Understanding Gait Control Dynamics: Ageing Effects on Falling Risks

Hanatsu Nagano

Abstract

Background

Due to the ongoing trend of an ageing population in many developed countries, falls among older adults during walking are emerging as an important social healthcare issue due to high injury rates and associated medical costs. Unsuccessful recovery from balance loss leads to falls, and four critical swing phase gait events are toe-off, heel contact, minimum foot clearance (MFC) and minimum lateral margin (MLM). Dynamic balance at these four gait events were examined to characterise biomechanical evidence in older adults' walking patterns to understand the frequent occurrence of falls.

Falls prevention strategies need to be effective and practical in addition to cost advantages and ease of engagement, if a measure is to be adopted as a long-term habitual intervention. This project also investigated whether a shoe insole is effective in reducing falls risks. As insoles can be applied to any shoes at lower cost, insole interventions could possibly prove an ideal approach to falls prevention during walking.

Study Designs and Aims

Gait data were collected from 26 older adults and 30 young controls, sampled by an Optotrak® 3-D system (100Hz) with two AMTI force plates recording ground reaction forces and centre of pressure (CoP) (1000Hz). The current research increased and decreased step width (±50% relative to preferred width) as experimental conditions.

The first aim was to examine ageing effects on balance control at the four events. Based on the Safety Zone Model, 'secured' balance is defined when the centre of mass (CoM) projects within the Safety Zone, the area defined by two feet positions in the transverse plane. Another aim was to investigate a shoe insole's dorsiflexion and eversion support effects on tripping and lateral balance loss prevention while reducing lower limb injuries including inversion sprain and knee joint osteoarthritis.

Findings and Implications

Step width manipulation only affected the older group. A shorter medio-lateral Safety Zone was found in narrow walking. Older adults walked more slowly with reduced step length to preserve balance. Wide walking showed functional benefits in a greater medio-lateral Safety

Zone, but walking velocity and step length were still lower than preferred walking. The shorter anterior-posterior Safety Zone due to slower gait was compensated by the non-dominant lead foot toeing-in, which extended the anterior margin when older adults walked with controlled widths. Age-specific asymmetry in toeing-in of the non-dominant foot, however, reduced the Safety Zone when the non-dominant foot was trailing. Under width controlled conditions, less variable step width and medio-lateral Safety Zone were observed but older adults also showed more variable medio-lateral CoM movement, non-dominant MLM and the foot's CoP control, identified as further evidence of age-related impairment in medio-lateral balance.

The insole's dorsiflexion support elevated swing foot clearance at MFC and eversion reduced lateral CoP displacement and increased the safety lateral margin at MLM, suggesting successful reduction in balance loss risks. Both dorsiflexion and eversion enhanced the efficient use of mechanical energy during loading following heel contact. The tested insole's features can potentially be incorporated into shoe-insole designs to enhance safer walking for older adults.

Student Declaration

I, Hanatsu Nagano, declare that the PhD Thesis entitled "**Understanding Gait Control Dynamics: Ageing Effects on Falling Risks**" is no more than 100,000 words in length, including quotes, and exclusive of tables, figures and appendices, bibliography, references and footnotes. This thesis contains no material that has been submitted previously, in whole or in part, for the award of any other academic degree or diploma. All work in this thesis was conducted by me.

Hanatsu Nagano

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Equations

$CoF = \frac{F(fri)}{F(N)}$ 1

$RCoF = \frac{GRFhor}{GRFver}$ 2

$\mathrm{T} = \mathrm{F}_{(\mathrm{f})} \times \sin\theta \times \mathrm{d} = -\mathrm{F}_{(\mathrm{C})} \times \sin\theta \times \mathrm{d}$ 3

$\mathrm{F}_{(\mathrm{C})} = \frac{\mathrm{T}}{\mathrm{M}_{(\mathrm{C})} \times \sin\theta \times \mathrm{d}}$ 4

$\mathrm{A}_{(\mathrm{C})} = \frac{-\mathrm{F}_{(\mathrm{f})}}{\mathrm{M}_{(\mathrm{C})}}$ 5

$\mathrm{F(f)} = GRFver \times CoF - GRFhor$ 6

$\mathrm{A(c)} = \frac{\mathrm{F}_{(\mathrm{f})}}{\mathrm{M}_{(\mathrm{c})}}$ 7

$\mathrm{A(f)} = \frac{\mathrm{F}_{(\mathrm{f})}}{\mathrm{M}_{(\mathrm{f})}}$ 8

$\frac{t_y - C_{y\prime}}{t_x - C_{x\prime}} = \frac{C_{y\prime} - st_{y\prime}}{C_{x\prime} - st_{x\prime}} = \frac{t_y - st_{y\prime}}{t_x - st_{x\prime}}$ 9

CoM $(\mathrm{C}_{x'}, \mathrm{C}_{y'}) = (V_{(cx)} \times \mathrm{t} + 0.5 \times A_{(cx)} \times \mathrm{t}^2, V_{(cy)} \times \mathrm{t} + 0.5 \times A_{(cy)} \times \mathrm{t}^2)$ 10

Slipping foot $(\mathrm{st}_{x'}, \mathrm{st}_{y'}) = (st_x + 0.5 \times A_{(fx)} \times \mathrm{t}^2, st_y + 0.5 \times A_{(fy)} \times \mathrm{t}^2)$... 11

$t = \frac{-b \pm \sqrt{b^2 - 4ac}}{2a}$ 12

$F_{(f)} = GRFver \times CoF - GRFhor$ 13

$A_{(c)} = \frac{F_{(f)}}{M_{(c)}}$ 14

$A_{(f)} = \frac{F_{(f)}}{M_{(f)}}$ 15

$\frac{h_y - C_{y\prime}}{h_x - C_{x\prime}} = \frac{C_{y\prime} - sh_{y\prime}}{C_{x\prime} - sh_{x\prime}} = \frac{h_y - sh_{y\prime}}{h_x - sh_{x\prime}}$ 16

$(\mathrm{C}_{x'}, \mathrm{C}_{y'}) = (\mathrm{V}_{(\mathrm{cx})} \times \mathrm{t} + 0.5 \times \mathrm{A}_{(\mathrm{cx})} \times \mathrm{t}^2, \mathrm{V}_{(\mathrm{cy})} \times \mathrm{t} + 0.5 \times \mathrm{A}_{(\mathrm{cy})} \times \mathrm{t}^2)$ 17

$(sh_{x'}, sh_{y'}) = (sh_x + 0.5 \times A_{(fx)} \times t^2,\ sh_y + \sqrt{{V_{(fx)}}^2 + {V_{(fy)}}^2} \times t + 0.5 \times A_{(fy)} \times t^2)$ 18

$\sqrt{(sh_{x\prime} - sh_x)^2 + (sh_{y\prime} - sh_y)^2} = \sqrt{{V_{(fx)}}^2 + {V_{(fy)}}^2} \times t + 0.5 \times \sqrt{{A_{(fx)}}^2 + {A_{(fy)}}^2} \times t^2$ 19

$F_{(f)} = A_{(f)} \times M_{(f)}$ …………………………………………………………………… 20

$A_{(c)} = \frac{A_{(f)} \times M_{(f)}}{M_{(c)}}$ …………………………………………………………… 21

$X = \left[R_y - \frac{R_x(R_y - L_y)}{R_x - L_x}\right] / \left[\frac{V_{(cy)}}{V_{(cz)}} - \frac{R_y - L_y}{R_x - L_x}\right]$ ……………………………………………… 22

$Y = \frac{V_{(cy)}}{V_{(cz)}} \times \left[R_y - \frac{R_x(R_y - L_y)}{R_x - L_x}\right] / \left[\frac{V_{(cy)}}{V_{(cz)}} - \frac{R_y - L_y}{R_x - L_x}\right]$ ……………………………… 23

$D = \sqrt{\boldsymbol{X^2} + \boldsymbol{Y^2}}$ …………………………………………………………… 24

ART (t); $t = \frac{-V(c) + \sqrt{V(c)^2 + 2 \times A(c) \times D}}{A(c)}$ …………………………………………………… 25

$t = \frac{(ty - Cy)}{V(cy)}$ …………………………………………………………………… 26

$(tx - Cx) = V(cx) \times t$ …………………………………………………………… 27

$C(vx) = 0.5 \times \frac{F(l)}{M(c)} \times t^2$ ………………………………………………… 28

θ = atan [(ty-Cy)/(tx-Cx)] ………………………………………………………… 29

Abbreviations and Symbols

AP	anterior-posterior
ART	available response time
BoS	base of support
CoF	coefficient of friction
CoM	centre of mass
CoP	centre of pressure
FoF	fear of falling
GRF	ground reaction force
HC	heel contact
HO	healthy older
HY	healthy young
IP	inverted pendulum
IQR	interquartile range
RCoF	required coefficient of friction
MFC	minimum foot clearance
MLM	minimum lateral margin
ML	medio-lateral
MTP	minimum toe point
SD	standard deviation
TO	toe-off

Chapter 1

Introduction

Independent mobility is essential to a healthy and rewarding lifestyle because it provides the locomotor abilities for our everyday work, recreation, and home based activities. One cost, however, of the physical and psychological health benefits of mobility is the risk of sustaining a fall. Falls become an increasingly significant health risk with ageing due to declines in the neuromuscular control of walking that compromise stability and, as a consequence, increase the risk of falling. The increased incidence of falls in older adults and the associated mortality, morbidity, and disability are a major concern for the Australian healthcare system (Moller 2003). In one report falls-related expenditure in Australia was estimated to be $3billion per annum, possibly higher than for any other category of injury, including motor vehicle accidents (Commonwealth Department of Health and Aged Care, 2001; Cassell and Clapperton, 2008; SRDC 1999). As a consequence, there is in progress a worldwide research effort to better understand the causes of falls and, subsequently, devise interventions to prevent them. Consistent with this broad initiative this Thesis had two principal aims. The first was to understand how ageing changes the biomechanical characteristics of balance during walking that may cause falls. The second aim was to investigate methods of falls prevention and in this project the protective effects of a shoe insole were determined experimentally.

Categorisation of falls based on their direct cause has significantly advanced the application of gait biomechanics to the problem of falls prevention. Biomechanical studies have provided clear evidence that increased falls risks in older adults is due to changes in walking mechanics that cause balance loss. Balance loss is the primary direct cause of falls, accounting for up to 87% of all incidents (Blake et al., 1988; Berg et al., 1997; Prince et al., 1997). Balance-loss induced destabilisations occur primarily in two of the three principal axes of motion, i.e., anterior-posterior and medial-lateral. Approximately 53% of falls are due to tripping-related forward balance loss, the leading cause. About 25% of falls are associated with backward balance loss due to slipping (Blake et al., 1988; Prince et al., 1997; Smeesters et al., 2001). Slow walking, fainting, unexpected stepping down and slipping increase the risk of lateral balance loss (Smeesters et al., 2001). In addition to mechanical disturbances to the centre of mass (CoM) there are other causes of balance disturbance not associated directly with lower-limb trajectory control, such as neurological disorders, visual impairments, auditory degeneration, medications and dizziness, all of which can negatively affect gait stability (Lord and Daybew, 2001; Nicklason et al., 2002).

Previous research has documented ageing-related changes to walking that are proposed to be adaptive in securing balance. Older adults for example, walk with shorter and wider steps, slower velocity and prolonged double support time (Whittle, 2007; Winter, 1991). While balance control in ageing has been documented there remain two important limitations to previous research that form the focus of the project described below. First, while medio-lateral (ML) stability has been studied there are relatively few accounts of stability loss in the anterior-posterior (AP) direction. A more complete stability model should, therefore, encompass balance loss in both primary axes of motion, ML and AP, therefore transverse plane. Tripping and slipping, for example, would be expected to induce greater AP stability loss than ML.

The second feature of stability modelling investigated in the present experiment was the capacity to maintain stability during the swing phase of the gait cycle when only one foot maintains ground contact. Previous studies have revealed that the capacity to control the CoM within the base of support (BoS) reduces with ageing, especially for those with history of falls (Åberg et al., 2010; Lugade et al., 2011). Traditionally the effective BoS during walking has been defined during double support, when step width can be easily measured as the distance between the feet. When one foot is off the ground during the swing phase, however, the CoM position relative to the BoS is less easily modelled because the supporting base formed by feet changes shape and extent as the swing phase progresses.

Walking has been described by Drury and Wolley (1995) and others (Brooks, 1986; Soderberg, 1986) as a …"series of controlled falls requiring a repeated pattern of losing and regaining a balanced stance" (p. 714). The process of maintaining balance, i.e., "controlling falls" is, however, inextricably linked to the second key objective of gait, progression. Walking can, therefore, be characterised as essentially "…progression of the body in the desired direction and not falling down" (Drury and Wooley, 1995; Das and McCollum, 1988). This concept is extended further in this project by demonstrating how the temporal and spatial dimensions of foot positioning, more commonly associated with the primary task of progression, can equally be viewed as serving to preserve stability by maintaining the body's CoM within the supporting base formed by the feet. Thus, the stride phase variables of gait biomechanics may be controlled equally by the motor system to preserve balance.

While balance has, therefore, traditionally been considered essential to understanding functional, adaptive human gait control a broad range of biomechanical variables and associated concepts have been used to define “balance”. In this thesis a comprehensive, unified, model has been developed that demonstrates how *all* spatio-temporal gait parameters influence the capacity to maintain balance; the Safety Zone Model (hereafter also referred to as “The Model”). As implied by Drury and Wolley (1995) the fundamental proposition underlying The Model is that balance depends on step width variability that narrows or lengthens the base of support in combination with step velocity that increases or decreases the time available in which to redirect the whole body centre of mass. “Balance” is, therefore, proposed as the primary concept in understanding adaptive locomotor control.

The first research question addressed in this thesis was how ageing affects balance control during walking. To answer this question, the Safety Zone model was employed incorporating the two key features designed to overcome the limitations of previous research outlined above. Specifically the Model: (i) accommodates loss of stability in two of the principal axes of motion, AP and ML and; (ii) incorporates a stability boundary during the vulnerable swing phase, when the base of support is changing continuously and balance loss is more likely than during the relatively stable double support phase. By incorporating these features the effects on balance of tripping and slipping to the CoM were biomechanically modelled by measuring experimentally the AP and ML CoM displacement relative to a “safety zone” base of support formed by the feet at four key swing phase events (i.e. toe-off, minimum foot clearance, minimum lateral margin and heel contact).

Increased step width has been acknowledged as a fundamental ageing response that assists in maintaining ML balance (Whittle, 2007). A greater ML BoS area allows a wider range of the CoM movement and, accordingly, can be considered to reduce the risk of lateral balance loss (Åberg et al., 2010). It was, therefore, considered important to observe how older adults change their Safety Zone characteristics and associated dynamic balance when balance is disturbed. The current research employed step width control as an experimental condition to impose destabilizing walking conditions to examine how ageing-related gait adaptations emerge. Relative to baseline data for preferred walking, conditions in which participants were requested to walk with narrow and wide steps (±50%) were included as experimental conditions.

As discussed briefly there is a pressing need for effective and practical falls prevention interventions. Critically important in falls prevention are cost and ease of engagement, if a measure is to be adopted as a long-term habitual intervention (e.g., Yardley et al., 2008). Walking, for example, is proposed as an ideal exercise intervention for older adults due to its adequate intensity, low cost and generally easy engagement. One limitation to walking for exercise is, however, a higher risk of falling when walking in a more obstructed environment comprising irregular and varied ground surfaces. Shoe designs have previously been investigated (e.g. Menant et al., 2008) as an important safety consideration for older adults because they provide the coupling or interface between the locomotor system and the highly varied support surfaces upon which we move. This project also addresses the question of whether a shoe insole is effective in improving the balance of older adults by reducing tripping and lateral balance loss. Effects of improved shoe design are only of benefit when specifically designed shoes are worn, while insoles can be applied to any shoes at lower cost. Insole interventions could possibly prove an ideal approach to reducing balance loss during walking.

A second research question was to whether a custom designed supporting insole could control older adults' walking to improve stability and reduce the risk of injuries related to walking (i.e. inversion sprain and knee osteoarthritis). The insole tested in the current study was designed to change two ankle motions: dorsiflexion and eversion. To avoid tripping over obstacles increased foot-ground clearance, especially at the minimum clearance point (MFC) is essential and dorsiflexion is known to elevate the swing foot (Moosabhoy and Gard, 2006). Increased eversion was expected to regulate excessive lateral deviation of centre of pressure (CoP) trajectory and therefore reduce lateral balance loss and the risk of inversion sprain. In addition, both these ankle motions could reduce impact at heel contact and assist a more energy efficient loading response following heel contact to relieve stress on knees following heel contact (see section 4.3.1). If shown to be effective in improving gait stability the ankle supporting functions of shoe-insoles have the potential to become an innovative approach to fostering safer walking to support older adults' independent lifestyles and contribute to reducing medical costs.

1.1 Organisation of the Thesis

The thesis structure is as follows. Chapter 2 concerns the epidemiology of falls in the older population, it provides key terms and definitions of gait analysis and reviews ageing effects on gait, in particular, falls risk. A Safety Zone Model is developed in Chapter 3 to characterise dynamic balance. In addition, balance loss simulation methods are separately developed for the four examined events, including toe-off, minimum foot clearance (MFC), minimum lateral margin (MLM) and heel contact. With respect to the insole intervention, previous footwear research and the specification of the tested insole is provided in Chapter 4. Specific research questions and expected outcomes are summarised in Chapter 5. The following Chapter 6 explains the research methods. Results are reported in Chapters 7 to 10. Each Result chapter begins with a brief review of the research question, followed by statistical analysis and a chapter summary. All interpretations of the Results chapters and an extended discussion with future research directions comprise the final chapter.

Chapter 2.

Literature Review

2.1 The Epidemiology of Ageing and Falls

In Australia, 1.5 million falls are estimated among older adults every year, and depending on the study, 9-20% of falls cause severe injuries, such as hip fractures (Stevens et al., 2006). Nationwide at least 100,000 severe injuries among older adults are due to falls, ranked as the leading cause of non-fatal injury (27%) and also the third leading direct cause of injury-related death (Watson and Ozanne-Smith, 2000). The annual medical costs of falls is estimated at close to $3 billion and on an individual basis, the average cost per falls-related hospitalisation is reported to exceed $20,000 (Australian Government, 2007; Hill et al., 1999; Stalenhoef et al., 2002; National Injury Prevention Advisory Council, 1999). Medical costs are predicted to triple in the next 40 years due to the increasing proportion of older adults relative to the population as a whole from 13.5% now to 22% by 2050 (Commonwealth of Australia, 2010).

In addition to costs, falls have a negative impact on an individual's quality of life, for example, half of the hip fractures due to falling result in permanent loss of independent lifestyles and 20-30% of those eventually result in death (Cassell and Clapperton, 2008; Hung et al., 2012; Sherrington and Menz 2003). In preparation for a demographically ageing society, a reduction in falls should be a top priority. Successful establishment of any innovative intervention has the potential to create a positive impact on the falls problem worldwide. In Australia, every 1% reduction in falls could prevent 45,000 older adults from falling and save an associated $90 million every year (Australian Government, 2007; Hill et al., 1999; Stalenhoef et al., 2002; National injury prevention advisory council, 1999).

Current falls interventions include strength and balance exercises, medication management and installation of home safety devices. However, a recent Cochrane Collaboration review of falls interventions conducted on 111 randomized trials (55,303 participants) revealed moderate adherence difficulties with single intervention approaches such as exercise programs (Canadian Falls Prevention Curriculum, 2011; Yardley et al., 2008). Home safety interventions such as handrails were found not to reduce falls rates except in high falls risk populations, suggesting future interventions should also target intrinsic risk factors. According to previous studies (Berg et al., 1997; Sherrington and Menz, 2003; Koepsell et al., 2004; Li et al., 2006; Curry et al., 2003; WHO, 2007), about half of falls in older people occur while they are walking outdoors. Despite the risk of falling, maintaining mobility by walking is recommended for older adults. Further understanding of age-related effects on gait and balance is, therefore, essential for the assessment of ageing-related falls risk and developing effective prevention strategies. The conventions employed in previous gait research are summarised in the following section.

2.2 Terms and Definitions of Gait Cycle

The term ‘gait’ refers to ‘manner’ or ‘style’ of walking (Whittle, 2007). Although the word ‘walking’ is often used interchangeably, ‘gait’ is more suitable in the study of human locomotion. For analytical purpose, the minimal unit of investigation which contains all the characteristics of the normal gait cycle was first, essentially defined. One gait cycle is defined as ‘the time interval between one gait event and the next occurrence of the same gait event of the same limb’, and majority of the studies employ heel contact as the initiation and termination of one gait cycle (Begg et al., 2006; Rose et al., 2006; Whittle., 2007). Applying this common manner, an alternative definition of gait cycle can be ‘Heel contact to the next heel contact of the same foot’. Heel contact is defined as the instant when the heel of the foot or shoe makes initial contact with the ground (Winter, 1991). For abnormal or pathological gait where the heel may not be the first part to make contact with the ground, and as such the term initial contact, foot strike, and foot contact are used instead. Toe off is defined as the instant when the toe of the foot or shoe leaves the ground (Winter, 1991). The term ‘foot off’ is, in some gaits, preferred for the same reason. Heel contact and toe-off events determine two major gait phases i.e., stance and swing.

2.2.1 Axis and Joint Motions

Gait study is frequently based on the quantitative information measured in 3-dimensions. To express motions or forces in 3-dimensions, a common coordinate system is usually used in most of the gait analyses (Figure 2.2.1).

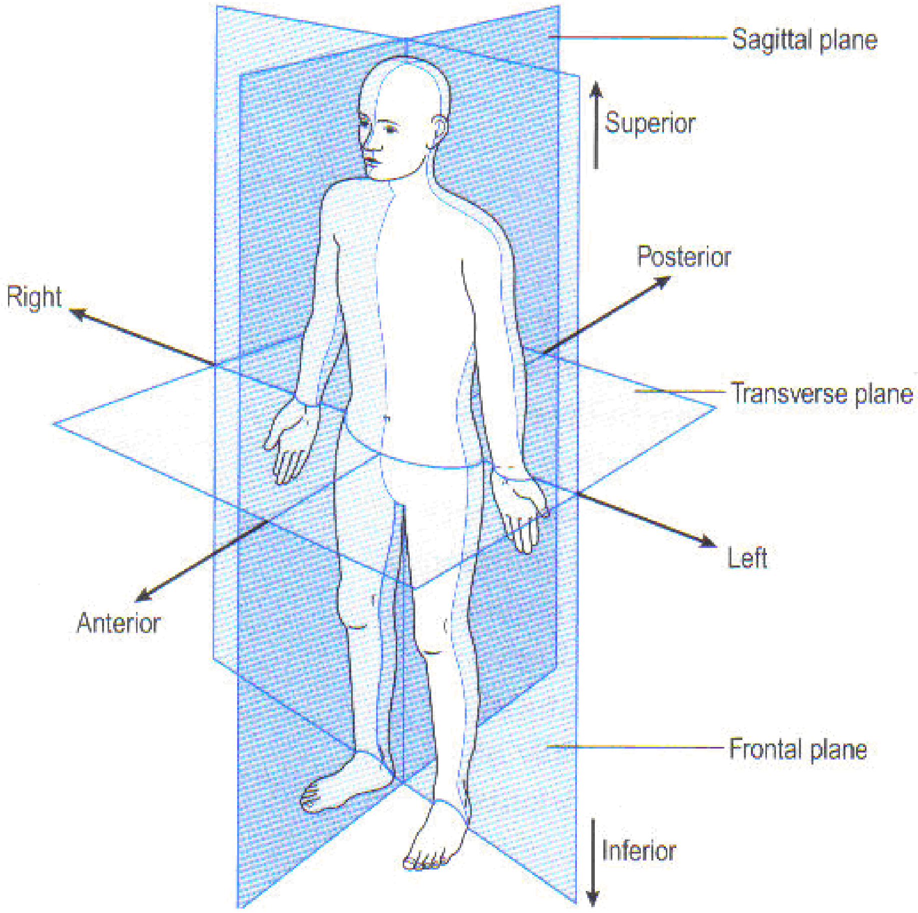

Figure 2.2.1 *Three dimension coordinate system. Forward-backward = anterior-posterior; right-left = medio-lateral (adapted from Whittle, 2007).*

2.2.2 Stance and Swing Phases

Stance phase is the time when the foot is in contact with the ground; heel contact (described as initial contact in Figure 2.2.2) to toe-off; whereas the swing phase is the time when the foot is off the ground and moving forward in the air; toe-off to the next heel contact (or initial contact) (Whittle., 2007). Rose et al. (2006) reported the proportion of the two phases for normal adults is 62% stance and 38% swing phase. Both phases have a number of subdivisions as illustrated in Figure 2.2.2.

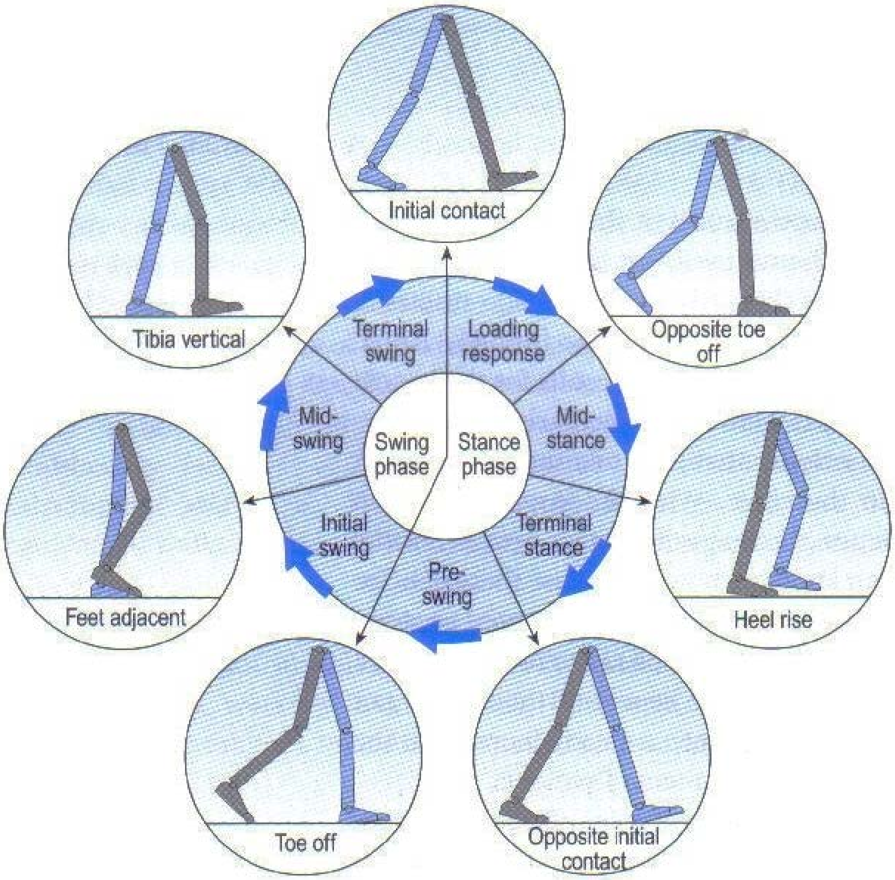

Figure 2.2.2 *Stance and swing phase of one gait cycle and their further sub-events relative to the stance limb. Heel contact is indicated as initial contact (adapted from Whittle, 2007).*

Stance foot generates ground reaction forces (GRF). Figure 2.2.3 compares the vertical GRF between the two older female populations classified by lower limb strength. Peak F1 and F3 are respectively associated with braking and push-off forces. While differences in lower limb strength between the two groups did not affect GRF during slower walking (A), clear differences are seen in faster walking (B). GRF has been therefore considered generally larger in the population with faster gait speed (Nilsson and Thorstensson, 1989). Both anterior-posterior (AP) and medio-lateral (ML) GRF components are also generally greater in fast and vigorous gait although less investigated compared to vertical GRF during normal walking.

