the H-K-B model is derived and tested is without factors associated with balance and gravity, limiting its application to gait.

The literature on spatial dynamics of inter-limb coordination (e.g. bilateral MFC's) is extremely limited, as most experimental work has investigated inter-limb relations at the temporal level. Applying and testing spatial models that describe inter-leg coordination dynamics are required. Nevertheless, the H-K-B model as applied to inter-limb experimental work has provided rich insight for (i) identifying the important parameters at the spatial level and (ii) revealing key principles governing inter-limb coordination.

2.6.4 Applications of bimanual coordination principles to spatial gait coupling

Gait requires inter-limb coordination of various body parts and limb segments on a rhythmical basis. Applying the principles of inter-limb coordination to the spatial nature of foot trajectory raises several hypotheses. Symmetry is influenced by two competing conditions, one of flexibility and the second, stability (Turvey et al., 1993). The outcome of these competing conditions is a fluctuating process about an intended symmetry. These asymmetric fluctuations are characterised in Figure 2.6.4.1. Although the diagram reveals how the left and right limbs fluctuate in relative phase (right > left; right = left; right < left where the intended symmetry is when right = left), the concept may reveal how inter-limb MFC fluctuations (as derived from left and right MFC events) are related to intra-limb MFC events. It is expected that variability in fluctuations will increase with task complexity

(Turvey et al., 1993). Inter-limb instability has been related to postural stability (Whitall and Clark, 1994).

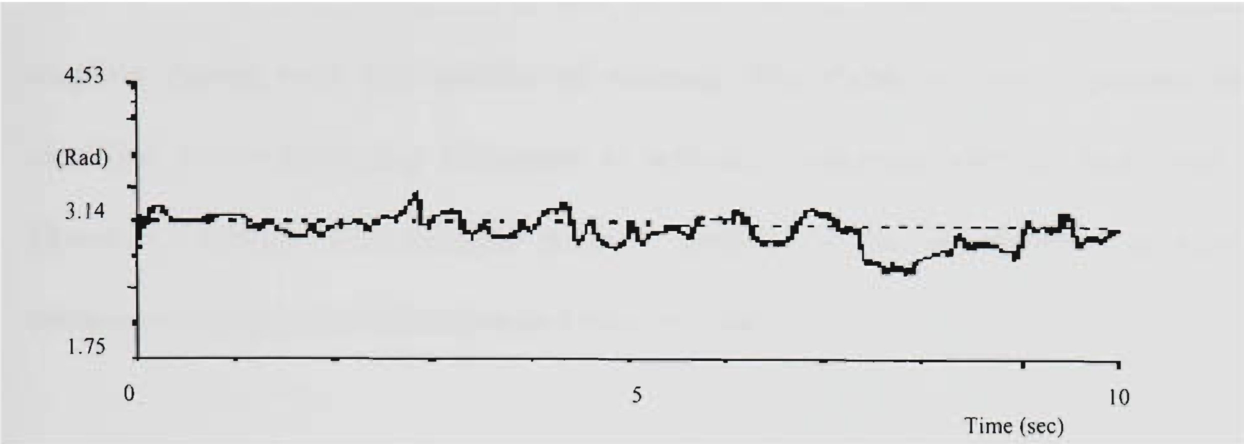

Figure 2.6.4.1 Normal fluctuations of relative phase from bimanual arm movements. The fluctuations occur about an intended anti-phase symmetry relationship (displaced in time by 180°). Note the slow drifting pattern in the line graph. (Source: Turvey et al., 1993).

2.6.5 Contradictions of spatial coupling models

Most of the experimental design tasks involve simultaneous motion of the upper limbs, of far less complexity in comparison to walking. Bipedal gait coordination involves sequential loading and unloading, therefore unique experimental designs are required to investigate principles derived from the H-K-B model. One such experimental design has recently shown that the symmetric organization in walking may not be primarily due to spatio-temporal constraints, but determined by force loading receptor mechanisms (Reik and Carson, 2001). This finding contradicts symmetry group theory models, suggesting that bipedal gait patterns have evolved in such a way that the collective behaviour of the limbs is constrained also, by load response mechanisms. Further, reports suggest that the swing phase is not a passive movement but is under neural control (Whittlesey, van Emmerik and Hamill, 2000). If it were a passive movement, it would be expected that inter-limb coordination would show instability in spatial trajectories of the foot.

2.6.6 Entrainment and coupling strength

Entrainment is the ability of one limb to match the pattern of the other limb. It has been shown (Whitall and Clark, 1994) that infants display weak entrainment, or inter-limb coupling during their first months of walking. The ability to control posture has been identified as a contributing subsystem to refining entrainment (Whitall and Clark, 1994). Therefore, persons with reduced postural control can be hypothesized to have weak entrainment leading towards between limbs asymmetry.

Turvey et al., (1993) explores the notion that inter-limb coordination involves the interplay of three levels. These three levels relate to: (i) the physical properties and inherent tendencies of the coordinating limbs; (ii) the relative inter-limb coordinating pattern performed; and, (iii) the intentional pattern (e.g. the timing goal of gait is to coordinate bilateral events that are phase shifted by half a cycle; see Figure 2.2.2.1). Catteart, Barthe, and Clarac (1994) indicate two levels of CPG networks responsible for inter-and intra-limb control within the crayfish locomotor system. At one level, there exists a CPG network that is responsible for coordinating intra-limb processes. A higher level CPG network has the role of coordinating the independent intra-limb CPG networks. These concepts are derived from empirical research not related to human gait, however, they imply that in walking, a possible mechanism exists that seeks to entrain the limbs.

2.7 NON-LINEAR DYNAMICS

This section reviews non-linear methods that have been applied to variability patterns found within biological processes. These methods will be explained in this section and will be examined in view of their potential for assessing the processes of minimum foot clearance.

2.7.1 Issues in describing locomotor control

In the previous sections, references were made to long-range correlations and non-linear processes within gait parameters, which highlights the spurious aspects of assuming linearity in gait data. Dingwell et al. (1999) summed up the limitation of traditional linear methods by expressing that using traditional measures of variability, important stride-to-stride information is lost. This is due to normalising deviations over multiple gait cycles, such as when calculating the standard deviation and coefficient of variation (CV). This is supported by heart rate variability research showing that patients with clinically different problems can demonstrate the same standard deviation (Peng et al., 1993). However, when viewing 'beat-to-beat' time series graphs a marked difference exists between patients (Goldberger et al., 2000). Also, there is evidence from various bimanual tasks to suggest that inter-limb interactions are nonlinear (Kelso, 1995). A non-linear set of data requires non-linear methods of analysis. The general message from these authors suggests that traditional methods of variability based upon linearity, stationarity and equilibrium may miss important information and it is necessary to consider newer approaches that describe variability. Dingwell et al. (2001) further states that a thorough understanding of locomotor control requires an understanding of how movements are controlled from one moment to the next over an extended period of time.

Hausdorff et al. (1995) and Dingwell et al. (1999) are two groups who are pioneering the dynamic approach into gait analysis. The choice of the parameters and the methods that they adopt reflect different rationales for capturing gait unsteadiness. Hausdorff's group has been applying DFA, while the group of Dingwell et al. utilise state space reconstruction models (deriving a term known as the Lyapunov exponent). The DFA method takes advantage of interval data from cyclic periods, whereas the Lyapunov exponent is calculated from continuous trajectory data. Both methods take advantage of extended walking periods, commonly at least ten minutes, which involves hundreds of consecutive strides. Other investigations examining variability in gait have been limited up to several consecutive strides (e.g. Maki, 1997; Gabell and Nayak, 1984). Ground reaction force investigations into gait involve even greater constraints for examining consecutive foot falls (e.g. Crowe et al., 1996; Giakis et al., 1997).

Two promising methods used to characterise dynamic states in biological behaviour that can be applied to MFC data come from Hausdorff et al., (1995) and, Brennan et al. (2001). The former developed the DFA to identify and characterise the ordering of time series data in the form of stride-to-stride intervals during gait. The other method, the Poincaré plot, has previously become popular for visually interpreting nonlinear dynamics. Brennan et al. (2001) have been able to draw out the quantitative information from the salient features of the plot from an ellipse fitting technique developed for heart rate intervals. This is a new application for quantifying unsteadiness in gait.

Both methods have an advantage over traditional methods because they are able to describe the fluctuating process underlying the variance in the data. These methods may be beneficial for characterising the instability underlying gait unsteadiness, thus, may strengthen support of cause and effect relationships between predictors of falls, and the occurrence of falls.

2.7.2 Fractal geometry and their breakdown in disease and aging

Fractals represent the formation of geometric structures and patterns found in the complex systems that occur in natural phenomena. Fractal geometry has developed from statistical physics concepts first introduced by Benoit Mandelbrot, an early 20th century mathematician

(Meakin, 1998). Fractals are based upon considering how structures of one length scale relate to those of other length scales by focusing on the common universal properties instead of detailed analysis of the microscopic components (Meakin, 1998). The connection between microscopic and macroscopic components is important as it underlines much of the theoretical applications of the scaling law models used for studying the fractal nature. These models have been used for studying both atomic construction and the shape of coastlines.

Power law relationships are the primary concept behind fractals and scaling. Fractals are the study of shapes and the underlying structure of these shapes exhibits self-similar features at many different magnifications. In very general terms, to determine if a structure is self-similar, a subset of different magnifications of an original time series pattern is rescaled to the original data set. The shape of these rescaled structures or patterns are then compared. A power law function (equation 2.2) demonstrates scale invariance which means that $y(x)$ has the same shape for all scales and the exponent 'a' does not depend upon the units of measurement that characterise x or y. If both sides of the equation are equal, then the original data series is self-similar.

$$y(x) = cx^{a} \tag{2.2}$$

The evidence of power-law scaling is based upon a linear relationship obtained from the logarithmic version of equation 2.3.

$$\log y(x) = \log c + a \log x \tag{2.3}$$

By plotting equation 2.2 over a sufficiently large range of scales provides evidence of power law relations and can be quantified by the scaling exponent 'a' and the amplitude 'c' (Meakin, 1998; see Figure 2.7.2.1). This quantitative measure has proved to be an invaluable tool for describing complex structures in nature that were previously believed to be disordered patterns.

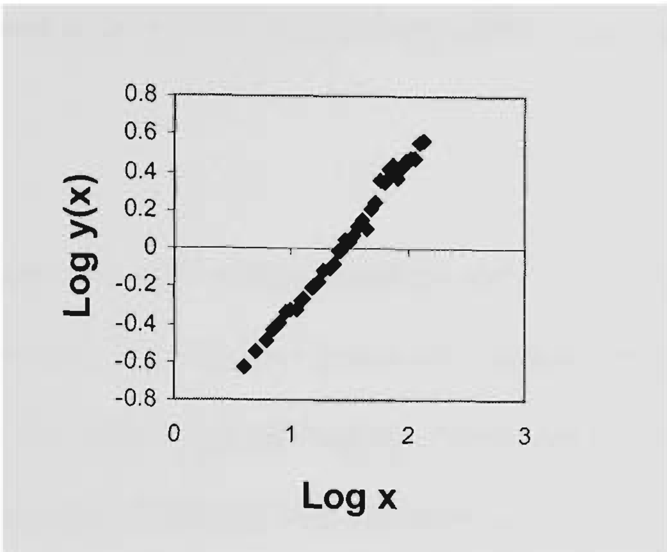

Figure 2.7.2.1 presence of power law scaling. The plot of equation 3: log y(x) = log $c + a$ log x; where a is the gradient of the line on the log-log plot, and c represents the coordinate where the line cuts the y intercept. Or, the gradient 'a' may be represented by $y(x) \cong x^{a}$.

Fractal processes in physiology appear to be evidence of complex scale free systems (many input processes acting on different time scales[1]) that prevent a system from achieving a locked state of equilibrium. In other words, fractals are a sign that a system is flexible to adapt when appropriate (Peng et al., 1993). Long-range correlations, a feature of fractal like processes in physiology, may lack a characteristic scale for the purpose of preventing excessive mode-locking within a self-organised system, a prospect that would restrict the functional responsiveness (plasticity) of the organism (Peng et al., 1993). When fractal

[1] An example of a scale free physiologic system is the heart rate regulation that is affected by semi-autonomous systems operating at disparate time scales. For example, such influences would be related to the respiratory, circadian rhythm, and hormonal system. All these input systems operate at different (multiple) time scales. Within these input system scales, they can (i) undergo random changes, creating a superposition of random inputs, and (ii) take on different amount of influence to the whole system. (Peng et al., 1993)

processes breakdown it signifies that a system has become reliant upon fewer input scales, thus leading to highly periodic (e.g. Brownian noise) behaviour of the output signal. Alternatively fractal breakdown in aging effects in health have also shown random, uncorrelated structure (Hausdorff et al., 2000). Therefore they do not characterise fractal like features.

Healthy gait has been characterised by persistent long-range correlations in the stride-to-stride intervals, decaying in a power law fashion (Hausdorff et al., 1995). The authors have attributed the underlying mechanisms behind long-range correlations to a type of 'memory' effect operating within the neurophysiological control process, across hundreds of strides. Long-range correlations are 'fractal-like' in that they define a self-similar structure across different length scales. Elderly and pathological gait has shown breakdown in long-range correlations of the stride-to-stride interval times, such that output signals (time series) become more 'random-like' and are characterised by 'white noise' fluctuations (Hausdorff et al., 1997a). An important finding by the authors demonstrate that when comparing between two subjects, variability measured by standard deviation of a gait parameter may be almost identical, however, the correlations inherent within the time series fluctuations can be significantly different.

2.7.3 Detrended fluctuation analysis (DFA)

The following describes the process of quantifying the fractal features of a dynamic pattern. Using an adaptation of the equation (2.2 and 2.3) discussed previously (section 2.7.2), a self-similarity scaling parameter (α) is computed from the intra-limb MFC time series data. The model involves several steps and is sometimes referred to as a root mean square analysis of a 'random walk' (periodic type fluctuations, i.e. Brownian noise). DFA accounts for those problems commonly associated with 'bounded' time series data in biological processes (i.e. data that is not free to fluctuate arbitrarily but fluctuates in a random manner about a self-similar structure) by mapping it with a 'self-similar process' through integration (Goldberger, Amaral, Glass, Hausdorff, Ivanov, Mark, Mietus, Moody, Peng and Stanley, 2000). Brownian motion is a concept applied to the investigation of fractal-like properties in gait and other biological processes, such as heart rate. In this case, "the random force (noise) acting on particles is bounded, similar to physiologic time series. However, the trajectory (an integration of all previous forces) of the Brownian particle is not bounded and exhibits fractal properties that can be quantified by a self-similarity parameter" (Goldberger et al., 2000). Therefore, the first step requires integration of the MFC. This integration step maps the original MFC time series to a self-similar process (equation 2.4), where p_i is the ith value for a given parameter p (MFC in this case); and, $p_{(ave)}$ is the average of the given parameter value (MFC mean).

$$y(k) = \sum_{i=1}^{k} \left[p_i - p_{(ave)}\right] \qquad (2.4)$$

Because most physiological systems are non-stationary (mean, SD, higher moments are not invariant over time), the integration procedure (equation 2.4) will exaggerate these non-stationary effects. To overcome this, a second equation from root mean square analysis of a random walk is adapted which determines self-similarity of the fluctuations in the integrated time series (y(k)). The vertical characteristic scale of the integrated time series is measured, requiring scaling over a number of different sized windows of equal length (n). Each window size (n) calculates the y coordinates by fitting a least squares line to the data (Figure 2.7.3.1). The y coordinate of the straight line segments is denoted by $y_n(k)$. The integrated time series, y(k), is detrended by subtracting the local trend, $y_n(k)$ in each window. For a given window size (n), the characteristic fluctuation for the integrated and detrended time series is calculated by equation 2.5:

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}[y(k) - y_n(k)]^2} \qquad (2.5)$$

This is repeated over all window sizes (n), creating a relationship between the average fluctuations F(n), and window size (n), where N is the length of the total data set. If the fluctuations F(n) at different windows scale as a power-law with window size n, the integrated time series is self-similar. A linear relationship on a double log graph indicates the presence of self-similarity, where the slope of the gradient relating log F(n) to log n determines the value of the self-similarity parameter, α, where $F(n) \cong n^{\alpha}$ (see Figure 2.7.2.1).

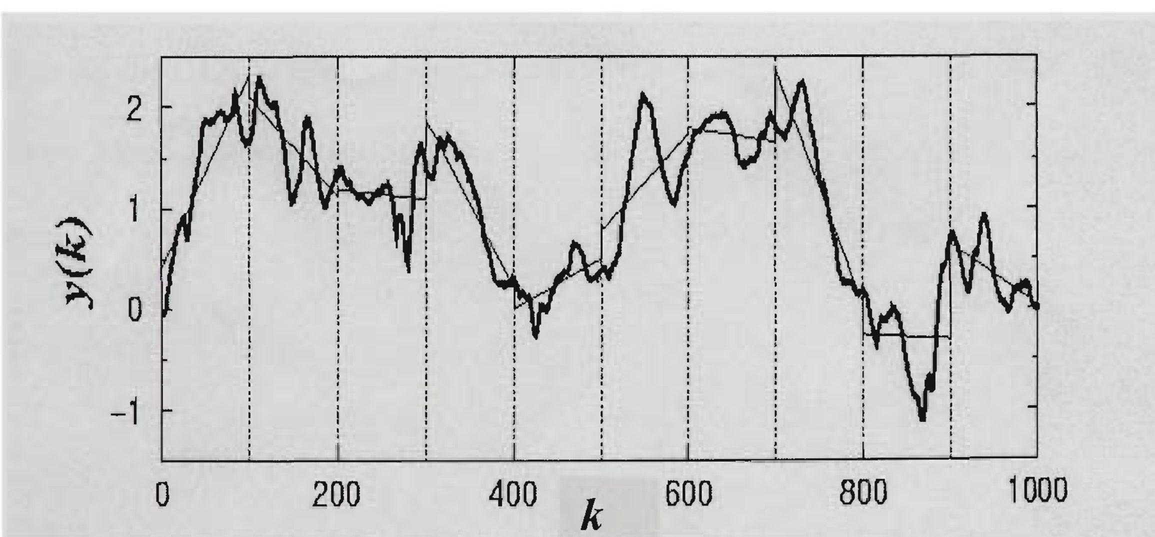

Figure 2.7.3.1 Local detrending of the DFA algorithm. The integrated time series y(k) (equation 1). The vertical dotted lines represent boxes of size n = 100. The solid lines in the box represent the local trend $y_n(k)$. (Source: Golberger et al., 2000).

This method can be appropriately used when there is a sufficient length of data, such as that provided by a ten-minute walk (Peng, Havlin, Stanley and Goldberger, 1995). The method describes the correlations within the fluctuations from one stride interval to the next. A value of 0.75 represents long-range correlations, while a value closer to 0.5 is representative of uncorrelated, random occurrences.

The DFA method is designed to be robust against non-stationary drifts in biological time series data, which other methods have demonstrated some inaccuracies of scaling due to sensitivity of non-stationarities in the data, such as Fast Fourier Transformation and Hurst's exponent analysis methods (Peng et al., 1995). Nevertheless, non-stationary drifting may produce some inconsistencies in the DFA results due to large scale fluctuations when linear detrending is performed. To overcome this likelihood, higher order detrending can be performed, which reduces the magnitude of the large scale fluctuations that non-stationary effects may cause. Figure 2.7.3.2 illustrates the method that has been used by Hausdorff et al. (1996), who found that in stride interval correlations, higher order detrending had almost no effect on the DFA scaling (without detrending, $\alpha = 0.895$; with higher order detrending $\alpha = 0.889$). Thus suggesting that the DFA accounts for any non-stationary drifts inherent in a data series.

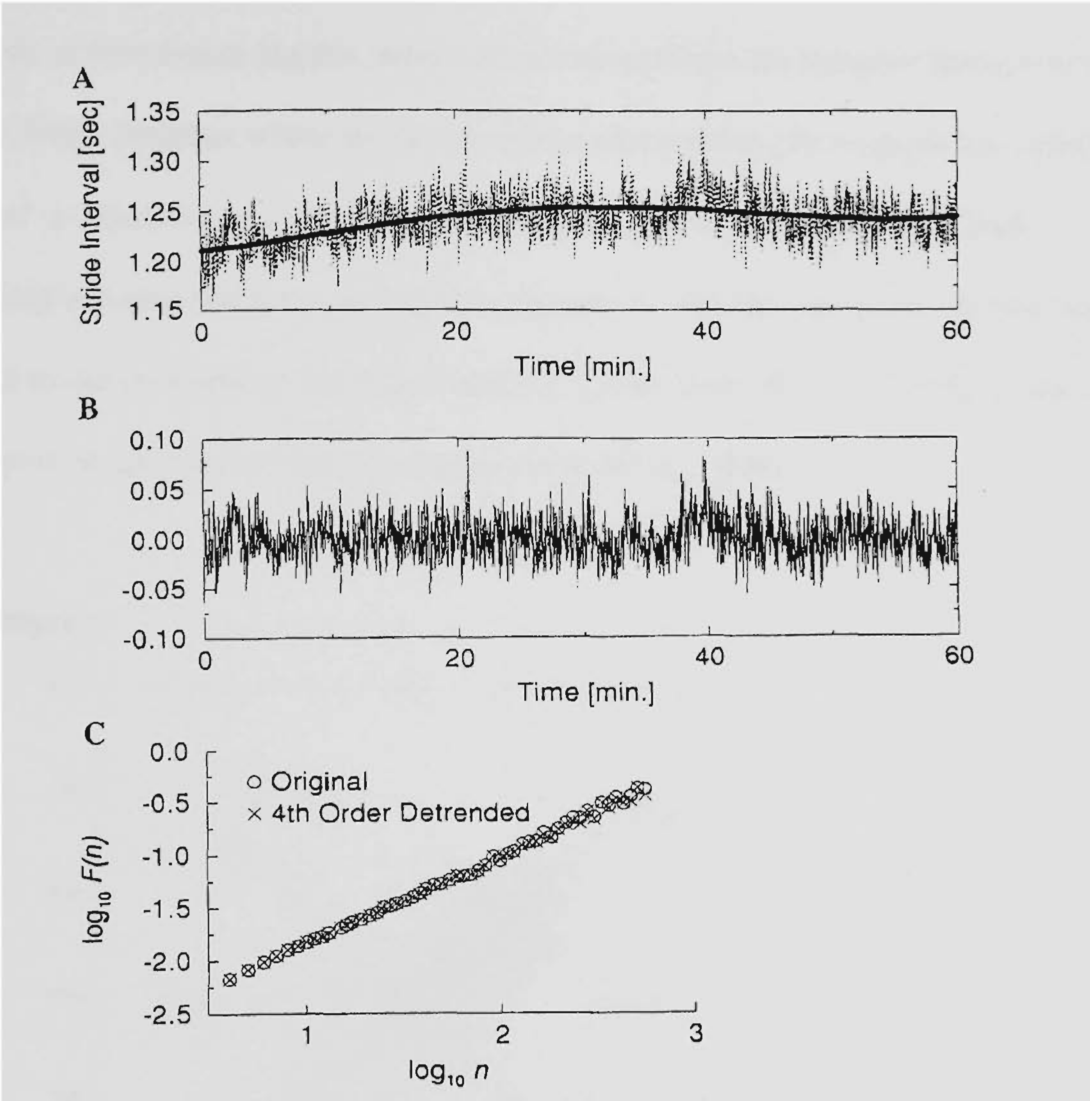

Figure 2.7.3.2. Detrending using higher order polynomials in the DFA. Panel A shows a fourth order polynomial fitted to a non-stationary time series. Panel B illustrates the detrended fluctuations about the 4th order polynomial line fitted to the time series in A. Panel C demonstrates the comparable scaling between fourth order detrending and linear detrending, indicating the robust features of the DFA against non-stationary effects within a data series. (Source: Hausdorff et al., 1996).

2.7.4 Poincaré plot descriptive measures of variability

Figure 2.7.4.1 describes the computation of the variability ratio 'μ' by calculating two measures of variability: SD_1 and SD_2, along two perpendicular axes, X_1 and X_2 (Brennan et al., 2001). The plot is termed the Poincaré plot, named after the French mathematician; Henry Poincaré. Other terms have also been used to describe this type of plot, for example a Poincaré return map (Kelso, 1995). Poincaré used this qualitative map to overcome the limitations of Newtonian physics, which do not account for the dynamic features of a system (Clarke, 1995). Systems whose dynamics can be characterised by a parameter captured at an instant of a repeating cycle can be mapped using the Poincaré plot (Clarke, 1995). A relationship is connected from one point to the next by a certain function, 'G' (the function G is related to the concepts of the H-K-B model), in the form of: $P_{n+1} = G(P_n)$, such that each point is plotted against the next consecutive point (Kelso, 1995).

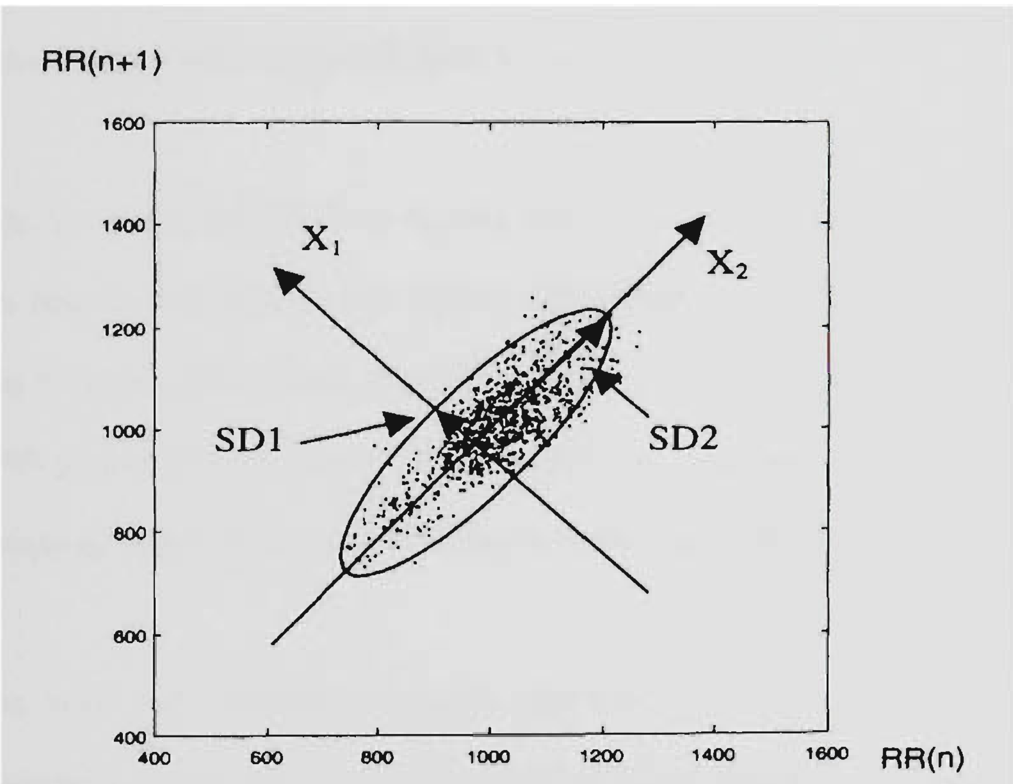

Figure 2.7.4.1. An example of a Poincaré plot detailing the ellipse fitting technique. The standard deviation of the distance of the points from each axis (X_1 and X_2) determines the width (SD1) and Length (SD2) of the ellipse. See text for definitions. (Adapted from Brennan et al. 2001).

The standard deviation (SD_1) about the width of the cloud represents the short-term variability (equation 2.6).

$$SD_1^{\,2} = \tfrac{1}{2}(SDSD^2) \qquad (2.6)$$

The standard deviation (SD_2) measured about the length of the cloud represents the long-term variability (equation 2.7).

$$SD_2^2 = 2SD(RR)^2 - ½(SDSD^2) \qquad (2.7)$$

RR is the time interval between repeating cyclic events, such as heart rate intervals, and SDSD is the standard deviation of the successive differences of the RR intervals. SD(RR) is the standard deviation of the RR interval *z*-scores.

A similar ratio is described by Kamen and Tonkin (1995) and termed the aspect ratio. The ratio described by Kamen and Tonkin differs from Brennan et al. (2001), where the latter maps the points of the cloud to the longitudinal axes (X_2, Figure 2.7.4.1). Kamen and Tonkin (1995) project the points onto the horizontal x-axis, as opposed to the more recent technique of mapping them to an axis of +45 degrees to the horizontal (Brennan et al., 2001).

From heart rate variability research, Kamen et al. (1995) utilised the Poincaré plot to successfully reveal distinctive differences between patients with heart failure and healthy patients. Decreased standard deviation of the data mapped onto the 'x-axis' (approximately similar to SD_2) and SD_1 has demonstrated a significant difference between healthy subjects and those with chronic heart failure. Kamen et al. (1995) also applied nomenclature classification of the qualitative aspects of the clouds to distinguish between heart failure and healthy heart function. This nomenclature is related also to the quantitative measures of variability associated with the plot (see Table 2.7.4.1). A long thin cloud observed from the Poincaré plot is representative of significantly decreased short-term variability in relation to the long-term variability. In contrast a circular shaped plot indicates equality of variability for both short-and long-term distributions.

Table 2.7.4.1. Quantitative guide for Poincaré plot nomenclature used by Kamen and Tonkin (1995). The values in this table relate to the units (msec) from the illustrated plots in figures 2.7.4.2. and 2.7.4.3.

	SD1 Short term variability	SD2 Long term variability	SD2/SD1 Aspect ratio
Cluster- open		> 80 ms	
Cluster- normal	> 10 ms	20 – 80 ms	< 2.2
Cluster- tight		< 20 ms	
Fat cigar	> 10 ms	20 – 80 ms	> 2.2
Cigar	< 10 ms	20 – 80 ms	> 2.2
Comet	A pattern reminiscent of a comet due to a large head and a narrow tail. There are no quantitative guidelines applied by Kamen and Tonkin (1995).		

Figures 2.7.4.2 and 2.7.4.3 show the different shapes in the plots created by various dynamic systems controlling the heart rate, which have been analysed by Kamen et al. (1995). These qualitative aspects of the Poincaré clouds have formed a nomenclature rationale used by the authors to classify the functional status of the heart rate. All ‘cloud patterns’ other than a ‘normal cluster’ were associated with some type of pathology condition of the heart.

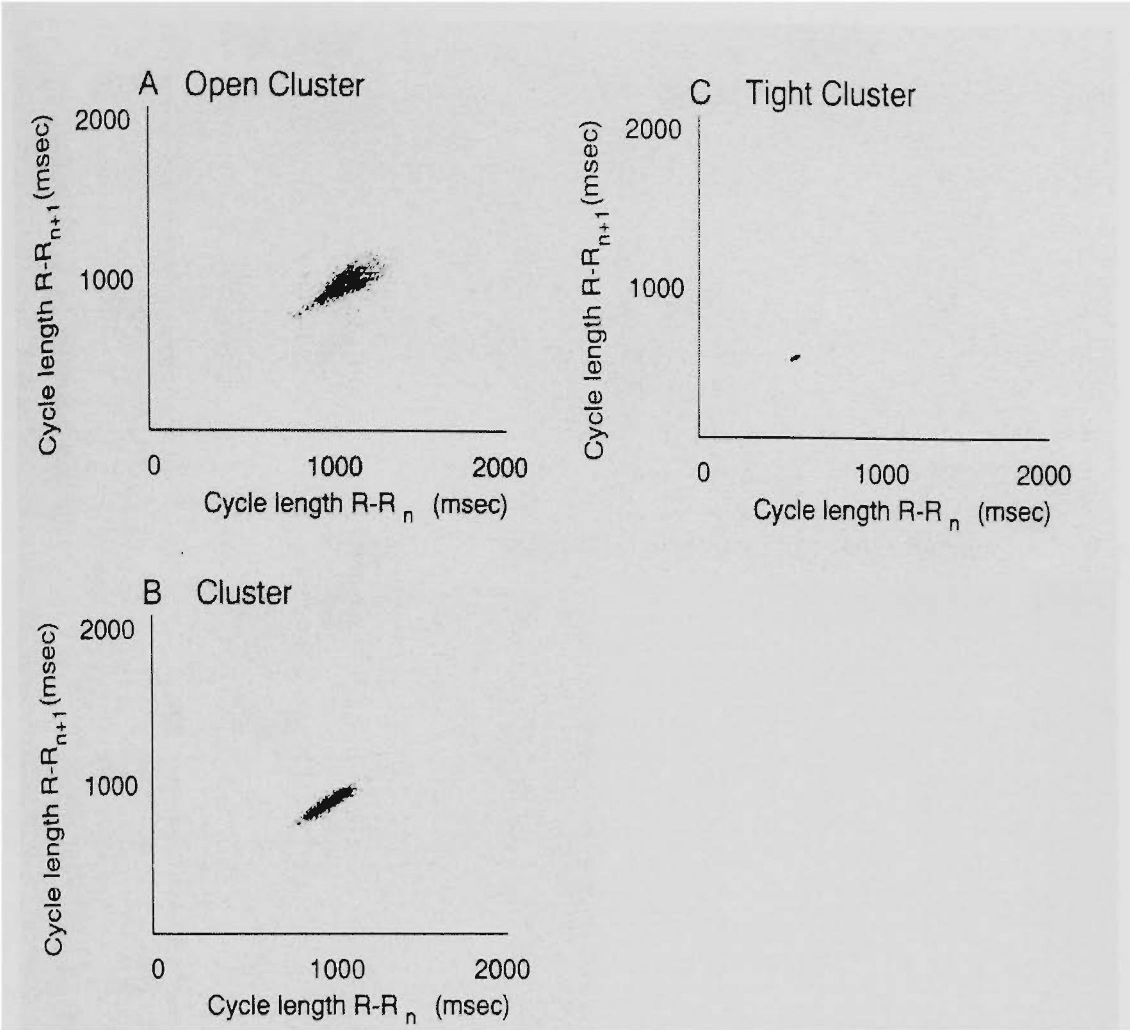

Figure 2.7.4.2 Cluster Patterns: (A) represents an 'open cluster' pattern – found in a patient with heart failure. (B) is called a typical 'cluster' representative of healthy HRV data. (C) indicates a 'tight cluster' and was found in heart disease patients. See text for definitions (Source: Kamen and Tonkin, 1995).

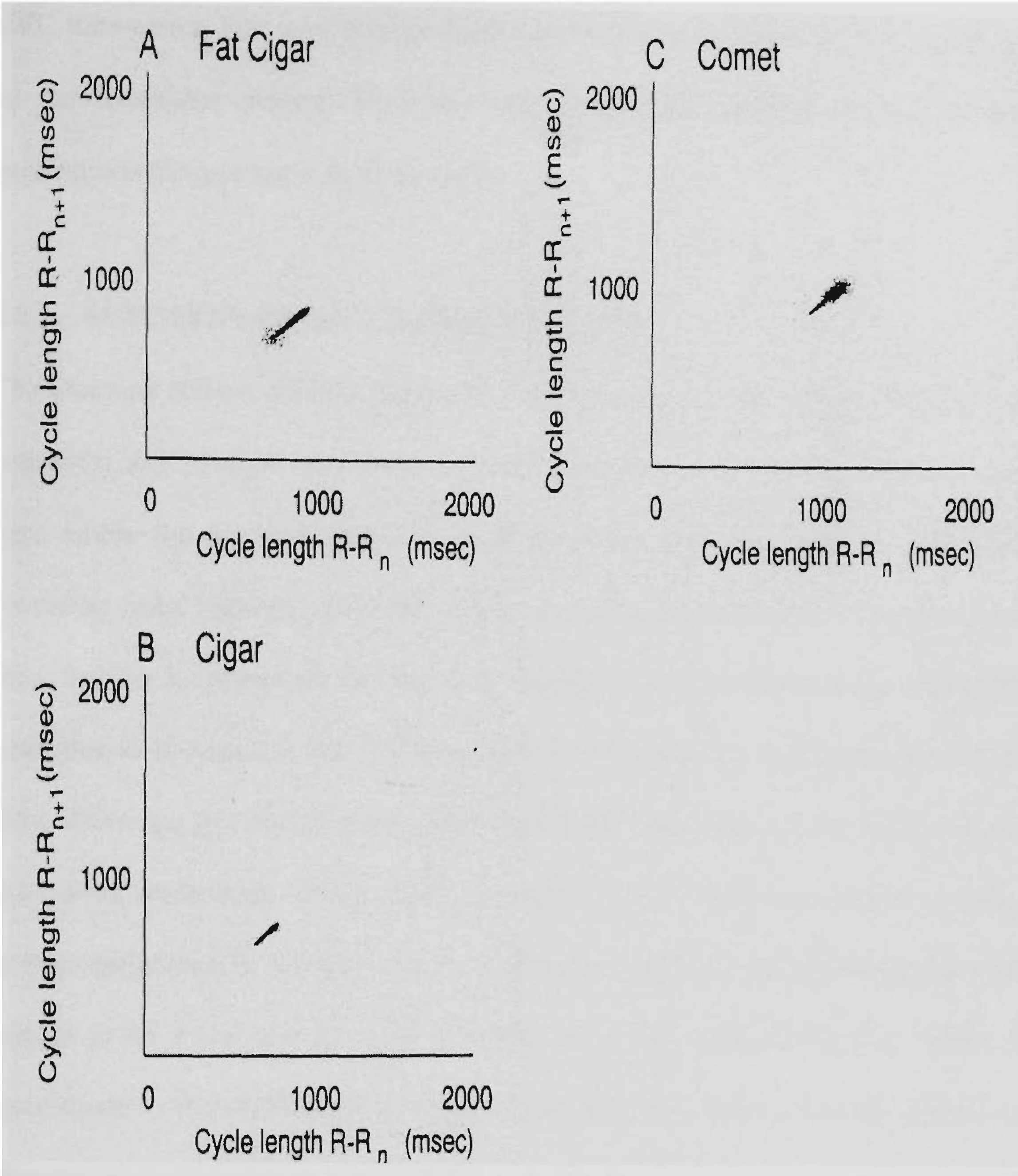

Figure 2.7.4.3. Elongated and non-Gaussian Poincaré plot types. (A) Fat cigar (B) cigar (C) comet typically illustrates non-stationary data. See text for definitions (Source: Kamen and Tonkin, 1995).

Similarities between walking and heart rate dynamics can be related to the complex nature of the interacting physiological systems influencing their respective functioning. Kamen et al. (1995) demonstrate that the Poincaré plot is a promising tool to probe these complex interactions which separate between healthy and pathological function. Based upon the nomenclature by Kamen et al. (1995) on heart rate variability, the Poincaré plot 'clouds' of

MFC time series, that may diverge from a normal cluster pattern could suggest aging effects on the locomotor system. Unfortunately, comparable research studies of Poincaré plot applications for gait analysis do not exist.

2.8 SUMMARY OF LITERATURE REVIEW

The literature review acknowledges the progress made in the various research fields of falls aetiology, gait analysis and measurement of motor control and dynamic systems. Various gaps within the research literature were presented, which requires further exploration for providing links between cause and effect relationships associated with gait parameters and falls. Section 2.1 demonstrated that falls are linked to gait performance. The gait parameters described in section 2.2 and 2.3 indicate that variability in certain parameters are linked to falls. However, gait analysis has not analysed the symmetric nature of gait in detail, which has caused conflicting reports about its presence. The locomotor system reveals a complex process underlying its control, which provides the rationale for long-range correlations being present in the moment-to-moment fluctuations in gait (Hausdorff et al. 1995). Long-range correlations in the stride intervals of gait begin to break down in elderly groups, suggesting a decrease in motor control. The literature on the MFC event in section 2.5 revealed a strong rationale for being associated with tripping during walking, which has implications for the falls problem in the elderly population. The MFC has not been investigated from a bilateral or dynamic variability perspective and represents a gap in the research needing to be addressed.

The final section reviewed two concepts having been applied to dynamic data sets in physiology. The DFA has been applied to stride time intervals, however there is no current research investigating its use on the minimum foot clearance event. It has not been explored for describing inter-limb processes of gait. The Poincaré plot is also new to gait analysis, and has been applied in the past to inter-limb coordination research (Kelso, 1995). No research literature exists for the applications of the Poincaré plot to gait. No research has been found in the literature that has applied dynamic analysis methods outlined within this section to the minimum foot clearance features of gait. The potential benefits of the DFA and Poincaré plot methods for understanding the phenomenon of gait unsteadiness and control will be assessed in this investigation. The methods used to derive parameters from the aforementioned non-linear applications, such as DFA and Poincaré parameters, will be expanded upon in the next chapter.

Chapter 3

METHOD

3.1 SUBJECTS

Subject consent was made in conjunction with the approved ethics protocol granted by the Human Research Ethics Committee, Victoria University, Melbourne.

Ten 'healthy young' male (n = 5) and female (n = 5) people (mean age 33.3 ± 2.9 years) responded to research volunteer notices displayed within the University community. Eight 'healthy older' community dwelling men (n = 5) and women (n = 3) aged between 60 and 80 years (mean age 72.75 ± 3.6 years) were recruited from gymnasiums throughout the metropolitan area. Inclusion criteria for the elderly subjects was (i) to have recent experience having walked on a treadmill and (ii) being active in daily routines. Exclusion criteria were a history of any falls in the previous 12 months prior to testing or the presence of any medical problems affecting balance and mobility. Such conditions included arthritis, osteoporosis, diabetes, joint replacement, vision impairment or medications that may affect their mobility. Further tests were conducted to provide an objective means for classifying elderly subjects as being free of any locomotion impairment. It was beyond the scope of this investigation to conduct comprehensive medical screening by geriatricians to assess the health and onset of aging for the older subject group.

Treadmill experience was required because it would act as a control for any unsteadiness that may have occurred when walking on unfamiliar 'terrain'. Interviewing subjects determined a verbal acknowledgement that they were able to walk on a treadmill for at least 20 minutes at a constant walking speed. The speed conveyed to the subjects was to be their normal walking speed. Subjects are required to wear light, comfortable clothing and their own flat walking/jogging shoes.

Although subjects in the older group were considered healthy in terms of functional ability, some reported minor medical ailments that require maintaining by medical treatment.

3.2 INSTRUMENTATION AND PROCEDURE

3.2.1 Preliminary investigation

The following investigations proved useful for determining the type of marker to be used and the appropriate experimental set up. This investigation included analysis of: (a) location of camera position; (b) camera settings; (c) determination of marker type (LED or passive); (d) harness type and location of attachment; (e) familiarization with equipment; (f) accuracy testing; (g) ground surface prediction by the geometric model and Q-basic software.

After experimenting with reflective markers, it was decided that the complexity of the set up increased when trying to avoid light crossing from opposing sides of the treadmill. Hence, Light Emitting Diodes were tested by the laboratory technicians at Victoria University and were subsequently administered to avoid the issue of unwanted light.

3.2.2 Protocol procedure

The protocol administered was common to both groups, however, there was a falls efficacy survey and an up and go test administered to the elderly group only. The experimental set up and procedure was conducted at the Victoria University Biomechanics Laboratory (Flinders St. Campus).

3.2.3 Subject tests

Subjects received a consent form to fill out including an overview of the testing protocol (see appendix A). Subjects were then tested on the following: leg length (greater trochanter to lateral malleolus), height, weight, range of movement at the ankle joint, overground walking speed, fear of falling survey (Tinetti et al., 1990), and an 'up and go' test (Shumwell-Cook, Brauer and Wallacott, 2000). Standard tests included age, body mass, height, and body mass index (BMI; kg/m^2). Leg length was measured to test discrepancies between left and right sides. This was taken from the greater trochanter head to the lateral aspect of the maleolus. Ankle range of motion was obtained from maximum goniometer measures during passive dorsiflexion and plantarflexion.