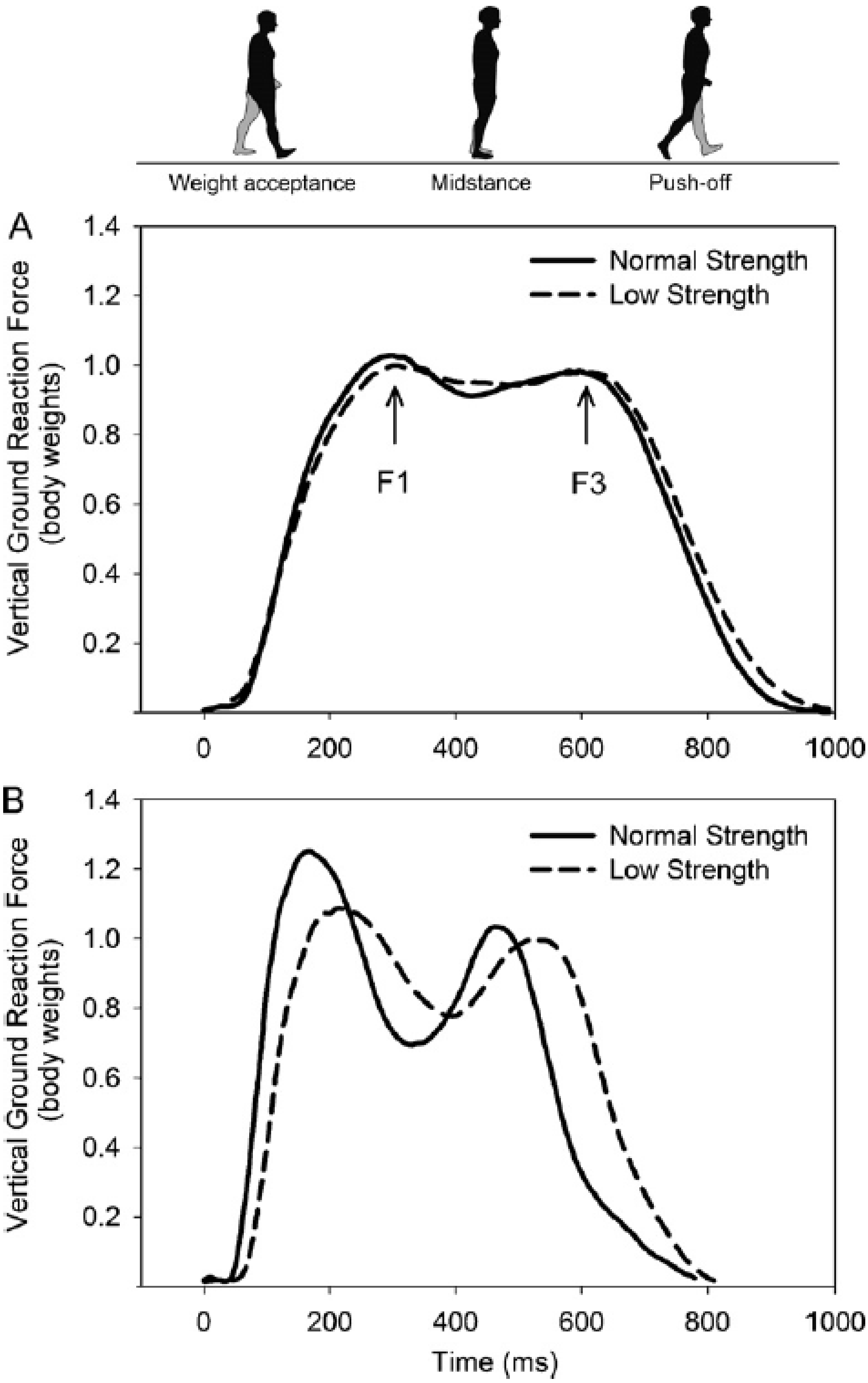

Figure 2.2.3 *Comparison of vertical ground reaction forces between (A) walking at 0.8m/s and (B) walking as fast as possible; Normal strength = older females with normal strength and low strength = older females with relatively lower strength. Values normalised to body weights. F1 = braking, F3 = push-off (adapted from LaRoche et al., 2011).*

While gait descriptions (Figure 2.2.2) and ground reaction forces (Figure 2.2.3) are relative to the stance limb, it is also possible to use swing phase events to divide the swing phase of gait cycle. Vertical toe-trajectory is for example, one method to define swing phase events.

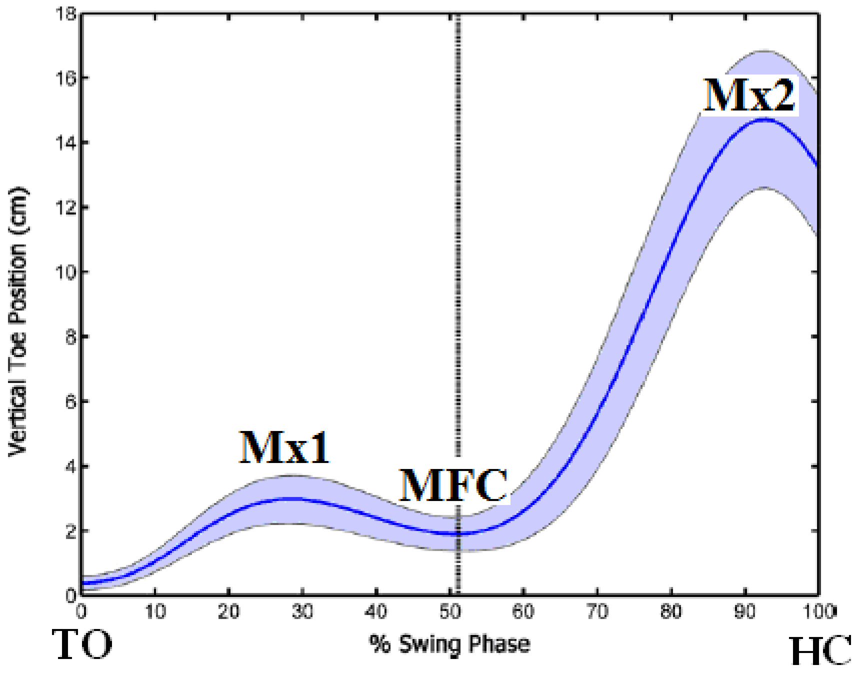

Figure 2.2.4 Swing phase foot trajectory events: Mx1, MFC and Mx2: TO = toe-off and HC = heel contact (adapted from Moosabhoy et al., 2006).

As visualised in Figure 2.2.4, , the first and the second (maximum) clearance has been described as Mx1 and Mx2, while the local minimum between the two peaks has been known as minimum foot clearance (MFC) where the risk of tripping has been considered to be frequent (Begg et al., 2007). As this example shows, falls need to be categorised into several different types to analyse gait features that increase the risk of particular types of falls and identification of specific gait events for each type of fall is necessary to advance gait biomechanics for an issue of falls among older adults.

2.3 Categorisation of Falls based on Gait Events and Balance Loss Directions

To progress our understanding of falls, it is essential to examine critical gait events for different falls categories to develop biomechanical models of falling. Up to 78% of falls, for example, are attributable to either tripping or slipping (Blake et al., 1988; Berg et al., 1997; Prince et al., 1997) and the mid-swing phase event, minimum foot clearance (MFC) and the terminating swing phase event, heel contact, have been investigated, respectively (section 2.7; Begg et al., 2007; Lockhart et al., 2007). Although it has been less well explored than MFC and heel contact characteristics, Myung (2003) reported that at the event prior to toe-off, there is a risk of forefoot posterior slipping that may lead to falls.

In addition to direct cause, such as tripping or slipping, fall direction can also be used to classify a fall. Almost all falls during walking can be attributed to balance loss, either anterior, posterior or lateral, except in unusual balance loss events such as a sudden collapse of the walking surface. Tripping at MFC and posterior forefoot slipping prior to toe-off mainly induce anterior balance loss while posterior falls are most frequently caused by slipping after heel contact (Myung, 2003; Smeesters et al., 2001). Thus, for further understanding of AP balance loss, critical gait events including toe-off, MFC and heel contact should be investigated. In relation to lateral falls, it is interesting that there have been few previous investigations (Lugade et al., 2011). As detailed in Section 2.5 and Chapter 3, falls direction is determined by the support base boundary (Hof et al., 2005).

The current research explores falls biomechanics from the perspective of dynamic balance at the critical gait events. Critical gait events in relation to lateral balance loss have not however, yet been clearly identified. This limitation has hindered research on lateral balance control and accordingly, the first objective is to detect those gait events leading to lateral instability. As suggested by Perry et al. (2008) and Åberg et al. (2010), minimum lateral margin (MLM) is the mid-swing event with the shortest ML distance between CoM and the stance foot (section 2.8), and accordingly, an association between MLM and lateral balance perturbation is predicted. The four key swing phase events examined in the current study were toe-off, MFC, MLM and heel contact. Biomechanical characteristics of these events were first characterised for ageing and step width effects but independent of body balance, but later, re-examined by incorporating dynamic balance assessment method using the Safety Zone. Preceding the introduction of previous findings about the four critical swing phase events and the rationales for balance loss description by the Safety Zone Model provided, it is meaningful to review general causes of gait changes and balance impairment due to ageing.

2.4 Ageing Effects on Gait

Age-related changes to gait and balance can be attributed to deterioration in sensorimotor function (Menz et al., 2007; Perry et al., 2008). These functional declines are considered the primary intrinsic cause of falls in older adults (Richardson et al., 2003). Ageing induces a systematic and progressive decline in foot sensitivity (Perry, 2006), lower limb muscle strength and power (Puthoff and Nielsen, 2007), lower limb joint flexibility (Kang and Dingwell 2008), and muscle activation (Solnik et al., 2010). Strength in the quadriceps, hamstrings, dorsiflexors and plantarflexors and reduced joint range of motion (RoM) have been identified due to decreased maximum flexion and extension of each lower limb joint (Perry et al., 2007). Motor control also declines with ageing and negatively affects reaction speed, precision of end-point control, (e.g. the foot) and movement variability (Hausdorff, 2005). Muscle activation during walking is modified in older adults, as reflected in decreased activation of some muscles and increased co-activation of others (Hortobagyi et al., 2009). For example, deficits in dorsiflexor activation have been observed in older adults and linked to changes to the foot's trajectory during the swing phase that may increase the risk of tripping (Morse et al., 2004). With ageing, afferent sensory signals from the feet are also attenuated due to reduced plantar fat pad thickness, flattening of the longitudinal arch and the development of claw and hammer toes (Burnfield et al., 2004). Reduced foot sensitivity has, furthermore, been associated with an increased risk of falling due to balance loss (Zehr and Stein, 1999). Most important is that the sensorimotor changes with ageing identified above influence the walking pattern and impair dynamic balance.

Reduced walking speed is the most well documented ageing effect, due to shorter step length with prolonged double support time (Hollman et al., 2007; 2011; Macellari et al., 1999; Whittle, 2007; Winter, 1991). Another key adaptation in spatio-temporal gait parameters is that older adults tend to adopt wider steps to preserve balance possibly with additional mechanical energy costs (Hurt et al., 2010; Ko et al., 2007; Pandy et al., 2010). It is, however, important to note that falls are more frequent in "older" and frailer individuals who usually have increased step width, suggesting that age-related balance impairments cannot be fully compensated by larger step width (Åberg et al., 2010; Wezenberg et al., 2011). In the present experiment the gait task was manipulated by changing step width, both narrower and wider than preferred, to challenge balance and investigate how with ageing, gait is adapted to maintain stability.

2.5 Gait Adaptations to Challenging Walking Conditions among Older Adults

In challenging walking conditions, not only do physiological and neurological factors increase the difficulty of processing unfamiliar sensory inputs, but older adults' gait has also been described as "cautious" or "fearful" (Prince, 1997; Wass et al., 2005). The usual term to describe age-specific effects on gait adaptations due to psychological preparatory responses against potential hazards is 'fear of falling' (Bock and Beurskeus 2010; Dunlap et al., 2012; Hollman et al., 2007; Ko et al., 2007; Nagano et al., 2012; Nordin et al., 2010; Schrager et al., 2008). Fear of falling has been identified as a primary factor in inducing cautious gait adaptations, such as shorter step length and associated reduction in gait velocity, leading to impaired balance (Menz et al., 2007). Multiple factors, including self-awareness of decline in physical condition such as lower limb muscular strength, slower reaction speed, visual or auditory deterioration and poor balance, in addition to history of falls or activity restrictions may trigger caution-based gait adaptations (Donoghue et al., 2012; Scheffer et al., 2008; Vellas et al., 1997; Zjistra et al., 2007). Onset of fear of falling tends to initiate a cycle of negative 'chain reactions', for example, regulation of outdoor activities leading to loss of sociality, depression, accelerating physical decline and further escalation of fear of falling.

Gait patterns of individuals in a challenging walking environment include slower gait velocity due to a reduction in step length, prolonged double support time and increased step width, all of which are typical age-related effects (Donoghue et al., 2012; Menz et al., 2007). In previous gait studies age-related gait adaptations to spatio-temporal parameters were minimal in unobstructed walking, but under more challenging conditions, however, such as dual-task locomotion, treadmill walking, or walking on a narrow or elevated surface, ageing effects were accentuated (Bock and Beurskeus 2010; Dunlap et al., 2012; Hollman et al., 2007; Ko et al., 2007; Lythgo et al., 2007; Nagano et al., 2012; Nordin et al., 2010; Schrager et al., 2008). Fear of falling in older adults on a particular walking surface seems to be reflected in changes to spatio-temporal parameters. In the current study incorporating decreased and increased step widths as experimental conditions, assessment of spatio-temporal parameters was undertaken to reflect perceived challenges by older adults. As the following section illustrates, a purpose of caution-based gait adaptations seen in older adults can be hypothesised to ensure the centre of mass (CoM) location within safety zone. For balance assessment, either 15-segment body CoM (e.g. Bhatt et al., 2011) or pelvis segment CoM (e.g. Kingma, 1995) has been frequently used.

2.6 Biomechanics of Balance in Locomotion

Ageing changes gait due to declines in sensorimotor function and increased fear of falling (Menz et al., 2007). Balance has been previously assessed by inspecting the whole body CoM kinematics relative to the base of support BoS during double support and single support. Position, velocity and acceleration of the CoM relative to the BoS determine dynamic balance (Hof et al., 2005; Lee and Chou, 2006; Winter et al., 1996). The CoM is the 'point' representation of the whole body (Lee and Chou, 2006) and the BoS is defined as the contact area formed by the feet during stance. Body balance is considered to be secured when the CoM is within the BoS but a difficulty to this model is in defining the spatial area between the stance feet (Figure 2.6.1).

During double support step length and step width approximately define the BoS but in normal gait the CoM does not remain above either stance foot, and usually projects within the transverse area between the feet (Granata and Lockhart, 2008). The CoM movement within this area is not hazardous, and during double support the area between the feet (Figure 2.6.1, left) can, therefore, be viewed as a safety zone. This traditional concept has been adequate for analysing balance during double support, but limitations to using the CoM-BoS concept in analysing dynamic balance during single support is visualised as only the contact area of the stance foot defines the BoS but the location of swing foot is totally disregarded (Figure 2.6.1, right). Based on this definition, during single support, balance loss was characterised as the CoM moving away from the stance foot (Hof et al., 2005). While gait can be described as a continuum of losing and regaining balance (Drury and Wolley, 1995), it is necessary to differentiate balance loss as leading to falls from the functional part of gait cycle. Detailed in Chapter 3, the Safety Zone Model is applicable for examination of dynamic balance during single support, in which spatial area between the two feet in the transverse plane regardless of whether feet are in contact with the walking surface, is utilised to characterise balance based on the traditional concept during double support (Figure 2.6.1, left).

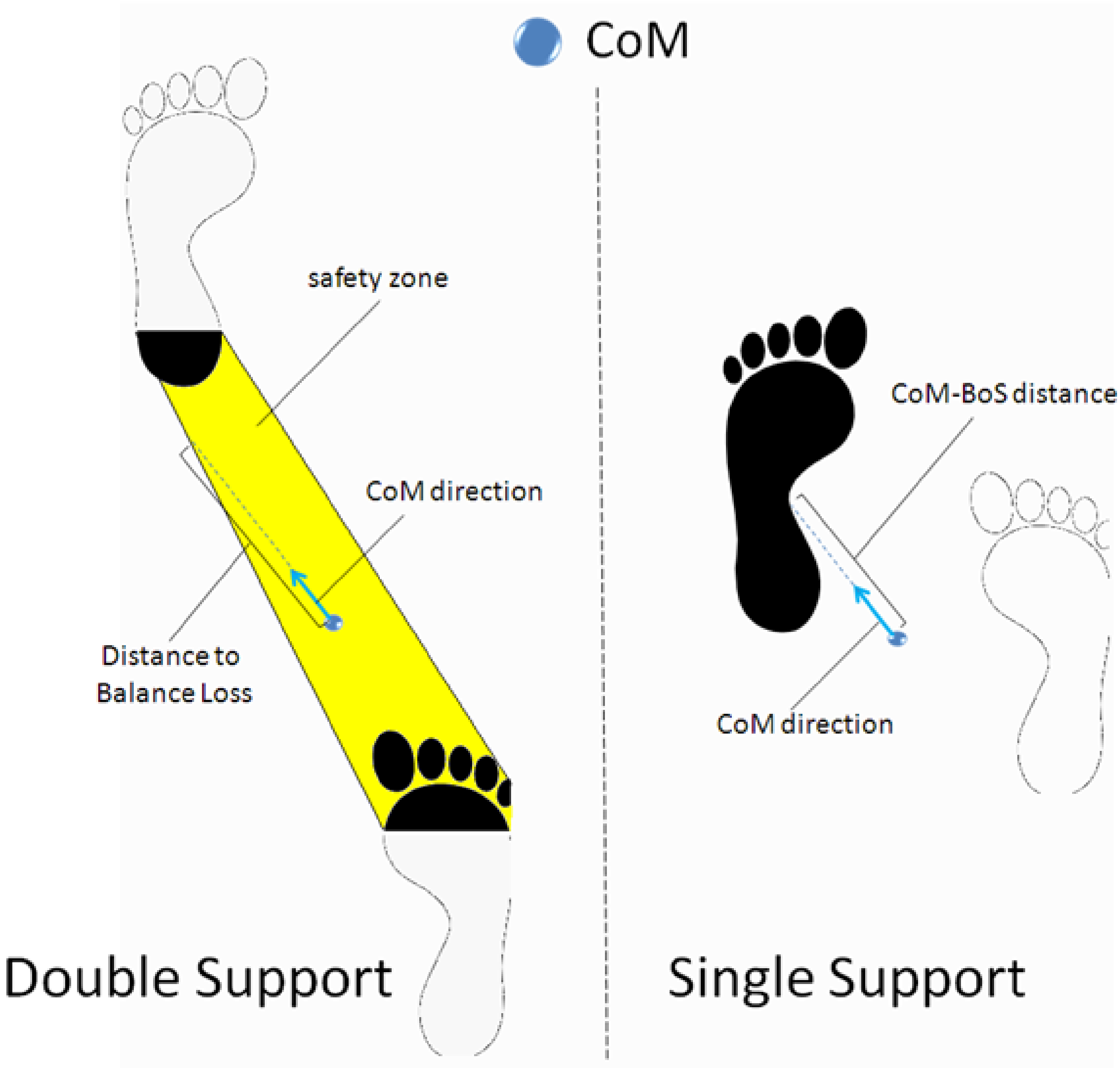

Figure 2.6.1 Traditional dynamic balance measurement using the BoS during double support (left) and single support (right). Distance to Balance Loss and the CoM-BoS distance are based on the CoM movement. Black shaded area; foot contact area. Non-shaded area; foot out of contact with the walking surface (adapted from Lugade et al., 2011).

Despite the limitation in characterising dangerous balance loss during single support based on the traditional model of balance measurement using the CoM and the BoS (Figure 2.6.1), effects of ageing and ageing and fear of falling on balance were systematically examined at heel contact and toe-off (Table 2.6.1). Lugade et al. (2011) defined heel contact as the initial event of double support time and therefore, the spatial area between the two feet was counted as the BoS. In contrast, toe-off was treated as the initial event of swing phase and only the lead foot's contact area determined the BoS, examined by the concept illustrated in the right panel of Figure 2.6.1.

At heel contact, distance from the CoM location to the BoS boundary (Distance to Balance Loss in Table 2.6.1 and Figure 2.6.1) tends to reduce with both ageing and fear of falling. The BoS area shows a similar response, in being the largest in healthy young and smallest in older adults with fear of falling. Available response time (ART) at heel contact, the *temporal* period from the current CoM location to the point on the BoS boundary at which the CoM will eventually cross based on the current velocity, was not differentiated between young, healthy older and older adults with fear of falling because reduction in gait velocity is accompanied with shorter “Distance to Balance Loss”.

Table 2.6.1. Effects of ageing and history of falls on lateral balance; Older FoF = older adults with fear of falling; HC = heel contact; TO = toe-off (adapted from Lugade et al.2011) ****significantly different from Healthy Young † significantly different from Healthy Older***

		Healthy Young	Healthy Older	Older FoF
	Gait Velocity (m/s)	1.38 ± 0.14	1.26 ± 0.20	***†1.02** ± 0.10
Double Support (HC)	Distance to Balance Loss (cm)	23.0 ± 4.1	***18.7** ± 4.0	***17.5** ± 2.6
	ART (s)	0.16 ± 0.03	0.15 ± 0.04	0.17 ± 0.03
	BoS area (cm²)	475.0 ± 59.8	435.4 ± 57.2	***401.9** ± 71.7
Single Support (TO)	CoM-BoS distance (cm)	17.2 ± 3.7	15.3 ± 6.7	***11.3** ± 4.0
	Time to Contact (s)	0.12 ± 0.03	0.11 ± 0.04	0.11 ± 0.04
	BoS area (cm²)	218.0 ± 34.2	219.8 ± 35.7	227.7 ± 40.0

Toe-off was contrarily, investigated by Lugade et al. (2011) as the initial event of single support. Because only lead stance foot is counted as the BoS, similar foot contact area was found in all the three examined groups (Table 2.6.1). The CoM-BoS distance represents the shortest distance from the current CoM location to any part of the *lead* stance foot’s contact area. The trend of the reduced CoM-BoS distance due to ageing and fear of falling was thought to mirror impaired ability to separate the CoM from the BoS in the AP direction (Lugade et al., 2011).

The only study engaged specifically in ML balance involving older adults with and without fear of falling inspected gait initiation patterns during sit-to-walk performance (Åberg et al., 2010). In this study, step width and the ML CoM excursion of the initial step after standing up from a chair was monitored. The study was unique in focusing on the ML CoM movement within the two lateral boundaries of the BoS, but again, due to the limitation of the traditional CoM-BoS model, the investigation was exclusive to single support period (Figure 2.6.2).

Compared to healthy older adults, the fear of falling group took larger step width and associated increase in the ML CoM excursion, but the ML CoM velocity was not differentiated between the two groups. This implies that older adults with 'fear of falling' take temporarily longer step because of the greater distance to travel (Figure 2.6.2). This adaptation could effectively prolong time to redirect the CoM and consequently increase ART to lateral balance loss (Ko et al., 2007; Donelan et al., 2001). From this result again, increased step width can be considered an effective adaptation to secure balance by prolonging ML.

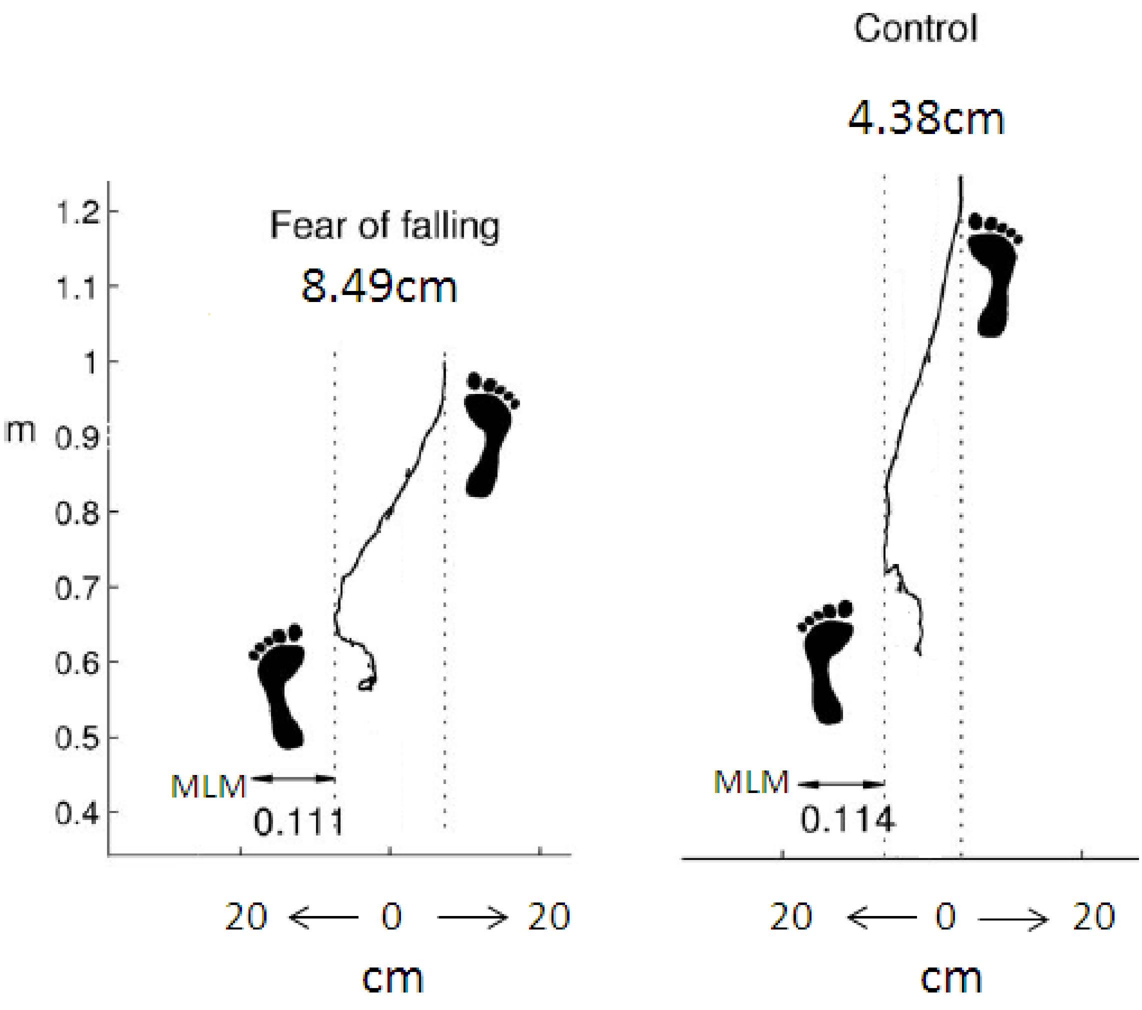

Figure 2.6.2 Effects of fear of falling on the ML CoM motion during sit-to-walk performance. The ML distance of CoM indicated between the two broken lines. MLM is minimum lateral margins, the shortest distance from the lateral boundary of the stance foot contact area to the CoM (adapted from Åberg et al., 2010).

Another important contribution of Åberg et al. (2010) was the report of minimum lateral margin (MLM), the mid-swing event where the lateral distance from the CoM to the stance foot becomes minimum, as shown in Figure 2.6.2. As further detailed in section 2.8, MLM can be considered as the event at which there is a high risk of the CoM being laterally dislocated from the stance foot (Perry et al., 2008). MLM is, therefore, an important concept in understanding lateral balance loss. In this Thesis MLM is further explored in relation to balance control (Figure 2.8.1).

Summarising the information about medio-lateral (ML) balance, enlarged step width is typical in the population with less stable gait but it seems effective in extending ML margin of the CoM excursion, therefore improving ML balance. As implicated by Figure 2.6.2, narrower step width accompanies decreased ML distance for the CoM to travel, possibly increasing the risk of lateral balance loss. These are the rationales for the current study to employ 50% narrower and wider walking as experimental conditions. Both step width manipulations are likely to accompany unfamiliarity (Donelan et al., 2001; Wezenberg et al., 2011), possibly causing fear-associated gait adaptations (section 2.5). In addition to unfamiliarity, narrower walking seems to provide functionally more challenging walking conditions, while wider walking may advantage ML balance due to a wider BoS. Healthy older participants in the current project were not expected to reveal age-associated impairments in balance control when preferred-width walking. With decreased or increased step width they were anticipated to demonstrate impaired balance control as suggested by the previous studies (Donelan et al., 2001; Wezenberg et al., 2011).

The primary advantage of the Safety Zone approach over the traditional BoS-based balance assessment is its applicability to any part of gait cycle including previously unexplored single support. Step width manipulations change heel contact patterns and associated gait characteristics during double support, but single support phase could be more unstable and association between step kinematics and balance during single support can be investigated by applying the safety zone concept. In the current project, four key swing phase events influencing balance were investigated by the safety zone concept as further explained in the following section 2.7 and 2.8 for details about these gait events.

The examined events are illustrated in Figure 2.6.3 below. Toe-off and heel contact are both swing phase initiating and terminating events. These two events commonly accompany the risk of slipping. Minimum foot clearance (MFC) is the mid-swing phase event with low vertical height of swing foot ground clearance and tripping related balance loss is attributed to MFC (Begg et al., 2007; Winter, 1991). Minimum lateral margin (MLM) is another mid-swing event usually following MFC before heel contact where the lateral CoM margin to stance foot becomes minimum (Figure 2.6.2). Unexpected lateral balance perturbation tends to result in lateral balance loss at MLM.

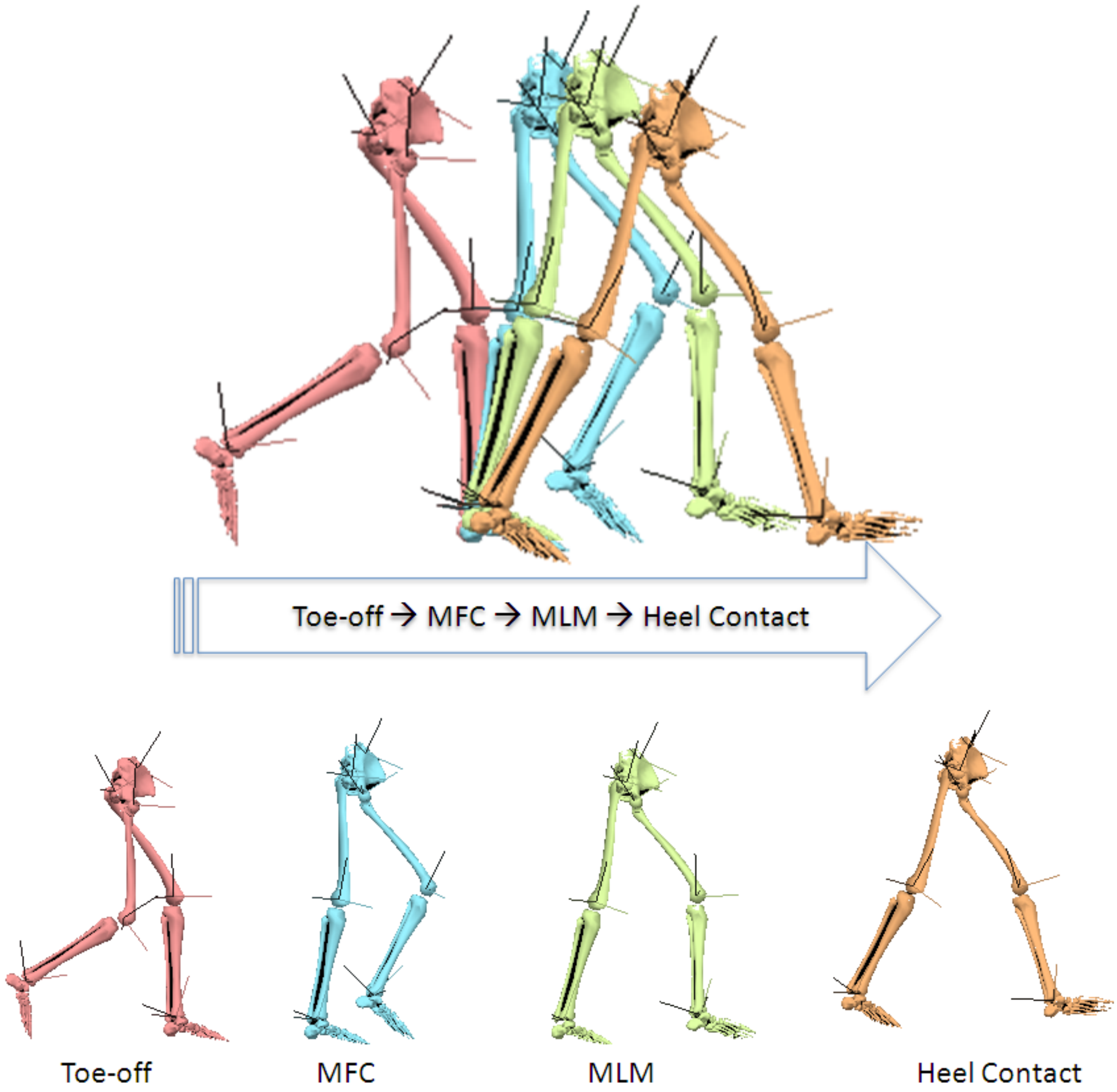

Figure 2.6.3 Illustration of four swing foot events from the sagittal view. Data taken from single subject.

For the present purposes balance is defined as a biomechanical state in which the CoM projection onto the transverse plane remains within a safety zone that changes continuously throughout the gait cycle (Hof et al., 2005; Patla et al., 1991; Terry et al., 2011).

In this sense the Safety Zone Model described here is important in being a dynamic model, changing as continuously throughout the gait cycle, rather than the entirely static concept previously investigated that define balance using variables that are a “snapshot” of gait at a single point in the cycle at double support. Before discussing the Safety Zone approach further, previous research into the four key swing phase events in Figure 2.6.3 is introduced in relation to falls.

2.7 Disturbances to Balance: Tripping and Slipping

Tripping and slipping are the two major causes of AP balance loss (Blake et al., 1988; Berg et al., 1997; Prince et al., 1997). Tripping is associated with anterior balance loss and is the leading cause of falls among older adults (Blake et al., 1988; Berg et al., 1997; Prince et al., 1997). Strategies to reduce the risk of tripping could therefore dramatically decrease falls rates. Tripping will not occur if swing foot ground clearance is greater than the height of an obstacle. Slipping is the second leading cause of falling, accounting for a quarter of falls in older adults and the primary cause of posterior balance loss (Berg et al., 1997; Leclercq, 1999; Yoon and Lockhart, 2006). Hip fracture is commonly associated with slipping falls, recognised as the most serious injury, as permanent loss of some body functions was reported in 50% of cases (Curry et al., 2003; Smeesters et al., 2001; WHO, 2007). Except for the study by Myung (2003), investigations of slipping have been limited to anterior slipping immediately after heel contact, which causes posterior and sometimes the lateral CoM displacement relative to the BoS (Smeesters et al., 2001).

2.7.1 Tripping Biomechanics at Minimum Foot Clearance

Tripping is defined as an event in which the most distal feature of the swing limb, usually the lowest part of the shoe or foot, makes unanticipated contact with either the supporting surface or objects on it with sufficient force to destabilise the walker. Successful swing foot clearance is possible when vertical displacement is sufficient to avoid obstacle contact. Begg et al. (2007) investigated the biomechanics of foot trajectory control and discovered those biomechanical variables that *predict* the probability of tripping. Figure 2.7.1 illustrates a typical swing foot trajectory in the sagittal plane. Swing foot trajectory control in unconstrained walking reflects the risk of contacting undetected obstacles and there is no trajectory modulation associated with intentional obstacle negotiation. The key swing phase events described in Figure 2.7.1 are; Mx1, the first maximum peak of vertical height, observed at around 25% of the swing phase; MFC, the lowest height between Mx1 and Mx2 at around 50% (MFC_t = 50%) of the swing phase; and Mx2, the highest location throughout the swing phase usually observed at around 90% of the swing phase (Nagano et al., 2011). Minimum toe point (MTP) is the most proximal and inferior surface of foot, most commonly used as the point of investigation for swing foot clearance (Begg et al., 2007).

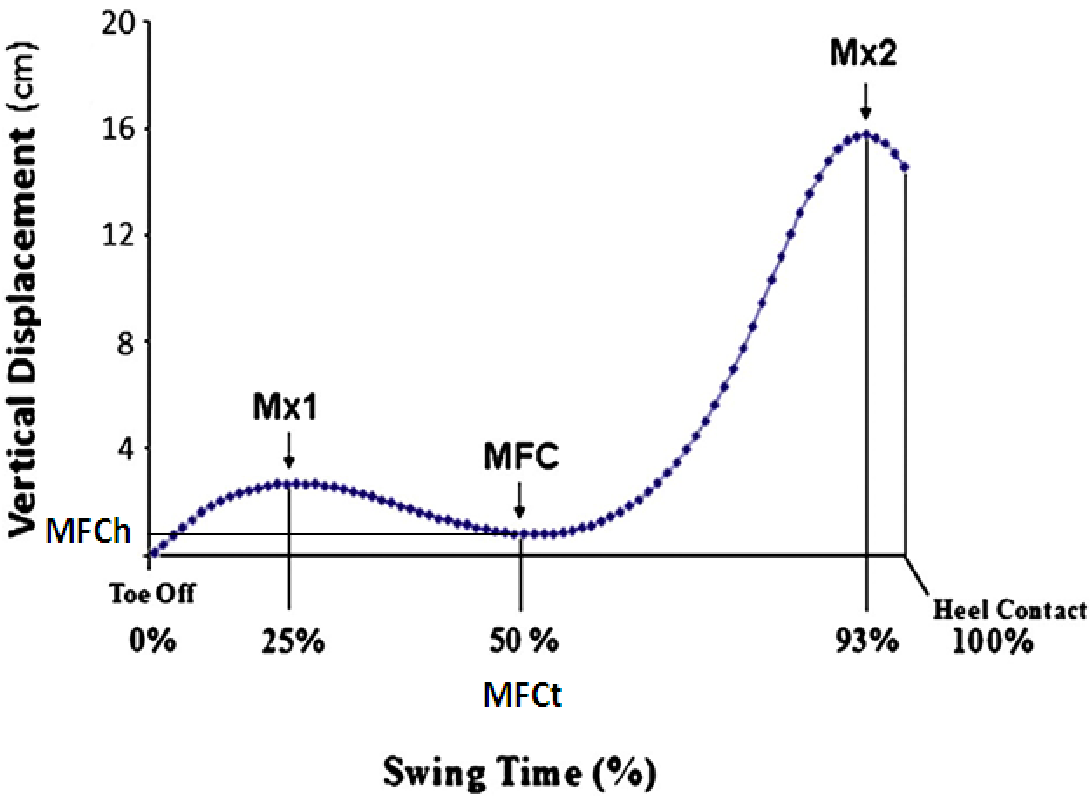

Figure 2.7.1 Swing foot MTP trajectory (adapted from Nagano et al., 2011). MFC_h = vertical displacement of MTP at MFC; MFC_t = timing of MFC (%) during swing time.

Vertical height of the swing foot from the walking surface at minimum foot clearance (MFC_h) is generally within 1.0-2.0 cm of the ground regardless of the walking surfaces, tasks, limb dominance or age, and the foot is moving with a velocity approximately three times walking speed (Begg et al., 2007; Nagano et al., 2011; Sparrow et al., 2008; Winter, 1991). For these reasons, tripping-related balance loss at MFC can be most likely to result in hazardous balance loss in the anterior direction and possibly falls compared to other parts of gait cycle (Smeesters et al., 2001). Tripping at MFC causes greatest momentum and recovery is, therefore, more challenging than at any other swing phase event. Immediately after toe off, foot-ground clearance is lower than at MFC but the swing foot is positioned further posterior to the stance foot and usually provides sufficient anterior margin to prevent anterior balance loss. The swing foot at MFC is, in contrast, parallel to the stance foot, creating a smaller anterior margin to the frontal boundary of the supporting base.

Hip flexion, knee flexion and ankle dorsiflexion are lower limb motions that increase MFC_h and reduce tripping risk (McFadyen and Prince, 2002; Moosabhoy and Gard., 2006). Dorsiflexion is most effective in raising MFC_h (Moosabhoy and Gard, 2006) and compared to knee or hip flexion, it can be speculated that ankle motion should effectively raise MFC_h without affecting the entire gait motion. Practical gait modifications to attain increased MFC_h should minimally disturb natural walking patterns and should be constantly attainable in every gait cycle without intentional control. Therefore, dorsiflexion support can be the most practical joint motion to raise MFC_h and reduce tripping risk, which is applied in the shoe insole tested in the current project (Chapter 4).

Attaining consistent MFC_h throughout multiple swing phases is evidence of effective MFC control and tripping risk is considered to be lower with reduced MFC_h variability (Best and Begg, 2008; Begg et al., 2007). Failure to adequately compensate for surface height variability by adjusting MFC_h makes it the point of highest risk for a trip. Based on these findings, non-normal distributions of foot-ground clearance have been used to determine the precise probability of tripping while recent results have identified asymmetries (between-foot differences) which were important measures for prediction (Nagano et al., 2011).

2.7.2 Ageing Effects on MFC

Previous studies have reported no significant differences in MFC_h between young and older adults. When each swing foot’s ground clearance is plotted in a histogram, however, the most informative presentation of MFC central tendency and distribution characteristics is provided (Figure 2.7.2). Greater MFC variability (IQR) and lower kurtosis are observed in older adults, suggesting less control of MFC_h (Begg et al., 2007). A further indication of inconsistency in gait control is MFC_h asymmetry between the two limbs. Di Fabio et al. (2004) have found that obstacle clearance characteristics between the two feet were more asymmetrical in the population with the higher risk of falls. For normal foot ground clearance, older adults show an age-specific response in increased asymmetry due to higher MFC_h in the non-dominant swing foot (Nagano et al., 2011). This age unique adaptation can be possibly considered as attempt to ensure safety because tripping in the non-dominant limb is harder to recover (Neptune et al., 2001; Perry et al., 2007; Pijnappels et al., 2008; Winter, 1980).

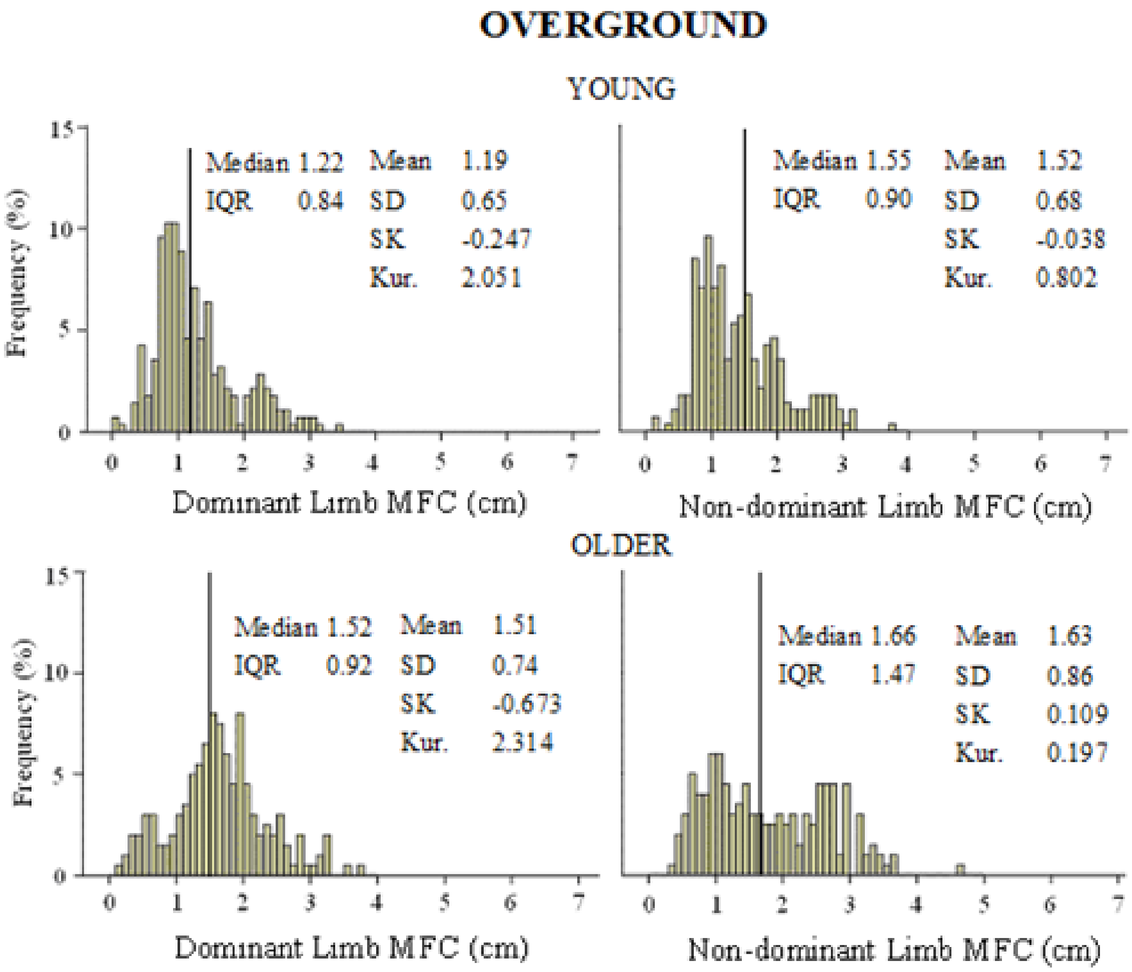

Figure 2.7.2 MFC histogram of both young and older adults walking on overground walking surface (adapted from Nagano et al., 2011).

2.7.3 Slipping Biomechanics

Slipping is due to unexpected loss of traction, again in such a way that the individual is destabilized. Slipping is a sliding motion on a walking surface of the bottom of the shoe of the stance foot, characterised by slip distance and velocity. During normal non-slipping gait, the horizontal ground reaction force (GRF) component is countered by the equal but opposite friction force. When available friction is outweighed by horizontal GRF, net anterior force is produced, accelerating a foot therefore causing a slip. Slipping can be, accordingly, defined as when horizontal GRF is greater than the friction force available.

Another description of slipping is when slip-resistance of the two interfaces (i.e. a bottom sole of a shoe and a walking surface) drops under the certain threshold. Slipperiness of a particular surface is described by coefficient of friction (CoF) and the ratio of vertical and horizontal GRF determines required coefficient of friction (RCoF) that indicates minimum CoF necessary to prevent slipping (Chang, 1999). The conventions about CoF and RCoF are described as follows.

$$CoF = \frac{F(fri)}{F(N)} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad 1$$

$$RCoF = \frac{GRFhor}{GRFver} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad 2$$

where CoF = coefficient of friction; F(fri) = friction force; F (N) = normal force; GRFhor = horizontal GRF component; RCoF = required coefficient of friction

When CoF is lower than RCoF, slipping initiates but under non-slippery walking conditions, RCoF measurements can also identify the risk of slipping for a particular walking pattern when unexpectedly encountering a slippery surface. RCoF assessment of normal walking reveals the three peaks in early stance and one negative peak toward the end of the stance phase, indicating where the risk of slipping arises (Figure 2.7.3).

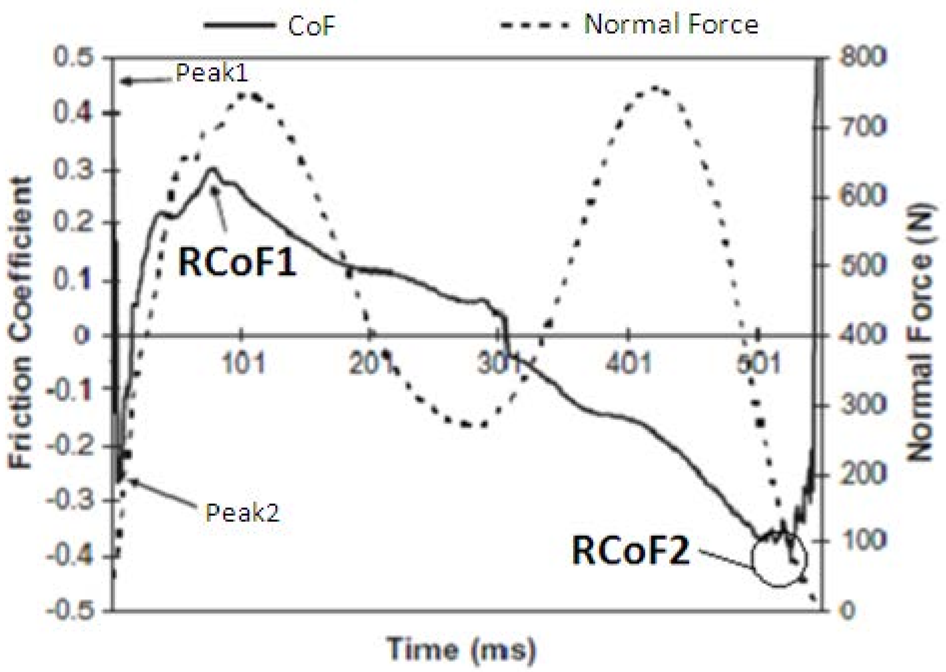

Figure 2.7.3. RCoF measurement throughout the stance phase. Peak 1 indicating potential for heel anterior slipping and Peak2 (negative peak) indicating potential for toe posterior slipping at push off (adapted from Chang et al., 2012)

Positive (> 0) and negative RCoF (< 0) indicate anterior and posterior slipping potentials respectively. Assessment of anterior slipping is possible 50ms to 120ms after heel contact, approximating RCoF1. The initial positive peak indicated with arrow is probably related to micro-slipping, an insensible heel sliding movement less than 30mm that can be distinguished from hazardous slipping events (DiDomenico et al., 2005; 2007). The second negative peak reflects posterior movement of the heel prior to slip initiation, typically within 40ms of heel contact (McGorry et al., 2008). The negative peak, RCoF2 prior to toe-off is related to posterior forefoot slipping, but limited research attention has been given in this area possibly due to the assumption that forefoot posterior slipping does not cause falls. The risk of falls from this action may, however, exist (Myung et al., 2003) in older and others with impaired reaction time and muscular strength particularly if more demanding gait tasks are involved (e.g. turning).

Prevention of anterior heel slipping is possible by decreasing RCoF1. Specific heel contact adaptations including increased heel contact (shank) angle and reduced foot contact angle are effective in reducing RCoF1, characterised as flatter and more vertical foot placement (Brady et al., 2000; McGorry et al., 2008; 2010). In normal unconstrained walking heel contact kinematics can be used to estimate the likelihood of anterior slipping in the case of unexpected foot placement on a slippery surface (Figure 2.7.4).

In summary, investigation of slipping risk of a particular gait pattern can be characterised by inspecting kinetic features at RCoF1 for anterior heel slipping after heel contact and RCoF2 for posterior forefoot slipping prior to toe-off. The current research has identified toe-off and heel contact as key gait events in relation to balance loss subsequent to slipping and therefore, GRF characteristics at both RCoF peaks in addition to heel contact (shank) angle and foot contact angle have been obtained to investigate effects of increased and decreased step widths and the shoe insole on slipping risks.

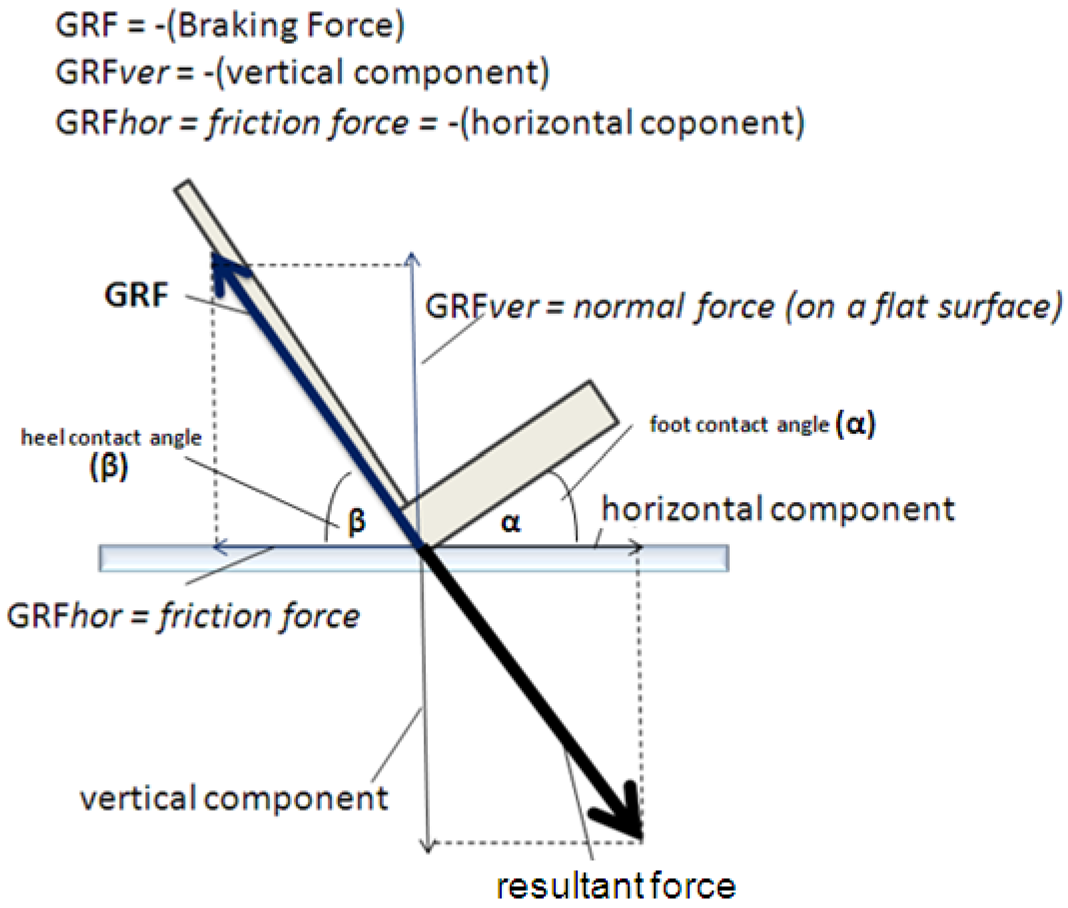

Figure 2.7.4 Ground reaction force (GRF) kinetics at heel contact and definitions of heel contact angle and foot contact angle.

2.7.4 Ageing Effects on Slipping

Older adults were reported to slip further with greater velocity while the CoM horizontal anterior velocity reduces with age (Lockhart et al., 2003). The risk of balance loss due to anterior heel slipping is increased in older adults because when the lead-foot slips forward, the reduced CoM velocity is disadvantageous in preventing the CoM from catching up with the BoS that is moving away from the CoM (You et al., 2001). Safety adaptations to reduce the risk of slipping after heel contact include greater heel contact angle and flatter foot contact (i.e. less dorsiflexion) as discussed (Brady et al., 2000; McGorry et al., 2010). It can be reasonably assumed that older adults have flatter foot contact because they show reduced dorsiflexion prior to heel contact compared to young adults (Nagano et al., 2011). It is unknown whether heel contact angle increases or decreases with age. In one perspective, shorter step length seen in older adults can be due to limited maximum lead-limb forward stroking. If swing limb is placed down in the more conservative length, higher vertical-shank floor contact angle can be expected in older adults. To achieve greater heel contact angle, the CoM should be located ideally straight above contact limb including foot, knee and hip joints.