The falls efficacy survey determined the level of confidence a person displays when carrying out daily activities. Fear of falling can be associated with changes in gait parameters and therefore a measure of this was accounted for by a 14 item efficacy survey (attached in appendix B). Briefly, each subject rated their level of confidence without an associated feeling of fear that they may lose balance and subsequently fall. This was rated on a scale of 10 from a list of 14 commonly occurring daily activities, such as getting dressed and crossing the street (Adapted from Tinetti et al., 1990; and Hill et al., 1999).

Differences in functional mobility may also cause differences in the MFC data and was thus measured by a timed 'up and go' test (UGT). An UGT is commonly used to assess mobility function and falls risk (Shumwell-Cook et al., 2000). A score of greater than 10 seconds was used in this protocol as an indicator of mobility impairment (Shumwell-Cook et al., 2000). The timed test involves begins upon initiating movement from a seated position on a chair.

The subject then walks forwards, around a marker located 3 meters from the chair and then returns to the sitting position to complete the timing.

The treadmill velocity for all subjects will be based upon their self-selected walking velocity conducted over a 20 meter walkway and calculated from an average of five timed trials. Habitual, or naturally selected gait speed was assessed over the middle 10 meters of a 20 meter walkway using infrared sensored gates. The average of 5 trials was recorded as the habitual gait speed. Gait speed is a common test for measuring physical function in elderly groups and is a valid indicator of a person's ability to perform activities of daily living (Potter, Evans and Duncan, 1995).

3.2.4 Marker placement

Following the preliminary tests, three light emitting diodes (LED) were then attached on the lateral sides of the subject's left and right shoes (Figure 3.2.4.1). Specifically, these LED markers were located at the great toe (TM; P1), 5th metatarsal head (MH; P2), and heel (P4) positions. Movement of the marker attachments during the trial will cause errors in MFC accuracy. Therefore, great care and skill was required to fix the diodes to the shoe. This was made easier by the design of the LED system developed by the Victoria University laboratory technicians. The LED markers provided a tracking reference of the foot during the swing phase of walking. The position of the TM and MH were used to describe the position P3 (Figure 3.2.4.1). P3 is defined as the minimum toe point (MTP), which is derived geometrically after digitizing (see Section 3.3.3). P3 is the point of the shoe that travels closest to the ground during the swing phase. P4 was used to represent heel position that may be required to validate the ground reference.

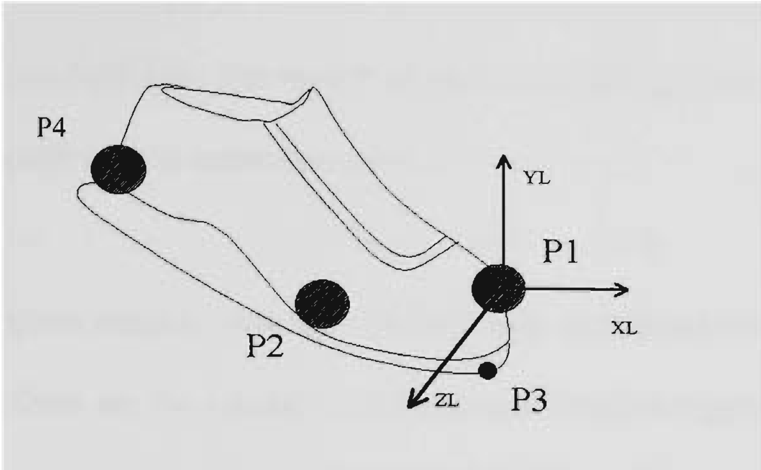

Figure 3.2.4.1. Marker positions. P1 = toe marker; P2 = 5th metatarsal head marker; P3 = minimum toe point at MFC; P4 = heel marker.

3.2.5 Foot model and calibration

Before the testing took place, a view of both the left and right shoes with the markers attached, was captured with the shutter speed turned off to maximize light contrast. A black background highlighted the white strips of lightweight adhesive tape outlining the 'out-sole' surface of the shoe. This was necessary for defining the geometric model. Subjects were asked to place their foot on the edge of a black box situated on the treadmill. The monitor was then checked to ensure a clear outline was visible.

3.2.6 Treadmill walking

Young subjects were asked to walk for 30 minutes continuously. The elderly group was asked to walk for at least 20 minutes, and if they felt comfortable then 30 minutes. The difference in instructions given to the subjects was based upon rates of exertion. Both subject groups were deemed physically able to comfortably walk for this period based upon health, treadmill experience and activity levels.

Treadmill minimises changes in walking speed throughout a walking trial due to the constant velocity of the treadmill belt. The benefit of treadmill walking allows repetitive gait cycles to be analysed through a fixed camera position.

Prior to the subjects stepping onto the treadmill they were fitted with the safety harness (see section 3.2.8). Once on the treadmill, subjects were briefed regarding the operation of the treadmill and how to operate the stop buttons should they wish to stop the belt motion. A kill switch is attached to a string that is then clipped onto the subject's safety harness. The subjects began treadmill walking from a standstill position, holding on to the rails, and then the treadmill belt was accelerated in a slow and smooth manner. Once subjects felt comfortable, they were asked to begin walking with freely moving arms until the overground speed was reached. Self-selected walking speed is commonly required during gait testing, as it is a good indicator of gait performance (Potter et al., 1995). Some subjects felt that the treadmill speed was faster than their normal overground walking speed, thus the testing speed was adjusted to suit their preferred pace.

Filming did not begin until the subjects had acknowledged that they had developed a comfortable rhythm, that they believed closely represented their overground gait. A person stood to the front and side of the treadmill should any assistance be required in case of an accident.

Once the testing time was agreed to end, the subject pushed the stop button, which slowed the belt speed down gradually while the subjects used the rails until the belt stopped. This procedure was demonstrated to the elderly subjects prior to testing. Verbal instructions were deemed adequate for the young subjects.

3.2.7 Instrumentation

3.2.7.1 Experimental set up

The camera set up follows recommended 2-D videography principles (Dainty, Gagnon, Lagasse, Norman, Robertson and Sprigings, 1987). The walking trial was performed on a 'Trimline 7600.1E (230V, 10A and 50Hz) motorized treadmill. The experimental set up (Figure 3.2.7.1) and procedure was conducted at the Victoria University Biomechanics Laboratory (Flinders Street Campus). Cameras 1 and 2 are positioned bilaterally, 9 meters from the midline of the treadmill. This distance is expected to reduce perspective error. The optical axis of cameras 1 and 2 is perpendicular to the plane of the treadmill belt motion to reduce parallax error. Shutter speed was set at 1/1000s, camera speed at 50 Hz with a Panasonic colour CCTV digital camera (model: WV-CL350; power: 220-240V ~50Hz 5.5W, company: Matsushita Communication Industrial Company Ltd.) video camera. The hardware configuration is shown in Figure 3.2.7.2.

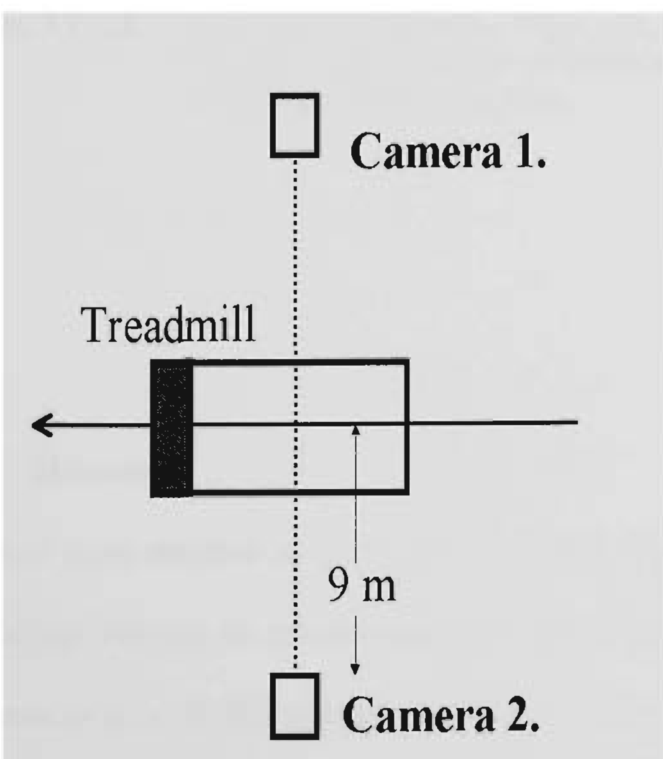

Figure 3.2.7.1 Experimental set up.

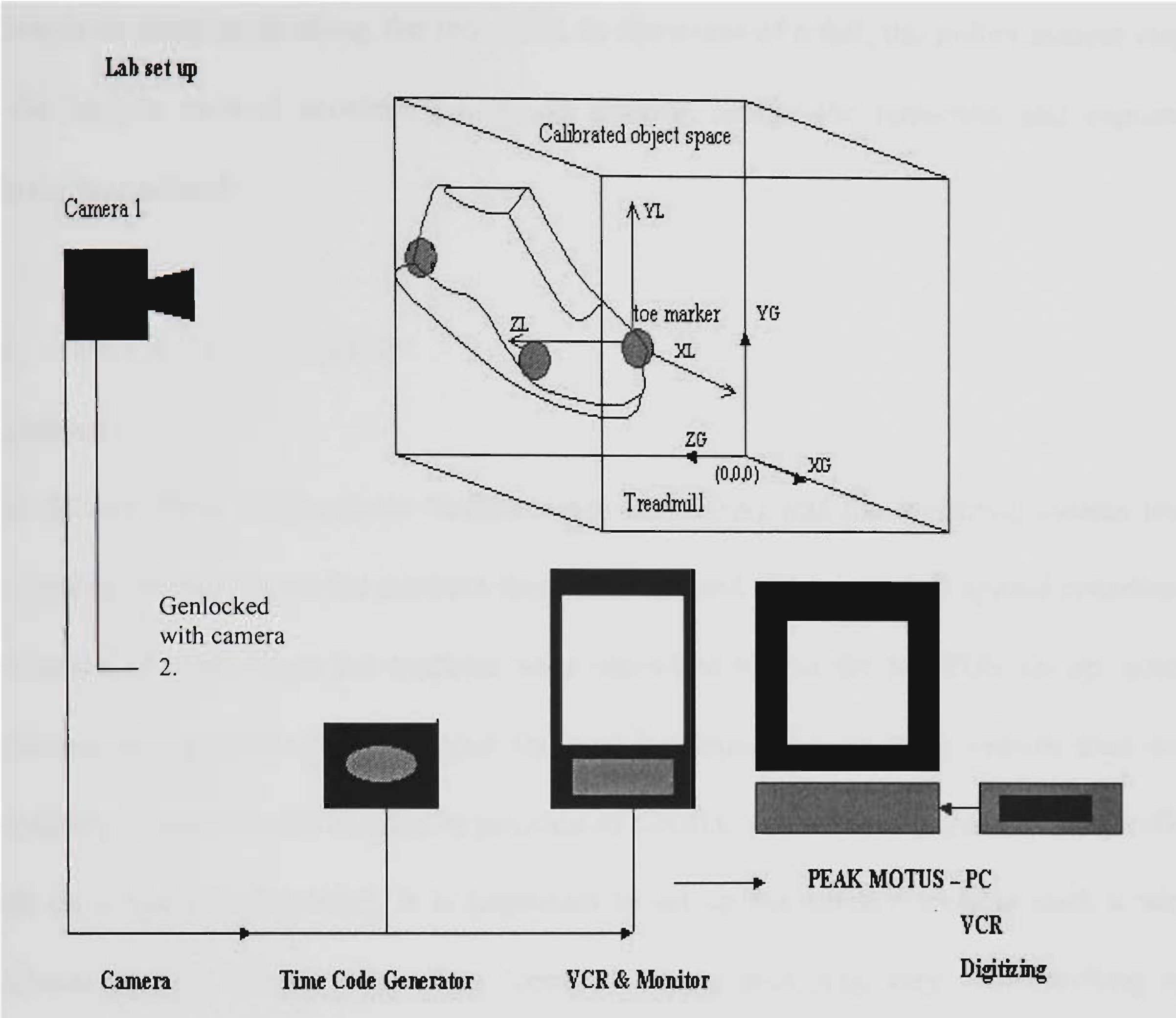

Figure 3.2.7.2. Data collection process. Within the calibrated object space: XG, YG, ZG, refer to the global reference system; XL, YL, ZL, refer to the local reference system.

3.2.7.2 Harness

All subjects were required to wear a harness regardless of age. This is to control for any affects the harness has on constraining gait. The harness was attached to a pulley system, which operates in a similar manner to a car seat belt device. The pulley was attached to a secured 'eye bolt' screwed into the concrete foundation in the ceiling. Because the pulley is spring loaded, no constraint was placed upon the subject, as they were free to move either forwards or backwards along the treadmill. In the event of a fall, the pulley system responds to the body's vertical acceleration due to gravity, brakes the retraction and captures the subject immediately.

3.3 DATA COLLECTION

Digitising:

Peak Motus (Peak Performance Technologies Inc, USA) was the digitizing system used for converting the motion of the markers from the captured video into 2-D spatial coordinates as a function of time. Once the markers were identified within the MOTUS set up, automatic digitizing was performed throughout the walking trial. The tracking system uses contrast sensitivity to monitor the change in position of circular white objects, generated by reflective light on a black background. It is important to set up the camera view in such a way that produces circular 'dots' rather than 'comets' during foot trajectory. The tracking system colors in the dots and predicts their centre position. It is important for the dots to remain circular throughout the foot trajectory to avoid erroneous centre position of the marker.

To convert the raw spatial coordinates to real coordinates, calibration of a known distance within the field of view was performed. This was achieved by videoing and then digitizing two LEDs at a length of 1.00 meter apart in the middle of the treadmill. After filtering the raw coordinates were scaled according to the calibration factor.

The foot model was also digitised, however, calibration was not required in this case as the geometric model is derived from the angles of the triangle and not the distances. Eight manually digitised frames of the foot model were captured to provide spatial information required by the geometric model (see Figure 3.3.1). Hence, careful positioning of the centre position on the P1 and P2 markers and the out-sole point (P3) was required. An additional 5 points, similar to P3 were digitised along neighboring positions on the shoe out-sole to verify that we had obtained the most distal and inferior edge of the shoe closest to the ground at MFC.

Filtering:

An optimal cut-off frequency filter option from the Butterworth Digital Filter (Peak 5 Manual, 1993) was used to smooth the amplitude noise that ranged between 4-8Hz occurring in the raw data. This option in the Peak system follows the Jackson Knee Method (PEAK 5, 1993) procedure. Because of the smooth trajectory at MFC, noise effects are expected to be minimal.

MFC prediction – geometric model:

The filtered raw coordinates of the foot model markers were exported from an ASCII file to a Microsoft ™ Excel file. Here, the calculation of the relative distances and the angles within the triangle are processed. Refer to Figure 3.3.1 for the design of the geometric model. The following equations demonstrate how y(P3), y(P1) and d are calculated.

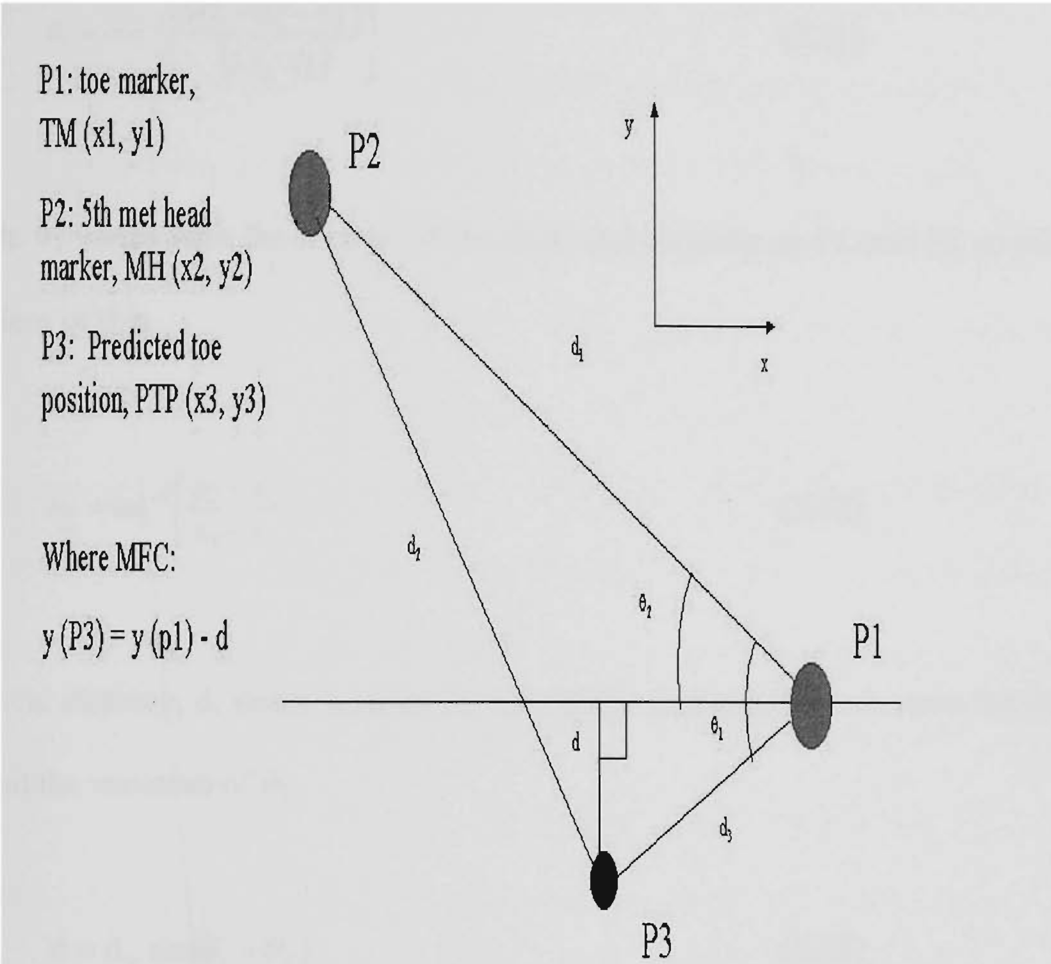

Figure 3.3.1. The geometric model. P1, toe marker; P2, 5th metatarsal head marker; P3 minimum, minimum foot clearance point (MFC_P); d1, d2, d3 and θ_1 are constant values derived from the foot model; θ_2 varies with the gradient of d1.

Mean distances for d1, d2, d3, are calculated using Pythagoras' Theorem. P1, P2, and P3 are located in 2-D plane using (x, y) coordinates.

$$d_1 = \sqrt{(x_1 - x_2)^2 + (y_1 - y_2)^2} \qquad (3.8)$$

$$d_2 = \sqrt{(x_2 - x_3)^2 + (y_2 - y_3)^2} \qquad (3.9)$$

$$d_3 = \sqrt{(x_1 - x_3)^2 + (y_1 - y_3)^2} \qquad (3.10)$$

The value of θ_1 (refer to equation 3.11) remains constant and is calculated by the cosine rule.

$$\theta_1 = \cos^{-1}\left[\frac{\left(d_1^2 + d_3^2 - d_2^2\right)}{\left(2.d_1 .d_3\right)}\right] \tag{3.11}$$

Angle θ_2 varies with the motion of the foot, and depends on P1 and P2 coordinates (i.e. the gradient of d_1):

$$\theta_2 = \tan^{-1}\left(\frac{y_2 - y_1}{x_2 - x_1}\right) \tag{3.12}$$

Vertical distance, d, varies with the motion of the foot and depends upon the constants d_3 and θ_1 and the variation of θ_2:

$$\mathrm{d} = \mathrm{d}_3.\sin(\theta_1 - \theta_2) \tag{3.13}$$

The vertical coordinate of the predicted shoe out-sole position, is given by:

$$\mathrm{y(P3)} = \mathrm{y(P1)} - \mathrm{d} \tag{3.14}$$

During foot trajectory, the position y(P3) reaches a minimum point, such that it is the closest point to the ground during mid swing. Therefore MFC is related to y(P3) in the following equation:

$$MFC = \mathrm{y(P3)}_{\mathrm{min}} - \mathrm{y}_{\mathrm{g}} \tag{3.15}$$

where y_g is the ground reference point. This point is derived from different standpoints. It can be obtained by calculating the minimum value of P3, the shoe out-sole at toe off, or it can be found by manually digitizing the ground position of the treadmill using a calibration device. Both methods were used as a comparison in this study. The trajectory of P3 and P1 is illustrated in the following diagram (Figure 3.3.2). The graph shows a larger vertical position for the toe marker (P1), hence the more accurate value of the shoe out-sole point.

The values derived from equations 3.8 to 3.15 show how the point P3 is related to: the ground (y_g); d1; d2; d3; theta 1 (θ_1); and theta 2 (θ_2). The values are inserted into a software program (q-basic). The program is able to track P3 from the spatial coordinates of the toe (P1) and the 5th metatarsal head (P2). The position of MTP (P3) at the MFC event is related to the angle θ_2 from the coordinates of P1 and P2. A second program is used to isolate the frame at which P3 reaches a minimum value (MFC) within each stride, and provides a series of stride-to-stride MFC values. Figure 3.3.2 shows the variables (A - E) that the q basic software program provides. Variable C is the MFC, where the treadmill height is determined by the average MTP value at toe off.

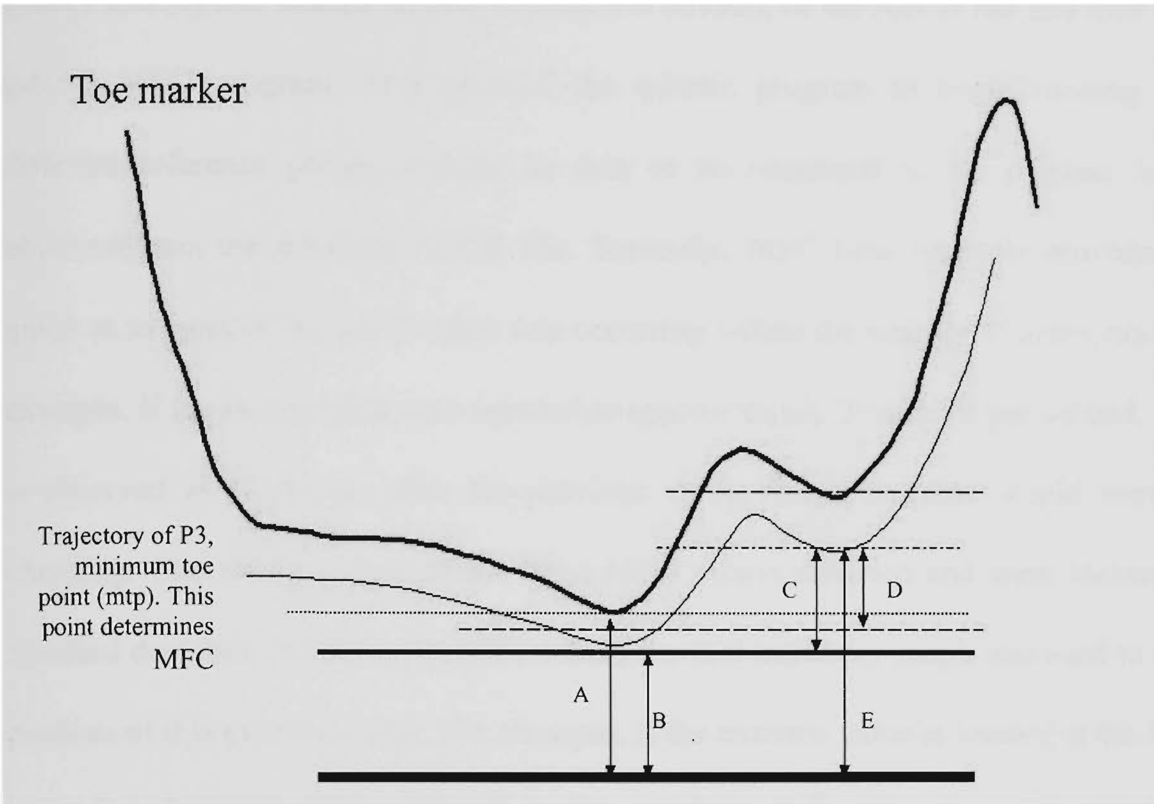

Figure 3.3.2. Trajectory of the toe marker and the minimum foot clearance point (MFC). A – absolute position of the average minimum value of MFC at toe off; B – absolute position of the manually digitized ground reference point; C – distance between MFC and the manually digitized ground reference point; D- distance between MFC and the average minimum value of MFC at toe off; E – absolute value of MFC.

3.4 MEASURING MFC

3.4.1 Formulating intra-limb MFC data

Within each limb, left and right, MFC data series values were calculated and then screened in the following manner. The benefit of the time-space graph showing the shoe out-sole point trajectory is that the calculated MFC values can be validated by assessing the frame number and the form of the 'out-sole point' curve at MFC. Due to the nature of the q-basic program, data points are overlooked at every 4000 frames, which can occasionally mean erroneous MFC values. Hence, to eliminate any incorrect values in the MFC time series, three checking methods were adopted. Firstly, an editing procedure was followed by deleting the first (a) 150 frames and (b) 300 frames (approximately 4-6 strides), of the ASCII file and then running the q-basic MFC program. This allowed the q-basic program to begin running from three different reference points, serving as data to be compared to the original MFC values obtained from the unedited ASCII file. Secondly, MFC time intervals provided a general guide as to whether the MFC value was occurring within the vicinity of mean stride time. For example, if the mean stride time equated to approximately 50 frames per second, and a value is observed at 30 frames after the previous value, then this value would warrant further checking. The third method: if the three MFC values matched and were located outside 3 standard deviations of the mean MFC value, the foot trajectory graph was used to validate the position of this extreme value. For example, if the extreme value is located at the MFC event, and it is indeed very large or small, further checking of the video provided visual qualitative clarification that this was in fact a probable occurrence.

3.4.2 Formulating inter-limb MFC data

Because the left and right intra-limb digitized data is captured independently, a reference system was required to synchronize the time series data. Synchronisation of the cameras and the time code generator (refer to Figure 3.2.7.2), and a formula derived in Excel (Microsoft ™) provided this reference system. This ensured that sequential left and right MFC events were appropriately converted without any shifting effects within inter-limb data sets.

3.5 MFC PARAMETERS

3.5.1 MFC descriptive parameters

Important information of the MFC distribution was investigated, including measures of normality, central tendency and dispersion. This information was obtained from SPSS™ and MS Excel worksheets.

Qualitative assessment of normality involved inspection of histograms, normal probability and detrended probability plots (Hair, Anderson, Tatham and Black, 1998). Because the MFC data sets are larger than 50 samples, the Kolmogorov-Smirnov statistic with a Lilliefors significance level was used for quantitatively assessing normality, which was assumed if the significance level was greater than .05. Based upon previous MFC studies, data obtained from both young and elder groups walking on a treadmill demonstrate that the distribution is non-gaussian (Best et al., 1999). Relatively large non-zero values of skewness and kurtosis are characteristic of these distributions and will also be explored.

Given the assumption that MFC is a non-normal distribution, a measure of central tendency is assessed by calculating the mean, median and mode. Similarly, measures of dispersion include the range, standard deviation and coefficient of variability (CV). Hausdorff et al. (1997) has used these measures of dispersion to demonstrate differences between fallers and non-fallers (refer to section 2.3.1). The coefficient of variation (CV) is calculated from the following equation, adapted from Winter (1991), and is expressed as:

$$\mathrm{CV} = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N} s_i^2}}{\frac{1}{N}\sum_{i=1}^{N}\left|X_i\right|} \qquad (3.16)$$

where N is the number of MFC sample values collected over the trial, X_i is the average (mean) 'measured value' at the i^{th} event, and s_i is the standard deviation about the mean X_i. The CV can give a quantitative and visual representation of variability for MFC over the walking trial.

Remodeling the non-normal data series into a normal distribution was not attempted because it may ignore the true nature of the phenomenon occurring with MFC.

3.5.2. MFC dynamic variability parameters

For analysis of the inherent dynamics of both intra-and inter-limb parameters of interest, different approaches were adopted when formulating the data series. Treatment of the data relevant to the parameters listed in Table 3.5.2.1 was specifically addressed in the following sections. Table 3.5.2.1 identifies the parameters and their sources that were used to explore inter- and intra-limb MFC characteristics.

Table 3.5.2.1. Intra-and inter-limb parameters under investigation and their cited methods of calculation (+ indicates parameters specific to dynamic measures).

Parameter	Authors
Minimum foot clearance (MFC)	Winter, (1990); Dingwell et al. (1999); Best et al. (1999); Karst et al. (1999)
MFC standard deviation (SD) and coefficient of variation (CV)	Winter et al. (1991); Best et al. (1999); Dingwell et al. (1999); Hausdorff et al. (1997).
Symmetry Index (SI)	Becker et al. (1999); Giakis et al. (1997).
+Inconsistency of the Variance (IV)	Hausdorff et al. (2001).
+Nonstationary index (NSI)	Hausdorff et al. (2001); Hausdorff et al. (2000).
+Scaling exponent α (DFA)	Hausdorff et al. (1995).
+Relativity ratio μ (D1/D2)	Brennan et al. (2001).
+D2	Kamen et al. (1995).

3.5.2.1 Normalising each data series

Because each MFC time series can be composed of different mean and variability characteristics, each time series was treated to minimize these effects. Each observation of successive intra-limb MFC events was normalised with respect to the sample's mean and standard deviation. In essence, this procedure forms a series of z-scores (i.e. LZ_1, LZ_2, LZ_3,..; where LZ_1 represents the 1st MFC z-score value of the left limb). Each data series now has a mean of 0 and a standard deviation of 1. This controls for any inherent differences between the left and right sides, such as higher MFC values or larger overall variability magnitude.

3.5.2.2 Intra-limb dynamic variability measures

Measuring IV and NSI:

Moment-to-moment analysis may be categorized as a cluster of MFC events changing over time. To represent this occurrence, the time series is divided into blocks of 5 MFC values each, where each block receives a measure of its mean and standard deviation. The NSI is simply the calculated standard deviation of the block means. This measure describes how the local averages change throughout the walking trial and is independent of the overall variance. The IV is the calculated standard deviation of the summed standard deviations within each block and also provides insight how the local variance changes with time (Hausdorff et al., 2001). In both cases, higher values represent an unstable control from one moment to the next where a moment represents a period of time expanding 5 strides. The calculation of NSI and IV was performed using MS Excel™ spreadsheets.

Measuring relativity ratio (μ), D1 and D2:

The Poincaré plot descriptive statistic was based upon both raw data and normalised data (see Section 3.5.2.1). The original MFC values are used for describing intra-limb Poincaré plots.

To account for between limb differences, inter-limb plots are derived from normalised left and right MFC data with respect to each side's data series mean and standard deviation. The descriptive statistics and plot shape is unaffected by this normalization procedure, only the plot shape undergoes a shift towards the origin of the plot axis. The classification procedure (listed in Table 3.5.2.2.1) for qualitative assessment of the plots (nomenclature) came from modifications of previous research (Kamen and Tonkin, 1995). The quantitative descriptive parameters D1 and D2 were calculated from the equations previously listed in section 2.7.3. Those equations are restated here for simplicity:

$$D_1^2 = ½(SDSD^2) \quad (3.17)$$

$$D_2^2 = 2SD(MFC)^2 - ½(SDSD^2) \quad (3.18)$$

Where SDSD is the standard deviation of the successive differences of the MFC events. SD(MFC) is the standard deviation of the MFC *z*-scores. Kamen and Tonkin (1995) used an aspect ratio to describe the relationship between two different variability measures of the Poincaré cloud (see Section 2.7.4). A similar, but an alternate ratio was applied in this dissertation that calculated the quotient of D1/D2, termed the variability ratio, 'μ':

$$\mu = \frac{D_1}{D_2} \quad (3.19)$$

These equations were then formalized in an MS Excel™ spreadsheet and applied to the normalised data. The plot itself was derived from the 'untreated' MFC values, plotting MFC_n against MFC_{n-1}. This gives a similar shape to the 'treated' plot but without the shifting associated with negative normalised values.

Table 3.5.2.2.1. Quantitative guide for Poincaré plot nomenclature. The values in this table relate to normalised MFC data sets as described in section 3.4.2.1.

	SD1 Short term variability	SD2 Long term variability	SD2/SD1 Aspect ratio
Blob		>0.8	>0.9
Squashed Blob		>0.8	<0.9
Cluster- normal	> 0.2	0.2 – 0.8	< 0.65
Cluster- tight		< 0.2	
Fat cigar	> 0.1	0.2 – 0.8	> 0.65
Cigar	< 0.1	0.2 – 0.8	> 0.65
Comet	A pattern reminiscent of a comet with a large head and a narrow tail.		

Measuring DFA:

The DFA calculates the correlations in a set of data as outlined in section 2.7.3. Equation 2.4 from section 2.7.3 can be adapted in the following way:

$$y(k) = \sum_{i=1}^{k} \left[MFC_{Li} - MFC_{(L\ ave)} \right] \qquad (3.20)$$

where MFC_{Li} is the i^{th} MFC z score value of the left limb, and $MFC_{L\ ave}$ is the average MFC value obtained for the left limb. This will then be applied to equation 2.5 (see section 2.7.3). To determine that the scaling value is not due to non-stationary effects commonly found in biological data, a process of 'validation' is followed. Firstly, a visual inspection of time series graphs can provide information about stationary effects and the occurrence of extreme outliers within the data series. Following this step, the calculation of α can be deemed a robust measure of the inherent long-range correlations when applied over different sized scaling regions and for 'higher order detrending'. This was done for four scaling regions that are most commonly used in previous studies (Hausdorff, Purdon, Peng, Ladin, Wei and Goldberger, 1996), (i) default region[1], (ii) sliding scale region (see footnote1), (iii) $10 < n < 50$, (iv) $10 < n < 20$. A robust measure is related to convergence for each of these scaling regions calculated over higher order polynomial detrending (i.e. 2nd, 3rd, and 4th order).

3.5.2.3 Inter-limb dynamic variability measures

For descriptive MFC parameters, exploring the inter-limb differences are represented as a relationship between the ensemble averages of the left and right limbs. This was achieved by using the symmetry index (SI) from equation (1) in section 2.6.1, which is restated here in its application to MFC:

$$SI = \{(MFC_r - MFC_l) / 0.5*(MFC_r + MFC_l)\}*100 \qquad (3.21)$$

where MFC_r is a specified parameter related to the right MFC data. A negative value for the SI indicates that the parameter for the left limb is greater than the parameter value of the right limb.

Measuring inter-limb IV and NSI:

The order of the time series for obtaining the inter-limb NSI and IV will be:

Lzn, Rzn, Lzn+1, Rzn+1, …

[1] The default region and scaling region are 'option' applications from the DFA software program. The default region $4 < n < N/4$ (non-overlapping windows), and the sliding window region is carried out over all possible window sizes overlapping. (Source: Goldberger et al., 2000).

where z is the normalised score obtained from section 3.5.2.1, and Lzn is the nth MFC value of the left limb, and Rzn is the nth MFC value of the right limb.

Measuring inter-limb relativity ratio (μ), D1 and D2:

Because the Poincaré plot is based upon the immediate relationship between two points over an entire data series, these plots are appropriate for describing the inter-limb relations between successive MFC occurrences (i.e. left n against the previous right n-1).

3.6 EXPERIMENTAL ERROR

This section explores sources of error affecting both the accuracy and precision of reconstructed MFC values. The implications of the results presented will be discussed in section (5.1). The predominant source of error affecting the reconstruction of MFC values relates to digitising processes.

There are three areas where error can be found in the process of obtaining MFC values. The first source of error is variable and relates to the tracking of the two markers positioned on the shoe during the trial and can cause fluctuations about the true position of the MFC. The second source of error is systematic and is associated with human interpretation when digitizing the foot to obtain the geometric model and is representative of systematic error. Thirdly, the prediction of the ground can be determined from different methods and is related to systematic error of the MFC. This is also systematic error.

3.6.1 Camera resolution

Using the set up described in section 3.2.7 the camera resolution was calculated using a wall marked with a grid to represent the absolute vertical and horizontal edges within the camera view. The field of view closely approximated the area representing that to be used in the testing protocol. Eight light emitting diodes (LEDs) were manually placed at precise (± 0.5 mm) edge marks located on the grid. The results were calculated after filtering the scaled raw coordinates. The root mean square (RMS) error was then calculated from the reconstructed marker coordinates.

3.6.2 Aperture

Aperture is important when digitising moving markers with PEAK Motus™. To obtain appropriate camera aperture for this study, two LED markers were positioned (240 mm apart) on a flat rectangular object free to rotate about a fixed plane, parallel to the sagittal view and positioned 9 meters from the camera (i.e. as per set up procedure, section 3.2.7). Trials consisted of capturing various apertures of the rotating object, at similar velocities to those expected during walking. Subjective analysis of the video focused on the changes in the shape of the markers, with the aim of determining the appropriate aperture that consistently produced a spherical shape.

3.6.3 Calculating error

The two types of error quantified in the following procedure are commonly referred to as the root mean square (RMS) error and the coefficient of variation (CV). The latter method determines the variability about the mean value, whereas the root mean square considers the error associated with the true value (Schmidt and Lee, 1999).

3.6.3.1 Static error

A similar set up to the one listed previously was used for determining the accuracy when markers were not in motion. The rectangular object was placed in a static horizontal position to measure the mean horizontal distance and then at a vertical position to measure the mean vertical distance.

3.6.3.2 Dynamic error

Referring to section 3.3.3, the calculation of P3 requires digitization of the foot model. From the foot model coordinates the value d3 is obtained along with the value θ_1. These two values remain as constants, which are then combined with the changing value of θ_2, obtained from the motion of the two markers of the foot during the walking trial. Therefore, only the angle θ_2 is the limitation to the accuracy of MFC during digitization of the walking trial. The following experiments apply to the possible errors associated with θ_2, which come from the digitization of the two moving markers P1 and P2.

Adopting the same set up as per section 3.2.7, the object described in 3.6.2 was rotated within different planes to the camera axis. A geometric reconstruction of the vertical and horizontal position of each marker from one frame to the next allowed for the relative distances P1 to P2 to be derived throughout the rotation of the object.

3.6.3.3 MFC error from foot rotation during the swing phase

Foot rotations have been investigated in the gait cycle and have revealed tri-planar motion during the swing phase (Figure 3.6.3.3.1). The consequences of foot rotation on the calculation of MFC can be demonstrated in Figure 3.6.3.3.2. The latter figure illustrates the changes in length between P1 and P2 (this refers to the distance d1 in equations 3.9 – 3.12) with tri-planar motion. The reconstruction of MFC is obtained by the application of: the predetermined geometric model coordinates upon the digitised coordinates of P1 and P2 from the walking trial. The angle between P1 and P2 (represented by θ_2 in equation 3.13) varies during the gait cycle and determines the orientation of the geometric model. Figure 3.6.3.3.2 shows that slight variations in the angle can occur with marker rotation in the XZ and YZ planes. Therefore, calculation of the angle $\theta 2$ and the change in d1 was obtained from pilot studies to determine the extent of variation in $\theta 2$ that is caused by foot rotation in the XZ and YZ planes. A pilot study using a test subject walking for 5 minutes on the treadmill provided the results for investigating the errors associated with MFC error. The same instruments and data collection protocol was used as per sections 3.2 and 3.3.

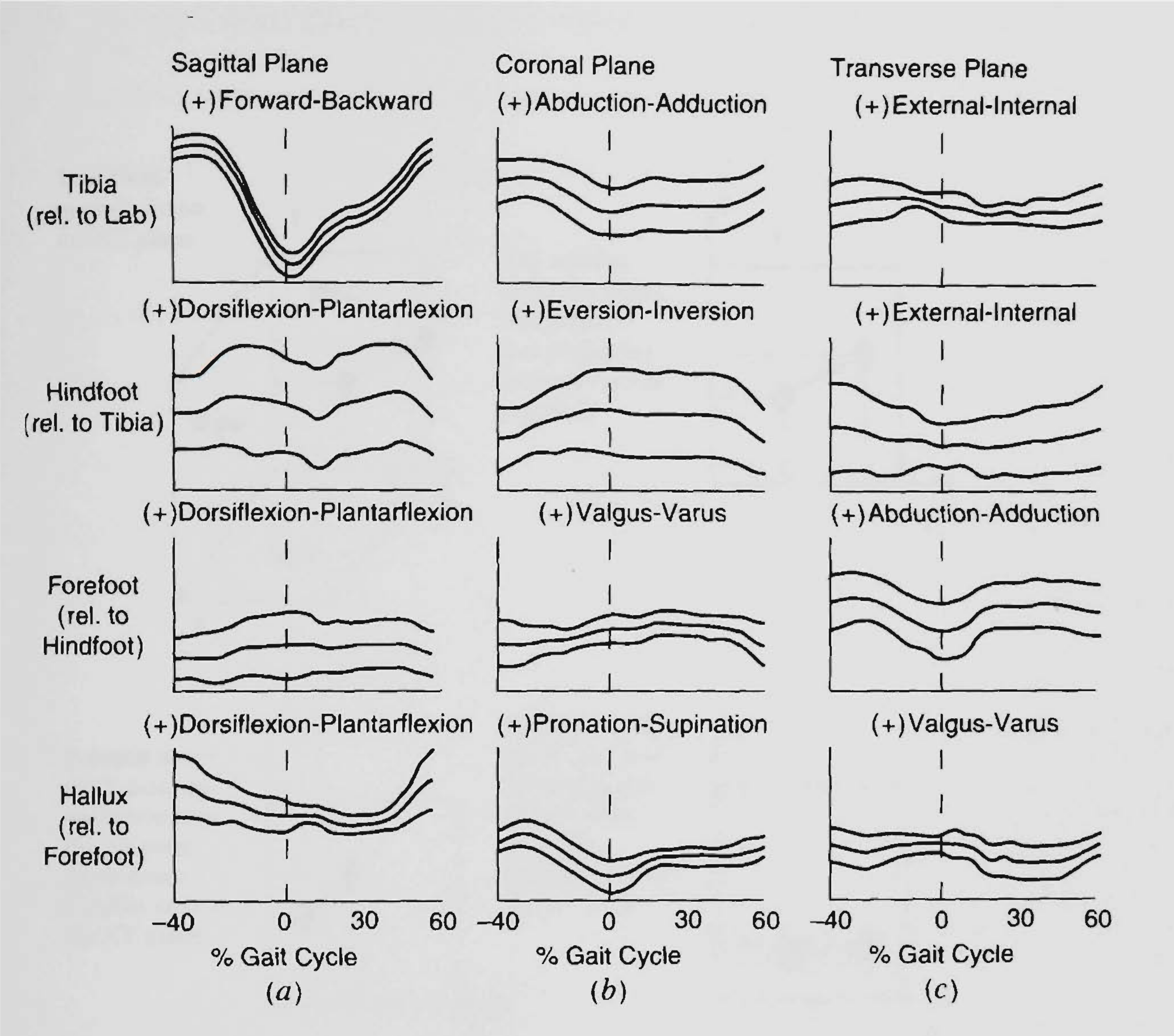

Figure 3.6.3.3.1. Rotations of the lower limb relative to the proximal joint in three planes (column: A - XY plane; B – ZY plane; C - ZX plane). Foot trajectory during swing phase is the portion preceding 0% of the gait cycle. Minimum foot clearance occurs at approximately mid swing (i.e. ≅ -20% of the horizontal axis). Results are taken from a sample of 3 subjects. Because rotations within the foot are within the shoe, the top row of results is of most relevance to foot rotation relative to the camera axis. Due to the nature of the ankle joint, motion in the XY plane seems more than likely to produce motion in the transverse and coronal planes also. The graphs demonstrate that during the swing phase, rotation in these planes is present. Individual differences will determine the magnitude of rotation. (Source: Johnson, Kidder, Abuzzahab and Harris, 1996).