2.8 Minimum Lateral Margin (MLM)

Toe-off and heel contact influence slipping risks when encountering a walking surface with an unexpectedly low CoF while at MFC, tripping related anterior balance loss is most likely. In contrast, the limited studies have investigated ML balance. In understanding ML balance minimum lateral margin (MLM) is critical, defined as the shortest ML distance between the CoM and the stance heel during the mid-swing phase (Figure 2.8.1). In definition, ML acceleration and velocity of the CoM should be zero at MLM because the CoM medial redirection takes place approximately at MLM. Same as MFC, MLM_d (distance) and MLM_t (timing) describe MLM. Failure to provide successful medial redirection of the CoM at MLM is linked to the increased risk of lateral balance loss. In other words, unexpected ML balance perturbation at MLM is more likely to result in lateral balance loss than for any other part of swing phase. Although Åberg et al. (2010) measured MLM_d between older adults with and without fear of falling as described earlier in Figure 2.6.2, this concept has not been yet fully explored. One motivation for the current research in relation to ML balance is more advanced examination of MLM including MLM_d and MLM_t. Furthermore, as balance maintenance is considered more difficult during single support, the period from toe-off to MLM (MLM_t) can be considered critical in terms of ML balance. Lateral CoP displacement and average medial acceleration on the CoM during this period have been thus captured for more comprehensive understanding underlying the mechanism of lateral balance perturbation at MLM.

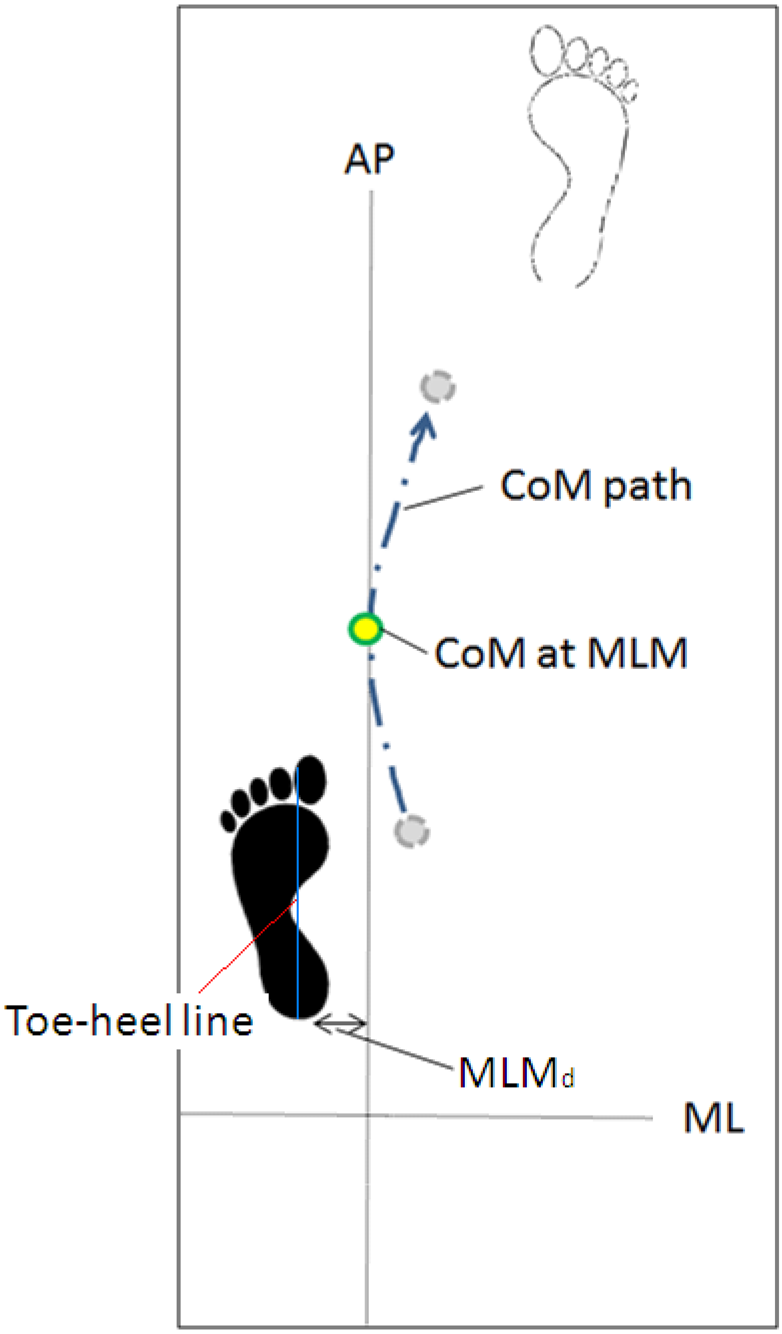

Figure 2.8.1 Image of minimum lateral margin (MLM). MLM_d = minimum medio-lateral distance between stance heel and the centre of mass (CoM). Stance foot = shaded (black) foot, swing foot = non-shaded foot. AP = anterior-posterior, ML = medio-lateral. Time percentage (%) relative to the entire swing phase at MLM is MLM_t.

Chapter 3

A Safety Zone Model of Balance When Walking

The previous section 2.6 has illustrated the traditional model of balance loss assessment method based on the centre of mass (CoM) and the base of support (BoS), but the limitation of the model in application of balance during single support has been also clarified for balance during single support. In this chapter, the Safety Zone Model is first defined and then, its application to assess balance at the swing phase events described in section 2.7 and 2.8 is explained. Furthermore, utility of safety zone model for balance loss simulations at these four events are described.

Previous biomechanical investigations of balance control when walking were limited because they did not capture the "dynamic" or time-varying nature of balance. Stability models are restricted to a theoretically defined event at double support which provides a "snapshot" of balance rather than balance in which the time-varying the base of support (BoS) changes in shape and extent as the gait cycle progresses. To address the research questions in this project a comprehensive *dynamic* Safety Zone Model was developed to more thoroughly investigate ageing effects on walking balance.

3.1 Modelling the Safety Zone

Patla et al. (1991) first introduced the idea of 'virtual base of support' based on the transverse locations of the two feet during single support. This concept is necessary to further explore balance during single support. Regardless of whether or not the foot is in contact with the walking surface, the area between the two feet could be considered a safety zone, where the CoM can travel without hazardous balance loss (Figure 3.1.1). The Safety Zone during single support will represent the BoS in the case of swing foot placement vertically. Balance control during walking depends, therefore, critically, on controlling the CoM within the Safety Zone but in this model the swing foot's trajectory continuously changes the Safety Zone boundaries, both along anterior-posterior (AP) and medio-lateral (ML) directions. Furthermore, centre of pressure (CoP) excursion controls the CoM.

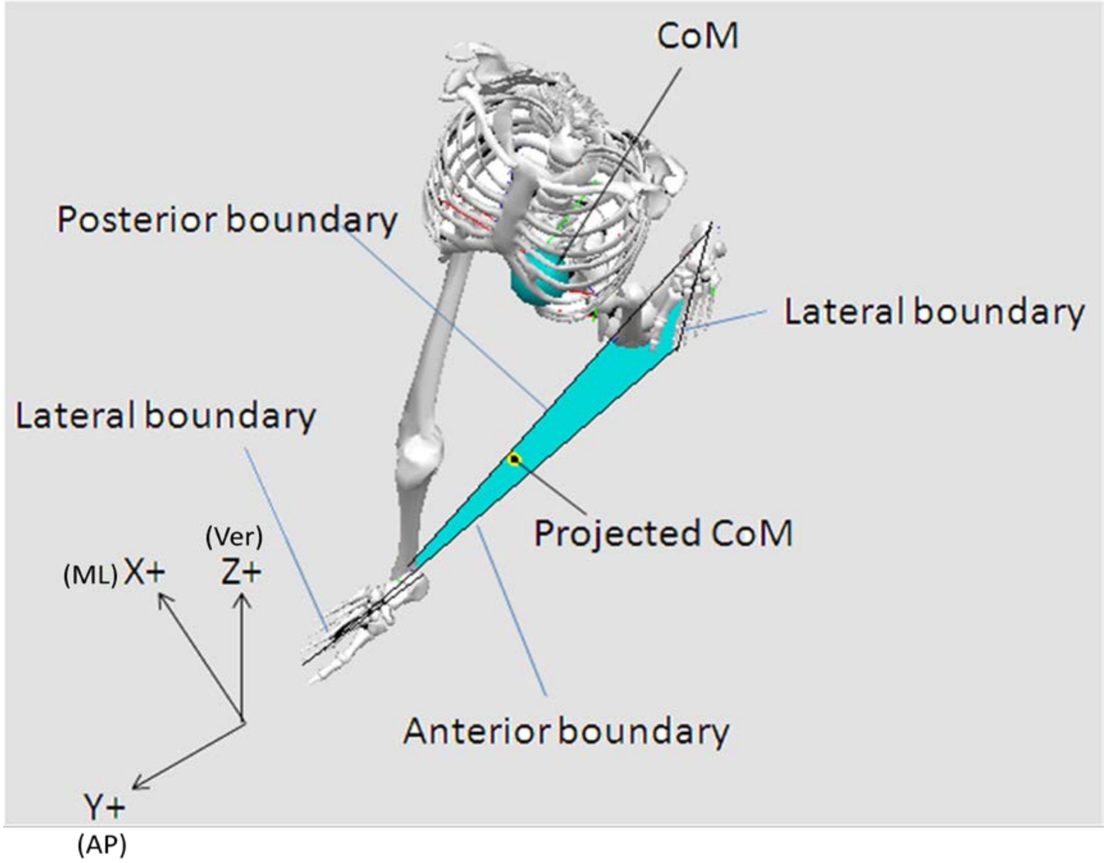

Figure 3.1.1 Image of the Safety Zone Model. The Safety Zone = light blue shaded area. X+ = lateral(right), y+ = anterior-posterior, z = vertical.

First, in the Safety Zone Model, the spatial transverse area between the two toe-heel lines of both feet (Figure 3.1.2) represents *conservative* boundaries for lateral CoP movement. The line between two toes defines the anterior boundary of the Safety Zone while the two heels define the posterior boundary.

The typical CoP trajectory in normal gait begins approximately at the heel, passing through the plantar surface area by taking a slight lateral curve and disappearing at toe (Shanthikumar et al., 2010). Despite functional necessity in lateral CoP movement to redirect CoM medially, any excessive lateral CoP displacement can be a risk factor for lateral balance perturbation (Han et al., 1999; Rougier, 2009; Rugier, 2008; Winter et al., 1996). At heel contact, for example, as little as extra 2cm lateral deviation of CoP can cause lateral balance loss (Rietdyk et al., 1999).

Compared to the relatively stable double support phase, it can be more difficult to maintain balance during single support. ML CoP displacement therefore tends to be minimised during single support (Hof et al., 2005) as shown in Figure 3.1.2 as the typical CoP path during normal gait. In an event of the swing foot's unexpected placement, sudden onset of CoP in the area lateral to the typical CoP path (e.g. the very lateral edge of the foot) could raise a risk of inversion sprain and associated risk of ML balance perturbation. Typical CoP path instead of the entire contact area may be therefore more appropriate to define safety zone especially during single support. For this reason, the lateral boundaries of the Safety Zone Model are defined by toe-heel lines of both feet (Figure 3.1.2).

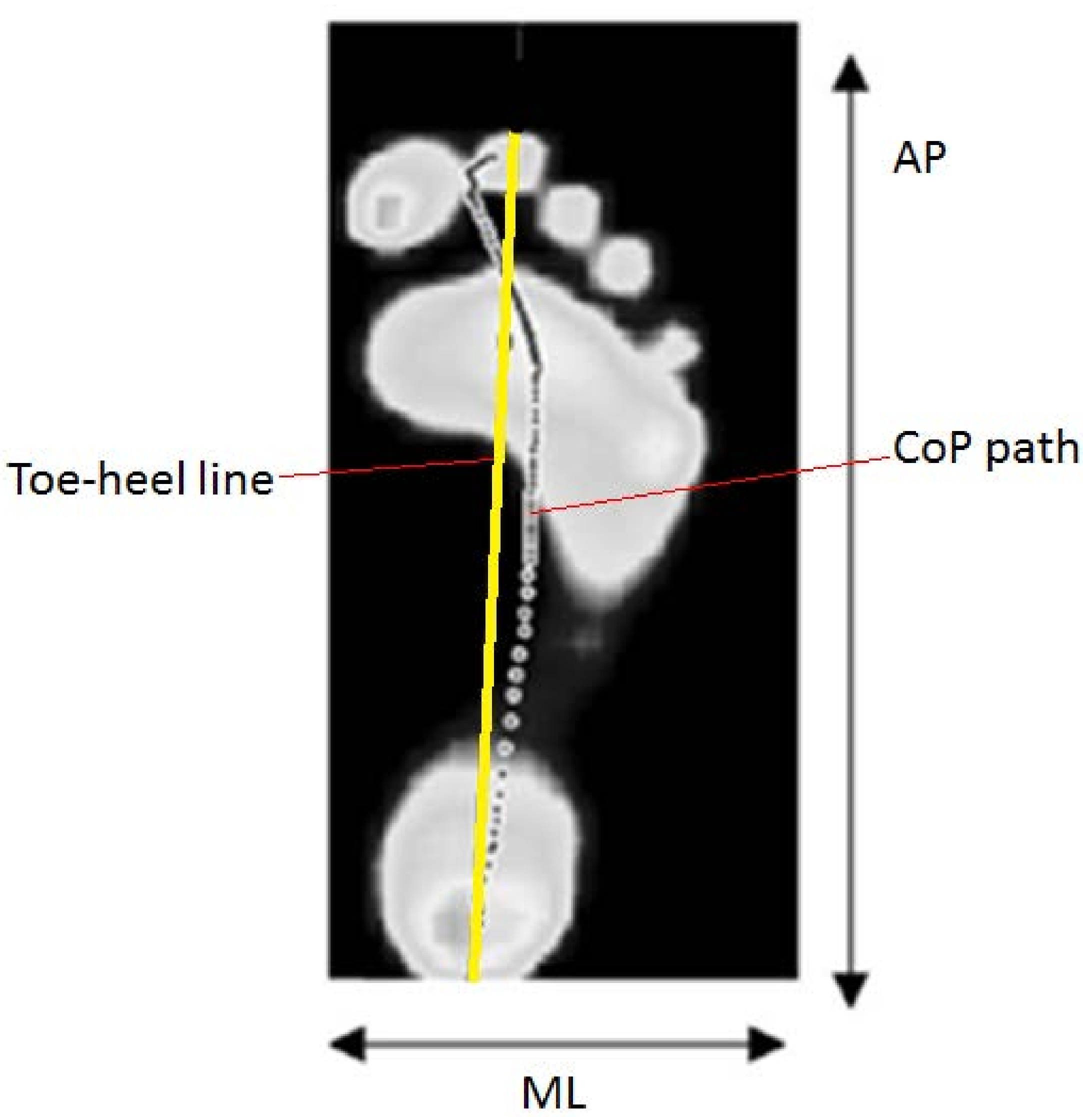

Figure 3.1.2 Toe-heel line and typical CoP path during stance: from heel contact to toe-off (Cook et al., 2008).

3.2 Balance During Single Support Phase

Once the Safety Zone is defined, hazardous balance loss during single support can be distinguished from balance loss as functional requirement of gait by inspecting whether the transverse CoM location is within the safety zone. A systematic method by which to characterise dynamic balance using the Safety Zone is to analyse the CoM location relative to both feet (Figure 3.2.1). Locations of both feet relative to the CoM determine the shape of Safety Zone and represent the margin to each boundary. Based on the CoM velocity, a 'Balance Loss Point' can be estimated and Distance to Balance Loss calculated. Computation of ART is then possible by dividing Distance to Balance Loss by the average linear CoM velocity from the current location (0, 0) to the Balance Loss Point.

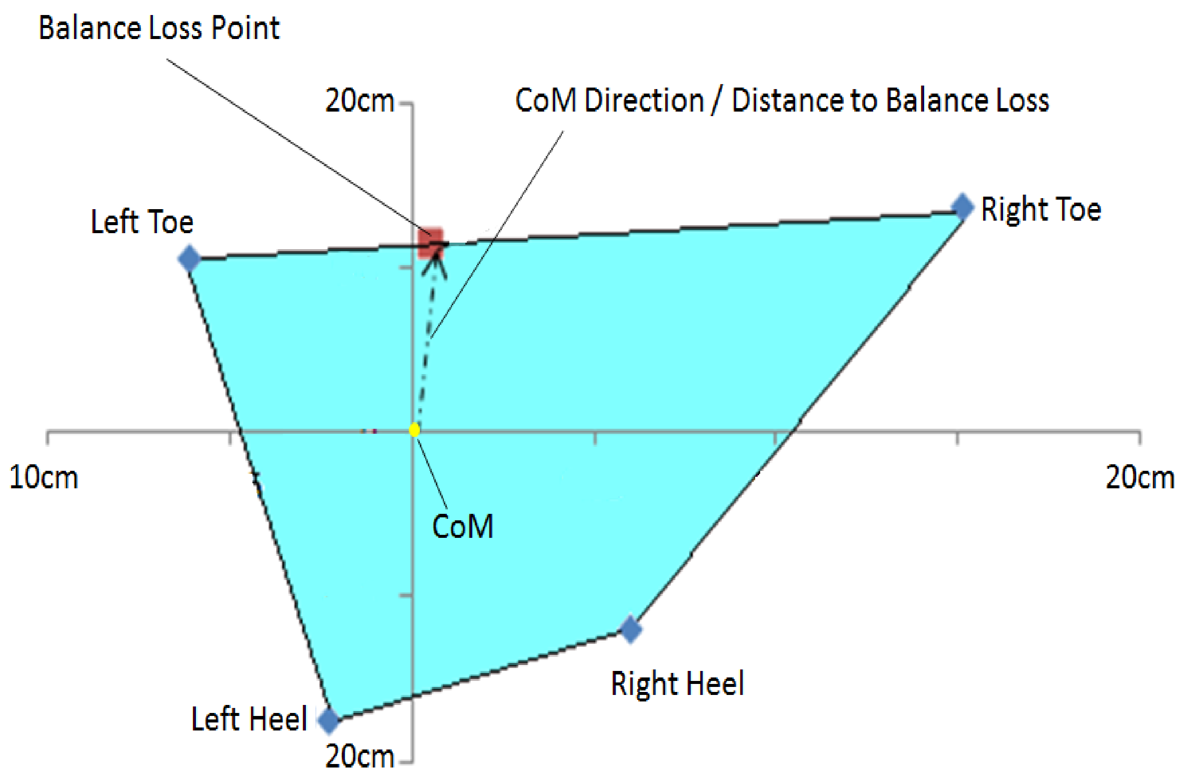

Figure 3.2.1 Safety Zone based balance assessment at right swing MFC. The Safety Zone (coloured area); defined by the transverse positions of toes and heels of the both feet relative to the CoM (0, 0). The CoM direction is based on the CoM transverse linear velocity. Distance to Balance Loss = distance to boundary of the Safety Zone based on the CoM direction.

The Safety Zone Model allows investigation of dynamic balance during single support and four key swing events (toe-off, heel contact, MFC and MLM) will be compared between the young and older populations; the dominant and non-dominant limb parameters;

and the three step widths walking conditions: preferred, narrow and wide. Figure 3.2.2 illustrates the typical Safety Zone characteristics at the swing phase events examined.

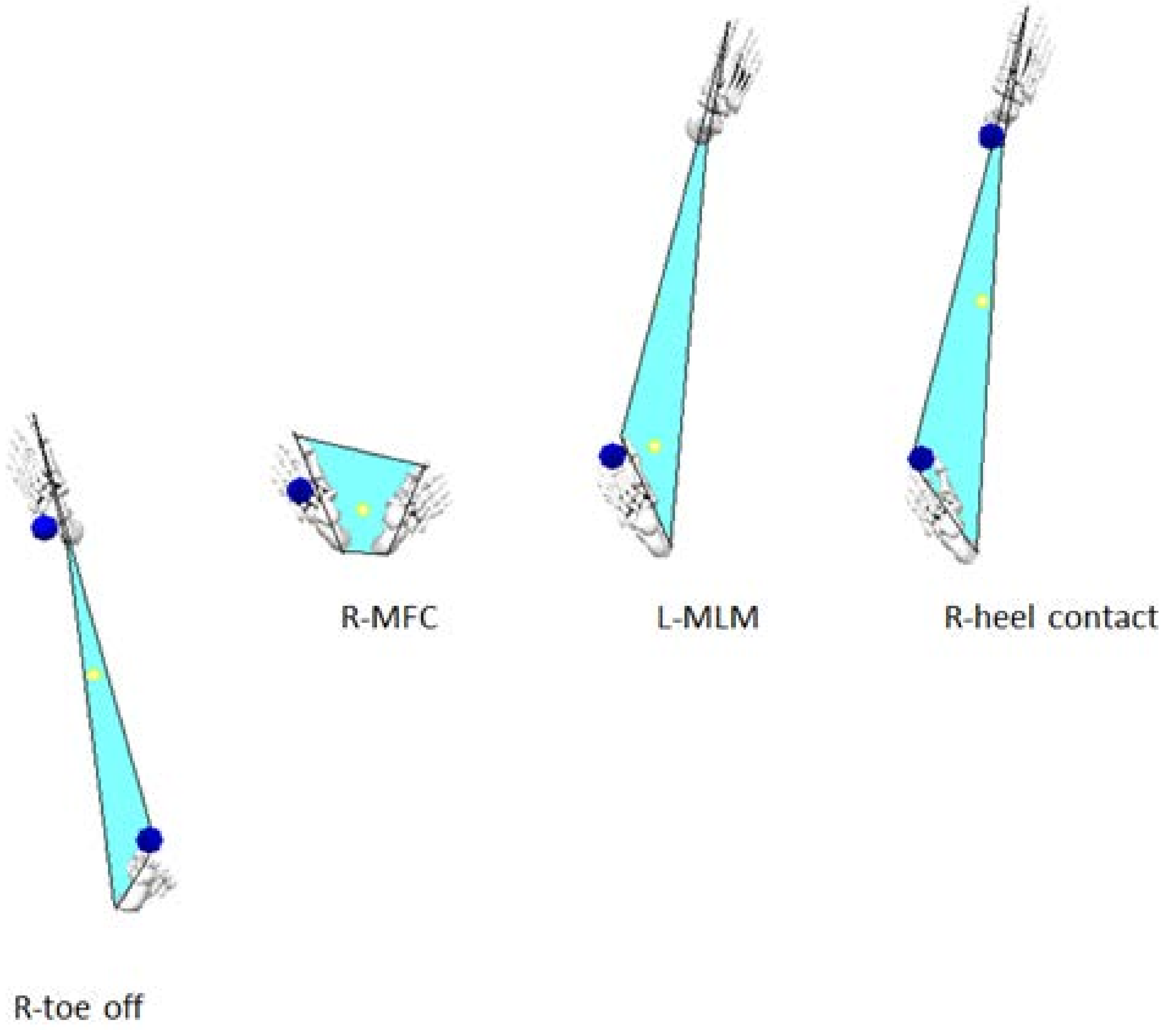

Figure 3.2.2 Safety Zone during the swing phase. Blue dot = CoP, yellow dot = CoM, light blue shaded area = Safety Zone.

3.3 Inverted Pendulum (IP) Model

The Safety Zone Model visualises the CoM movement in the transverse plane but for balance loss simulation, it is also important to understand the CoM movement in the sagittal plane. The inverted Pendulum (IP) model has been the predominant theory to explain human walking during single support. As illustrated in Figure 3.3.1, the lower end of pendulum is the foot (lateral malleolus) and the other end is the CoM of the pelvis segment or greater trochanter. The IP model can be characterised by sagittal motion of the CoM relative to the stance foot. The Safety Zone Model incorporates the IP model illustrated below (Figure 3.3.1) to simulate balance loss due to tripping and slipping.

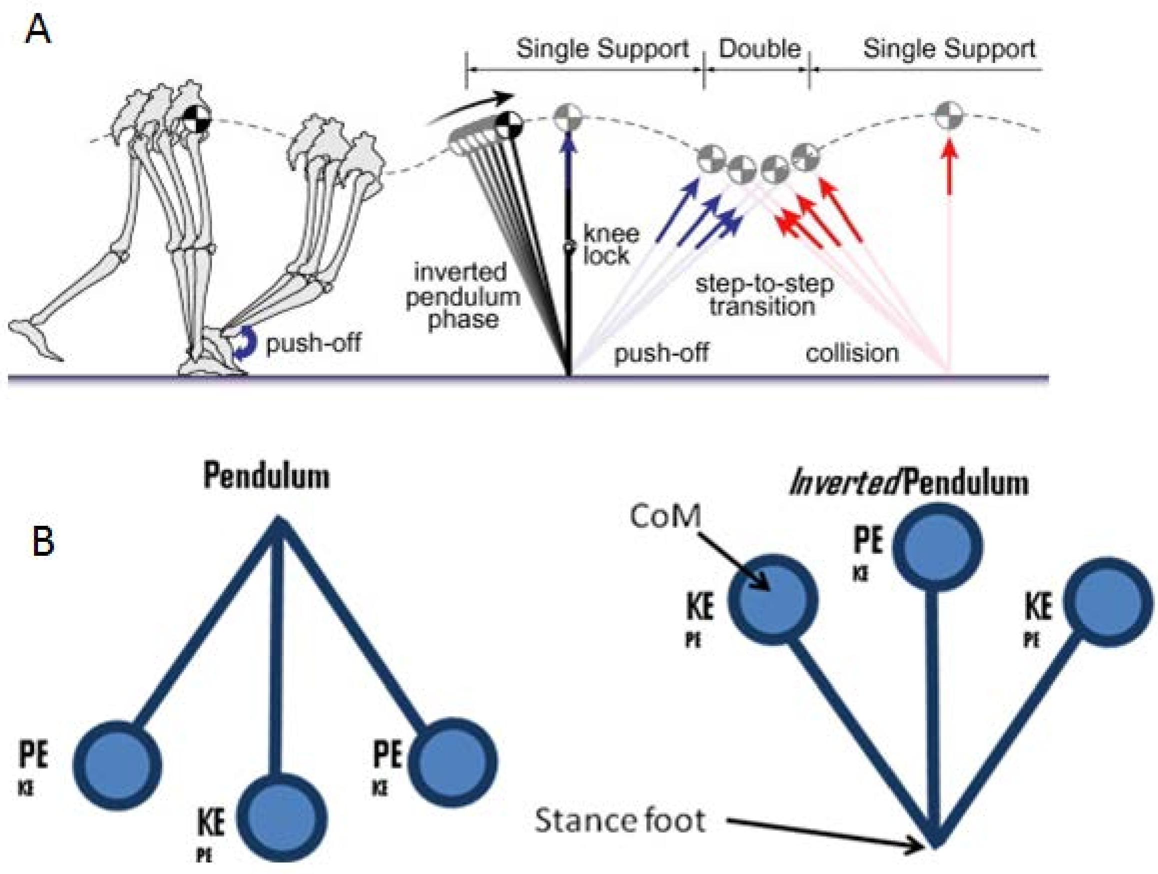

Figure 3.3.1 Dynamic walking theory. (A) Inverted pendulum (IP) mechanics during single support phase and step-to-step transition during double support phase. No energy required for IP movement. Energy is, however, required for step-to-step transition (adapted from Kuo, 2007); (B) mechanical energy exchange of pendulum and inverted pendulum mechanics. PE = potential energy, KE = kinetic energy.

In the IP model (Figure 3.3.1 B), lower limbs are considered as rigid segments and no mechanical energy expenditure is required to transport the CoM anterior during single support, only the exchange between potential energy and kinetic energy (Nagano et al., 2012). Step-to-step transition during double support (Figure 3.3.1 A), in contrast, links one IP cycle with the following IP cycle of the opposite limb and energy expenditure is necessary to produce the ground reaction forces (GRF) required to shift the CoM into the next IP cycle (Kuo, 2007). The established method of calculating energy efficiency of step-to-step transition during double support time is known as recovery rate, in which the mechanical energy during double support time is computed, described as Δ(KE+PE) in Figure 3.3.2

(Cavagna et al., 1976; Collett et al., 2007; Vereecke et al., 2006; Ortega and Farley, 2003; Detremblur et al., 2005; Schepens et al., 2004), reported to be 60-70% in normal gait (Collett et al., 2007; Nagano et al., 2012; Schepens et al., 2004).

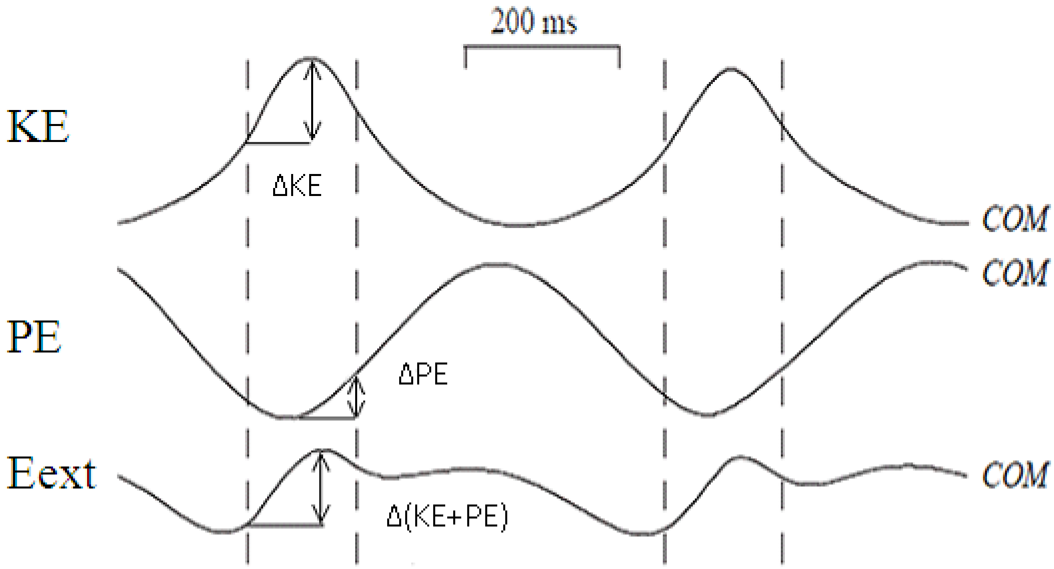

Figure 3.3.2 Illustration of recovery rate (%). Symbols: ΔKE = increase in KE during the double support phase, ΔPE = increase in PE during the double support time. Eext: input of external mechanical energy. Δ(KE+PE) = increase in the total mechanical energy (KE + PE) during the double support time. (adapted from Schepens et al., 2004)

3.4 Balance Loss Simulation based on Force Couple

In this section the background is developed for balance loss simulation using the experimental data. The Safety Zone Model will be applied to balance loss simulation based on unexpected external perturbations including slipping before toe-off and after heel contact, tripping at minimum foot clearance (MFC) and lateral balance loss at minimum lateral margin (MLM). Movement of the centre of mass (CoM) and foot locations in the transverse plane describe dynamic balance. External disturbances can be described as a kinetic input to the foot and the CoM. Kinetic inputs precede kinematic changes in gait and force determines acceleration, velocity and associated displacement of both the CoM and feet. Slipping simulation must consider movements of both the CoM and the Safety Zone boundary. In contrast no Safety Zone boundary simulation is required for tripping on a fixed object because the foot does not move. Similarly, the Safety Zone boundary will remain unchanged due to lateral balance perturbations that do not affect foot placement. Mechanical energy inputs to the current CoM movement are, however, modelled.

Tripping simulation and anterior and posterior slipping begin with a sagittal plane description of kinetic input and its kinematic effect on the CoM (Figure 3.4.1 below). Later the changing kinematic relationship between the CoM and the Safety Zone in the transverse plane is described (Figures 3.4.2, 3.4.3, 3.4.4). Lateral balance loss simulation at MLM is undertaken only in the transverse plane, using the stance foot location and lateral force. Slipping and tripping simulation of the estimated CoM movements due to kinetic inputs to the foot are described below (Figure 3.4.1).

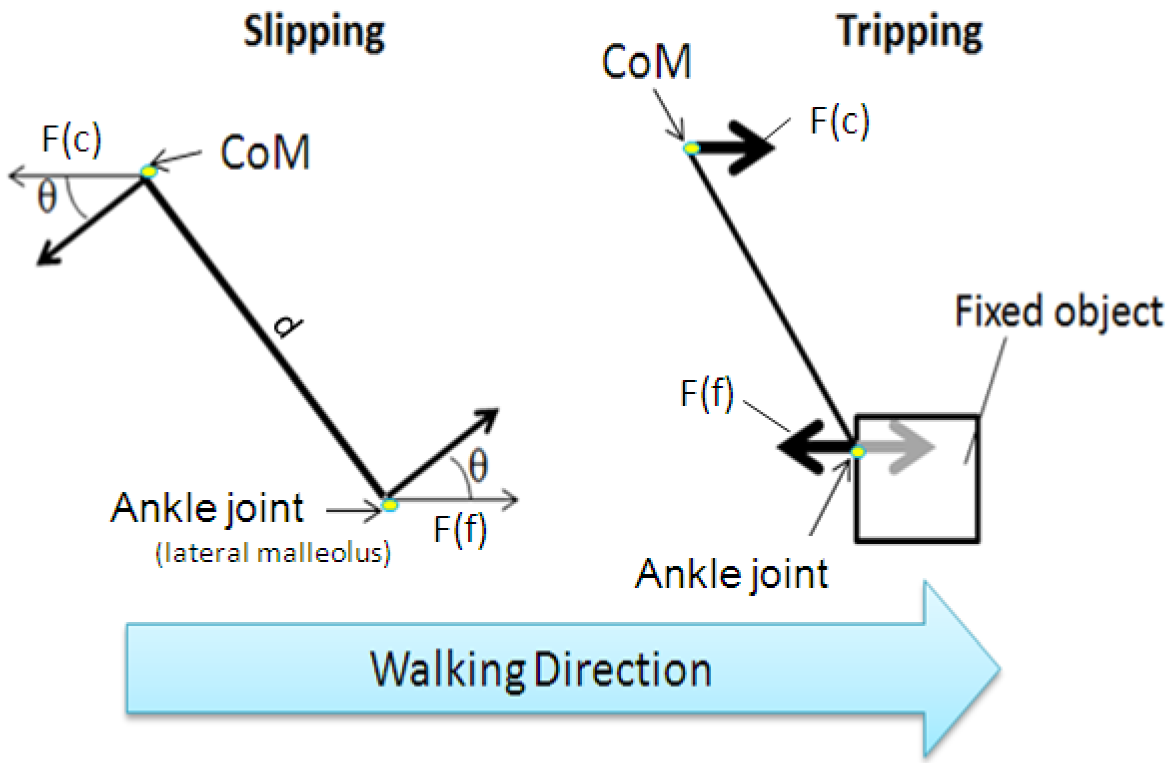

Figure 3.4.1 Force couple between the foot (lateral malleolus) and the pelvis CoM. (Left) during slipping (right) tripping over a fixed object. Arrows indicate forces on the foot and the CoM.

Estimation of the whole body CoM has been based on the 15 segment models' kinematic or kinetic centre (Gutierrez-Farewik et al., 2006), but the pelvis segment CoM has been also used for the transverse plane CoM motion for balance assessment of normal walking (Kingma et al., 1995; Mengi et al., 2011; Rabuffetti and Baroni, 1999). Pelvis CoM is also more adequate for the IP model of the sagittal gait description (Figure 3.3.1) than the whole body CoM that does not have segment mass. As shown in Figure 3.4.1, both slipping and tripping can be described as 'force couple'. Gait simulation therefore, begins with describing disturbances on the foot as force inputs to estimate simultaneous effects on the

CoM. In this process, segment mass is necessary for the CoM and for that purpose too, the pelvis CoM is suitable for the current research.

Slipping can accelerate the CoM opposite to force applied by the foot, as illustrated in the left panel of Figure 3.4.1. In the sagittal plane view, using the lower limb as the axis between the greater trochanter and lateral malleolus, foot sliding transfers force to the CoM via an equal and opposite torque, causing force couple. To compute the CoM acceleration due to slipping the force has to be divided by the pelvis segment mass. As detailed in Chapter 6 the pelvis CoM is used for estimating acceleration.

Calculation of torque requires the leg length (axis in the IP model), from lateral malleolus to greater trochanter. However, as described in Figure 3.4.1 (left), the force component parallel to the walking surface can be obtained by applying the same cosine (θ) factor to the force perpendicular to the axis. Following the equations below, therefore, in the IP description, force generated on the CoM by net force on a slipping foot does not require moment arm for calculation, but the same force can be assumed at both the pelvis and foot.

$$T = F_{(f)} \times \sin\theta \times d = -F_{(C)} \times \sin\theta \times d \quad \ldots\ldots \quad 3$$

$$F_{(C)} = \frac{T}{M_{(C)} \times \sin\theta \times d} \quad \ldots\ldots \quad 4$$

$$A_{(C)} = \frac{-F_{(f)}}{M_{(C)}} \quad \ldots\ldots \quad 5$$

where T = torque, F(f) = net horizontal force, d = lower limb length, F(c) = force working on the CoM, M(c) = mass of pelvis segment, A(c) = acceleration on the CoM due to a perturbation on a foot.

The right panel of Figure 3.4.1 illustrates force couple between the pelvis and foot when tripping. A swing foot’s impact force on a fixed object will generate an equal and opposite reaction force, producing linear anterior acceleration of the CoM. Applying the same technique, the leg length (moment arm) is not required to estimate the CoM acceleration due to tripping i.e. $F_{(c)} = -F_{(f)}$.

3.4.1 Toe (Forefoot) Posterior-Slipping

Posterior toe slipping occurs when push-off force outweighs the available friction force (Myung, 2003). Simulation of anterior balance loss due to toe posterior slipping begins with the GRF characteristics at RCoF2, the negative RCoF peak before toe-off. Friction force on a flat surface is determined by the vertical GRF and CoF.

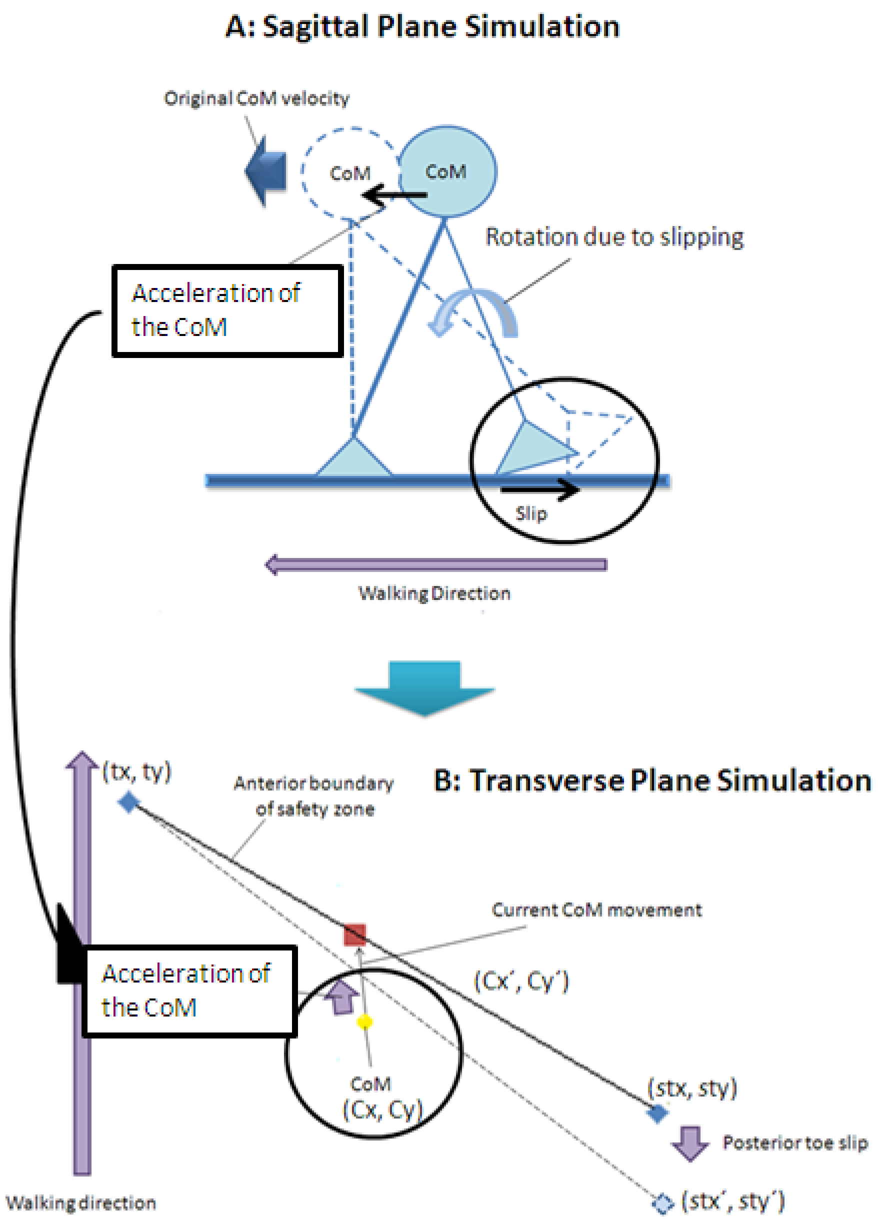

Figure 3.4.2 Illustration of anterior balance loss due to posterior forefoot slipping. (A) The sagittal illustration of toe posterior slipping (B) the transverse illustration of toe posterior slipping. (C_x, C_y) = the current CoM coordinates; ($C_{x'}$, $C_{y'}$) = the CoM coordinates at anterior balance loss initiation; (st_x, st_y) = the slipping foot coordinates at RCoF2; ($st_{x'}$, $st_{y'}$) = the slipping foot coordinates at anterior balance loss initiation.

As long as slipping does not occur during walking, CoF is greater than RCoF, but on the assumption of CoF lower than RCoF, acceleration of slip can be computed, by dividing net posterior force by the foot segment mass. Posterior movement of the forefoot then takes place i.e., forefoot posterior slipping. The anterior border of the Safety Zone, the line between the two toes, consequently begins posterior movement, accelerating the CoM toward the Safety Zone's *anterior* boundary.

Figure 3.4.2 illustrates ART calculation for anterior balance loss due to toe posterior slipping. In the sagittal plane (top), acceleration of the slipping foot and the CoM can be estimated. In the transverse plane, obtained acceleration due to torque generated by posterior forefoot slipping creates the CoM anterior acceleration, while the anterior boundary of the Safety Zone moves posterior due to posterior forefoot slipping. Thus, the posterior movement of the anterior boundary of the Safety Zone and the anterior CoM movement are the two factors causing anterior balance loss due to posterior forefoot slipping. Computation methods for ART due to forefoot posterior slipping comprise two processes, as follows.

3.4.1.1 Sagittal plane

$$\mathrm{F(f)} = GRFver \times CoF - GRFhor \quad \ldots\ldots \quad 6$$

$$\mathrm{A(c)} = \frac{\mathrm{F_{(f)}}}{\mathrm{M_{(c)}}} \quad \ldots\ldots \quad 7$$

$$\mathrm{A(f)} = \frac{\mathrm{F_{(f)}}}{\mathrm{M_{(f)}}} \quad \ldots\ldots \quad 8$$

where F(f) = force working on a foot due to slipping, A(c) = acceleration on CoM due to slipping, M(c) = pelvis segment mass, A(f) = acceleration of foot due to slipping, M(f) = foot segment mass

3.4.2.2 Transverse plane

Anterior balance loss due to posterior toe slipping can be defined as when both toe (t_x, t_y; $st_{\acute{x}}$, $st_{\acute{y}}$) and the CoM ($C_{\acute{x}}$, $C_{\acute{y}}$) locations are on a straight line indicated by the same slopes of any lines drawn between these three points.

$$\frac{t_y - C_{y\prime}}{t_x - C_{x\prime}} = \frac{C_{y\prime} - st_{y\prime}}{C_{x\prime} - st_{x\prime}} = \frac{t_y - st_{y\prime}}{t_x - st_{x\prime}} \quad \ldots\ldots \quad 9$$

Using ART (t), the following equations determine the CoM and toe coordinates at anterior balance loss initiation due to forefoot posterior slipping.

$$\mathrm{CoM}\ (C_{\acute{x}}, C_{\acute{y}}) = (V_{(cx)} \times \mathrm{t} + 0.5 \times A_{(cx)} \times \mathrm{t}^2, V_{(cy)} \times \mathrm{t} + 0.5 \times A_{(cy)} \times \mathrm{t}^2) \quad \ldots\ldots \quad 10$$

Slipping foot ($st_{\acute{x}}$, $st_{\acute{y}}$) = $(st_x + 0.5 \times A_{(fx)} \times t^2, st_y + 0.5 \times A_{(fy)} \times t^2\ )$ …………… 11

Solving quadratic equation (9) for the case of balance loss, using the CoM (10) and the slipped foot coordinates (11) ART (t) can be calculated as follows (detailed in Appendix IX).

$t = \frac{-b \pm \sqrt{b^2 - 4ac}}{2a}$ ………………………………………………………………… 12

where $a = 0.5(\mathrm{A(c)} - \mathrm{A(f)}, \mathrm{b} = \mathrm{V(c)}, \mathrm{C} = -\mathrm{C} - \mathrm{st} + \mathrm{st}'$

3.4.2 Heel Anterior Slipping

The fundamentals of heel anterior slipping simulation are similar to forefoot posterior slipping in that a sagittal plane analysis is first necessary to estimate the CoM acceleration due to slipping. Following this, the effect of the CoM acceleration can be calculated in the transverse plane to compute time for the CoM to reach the ‘posterior’ boundary. As for forefoot posterior slipping, posterior balance loss will occur when the heels and the CoM are in-line.

Subsequent to anterior heel slipping, the non-slipping trail foot is assumed to remain stationary while both the CoM and lead slipping foot move. For the simulation of anterior heel slipping, simulated friction force should be considered lower than horizontal GRF at RCoF peak after heel contact (RCoF1; Figure 2.7.3), which allows calculation of net force causing a foot to slip. Vertical GRF and simulated CoF determine the maximum friction force and when horizontal GRF is greater, net anterior force is produced, therefore anterior acceleration. Posterior boundary of safety zone moves in the anterior direction and if eventually reaching the location of the CoM, posterior balance loss begins.

Posterior balance loss due to anterior heel slipping can be conceptualised by the relationship between the CoM and the BoS as described in Figure 3.4.3 below (Hof et al., 2005; You et al., 2001). As with the posterior forefoot slipping model described earlier, net anterior force causing heel anterior slipping can be divided by pelvis mass to compute posterior acceleration. Given that posterior balance loss is predicted when the coordinates of both heels and the CoM in the transverse plane are aligned, available response time (ART) can be computed. Left of Figure 3.4.3 shows that posterior torque generated by foot anterior slipping motion moves the CoM posterior but this occurs only when linear posterior acceleration due to torque eventually causes net posterior velocity on the CoM, otherwise the CoM will slow. Even if CoM is still moving anterior, posterior balance loss could occur when the slipping foot travels faster than the CoM (You et al., 2001). The fundamental simulation methods are same as outlined in the section above for forefoot posterior slipping including equations 1-5. The only difference in the calculation is that the contacting foot is assumed to have heel contact velocity, in contrast the forefoot is initially stationary, prior to slipping.

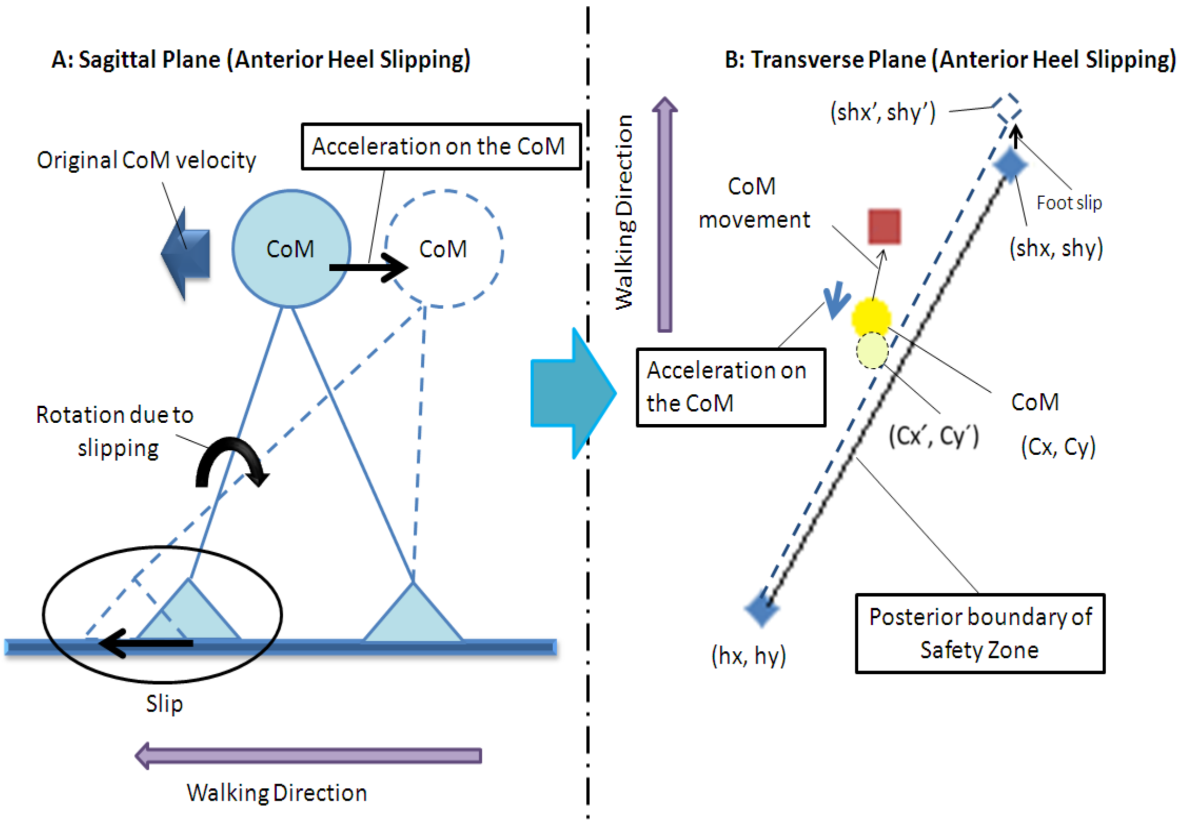

Figure 3.4.3 Illustration of posterior balance loss due to anterior heel slipping. (A) The sagittal illustration of foot anterior slipping. Net anterior force causing anterior slipping and generating posterior torque on the CoM. (B) The transverse illustration of the CoM and posterior boundary of the Safety Zone.

3.4.2.1 Sagittal Plane

$$F_{(f)} = GRFver \times CoF - GRFhor \quad \ldots \quad 13$$

$$A_{(c)} = \frac{F_{(f)}}{M_{(c)}} \quad \ldots \quad 14$$

$$A_{(f)} = \frac{F_{(f)}}{M_{(f)}} \quad \ldots \quad 15$$

where $F_{(f)}$ = force working on a foot due to slipping, $A_{(c)}$ = acceleration on the CoM due to slipping, $M_{(c)}$ = pelvis segment mass, $A_{(f)}$ = acceleration of foot due to slipping, $M_{(f)}$ = foot segment mass

3.4.2.2 Transverse Plane

When both heels (h_x, h_y; sh_x, sh_y) and the CoM ($C_{\dot{x}}$, $C_{\dot{y}}$) locations are on a straight line (equation 14), indicated by the same slopes of any lines drawn between these three points, coordinates of the CoM and a slipping foot can be expressed as follows (equation 15 and 16):

$$\frac{h_y - C_{y\prime}}{h_x - C_{x\prime}} = \frac{C_{y\prime} - sh_{y\prime}}{C_{x\prime} - sh_{x\prime}} = \frac{h_y - sh_{y\prime}}{h_x - sh_{x\prime}} \quad \ldots\ldots \quad 16$$

$$\text{CoM } (C_{\dot{x}}, C_{\dot{y}}) = (V_{(cx)} \times t + 0.5 \times A_{(cx)} \times t^2, V_{(cy)} \times t + 0.5 \times A_{(cy)} \times t^2) \ldots\ldots \quad 17$$

Slipping foot

$$(sh_{\dot{x}}, sh_{\dot{y}}) = (sh_x + 0.5 \times A_{(fx)} \times t^2,\ sh_y + \sqrt{V_{(fx)}{}^2 + V_{(fy)}{}^2} \times t + 0.5 \times A_{(fy)} \times t^2\ ) \quad \ldots\ldots \quad 18$$

Heel slipping distance to balance loss can be then described as in equation 19

$$\sqrt{(sh_{x\prime} - sh_x)^2 + (sh_{y\prime} - sh_y)^2} = \sqrt{V_{(fx)}{}^2 + V_{(fy)}{}^2} \times t + 0.5 \times \sqrt{A_{(fx)}{}^2 + A_{(fy)}{}^2} \times t^2 \ldots\ldots \quad 19$$

where ($C_{x'}$, $C_{y'}$) = the CoM coordinates at posterior balance loss, $V_{(cx)}$ = medio-lateral CoM velocity, $V_{(cy)}$ = anterior-posterior CoM velocity, $A_{(cx)}$ = medio-lateral CoM acceleration, $A_{(cy)}$ = anterior-posterior CoM acceleration, t = available response time to posterior balance loss due to a simulated slip, ($sh_{x'}$, $sh_{y'}$) = coordinates of a slipping heel at posterior balance loss, (sh_x, sh_y) = coordinates of a slipping heel at RCoF1, $A_{(fx)}$ = medio-lateral acceleration on a foot, $A_{(fy)}$ = anterior-posterior acceleration on a foot, $V_{(fx)}$ = medio-lateral heel contact velocity, $V_{(fy)}$ = anterior-posterior heel contact velocity,

3.4.3 Tripping

As illustrated in the inverted pendulum (IP) model (Figure 3.4.4) falling due to tripping tends to be anterior (Smeesters et al., 2001). Tripping reaction forces create a torque that increases the CoM angular velocity toward the Safety Zone anterior boundary. Torque working on the CoM in the IP model yields a linear acceleration component. Figure 3.4.4 illustrates that reaction forces on the tripping foot should be equivalent to anterior force on the CoM when tripping.

ART computation for anterior balance loss involves two stages; 1) sagittal plane simulation to obtain anterior acceleration on the CoM due to tripping and 2) transverse plane simulation to compute ART for the CoM to reach the anterior boundary. Computation of anterior balance loss due to tripping at MFC requires swing foot mass and horizontal toe acceleration at MFC to compute impact force.