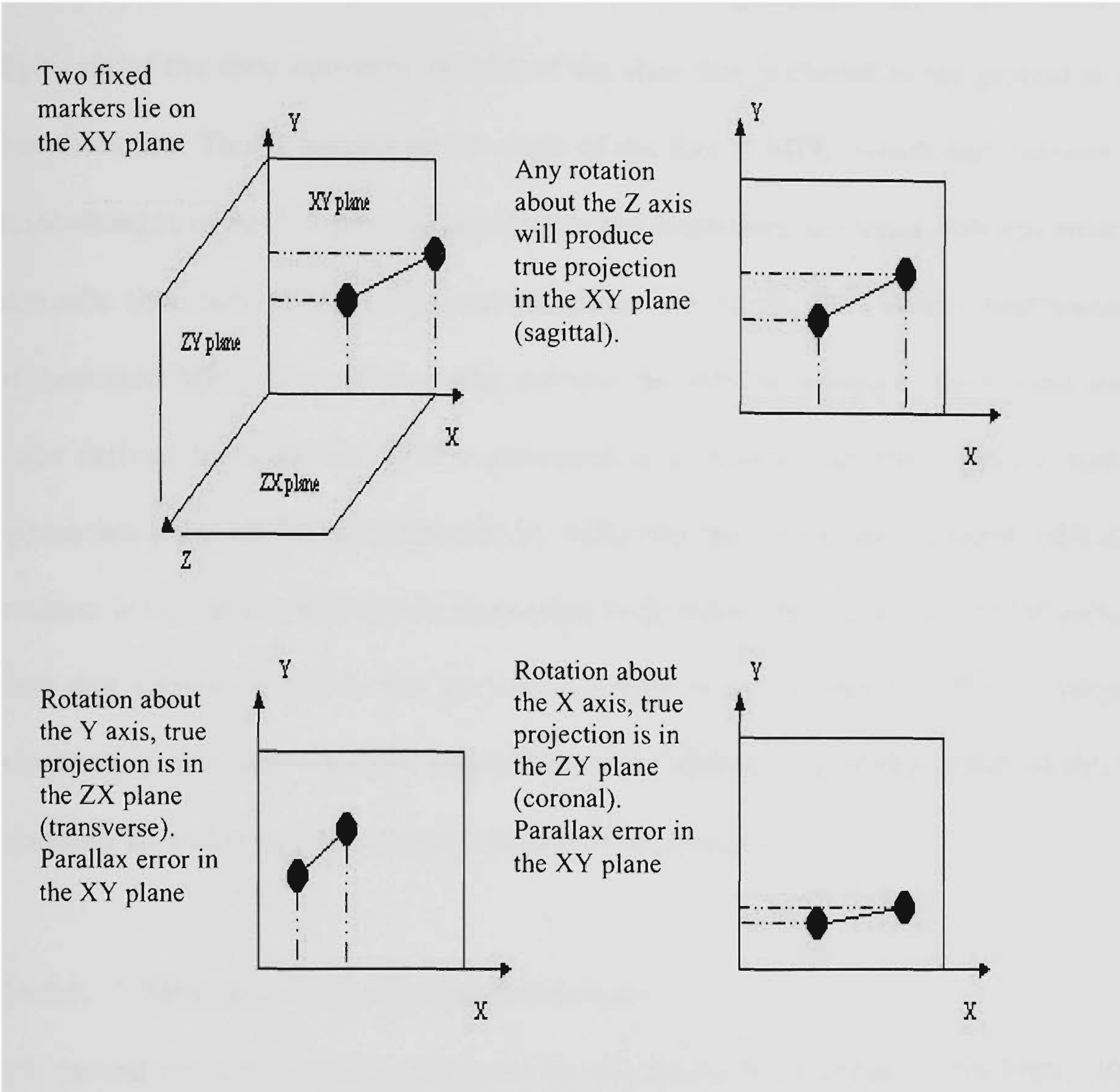

Figure 3.6.3.3.2. Diagrams showing the errors that may be incurred in three planes of motion when filming from a 2-D (sagittal, XY plane) view. The two markers represent P1 and P2 from the foot model. The change in the gradient (theta 2) joining the two markers can be observed by out of plane (XY) rotation.

3.6.3.4. MFC error due to foot model digitising errors

Another limitation of reporting inaccuracies is a systematic error associated with the digitising of the shoe out-sole, the part of the shoe that is closest to the ground at minimum foot clearance. This is subject to the angle of the foot at MFC, which may involve stride-to-stride changes in the XY plane. To minimise the limitations involved with this issue, 6 points along the shoe outsole were digitised (minimum toe points; MTP's) and represented six sets of generated MFC data series. After running the q-basic program, the lowest mean MFC value derived from the six MTP's generated (six independent foot models), indicated the appropriate point on the shoe outsole. By following this procedure, between limb differences in mean MFC values will not be associated with misrepresenting the correct portion of the shoe that comes closest to the ground. To determine whether the MTP's are accurately representing the shoe out-sole, observation of a spatial-time series graph of the 6 MTP's trajectory should reveal convergence at the same vertical point.

3.6.3.5. MFC error due to ground reference

The ground reference value is obtained by calculating the average y coordinate value of the MTP at toe off. The ground reference point is limited to the level of accuracy obtained in sections 3.6.3.2 and 3.6.3.3. The standard deviation was calculated for the y coordinate MTP value at toe off. A low standard deviation of the MTP at toe off suggests a consistent ground reference value, which is then limited in accuracy only to the presence of foot model digitizing error (see section 3.2.5).

3.7 STATISTICAL ANALYSIS

A common problem in biomechanics and motor control as emphasized by Mullineaux et al. (2001) relates to the sample size and the number of feasible trials. The nature of this investigation is no exception because of intensive data- collection, analysis and interpretation. Using a 50Hz camera, it takes approximately 25 hours to digitize one subject, including both left and right limbs from a captured 30 minute period of walking. Hence the sample size in this study is limited.

Mullineaux et al. (2001) suggests the use of effect size to complement results of statistical significance and eliminating violations of assumptions that underpin statistical inference. Effect size is defined by Hair et al. (1998) as the estimate of the degree to which a phenomenon being studied, such as the difference between means, exists in the population. The use of effect sizes is important when considering research that deals with small sample size and large variability. Effect sizes based upon t-test group comparisons are calculated from the following equation, where $\overline{X}_1$ (young) and $\overline{X}_2$ (elderly) are the group means and SD_1 is the standard deviation of the young group:

$$ES = \frac{\overline{X}_1 - \overline{X}_2}{SD_1} \qquad (3.22)$$

For the differences between means, the effect size will represent the number of standard deviations that the two groups will differ by.

The required sample size to obtain a power of 0.8 with an alpha level set at .05 is obtained using the following equation, where $\overline{X}_1$ (young) and $\overline{X}_2$ (elderly) are the group means, Z_α

relates to the level of significance (p = .05) and Z_β relates to the level of power (β=0.8). The assumptions are that N_1=N_2 and SD_1= SD_2, where in these cases SD will be related to the current young groups SD value for the parameters investigated.

$$N = \frac{2(SD^2)(Z_\alpha + Z_\beta)^2}{(\overline{X}_1 - \overline{X}_2)} \qquad (3.23)$$

$$N = \frac{2(SD^2)(1.96 + 0.84)^2}{(\overline{X}_1 - \overline{X}_2)} \qquad (3.24)$$

Significant differences will be stated when p values fall below the .05 level. Quoting only statistically significant differences between means does have limitations when the sample size is small (Franks and Huck, 1986). To make meaningful inferences about the results, p values will and the effect size will be quoted.

It is recommended that when dealing with less than 50 samples, the Shapiro-Wilks statistic is a guide for assessing normality of grouped parameters (Coakes and Steed, 1999). Means and standard deviations will be provided for all group measures. Within limb analysis of normality was performed by calculating the Z score values of skewness and kurtosis (Vincent, 1995) and the Lilliefores statistic (Coakes and Steed, 1999).

Providing further support for determining whether violations of MFC normality is when the skewness and kurtosis scores are converted to Z scores by dividing their scores by the standard error. This procedure allows a true evaluation of the scores where normality is acceptable within the range of ± 2 (Vincent, 1995). Hair et al. (1998) use a slightly more conservative approach by suggesting that z scores beyond ± 2.58 represent the critical value for rejecting the assumption of normality.

Obtaining skewness and kurtosis as a raw value requires converting each value in the data set to a Z score. Each new value is then treated by equations (3.25 and 3.26) to obtain the skewness and kurtosis of the data, where Z relates to the Z score and N is the size of the data set (Vincent, 1995).

$$Skewness = \frac{\sum Z^3}{N} \tag{3.25}$$

$$Kurtosis = \left(\frac{\sum Z^4}{N}\right) - 3 \tag{3.26}$$

The standard error for skewness and kurtosis is expressed by equations 3.26 and 3.27:

$$SE_{skew} = \sqrt{\frac{6}{N}} \tag{3.27}$$

$$SE_{kurt} = \sqrt{\frac{24}{N}} \tag{3.28}$$

To obtain the Z value we divide the raw scores of skewness or kurtosis by the appropriate standard error, obtaining the following equations:

$$Z_{skew} = \frac{\sum Z^3}{\sqrt{\frac{6}{N}}} \tag{3.29}$$

$$Z_{kurt} = \frac{\sum Z^4}{\sqrt{\frac{24}{N}}} \tag{3.30}$$

Non-parametric Wilcoxen Ranked Sum W tests were used to assess between group differences for all parameters. Given that the sample sizes are small, the assumption of normality is difficult to determine and these tests make no underlying assumptions about the distribution of the data. Also, justification of non-parametric analysis relates to the two non-linear methods applied to the MFC data analysis. The authors of both methods have practiced with similar methods to determine between group results (Hausdorff et al., 2001; and Kamen et al., 1995). Between group effect sizes were calculated along with the number of subjects required within each group to obtain power of 0.8 at a .05 significance level. Spearman's 'r' correlation coefficient was used to quantify univariate associations, as indicated. Unless indicated otherwise group results are reported as means ± SD. Statistical analysis were performed using SPSS software (version 10.0).

Chapter 4

RESULTS

4.1 EXPERIMENTAL ERROR

4.1.1 Camera resolution

The results for the camera resolution in the vertical and horizontal fields are presented in table 4.1.1.

Table 4.1.1.1 Camera resolution

	Pixels/mm	SD	RMS (error)
Horizontal	0.41	0.01	0.27
Vertical	0.41	0.01	0.26

4.1.2 Aperture

With low apertures the marker typically illustrated a comet like shape. With high apertures, the shape of the marker was round, but reduced in size. Subjectively, the appropriate aperture represented a sphere at the brightest threshold, approximately 2.8 on the camera settings (Panasonic WV-CL350 digital camera).

4.1.3 Digitising error

Results from experimental procedures outlined in sections 3.6.3 will be reported in the following two subsections 4.1.3.1 and 4.1.3.2.

4.1.3.1 Static error of two markers

The error associated with digitizing the LEDs used in the experiment are reported in Table 4.1.3.1.1. The test showed that the average distance between the markers in both the horizontal and vertical directions is accurate to the 'true' measured distance of 240 mm. The variability about the measured average is almost within 1 mm for the vertical plane.

Table 4.1.3.1.1. Digitised error in static marker position. Reported errors related to the change in digitized distance between the two markers.

	Mean (mm)	SD (mm)	RMS error	% Error (CV)
Horizontal	240.10	0.90	0.88	0.37
Vertical	240.07	0.62	0.62	0.26

4.1.3.2 Dynamic error of two markers

The error associated with digitizing the LEDs used in the experiment are reported in Table 4.1.3.2.1. The mean distances of the markers were calculated as 240.24 mm, with a standard deviation of 0.945 mm (0.4 %). Table 4.1.3.2.1 results show only a slight increase in the CV and RMS error when compared with results when the object is stationary (Table 4.1.3.1.1). The root mean square (RMS) error can increase by a magnitude of 20, when the object is rotated away from the sagittal plane (i.e about the XZ and YZ planes, see section 3.6.3.3.2).

Table 4.1.3.2.1. Digitisation error associated with moving markers. The error is associated with the change in distance between the two markers digitized.

Rotation Speed	Mean (mm)	SD	RMS error	% error (SD/mean)
Slow Rotation				
90° to camera axis	240.2	0. 9	0.9	0.4
30° to camera axis	225.9	11.1	17.9	4.9
20° inversion	224.2	5.5	16.8	2.5
20° eversion	239.2	4.6	4.1	1.9
Fast Rotation				
30° to camera axis	224.8	10.7	18.6	4.8

4.1.4 MFC error from foot rotation during the swing phase

The following results are from pilot testing of a subject walking for 5 minutes (see Section 3.6.3.3). The measurement of MFC as described in section 3.3.3 is dependent upon the assumption that the foot model coordinates d1, d2, and d3 remain invariant during treadmill walking. Section 3.6.3.3 demonstrates that foot motion during the swing phase of walking is unlikely to be strictly planar and thus the values d1, d2 and d3 are expected to vary slightly. Because of the nature of the experimental design, investigation of d1 is the only means of gaining an insight into the expected error of MFC when using 2-D analysis methods during treadmill walking. Figure 4.1.4.1 describes the changing magnitude of d1 (distance between p1 and p2) throughout the gait cycle. Although the pattern is consistent, the change in d1 at MFC is shown to fluctuate by approximately 0.5 cm.

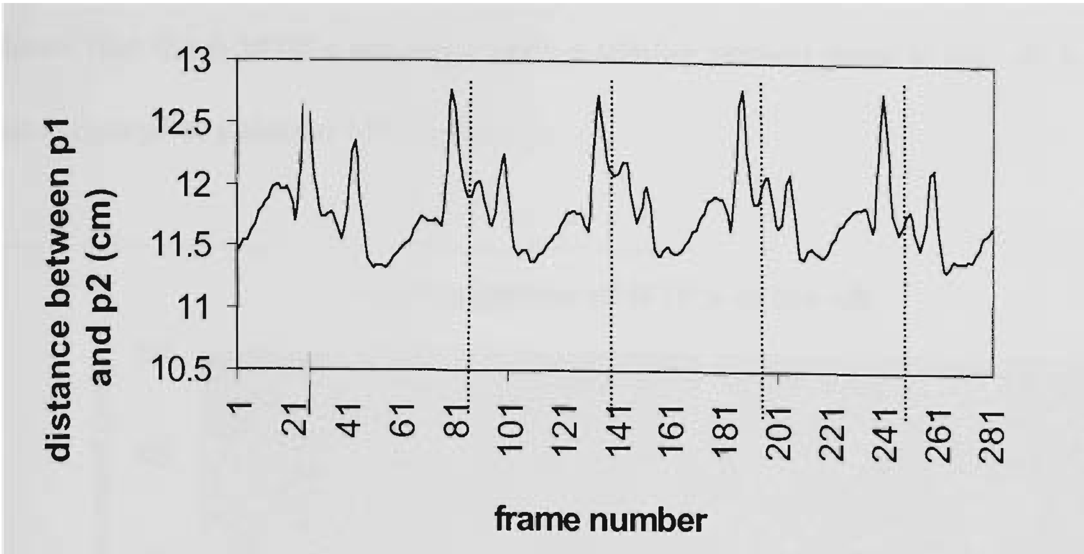

Figure 4.1.4.1. Change in the distance between the toe and 5th metatarsal head markers over several gait cycles. Vertical dotted lines represent successive MFC events.

Table 4.1.4.1 shows the results for changes in d1 at the MFC event and throughout the full gait cycle. The results show that the variability in d1 at the MFC event is similar to the variability in d1 occurring throughout the gait cycle.

Table 4.1.4.1. Changes in the relationship between the toe and 5th metatarsal head markers over several gait cycles.

	Mean	Standard deviation
Change in d1 Full gait cycle	11.82 cm	± 0.33
Change in d1 At MFC	11.79 cm	± 0.28

4.1.5 MFC error due to foot model digitizing error

Figure 4.1.5.1 demonstrates the time series graphs of the 6 MTP's digitized over four MFC events. The results are taken from the pilot study discussed in Section 3.6.3.3. The graph shows that the 6 MTP's converge upon a similar vertical point at toe off, however, they do not converge at a similar MFC point.

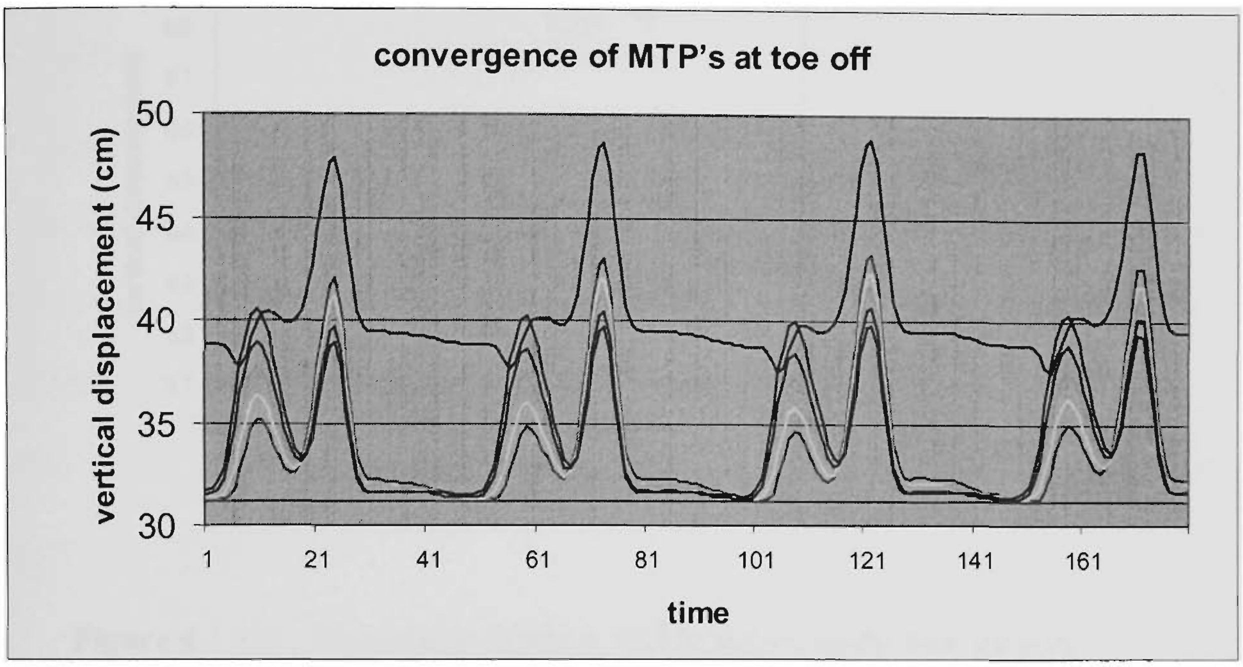

Figure 4.1.5.1. The trajectory patterns of the 6 digitised points over three MFC cycles. Note: all digitized points converge at the same point at toe off (1st minimum point). The lowest MTP is the last MTP to leave the ground.

Figure 4.1.5.2 shows the 6 digitized MTP's of a subjects right and left shoes. Figure 4.1.5.1 demonstrated that not all points from the shoe-outsole surface converge at the same vertical point at MFC. Therefore, the 6 MTP's were for determining the lowest point of the shoe out-sole surface that consistently came closest to the ground.

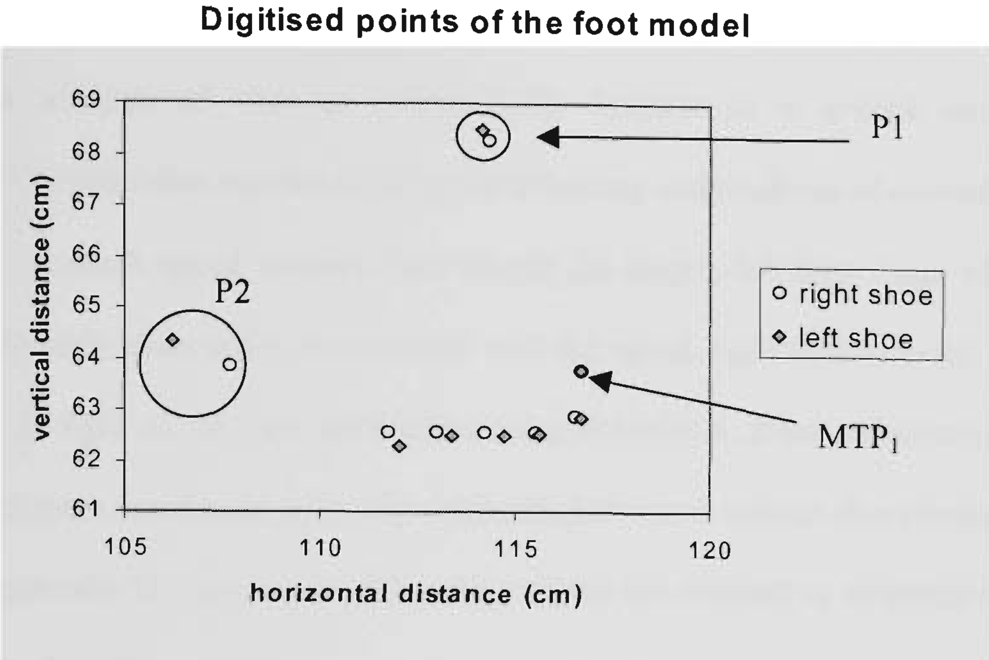

Figure 4.1.5.2 Digitisation of the 6 MTP's located on the shoe out-sole of the left and right feet (e.g. MTP$_1$, MTP$_2$, ...). The point closest to the toe marker (P1) is MTP$_1$. The fifth metatarsal head is represented by the digitized points at P2. The 6 points on the shoe surface provided the coordinates for 6 different foot models per limb.

4.1.6 MFC error due to ground reference position

The result for the standard deviation of the toe off position was obtained from the pilot study outlined in section 3.6.3.3. The appropriate MTP was determined from a graph from the lowest average MFC value derived from the 6 MTP coordinates. The standard deviation calculated for the toe off position was ±0.03 mm. Further results from the two neighbouring MTP coordinates were also investigated and revealed similar results.

4.2 SUBJECT CHARACTERISTICS

Table 4.2.1 compares the characteristics between the groups, p values are reported from multiple analysis of variance (MANOVA). Because of a limited sample size, three MANOVA tests were conducted using the following combinations of dependent variables: (i) weight, treadmill speed, average limb length; (ii) height, left limb range of plantar flexion, age; (iii) body mass index, overground walking speed, right limb average range of plantar flexion. In light of the limitations when using MANOVA, these combinations of dependent variables were considered relatively unrelated. Individual subject characteristics are provided in the appendix D. The young adult subjects were not required to complete the timed up and go test or the efficacy survey.

There were no significant differences between height, weight or body mass index measures. Generally, the elderly group was shorter and lighter while the BMI is comparatively similar. Not surprising, the age difference between the two groups is significant (p = .000).

Flexibility of the ankle joint is significantly different between the elderly and the young adult groups for both the left (p = .002) and the right (p = .003) limbs. There was no significant difference with limb length, or limb length discrepancies.

A trend is evident in the overground walking speed (p = .101), which is consistent with the literature on aging affects. A paired t-test was applied to investigate within group differences between treadmill walking and overground walking speeds. Interestingly, the elderly group showed a significant (p < .01) decrease in walking speed when walking on the treadmill. The ankle flexibility is significantly less in the elderly group, when comparing for both the left (p = .011) and right (p = .007) limbs.

Table 4.2.1. Group differences in baseline characteristics of subjects. NS is not significant at $p < .05$.

	Young	Elderly	*p* value
Number	10	8	-
Gender (% women)	50%	37.5%	-
Height (meters)	1.75 ± 1.02	1.64 ± 0.10	.028
Weight (kg)	71.40 ± 10.50	67.10 ± 8.40	.358
Body mass index (kg/m^2)	23.30 ± 2.50	25.10 ± 2.80	.152
Age (yrs)	30.30 ± 2.90	73.10 ± 3.80	.000
Average limb length (m)	0.84 ± 0.06	0.78 ± 0.06	.050
Left ankle ROM	84.10 ± 10.70	64.60 ± 11.80	.011
Right ankle ROM	88.80 ± 11.30	62.00 ± 19.40	.007
Overground gait speed (km/h)	5.43 ± 0.54	4.98 ± 0.55	.101
Treadmill speed (km/h)	5.43 ± 0.54	4.18 ± 0.47	.000
Tinetti efficacy scale (/140)	-	137.50 ± 3.78	-
Timed up & go test (sec)	-	8.49 ± 1.16	-

4.3 NORMALITY TESTING

This section reports on the descriptive measures characterizing the MFC distribution. Comparisons are made between: (i) subject groups; and (ii) limbs. The first part analyses the normality of the distribution and is followed by exploring the descriptive measures of the MFC distribution.

4.3.1 Normality testing of the MFC distribution

In order to assess aging affects between the measured parameters it is important to investigate the underlying structure of the MFC data. It has been shown that MFC data collected from treadmill walking is not normally distributed (Best et al., 1999) and therefore parametric statistics may not be appropriate. The nature of the data for individual subject's left and right MFC distribution is described in Table 4.3.1.1.

The Kolmogorov-Smirnov statistic with a Lilliefores significance level is a common test of normality for sample sizes larger than 50 (Coakes and Steed, 1999). Where significance is greater than .05, the distribution is considered normal (Coakes and Steed, 1999). The results of these tests and the results from standardized skewness and kurtosis tests (see Section 3.7) are presented in Table 4.3.1.1. In most cases, values exceeding ±2 for standardized skewness and kurtosis are supported by the results from the Kolmogorov-Smirnov statistic.

Table 4.3.1.1. Individual intra- limb MFC normality results for the Lilliefors statistic, and the standardized z scores of skewness and kurtosis. All values below the critical ± 2 for skewness and kurtosis are represented with #. Values from the Lilliefors statistic that are above the critical p > .05 level are represented by *. Cases without a * or # represent a non-normal distribution.

Subject	Lilliefors *p* value		Skewness Z value		Kurtosis Z value	
Young	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**
Y1	.008	.000	2.30	6.25	4.55	5.02
Y2	.004	.000	6.48	14.90	8.74	22.44
Y3	.091*	.000	7.09	8.82	14.09	7.08
Y4	.000	.000	7.76	7.94	3.89	6.28
Y5	.013	.000	7.51	7.27	7.26	0.78#
Y6	.000	.000	14.25	11.10	15.89	6.62
Y7	.000	.000	12.51	12.01	16.57	9.31
Y8	.200*	.098*	0.54#	2.68	3.96	1.67#
Y9	.000	.000	19.04	22.22	25.24	50.19
Y10	.000	.002	16.19	3.46	11.88	8.29
Elderly						
E1	.000	.000	13.38	13.51	27.74	11.74
E2	.000	.000	8.02	1.17#	10.83	24.17
E3	.000	.008	15.32	20.61	31.62	70.44
E4	.200*	.000	2.69	10.32	4.72	7.69
E5	.002	.006	4.23	4.62	6.18	3.73
E6	.000	.000	7.49	3.65	6.75	1.19#
E7	.056*	.000	5.65	8.42	4.04	10.58
E8	.001	.000	8.36	16.83	11.59	27.14

4.4 DESCRIPTIVE STATISTICS

4.4.1 Measures of central tendency and dispersion

Table 4.4.1.1 lists the measures of central tendency and dispersion for measures of mean and standard deviation and also the median and the inter-quartile range. Measures of central tendency and dispersion are reported for individual intra-limb MFC data. To account for both normal and non-normal data distributions measures are reported as mean ± standard deviation and median ± inter-quartile range.

Group averages show the median MFC is lower than the mean MFC. This is due to positive skewness occurring in most data distributions. Subject Y4 was the only subject to return an alternate result. This is not surprising because the between limb central tendency difference is extremely small for this subject. In most cases only minor differences exist between the mean and median. Hence for all other cases, clarifying the lowest MFC limb is consistent between the mean and median statistics. When differentiating between the low MFC limb and the high MFC limb the mean better represents central tendency. From this point on, the limb that is described by the low MFC mean will be referred to as the low MFC limb.

Table 4.4.1.1 Individual intra-limb MFC central tendency and dispersion. *indicates limb at risk of contacting an unseen obstacle while walking. # indicates discrepancy between measures of central tendency - mean and median statistics - when identifying the lowest MFC limb.

Subject	Mean ± standard deviation		Median ± inter-quartile range	
Young	**Left**	**Right**	**Left**	**Right**
Y1	1.26 ± 0.20	0.66 ± 0.19*	1.26 ± 0.26	0.64 ± 0.25
Y2	2.28 ± 0.39	1.23 ± 0.26*	2.26 ± 0.50	1.21 ± 0.32
Y3	1.45 ± 0.25	1.33 ± 0.28*	1.44 ± 0.33	1.31 ± 0.35
Y4	1.50 ± 0.40*	1.50 ± 0.56	1.45 ± 0.50	1.44 ± 0.73#
Y5	1.53 ± 0.24	1.16 ± 0.35*	1.52 ± 0.31	1.12 ± 0.46
Y6	0.89 ± 0.28*	0.91 ± 0.33	0.86 ± 0.34	0.88 ± 0.43
Y7	1.46 ± 0.33	1.38 ± 0.35	1.43 ± 0.42	1.35 ± 0.44
Y8	1.53 ± 0.18	0.89 ± 0.19*	1.53 ± 0.24	0.88 ± 0.25
Y9	1.17 ± 0.22	0.57 ± 0.21*	1.14 ± 0.26	0.54 ± 0.25
Y10	1.34 ± 0.29	1.23 ± 0.21*	1.30 ± 0.36	1.23 ± 0.26
Group	**1.44 ± 0.28**	**1.09 ± 0.29**	**1.42 ± 0.36**	**1.06 ± 0.31**
Elderly				
E1	1.47 ± 0.30	1.00 ± 0.36*	1.45 ± 0.36	0.95 ± 0.46
E2	1.49 ± 0.35	0.78 ± 0.23*	1.47 ± 0.43	0.76 ± 0.29
E3	1.74 ± 0.35*	1.76 ± 0.31	1.73 ± 0.45	1.74 ± 0.41
E4	1.32 ± 0.20	0.57 ± 0.22*	1.32 ± 0.26	0.55 ± 0.29
E5	0.72 ± 0.20*	1.83 ± 0.30	0.72 ± 0.25	1.82 ± 0.40
E6	0.95 ± 0.24*	1.43 ± 0.45	0.93 ± 0.30	1.39 ± 0.59
E7	1.18 ± 0.26	0.86 ± 0.21*	1.17 ± 0.36	0.85 ± 0.28
E8	1.52 ± 0.29	1.02 ± 0.37*	1.51 ± 0.39	1.07 ± 0.47
Group	**1.30 ± 0.27**	**1.16 ± 0.31**	**1.29 ± 0.33**	**1.14 ± 0.46**

Eight of the ten (80%) young subjects and five from the eight (63%) elderly subjects have a higher left MFC limb in comparison to the right limb. A Wilcoxen Rank Sum W test reveals a significantly (p = .023) higher MFC in the left limb for the young group. There is no significant difference between the left and right limbs for the elderly group.

Comparing within groups, differences between male and female did not reveal significant differences (elderly, p = .278; young, p = .173) when pooling the intra-limb mean MFC results. An associated test on the DFA revealed similar results (elderly, p = .329; young, p = .257). Based upon these findings, and the limited sample size, further analysis did not differentiate between male and females within or between groups.

4.4.2 Measures of skewness and kurtosis.

Histograms for all subjects are displayed in Figures 4.4.2(A) - 4.4.2(E). Common characteristics of most histograms are features of the positive skewness (S) and kurtosis (K) of the distributions. A high coefficient of variation (CV) can be observed from low central tendency and a relatively large dispersion characteristics of the histograms. See Table 4.4.2 for the quantitative values of S, K and CV. Skewness is evident by the values to the left of the curve not ranging (distribution closer to the mean) in the same manner as the values to the right of the curve (distribution further from the mean). Generally, comparisons between the left and right for most subjects show that distributions closer to zero appear to have higher kurtosis (more data points around the mean than normal K>0; leptokurtic) and positive skewness (S>0).

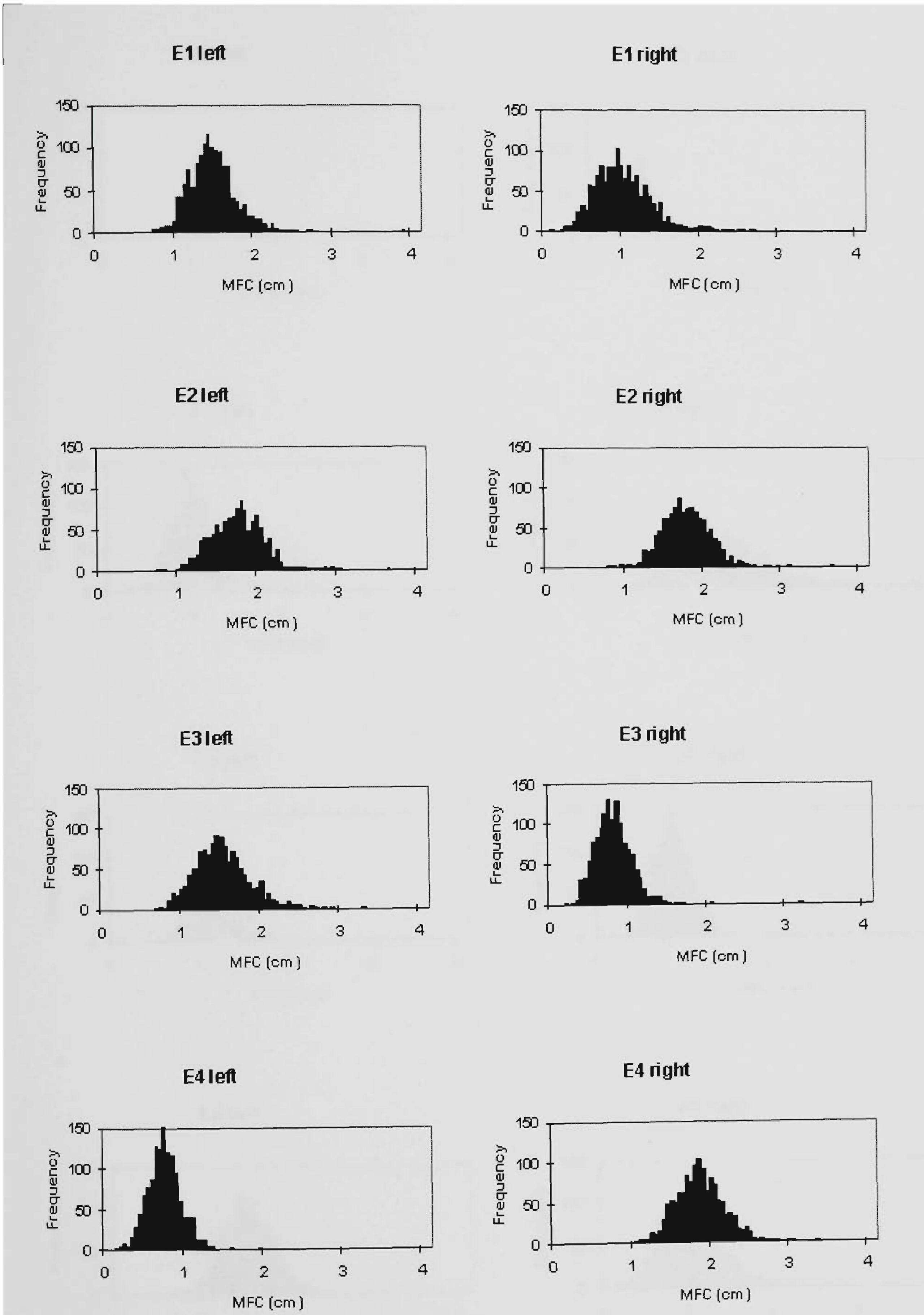

Figure 4.4.2.1(A). Elderly subjects E1 to E4 left and right MFC histograms.

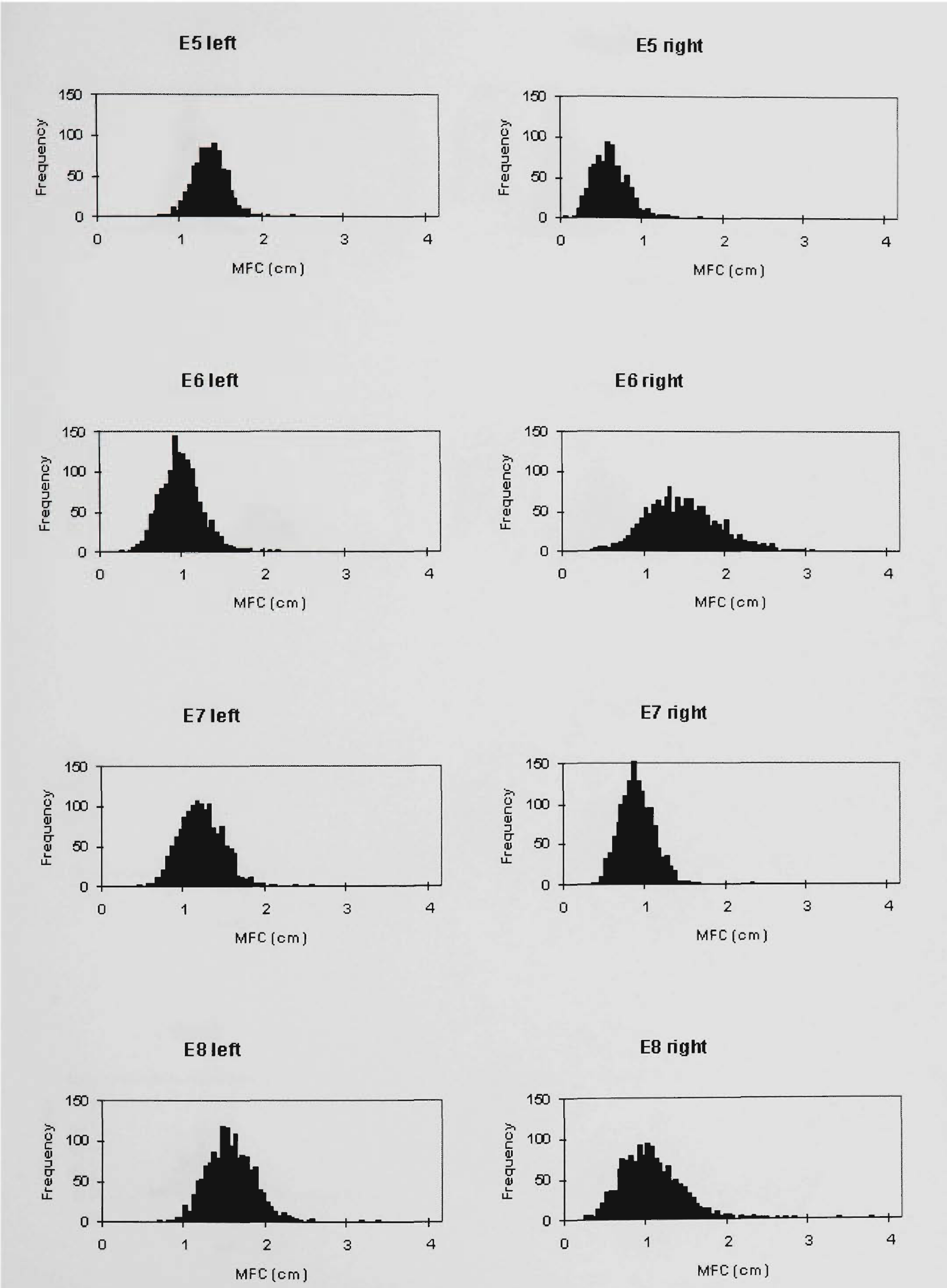

Figure 4.4.2.1 (B) Elderly subjects E5 to E8 left and right MFC histograms.

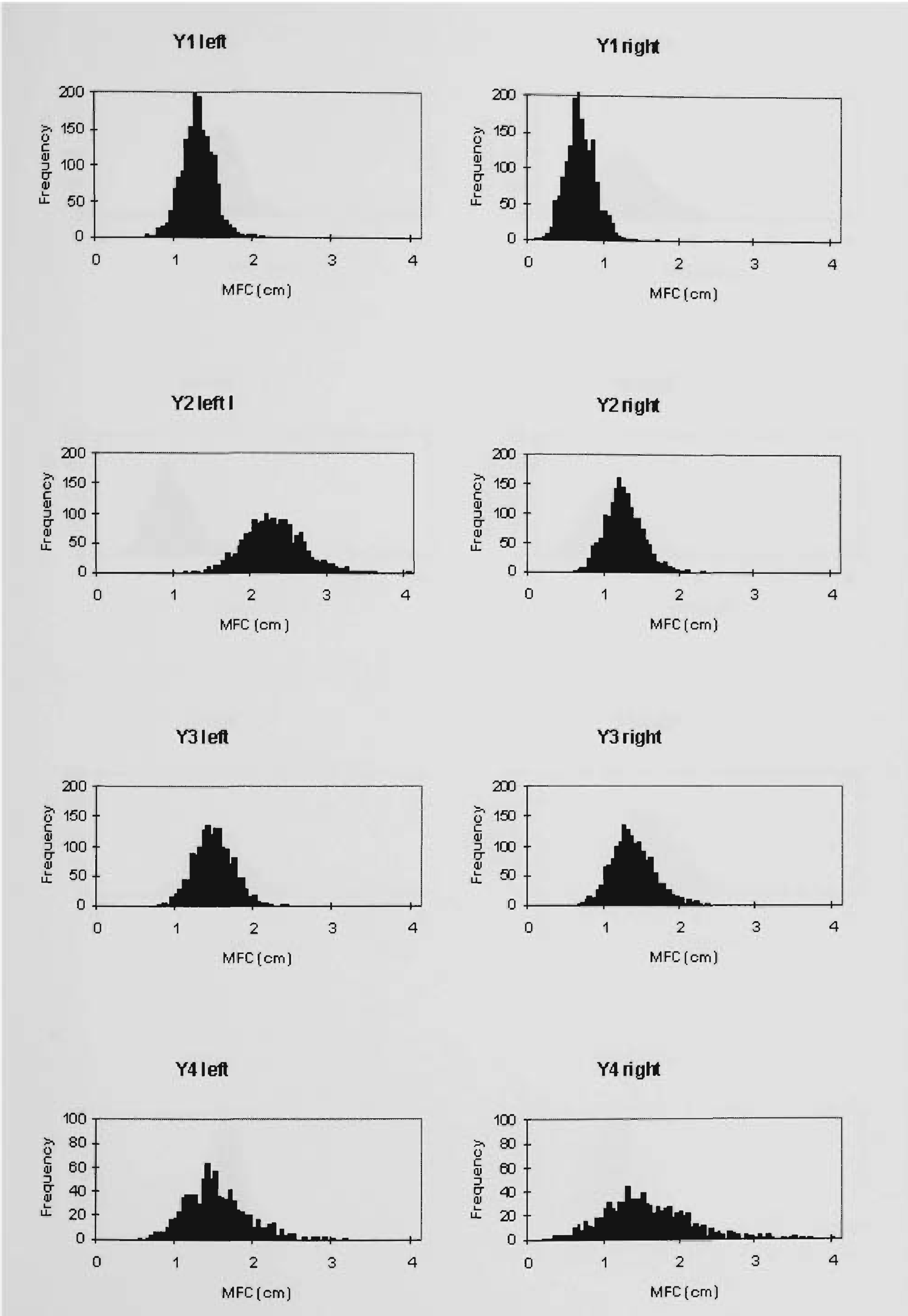

Figure 4.4.2.1 (C). Young subjects Y1 to Y4 left and right MFC histograms.