3.4.3.1 Sagittal Plane

$$F_{(f)} = A_{(f)} \times M_{(f)} \quad \ldots\ldots \quad 20$$

$$A_{(c)} = \frac{A_{(f)} \times M_{(f)}}{M_{(c)}} \quad \ldots\ldots \quad 21$$

where $F_{(f)}$ = reaction force working on swing toe due to tripping; $A_{(f)}$ = acceleration working on swing toe; $M_{(f)}$ = mass of foot segment

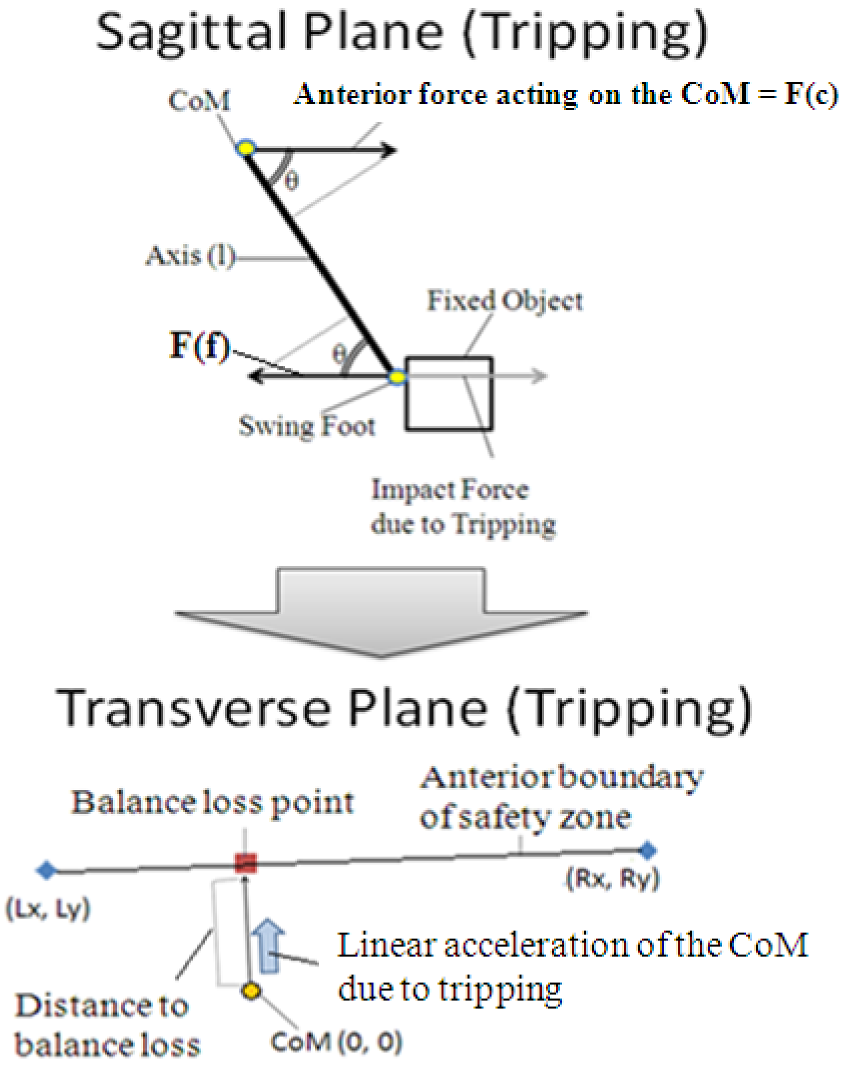

Figure 3.4.4 Illustration of anterior balance loss due to tripping. (Top) Sagittal illustration of tripping. $F_{(c)}$ = force acting on the CoM due to tripping; $F_{(f)}$ = reaction force working on a tripped foot,(Bottom) Anterior balance loss simulation in the transverse plane. (R_x, R_y) = right toe coordinate; (L_x, L_y) = left toe coordinate.

3.4.3.2 Transverse Plane

Available response time (ART) can be calculated by dividing distance to balance loss by the average CoM velocity from the current (0, 0) location to the balance loss point (Fig. 3.4.5, bottom). Balance loss occurs at the intersection of the CoM trajectory and the anterior boundary, (X, Y) below. Distance to balance loss (D) and ART (t) can be expressed as follows.

The intersection's coordinate (X, Y);

$$X = \left[R_y - \frac{R_x(R_y - L_y)}{R_x - L_x}\right] / \left[\frac{V_{(cy)}}{V_{(cz)}} - \frac{R_y - L_y}{R_x - L_x}\right] \quad \ldots\ldots\ldots\ldots \quad 22$$

$$Y = \frac{V_{(cy)}}{V_{(cz)}} \times \left[R_y - \frac{R_x(R_y - L_y)}{R_x - L_x}\right] / \left[\frac{V_{(cy)}}{V_{(cz)}} - \frac{R_y - L_y}{R_x - L_x}\right] \quad \ldots\ldots\ldots\ldots \quad 23$$

Distance to balance loss (D);

$$D = \sqrt{\boldsymbol{X^2 + Y^2}} \quad \ldots\ldots\ldots\ldots \quad 24$$

ART (t);

$$t = \frac{-V(c) + \sqrt{V(c)^2 + 2 \times A(c) \times D}}{A(c)} \quad \ldots\ldots\ldots\ldots \quad 25$$

where R/L(x, y) = right/left foot coordinate; (V(cx), V(cy)) = velocity of CoM (x, y), A(c) = acceleration on the CoM.

3.4.4 Lateral Balance Loss

Balance loss due to simulated lateral force can be modelled only in the transverse plane because the CoM cannot be displaced medio-laterally at MLM. If there is a lateral force on the CoM, lateral balance loss may occur due to the MLM being associated with the least lateral distance from the CoM to the stance foot. It was, therefore, possible to compute the minimum lateral force required to cause lateral balance loss by forcing the CoM toward the stance toe (Figure 3.4.5).

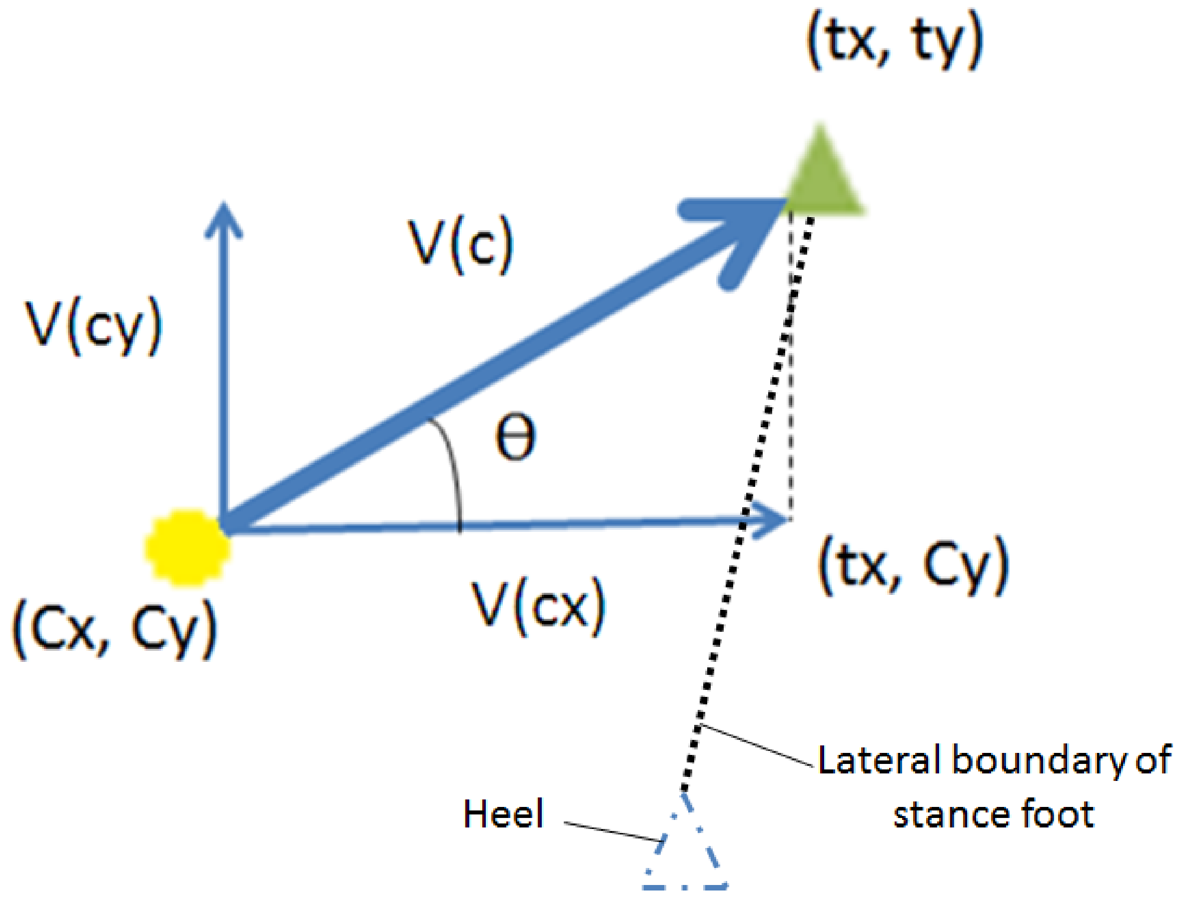

Figure 3.4.5 Lateral balance loss simulation in the transverse plane. CoM location (Cx, Cy); stance toe location (tx, ty); V(c) = resultant CoM velocity in the transverse plane due to lateral push; V(cx) = CoM lateral velocity due to lateral push; V(cy) = CoM anterior velocity.

Lateral force should generate V(cx) necessary for directing resultant velocity (Cv) toward the stance toe (tx, ty). Time taken for the CoM to reach stance toe (t) can be calculated from the anterior CoM velocity (V(cy)), AP location of the CoM (Cy), and the stance toe position (ty).

$$t = \frac{(ty - Cy)}{V(cy)} \quad \ldots\ldots \quad 26$$

For the essential condition of lateral balance loss, the CoM accordingly has to travel to the lateral toe location in t seconds.

$$(tx - Cx) = V(cx) \times t \quad \ldots\ldots \quad 27$$

To compute the average lateral CoM velocity (V(cx)), estimated lateral push needs to be divided by pelvis segment mass.

$$C(vx) = 0.5 \times \frac{F(l)}{M(c)} \times t^2 \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad 28$$

where F(l) = the minimum amount of lateral push to cause the CoM lateral balance loss

Following the computation methods above, F(l) can be obtained as the minimum force required to cause the CoM to be dislocated from the stance foot laterally.

Chapter 4

Footwear Interventions to Reduce Falls Risk and Lower Limb Injuries

Various strategies have been devised to reduce falls rate. One approach is supervised exercise interventions, for example, Hess and Woollacott (2005) reported that 10 weeks of intensive lower-limb muscular training (flexors/extensors) improved dynamic balance of older adults and reduced falls risk. Bhatt and Pai (2009) showed that multiple exposures to induced-slipping in a secure environment were an effective training to reduce slipping falls. Hass et al. (2004) reported that Tai Chi training for 48 weeks improved the CoM-BoS relationship and helped to prevent lateral balance loss. Despite these positive outcomes, without constant supervision and a sense of obligation, older adults are less likely to engage in exercise interventions (Yardley et al., 2008; Simek et al., 2012).

For this reason, practical falls prevention strategies may not be adopted voluntarily unless all the following requirements are met: low cost, effortless, immediate effect, and easy application (Yardley et al., 2008). Footwear interventions fulfil these requirements and can assist outdoor walking, a recommended exercise for older adults in maintaining active lifestyles despite the higher risk of falling while walking outside (Berg et al., 1997; Sherrington and Menz, 2003; Koepsell et al., 2004; Li et al., 2006; Curry et al., 2003; WHO, 2007). Footwear designs to reduce the risk of falling are, therefore, worthy of further research attention.

4.1 Footwear Interventions

In 63% of falls, older adults had footwear with inappropriate features (Gabell et al., 1985; Munro et al., 2009). Although older adults are broadly aware of the link between improper footwear and the increased risk of falling, they tend to choose footwear based on comfort rather than safety (Australian Government, 2007; Munro et al., 2009). This is reflected in the tendency to avoid shoes with fixation or heel counter, and sandals are commonly worn (22%) when they experience falls (Sherrington and Menz, 2003). Except supervised situations, such as rehabilitation facilities or hospitals, it is difficult to control everyday footwear, implying that comfortable safe shoes are required to be worn voluntarily.

4.2 Modification of Shoe Features

Menant et al. (2008) examined the effect of footwear features on balance, spatio-temporal gait parameters, and other gait data. Elevated heel (high heel shoes) is a factor in causing lateral instability, with caution-related adaptations in spatio-temporal parameters including longer double support time and reduction in gait velocity (Menant et al., 2008). Hard soles can more effectively provide tactile sensation to improve reactions to maintain dynamic balance than a standard sole. Heel collars also improve balance by providing increased tactile sensation around the ankle via extended contact area of the collar (Menant et al., 2008). A high collar could, however, increase tripping as reduced swing foot height at minimum foot clearance ($MFC_{h)}$ was reported as an effect of a high collar and based on the CoM-BoS inspection, while a high collar is advantageous for ML balance loss it may negatively affect AP balance. The shoe's tread could increase step length and associated gait velocity, assisting in lengthening the posterior margin the base of support (BoS) relative to the centre of mass (CoM) (Menant et al., 2008). The details of tread patterns were not, however, described by Menant et al. (2008) and accordingly, it is only possible to speculate that modifications to tread pattern can affect balance when walking.

Aschan et al. (2005) examined slip-resistance of 23 materials commonly used in shoe outsoles, especially the heel (Kim et al., 2001). European standards for professional footwear specify PU (polyurethane), TPU (thermoplastic polyurethane) or heat-proof NBR (nitrile-butadiere rubber) as anti-slip outsole materials for the oil surfaces but on icy surfaces other materials have better slip resistance (Aschan et al., 2005). Thus, no single material seems to be the solution for all walking surfaces. Application of multiple anti-slip materials in outsole design may be advantageous but with increased manufacturing cost.

4.3 Modification of Insoles and Other Orthotics

Compared to shoe modifications, insoles can be more cost effective and applicable for various types of shoes. Previous insole intervention can be classified into the three different types: 1) modifying ankle orientation inside the shoe; 2) providing greater contact area between foot and insole; and 3) stimulating cutaneous receptors.

Ankle motions can be described fundamentally by a combination of the four motions described below (Figure 4.3.1). Although all four ankle motions are necessary in walking, adequate dorsiflexion and eversion are considered to be more effective in enhancing safe gait and improving balance.

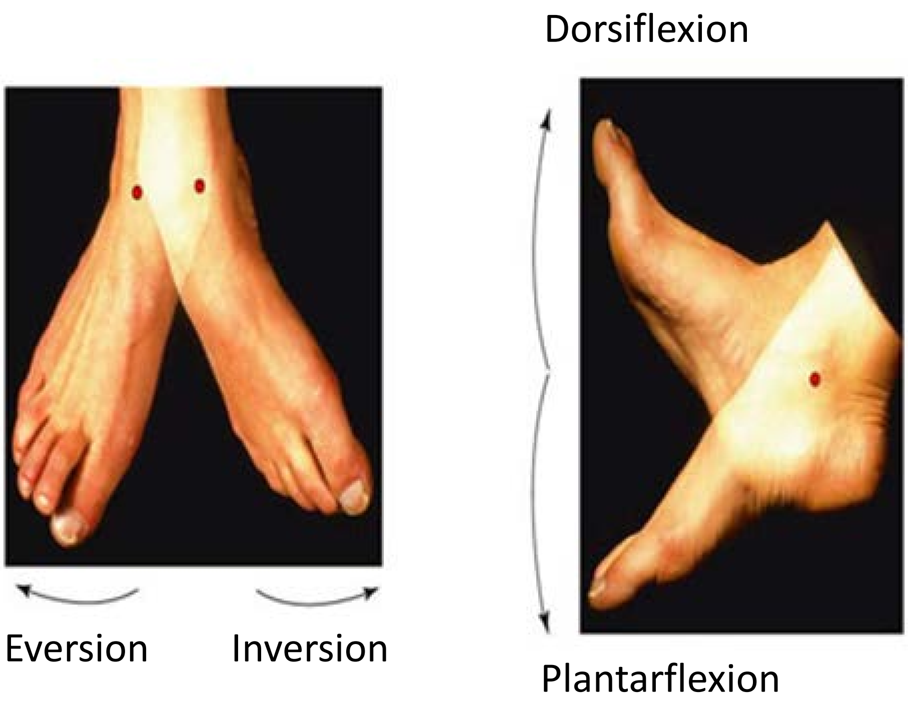

Figure 4.3.1 The four fundamental ankle motions (adapted from www.neuronarc.com)

4.3.1 Ankle Orientation Inside the Shoe

Dorsiflexion: Positive effects of ankle dorsiflexion have been reported for enhanced reaction, greater swing foot-ground clearance, ulcer prevention and shock absorption. Dorsiflexion is considered superior to plantarflexion for dynamic balance by augmenting sensory feedback. Dorsiflexion stretches the triceps surae, enhancing proprioception, therefore enabling quicker balance restoration, while in contrast, plantarflexion was reported to disturb ML balance (Hock et al., 2011; Macklin et al., 2012; Mecagni et al., 2000; Rougier et al., 2009; Silver-Thorn et al., 2012; Willems et al., 2005). Dorsiflexion supporting orthotics has been reported to increase swing foot-ground clearance (Kao and Ferris, 2009).

Increased MFC_h is the most fundamental adaptation to reduce the risk of tripping (Begg et al., 2007). Moosabhoy and Gard (2006) reported that one degree increase in dorsiflexion can elevate MFC_h by approximately 0.3cm, the most effective joint motion to raise MFC_h. Throughout the entire swing phase, dorsiflexion constantly increases swing foot-ground clearance. For example, there is the evidence that increased MTP height at Mx1 also increased MFC_h when older adults walked cautiously on a treadmill (Nagano et al., 2011). Higher swing foot-ground clearance at Mx1 can be also achieved by dorsiflexion (Moosabhoy and Gard, 2006).

Age-related changes in gait pattern begin with deterioration of sensorimotor function but foot deformity is common in older adults, often accompanying pain-associated undesirable gait adaptations. The major cause of foot deformity is ulcer development due to excessive and continuous plantar pressure at the metatarsal regions and Hallux (Figure 4.3.2).

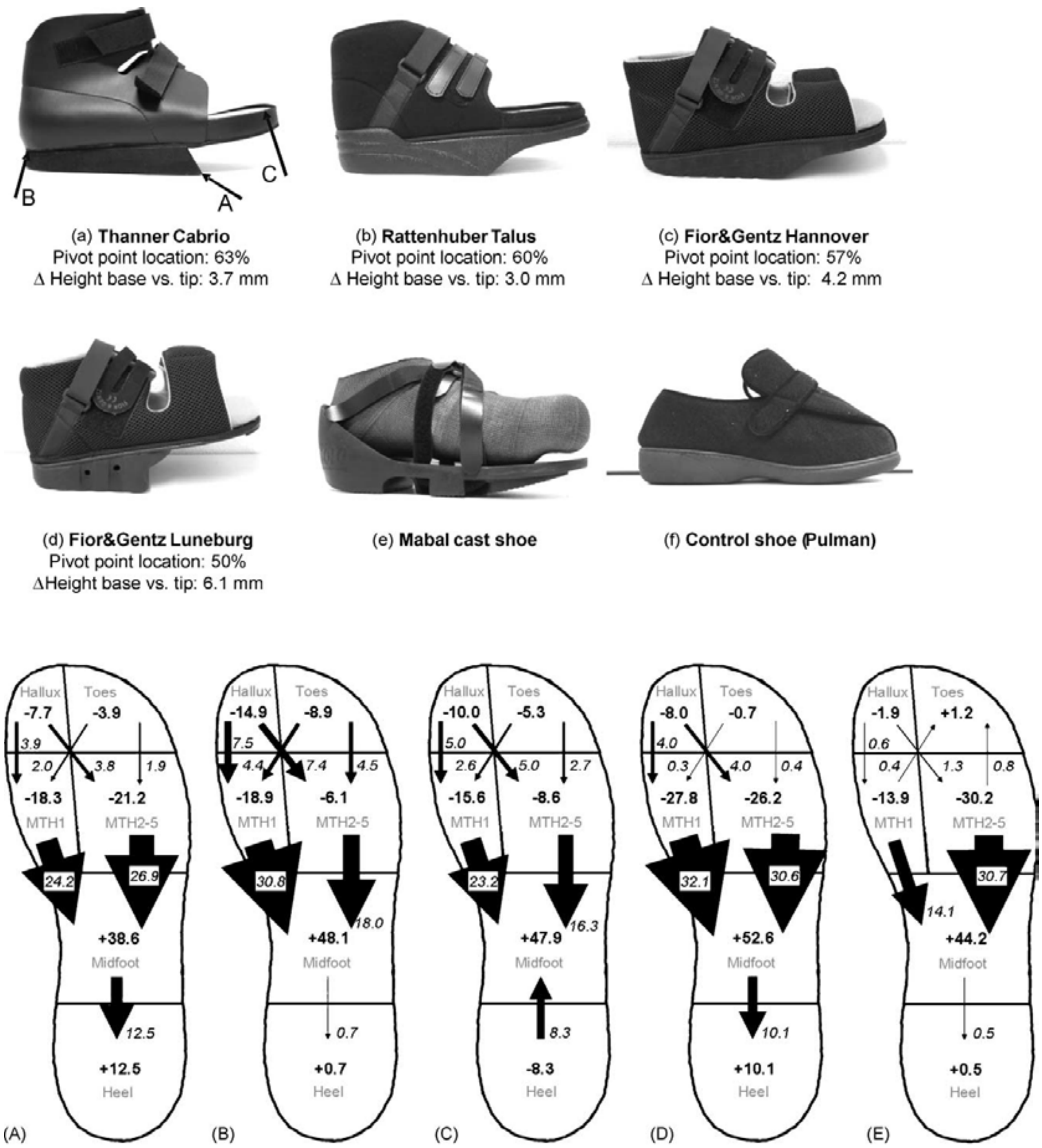

Figure 4.3.2 Forefoot offloading orthotics with dorsiflexion support and their effects on plantar pressure distribution. Number in arrow indicating percentage of pressure shift (Kao and Ferris, 2009).

Reduction in plantar pressure is essential for healing ulcers. A variety of dorsiflexion orthotics shift plantar pressure to the mid-foot region as shown in Figure 4.3.2 above (Bus et al., 2004; 2009) while plantarflexion was reported to increase concentration of plantar pressure at the metatarsal region (Eisenhardt et al., 1996).

In addition to offloading pressure from the common sites of ulcer development, reduction in impact forces at loading is important for preventing pain development especially in knee joints. Softer loading is achieved by absorbing impact as elastic energy in the extended Achilles tendon (Lichtwark and Wilson, 2006; Lidtke et al., 2010; Silver-Thorn et al., 2011; Ventura et al., 2011). Greater dorsiflexion at heel contact can be therefore advantageous in restoring shock as elastic energy and the effective release of this energy by ankle plantarflexion toward push-off provides more efficient energy transfer (Han et al., 1999). Furthermore, a dorsiflexed ankle at heel contact lengthens the time from heel contact to foot-flat compared to flatter contact. Following the force-time relationship, peak force relative to the impulse after heel contact essentially increases if the time to foot flat reduces. Based on the discussion here, greater dorsiflexion at heel contact may be useful in the impact reduction on knee joints and possibly ease knee joint pain due to osteoarthritis.

One concern with dorsiflexion supporting orthotics is the long-term effect of continuous extension, which can potentially cause impairment in the muscular strength of plantarflexors due to the dorsiflexion support feature (Kao and Ferris, 2009). Silver-Thorn et al. (2012) tested the effect of dorsiflexion and plantarflexion on spatio-temporal parameters and lower limb kinematics/kinetics, and based on their results, an additional dorsiflexion/plantarflexion up to 5° is unlikely to cause any negative effects on joint moments and spatio-temporal parameters. If gait kinematics and kinetics are maintained, it can be speculated that long-term dorsiflexion/plantarflexion not exceeding 5° may not affect muscular strength.

Eversion: Adequate eversion support can be hypothesised to improve ML balance by assisting medial redirection of CoM and preventing centre of pressure (CoP) lateral displacement from exceeding the range necessary for ML stability (Bus et al., 2009; Corriveau et al., 2000; Han et al., 1999; Has et al., 2004; Pandy et al., 2010; Rietdyk et al., 1999; Shankthikumar et al., 2010). Inversion sprain (Figure 4.3.3 top) is difficult to recover and can damage ankle ligaments, often directly leading to lateral balance loss. Inversion sprain occurs more frequently in inversed ankle with unexpectedly strong onset of CoP on the lateral area of the bottom of the foot (Willems et al., 2005). Lateral CoP deviation can also cause loss of effective afferent feedback because mechanoreceptors are concentrated along the typical CoP path (Nurse and Nigg, 2001).

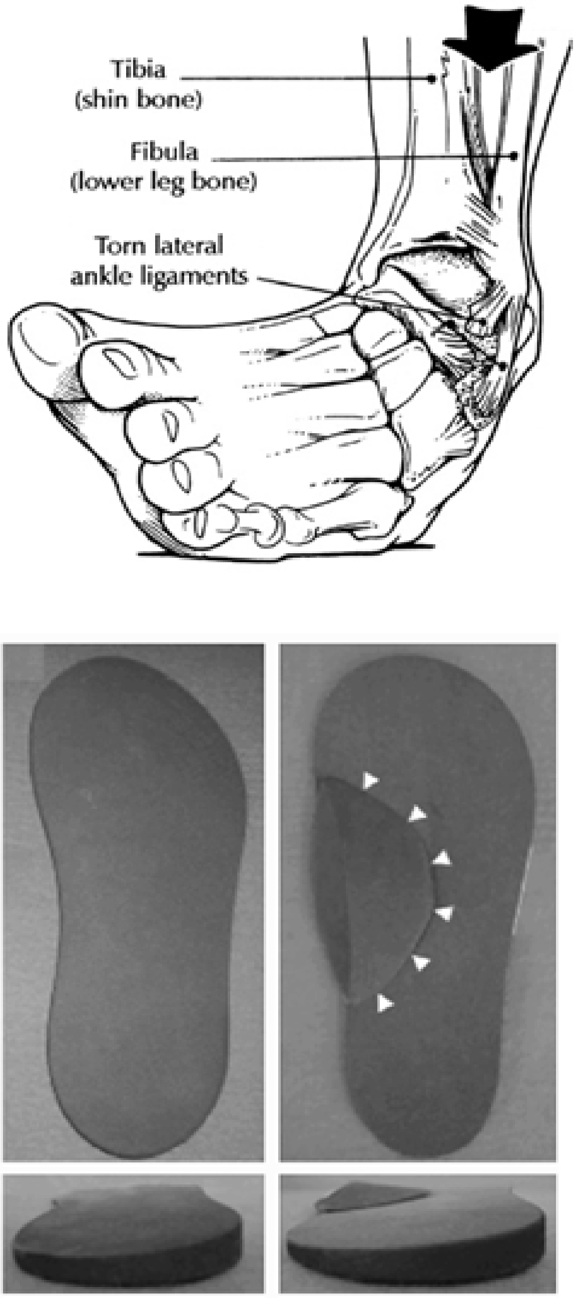

Figure 4.3.3 (Top) illustration of inversion sprain; (bottom) eversion (lateral wedge) insole (6°) with (right) and without (left) medial arch-support (Nakajima et al., 2009)

Assisting eversion can reduce the risk of inversion sprain by limiting the lateral CoP displacement. As visualised in Figure 3.1.2, CoP trajectory initiates at the heel and ends at the toe with a slight lateral curve up to the plantar area relative to the straight toe-heel line. The degree of the lateral curve reflects a risk of inversion sprain (Willems et al., 2005). Eversion support is expected to offload lateral weight concentration, possibly shifting CoP trajectory more medially. No study has investigated this concept to date but it is worthwhile observing how eversion support could change CoP trajectory. Previous studies of shoe insoles with eversion features have tested eversion angles (5˚-15˚) but 5˚ has been most commonly used (Radzimski et al., 2012).