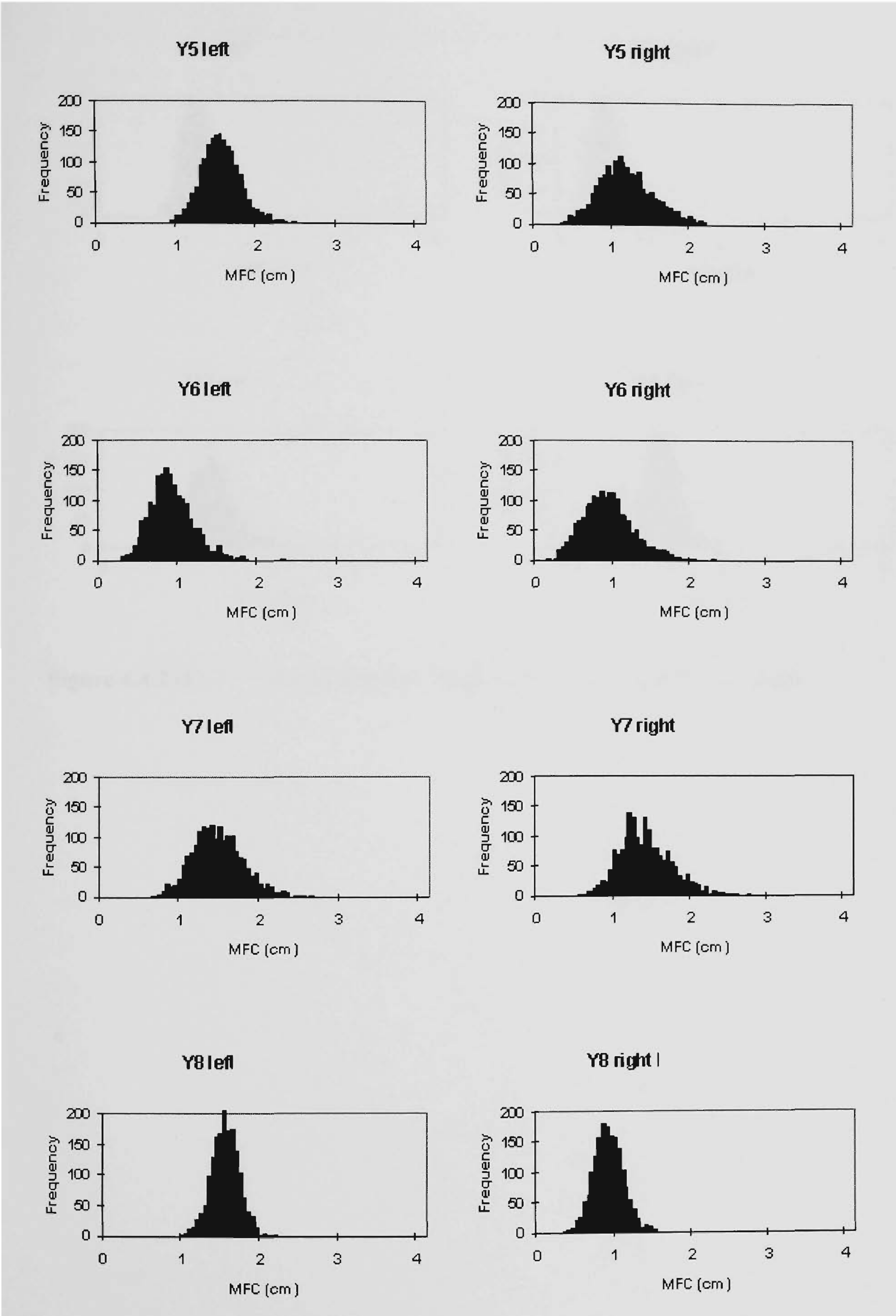

Figure 4.4.2.1 (D). Young subjects Y5 to Y8 left and right MFC histograms.

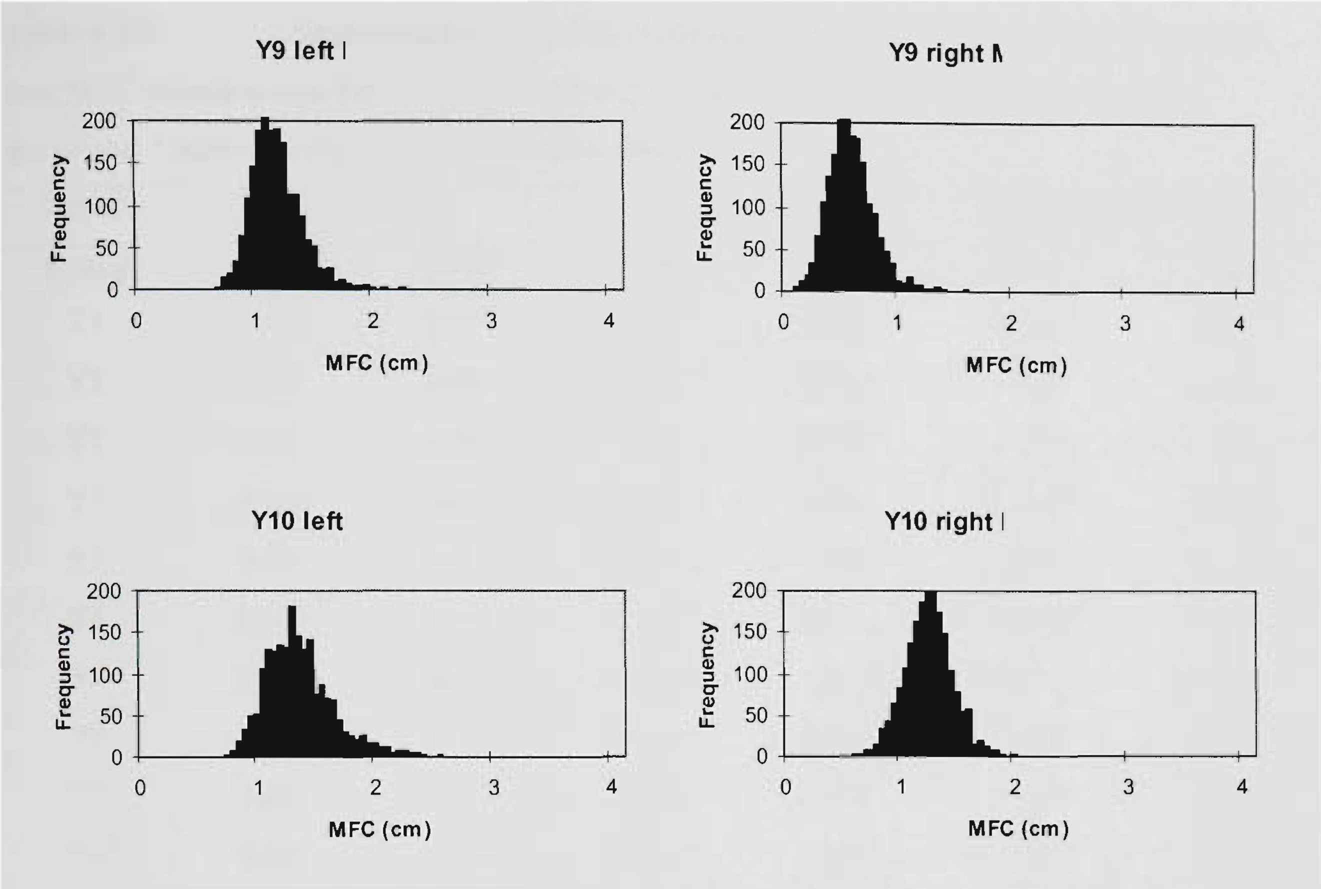

Figure 4.4.2 (E). Young subjects Y9 and Y10 left and right MFC histograms.

Table 4.4.2. Skewness, kurtosis and coefficient of variation of the left foot and right foot MFC distributions for all subjects. Group results are provided as mean ± standard deviation; * indicates the lowest mean MFC limb.

Subject	Skewness		Kurtosis		Co efficient of Variation	
Young	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**
Y1	0.14	0.37*	0.54	0.60*	15.53	29.21*
Y2	0.39	0.89*	1.04	2.68*	17.25	21.03*
Y3	0.44	0.54*	1.73	0.87*	17.28	21.16*
Y4	0.64*	0.66	0.64*	1.04	26.43*	37.37
Y5	0.45	0.43*	0.86	0.09*	15.81	30.59*
Y6	0.83*	0.64	1.84*	0.77	31.58*	36.64
Y7	0.72	0.69*	1.90	1.07*	22.79	25.04*
Y8	0.03	0.16*	0.47	0.20*	11.56	21.83*
Y9	1.07	1.25*	2.84	5.67*	18.57	37.27*
Y10	0.90	0.19*	1.32	0.92*	21.55	17.22*
Group	**0.56 ± 0.33**	**0.58 ± 0.33**	**1.32 ± 0.76**	**1.39 ± 1.66**	**19.84 ± 5.88**	**21.72 ± 4.21**
Elderly						
E1	0.87	0.88*	3.63	1.54*	20.25	36.21*
E2	0.58	0.08*	1.55	3.47*	19.90	17.40*
E3	1.04*	1.39	4.28*	9.53	23.80*	29.76
E4	0.18	0.70*	0.64	1.05*	14.82	38.07*
E5	0.35*	0.38	1.03*	0.62	28.15*	16.39
E6	0.48*	0.42	0.87*	0.15	25.85*	31.15
E7	0.37	0.55*	0.53	1.39*	22.15	24.55*
E8	0.53	1.07*	1.47	3.44*	18.86	36.63*
Group	**0.55 ± 0.28**	**0.69 ± 0.42**	**1.75 ± 1.42**	**2.65 ± 3.03**	**27.73 ± 7.56**	**28.77 ± 8.54**

The within group Wilcoxen Rank Sum W test revealed a significantly higher CV in the right limb for the young subject group. There are no differences between the left and right limbs within either of the groups for all other parameters. This was also confirmed by t-tests.

4.4.3 Differences between groups for pooled MFC data

Figure 4.4.3.1 shows group comparisons for pooled left and right within limb descriptive statistics. Wilcoxen Rank Sum W tests do not reveal significant differences between the groups. The graphs do not reveal large effect sizes (ES) between the groups for all the descriptive parameters characterising the MFC distribution. The strongest difference of the results was a weak to moderate effect size for kurtosis (ES = 0.47). In general, the group means show that the elderly group has a lower MFC, higher SD, CV, S, and K when compared with the young group.

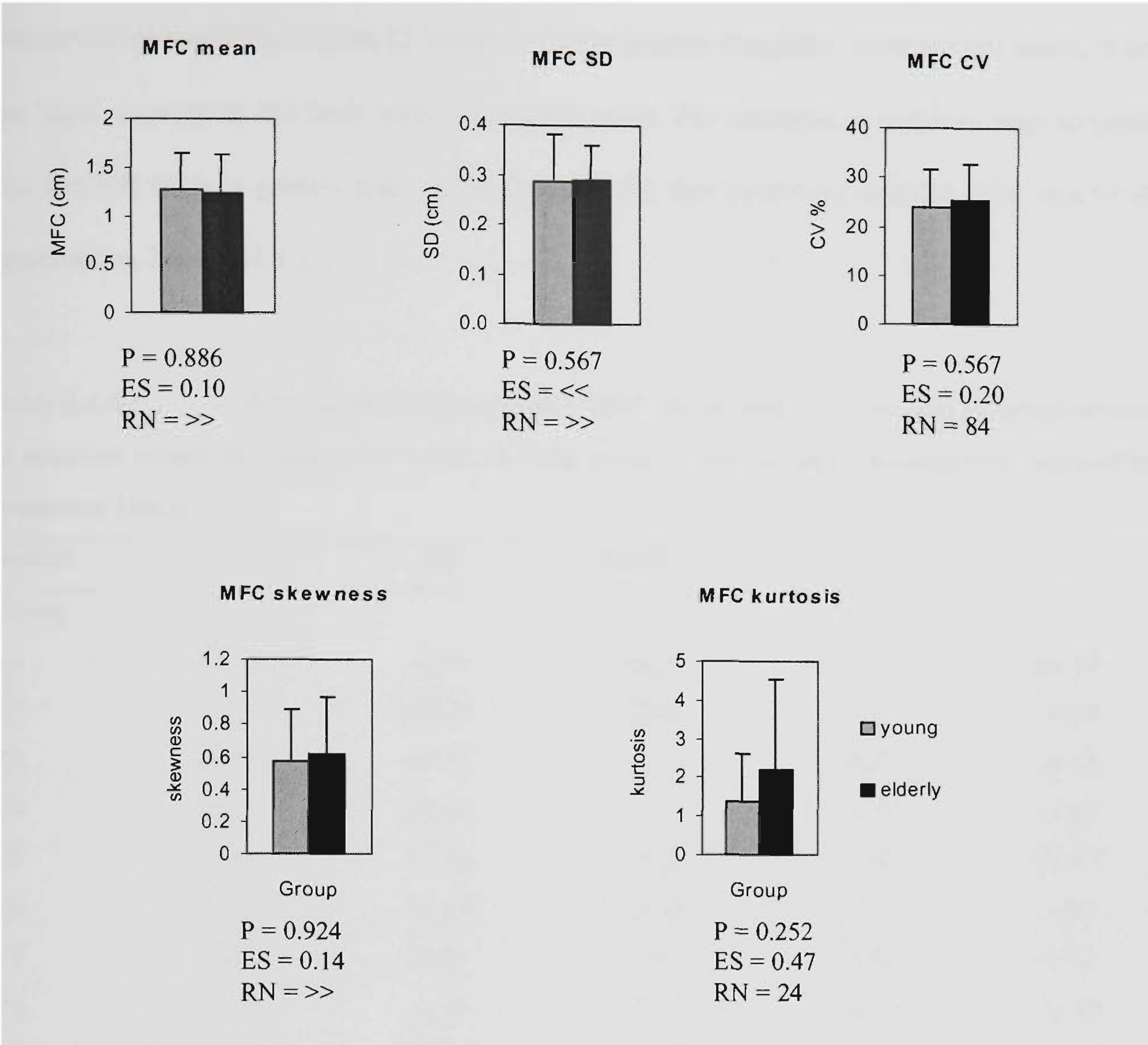

Figure 4.4.3.1. Between group differences for pooled intra- limb MFC data. The five column graphs from top left to right are for: mean, standard deviation, and coefficient of variation. The bottom panel from left to right are: skewness and kurtosis. The effect sizes (ES) and significance values (P) are indicated below each graph. RN represents the required sample size to obtain power of 0.8 at a significance level of .05.
Symbolisms: >> indicates very large values; << indicates very small values.

4.4.4 Symmetry Index (SI) for MFC descriptive parameters

In some cases the histograms show small differences in shapes of the distribution between limbs while in other cases the between limb differences are large. This was observed in Table 4.4.2 and Figures 4.4.2 (A)-(E). To quantitatively examine the inequality between limbs the symmetry index (SI) described in section 3.5.2.3 was applied to the left and right MFC descriptive parameters. Higher SI values indicate greater inequality between the limbs, while the 'sign' represents the limb with the higher value. For example, a negative sign indicates that the left limb is greater than the right limb for that particular statistic. The results are presented in Table 4.4.4.1.

Table 4.4.4.1 Asymmetry of descriptive MFC parameters. Group means are presented as absolute symmetry index (SI) values. Group mean ± SD represent the absolute value of the symmetry index.

Subject	Mean	SD	Skewness	Kurtosis	CV
Young					
Y1	-62.92	-2.06	92.34	9.84	61.15
Y2	-59.37	-40.74	78.78	87.89	19.74
Y3	-8.63	11.63	21.72	-66.20	20.15
Y4	0.27	34.48	2.31	46.98	34.28
Y5	-27.85	37.58	-3.19	-161.05	63.69
Y6	1.56	16.29	-24.88	-82.39	14.82
Y7	-5.00	4.42	-4.12	-56.09	9.39
Y8	-53.10	9.16	133.33	-81.55	61.50
Y9	-69.43	-2.80	15.40	66.16	66.98
Y10	-8.27	-30.40	-129.50	-35.59	-22.36
Group	**29.64 ± 28.45**	**18.96 ± 15.30**	**50.56 ± 52.79**	**69.37 ± 40.07**	**37.41 ± 23.22**
Elderly					
E1	-38.47	18.84	1.02	-81.07	56.52
E2	1.20	-12.27	-149.09	76.24	-13.41
E3	-62.49	-41.64	29.47	76.08	22.27
E4	-79.77	10.22	117.38	47.90	87.94
E5	86.90	38.42	8.72	-49.54	-52.79
E6	40.64	58.35	-13.78	-140.00	18.58
E7	-31.01	-21.19	39.39	89.53	10.28
E8	-39.53	26.32	67.25	80.33	64.04
Group	**47.50 ± 27.84**	**28.41 ± 16.52**	**53.26 ± 54.14**	**80.09 ± 28.48**	**40.73 ± 28.47**

The data in Table 4.4.4.1 illustrates the strategies to reduce ground contact risk. In subject E8, the MFC asymmetry value is negative, indicating that the right limb is lower. The SI values are positive for all other listed parameters. The positive value in the standard deviation indicates that the low MFC limb, the right in this case, has higher standard deviation, higher skewness and kurtosis. The positive SI value in the coefficient of variation also indicates that the right limb has increased SD relative to the mean.

The SI investigates whether there exists a common within-subject strategy that can be revealed by relationships between the MFC SI and the SI values associated with other descriptive parameters. Table 4.4.4.2 explores these relationships among the descriptive statistics for the SI. The correlation matrix calculated the relationship among these parameters using Spearman's 'r', rank order test. The only significant correlation returned is the moderate relationship between the mean and the coefficient of variation (r = 0.512, p = .018).

Table 4.4.4.2 Correlations between SI measures for descriptive parameters for all subjects. *p* values are in parenthesis, NS – not significant at the .05 level. MFC – minimium foot clearance; SD – standard deviation; S – skewness; and, K – kurtosis.

	MFC	SD	S	K
SD	NS			
S	NS	NS		
K	NS	NS	NS	
CV	0.552 (.018)	NS	NS	NS

4.4.5. Between group differences for SI

To determine group differences of asymmetry the following bar charts are provided (Figure 4.4.5.1). The relative effect size and significance level from Wilcoxen Ranked Sum W tests are provided below each chart.

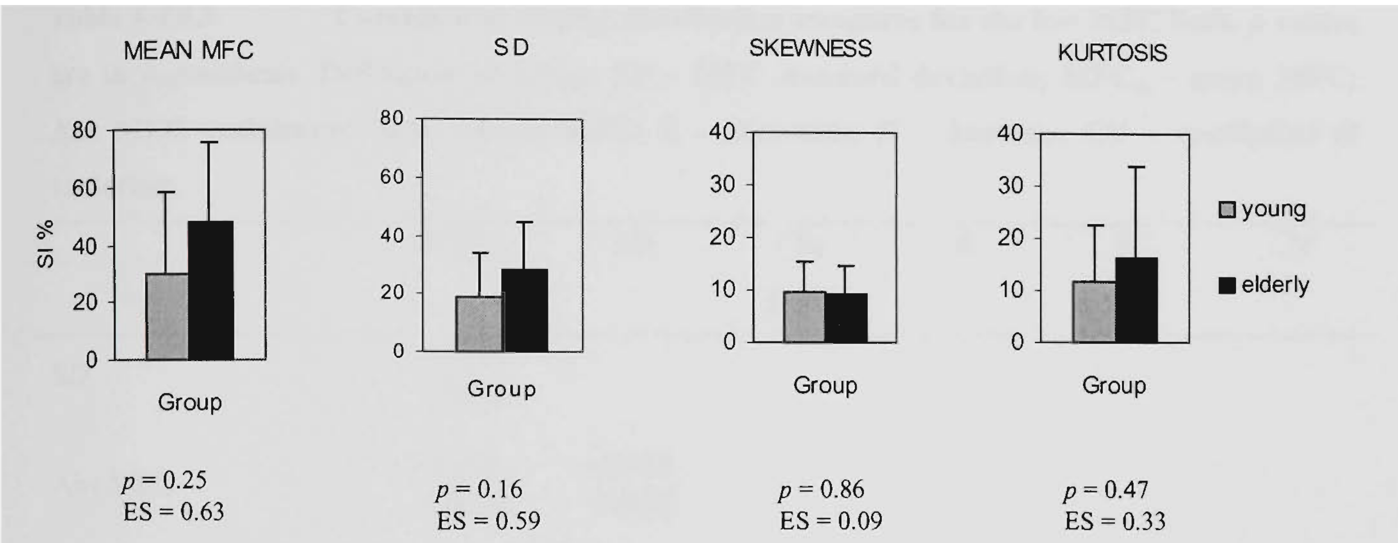

Figure 4.4.5.1 Symmetry index (SI) values for between group comparisons. The group SI values are absolute in that they do not indicate which limb has the higher value of the specified parameter. Effect sizes (ES) and significance values (*p*) are shown below the graphs.

No significant differences exist between groups for the SI descriptive statistics of the MFC. Apart from skewness, higher mean values of absolute SI accompany the elderly group. The between group differences of SI for the mean and standard deviation show a moderate to strong effect size (MFC $_{MEAN}$ = 0.63; MFC $_{SD}$ = 0.59). The coefficient of variation, skewness and kurtosis show weak effect sizes.

4.4.6 Low MFC limb correlations with MFC descriptive parameters

Correlations within the low and high MFC limb descriptive parameters were explored for determining whether there exists a phenomenon present within the low MFC limb that may not exist within the high MFC limb. This approach is different to the symmetry index because it does not consider the relative effects that can be caused by values associated with the higher MFC limb. Two correlation matrices determined the correlations between the descriptive parameters within the (i) low MFC limb (Table 4.4.6.1), and (ii) the high MFC limb (Table 4.4.6.2).

Table 4.4.6.1 Correlations among distribution measures for the low MFC limb. *p* values are in parenthesis. Definition of terms: SD – MFC standard deviation; MFC_m – mean MFC; Abs MFC –minimum value of the MFC; S – skewness; K – kurtosis; CV – coefficient of variation.

	MFC_m	SD	Abs MFC	S	K	CV
SD	.710 (.001)					
Abs MFC	0.673 (.002)	-0.438 (.069)				
S	NS	0.411 (.090)	NS			
K	NS	NS	NS	.849 (.000)		
CV	-0.591 (.010)	NS	-0.759 (.000)	.505 (.033)	NS	
Treadmill Speed	NS	NS	NS	NS	NS	NS

The data from table 4.4.6.1 shows that skewness has a significant positive correlation with kurtosis (r = .849, p = .000) and the coefficient of variation (r = .505, p = .033) when the low MFC limb is taken into consideration. The low MFC limb also has a strong relationship with the CV (r = -.591, p = .01), SD (r = 0.710, p = .001) and absolute MFC value (r = 0.673, p = .002). The absolute MFC (Abs MFC) for the low MFC limb is also significantly correlated with the coefficient of variation (r = -0.759, p = .000). The standard deviation shows a trend towards being correlated with skewness and the absolute MFC.

Table 4.4.6.2 shows that the correlations are different for the high MFC limb. In contrast to the low MFC limb, MFC_m did not show a correlation with the standard deviation (SD) or the coefficient of variation (CV). The CV is significantly correlated with the SD, absolute MFC and skewness.

Table 4.4.6.2 Correlations among MFC distribution measures for the high MFC limb. *P* values are in parenthesis. Definition of terms: SD – standard deviation; MFC_m – mean MFC; Abs MFC –minimum value of the MFC; S – skewness; K – kurtosis; CV – coefficient of variation.

	MFC_m	SD	Abs MFC	S	K	CV
SD	NS					
Abs MFC	0.518 (.028)	NS				
S	NS	NS	NS			
K	NS	NS	NS	.622 (.006)		
CV	NS	0.750 (.000)	-0.604 (.008)	.610 (.007)	NS	
Treadmill Speed	NS	NS	NS	NS	NS	NS

Common findings in Tables 4.4.6.1 and 4.4.6.2 are consistent for both skewness and treadmill speed. Treadmill speed is not correlated with any of the parameters. Skewness is significantly correlated with the CV for the low MFC limb ($r = 0.505, p = .033$) and the high MFC limb ($r = 0.610, p = .007$). The correlation matrices showed that skewness and kurtosis are not associated with the mean of the low MFC limb or the high MFC limb.

4.4.7 Between group differences for the low MFC limb

Figures 4.4.7.1 and 4.4.7.2 illustrate associations of MFC and MFC descriptive parameters between groups. There are significant differences within both groups when comparing the mean MFC heights. The young group's high MFC limb (1.44 ± 0.35) was significantly higher (p = .005) than the low MFC limb (1.08 ± 0.32). In comparison, the elderly group's high MFC limb (1.50 ± 0.21) was significantly different (p = .005) to the low MFC limb (0.95 ± 0.35). No other significant differences were found within each group when comparing the relative parameters. Effect sizes were strong within the elderly for differences between high and low MFC limbs for standard deviation and skewness. Figure 4.4.7.1 illustrates between group differences, such that the elderly have lower standard deviation and higher skewness and kurtosis associations with the low MFC limb. The column graphs generally show compensating strategies from comparisons between the limbs across both groups. Although none of these differences are statistically significant (p<.05) SD is lower, and S and K are higher in the low MFC limb compared to the high MFC limb. These differences appear more evident in the elderly group.

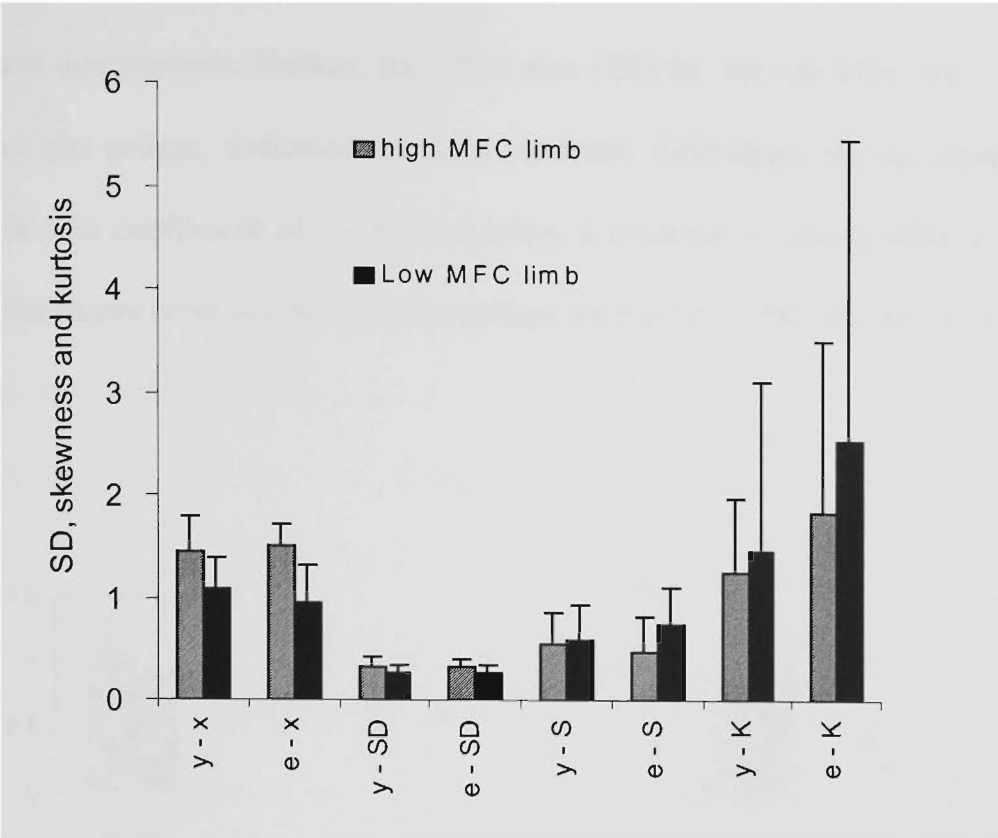

Figure 4.4.7.1. The presence of a compensating strategy at the low MFC limb. Horizontal categories indicate mean (x), standard deviation (SD), skewness (S) and kurtosis (K) for groups y (young) and elderly (e).

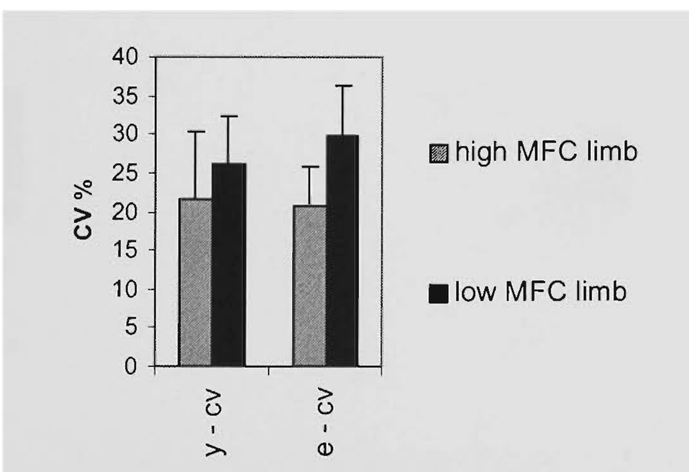

Figure 4.4.7.2. The low MFC and high MFC limb results for the CV. Definition of terms: y – young; e – elderly.

For the low MFC limb, the between group differences are displayed in Figure 4.4.7.3. No significant differences existed between the groups for the mean, coefficient of variation, skewness and kurtosis. Further, the effect size (ES) for the low MFC limb when comparing between the groups, indicated weak to moderate differences for the mean, skewness and kurtosis. The coefficient of variation displays a moderate to strong effect size. The standard deviation shows no effect between the groups for the low MFC limb and is not presented.

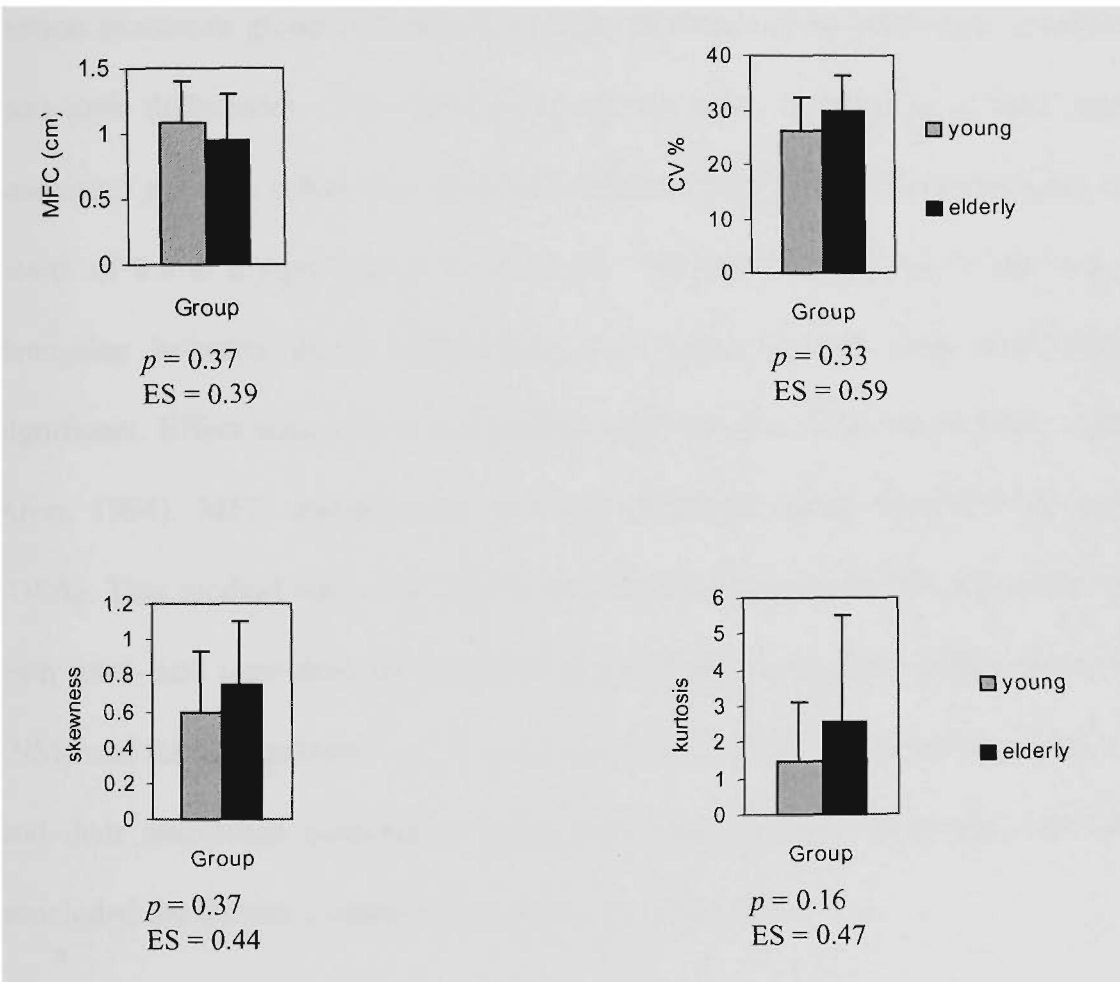

Figure 4.4.7.3. Between group results for the low MFC limb. p values and effect sizes (ES) are provided below each graph.

4.5 DYNAMIC VARIABILITY MEASURES

This section uses several methods to quantitatively describe the characteristics of the MFC dynamics. The results have so far demonstrated that comparing between the left and right limbs may not be appropriate based upon the likelihood that certain characteristics are more so related with the low and high MFC limbs. Thus, comparing between the low and high MFC limbs suggests a more suitable means for investigating bilateral MFC parameters. This section examines group differences in light of these issues while also considering 'pooled' intra-limb differences. The effect of aging has been reported in column graphs with the associated p value, effect size (ES) and required sample size (RN) necessary to obtain high power of 0.8 at a significance level of .05. Wilcoxen Rank Sum W test was performed to determine between group differences. A p value of $<.05$ was considered statistically significant. Effect sizes of 0.6 and greater was termed a moderate to strong effect (Aron and Aron, 1994). MFC unsteadiness was first described using detrended fluctuation analysis (DFA). This method was only applied to intra-limb processes. The following section reports both intra- and inter-limb dynamics from quantitative measures of the non-stationary index (NSI) and the inconsistency of the variance (IV). Finally, examination of the Poincaré plots and their associated quantitative descriptive measures for intra- and inter-limb processes concluded the dynamic unsteadiness analysis of the MFC.

4.5.1 Detrended fluctuation analysis (DFA)

4.5.1.1 Investigating DFA properties within MFC

A representative elderly subject serves as an example of the aging effects upon the relative MFC rhythm dynamics between limbs (Figure 4.5.1.1.1). Although visual inspection of the graphs reveal only small differences between the left and right MFC patterns, the right MFC limb is generally lower and demonstrates a more 'consistent' structure in comparison to the left limb, indicating asymmetry.

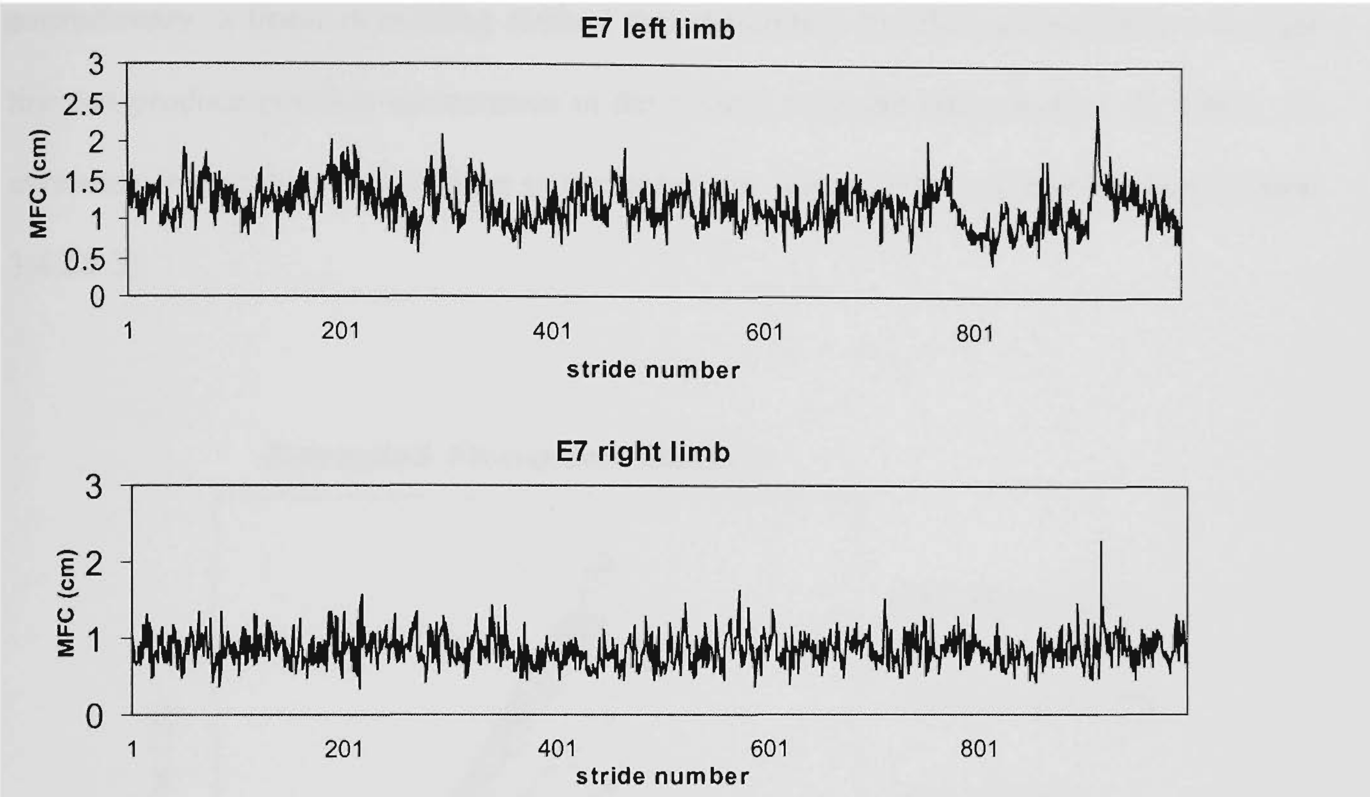

Figure 4.5.1.1.1 A representative example of an elderly (E7) subject who is displaying different structuring of the MFC dynamics between limbs. The left limb MFC is generally higher throughout the trial, and displays a wavelike pattern. In contrast, the right limb MFC is consistently lower and is without the same presence of a wavelike structure.

Detrended fluctuation analysis (DFA) demonstrates how the MFC differs between limbs for subject E7. The slope of the line relating log F(n) to log n is steeper for the left limb (Figure 4.5.1.1.1). In contrast, the right limb is less steep and represents fluctuations that are less highly correlated. For an inspection of all subjects line graphs, see appendix E.

The example in Figure 4.5.1.1.2 demonstrates a near perfect correlation for the regression line from the log F(n) to log n graph of subject E7. All the regression lines fitted to each data series are highly correlated. This may not always be the case and the accuracy of the scaling exponent is determined by a number of factors that influence the linear relationship between log n and log F(n). To overcome the factors that can lead to inaccuracies, several methods are employed to determine the most robust measure of the scaling exponent 'α'. When data is nonstationary, a linear detrending method that determines the fluctuations about a straight line can produce possible inaccuracies in the scaling exponent (Hausdorff et al., 1996). To overcome this problem, detrending is performed for higher order polynomials (see Section 3.4.2.2.3).

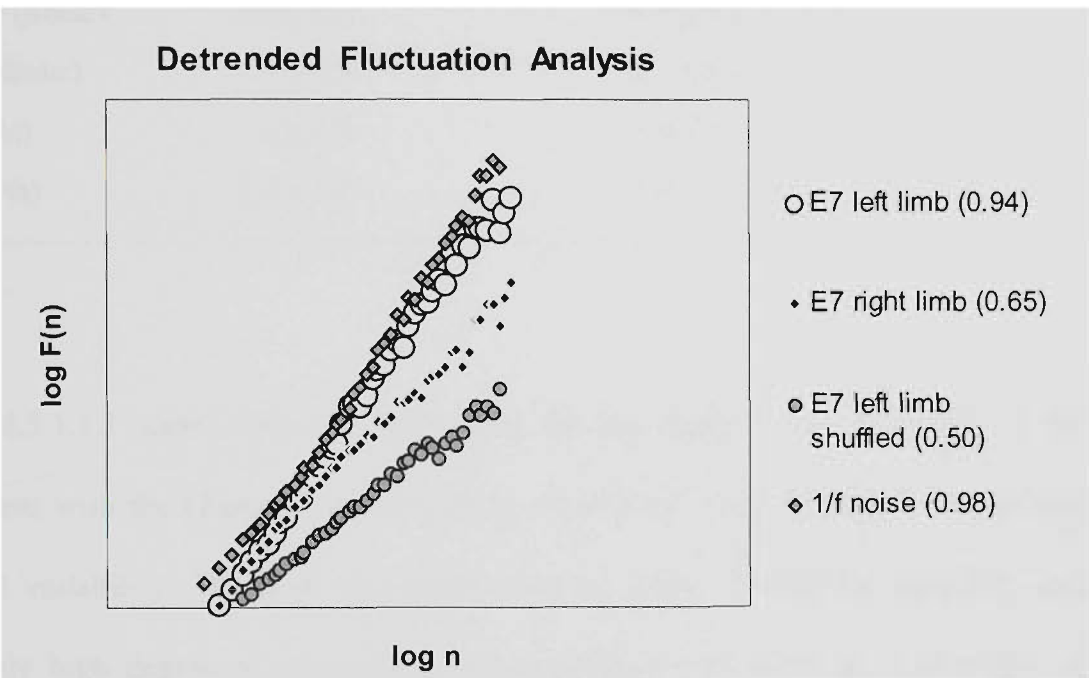

Figure 4.5.1.1.2. An example of detrended fluctuation analysis of the elderly subject E7, whose time series graphs are displayed in Figure 4.5.1.1.1. Four different data series are displayed on the log plot. The slope of the line relating log F(n) to log n determines the value of the scaling exponent. Gradients approaching 1.0 indicate 1/f type noise, those between 0.65 and 0.9 represent long-range correlations in the data, 0.5 represents random correlations (white noise). From the graph, the E7's left limb shows a DFA closer to 1/f noise ($\alpha = 1.0$), whereas the right limb is closer to white noise ($\alpha = 0.5$). When E7's left limb's MFC data is randomly shuffled, the DFA recognizes the random order, such that $\alpha = 0.5$.

The CV of the four DFA values calculated from higher order detrending is presented in Table 4.5.1.1.1. The mean value from the group of four DFA values from different order detrending represented the 'official' value reported in Table 4.5.1.2.1 (see section 4.5.1.2).

Table 4.5.1.1.1 Effect of higher order detrending on α. CV is the coefficient of variation.

	CV in α from four DFA values	
	Default Option:	Simulated 1/f noise (Hausdorff et al. 1996)
1st order (linear)	4.0 ± 4.1	0.1 ± 0.4
2nd (quadratic)	1.3 ± 0.9	0.5 ± 1.1
3rd (cubic)	1.7 ± 1.6	1.0 ± 1.6
4th (fourth)	2.2 ± 1.9	1.6 ± 1.9

Table 4.5.1.1.1 shows that the detrending for the higher order polynomials for MFC is consistent with the changes encountered by Hausdorff et al. (1996) for stride-to-stride time interval variability. The high discrepancy for the linear detrending procedure indicates the relatively high degree of non-stationarity associated with stride-to-stride MFC data series. The consistency for the 2nd to 4th order detrending values from the value reported allows a certain degree of confidence when determining between group differences.

The accuracy of the DFA is also related to the scaling region employed. Hausdorff et al. (1996) has used a number of different window regions (see Figure 2.7.3.1 and equation 2.5 of section 2.7.3 for further information) to determine the presence of fractal processes within stride interval time series data. To determine the most appropriate region for MFC data three different regions were investigated. The effect of different scaling regions is reported in Table 4.5.1.1.2, which calculates the CV% from four levels of higher order detrending.

Table 4.5.1.1.2 Change in scaling exponent for scaling region when detrending over 1st to 4th order polynomials. Definition of terms: CV% is the coefficient of variation; s represents the sample size.

	Window region size		
	Default Option:	**10 < n < 50**	**10 < n < 20**
CV%	2.91 (s = 36)	5.86 (s = 20)	6.20 (s = 20)

The CV% values associated with polynomial (1st to 4th order) detrending, demonstrates the default option to be the most robust window size from the three options (Table 4.5.1.1.2). The default scaling region is given by: 2k+2 < n < N / 4, where k represents the order of detrending applied, N is the length of the MFC series, and n is the window size (Goldberger et al., 2000).

MFC data may possess outliers of higher magnitude in comparison to time interval data sets as used by Hausdorff et al., (1996). Therefore, to check whether values greater than ±3 SD (after normalizing) may have an effect on the scaling exponent 'α', the young subjects data served as a pilot investigation where any values exceeding ±3 SD from the mean were filtered out. Table 4.5.1.1.3 shows the results for the change in the scaling exponent 'α' compared to the original value. The Table shows no significant difference (p > .05) between the original and the screened data from Wilcoxen Rank Sum W test. Some differences are apparent within the left limb DFA comparisons, showing in general a reduced value for the screened data. The results for the right limb remain consistent. When screening for values greater than three standard deviations from the mean the change in the scaling exponent should remain consistent.

Table 4.5.1.1.3 Change in scaling exponent after removing outliers > ±3SD from the mean MFC.

Subject	DFA			
Young	**Left Original**	**Left Screened**	**Right Original**	**Left Screened**
Y1	0.68	0.69	0.82	0.82
Y2	0.82	0.81	0.82	0.82
Y3	0.74	0.75	0.82	0.81
Y4	0.84	0.81	0.85	0.85
Y5	0.73	0.72	0.77	0.76
Y6	0.74	0.74	0.70	0.71
Y7	0.81	0.80	0.81	0.81
Y8	0.75	0.75	0.82	0.80
Y9	0.77	0.72	0.80	0.79
Y10	0.77	0.75	0.79	0.80
Group	**0.77 ± 0.05**	**0.75 ± 0.04**	**0.80 ± 0.04**	**0.80 ± 0.04**

4.5.1.2 Subject results for DFA

Table 4.5.1.2.1 reveals the presence of long-range correlations in the MFC event for all subjects. The accuracy of each DFA value is represented by the coefficient of variation (CV) from four levels of detrending. Group means and standard deviations are provided. The average of the four DFA values obtained from 1st to 4th order detrending for the default scaling region represented the reported value.

Table 4.5.1.2.1. DFA values for subjects and the CV% associated with the four DFA values obtained from 1st to 4th order detrending.

Subject	DFA			
Young	**Left**	**% change (SD/x)*100**	**Right**	**% change (SD/x)*100**
Y1	0.68	3.13	0.82	1.07
Y2	0.82	0.80	0.82	3.28
Y3	0.74	0.13	0.82	3.28
Y4	0.84	2.41	0.85	2.79
Y5	0.73	1.58	0.77	6.51
Y6	0.74	1.64	0.70	1.81
Y7	0.81	2.85	0.81	4.80
Y8	0.75	4.47	0.82	4.24
Y9	0.77	2.66	0.80	0.71
Y10	0.77	12.34	0.79	9.02
Group	**0.77 ± 0.05**	**3.20 ± 3.44**	**0.80 ± 0.04**	**3.75 ± 2.54**
Elderly				
E1	0.74	1.57	0.72	9.69
E2	0.73	1.67	0.83	1.47
E3	0.69	2.98	0.66	4.17
E4	0.80	3.36	0.82	0.55
E5	0.69	1.22	0.86	0.44
E6	0.73	1.08	0.82	2.40
E7	0.94	2.21	0.65	1.52
E8	0.76	1.60	0.82	1.47
Group	**0.76 ± 0.08**	**1.96 ± 0.82**	**0.77 ± 0.08**	**2.71 ± 3.05**

To determine whether the long-range correlations are occurring by chance, surrogate data sets were obtained by randomly shuffling the naturally occurring order of the MFC events. The earlier example for describing the derivation of the scaling exponent shows the change in α on a log n to log F(n) graph after random shuffling (Figure 4.5.1.2.1). For the young subjects left and right limb MFC data (n=20) the scaling exponent 'α' is reduced to a mean of 0.495 ± 0.03 after random shuffling, hence demonstrating uncorrelated white noise (α = 0.5). Figure 4.5.1.2.1 illustrates the significant difference between the DFA and the surrogate DFA obtained by Wilcoxen Ranked Sum W test.

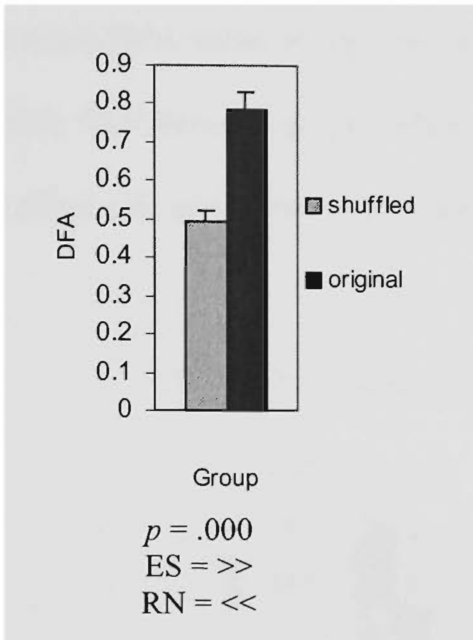

Figure 4.5.1.2.1 DFA of original data compared with the same data series after random shuffling. The results are taken from all young subjects intra- limb MFC time series (n = 20). Symbolisms: >> very large; << very small.

4.5.1.3 Group Differences in DFA

Group summaries of the DFA are presented in Figures 4.5.1.3.1 and 4.5.1.3.2. These graphs represent: (4.5.1.3.1 A) the pooled intra-limb (left and right) data; (4.5.1.3.1 B) the DFA obtained when related to the low MFC limb; (4.5.1.3.2 A) the lowest DFA value recorded within subjects; and (4.5.1.3.2 B) the symmetry index.

From the graphs it is evident that the pooled intra-limb data reveals no significant differences between the two groups (E = 0.766 ± 0.08; Y = 0.783 ± 0.05). To account for individual differences, such that unsteadiness effects may be predisposed towards a certain limb, group differences were compared in three ways.

By comparing the two groups in relation to the low MFC limb (Figure 4.5.1.3.1 B), the elderly group shows a lower mean DFA value in comparison to the young adult group (E = 0.728 ± 0.06; Y = 0.803 ± 0.03). This between group difference is significant (p = .019) that is also supported by a strong effect size and a power of greater than 0.8 (ES = 1.6, required N = 7).

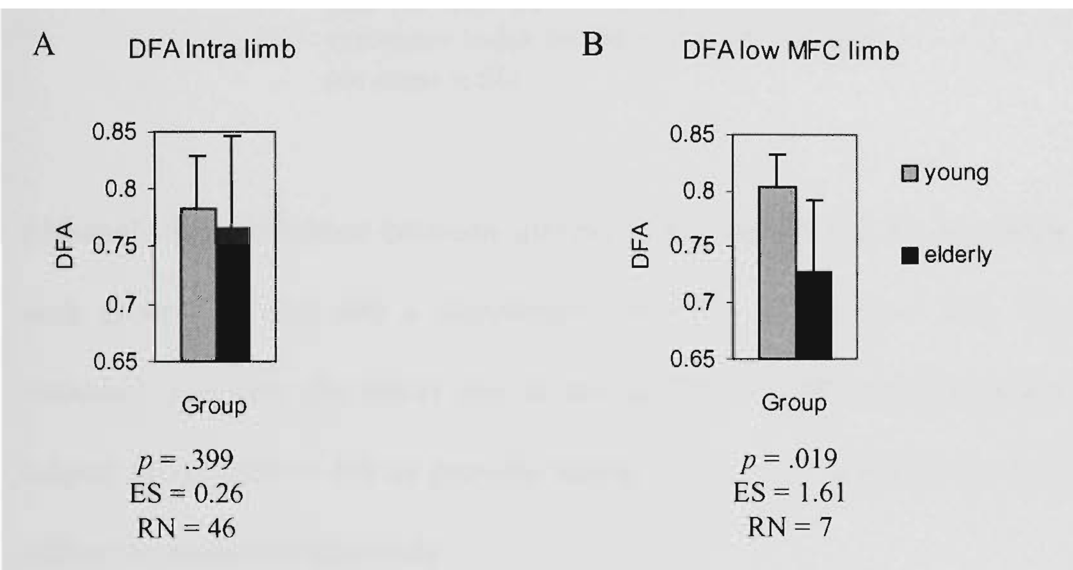

Figure 4.5.1.3.1. Between group differences for the DFA. (A) pooled intra- limb (n = 36); (B) the low MFC limb (n = 18). Values are mean ± SD.

The mean scaling exponent is lower in the elderly (E = 0.718 ± 0.05; Y = 0.761 ± 0.05) when considering the lowest DFA value between limbs (see Figure 4.5.1.3.2). Although this difference is not significant (p = .074) it does show a strong effect size (ES = 0.85). Figure 4.5.1.3.2 B shows that the mean SI value for the elderly group more than twice the value of the young group (E = 12.51 ± 11.65; Y = 5.63 ± 5.76). The subject results for the symmetry index are plotted in Figure 4.5.1.3.3. The plot demonstrates the greater spread between left and right DFA values within the elderly group.

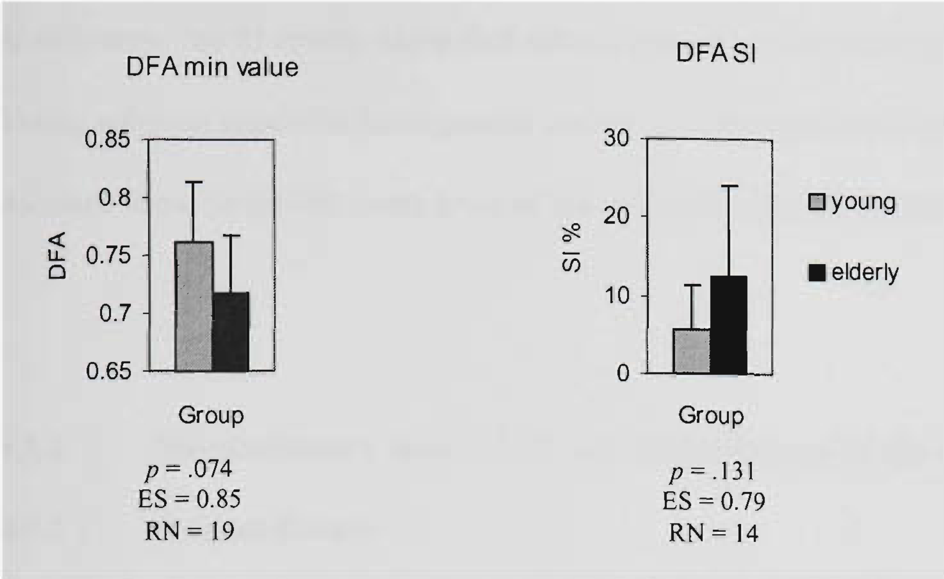

Figure 4.5.1.3.2. Group averages of the DFA for the lowest between limb DFA value (n = 18); and the symmetry index for DFA (n = 18). Values are mean ± SD.

Although this difference between groups is not significant, by removing one subject from each group (Y1 and E4) a significant difference exists ($p < .05$). Nevertheless, without excluding subjects, the effect size is strong (ES = 0.79) and the required sample size per subject group (RN = 14) to provide strong power (at a significance level of .05) is almost within the realms of this study.

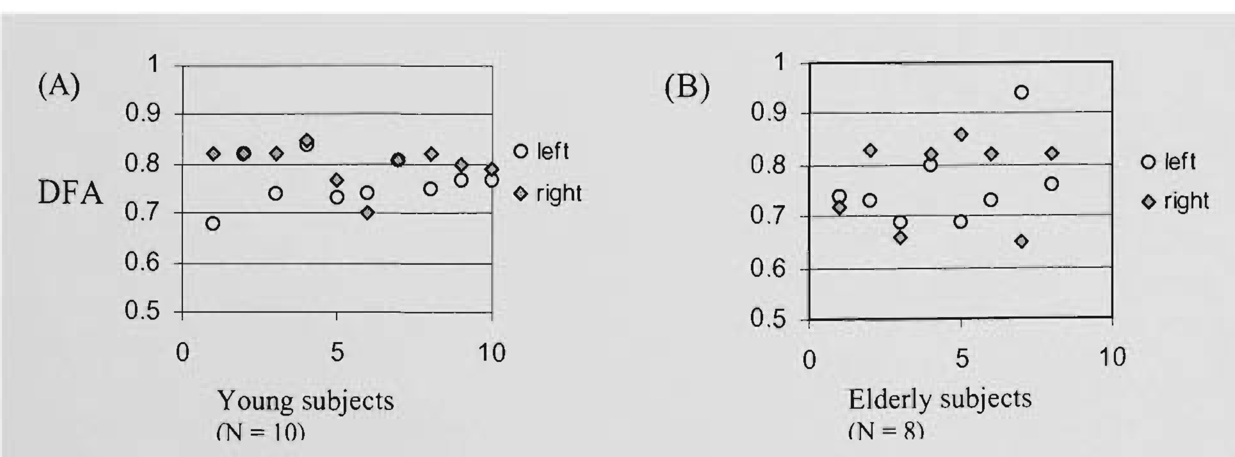

Figure 4.5.1.3.3. Limb inequalities in DFA values for young and elderly. (A) DFA v young subjects; (B) DFA v elderly subjects.

In summary, the SI results show that elderly persons are asymmetric in the DFA of the MFC. Young subjects appear to have greater control over the low MFC limb. In contrast, the elderly does not demonstrate the same level of control of the low MFC limb.

4.5.2 Non-stationary index (NSI) and inconsistency of the variance (IV)

4.5.2.1 Subject Results

Subject results for the NSI and IV for all subjects are listed in Table 4.5.2.1.1. The results show individual differences within both groups for both intra- and inter-limb processes.

Table 4.5.2.1.1. NSI and IV subject results for intra- and inter-limb data.

Subject	NSI			IV		
Young	**Left**	**Right**	**Inter**	**Left**	**Right**	**Inter**
Y1	0.64	0.70	0.50	0.34	0.32	0.40
Y2	0.68	0.67	0.64	0.35	0.38	0.36
Y3	0.61	0.64	0.53	0.36	0.34	0.38
Y4	0.71	0.73	0.66	0.31	0.32	0.33
Y5	0.60	0.77	0.63	0.37	0.27	0.34
Y6	0.74	0.74	0.71	0.34	0.29	0.32
Y7	0.70	0.69	0.64	0.33	0.33	0.36
Y8	0.63	0.68	0.56	0.33	0.32	0.34
Y9	0.67	0.66	0.66	0.35	0.41	0.37
Y10	0.84	0.74	0.71	0.23	0.30	0.30
Group	**0.68 ± 0.07**	**0.70 ± 0.04**	**0.62 ± 0.07**	**0.33 ± 0.04**	**0.33 ± 0.04**	**0.35 ± 0.03**
Elderly						
E1	0.69	0.73	0.67	0.36	0.29	0.34
E2	0.64	0.69	0.62	0.38	0.33	0.34
E3	0.60	0.53	0.52	0.38	0.41	0.41
E4	0.64	0.71	0.56	0.29	0.29	0.37
E5	0.62	0.69	0.61	0.33	0.32	0.35
E6	0.62	0.75	0.61	0.35	0.30	0.35
E7	0.75	0.59	0.60	0.28	0.34	0.34
E8	0.69	0.68	0.63	0.33	0.35	0.34
Group	**0.66 ± 0.05**	**0.67 ± 0.07**	**0.60 ± 0.05**	**0.34 ± 0.04**	**0.33 ± 0.04**	**0.36 ± 0.02**

4.5.2.2 Between group differences

As in the previous section, between group differences are explored using different approaches and are displayed in Figures 4.5.2.2.1 and 4.5.2.2.2 for the NSI and the IV. There is a strong effect (ES = 1.13) and a significant difference (p = .045) between the elderly and young groups for the NSI when considering the low MFC limb (E = 0.64 ± 0.07; Y = 0.7 ± 0.04). In contrast, the IV demonstrates a weak effect size (ES = 0.26) and no significant difference between the young and elderly groups for the low MFC limb, although the elderly group IV mean is slightly higher (E = 0.343 ± 0.04; Y = 0.332 ± 0.04).

Although no significance exists, comparing between grouped intra (p = 0.293) and inter-limb (p = .248) means for the NSI, the young group (Y = 0.69 ± 0.06; 0.62 ± 0.07) is slightly higher than the elderly group (E = 0.66 ± 0.06; 0.60 ± 0.09).

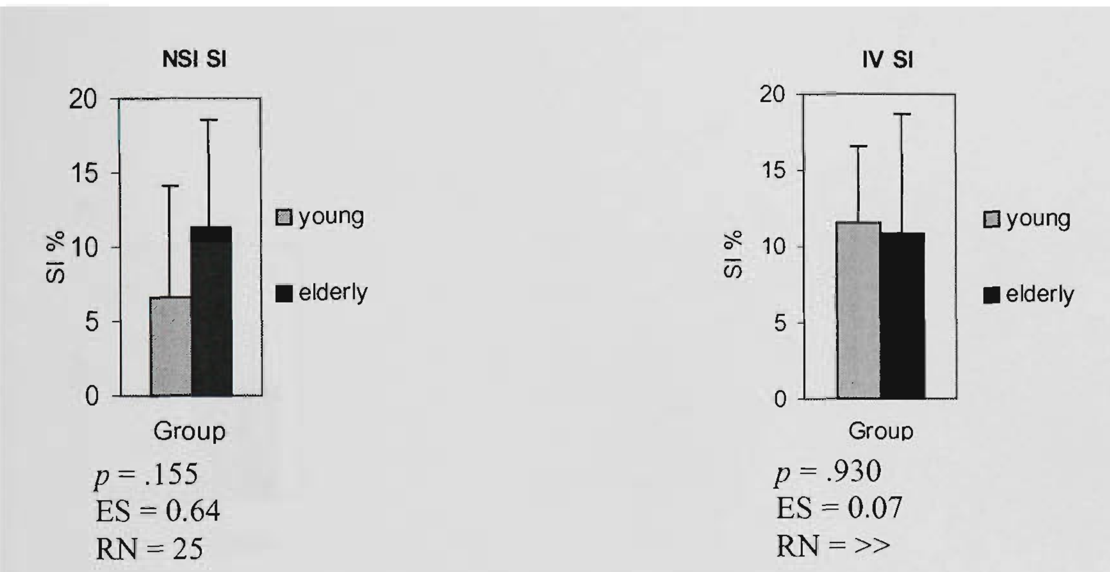

Figure 4.5.2.2.1 Between group differences for the NSI and IV symmetry index (SI). p values, effect sizes (ES) and required sample size (RN) for a power of 0.8 are listed below the graphs.

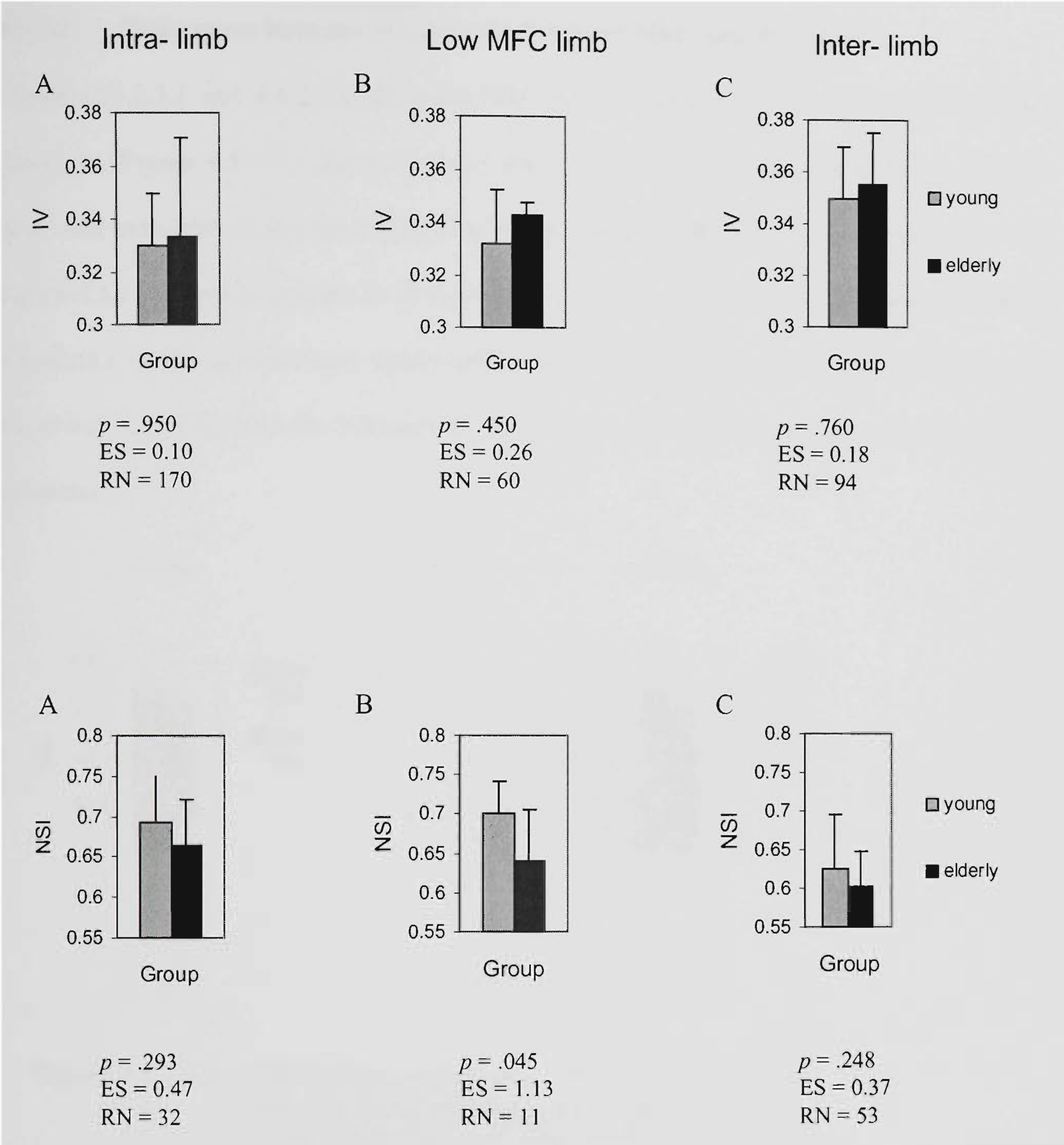

Figure 4.5.2.2.2. Group differences for NSI and IV. Graph 'A' relates to the pooled left and right data. 'B' compares group differences for the low MFC limb. 'C' is related to inter limb MFC interactions. p values, effect sizes (ES) and required sample size for a power level of 0.8 are listed below the graphs.

4.5.2.3 Differences between intra- and inter-limb MFC coordination

Figures 4.5.2.3.1 and 4.5.2.3.2 show the NSI and IV respectively for intra- and inter-limb processes. Figure 4.5.2.3.1 shows that the NSI parameter is significantly different between inter- and intra-limb processes within both subject groups (Elderly, $p = .02$; Young, $p = .02$). Figure 4.5.2.3.2 reveals that the IV is not significant between limb processes however there is a moderate to strong effect size within both groups. There are slightly higher effect sizes for the elderly group in both the NSI and the IV when comparing between intra- and inter-limb processes.

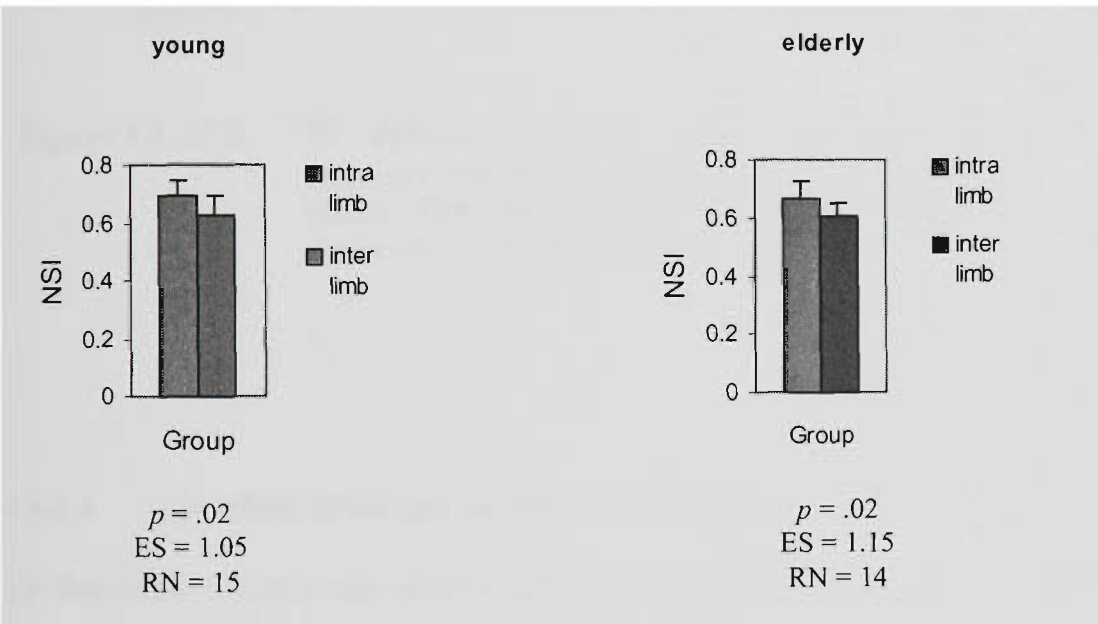

Figure 4.5.2.3.1. NSI differences between intra- and inter-limb processes within the young and elderly groups. *p* values, effect sizes (ES) and required sample size for a power of 0.8 are listed below the graphs.

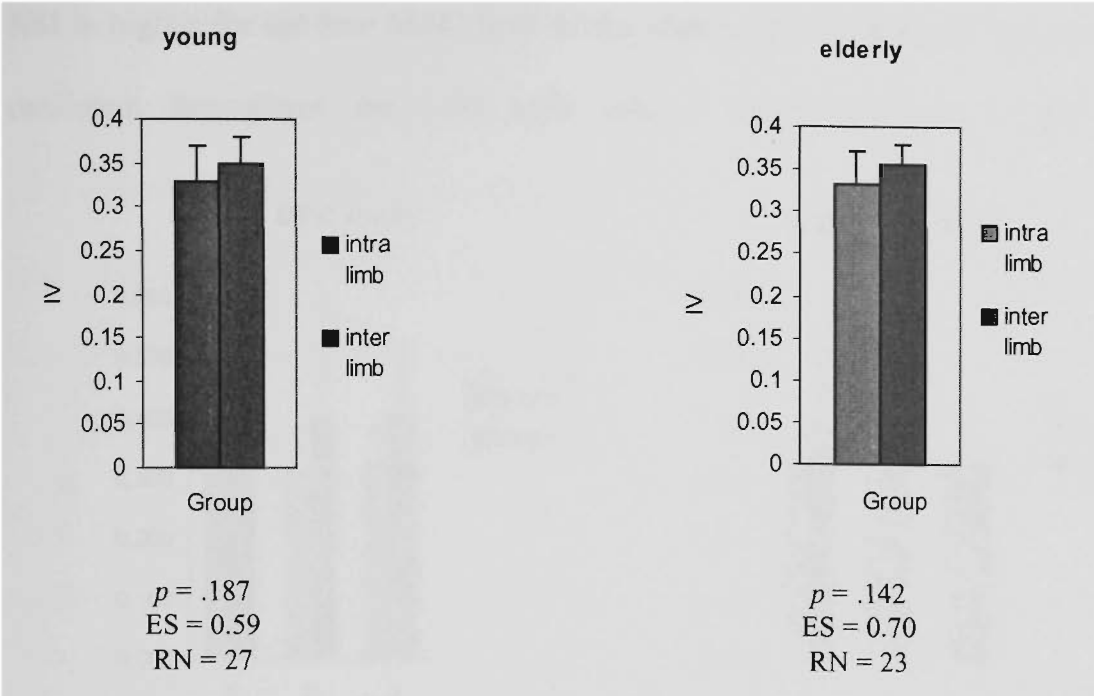

Figure 4.5.2.3.2. IV differences between intra- and inter- limb processes within the young and elderly groups. *p* values, effect sizes (ES) and required sample size for a power of 0.8 are listed below the graphs.

4.5.2.4 The effect of fatigue on MFC unsteadiness

To determine whether the elderly group became unsteady in the low MFC limb due to fatigue, the NSI and IV was calculated for the three equal sections of the trial. Figure 4.5.2.4.1 compares the two groups for changes in unsteadiness for the IV and 4.5.2.4.2 similarly for the NSI. There are between group differences indicated by the effect sizes throughout the duration of the trial.

Figure 4.5.2.4.1 illustrates that the young group's IV remains consistent for both the high and low MFC limbs. In comparison the elderly group demonstrates an increase in the IV for the low MFC limb while the high MFC limb reduces. For the elderly group, differences between the second and third segments of the trial are consistent for the IV. In Figure 4.5.2.4.2 the

NSI is higher for the low MFC limb in the elderly group, however the group NSI remains consistent throughout the trial with only a minor increase in the third segment.

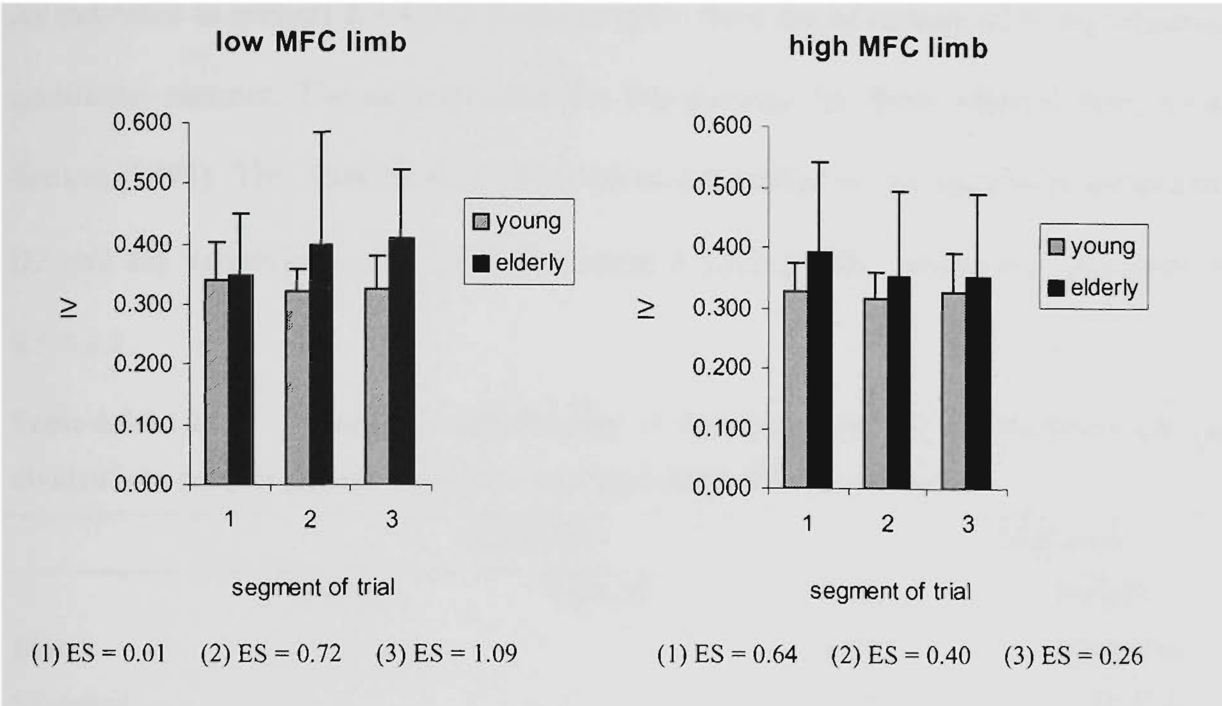

Figure 4.5.2.4.1. The effect of fatigue on the (A) low and (B) high MFC limb for the IV. The horizontal axis represents the stage of the trial. The between group effect sizes (ES) for each of the 3 phases of the trial are listed below the graphs.

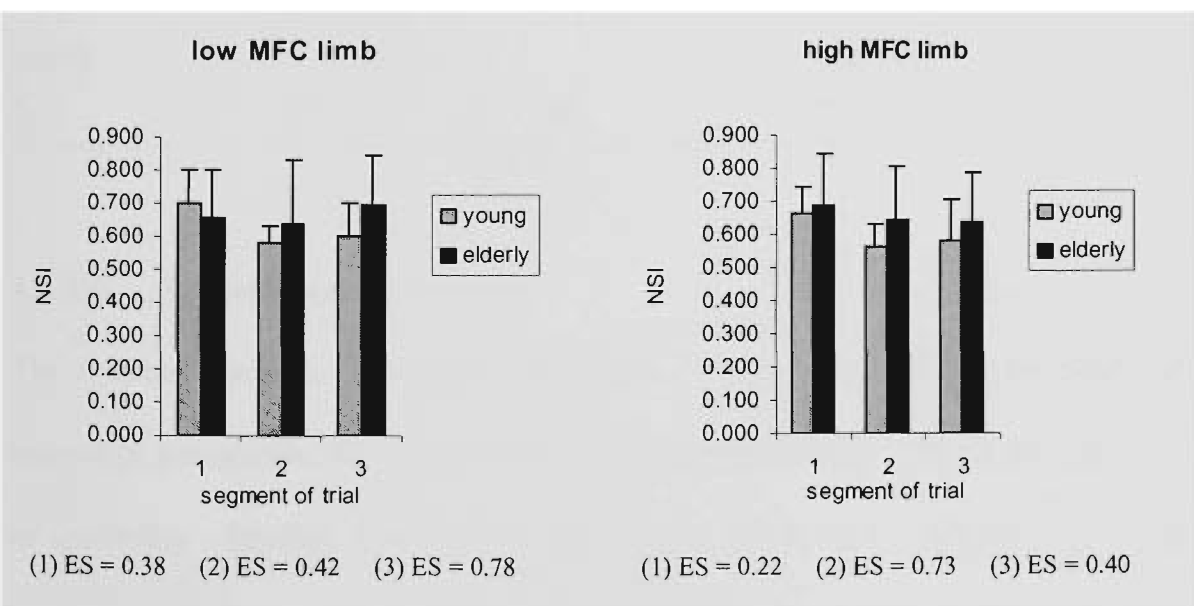

Figure 4.5.2.4.2. The effect of fatigue on the (A) low and (B) high MFC limb for the NSI. The horizontal axis represents the stage of the trial. The between group effect sizes (ES) for each of the 3 phases of the trial are listed below the graphs.

4.5.3 Poincaré plot parameters

4.5.3.1 Poincaré plot nomenclature

As indicated in section 2.7.4, the Poincaré plots have the advantage of being interpreted in a qualitative manner. The nomenclature for this purpose has been adapted from Kamen and Tonkin (1995). The classification of the plots are related to the variability measures of D1, D2 and the variability ratio, 'μ' (see section 3.5.2.2.2). The results are presented in Table 4.5.3.1.1.

Table 4.5.3.1.1. Frequency distribution of the nomenclature for Poincaré plot patterns divided into subject groups and intra- and inter-limb MFC processes.

	Intra-limb		Inter-limb	
	Young	**Elderly**	**Young**	**Elderly**
Blob			4 (40%)	5 (62.5%)
Squashed			5 (50%)	3 (37.5%)
Tight	3 (15%)	3 (22%)		
Normal	5 (25%)	7 (44%)		
Fat	7 (35%)	3 (22%)		
Cigar	4 (20%)	3 (22%)		
Comet	1 (5%)		1 (10%)	
N =	20	16	10	8

4.5.3.2 Subject Poincaré plots

The Poincaré plots for all subjects (Figures 4.5.3.2.1A - 4.5.3.2.1E) are adjoined with their respective parameters: D1 (short- term variability), D2 (long- term variability) and 'μ' (ratio of variability - D1/D2). Some clear differences between intra- and inter-limb processes are apparent from these plots. Likewise, within subject differences were seen between left and right limbs. Generally, larger plots indicate higher variability. The exception being inter-limb plots, as they are formed by normalised MFC values (z scores) to remove between limb effects caused by differences in mean and variance.

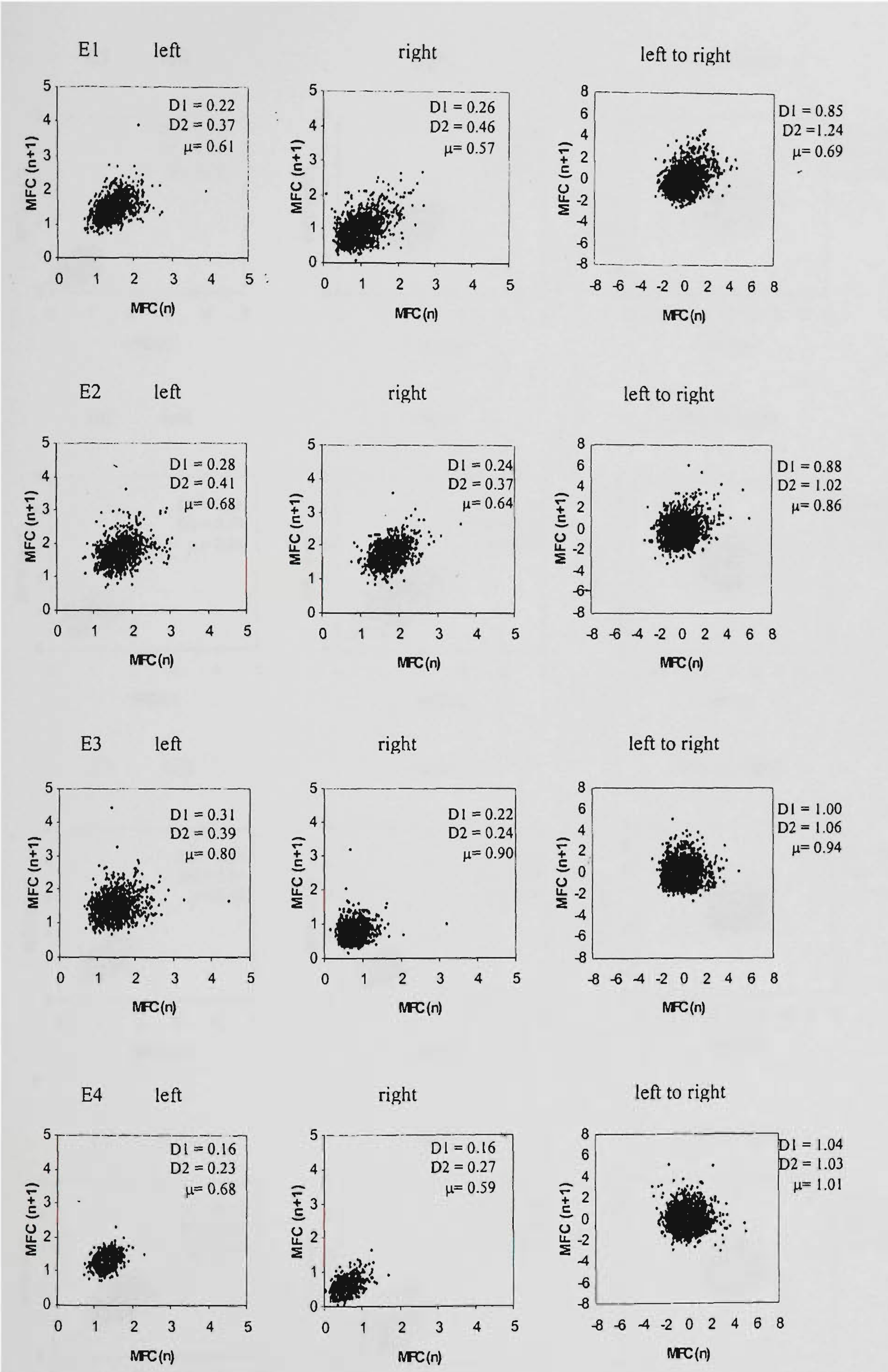

Figure 4.5.3.2.1 (A) Poincaré plots – elderly subjects E1 to E4. See section 4.5.3 for definitions.

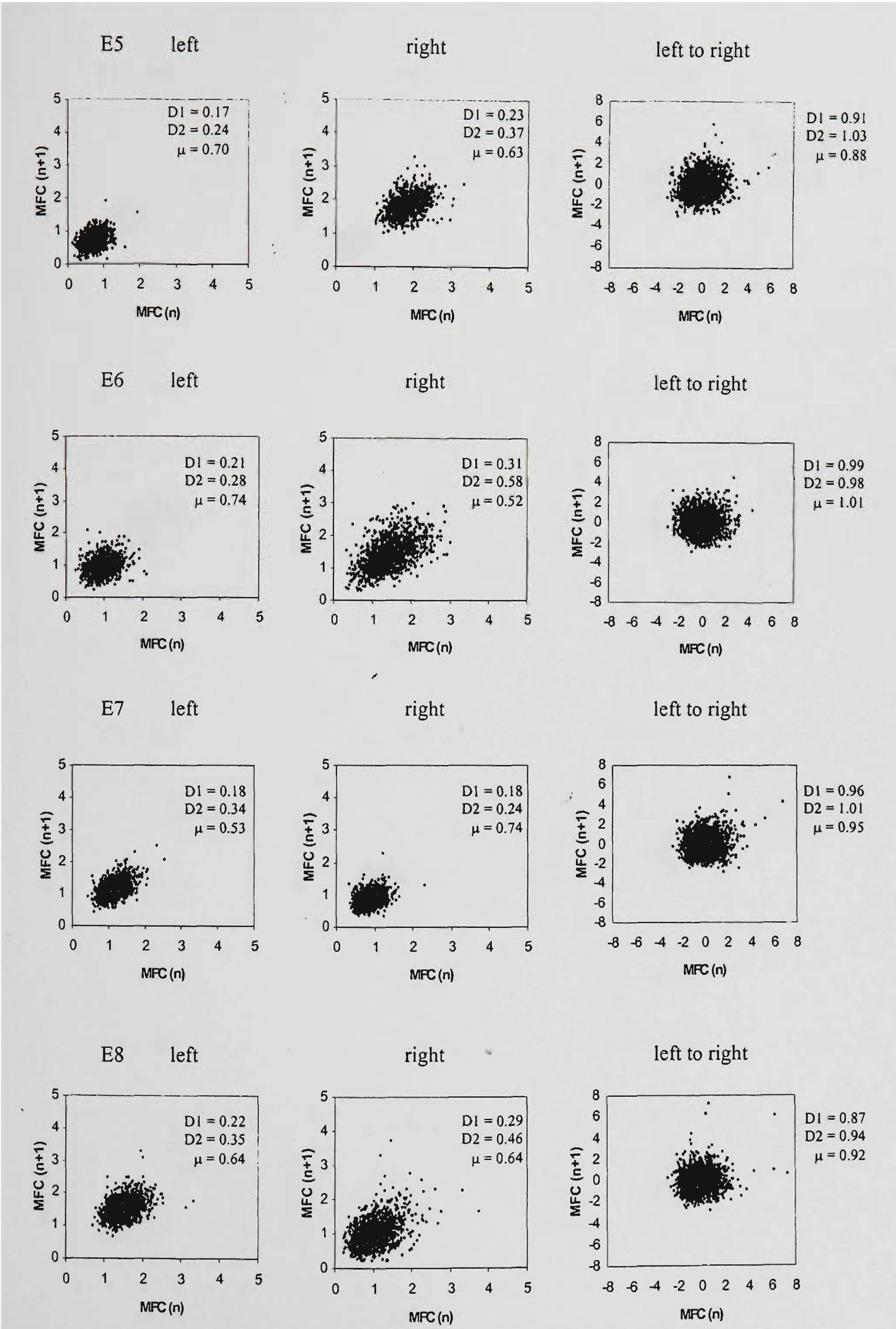

Figure 4.5.3.2.1 (B) Poincaré plots – elderly subjects E5 to E8. See section 4.5.3 for definitions.