As in dorsiflexion, ankle eversion is designed to absorb the impact during the early loading response (Han et al., 1999; Silver-Thorn et al., 2011). Mechanical energy efficiency during loading response can quantify the rate of impact force oscillated through stance for swing initiation. Some part of mechanical energy not used for swing initiation can be considered to be absorbed into shank segment as vibration that damages knee (Lidtke et al., 2010). Efficiency of mechanical energy transfer from heel contact throughout loading response can be calculated by recovery rate (Figure 3.3.2) and the computation method described in Chapter 6 (equation 24).

Dorsiflexion at heel contact facilitates knee flexion and both of these lower joint motions help reduce impact on knees. Based on previous studies (Nakajima et al., 2009; Radzimski et al., 2012), an additional 6% eversion could reduce the average knee adduction moment during the stance phase by 7.7% and with arch-support, 13.3%. Knee adduction moment has been recognised as the kinetic component of gait that is linked to knee osteoarthritis. Adequate dorsiflexion and eversion supporting functions are thus evidently effective in preventing knee pain. Nearly half of the older population was found to have progressive knee osteoarthritis (Murphy et al., 2008), and for this reason, footwear designs to reduce knee adduction moment is demanded (Rafiaee and Karimi, 2012; Kerrigan et al., 2002; Nakajima et al., 2009; Toda and Tsukimura, 2004).

4.3.2 Increasing Contact Between Foot and Insole

Many custom-made shoe insoles are moulded based on each individual’s foot shape. Increased contact area between a foot and insole is effective for pressure distribution. Arch support is especially the common function for many custom-made shoe insoles and promotes pressure distribution. Prevention of ulcer development and also joint pains can be expected by sufficient pressure distribution. Custom-made is the preferred method to mould individual's foot but more cost effective solution can be application of semi-custom moulded shoe-insole by applying cryogenic freezing or heat-moulding technology.

Inner material of shoe-insole determines its quality such as elasticity, density, or resilience. Ethylene vinyl acetate (EVA) foam has been popularly used for its wider range of density (recommended to be 300-400m/s^3) ideal for semi-custom moulding of shoe-insole production (Crabtree et al., 2009).

4.3.3 Stimulating Cutaneous Receptors

An insole provides the interface on which a foot is directly placed. Modifications to an insole surface can possibly promote effective afferent feedback. Despite various positive effects by dorsiflexion support as described, dorsiflexion reduces plantar pressure and therefore could impair sound afferent feedback although offloading of plantar pressure from the metatarsal regions is important for ulcer prevention (Menz et al., 2005; Nurse et al., 2001; 2005). Thus insole surface modifications to enhance afferent feedback can be particularly worth consideration for dorsiflexion supporting insole.

Priplata et al. (2003) tested the effect of vibrating insoles on balance control by stimulating the plantar area and older adults demonstrated great improvement in dynamic balance. Lafond et al. (2004) have, however, questioned the methodology of this study such as using only single reflective marker to estimate the movement of head-arm-trunk segment for balance assessment. Because of this counterargument, the validity of this effect is still unknown, yet supporting the possibility that stimulation on plantar surface area may improve dynamic balance.

Bancroft et al. (2011) and Nurse et al. (2005) suggested the possibility of textured insole to enhance proprioceptive reaction. Addition of texture around plantar surface area may assist in providing more tactile sensation. When wearing textured insoles (Figure 4.3.4; right), reduced muscular work (soleus and tibialis anterior) and longer time to peak impact were found (Nurse et al., 2005). Longer time to peak impact is advantageous in allowing more time for feed-forward adaptations and considered beneficial for balance maintenance, although the study also stated that the overall effect of textured insole on the risk of falls is still not clearly understood. It is, however, revealed that textures can stimulate cutaneous receptors and acuteness of pressure can be controlled by changing the material, shape or spacing of each texture.

The balance enhancing function of insole relies on cutaneous stimulation on the bottom of the foot. Sole Sensor (U.S. patent issued in 2001, licensed to Hart Modbility, Inc) is a science-based, commercially available shoe insole that takes advantage of cutaneous sensation to improve reaction speed (Maki et al., 2008). A 3mm diameter plastic tube is placed peripherally except the front, such that cutaneous stimulation occurs only when CoP reaches the BoS boundaries (Figure 4.3.4).

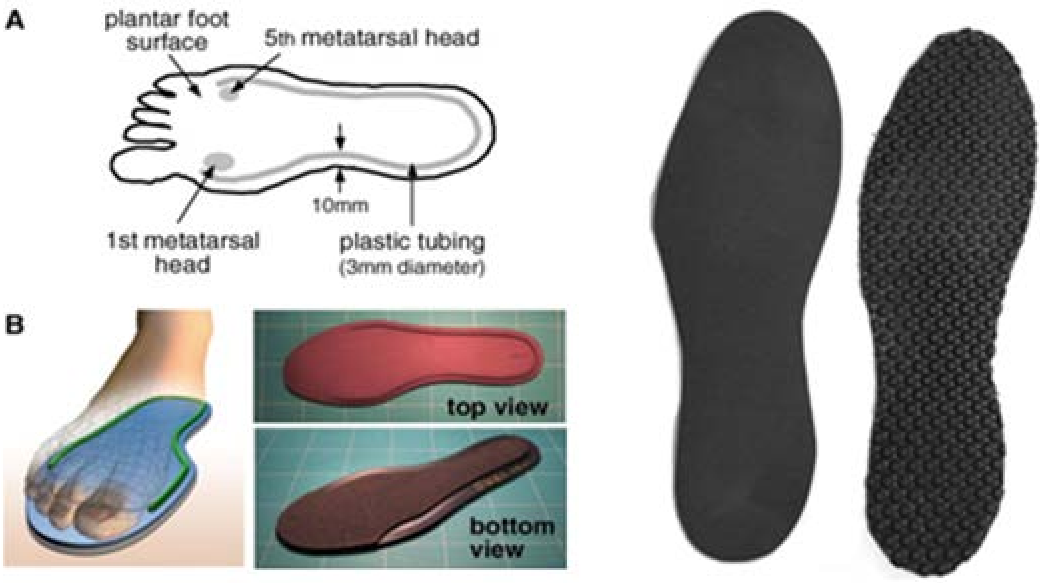

Figure 4.3.4 (Left) SoleSensor; (Right) textured Insole, 8mm from one texture to the next (adapted from Nurse et al., 2005).

4.4 Shoe Insole Tested in the Current Project

In this project insole effects on tripping, slipping and lateral balance loss will be investigated. As discussed above, dorsiflexion was expected to increase swing foot-ground clearance and reduce the likelihood of the swing foot contacting an obstacle. It was therefore hypothesised that the insole would increase MFC_h due to increased dorsiflexion. Eversion helps prevent lateral CoP deviation and increase MLM_d. It was therefore hypothesised that the insole would improve ML balance. Insole effects on CoM medial acceleration and MLM_t have been also reported. Combination of dorsiflexion and eversion are, as described, important ankle joint motions to enhance natural loading response, which will reduce impact on knees, reflected in knee adduction moment. Time to foot flat has been also obtained to indicate whether force is adequately distributed over a certain period of time.

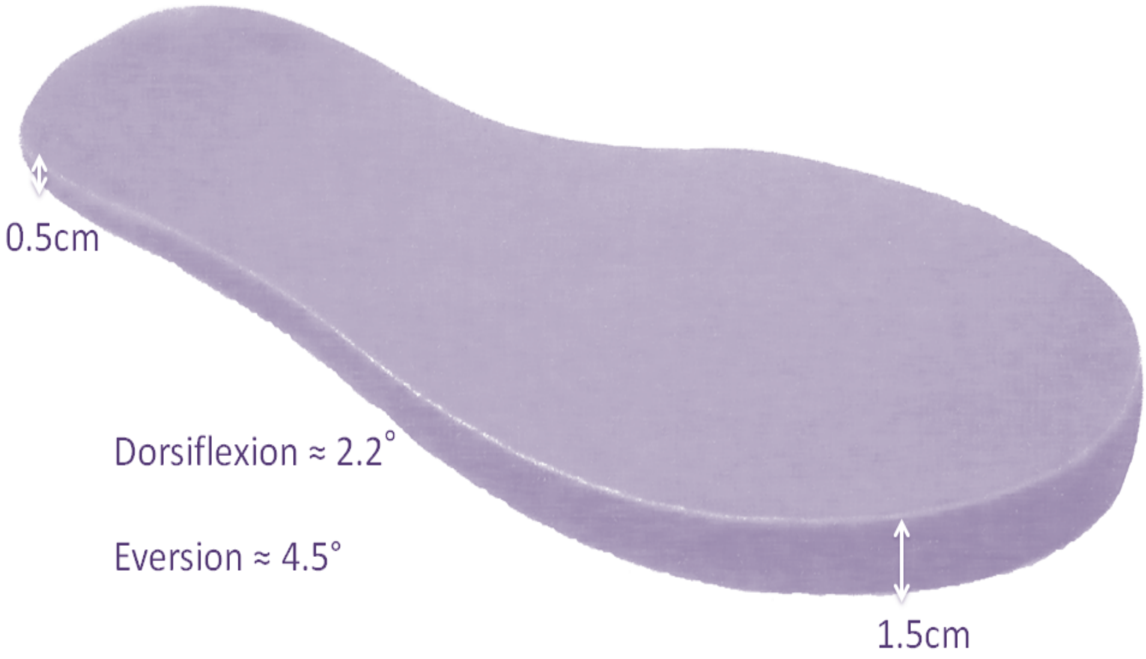

Figure 4.4.1 Specification of the insole (26cm) tested in the current study. Length 26cm, width at heel 6cm. Inner heel at zero height. Inclination by dorsiflexion and eversion maintained on the entire insole surface.

The specification of the insole used in the current experiment was to support 2.2 degrees of dorsiflexion and 4.5 degrees of eversion by providing inclination of the insole interface (Figure 4.4.1). The insole surface was maintained flat and the inner material was sufficiently durable to maintain the same inclination. Relative to the inner part of heel, the most outer edge was 0.5cm higher and this eversion angle was applied across the entire insole surface. The most anterior inner surface was elevated relative to the inner heel by 1.0cm and the highest section was the most lateral and anterior toe section (1.5cm). Four insole sizes were available to match the participants' shoes.

4.4.1 Mechanical Energy Recovery

To quantify efficient transfer of mechanical energy during loading response, recovery rate can be used (Figure 3.3.2). Double support phase is described as step-to-step transition of CoM and mechanical energy cost is necessary to produce GRF to link the CoM of one IP cycle with the next IP cycle (Kuo, 2007). Recovery rate is the established method of calculating the energy transferred to oscillate loading response from heel contact to toe-off (Cavagna et al., 1976; Collett et al., 2007; Vereecke et al., 2006; Ortega and Farley, 2003; Detremblur et al., 2005; Schepens et al., 2004).

Recovery rate can quantify effects of dorsiflexion and eversion support on efficient use of mechanical energy. Greater mechanical energy efficiency can be expected in the tested insole. More efficient loading response is only possible by utilising shock at heel contact to initiate toe-off without dispersing mechanical energy transferred to other lower limb joints (i.e. knee). It is therefore considered that higher recovery rate is advantageous in reducing impact generated at heel contact.

Chapter 5

Research Questions and Hypotheses

Ageing Effects on Dynamic Balance

The primary research question addressed in this research was how ageing influences balance during gait. Age-associated gait adaptations can be classified as due to either safety adaptations or age-related impairment. Safety-related adaptations can be generally described as more ‘conservative’ or ‘cautious’ gait styles that tend to be accentuated in challenging walking conditions (Nagano et al., 2012). This type of gait adaptation is employed voluntarily by older adults and may preserve balance but at a cost of, for example, increased energy cost and slower walking. Another age-related gait changes reflect impaired balance maintaining ability such as inconsistency of gait over multiple gait cycles and also between the two lower limbs’ motions. Increased variability and asymmetry in gait parameters are generally expected in the older population. In relation to step width manipulations, relative to preferred width both 50% narrower and wider width walking would be perceived as ‘unfamiliar’ walking conditions, possibly leading to both types of gait adaptation. Wider step walking was expected to be more advantageous than narrow walking in terms of balance control (Whittle, 2007).

Specific hypotheses for the effects of ageing and step width are outlined here for various gait parameters. From previous research (Whittle, 2007; Winter, 1991), older adults were hypothesised to show typical ageing effects on spatio-temporal parameters including slower gait velocity due to shorter step length, enlarged step width and prolonged double support time. These effects were expected to appear clearly under width-altered walking conditions especially in narrower walking and also in wider walking to a lesser degree.

Foot-ground kinetics at braking and propulsive phases (Figure 2.2.3) can be expected to influence gait velocity and associated step length, because ground reaction forces (GRF) influence both propulsion and braking. Older adults were expected to reduce horizontal GRF and as a consequence, reduce RCoF. In other words, a lower slipping risk can be anticipated for older adults. The hypothesis concerned with heel contact dynamics and associated anterior heel slipping potential after heel contact is that due to reduced dorsiflexion prior to heel contact (Nagano et al., 2011; Perry et al., 2007), lower (flatter) foot contact angle can be expected as an ageing effect. Reduced step length would, therefore, be expected to elicit a more conservative swing limb forward motion and foot placement closer to the CoM. The lower limb joints would be then aligned more vertically increasing the shank-floor contact angle (Section 2.7.3).

In terms of minimum foot clearance (MFC) characteristics, no ageing effect on height (MFC_h) was expected based on the previous investigations (Begg et al., 2007; Nagano et al., 2011). Asymmetrically greater non-dominant MFC_h is, however, expected in older adults as a safety adaptation (Nagano et al., 2011). A further safety adaptation is MFC timing (MFC_t). For controlled-width walking in older adults, to avoid the swing foot being located parallel to the stance foot, MFC(t) was expected not too be consistent with the previously reported timing at 50% swing (e.g. Winter, 1991). Safety adaptations for minimum lateral margin (MLM) include greater MLM_d, smaller lateral CoP displacement and the higher medial CoM acceleration (Section 2.8.) These MLM characteristics would be seen for older adults especially under challenging controlled step width conditions.

When forming a hypothesis about the fundamental ageing effect on safety zone shape, it is useful to refer to spatio-temporal parameters. During double support, step length and width approximately represents AP and ML length of the safety zone, respectively. Shorter step length and wider width are typical in older adults' gait relative to young adults (Whittle, 2007; Winter, 1991). It is therefore possible that the general safety zone shape of older adults to be shorter in AP but wider in ML length compared to young adults. In terms of age-related deficits in balance control a variable and asymmetrical safety zone shape was expected. Increased safety zone variability was anticipated due to inconsistent foot placement or/and the less stable CoM control in the transverse plane. While reduced variability in step length and step width reflect more consistent foot placement, if still high variability in the CoM-foot distance is observed, the reduced CoM control is attributable. Such predictions are adequate during narrow walking where the lines may assist lower SD in step width while narrower Safety Zone negatively affects the ML CoM control.

Age-related loss of balance should be reflected in the safety zone parameters, possibly reflected in greater SD in the CoM-foot length. Ageing has been recognised important when comparing the non-dominant to the dominant limb (Nagano et al., 2011; 2012). Based on previous findings age-related gait asymmetry may be related to the reduced burden on non-dominant gait control compared to the dominant limb (Nagano et al., 2011; 2012). It can therefore, be hypothesised that older adults may exhibit asymmetry in the safety zone which could possibly offload the balance maintaining burden from the relatively weaker non-dominant limb.

Effects of Shoe-insoles to Enhance Safe Walking

In regards to the tested shoe-insole, dorsiflexion support of 2.2˚ is unlikely to affect spatio-temporal parameters according to the previous study (Silver-Thorn et al., 2012). No previous studies on eversion supporting function up to 15˚ have focused on effects on spatio-temporal parameters. On the assumption that the previous studies did not inspect effects of eversion on spatio-temporal parameters because of lack of expectation that eversion can significantly change spatio-temporal parameters, the current project can hypothesise that 4.5˚ of eversion support may not affect spatio-temporal parameters.

The purpose of dorsiflexion support in relation to balance is to elevate MFC_h and prevent swing foot from tripping (Moosabhoy and Gard, 2006). Although it is unknown how much dorsiflexion support can be obtained at MFC by adding 2.2˚ support during static standing, roughly 0.3cm increase in MFC_h per every one degree of dorsiflexion could be expected (Moosabhoy and Gard, 2006). Eversion would be useful in regulating lateral CoP displacement and promoting the medial CoM redirection. As improved ML balance, less CoP displacement, greater MLM_d and the increased medial CoM acceleration can be possibly seen if the insole could reduce the risk of lateral balance loss.

As described in Chapter 4, both dorsiflexion and eversion are known as ankle joint motions that enhance energy efficient loading response following heel contact and consequently, reduction in impact on knees could be expected. Greater recovery rate and lower knee adduction moment can be therefore predicted. Such adaptations would be attained by increased dorsiflexion and knee flexion joint motions and also prolonged time to foot flat from heel contact.

Chapter 6

Research Methods

6.1 Participants

The participants included 30 healthy young male (18-35 yrs) and 26 healthy older male adults (> 60 yrs) (Calisaya et al., 2010; Lord et al., 1996). The sample size for the project was controlled to exceed the targeted power of 0.8 with effect size of 0.8 (Thomas et al., 2005; Appendix II and VIII). Physical characteristics were height (young: 1.77 ± .06m, older: 1.74 ± .07m) and mass (young: 75.7 ± 3.5kg, older: 76.7 ± 7.9kg). Height and body mass were not statistically different, while the mean ages of the two groups were significantly different ($F = 1027.5$, $p = .000$). Five from the young and four from the older groups were classified as left limb-dominant, determined by the procedure used by Seeley et al. (2008), in which the dominant limb was defined as the preferred foot in kicking a ball. All participants were free from any pathological conditions listed in the Informed Consent (Appendix III) and Questions to Participants (Appendix IV). Older participants were limited to vigorous and healthy individuals who maintained independent lifestyles and were capable of walking actively for 30 minutes or more without a break confirmed by general health survey (Appendix V). They also reported no falls in the past two years and no traumatic injuries that would affect their gait. Young participants were university community volunteers while the older group was recruited through advertisements in a local newsletter (Appendix I). During the initial telephone screening, volunteers were confirmed to have a shoe-size between 26cm and 29cm and the same shoes were provided to eliminate any potential shoe effects on gait. Various sizes of shoe insole matched the shoe sizes. The Human Research Ethics Committee, Victoria University, approved the experimental protocol (Appendix II).

6.2 Apparatus and Procedure

Three Optotrak (Optotrak®, NDI, Canada) cameras and two force plates (AMTI) were set up as illustrated (Figure 6.2.1, A). Force data were collected at 1000Hz. Joint anatomical landmarks were tracked at 100Hz (Figure 6.2.1, B). Prior to collecting walking data, static trials in anatomical position were taken for 5 seconds, later used for modelling to estimate joint kinematics and kinetics. Body CoM was estimated following the traditional model (e.g. Gutierrez-Farewik et al., 2006; Lee and Chou, 2006). For safety zone analysis, the pelvis segment CoM motion was modelled by anterior superior iliac crests, posterior inferior iliac spines and greater trochanters (Kingma et al., 1995). The transverse motion of pelvis segment centre of mass (CoM) has been employed for balance assessment for normal gait and useful for force couple simulation based on the inverted pendulum model as described in Figure 3.4.1 (Kingma et al., 1995; Kuo, 2007; Mengi et al., 2011; Rabuffetti and Baroni, 1999).

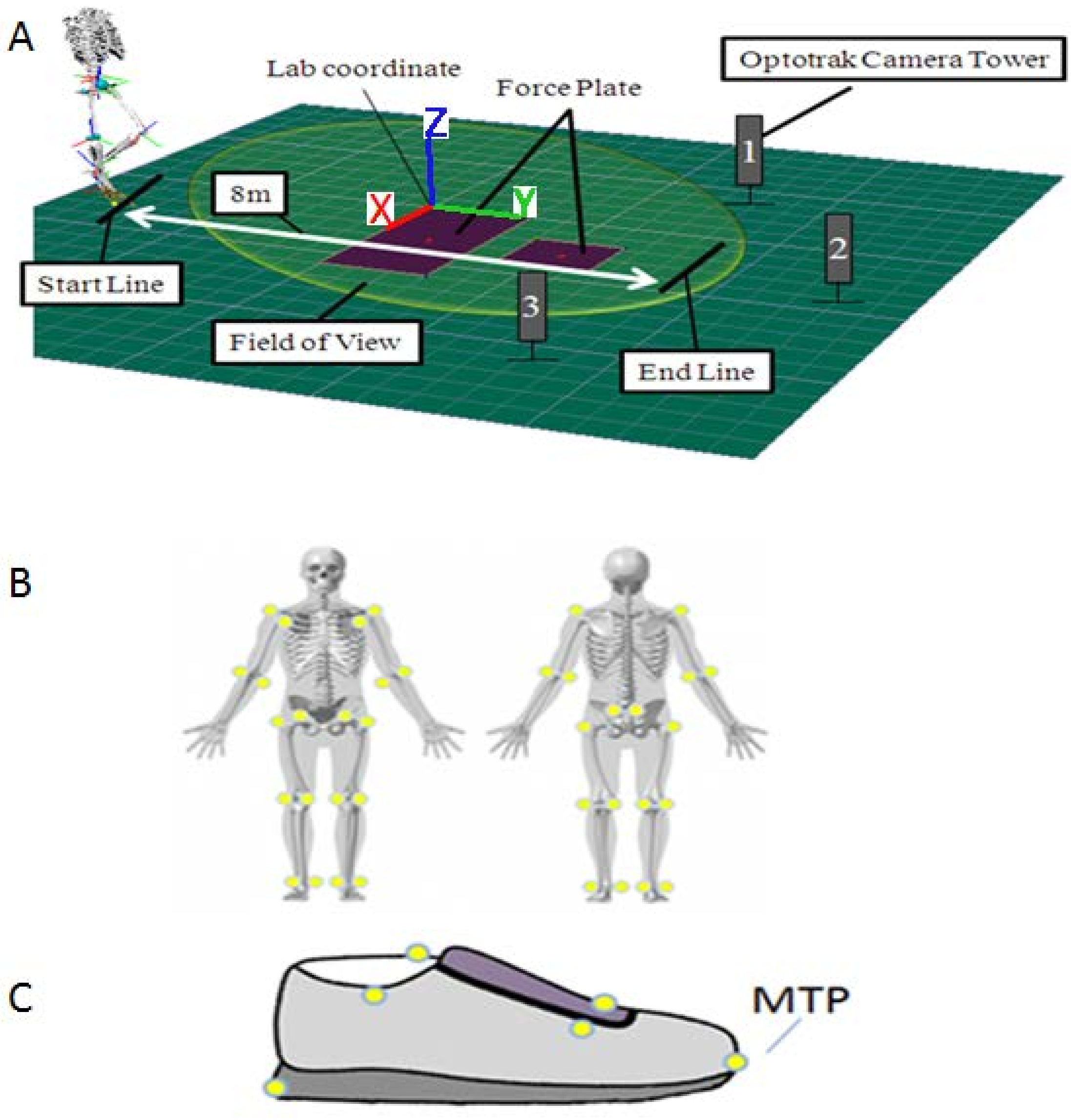

Figure 6.2.1 (A) laboratory setup and coordinate system. Anterior walking direction (Y+) and the right perpendicular to the walking direction (X+;. (B) Tracked IRED markers on joint landmarks for kinematic and kinetic data collection; (C) Foot modelling: heel, lateral and medial malleolus, 2nd and 5th metatarsal heads, MTP = minimum toe point, used for measurement of MFC (Begg et al., 2007)

A lower body skeleton model was developed using Visual 3D as shown in Figure

6.2.2. The femur was based on the locations of the greater trochanter, quardrate tubercle, lateral and medial epicondyle. The shank was defined by the lateral and medial condyles and lateral and medial malleolus of tibia. The foot complex was built using the heel and the 2nd & 5th metatarsal heads, toe and lateral and medial malleolus. The thorax was modelled on the anterior superior iliac crests and posterior inferior iliac spines of the pelvis segment and greater tubercle and lesser tubercle of both humerus bones.

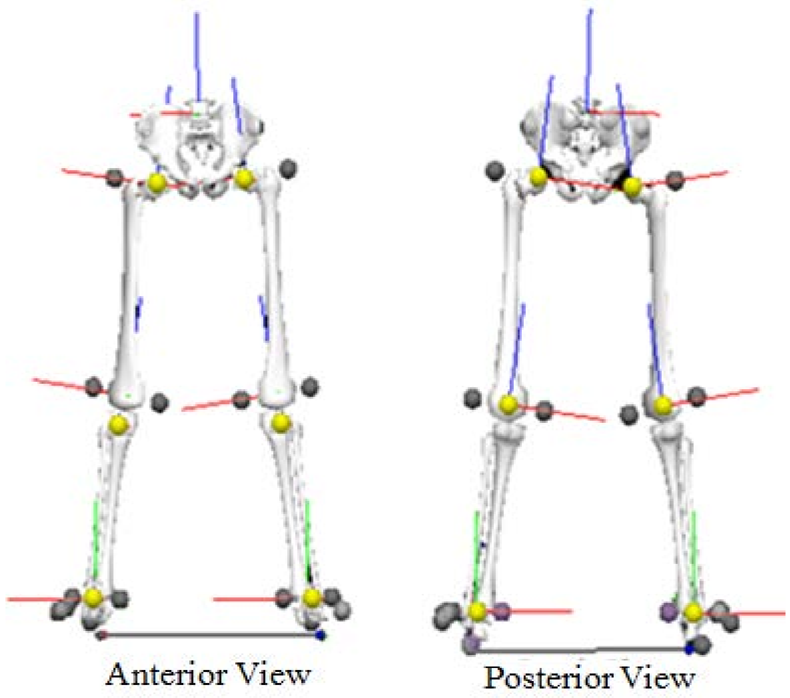

Figure 6.2.2 Image of imarinary markers registration of the lower body. Hip (lateral surface on both anterior and posterior iliac spine), knee (medial & lateral knee and patella), and foot (medial & lateral malleolus, 2nd & 5th metatarsal phalanges, minimum toe point, and heel). Yellow markers indicate the joint centres and gray markers indicate registered imaginary markers.

The four experimental conditions were preferred speed unconstrained walking, narrow step walking, wide step walking and insole walking with preferred step width. All participants began with unconstrained walking, in which they walked straight at preferred speed along the 8m walkway (Figure 6.2.1 A). The following conditions (wide, narrow and insole) were performed in a randomised order. Narrow and wide walking conditions were ±50% relative to mean step width in unconstrained walking (Figure 6.2.3). In both width-controlled walking conditions, two parallel tapes were attached to the walkway to indicate the target step width. The experimenter demonstrated width-controlled walking prior to the participant's trials to clarify the procedure.