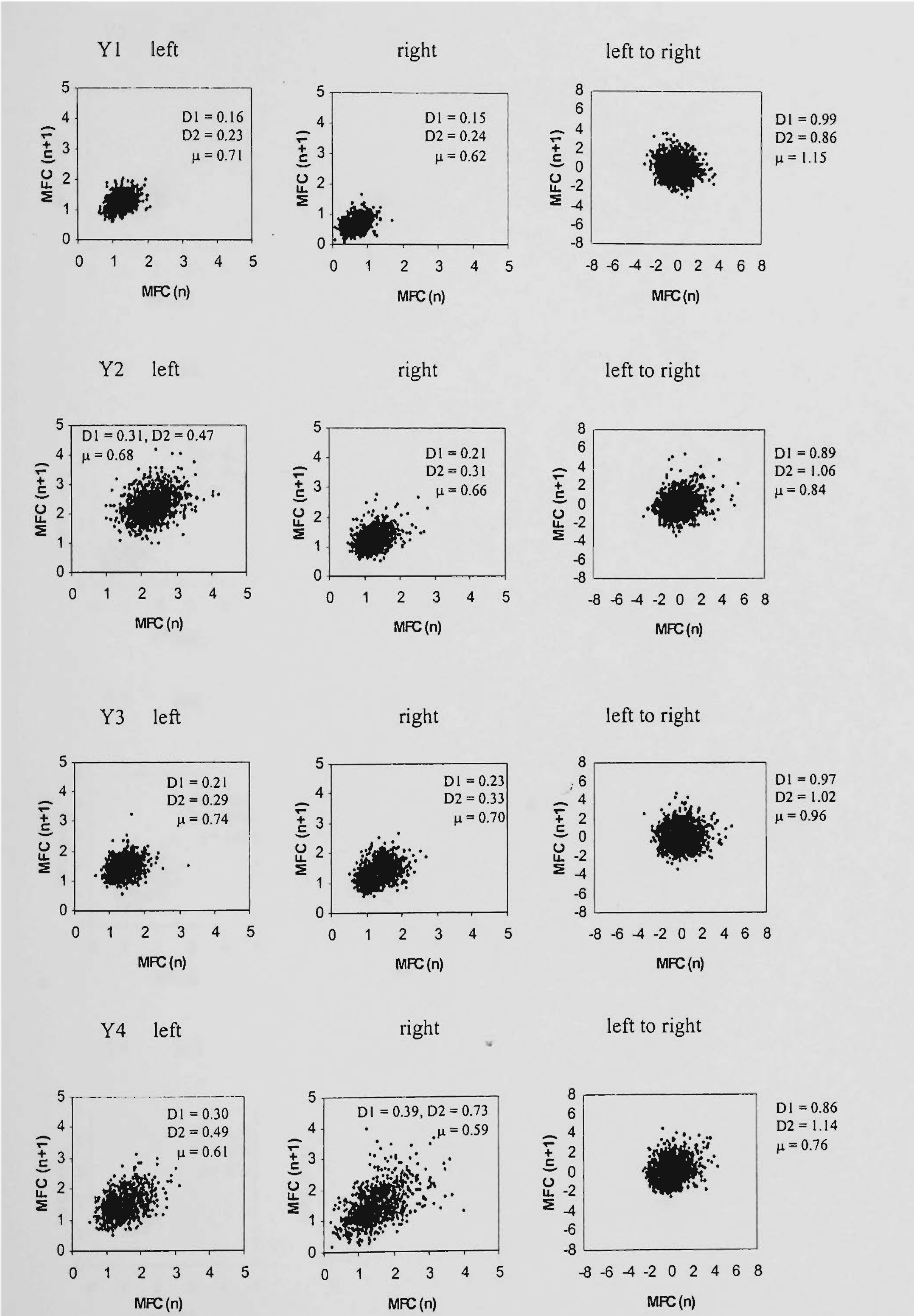

Figure 4.5.3.2.1 (C) Poincaré plots – young subjects Y1 to Y4. See section 4.5.3 for definitions.

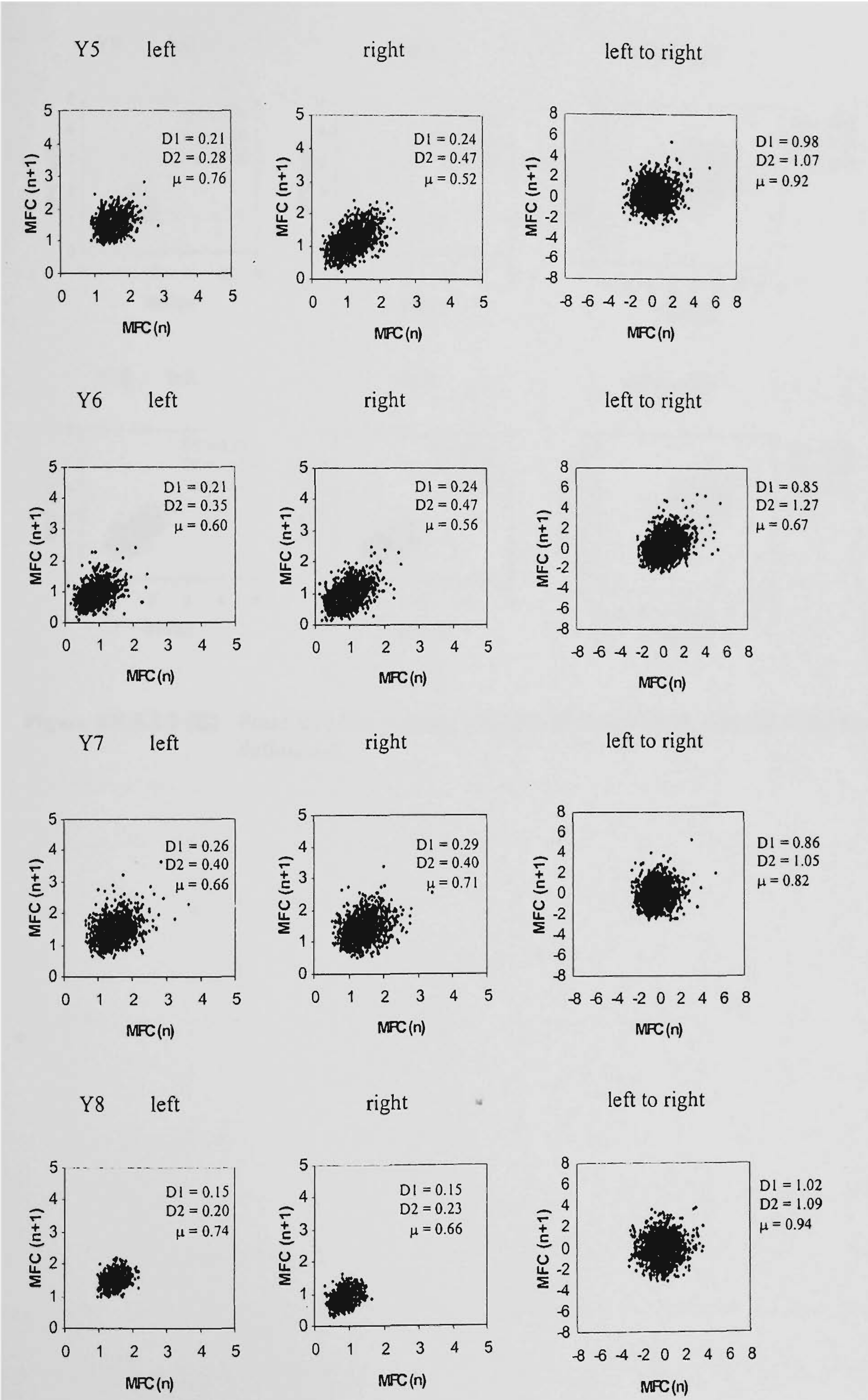

Figure 4.5.3.2.1 (D) Poincaré plots – young subjects Y5 to Y8. See section 4.5.3 for definitions.

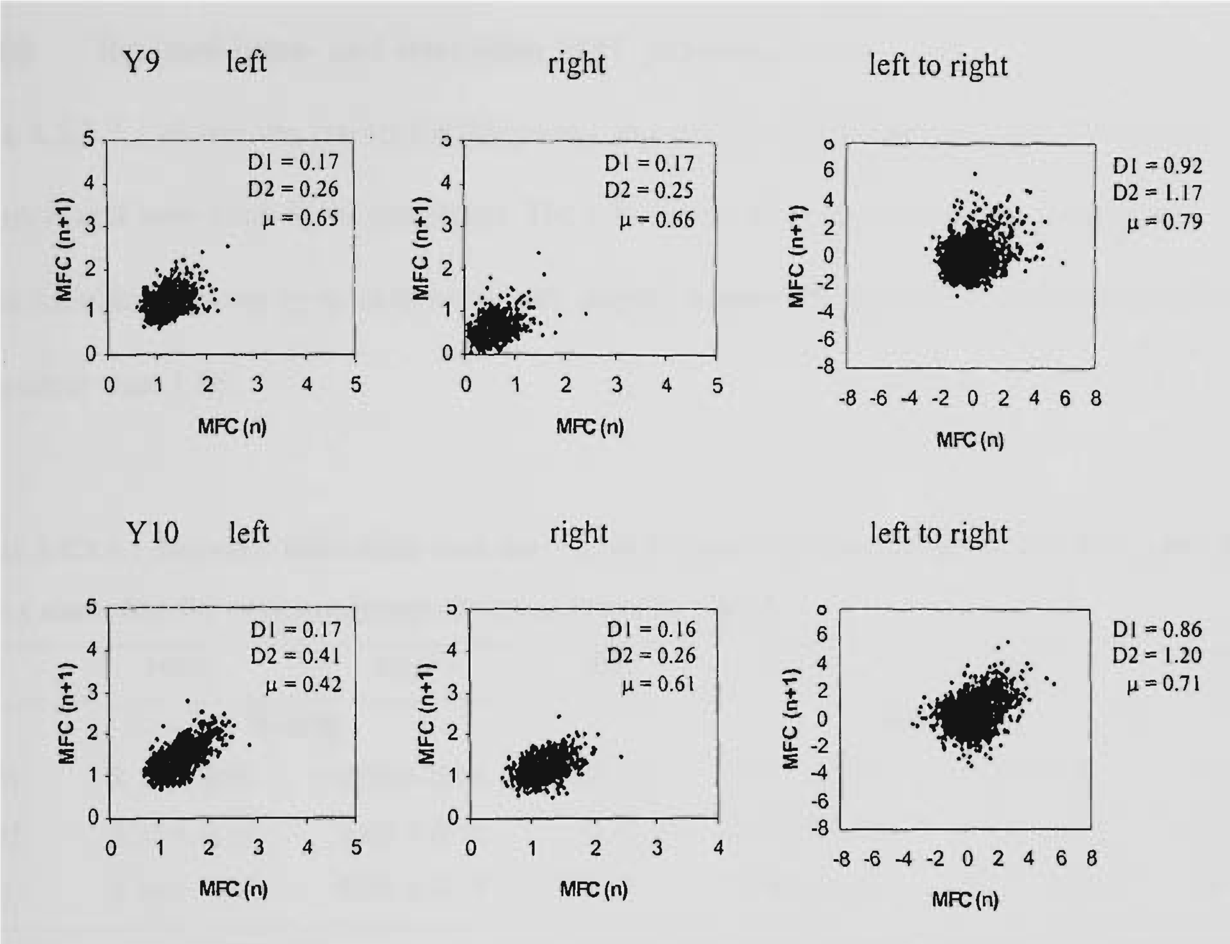

Figure 4.5.3.2.1 (E) Poincaré plots – young subjects Y9 and Y10. See section 4.5.3 for definitions.

4.5.3.3 Between intra- and inter-limb MFC processes

Table 4.5.3.3.1 shows the results for the young and elderly subject groups when comparing the intra- and inter-limb MFC processes. The effect sizes between groups are comparable, while the elderly group appears to have only slightly higher effect sizes. The effect sizes for μ are greater than 1.93.

Table 4.5.3.3.1 Between intra-limb and inter-limb Poincaré plot parameters: D1; D2; and 'μ'. Effect sizes (ES) for between group differences are provided.

	Intra-	Inter-	ES	Intra-	Inter-	ES
	Young			**Elderly**		
D1	0.22 ± 0.07	0.92 ± 0.06	10.81	0.23 ± 0.05	0.94 ± 0.07	11.76
D2	0.35 ± 0.13	1.09 ± 0.11	6.20	0.35 ± 0.10	1.04 ± 0.09	7.44
μ	0.64 ± 0.08	0.86 ± 0.14	1.93	0.66 ± 0.10	0.91 ± 0.10	2.44

4.5.3.4 Between group differences for Poincaré plots

The non-parametric Wilcoxen Rank Sum W test did not reveal any significant differences for D1, D2 and 'μ' between the groups for either intra- or inter-limb processes. Table 4.5.3.4.1 and Table 4.5.3.4.2 report those comparisons using between group effect sizes. A strong effect size characterised the mean difference for 'μ' between young and old for the low MFC limb (ES = 0.72). The key feature of the results is the interaction of D1 and D2, as they appear to be codependent evident by the variability ratio for both intra- and inter-limb MFC interactions. The elderly have a lower D2, higher D1 and 'μ' for inter-limb coordination (Table 4.5.3.4.2). A moderate effect size (ES = 0.53) characterizes the between group difference for D2. The elderly group has higher symmetry index (SI) values, showing that they have greater between limb inequalities for D1 and D2 in comparison to the young group

(Table 4.5.3.4.2). The between group differences for the SI are characterized by moderate effect sizes for both D1 and D2 (Table 4.5.3.4.2).

Table 4.5.3.4.1 Between group differences for intra-limb Poincaré plot parameters: D1; D2; and 'μ'. Effect sizes (ES) for between group differences are provided.

	Young	Elderly	ES	Young	Elderly	ES
	Left MFC			**Right MFC**		
D1	0.22 ± 0.06	0.22 ± 0.05	0.07	0.22 ± 0.07	0.24 ± 0.05	0.21
D2	0.34 ± 0.10	0.33 ± 0.07	0.14	0.37 ± 0.15	0.37 ± 0.12	0.06
μ	0.65 ± 0.10	0.67 ± 0.08	0.17	0.62 ± 0.07	0.65 ± 0.12	0.31
	High MFC			**Low MFC**		
D1	0.23 ± 0.08	0.23 ± 0.05	0.10	0.21 ± 0.05	0.22 ± 0.05	0.04
D2	0.37 ± 0.16	0.38 ± 0.10	0.04	0.33 ± 0.09	0.33 ± 0.10	0.19
μ	0.64 ± 0.11	0.63 ± 0.09	0.13	0.64 ± 0.06	0.69 ± 0.10	0.72

Table 4.5.3.4.2 Between group differences for inter-limb Poincaré plot parameters: D1; D2; and 'μ'. Effect sizes (ES) for between group differences are provided.

	Young	Elderly	ES	Young	Elderly	ES
	Symmetry index (SI)			**Inter-limb MFC**		
D1	12.37 ± 11.85	20.24 ± 14.75	0.59	0.92 ± 0.07	0.94 ± 0.07	0.24
D2	23.14 ± 19.05	33.70 ± 19.36	0.55	1.09 ± 0.11	1.04 ± 0.09	0.53
μ	13.31 ± 13.38	14.51 ± 12.36	0.09	0.86 ± 0.14	0.91 ± 0.10	0.43

Figure 4.5.3.4.1 highlights the between group trends when comparing the low and high MFC limb Poincaré parameters. The key feature from the graph is the difference within the elderly group between the low MFC limb and the high MFC limb for 'μ'. This difference is not reflected by the symmetry index from Table 4.5.3.4.2, indicating between limb co-dependencies. Figure 4.5.3.4.1 also shows that the group mean D1 and D2 are similar

between the low- and high- MFC limbs for both subject groups. Higher D1 and D2 SI values in the elderly group are therefore not confined to either the high- or low- MFC limb.

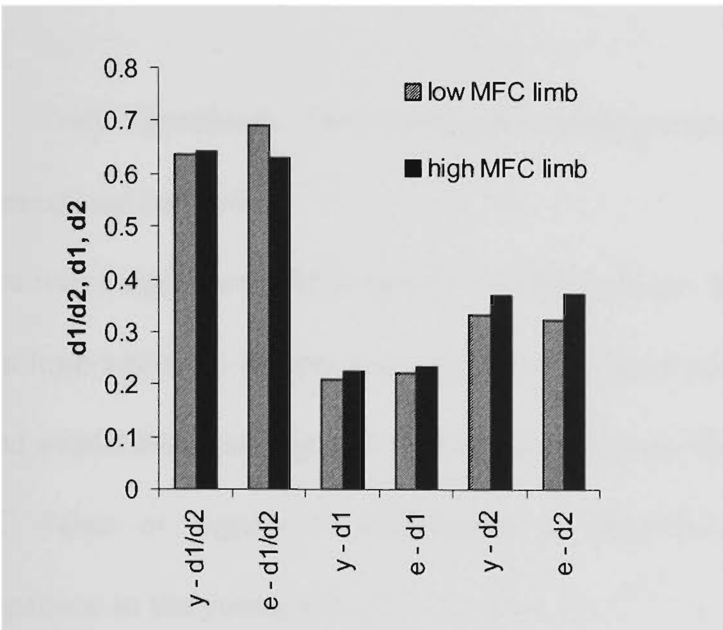

Figure 4.5.3.4.1. The between group differences when comparing the low and high MFC limbs for the Poincaré plot parameters. Symbolisms: y – young group; e – elderly group; d1/d2 – is the relativity ratio, 'μ'.

4.6 SUMMARY OF THE RESULTS

The project sought to test ten hypotheses outlined in section 1.4 using non-parametric methods (see section 3.7). Significance was accepted at the .05 level for both Spearman rank order correlations and the independent groups Wilcoxen Rank Sum W tests. Between group differences were explored among various limb classifications. The choice of the low MFC limb was of interest because it is hypothesized as being related to tripping while walking. The results are summarized below in relation to the hypotheses tested.

1 *Null Hypothesis: MFC distributions are not normal.*

MFC histograms confirmed that MFC data is a non-normal distribution by finding 86% of the analysed data to have a significant non-normal distribution using the Lilliefores statistic. The z values for skewness and kurtosis support this result.

2 *Null Hypothesis: The elderly and young groups have a similar mean MFC, SD, CV, skewness and kurtosis.*

There was a significant difference ($p = .000$) between the low MFC mean limb and the higher mean limb when the elderly and young groups were pooled. A significant difference was also found within the groups ($p < .05$). The elderly group did not reveal a significantly lower mean MFC value or significant differences in skewness, kurtosis or standard deviation in comparison to the young group.

There was no support for the hypothesis that low MFC mean values would be associated with high skewness and kurtosis or with decreased standard deviation. Within subject comparisons between low and high MFC limbs did not confirm that skewness and kurtosis are significantly associated with the low MFC limb.

Between group comparisons did not show significant differences in the histogram descriptive parameters associated with the low MFC limb. Skewness and kurtosis demonstrated a significant correlation within both the low MFC and high MFC limbs, with correlation higher for the low MFC limb.

3 *Null Hypothesis: There will be random correlations measured by DFA within both limbs MFC stride-to-stride events for both young and elderly groups.*

Long range correlations are represented by scaling exponents within the region of 0.75 measured by the gradient on the log F(n) verses log n plots. The mean DFA found for both groups is within this range for both young and elderly groups. The affirmation that these values indicate long-range correlations is evident by the significant difference (p = .000) of the change in 'α' after random shuffling the MFC time series order using the young subjects data. Long-range correlations are not due to outliers greater than three standard deviations from the mean, and are not due to non-stationary drifts in the data.

4 *Null Hypothesis: Correlations within the intra-limb MFC stride-to-stride events for the elderly will be comparative to the young group's DFA value.*

The DFA did not show a significant reduction (p > .05) in the elderly group when both limbs were grouped together. However, a significant difference (p = .019) existed between the two groups when considering the DFA associated with the low MFC limb. The absolute minimum DFA value also showed that the elderly were lower than the young group with a strong effect size (ES = 0.85), although significance (p = .074) was not found.

5 *Null hypothesis: There will be no difference between measures of unsteadiness for inter-and intra-limb MFC events between groups.*

There were no significant differences between groups in inter-limb relations of MFC unsteadiness. The non-stationary index and the inconsistency of the variation were similar between groups. The Poincaré parameters show trends towards increased inter-limb unsteadiness in the elderly group compared to the young group, however the effect size was only moderate for D2 and 'μ' and no significance was found.

When grouping the left and right limbs, no differences were evident in MFC unsteadiness between the groups. However, in similar context with the DFA results, the NSI revealed that the low MFC limb is significantly (p = .045) more unsteady in the elderly group. The variability ratio also supports these findings with a moderate to strong effect size. From a power analysis equation, a sample size of 16 within each group suggests that group differences for the low MFC mean limb can be supported with a power of 0.8 at a significance level of .05 for the unsteadiness parameters NSI, α and μ. The increased unsteadiness of the low MFC mean limb compared to the high MFC mean limb for the elderly group is supported by the following parameters: CV; DFA; μ; and IV.

6 *Null Hypothesis: Differences between left and right limbs are higher for the elderly group as measured by the symmetry index over all parameters that describe the nature of intra-limb MFC.*

There were no significant differences (p > .05) between the elderly and the young for measures of asymmetry. The symmetry index was generally found to be higher in the elderly for most parameters, with moderate to strong effect sizes. Specifically, between group effect sizes showing higher asymmetry in the elderly group greater than 0.5 were the mean, SD, DFA, NSI, D1 and D2. The skewness, kurtosis, CV, and μ were all only slightly higher for the elderly group with a weak to moderate effect.

7 *Null hypothesis: There will be no difference between measures of unsteadiness between inter-and intra-limb MFC events within both groups.*

Both groups demonstrated that inter-limb control is not as strong as intra-limb control. A significant difference between limb relations was supported for the NSI, whereas the IV showed only a strong effect between the two processes. Visual demonstration of the difference between the two limb-to-limb processes is clearly apparent in the Poincaré plots. The relativity ratio 'μ', of the Poincaré plot shows large effect sizes to support the within limb control being greater than between limb control.

8 *Null Hypothesis: Treadmill walking speed does not affect MFC unsteadiness parameters for both the young and elderly groups.*

Treadmill speed was not associated with any of the measures describing MFC unsteadiness. Nevertheless, an unexpected significant correlation (r = 0.552, p = .018) between the treadmill speed and the DFA was found after considering the low MFC limb. It is interesting that treadmill speed affected the DFA for the low MFC limb and not the high MFC limb. While the overgound walking speed was not significantly different between the two groups (ES = 0.33), the elderly group had a treadmill speed significantly slower than the younger adult group (p = .000).

9 *Null Hypothesis: MFC unsteadiness parameters are not correlated.*

In general, measures of MFC unsteadiness showed a relationship using the non-parametric Spearman's correlation matrix. The following relationships are significant when analysed using both left and right limb results combined. The SD has a strong positive correlation with D1 and D2, a moderate correlation with NSI. The CV is also a moderate positive correlation with the NSI, and D2, while it has a negative association with μ. The DFA is also negatively correlated with μ, as well as the IV. The DFA is positively related to the NSI. The NSI is negatively related to the IV, μ and D1. The IV is positively related to μ and D1. D1 and D2 share a strong positive relationship.

Chapter 5

DISCUSSION

Many studies of gait assume symmetry between the limbs and no studies have investigated MFC symmetry. This dissertation has built upon the current literature by demonstrating that the interaction between the limbs differ using the MFC to define the output function of gait. Two subject groups of young and elderly healthy adults completed one walking trial each on a motorized treadmill at the biomechanics lab at Victoria University. Minimum foot clearance data was extrapolated from bilateral spatial coordinates of the foot trajectory using 2-d video analysis procedures. The study utilized quantitative measures to characterise the changing history of the minimum foot clearance event. New methods describing the dynamic patterns produced by MFC series data demonstrate results that will be summarized and discussed in the following sections.

5.1 MFC ERROR AND IMPLICATIONS

5.1.1 Summary of MFC error

A number of contributing factors that may affect the minimum foot clearance value have been explored. There were three primary sources of error: 1) rotation of the foot during the swing phase; 2) foot model digitization; and 3) misrepresentation of the ground reference.

The rationale behind the application of the geometric model was based upon foot rotations during the swing phase being primarily limited to a two dimensional plane. Ideally, the distance between the two markers P1 and P2 should remain constant, producing an invariant d1 value. The results from section 4.1 (Figure 4.1.4.1) show that the change in d1 is apparent throughout the gait cycle, but appears to be changing in a systematic manner. The systematic changes in d1 during the swing phase are supported by Johnson et al. (1996), which is important because it shows that d1 will be relatively consistent at successive MFC events. The effect of this on the final MFC has been explored by investigating the change in θ_2 and it's relationship with d1 at MFC. The changes in d1 related to θ_2 may be due to changes in foot angle motion in the sagittal plane rather than changes in d1. The proportion of error in θ_2 due to changes in d1 at the MFC event is approximately similar to the change in d1 itself due to rotation, which is 2.5%. The effect of rotation on the variability of θ_2 (SD = ± 3.01 degrees) was approximately ± 0.1 degrees and, the error in MFC attributed to variations in θ_2 values due to foot rotation was negligible.

Foot model digitization accuracy involves systematic error affecting the mean MFC values. Systematic errors caused by foot model digitization and ground reference approximation can affect the mean MFC values. To analyse the variability that may be associated with human error when digitizing the shoe out-sole, a number of neighbouring points (see section 4.1.5) on the shoe out-sole were digitised. A scatter plot revealed the shoe out-sole contour, such that any irregularities are exposed and thus re-examined (Figure 4.1.5.2). Provided an accurate digitized foot model is used, all points should ideally converge upon the same vertical reference at toe off. This has been validated by Figure 4.1.5.1, which demonstrated that the ground reference is accurately predicted from the geometric model.

The analysis of these potential forms of error associated with the MFC allows a confident interpretation of the MFC data. But, further work will be needed to apply three dimensional analysis techniques and various models to describe the ground reference point and the shoe out-sole for making concrete assessments of the limitations to the geometric model. This type of approach will build upon research by Startzell and Cavanagh, (1999).

5.2 SUBJECT CHARACTERISTICS

The decreased range of motion for ankle joint mobility in the elderly group compared to the young group is supported in the literature. The MANOVA result revealed that the average overground walking speed is greater in the young ($p = .101$). Interestingly, the elderly needed to lower the treadmill speed from their overground speed, which created a significant difference between the two groups for treadmill walking speed ($p = .000$). This suggests that the elderly may not have been as comfortable walking on the treadmill in comparison to the young group and may have adjust their walking pattern.

The up and go test and the fear of falling survey demonstrated that the elderly group maintains a high degree of functional mobility and no fear of falling when performing daily activities. Within the group, the lowest rating in the efficacy survey was 130 from a possible score of 140. A possible limitation of the survey is related to the composition of the final score. For example, some subjects rated all items as a ten apart from two, which they nominated as a five. Alternatively, there were subjects that rated most items as an 8, 9 or 10 and hence returned a similar total score. These subjects could be deemed cautious rather than fearful, yet they produce a similar total score to those subjects who display a strong feeling of inadequacy when performing certain tasks.

The highest time for the up and go test was found to be 10.76 seconds. This time does represent some mobility impairment based upon the literature (Shumwell-Cook et al., 2000).

However, the subject did not report any physical impairments or dysfunctions that may have excluded them from this investigation. This subject also had an overground walking speed higher than the within group mean, and therefore consideration of these factors did not exclude this participant from further testing.

5.3 DYNAMIC PATTERNS OF INTRA- AND INTER-LIMB MFC DATA

5.3.1 Visual interpretation of the MFC patterns

Visual interpretation of the MFC patterns has shown differences between limbs, between individuals, and to a lesser extent, group differences. The quantitative measures of unsteadiness investigated are related to these graphical forms.

DFA shows how taking one magnification of the MFC time series graph can be similar in shape to a different magnification of the same graph. This structuring within the time series pattern becomes clearer when contrasting non-correlated processes with long-range correlations. To further examine non-stationary drifts, visual inspection of the non-stationary index (NSI) demonstrates a common structure that is unlike the time series graphs of stride-to-stride MFC values. The structure within the non-stationary index graphs is more apparent. This is due to slowly varying changes of the local means. In contrast to the NSI, the inconsistency of the variance (IV) illustrates greater consistency (less structure) in the fluctuations among neighbouring MFC values (see section 5.3.5 for an example of the graphical nature in the NSI and IV graphs). Higher consistency is associated with less instability from one stride (or step) to the next.

Histograms and Poincaré plots show how differences in shape exist, both within and between subjects, for both left and right intra-limb processes. The Poincaré plot has an additional feature by 'mapping' the inter-limb coupling features, which showed clear differences between intra- and inter-limb processes. Different information has been revealed from the various plots, for example the histogram illustrates a general outline of the MFC control tendency and strategy, while the Poincaré plot shows how short- term control aspects of walking are related to the more global changes (i.e. D2; long- term) in the motor control system. The Poincaré plot best demonstrates this relationship when comparisons are made between the inter- and intra-limb processes. A circular 'cloud' form characterises MFC as a more random correlation between successive values. These characteristics will be addressed more specifically throughout the following discussion. The purpose here is to highlight the visual aspects of the different graphs and their general patterns.

5.3.2 Characteristics of the MFC mean and variance

Given the various procedures within the literature for calculating MFC are different (e.g. Winter, 1992; Karst et al., 1999) to those used in this dissertation, it is worth noting that the descriptive (mean and SD) MFC results of this dissertation indicate similarities. There was no between group effects for either the mean or standard deviation for grouped left and right MFC data. The young subjects had a mean MFC of 1.26cm and a standard deviation of 0.29cm, whereas the elderly had a mean of 1.23cm, although a similar SD of 0.29cm. This finding suggests a common similarity between the young and elderly without an indication of aging effects. The variability about the intra-group mean for the elderly (SD = 0.36) was similar to the young (SD = 0.33) for the left MFC, however the right MFC was noticeably higher in the elderly group (Y = 0.32; E = 0.48; see Table F2, Appendix F), suggesting a lack of homogeneity within the elderly group. The coefficient of variation is only slightly higher in the elderly (CV = 25.25 %), in comparison to the young adults (CV = 23.79 %). The mean results are comparable to overground walking trials by Winter (1992) and Karst et al. (1999). The standard deviation is comparable to Winter (1992) and Dingwell et al. (1999).

5.3.3 Strategy in MFC indicated by skewness and kurtosis

The MFC data distributions indicate the presence of both positive skewness and kurtosis. The magnitude of both skewness and kurtosis z scores and the Lilliefors statistic indicate that minimum foot clearance data is not normally distributed (see Section 4.3.1). Similar results have been reported elsewhere from treadmill walking (Best et al., 1999), however, no comparisons can be made with over-ground walking. The presence of high skewness and kurtosis suggest that there may be a strategy to avoid ground contact during the swing phase of gait. This appears logical, as high kurtosis would appear to be associated with less variability about the frequently occurring central tendency values. However, the results did not indicate a correlation between SD and kurtosis. Positive skewness for MFC distribution indicates 'carefree' control above the central tendency, while it appears that the motor control system adjusts when proprioceptive reflexes sense a lower placement of the swing limb. The presence of both skewness and kurtosis may suggest a positive sign that the locomotor system is planning to minimise the likelihood of tripping over an unseen obstacle. Further, research into strategies for maintaining low MFC occurrences is in its infancy but Best et al. (2001) have found mean, SD, skewness can be used on their own or in combination to reduce trips.

5.3.4 Long-range correlations

The human gait is a complex process dependent upon the inputs from the motor cortex, basal ganglia, cerebellum, and spinal nerve centers, along with feedback mechanisms from the vestibular, visual and proprioceptive systems (Leonard et al., 1995). It is remarkable that all these aspects involved over an elaborate multiplicity of sensory scales can unfold to present self-similarity structures, a manifestation of power-law relationships and scaling (Meakin, 1998), in the MFC events. West et al. (1998) term this presence of scaling allometric control, a process that underlies regularity that is common in biological systems. In order to maintain regularity, biological systems with power-law scaling are not allowed to settle into a 'fixed state' but they are considered to be verging on transition, and are far from equilibrium (Kelso, 1995). This is a common theme for self-regulating systems such as the rhythmic generation of the MFC event during walking. West et al. (1998) consider the concepts of cybernetics, which implies that a self-regulating system utilizes the information producing the output and feeds it back into the input. These concepts were described in the literature review section 2.4. Therefore, in gait, coordination of the responses to the numerous multi scaled sensory output is dependent upon the ability of each sensory system to obtain quality feedback. The DFA results indicate that the interdependent functioning of efferent and afferent mechanisms show differences between limbs. These differences are greatest in the elderly.

The general presence of long-range correlations in the spatial output of the MFC over hundreds of strides is a new finding in gait research. The findings from this dissertation build upon previous studies in gait analysis that identified the presence of long-range correlations in time intervals using the DFA (Hausdorff et al., 1995). This original finding gained support from a similar technique derived from power-law scaling, the average Hurst exponent (West et al., 1998).

In a similar approach to the two aforementioned studies, the findings from this study were validated by a significant difference ($p < .001$) found when the left and right MFC data series for the young subjects were randomly shuffled, and then compared against their corresponding original data set. This result showed that long-range correlations were not simply due to the distribution of MFC data points.

Further validation methods to confirm the presence of power-law correlations in the MFC followed previous tests from Hausdorff et al. (1996). First, non-stationary drifts may cause false structuring of the MFC time series data from inconsistent drifts in the magnitude of the MFC mean that may be caused by changes in energy levels or gait speed. A certain degree of control over the latter can be due to the speed constraint imposed by the treadmill walking as speed fluctuations are generally minimized in comparison to gait parameter variability of over-ground walking (Dingwell et al., 1999). Changes in gait speed will be marginal and any effects on non-stationary drifts due to these changes can be discounted as an influence on non-stationarity found in the MFC time series.

Regardless of the control in gait speed, adaptations may occur over the duration of the walking trial and cause noticeable drifts. Thus, to eliminate the possibility that the long-range correlations were due to the effects of non-stationary drifting, 'α' was also investigated by detrending the original time series y(k) against a higher order line of best fit $y_n(k)$ within each window size n. A coefficient of variation indicates the degree of accuracy of the DFA over 1st to 4th order detrending (section 4.5.1.1). Ideally, 'α' should remain stable, which validates that the self-similar structure is not due to possible non-stationary effects.

In general, the low CV across higher order detrending validated the presence of self-similarity. Eight from 36 cases displayed CV of greater than 4 percent. The group mean CV for all calculated DFA values over 1st to 4th order detrending was approximately 3 percent. This value was nearly twice as high in contrast to Hausdorff et al. (1996), which was caused by large outliers. Subject Y10 provides a good example of a high CV value that indicated that self-similarity is possibly due to non-stationarities. Upon closer inspection of the time series graph indicates a definite drift (see appendix section E) that will be well characterised by a second order polynomial. Hence, DFA values generated from 2nd and 4th order detrending did reduce CV from 12.34% in the left limb to 3.4% and yielded a lower DFA value from 0.77 to 0.72. Similar results occurred for the right limb after higher order detrending.

An explanation for the second possibility, that long-range correlations are simply a consequence of integrated mechanisms acting over multiple time scales has previously been addressed in Hausdorff et al. (1996). The authors super-imposed surrogate data sets of white- and Brownian- noise, and found that the combined properties did not produce patterns demonstrating long-range correlation structures.

Higher order DFA detrending demonstrates that the scaling exponent remains constant, however, it does not display the same convergence as time interval data investigated by Hausdorff et al. (1995), the MFC remains robust over higher order detrending. Detrended data sets will have little effect on the scaling exponents of genuine self-similar processes. The results proved inconclusive and again for random fluctuations on multiple time scales. This study demonstrated that the order of the data is significantly different to when the order has been randomly shuffled.

Although long-range correlations have been found in stride time intervals of gait, the current finding of spatial (MFC) correlations demonstrates additional support for the presence of long-term memory control during gait. The slightly lower magnitude of fractal scaling in the MFC in comparison to the results commonly found in stride time correlations could relate to two issues. First, constraints imposed by walking by the treadmill is a likely possibility due to previous findings indicating that metronome paced walking is associated with the breakdown of long-range correlations in stride time intervals (Hausdorff et al., 1996). The metronome acts in a similar way to the treadmill in controlling gait speed, however the treadmill allows more freedom to change cadence. Second, larger contribution of stochastic processes involved in spatial control of the foot during the swing phase of gait. Another issue that can be raised is the set treadmill speed may not be the actually preferred speed desired by the participant. To oppose this issue, taking results from stride time interval data, Hausdorff et al. (1996) tested subjects for three different speeds. In general, the results suggest that speed does not alter the DFA. Hence, treadmill speed variations from natural speed can be considered minimal.

A correlation between treadmill and overground walking speeds was not significant, indicating that certain subjects within the group adapted to the treadmill differently than others when replicating their overground walking speed. This is especially important, because the elderly treadmill speed (3.8 ± 1.24 km/h) was marginally lower than their overground speed (4.53 ± 1.48 km/h). This may also be relevant to suggestions that the gait pattern for some people may have become modified from treadmill constraints. To reiterate sections 3.2.6 and 5.2, many elderly subjects found after their familiarisation period, that the overground speed certainly did not feel natural when walking on the treadmill and a reduced speed was more comfortable. All elderly subjects had prior experience with treadmill walking and thus any psychological affects associated with treadmill walking and gait pattern can be considered minor. In any case, long-range correlations were still apparent in slow walking subjects.

Identifying the mechanisms and their roles that are responsible for producing long-range, fractal correlations, is a far greater challenge than quantifying their presence. The evolution of long-range correlations has been described with a mathematical model that replicates a similar process (Hausdorff et al., 1995). By applying predetermined randomly assigned frequencies to different pattern generator modes, and the transition between neighbouring frequency modes can occur with equal probability, generation of long-term correlations is achieved, such that $\alpha = 0.75$. Without suggesting that the model replicates the locomotor control system, it does highlight a rationale existing behind the regulation of multiple frequency scales within the locomotor system that produces long-range, fractal like correlations found in the foot trajectory of gait. These frequencies reside in both afferent feedback system and the descending signals from central output mechanisms. Nevertheless, this model is a somewhat general version of the locomotor system and does not specifically address the underlying mechanisms of gait control.

5.3.5 Support from the non-stationary index and the inconsistency of the variance

From figure 5.3.5.1 the NSI shows how the local MFC values appear to drift over time rather than change in random like fashion supports the presence of 'random walk' (Brownian motion) behaviour found in biological processes (Goldberger et al., 2000).

To explain the changes occurring from one stride to the next the non-stationary index serves as a conceptual link. The mean MFC of the next five steps seems to follow a drifting trend, observed in the wavelike structure of the NSI graph shows how the local MFC values make minor adjustments in relation to a previous local average. The inconsistency of the variance does not appear to drift but rather remains more consistent, such that changes in the local fluctuations are apparently random rather than cyclic. An interesting negative association between the NSI and the IV may imply that with increased variability among the local MFC fluctuations, the ability of the local values to continue following a certain trend is prevented. It may be that the increased variability does not allow the motor program to settle into a memory or long-term flow.

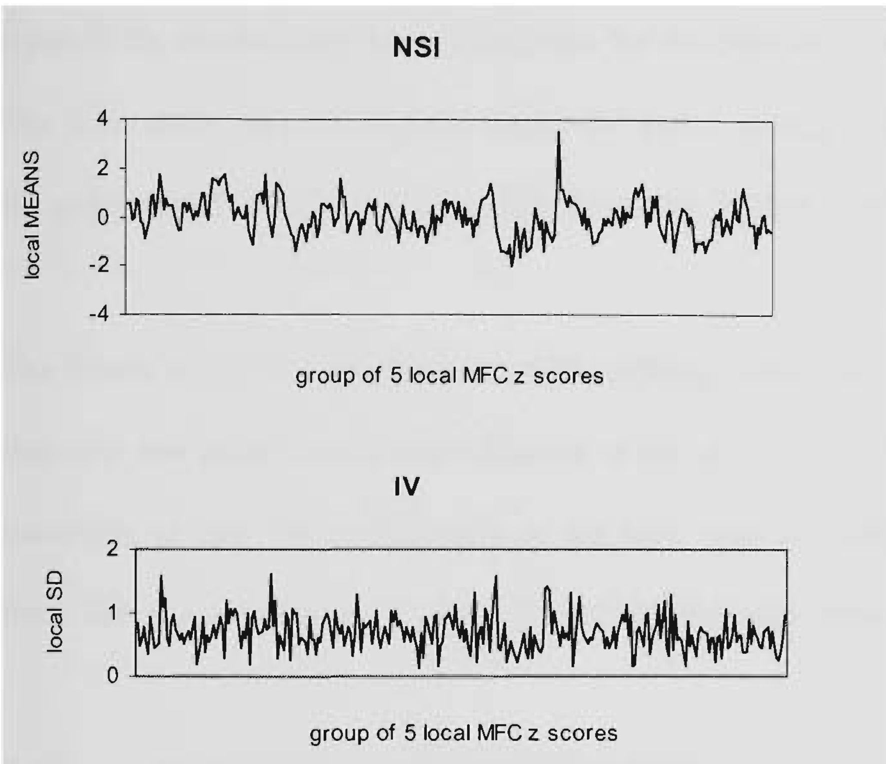

Figure 5.3.5.1 Characterising the fluctuating nature of the NSI and the IV taken from the left MFC data of subject E7. The horizontal axis is the mean of the local MFC z scores for the NSI; and the standard deviation of the local MFC z scores for the IV. Each point on the horizontal axis represents the mean of the five local MFC z score points. The next represents the mean of the next five local MFC z score points.

Although the measures of the NSI and the IV have been applied together elsewhere (Hausdorff et al., 2001) and exhibited similar features, a constructual explanation of the NSI, DFA and IV inter-related properties are not addressed. Further application of the NSI has demonstrated inconsistent characteristics of fluctuation dynamics among different subject groups. A pathological group with amyotrophic lateral sclerosis (ALS) has higher NSI values

(0.69 ± 0.05) in comparison to a healthy control group (0.67 ± 0.02), whereas subjects with Huntington's disease and Parkinson's disease were lower than the control group (HD 0.54 ± 0.03; PD 0.64 ± 0.03). The magnitude of the NSI values are comparable to those obtained in this study, excluding the finding from the ALS pathology group. Hausdorff et al. (2000) reported the mechanisms that may account for the unexpected finding are currently unknown. The same study reported highest values for fractal scaling in the control group followed by the group with Parkinson's disease (C 0.91 ± 0.05; PD 0.82 ± 0.06).

The results of the NSI signaling wave like drifting in the MFC suggests that when collecting data over few trials is not a true reflection of the gait. This is because at some time within the patterning of gait, the performance of the MFC may be higher than at another moment in time. This is also supported by the presence of long-range correlations.

5.3.6 Support from Poincaré plot statistics

It is interesting to note the moderate correlation between the DFA and variability ratio. To hypothesise how the values of D_2 and D_1 relate to unsteadiness in MFC is premature as this investigation will provide an original source for future comparisons. This may provide some support for validating the presence of fractal correlations in the MFC data. Fractal like processes, evident by long-range correlations, are in a way, a relationship between macroscopic and microscopic components (Meakin, 1998). The variability ratio is the quotient between the short- and long- term variability. Thus, the moderate correlation between the variability ratio and the DFA ($r = -0.468$, $p < .004$) may be justified from these common concepts.

5.4 ASYMMETRY

5.4.1 Is gait asymmetric?

One of the aims of this dissertation was to examine right and left limb symmetry or, in other terms the level of equality between the left and right for the parameters measured. From the visual inspections of the Poincaré plots and the histograms, the shapes between the left and the right limbs are often different within subjects. These shapes provide a valuable insight into symmetry because they are representing hundreds of gait cycles. Therefore, from these illustrations alone, qualitative assessment of reflections between the left and right shapes suggests the presence of asymmetry in both young and elderly. These shapes also indicate that asymmetry is not redundant as some subjects show remarkable similarities between limbs, and that a broader sense of symmetry may be evident when considering key features of the graphs. If we were to group all the young subjects left limb data together and compare it graphically to the right side, the prospect of symmetry increases. A similar case would be found for the elderly group. This has been a general finding from other studies (Gunderson et al., 1989; Crowe et al., 1997), which leads to the notion that grouped normative data is misleading. After encountering the effects of group comparisons when analyzing symmetry, Gunderson et al. (1989) stated that "symmetry can neither be assumed nor generalized for the variables studied but must be examined in view of idiosyncratic subject variability."

The quantitative descriptive statistics of the MFC data support the asymmetric features of the histograms. In general, many subjects displayed MFC asymmetry for the mean, standard deviation, skewness and kurtosis based upon recommended absolute symmetry index (ASI) values above 10% (Giakis et al., 1997). This is contrary to what Giakis et al. (1997) found by using the same index to determine asymmetry, that ground reaction force are symmetric from 10 young adult subjects. Ground reaction force symmetry was confirmed elsewhere Hamill et al. (1984). This dissertation found varied differences among the subjects within both groups for quantitative measures, supporting the visual interpretation of the histograms and Poincaré plots. Within subjects, the absolute symmetry index (ASI) values range from 0 to 161 percent. Although no significant differences were found between groups for asymmetry, the elderly did demonstrate higher absolute asymmetry for all parameters apart from skewness, which was similar for both groups. The high ASI values are evident within both groups. Similarly, values representing symmetry are also present in the elderly group. This is expected due to the functional status of the elderly group. All the mean ASI values for the descriptive parameters displayed by the elderly group were greater than 10 percent apart from skewness, which demonstrated a value of 9% for both subject groups.

No other studies have investigated bilateral symmetry characteristics of the minimum foot clearance event. Supportive evidence for these results comes indirectly from a study by Crowe et al. (1996). These authors investigated gait symmetry from oscillations in the body center of mass derived from ground reaction forces of consecutive left and right foot falls. The position of the center of mass over the base of support was the parameter they described as being an important objective of walking. This is somewhat similar to the concept of the minimum foot clearance applied to this dissertation. They concluded that gait is consistently asymmetric with small fluctuations. Given that the center of mass is asymmetric between left and right sides of the body during the stance phase of gait, it certainly has implications affecting the spatial trajectory of the foot during the contra-lateral swing phase. In addition, Giakis et al. (1997) reported medio-lateral asymmetries from ground reaction forces. The presence of medio-lateral sway asymmetry during stance is consistent with the findings by Crowe et al. (1996). These findings from ground reaction force asymmetries do not give direct support for the minimum foot clearance results, however, they do challenge the notion of Sadeghi et al. (2000), who suggest spatio-temporal parameters are not an informative insight into gait symmetry because they describe the effect rather than the cause. The view of Crowe et al. (1996) appears more relevant, who believe that parameters concerned with the body center of mass are most important.

The unsteadiness parameters that are most related to traditional measures of variance were generally higher than the parameters that analysed the moment-to-moment changes in variance. For example, the detrended fluctuation analysis (α), variability ratio (μ), non-stationary index (NSI), and inconsistency of the variance (IV) were all below 15% for group mean values. In contrast, the short-term variability statistic from the Poincaré plot (D1), coefficient of variation (CV), and standard deviation (SD) are of a high variability indicated by ASI values for the group means.

Examination of the extent of symmetry within subjects demonstrates some inconsistencies. This is likely due to inter-parameter correlations describing unsteadiness are not entirely redundant. For example, the DFA (α) and the variability ratio (μ) share a significant negative correlation for pooled data ($r = -0.468$, $p < .004$) however the correlation is not significant when observing the symmetry index ($r = 0.01$). This is demonstrated from the results of subject Y4, who is symmetric for the DFA (SI = 1.2), however, for the variability ratio this subject is considered asymmetric (SI = - 14.2). Hence asymmetries are unpredictable and difficulty arises when trying to seek an explanation by correlating variables. Further testing on surrogate data sets is required to gain understanding of the links between mathematical properties that characterise the unsteadiness parameters.

Studies analysing the MFC from unilateral methods will have reported a different result in comparison to the contra-lateral limb based on the findings from this study. Investigations from unilateral methods (e.g. Winter, 1992; Karst, et al., 1999; and Best et al., 1999) can be assumed to have reported values that are not representative of the entire bilateral system.

5.4.2 Is the MFC asymmetry compensating for functional purpose?

Sadeghi et al. (2000) argues that asymmetry of two corresponding parameters (e.g. local asymmetry of left and right MFC) may represent natural functional differences rather than appear as a consequence of abnormality. These functional differences may be attributed to adaptations and compensation within the link system. For example, if one limb has been amputated, the hip musculature will need to adapt to compensate for the absence of calf musculature in the prosthetic limb. Hence, asymmetry will more than likely be found when comparing the 'local parameters' of the left and right hip musculature. Hence, MFC may represent functional differences due to compensations in other areas of the link system.

Further exploration of compensating strategies takes a different perspective by analyzing the descriptive features of the MFC. The hypothesis being that decreased variability (standard deviation) and increased skewness or kurtosis compensates for low MFC values. These values will indicate a control strategy at the minimum foot clearance to reduce the likelihood of ground contact during the swing phase. Hence, given the possibility of asymmetry in SD, skewness or kurtosis, a conclusion may be drawn that this asymmetry is warranted for establishing safe walking, highlighting the fact that asymmetry is beneficial.

Subject results (Figure 4.4.7.1) indicate a compensating strategy that may be a generalised feature associated with low MFC values in both young and elderly persons. It does appear that comparatively lower values in standard deviation, together with higher skewness and kurtosis values are associated with the low MFC limb. It is interesting to note that this limb commonly appears to be the right limb for both young and elderly groups. A relationship was not found when Spearman's correlation was applied to a SI applied to leg length differences and then compared to MFC asymmetries. These results appear to suggest that for SD, S and K the presence of asymmetry is a functional benefit to help negotiate the terrain rather than a general sign of poor gait control. This compensation may be attributed to low MFC or MFC asymmetry (taking notice that MFC asymmetry is not present in all subjects).

Given that the ground is level, the asymmetry of the MFC means would appear not to be related to negotiating the terrain, but rather the propelling and stabilizing objectives of gait. One group of authors suggested that a low MFC may help to stabilise oscillations of the body center of mass by reducing rotations of the link system components (Crowe et al., 1996). MFC asymmetry may be a compensating effect seeking to achieve body center of mass symmetry from one stride to the next. Higher asymmetries in the elderly compared to the young may possibly be due to a history of perhaps minor ailments that have caused permanent between limb functional inequalities. Additionally, research evidence indicates compensations are required within the sensory systems due to the aging process, which can have an effect on controlling locomotor tasks (see sections 2.1.2 and 2.4). This could explain the difference of consistency in the SI between groups. The young generally display a higher MFC mean for the left limb, however the elderly group does not indicate such unanimous tendencies. Another prospect is that the left and right link systems have their own idiosyncrasies inherent within their respective motor synergies due to the anatomical and physiological variations between the limbs. This is also affected by the aging process, which would explain higher MFC asymmetries in the elderly.

5.4.3 Is the MFC asymmetry an effect of between 'link system' role differences?

MFC may be questioned as an appropriate parameter to use for investigating functional gait symmetry, given that it may be identified as a parameter that describes an effect of compensating mechanisms rather than an indicating cause of pathological asymmetry. This issue will be addressed in the remaining section of this chapter. Firstly, it is of interest that eight of the ten young subjects all have a higher left MFC limb, and that the other two subjects are within 2% of symmetry based upon the SI, suggests that there may exist a specific role function rather than compensating mechanisms producing MFC asymmetry. This finding begins the exploration of limb dominance theory and the concept of laterality and how it may be related to MFC asymmetry.

In brief, laterality (limb dominance) refers to a preferred link system to perform certain challenging tasks. Hart and Gabbard (1998) designed an inventory to identify this preferred link system, which is also called limb dominance. All tasks involved bilateral coordination/control, meaning that mobilizing one limb while stabilizing the other limb was a prerequisite to perform the task. For example, kicking a ball requires both stability and mobility of the link system. Another example from the inventory was drawing initials in a sand box. A common preference found by Hart and Gabbard (1998) was the right link system (RLS), such that the right limb was used to mobilize, and the left limb to stabilise. The relationship of the minimum foot clearance to limb dominance testing is evident. The minimum foot clearance is also an example of bilateral role differentiation. At successive bilateral steps, the performance of the left link system is compared to the performance of the right link system. The difference in height can be argued that it is not a measure of limb dominance performance. However, dynamic measures of fluctuating changes (unsteadiness parameters) reveal the level of control underlying each link system. Therefore, from hundreds of strides, the minimum foot clearance event can be regarded as a definitive limb dominance task for both the LLS and the RLS.

From the SI results into the unsteadiness parameters, large effect sizes between the two groups indicated elderly ASI were nearly 100% higher. The DFA, NSI, D1 and D2 show greater asymmetry for the elderly group, indicating the degree of control is asymmetric between the left and right link systems. Small differences can be explained as possible limb dominance features, however larger differences are more probably indicating breakdown in functioning within either the left or right link system. The link system that is demonstrating a significantly lower DFA value is most likely representing the non-dominant limb because of the breakdown in neural control. If this is the case, the DFA results demonstrate that all subjects apart from a minority (Y6, E1, E3 and E7) all have greater control over the right limb. This supports the research, which suggests that approximately 80% of the population prefer to use the right leg for mobilizing purposes (Peters, 1988).

It is not within the scope of the research design to suggest that the unsteadiness parameters are able to identify the dominant limb. Nevertheless, it is interestingly that young subjects generally showed greater stability for the right limb in the DFA, variability ratio, NSI and IV unsteadiness parameters. The Poincaré plots showed that two from the ten young subjects had lower values of control in the left limb. The elderly also had two subjects from the sample of eight who demonstrated greater control in the left limb. E7 is the most notable outlier, consistently showing greater control over the left limb. Correspondence with E7 following the analysis revealed that they preferred the right limb when kicking a ball. This was the only subject that refuted the speculation that large DFA asymmetry will expose the dominant limb. However, because there were no tests conducted for limb dominance, and E7 did report having undergone back surgery several years prior to testing, the concept of a relationship between the DFA and limb dominance is still a likely possibility. Further exploration of the quantification relationship between the unsteadiness parameters is needed.

Given the body's postural instability at the MFC event, the finding of long-range correlations demonstrates a self-organized, non-random structure when coordinating the complexity of this task. Although these correlations did not break down with aging, it appears that elderly groups are unable to maintain symmetrical structuring of correlations between limbs. A technique adopted from heart rate variability research (Brennan et al., 2001) supports higher asymmetry for the elderly in the long- and short- term variability of MFC data. Higher asymmetry was found in the elderly group for both short- term (young 12.37 ± 11.85; elderly 20.241 ± 14.75) and long- term (young 23.13 ± 19.05; elderly 33.70 ± 19.36) variability measures. It may be that elderly people begin to 'place all their eggs in one basket' by favoring a particular link system.

The asymmetry in the MFC control is definitely a cause and not an effect of the local parameters identified by Sadeghi et al. (2000). It does not appear rational to suppose that the breakdown in control over the performance of the MFC event, being a bilateral control task, is serving a functional purpose. If this breakdown is evident at MFC, then it would likely be reflected in other areas of the link system.

Getting back to the original question regarding asymmetry of the MFC mean being functional or compensating, suggestive that it may be more good than bad, notice needs to be taken from link system control asymmetries. Is it the limb with the greater control, possibly the more dominant limb, or is it present in the limb with poor control. Poor control over the limb that is lowest to the ground would not be a good adaptation, even if it was somehow compensating to improve body center of mass stability. Poor control over the high MFC limb may be energy inefficient, nevertheless it would ensure safe transition and negotiation of the terrain. The results from the elderly group suggest that breakdown in the control of the link system is possibly affecting the height of the MFC event. An equally likely alternative choice to distinguish between the functional task performance between the left and right link systems could have been related to CV. The CV is an important descriptive parameter because it indicates the relative risk factor within a limb because it not only considers the mean value, but it also accounts for the magnitude of the variability occurring for this value. A low MFC mean and a corresponding high standard deviation indicates a risk of ground contact.

Another alternative for higher asymmetry of the MFC means in the elderly can be related to task roles of laterality. If one limb is provided with an ability to propel the body with more energy, a corresponding high MFC value may arise. Alternatively, if one limb is able to stabilise with a high level of control but does not possess the same propelling energy, the contra-lateral MFC may be hypothesized as being comparatively lower. This concept may explain the mean differences commonly found between the left and right limbs, and may suggest that the elderly have greater differences between the roles carried out by the left and right limbs. The results show that the young subjects should be higher for the right limb if this is the case.

Gunderson et al. (1989) reported no relationship between limb dominance and gait asymmetry measured by knee range of motion. The reasons for the lack of effect may be related to the knee range of motion being an effect rather than a cause parameter. Alterations in knee joint angle, like other joints within the link system (see Figure 2.5.1.2), are functioning to serve the planned trajectory of the distal end point of the foot during the swing phase (Winter, 1992).

A hypothesis may be proposed at this point as to why the young subjects demonstrate good control over the limb traveling closest to the ground. Whereas the elderly subjects show inconsistencies mainly due to higher variability within the group. The dominant link system is composed of both relatively greater stability and mobility control. Therefore a possible strategy of the locomotor system in terms of negotiating the terrain may be to allow the limb with the greatest ability to adapt to perturbations to be the 'reconnaissance' limb. The non-dominant limb would serve as the following limb, receiving feedback from the contra-lateral dominant limb.

The neural exchange between the support limb and the mobilizing limb was highlighted by the cross-reflex mechanism in section 2.6.1. The inputs governing the self-similarity structure within each link system coordinating the MFC provide interesting speculation. For one, it signifies the presence of a motor synergy linking the support limb to the mobilized foot, which, in the act of undergoing dynamical changes and alterations is subject to the 'memory' of motor patterns, executed hundreds of strides earlier. The young subjects indicate 'memory' exists equally between limbs, however, the elderly demonstrate asymmetry. This inequality of bilateral control in the elderly group may be due to ageing processes.

The lack of control from left-to-right as opposed to right-to-right or left-to-left appears more prevalent in the elderly group, although this is based upon the relative differences of the two group's effect sizes (i.e. the elderly group has a higher effect size which indicates that their inter-limb relations are not as similar to the intra-limb relations in comparison to the younger group). To provide some support to the notion that the elderly group does not display the same degree of between limb coordination as the younger group the symmetry index was applied to all variables. The results indicate that the elder group does have larger differences when comparing one limb against the other. Although this is not as sensitive to the moment-to-moment measures provided by the non-stationary index, inconsistency of the variance and the Poincaré statistics that can be applied to the inter-limb coupling, it does propose the unlikelihood of entrainment when within one limb vast differences are occurring in comparison to the other.

To determine if this indeed was the case, determining the level of coupling or entrainment from the dominant limb to the non-dominant limb should be more controlled over the short term, than the alternative scenario. This dissertation did not go as far to explore the directional level of coupling existing between the limbs. A more general approach to describing the inter-limb coupling relationship is provided in a following discussion (Section 5.6).

In summary, the MFC is asymmetric for various parameters, occurring within both groups, but more prevalent in the elderly. It is suggested that the cause of these asymmetries is a manifestation of compensating and adapting requirements in gait. It is evident however that when analyzing between group differences from bilateral gait studies, simply grouping the left limb data together for all subjects can be fraught with masking the true nature of asymmetry. The MFC is an appropriate task for determining asymmetry because it clearly is a role that differentiates between the mobilizing and stabilizing features. The low MFC limb is not necessarily indicating the performance of a common task, such as propelling or stabilizing, however when it is considered with the relative level of control it provides new insight into the function of the locomotor system. Specifically, it represents the global output determined by the motor unit synergy that is controlling the particular link system. It also represents the limb most at risk of ground contact and given the predominance of tripping in the elderly, this serves as an appropriate choice. Further still, the minimum foot clearance is an end point control task for propelling the body in forwards progression, an objective equal to maintaining stability of the center of mass.

5.5 AGING AFFECTS ON GAIT UNSTEADINESS

The elderly have generally shown that they have less control over the low minimum foot clearance limb in comparison to the young group, which is possibly reflecting an unsteady walking pattern. There are no significant differences between the subject groups that can strongly confirm that elderly are more unsteady based upon the standard deviation and coefficient of variation parameters. The dynamic measures of MFC control, however, have indicated greater sensitivity to this occurrence. Moderate to strong effect sizes were found in μ, α, and the NSI. A weak effect size in the IV showed a slightly higher degree of unsteadiness of the elderly group for the low MFC limb. The NSI was the only parameter that resembled an effect between the young and the elderly groups when both left and right limbs were pooled. This indicates the importance of analyzing gait unsteadiness from a bilateral perspective and because of asymmetric gait patterns.

The standard deviation and coefficient of variation are the most widely reported parameters that have been successful in describing the relationship between falls and gait patterns (e.g. Maki, 1997; Gabell et al., 1984). This dissertation did not demonstrate a higher standard deviation in the elderly group for either the low MFC limb or when both limbs were pooled together. The elderly demonstrate inequalities between the limbs for the SD, indicated by a moderate to strong effect size (ES = 0.59). The coefficient of variation showed a moderate to strong effect for the low MFC limb. However, this difference may be more so related to the lower mean MFC value found within the elderly group. In any case, a low mean MFC value and a standard deviation that does not compensate, predisposes a high level of risk for tripping on an unseen obstacle. For this reason, the CV is an important parameter for indicating the risk of ground contact at the MFC event.

The effect of fatigue on unsteadiness (Figures and 4.5.2.4.1 and 4.5.2.4.2) was found to be an interesting factor for understanding the mechanisms that may contribute to the decreased control over the low MFC limb found in the elderly. The interesting aspect of this analysis relates to the opposite correlation between the NSI and the IV. The physical implications of a negative correlation can be readily interpreted (section 5.3.5), however, a positive correlation poses an irregularity in the relationship between NSI and IV. The physical representation associated with the NSI caused similar intrigue with Hausdorff et al. (2000). The authors found that among four subject groups, including three pathological and one control, the NSI placed the control group results between a pathological group at one end of the scale, and the two other pathological groups scaled below. The group situated above the control group displayed a comparatively lower DFA value. Previously, high NSI values were associated with good control because it seems to suggest the presence of drifting (Figure 5.3.5.1), and therefore a corresponding decrease in the IV was apparent. For the low MFC limb, the elderly display a higher NSI and a higher IV. Is this the result of fatigue?

It is of interest to note that the IV stabilised for the elderly group after the second segment, while a slight increase in NSI occurred simultaneously. This might imply that fatigue may not be associated with MFC control, but these changes may relate to the settling into a familiar walking pattern that is commonly found in treadmill walking (Matsas et al., 2000). MFC descriptive statistics have demonstrated stabilisation after 10 minutes of treadmill walking (Best et al., 1999). Therefore, the unsteadiness found in the low MFC limb is not likely to be due to fatigue.

The effect of treadmill walking on the pattern of gait can be argued for contributing to unsteadiness because it is an unnatural terrain. In contrast to the young group, the elderly subjects demonstrated a significantly ($p < .01$) lower treadmill speed in comparison to their overground speed. This may suggest that the elderly felt less comfortable on the treadmill in comparison to overground walking. However, the difference in speeds is not expected to be due to unfamiliarity of treadmill walking because the subjects were recruited from community health clubs and all the elderly subjects indicated that they have recently experienced treadmill walking and are competent to walk for at least 20 minutes.

The unsteadiness found within the elderly group for the low MFC limb may also be a conservative measure in comparison to overground walking. Dingwell et al. (1999) found that treadmill walking reduces variability and suggests that treadmill walking creates an artificially consistent gait pattern in comparison to overground walking. The author believes that by imposing a constant speed, the treadmill reduces the normal stride-to-stride variability. The task of obtaining comparitive MFC data from an extended overground walking duration is related to experimental set up limitations. Special camera tracking devices need to be able to follow the subject by traveling free of jitter over an extended distance.

The effect of walking speed, fear of falling, functional mobility level (up and go), and flexibility are factors that are associated with increased unsteadiness indicated by Hausdorff et al. (2001). Ankle range of motion was not correlated with the DFA in either the low MFC limb or the grouped left and right limbs, within and across both groups. Three elderly males were the only subjects who reported a lower than perfect score for the fear of falling survey adapted from Tinetti et al. (1990). The lowest score of 130 from a possible 140 was subject E3, who indicated a broad range of concerns when performing seven out of the 14 listed daily activities. Subject E1 reported a five out of ten for the task of using public transport. All subjects rated themselves as at worst fairly confident when performing daily activities. There was no correlation with the fear of falling survey and the up and go test times. There was no correlation between these measures and MFC unsteadiness.

Walking speed and the influence upon gait unsteadiness is a contentious issue. In this dissertation, treadmill and overground (baseline test) walking speeds did not correlate with the measures of MFC unsteadiness across the limbs when the left and right data was pooled. A significant, but nevertheless a moderate correlation ($r = 0.552$, $p = .018$) was apparent between the DFA and treadmill speed when considering the low MFC limb. The possible effect of treadmill speed and long-range correlations in the MFC event was not expected given the findings from previous research into overground walking speed and fractal scaling. Hausdorff et al. (1996) examined the changes in the DFA in normal, slow and fast walking. Hausdorff et al. (1996) found that walking speed did not affect the DFA scaling and therefore, if constraints are placed on the subject by the treadmill, the DFA results should be resistant to these effects.

5.6 MFC COUPLING

5.6.1 Intra- and inter-limb differences

For both subject groups, the quantitative and qualitative results of the Poincaré plots show the difference in the dynamics between intra- and inter-limb processes. The inter-limb features of the Poincaré plots described as blobs and squashed blobs in the nomenclature stand out from the intra-limb plots. The relativity ratio 'μ', demonstrates that for both elderly and young groups, coordinating inter-limb gait is a task of higher complexity relative to intra-limb. The values for both D1 and D2 indicate an increase in both the short- and long- term variability. The difference in variability is not due to the mean differences because the left and right limbs were normalised.

The Poincaré plot statistics describing the within and between limb differences are supported by the inconsistency of the variance (IV), and the non-stationary index (NSI). The negative correlation between the IV and NSI values is evident when describing the differences between the inter- and intra-limb dynamics of the MFC. A lower value for the non-stationary index and a higher value for the inconsistency of the variance is associated with the inter-limb process when compared to the intra-limb processes (see Table F4, Appendix F).

The inter-limb results for the IV and the NSI show less difference with the intra-limb IV and NSI results in comparison to those of the Poincaré plot statistics. The effect sizes comparing between the inter- and intra-limb dynamics are larger for the Poincaré plot statistics when compared to the IV and NSI. There is a strong correlation between the relativity ratio μ and the NSI ($r = -0.875$, $p = .000$), and the IV ($r = 0.823$, $p = .000$). This strong correlation between the dynamic parameters as well as the larger effect sizes for the Poincaré plot statistics, may suggest that the Poincaré plot is a more sensitive method for determining the dynamics of inter-limb coordination.

The lower NSI for inter-limb dynamics compare to the intra-limb dynamics is probably due to increased short- term variability that is occurring between successive left and right MFC events. Because of this, the inter-limb local means (five consecutive MFC values) are more inconsistent than intra-limb, therefore drifting local means will be less evident as measured by the NSI. This observation is opposite to that commonly found for intra-limb results. The rise in the inter-limb IV in comparison to the intra-limb IV supports the notion that the local means are not drifting. The lack of drifting for inter-limb dynamics suggests that the left and right limbs are not coordinating in a similar fashion to those observed from intra-limb.

The short- term variability statistic D1 from the Poincaré plot is slightly higher in the elderly group, but across both groups, the effect size between the intra-limb and inter-limb is very large (young: ES = 10.81; elderly: ES = 11.76). This supports the notion that successive left to right MFC events are loosely entrained in comparison to intra-limb entrainment. The long- term variability statistic D2 demonstrates a similar finding, which appears to present a contradiction. From the intra-limb results, increased long- term variability suggests a higher level of control function. This is due to high D2 values decreasing the relativity ratio. Given this rationale, a higher level of control would apparently exist within inter-limb dynamics, which is not the case as measured by the IV, NSI and μ. Hence, when describing inter- and intra-limb control using the Poincaré plot statistics, it is important to consider long- term variability in view of the short- term variability, such as that expressed by the relativity ratio, 'μ'.

The larger than expected dynamic variability for inter-limb processes, demonstrated by the relativity ratio, is surprising considering the principles of inter-limb coordination. The shape of the Poincaré plots and the values of μ, indicates that the coordination of the synergies between link systems (i.e. coordinating the LLS and the RLS) is a more complex task than coordinating repeating synergy cycles within a link system. Observations from von Holst recognized the simultaneous attraction and repelling mechanisms of control between coordinating limbs (Turvey et al., 1993). Further, Amazeen et al. (1998) suggests that principles of control at one level of coordination can be applied to more complex levels, such as within limb and between limb coordination. In contrast to these theories, the results from this dissertation show that the left and right link systems appear less influenced by their contra-lateral peer, and rather operate under independent mechanisms.

The 'blobs' and 'squashed blobs' defined in the nomenclature (Section 4.5.3.1) appear to reflect random like distributions on the Poincaré plot. Although the Poincaré plot method has been referred to in bimanual coordination publications (Kelso, 1995), unfortunately there is no comparitive data that can be found that has utilized the statistics of the Poincaré plot to differentiate between intra- and inter-limb control processes. Surrogate testing of randomly shuffled data is represented as a square shape plot. Therefore, the left to right interactions are not totally random and the 'squashed blob' may be representative of entrainment between the left and right link systems.

The concept behind the DFA method warranted an attempt to provide further information on the differences between inter- and intra-limb coordination. The DFA model is designed to characterise the structure within a pattern. The inter-limb sequence can be composed of two separate structures alternately sequenced together. Two surrogate data sets represented a hypothetical left and right MFC data distribution. One represents random noise and the other $1/f$ type noise. Thus, the two data sets modeled inter-limb sequencing and were then analysed using the DFA. The scaling exponent typically returned a value midway between the two DFA values for the independent surrogate sets of data. For example, 1/f noise is represented as a DFA value of 1.0, and random noise as 0.5, the inter-limb model returned a value of approximately 0.75. Hence, for the purpose of describing the nature of inter-limb coupling dynamics, the DFA was not able to indicate the short- term phenomenon. Instead, it described the existence of a structure within. For this reason, the DFA method was not used for describing inter-limb coordination.

The link between the relativity ratio and the DFA may allow comparisons from this dissertation to be made with alternative experimental research in bimanual limb coordination. Turvey et al. (1993) analysed inter-limb coordination using hand held pendulums that were oscillated to achieve anti-phase symmetry, which is when the two pendulums are displaced in time by 180° and has similarities to the contra-lateral phase differences in walking. Turvey et al. found that by altering the weight of one pendulum the complexity of the task demands increased, which was represented by the fluctuations about anti-phase symmetry. These fluctuations measured by Turvey et al. were calculated from power spectra laws, such that the fluctuations were described by a scaling factor as different forms of noise. When the pendulums were identically matched, the noise of the fluctuations ranged between Black to Brownian ($1/f^2$), indicating strong correlations. Brownian noise is represented by the DFA as 1.5. When task complexity increased due to inherent inequalities between the pendulums, the noise then moved towards $1/f$ type noise, which is similar to a DFA value of 1.0. The results from this dissertation reported DFA values below 1.0 for intra-limb processes. Given that the parameters NSI, IV and μ demonstrate that task complexity is higher for inter-limb coordination, the achievement of spatial symmetry when walking is far more complex compared with simpler tasks of swinging pendulums. The demands of achieving spatial symmetry between the link system synergies when walking may be related to the many degrees of freedom within each link system and the requirement to maintain the body center of mass over the base of support. Further research into timing differences between the left and right MFC events is required to make more definite comparisons between walking and other bimanual tasks described here.

The elderly appear to display a lower level of between limb coordination judging by the comparative higher values for the relativity ratio, 'μ' (Table 4.5.3.3.1). The Wilcoxen Rank Sum W test did not show a significant difference ($p = 0.214$) between the groups. An effect size of 0.42 for μ (see Appendix Table F4) suggests that entrainment between the limbs is not as strong in the elderly group. Similarly, a slightly lower effect size was found for the NSI (see Appendix Table F4). The lack of between limb entrainment in the elderly group may be due to a variety of possibilities. One possibility is related to the control of body posture from one stride to the next, which is reflected in the MFC variability when consecutive link systems are coordinating. Whitall et al. (1994) observed that inter-limb coupling stabilises as a result of increased postural control when investigating the development of walking in infants. Given that this may be the case found in these results, inter-limb coupling of the MFC may reflect oscillations of the body center of mass found by Crowe et al. (1996). Obviously further investigations into the relationship between intra- and inter-limb coupling during walking and the relationship between the body center of mass and the MFC event are required.

Another possibility for the slight differences found between the groups may relate to the concept of different levels coordinating gait. One form of central pattern generators have been associated with intra-limb coordination and a higher level of central pattern generators is suggested to act upon those within the limbs to provide inter-limb coupling (see Section 2.6.6). Given that this may be the case in humans as these findings are derived from cat gait research, it may imply that these higher levels of control may not be interacting appropriately in the elderly. However, the alternative is equally feasible – such that inter-limb control may be dependent upon the achievement of intra-limb coordination. Further research into this area is required to provide speculation on this issue. More research is needed in the field of inter-limb coordination that addresses weight bearing activities such as locomotion tasks as these are clearly regulated differently in comparison to the predominant experimental designs on upper arm or seated leg motion. Modeling the coupling of inter-limb weight bearing tasks may then contribute to understanding the influence of laterality during walking.

5.7 IMPLICATIONS FOR FALLS PREVENTION

The key findings and indications of this dissertation for how they may improve the current literature on understanding falls risk factors in elderly persons will be addressed in this section. The discussion will consider relating the findings from this dissertation to some of the diverse applications currently circulating within the falls prevention framework. This dissertation has generally addressed the issue of trip related falls by investigating the minimum foot clearance event, although some information may apply to other postural perturbations leading to a fall. It was alluded to in the introduction that trip related falls are the most commonly reported, and hence two related issues will be expanded upon: 1) a persons predisposition to tripping; 2) a persons ability to respond following a trip.

The likelihood of falling following a trip is determined by the increased frequency of tripping (Pavol et al., 1999). The causes that lead to a fall can be related to control and coordination over the leading limb, trailing limb, visual sensitivity, and awareness of the environment surrounds. The type of analysis conducted within this dissertation applies to the probability of tripping over an unseen obstacle. The different tendencies for MFC heights, and the level of control inherent within each link system, becomes relevant to this issue. There have been numerous studies into MFC, however, they remain as constructs for falls risk, where no literature has identified MFC as a risk factor for falling (Karst et al., 1999).

The indication that the low MFC limb in the elderly group is commonly under less control is a new finding that can propose fresh insights. The possibility that a loss of neural control within a link system is preventing the maintenance of adequate foot clearance height, has been supported by Dingwell et al. (1999). The authors found that neuropathic patients displayed a lower MFC in comparison to age-matched control subjects. Those results were obtained from unilateral analysis which does not fully describe the between subject differences as this dissertation has demonstrated. Loss of control associated with low MFC predisposes a person to the risk of tripping over an unseen obstacle.

Research into the biomechanics and neural control system of tripping responses is providing further understanding of how the limbs coordinate following an induced trip (e.g. Pavol et al., 2001; Schillings, van Wezel, Mulder and Duysens, 2000; Pijnappels, Bobbert and van Dieën, 2001). The answers that researchers are seeking relate to characteristics within the elderly that may be influencing the risk of falling following a trip. Although the researchers have not indicated that limb dominance has an influence on response selection following a trip, this dissertation proposes a link. Measures of unsteadiness in the MFC event have revealed a decrease in control within a particular link system of the elderly group. This might suggest that the specific link system will be more inclined to respond inappropriately upon striking an unseen obstacle.

Two general types of trip responses have been identified by researchers (Pavol et al., 2001; Schillings et al., 2000). The limb adjustments following a trip require reflexive input to coordinate an appropriate repositioning of the leading limb and the ability of the support limb to slow the rotation of the HAT (head, arms, trunk) system. Theoretically, the dominant link system will be able to reposition the foot and stabilise the HAT because it is provided with the functional superior mobilizing- and support- limb. Again, following the theory that the DFA results from this dissertation indicates the dominant link system, the elderly group indicates that the non-dominant link system has degenerated in comparison to the young subject group. It would appear probable that investigations into response selection following an induced trip or loss of balance protocol would indicate differences between link systems. A study by Pavol et al. (2001) investigated both limbs in view of determining the role of limb dominance in elderly persons when responding to an induced trip. The authors found no link between the limbs that may have suggested differences between the limbs when responding to a trip. In another study, Rogers, Hedman, Johnson, Cain and Hanke (2001) did not report any between limb differences when analysing the lateral step positioning following an induced sideways pull from a standing position. These findings seem to suggest that response selection is equally assigned within either limb, which is of contrast in view of the limb dominance theories.

The link between MFC unsteadiness and limb dominance and how it may be an influencing risk factor for trip related falls requires further investigation using persons with an experience of trip related falls. Given that there may be an association, the methods applied in this dissertation to determine unsteadiness or level of control will be important for assessing the success of intervention programs. The finding by Hausdorff et al. (2001) indicates that the breakdown of long-range correlations in elderly fallers can be reversed through exercise intervention programs. Hypothesising that poor control over the MFC event predisposes an individual to a fall, exercise and rehabilitation intervention programs will be able to become more specific. This might include exercises that focus upon bilateral support and mobilizing tasks, such as standing on an imbalanced surface while mobilizing the contra-lateral limb.

5.8 RECOMMENDATIONS FOR FUTURE RESEARCH

The presented results are new to the field of gait analysis and hence further investigation in this area is warranted to explore the control of gait and the phenomenon of falls aetiology in elderly populations.

The following recommendations will expand the scope of this dissertation:

- Use a variety of screening methods to determine the dominant limb
- Develop an inter-limb coordination model, taking advantage of the non-linear features of gait and inter-limb coordination, such as an applied version of the Haken – Kelso - Bunz (Beek et al., 2002) equations that can determine limb coupling. Such an approach may also require further exploration of surrogate testing using two independent data sets representing the left and right limbs respectively
- Increase the sample size for significance findings
- Examine pathological groups, especially groups within the community who have experienced falls from tripping
- Obtain overground walking MFC data and compare with treadmill MFC data

- Combine studies with trip response strategies and how laterality influences the performance of the response to a sustained trip
- Further studies of spatial symmetry will take advantage of the full pattern of the foot trajectory during the swing phase of gait. Timing information associated with bilateral foot trajectory will further the understanding of inter-limb coordination during gait.
- To determine the association between the body center of mass and the minimum foot clearance event

5.9 CONCLUSIONS

This dissertation has revealed new information that can have a positive influence on the prevention of falls in the elderly. The following conclusions are drawn from four key findings and are discussed with relevance to past literature in gait analysis and falls studies. These key points relate to:

- Assumption of gait symmetry is subject to individual differences and is generally violated from bilateral minimum foot clearance results
- The limb with the lower minimum foot clearance is significantly more unsteady in the elderly
- Long-range correlations indicate a 'memory' effect in the control of minimum foot clearance
- Inter-limb coordination issues

Firstly, gait is not symmetric as is often assumed in the literature. The minimum foot clearance is an outcome of a global coordination pattern and this parameter indicates that differences exist between limbs. Although minimum foot clearance asymmetry is more apparent in elderly groups, it is not exclusive to this group, which can be attributed to individual differences. Asymmetry can be attributed to reasons expressed elsewhere, such as inherent anatomical and physiological asymmetries, and asymmetries related to adjustments in the control system (Crowe et al., 1996).

The prevalence of minimum foot clearance asymmetry contrasts the respective average heights obtained at this event. Generally, most people do not appear to have symmetry in minimum foot clearance height, hence one limb may be termed 'at-risk' because it travels closer to the ground. The results demonstrate that elderly persons in general, have significantly greater unsteadiness within this limb in comparison to young adults. This is a new result to previous literature examining foot clearance and the concept of unsteadiness attached to the parameter.

A further finding from this dissertation that is new to gait analysis literature is the 'memory' effect underlying the control of the minimum foot clearance event. This has been demonstrated in both young and elderly groups. Previous research has described the minimum foot clearance in terms of traditional statistics, such as the mean, standard deviation and coefficient of variation. Long-range, fractal like structure proposes an insight into the control mechanisms, indicating that the distance between the foot and the ground is not a random occurrence from one step to the next as it is related to hundreds of strides earlier. Further, the breakdown in long-range correlations has been associated with motor control 'atrophy', and the presence of DFA breakdown in the 'at risk' or low MFC limb of the elderly group implies an increased risk of tripping.

An unexpected, although of interest to the literature of gait analysis, surrounds speculation of limb dominance and its presence. From previous studies, most people prefer to use the right limb to mobilize and the left limb to support. Applying this to the minimum foot clearance event suggests that the right limb for the general population should be under the greatest control in comparison to the left at the MFC event. If the DFA is used as a guide for identifying limb dominance, then all subjects confirmed the general rule that the right limb has greater control. Confirmation of this finding was beyond the scope of the dissertation.

The implications of limb dominance control and a low MFC limb may reveal valuable information for understanding gait and influence the prevention of falls. From the results of this dissertation, the dominant link system travels low to the ground and the non-dominant link system is generally higher. If this relationship is reversed, an increased likelihood of tripping appears to be a logical consequence.

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APPENDICES

VICTORIA UNIVERSITY

Victoria University of Technology

Sample Consent Form for Subjects Involved in Research

INFORMATION TO PARTICIPANTS:

We would like to invite you to be a part of a study into:

Bilateral foot trajectory during human gait.

CERTIFICATION BY SUBJECT

I,

of

certify that I am at least 18 years old and that I am voluntarily giving my consent to participate in the experiment entitled: **Bilateral foot trajectory during human gait**

being conducted at Victoria University of Technology by: **Dr. Rezaul Begg, Dr. Russell Best & Simon Taylor**

I certify that the objectives of the experiment, together with any risks to me associated with the procedures listed hereunder to be carried out in the experiment, have been fully explained to me by:

Simon Taylor

and that I freely consent to participation involving the use on me of these procedures.

Procedures:

- Walking over-ground for 15m to calculate normal walking speed
- Timed 'up and go' test (i.e. stand from a chair and walk 3 m, turn around, walk back, then return to sitting)
- Measurement of body height, mass, and leg length
- Attaching three light emitting diodes (LED's) to both shoes
- Video recording of foot motion while walking at a comfortable speed on a treadmill for about 20-30 minutes.
- A safety harness will be attached to all elderly subjects to eliminate the chance of injury if a fall is sustained during testing.

I certify that I have had the opportunity to have any questions answered and that I understand that I can withdraw from this experiment at any time and that this withdrawal will not jeopardise me in any way.

I have been informed that the information I provide will be kept confidential.

Signed: .... }

Witness other than the researcher: } **Date:** ....

....}

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 MCMC, Melbourne, 8001 (telephone no: 03-9688 4710).

Subject Assessment Form

Name: ______________________ Date of Birth: __________ Test Date: ______

Hieght: __________

Weight: __________

Ankle ROM (right / left):

Flexion: __________ / ________

Extension: __________ / ________

Leg Length (right / left): [greater trochanter to lateral malleolus].

__________ / ________ (cm)

Walking velocity: [walking distance 20m, timing over 10 meters].

1. ________ 2. ________ 3. ________ 4. ________ 5. ________

Average (sec): __________

Velocity (m/sec): ________

Velocity (km/h): ________

Up and Go test:

Time (sec): 1.________ 2. _______ Lowest time: ______________

Health Background Comments:

____

____

____

FALLS EFFICACY SCALE

On a scale from 0 to 10 with zero meaning not confident/sure at all, five being fairly confident/sure, and ten being completely confident/sure, how confident/sure are you that you can do each of the following without falling:

	Not Confident at all		Fairly confident		Completely confident
	*		*		*
• Get dressed and undressed	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Prepare a simple meal	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Take a bath or shower	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Get in and out of a chair	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Get in and out of bed	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Answer the door or telephone	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Walk around the house (inside)	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Reach into cupboards or wardrobes	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Light housekeeping (e.g. sweeping, dusting)	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Simple shopping (e.g. groceries)	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Using public transport (e.g. bus, tram)	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Crossing roads	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Light gardening OR hanging out the washing	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10
• Using front and/or rear exists of home	0..... 1 .....2 .... 3..... 4		.....5 .....6 .... 7..... 8 .....9 ...		10

List other situations where you are fearful:

- .... 0..... 1 .....2 .... 3..... 4 .....5 .....6 .... 7..... 8 .....9 ... 10
- .... 0..... 1 .....2 .... 3..... 4 .....5 .....6 .... 7..... 8 .....9 ... 10
- .... 0..... 1 .....2 .... 3..... 4 .....5 .....6 .... 7..... 8 .....9 ... 10

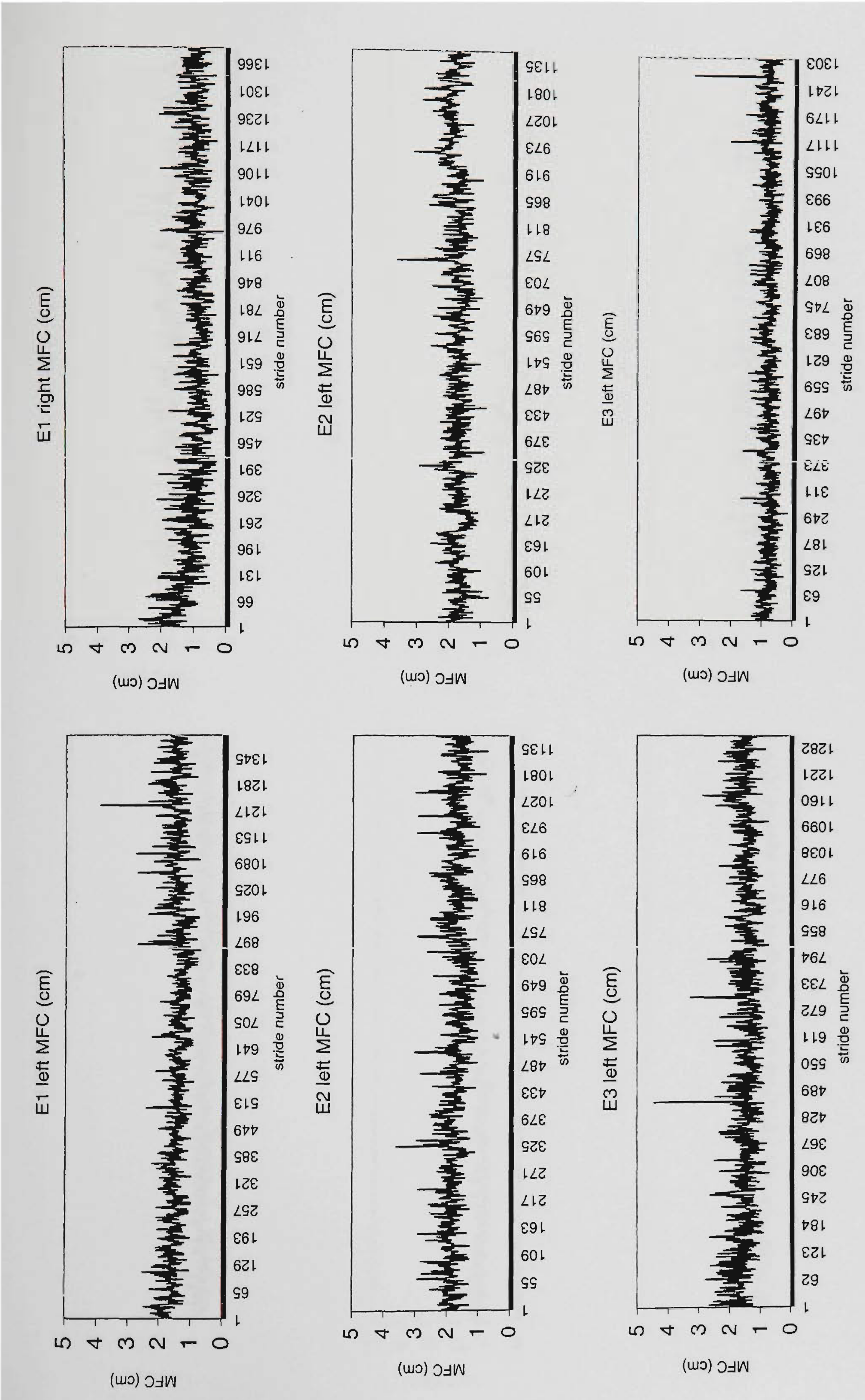
Adapted from Hill *et al.* (1996) Arch. Phys. Med. Rehabil., 77: P1025-9
Tinetti *et al.* (1990) J. Gerontol., 45: P239-43

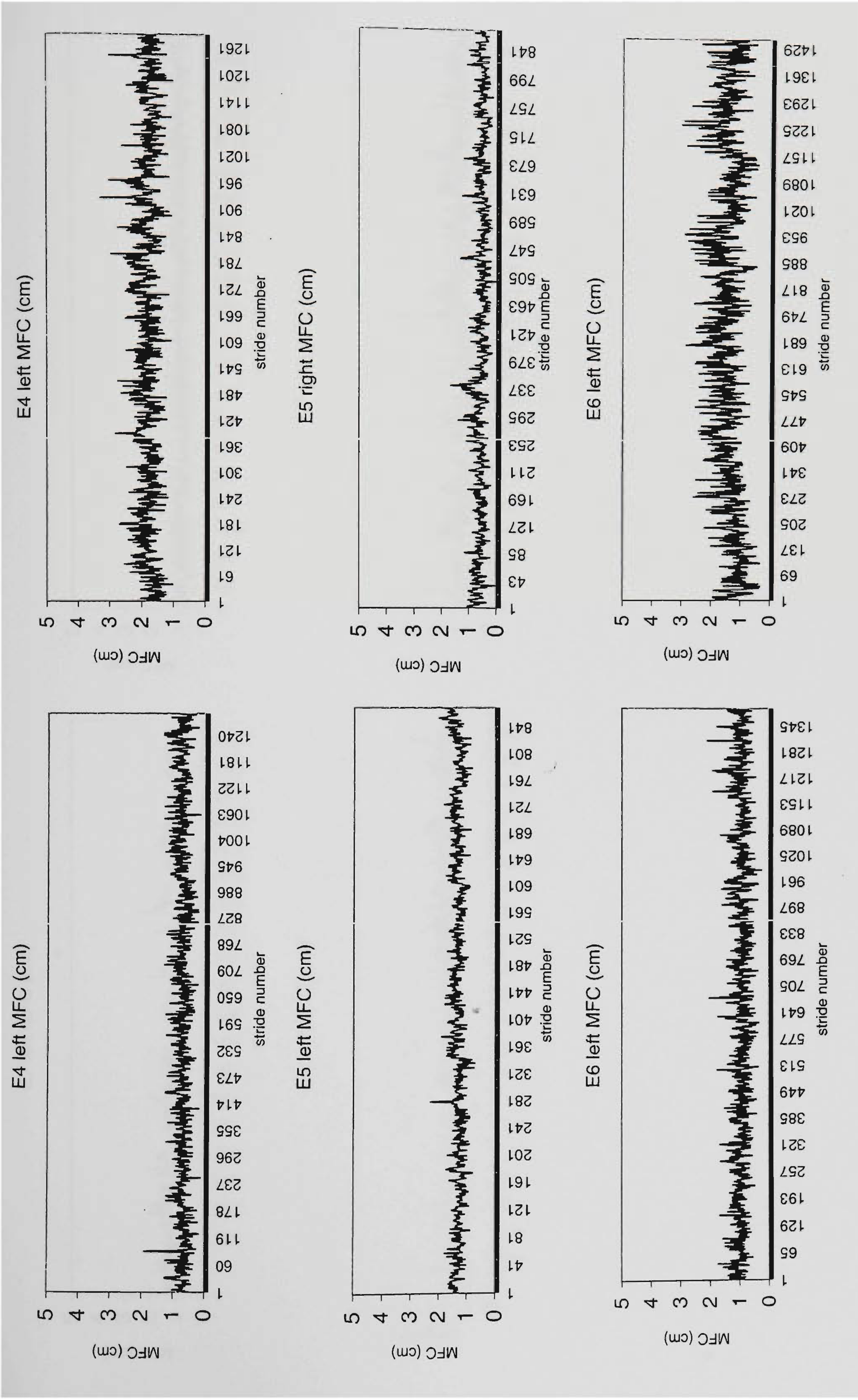
Table D1. *Elderly subject baseline characteristics. Symbolisms: ht – height (cm); wt – weight (kg); bmi – body mass index (kg/m^2); ll l left leg length (m); ll r - right leg length (m); l pf – left ankle plantar-flexion (degrees); r pf – right plantar-flexion (degrees); l df – left ankle dorsi-flexion (degrees); r df – right dorsi-flexion (degrees); l ROM - left ankle range of motion (degrees); r ROM – right ankle range of motion (degrees); OG sp – average overground walking speed (km/h); TM sp – constant treadmill speed for the walking trial (km/h); UAG – up and go test (sec); fear/140 – fear of falling survey scored out of 140points.*

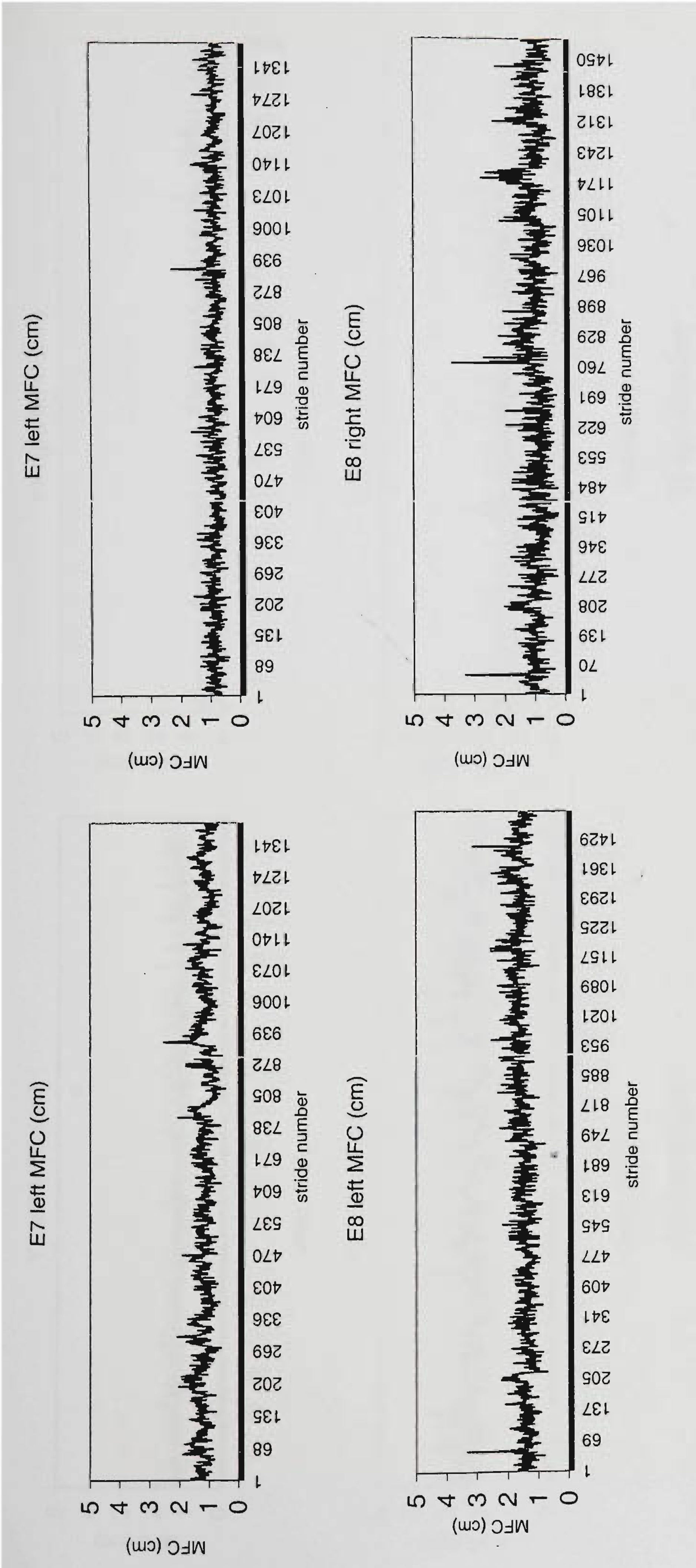
subject	sex	age	ht	wt	bmi	ll l	ll r	l pf	r pf	l df	r df	l ROM	r ROM	OG sp	TM sp	UAG	fear/140
E1	m	78.00	1.74	78.40	26.04	0.82	0.82	60.00	45.00	13.00	12.00	73.00	57.00	5.30	3.80	7.57	135.00
E2	m	77.00	1.62	69.00	26.45	0.70	0.70	60.00	60.00	5.00	5.00	65.00	65.00	4.73	4.00	8.50	130.00
E3	f	68.00	1.49	63.20	28.47	0.69	0.69	60.00	60.00	15.00	15.00	75.00	75.00	5.48	4.20	7.20	140.00
E4	m	70.00	1.77	65.40	20.92	0.84	0.83	45.00	45.00	5.00	5.00	50.00	50.00	4.06	4.10	9.48	135.00
E5	m	75.00	1.68	73.00	25.86	0.82	0.83	46.00	22.00	5.00	0.00	51.00	22.00	4.45	4.30	10.76	140.00
E6	f	76.00	1.53	49.80	21.27	0.82	0.82	60.00	65.00	10.00	20.00	70.00	85.00	4.76	3.70	8.36	140.00
E7	f	70.00	1.58	69.60	27.88	0.78	0.78	70.00	60.00	10.00	10.00	80.00	70.00	5.50	4.00	8.40	140.00
E8	m	71.00	1.70	68.40	23.81	0.78	0.77	48.00	60.00	5.00	12.00	53.00	72.00	5.52	5.50	7.61	140.00

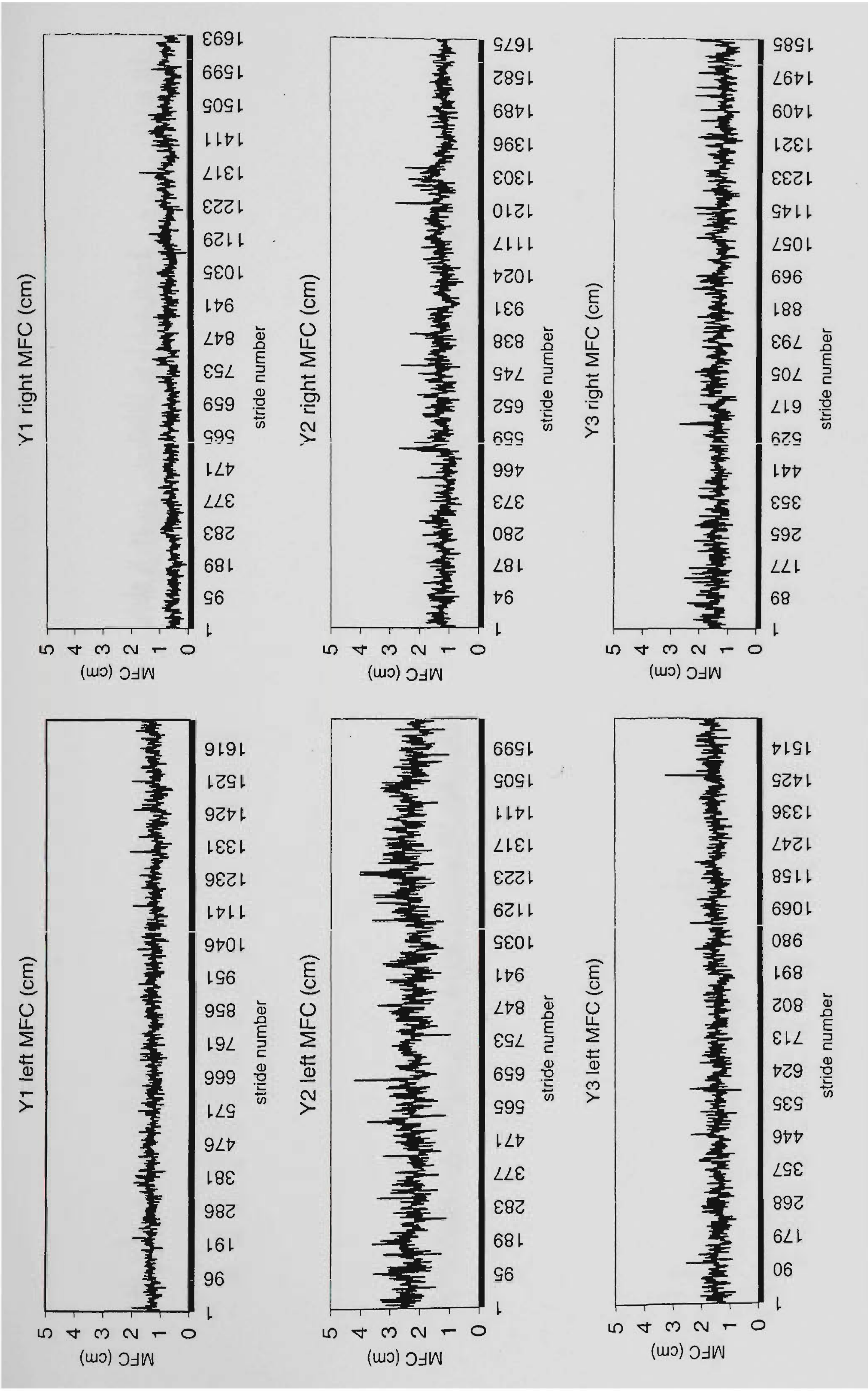
Table D2. Young subject baseline characteristics. Symbolisms: ht – height (cm); wt – weight (kg); bmi – body mass index (kg/m^2); ll l left leg length (m); ll r - right leg length (m); l pf -- left ankle plantar-flexion (degrees); r pf – right plantar-flexion (degrees); l df – left ankle dorsi-flexion (degrees); r df – right dorsi-flexion (degrees); l ROM - left ankle range of motion (degrees); r ROM – right ankle range of motion (degrees); OG sp – average overground walking speed (km/h); TM sp – constant treadmill speed for the walking trial (km/h).

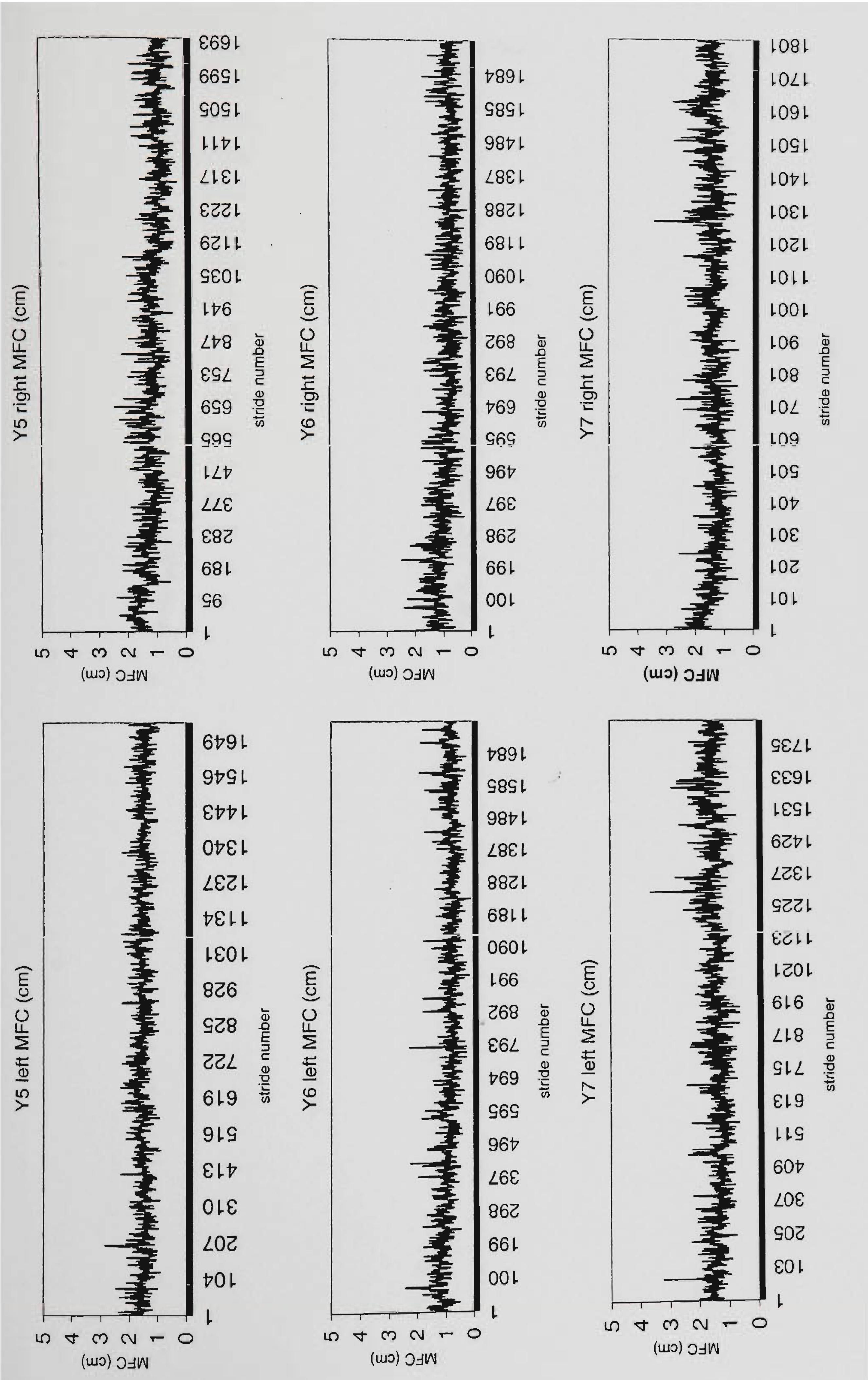
subject	sex	age	ht	wt	bmi	ll l	ll r	l pf	r pf	l df	r df	l ROM	r ROM	OG sp	TM sp
Y1	m	25.00	1.88	72.00	20.44	0.89	0.89	60.00	60.00	10.00	12.00	70.00	72.00	5.06	5.10
Y2	m	33.00	1.83	84.20	25.14	0.92	0.91	55.00	60.00	10.00	10.00	65.00	70.00	5.10	5.10
Y3	f	29.00	1.80	66.60	20.53	0.93	0.91	60.00	65.00	28.00	30.00	88.00	95.00	4.80	4.80
Y4	m	28.00	1.85	75.20	21.90	0.85	0.85	50.00	55.00	25.00	25.00	75.00	80.00	6.00	6.00
Y5	m	28.00	1.79	83.40	26.12	0.85	0.85	70.00	70.00	10.00	5.00	80.00	75.00	4.90	4.90
Y6	f	34.00	1.61	53.80	20.88	0.74	0.74	75.00	75.00	25.00	30.00	100.00	105.00	6.50	6.50
Y7	f	34.00	1.60	65.00	25.39	0.79	0.79	65.00	60.00	25.00	20.00	90.00	80.00	5.80	5.80
Y8	m	30.00	1.80	85.00	26.15	0.87	0.87	55.00	65.00	10.00	15.00	65.00	80.00	5.20	5.20
Y9	f	32.00	1.72	62.00	20.96	0.81	0.81	70.00	75.00	20.00	25.00	90.00	100.00	5.60	5.60
Y10	f	30.00	1.65	67.00	24.61	0.77	0.77	65.00	75.00	20.00	20.00	85.00	95.00	5.30	5.30


E1 left MFC (cm)
E1 right MFC (cm)
E2 left MFC (cm)
E2 left MFC (cm)
E3 left MFC (cm)
E3 left MFC (cm)
MFC (cm)
stride number


E4 left MFC (cm)
E4 left MFC (cm)
E5 left MFC (cm)
E5 right MFC (cm)
E6 left MFC (cm)
E6 left MFC (cm)
MFC (cm)
stride number


E7 left MFC (cm)
E7 left MFC (cm)
E8 left MFC (cm)
E8 right MFC (cm)
MFC (cm)
stride number


Y1 left MFC (cm)
Y1 right MFC (cm)
Y2 left MFC (cm)
Y2 right MFC (cm)
Y3 left MFC (cm)
Y3 right MFC (cm)
MFC (cm)
stride number


Y5 left MFC (cm)
Y5 right MFC (cm)
Y6 left MFC (cm)
Y6 right MFC (cm)
Y7 left MFC (cm)
Y7 right MFC (cm)
MFC (cm)
stride number

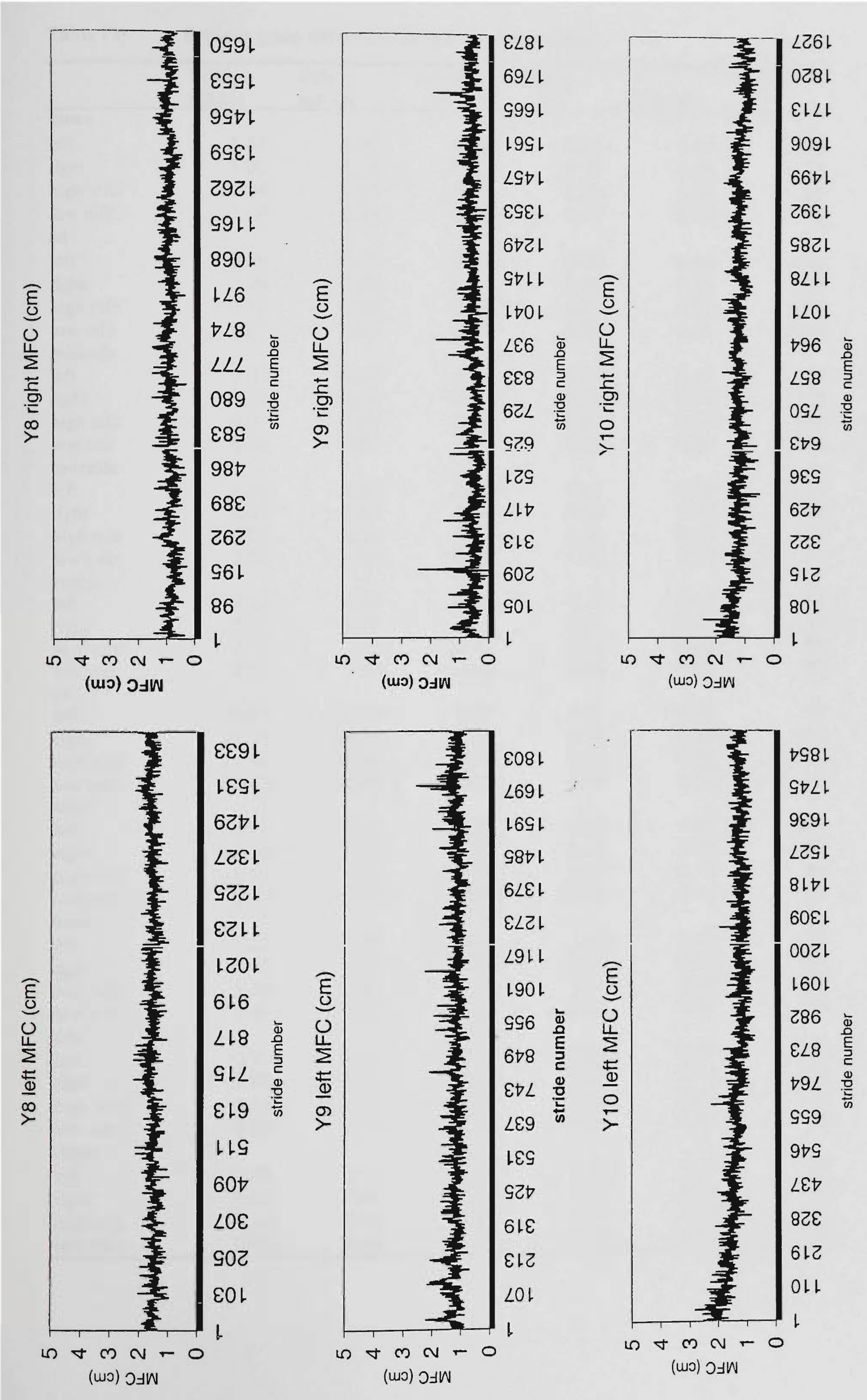

Y8 right MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 98 195 292 389 486 583 680 777 874 971 1068 1165 1262 1359 1456 1553 1650
stride number
Y9 right MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 105 209 313 417 521 625 729 833 937 1041 1145 1249 1353 1457 1561 1665 1769 1873
stride number
Y10 right MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 108 215 322 429 536 643 750 857 964 1071 1178 1285 1392 1499 1606 1713 1820 1927
stride number
Y8 left MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 103 205 307 409 511 613 715 817 919 1021 1123 1225 1327 1429 1531 1633
stride number
Y9 left MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 107 213 319 425 531 637 743 849 955 1061 1167 1273 1379 1485 1591 1697 1803
stride number
Y10 left MFC (cm)
MFC (cm)
0 1 2 3 4 5
1 110 219 328 437 546 655 764 873 982 1091 1200 1309 1418 1527 1636 1745 1854
stride number

Table F1. Between group differences for intra- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
mean						
left	1.44	1.30	0.36	0.33	0.41	39
right	1.09	1.16	0.32	0.47	0.18	71
high mfc	1.44	1.50	0.35	0.21	0.20	95
low mfc	1.08	0.95	0.32	0.35	0.39	38
sd						
left	0.28	0.27	0.08	0.06	0.06	318
right	0.29	0.31	0.11	0.08	0.11	159
high mfc	0.30	0.31	0.11	0.07	0.07	257
low mfc	0.27	0.27	0.07	0.07	0.01	3049
minmfc						
left	0.66	0.54	0.26	0.24	0.47	34
right	0.30	0.37	0.21	0.35	0.23	52
high mfc	0.63	0.60	0.29	0.27	0.12	135
low mfc	0.33	0.31	0.21	0.28	0.09	160
maxmfc						
left	2.91	3.02	0.67	0.93	0.14	94
right	2.59	2.93	0.70	0.70	0.49	32
high mfc	3.00	3.15	0.74	0.53	0.24	77
low mfc	2.50	2.80	0.55	1.00	0.39	28
range						
left	2.25	2.48	0.65	0.77	0.32	44
right	2.29	2.57	0.68	0.60	0.44	38
high mfc	2.37	2.55	0.81	0.54	0.28	68
low mfc	2.17	2.49	0.45	0.82	0.51	22
cv						
left	19.84	21.72	5.88	4.21	0.37	49
right	27.73	28.77	7.56	8.54	0.13	114
high mfc	21.44	20.60	8.78	5.19	0.12	165
low mfc	26.13	29.89	6.08	6.55	0.59	25
skew						
left	0.56	0.55	0.33	0.28	0.03	537
right	0.58	0.69	0.33	0.42	0.28	50
high mfc	0.54	0.49	0.32	0.32	0.18	87
low mfc	0.60	0.75	0.34	0.35	0.44	35
kurt						
left	1.32	1.75	0.76	1.42	0.39	28
right	1.39	2.65	1.66	3.03	0.54	21
high mfc	1.25	1.85	0.73	1.67	0.50	19
low mfc	1.46	2.55	1.66	2.93	0.47	24
dfa						
left	0.77	0.76	0.05	0.08	0.08	150
right	0.80	0.77	0.04	0.08	0.44	23
high mfc	0.76	0.81	0.05	0.08	0.66	20
low mfc	0.80	0.73	0.03	0.06	1.61	6
d1/d2						
left	0.65	0.67	0.10	0.08	0.17	99
right	0.62	0.65	0.07	0.12	0.31	37
high mfc	0.64	0.63	0.11	0.09	0.13	131
low mfc	0.64	0.69	0.06	0.10	0.72	15

Table F2. Between group differences for intra- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
d1						
left	0.22	0.22	0.06	0.05	0.07	240
right	0.22	0.24	0.07	0.05	0.21	89
high mfc	0.23	0.23	0.08	0.05	0.10	177
low mfc	0.21	0.22	0.05	0.05	0.19	84
d2						
left	0.34	0.33	0.10	0.07	0.14	135
right	0.37	0.37	0.15	0.12	0.06	275
high mfc	0.37	0.38	0.16	0.10	0.04	491
low mfc	0.33	0.33	0.09	0.10	0.08	185
nsi						
left	0.68	0.66	0.07	0.05	0.42	44
right	0.70	0.67	0.04	0.07	0.53	21
high mfc	0.68	0.69	0.07	0.05	0.06	325
low mfc	0.70	0.64	0.04	0.07	1.13	11
iv						
left	0.33	0.34	0.04	0.04	0.17	95
right	0.33	0.33	0.04	0.04	0.02	864
high mfc	0.33	0.32	0.04	0.03	0.09	197
low mfc	0.33	0.34	0.04	0.04	0.26	59
nsi 1st 3rd						
left	0.67	0.60	0.09	0.08	0.81	20
right	0.69	0.74	0.10	0.17	0.34	33
high mfc	0.67	0.69	0.08	0.15	0.22	50
low mfc	0.70	0.65	0.10	0.15	0.38	34
nsi 2nd 3rd						
left	0.56	0.61	0.06	0.10	0.63	19
right	0.59	0.68	0.05	0.23	0.61	9
high mfc	0.56	0.65	0.07	0.16	0.73	12
low mfc	0.58	0.63	0.05	0.20	0.42	14
nsi 3rd 3rd						
left	0.58	0.61	0.12	0.06	0.39	54
right	0.60	0.72	0.10	0.19	0.78	14
high mfc	0.58	0.64	0.13	0.15	0.40	36
low mfc	0.60	0.70	0.10	0.15	0.78	16
iv 1st 3rd						
left	0.32	0.34	0.05	0.06	0.24	58
right	0.34	0.40	0.08	0.17	0.48	21
high mfc	0.33	0.39	0.05	0.15	0.64	11
low mfc	0.34	0.35	0.08	0.10	0.10	140
iv 2nd 3rd						
left	0.32	0.32	0.04	0.03	0.25	69
right	0.31	0.44	0.03	0.22	1.01	4
high mfc	0.32	0.35	0.04	0.14	0.40	18
low mfc	0.32	0.40	0.03	0.19	0.72	6
iv 3rd 3rd						
left	0.33	0.34	0.06	0.08	0.01	1061
right	0.32	0.43	0.05	0.15	1.10	7
high mfc	0.33	0.35	0.06	0.14	0.26	39
low mfc	0.33	0.41	0.04	0.11	1.09	8

Table F3. Within group intra- and inter- limb differences.

	intra MEAN	inter MEAN	intra SD	inter SD	effect size	N for power of 0.8
d1/d2						
young	0.64	0.86	0.08	0.14	1.93	8
elderly	0.66	0.91	0.10	0.10	2.44	6
d1						
young	0.22	0.92	0.07	0.06	10.81	1
elderly	0.23	0.94	0.05	0.07	11.76	1
d2						
young	0.35	1.09	0.13	0.11	6.20	3
elderly	0.35	1.04	0.10	0.09	7.44	2
nsi						
young	0.69	0.62	0.06	0.07	1.05	15
elderly	0.66	0.60	0.06	0.05	1.15	14
iv						
young	0.33	0.35	0.04	0.03	0.59	26
elderly	0.33	0.36	0.04	0.02	0.70	22

Table F4. Between group differences for inter- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
d1/d2	0.86	0.91	0.14	0.10	0.43	42
d1	0.92	0.94	0.06	0.07	0.24	64
d2	1.09	1.04	0.11	0.09	0.53	33
nsi	0.62	0.60	0.07	0.05	0.37	52
iv	0.35	0.36	0.03	0.02	0.18	93

Table F5. Between group differences for intra- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
mean						
left	1.44	1.30	0.36	0.33	0.41	39
right	1.09	1.16	0.32	0.47	0.18	71
high mfc	1.44	1.50	0.35	0.21	0.20	95
low mfc	1.08	0.95	0.32	0.35	0.39	38
sd						
left	0.28	0.27	0.08	0.06	0.06	318
right	0.29	0.31	0.11	0.08	0.11	159
high mfc	0.30	0.31	0.11	0.07	0.07	257
low mfc	0.27	0.27	0.07	0.07	0.01	3049
minmfc						
left	0.66	0.54	0.26	0.24	0.47	34
right	0.30	0.37	0.21	0.35	0.23	52
high mfc	0.63	0.60	0.29	0.27	0.12	135
low mfc	0.33	0.31	0.21	0.28	0.09	160
maxmfc						
left	2.91	3.02	0.67	0.93	0.14	94
right	2.59	2.93	0.70	0.70	0.49	32
high mfc	3.00	3.15	0.74	0.53	0.24	77
low mfc	2.50	2.80	0.55	1.00	0.39	28
range						
left	2.25	2.48	0.65	0.77	0.32	44
right	2.29	2.57	0.68	0.60	0.44	38
high mfc	2.37	2.55	0.81	0.54	0.28	68
low mfc	2.17	2.49	0.45	0.82	0.51	22
cv						
left	19.84	21.72	5.88	4.21	0.37	49
right	27.73	28.77	7.56	8.54	0.13	114
high mfc	21.44	20.60	8.78	5.19	0.12	165
low mfc	26.13	29.89	6.08	6.55	0.59	25
skew						
left	0.56	0.55	0.33	0.28	0.03	537
right	0.58	0.69	0.33	0.42	0.28	50
high mfc	0.54	0.49	0.32	0.32	0.18	87
low mfc	0.60	0.75	0.34	0.35	0.44	35
kurt						
left	1.32	1.75	0.76	1.42	0.39	28
right	1.39	2.65	1.66	3.03	0.54	21
high mfc	1.25	1.85	0.73	1.67	0.50	19
low mfc	1.46	2.55	1.66	2.93	0.47	24
dfa						
left	0.77	0.76	0.05	0.08	0.08	150
right	0.80	0.77	0.04	0.08	0.44	23
high mfc	0.76	0.81	0.05	0.08	0.66	20
low mfc	0.80	0.73	0.03	0.06	1.61	6
d1/d2						
left	0.65	0.67	0.10	0.08	0.17	99
right	0.62	0.65	0.07	0.12	0.31	37
high mfc	0.64	0.63	0.11	0.09	0.13	131
low mfc	0.64	0.69	0.06	0.10	0.72	15

Table F6. Between group differences for intra- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
d1						
left	0.22	0.22	0.06	0.05	0.07	240
right	0.22	0.24	0.07	0.05	0.21	89
high mfc	0.23	0.23	0.08	0.05	0.10	177
low mfc	0.21	0.22	0.05	0.05	0.19	84
d2						
left	0.34	0.33	0.10	0.07	0.14	135
right	0.37	0.37	0.15	0.12	0.06	275
high mfc	0.37	0.38	0.16	0.10	0.04	491
low mfc	0.33	0.33	0.09	0.10	0.08	185
nsi						
left	0.68	0.66	0.07	0.05	0.42	44
right	0.70	0.67	0.04	0.07	0.53	21
high mfc	0.68	0.69	0.07	0.05	0.06	325
low mfc	0.70	0.64	0.04	0.07	1.13	11
iv						
left	0.33	0.34	0.04	0.04	0.17	95
right	0.33	0.33	0.04	0.04	0.02	864
high mfc	0.33	0.32	0.04	0.03	0.09	197
low mfc	0.33	0.34	0.04	0.04	0.26	59
nsi 1st 3rd						
left	0.67	0.60	0.09	0.08	0.81	20
right	0.69	0.74	0.10	0.17	0.34	33
high mfc	0.67	0.69	0.08	0.15	0.22	50
low mfc	0.70	0.65	0.10	0.15	0.38	34
nsi 2nd 3rd						
left	0.56	0.61	0.06	0.10	0.63	19
right	0.59	0.68	0.05	0.23	0.61	9
high mfc	0.56	0.65	0.07	0.16	0.73	12
low mfc	0.58	0.63	0.05	0.20	0.42	14
nsi 3rd 3rd						
left	0.58	0.61	0.12	0.06	0.39	54
right	0.60	0.72	0.10	0.19	0.78	14
high mfc	0.58	0.64	0.13	0.15	0.40	36
low mfc	0.60	0.70	0.10	0.15	0.78	16
iv 1st 3rd						
left	0.32	0.34	0.05	0.06	0.24	58
right	0.34	0.40	0.08	0.17	0.48	21
high mfc	0.33	0.39	0.05	0.15	0.64	11
low mfc	0.34	0.35	0.08	0.10	0.10	140
iv 2nd 3rd						
left	0.32	0.32	0.04	0.03	0.25	69
right	0.31	0.44	0.03	0.22	1.01	4
high mfc	0.32	0.35	0.04	0.14	0.40	18
low mfc	0.32	0.40	0.03	0.19	0.72	6
iv 3rd 3rd						
left	0.33	0.34	0.06	0.08	0.01	1061
right	0.32	0.43	0.05	0.15	1.10	7
high mfc	0.33	0.35	0.06	0.14	0.26	39
low mfc	0.33	0.41	0.04	0.11	1.09	8

Table F7. Within group intra- and inter- limb differences.

	intra MEAN	inter MEAN	intra SD	inter SD	effect size	N for power of 0.8
d1/d2						
young	0.64	0.86	0.08	0.14	1.93	8
elderly	0.66	0.91	0.10	0.10	2.44	6
d1						
young	0.22	0.92	0.07	0.06	10.81	1
elderly	0.23	0.94	0.05	0.07	11.76	1
d2						
young	0.35	1.09	0.13	0.11	6.20	3
elderly	0.35	1.04	0.10	0.09	7.44	2
nsi						
young	0.69	0.62	0.06	0.07	1.05	15
elderly	0.66	0.60	0.06	0.05	1.15	14
iv						
young	0.33	0.35	0.04	0.03	0.59	26
elderly	0.33	0.36	0.04	0.02	0.70	22

Table F8. Between group differences for inter- limb parameters.

	Young MEAN	Elderly MEAN	young SD	elderly SD	effect size	N for power of 0.8
d1/d2	0.86	0.91	0.14	0.10	0.43	42
d1	0.92	0.94	0.06	0.07	0.24	64
d2	1.09	1.04	0.11	0.09	0.53	33
nsi	0.62	0.60	0.07	0.05	0.37	52
iv	0.35	0.36	0.03	0.02	0.18	93

Table G1. Symmetry index values per subject for descriptive and dynamic parameters.

Subject	MFC	SD	CV	S	K	α	μ	D1	D2	NSI	IV
Y1	-62.92	-2.06	61.15	92.34	9.84	18.67	-12.35	-6.45	4.26	8.96	-6.06
Y2	-59.37	-40.74	19.74	78.78	87.89	0.00	0.76	-38.46	-41.03	-1.48	8.22
Y3	-8.63	11.63	20.15	21.72	-66.20	10.26	-5.82	9.09	12.90	4.80	-5.71
Y4	0.27	34.48	34.28	2.31	46.98	1.18	-14.15	26.09	39.34	2.78	3.17
Y5	-27.85	37.58	63.69	-3.19	-161.05	5.33	-37.55	13.33	50.67	24.82	-31.25
Y6	1.56	16.29	14.82	-24.88	-82.39	-5.56	-5.88	13.33	20.51	0.00	-15.87
Y7	-5.00	4.42	9.39	-4.12	-56.09	0.00	6.41	10.91	0.00	-1.44	0.00
Y8	-53.10	9.16	61.50	133.33	-81.55	8.92	-10.98	0.00	13.95	7.63	-3.08
Y9	-69.43	-2.80	66.98	15.40	66.16	3.82	1.99	0.00	-3.92	-1.50	15.79
Y10	-8.27	-30.40	-22.36	-129.50	-35.59	2.56	37.21	-6.06	-44.78	-12.66	26.42
E1	-38.47	18.84	56.52	1.02	-81.07	-2.74	-5.43	16.67	21.69	5.63	-21.54
E2	1.20	-12.27	-13.41	-149.09	76.24	12.82	-6.10	-15.38	-10.26	7.52	-14.08
E3	-62.49	-41.64	22.27	29.47	76.08	-4.44	12.39	-33.96	-47.62	-12.39	7.59
E4	-79.77	10.22	87.94	117.38	47.90	2.47	-14.41	0.00	16.00	10.37	0.00
E5	86.90	38.42	-52.79	8.72	-49.54	21.94	-10.38	30.00	42.62	10.69	-3.08
E6	40.64	58.35	18.58	-13.78	-140.00	11.61	-33.65	38.46	69.77	18.98	-15.38
E7	-31.01	-21.19	10.28	39.39	89.53	-36.48	32.94	0.00	-34.48	-23.88	19.35
E8	-39.53	26.32	64.04	67.25	80.33	7.59	-0.78	27.45	27.16	-1.46	5.88