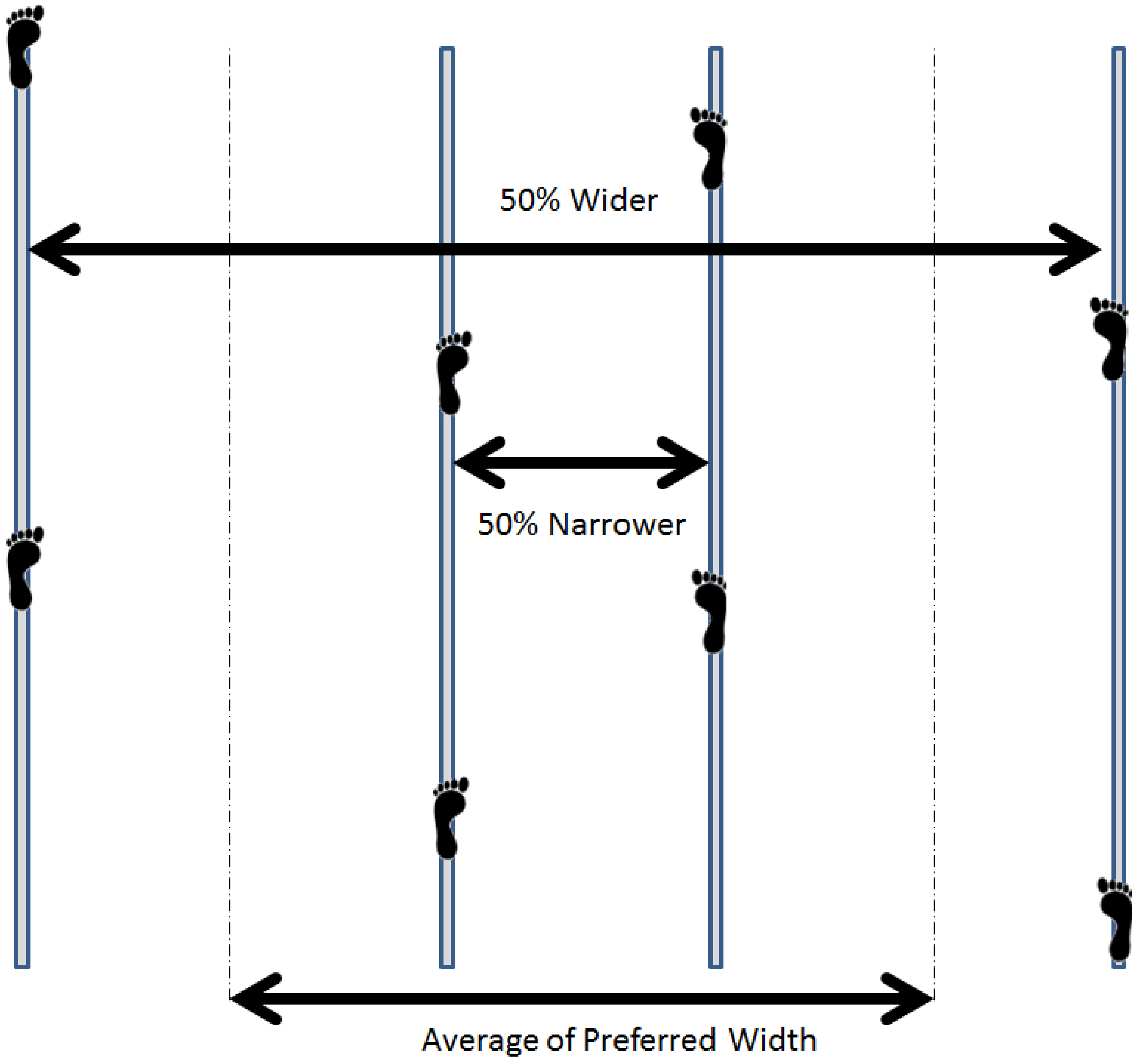

Figure 6.2.3 Image of width control. Average of preferred width ±50% narrower/wider walking tasks were controlled by indicating target step width.

Subjects were asked to contact the tape with their heels. Insole walking trials were conducted in exactly the same way as preferred walking. There were 30 trials in each condition and two or three complete gait cycles from both limbs were sampled on each trial (i.e. a total of 60 to 90 stride cycles per subject). A start position was determined such that at preferred walking speed either foot would strike the first plate and the other foot would land completely on the further (second) force plate.

Obtained raw data were first interpolated to compensate any occluded signals using a window of up to 10 frames (0.1s). A 4th order zero-lag Butterworth Filter with a cut-off frequency of 15 Hz was then applied to kinematic data (Nagano et al., 2011; 2012). To determine the gait cycle phases and obtain spatio-temporal parameters, toe off and heel contact were identified by applying the kinematic conventions (Fusco et al., 2008; O'Connor et al., 2007; Osaki et al., 2007), validated using GRF-based definitions with a vertical threshold of 5N for heel contact and toe-off as detailed below.

6.3 Definitions of Examined Gait Variables

This section operationally defines the gait variables employed in the Thesis.

6.3.1 Toe-off and heel contact

To obtain multiple gait cycles from a single walking trial, both toe-off and heel contact were defined based on kinematic data (O'Connor et al., 2007).

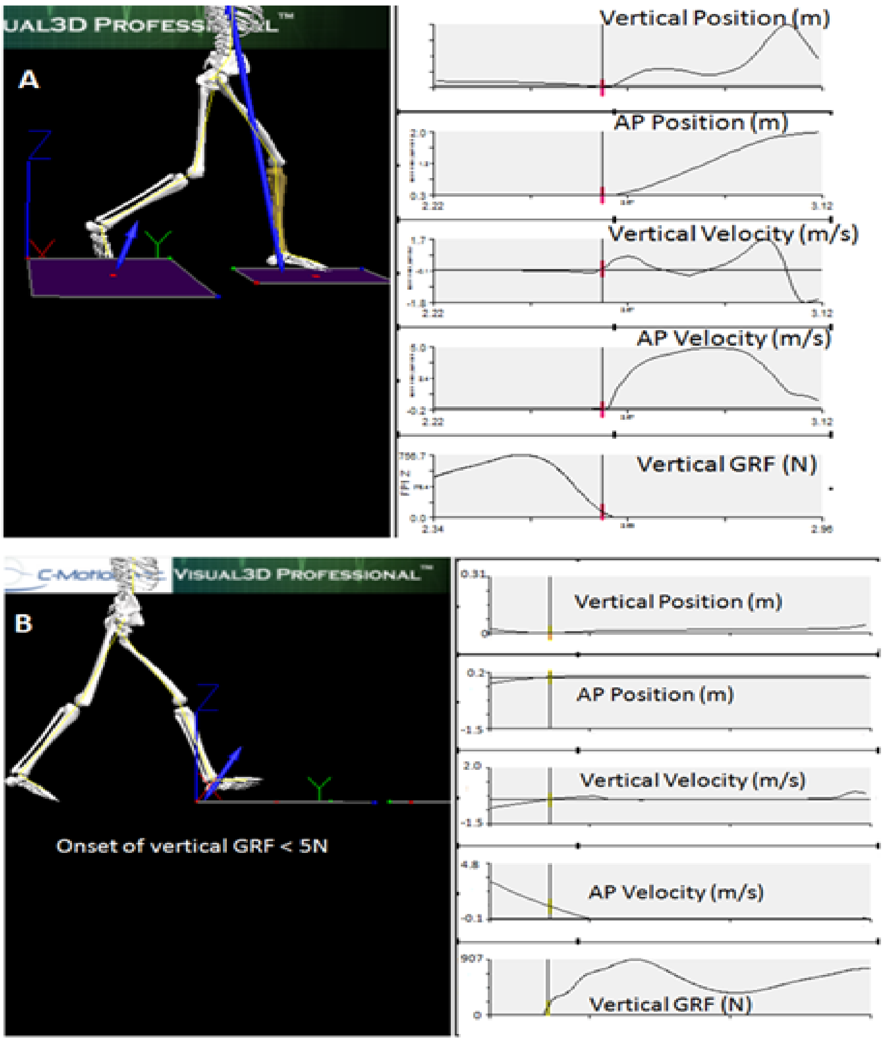

Figure 6.3.1 Definitions of (A) toe-off and (B) heel contact.(A)Toe marker kinematics and vertical GRF at toe-off. (B) Heel marker kinematics and vertical GRF at heel contact. The vertical line corresponds to toe-off as illustrated in the left image.

From the sagittal plane analysis, at toe-off (A), minimum vertical position is accompanied with zero anterior-posterior (AP) and medio-lateral (ML) velocity. In contrast, the minimum vertical heel with zero vertical velocity were characterised at heel contact. These events were selected based on a previous research in which a vertical ground reaction force (GRF) of 5N was the threshold for both events (McGorry et al., 2010).

6.3.2 Spatio-Temporal Gait Parameters

The stride cycle was defined as heel contact to the next heel contact of the same foot. One gait cycle has been divided into single and double support phases. Dominance of spatio-temporal parameters is based on the lead limb (Figure 6.3.2).

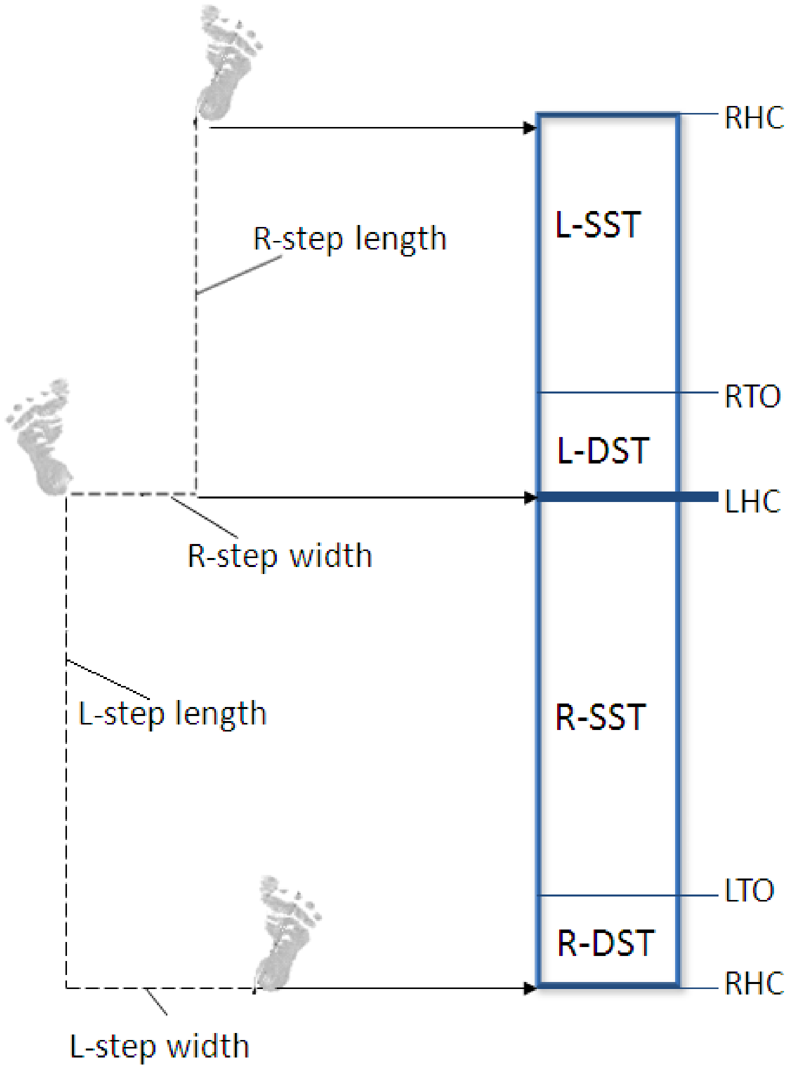

Figure 6.3.2 Illustration of spatio-temporal parameters. SST = single support time, DST = double support time, HC = heel contact, TO = toe-off.

1) Step length and width: displacements between two consecutive heel contacts in AP and ML directions, respectively.
2) Double support time: time from heel contact of one limb to contralateral toe-off, when both feet are on the walking surface. Double support time was analysed as percentage relative to step time from heel contact to contralateral heel contact.
3) Step velocity: average horizontal velocity of foot from toe-off to heel contact

6.3.3 Minimum Foot Clearance (MFC)

Local minimum of vertical minimum toe point (MTP) during mid-swing has been defined as MFC (Figure 2.7.1). At MFC, height of MTP, MFC-height (MFC_h) and timing (%) in swing phase, MFC-time (MFC_t) have been described by median ± IQR (Begg et al., 2007).

6.3.4 Minimum Lateral Margin (MLM)

MLM is mid-swing phase event following MFC as illustrated in Figure 6.3.3.

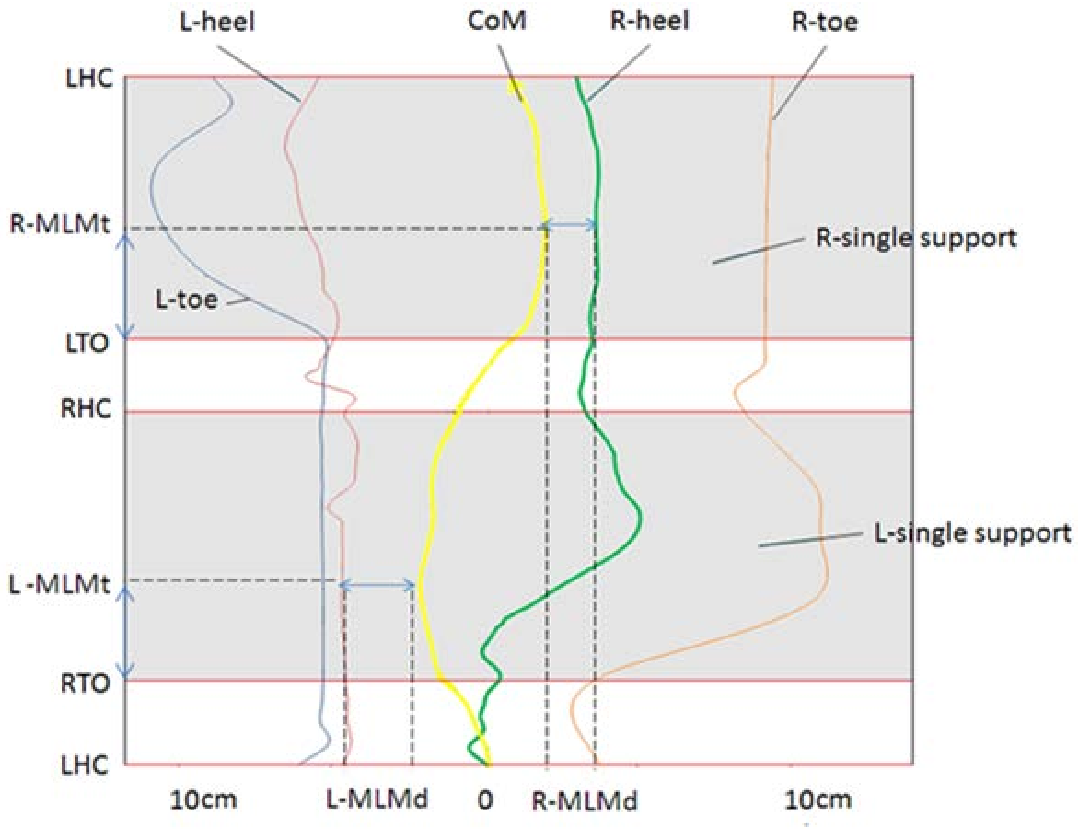

Figure 6.3.3 Locations of the CoM, heels and toes in the transverse plane. R- = right, L- = left. The average CoM location at toe-off = 0. HC = heel contact, TO = toe-off, MLM = minimum lateral margin.

Minimum lateral margin (MLM) was at shortest ML distance from stance heel to the CoM, confirmed by zero ML acceleration and velocity of the CoM (e.g. Perry et al., 2008; Åberg et al., 2010). As indicated, dominance of MLM follows the stance limb while dominance of MFC is, in contrast, determined by the swing foot.

6.3.5 Lateral CoP Displacement and Mean Medial CoM Acceleration

Time from toe-off to MLM was presumed as a critical period in terms of ML balance. Lateral CoP displacement and overall mean medial CoM acceleration in this period were reported.

6.3.6 RCoF1 and RCoF2

Characteristics of GRF in the three axes were obtained at RCoF1 and RCoF2 (Figure 2.7.3).

6.3.7 Heel Contact Dynamics

Heel contact angle was determined by the angle between shank (lateral knee to lateral malleolus) and the floor surface at heel contact (β in Figure 2.7.4). Foot contact angle (α in Figure 2.7.4) was the angle formed by floor and toe-heel line.

6.3.8 Lower Limb Joint Angles

Following the traditional definitions (Winter, 1991; Figure 6.3.4), knee and ankle angles were defined. Greater trochanter, lateral knee and lateral malleolus determined knee angle, while lateral knee, lateral malleolus, and MTP defined ankle angle. Greater knee angle indicates knee flexion and greater ankle angle indicates plantarflexion, respectively. In the comparison between non-insole and insole conditions, the lower limb joint angles were obtained at MFC and heel contact.

6.3.9 Knee Adduction Moment

Knee adduction moment was computed by inverse dynamics based on 3D modelling of lower limb joints. Rotational torque about y axis (Figure 6.3.4 right) was defined as knee adduction moment. Peak knee adduction moment during stance phase was also monitored.

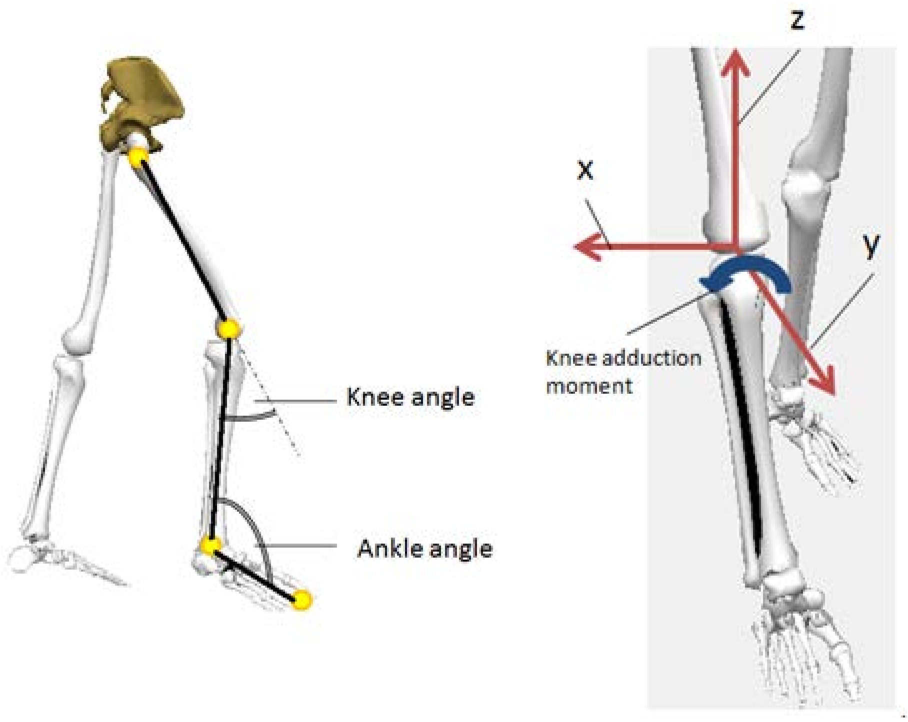

Figure 6.3.4 (Left) Definitions of knee and ankle angles. (Right) knee adduction moment

6.3.10 Time to Foot Flat

Foot flat was identified at the initial frame where the vertical MTP location after heel contact lowered to the height of static 'standing' trials, which were recorded for 3D modelling. Time was recorded from heel contact to foot flat (Mariani et al., 2012).

6.3.11 Recovery Rate (double support)

As described in Chapter 4 and Schepens et al. (2004), mechanical energy efficiency during loading response was measured by recovery rate and compared between non-insole and insole walking conditions (Figure 3.3.2).

$$\text{Recovery rate (\%)} = 100 * [\Delta\text{KE} + \Delta\text{PE} - \Delta(\text{KE} + \text{PE})]/(\Delta\text{KE} + \Delta\text{PE}) \qquad 24$$

6.3.12 Safety Zone

The Safety Zone was modelled by measuring the transverse positions of toes and heels from both feet relative to the CoM (0, 0) as illustrated in Figure 6.3.5.

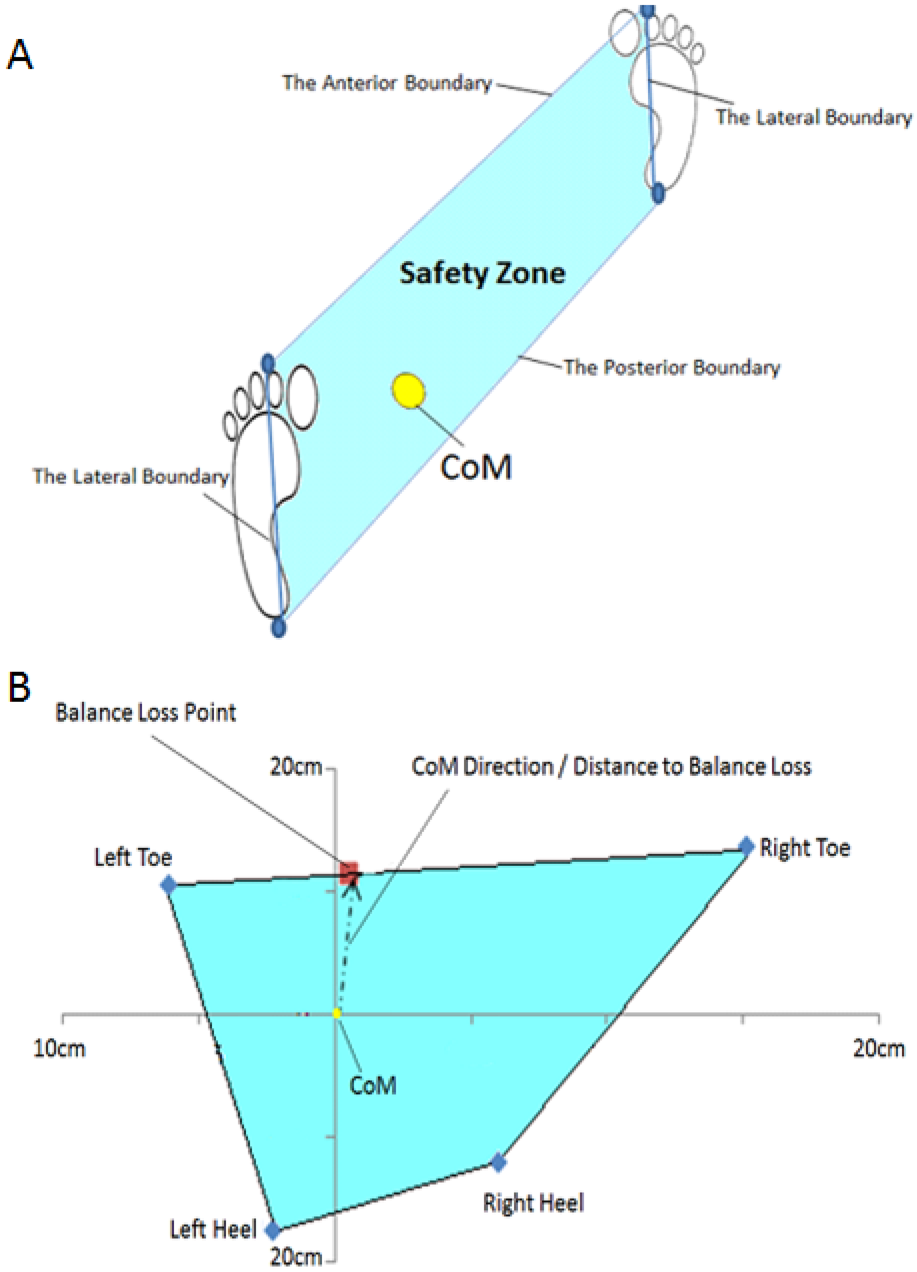

Figure 6.3.5 (A) Safety Zone and boundaries (B)Safety Zone based balance assessment at right swing MFC. The Safety Zone (coloured area); defined by the transverse positions of toes and heels of the both feet relative to the CoM (0, 0). The CoM direction based on the CoM transverse linear velocity. Distance to Balance Loss = distance to boundary of the Safety Zone based on the CoM direction.

The transverse CoM velocity was obtained in both ML and AP directions to estimate the direction of the CoM movement, which determines 'balance loss point' in Figure 6.3.5. Movement consistency of safety zone was characterised by both AP and ML variability of feet locations relative to the CoM (0, 0). Safety zone was visualised at toe-off, MFC, MLM and heel contact. Effects of ageing, limb dominance and step width on balance during swing phase were examined.

Balance loss simulation was attempted, using the methods explained in Chapter 3. Empirical gait data of both the young and older groups' preferred width walking were used as inputs of gait simulation to compute available response time (ART), defined as time taken from the current CoM location to either boundary of the Safety Zone. Slipping subsequent to heel contact and before toe-off at the two RCoF peaks (Figure 2.7.3), tripping at MFC and lateral balance perturbation at MLM were separately modelled using computation procedures, and obtained results were outlined in Chapter 9.

6.4 Design and Analysis

The experimental designs, analyses and results are presented in the following four Chapters. Chapter 7 '*Ageing and step width effects on gait and balance'*, Chapter 8 *'Ageing and step width effects on the Safety Zone'*, Chapter 9 '*Gait data computation for balance simulation'*, and Chapter 10 '*Effects of insole on balance loss and injuries risks'*. For the statistical tests, P-values less than .05 were accepted as significant, while $p < .01$ was further distinguished to highlight the strong effect. All dependent variables were described by central tendency (i.e. mean or median) and also its dispersion (i.e. S.D. or IQR) to characterise movement consistency (Thomas et al., 2005; Nagano et al., 2012).

The first analysis was designed to investigate the effects of age-related step width manipulations on balance during gait (preferred vs. narrow vs. wide step width). A 2×3×2 (age × width x limb) repeated measures mixed model analysis of variance (ANOVA) design was applied to show the effects of ageing, step width and limb dominance. The dependent variables were 1) spatio-temporal parameters: step width, step velocity, step length and double support time, 2) MFC data: MFC_h and MFC_t, 3) slipping related kinetic variables: GRF and RCoF at the two RCoF peaks and kinematic variables at heel contact: foot contact angle and heel contact angle 4) ML balance: MLM_d, MLM_t, lateral CoP displacement and average medial CoM acceleration. Normality of these fundamental position-time and GRF based data were confirmed normally distributed (Appendix X).Correlation analysis was performed to investigate any interdependency between spatio-temporal parameters.

Safety Zone analysis employed the same 2×3×2 (age × width x limb) statistical design to investigate distance between CoM and both toes and heels separately in AP and ML axes. Safety Zone analysis was conducted at toe-off, MFC, MLM and heel contact for both dominant and non-dominant limbs' parameters.

The insole effects were determined by comparing non-insole (preferred width walking) and insole walking conditions using 2 x 2 x 2 (age x footwear x limb) repeated measures mixed model analysis of variance (ANOVA). Dependent variables were 1) spatio-temporal parameters: step width, step velocity, step length and double support time, 2) MFC data: MFC_h, MFC_t; ankle and knee angles at MFC, 3) foot contact angle and heel contact angles, 4) ML balance: MLM_d, MLM_t, lateral CoP displacement and average medial CoM acceleration, and 5) variables for loading response: recovery rate, time to foot flat, knee adduction moment; ankle and knee angles at heel contact.

CHAPTER 7

Ageing and Step Widths Effects on Gait and Balance

7.1 Overview

Falls among older adults are largely due to age-related impairment in dynamic balance (Shkuratova et al., 2004). Older adults attempt to compensate impaired balance by increasing step width, and therefore, manipulation of step width was expected to influence older adults' gait, therefore older adults demonstrate safety-related gait adaptations in challenging walking conditions (Nagano et al., 2012; Chamberlin et al., 2005). In particular, 50% narrower walking was predicted to impose challenges on the balance of older adults due to narrower safety zone (Hurt et al., 2010; Pandy et al., 2010; Shkuratova et al., 2004), and safety gait adaptations such as slowing down to ensure balance were expected. Wide walking could provide functional benefits to ML balance (Whittle, 2007; Ko et al., 2007) although step width wider than preferred could also disturb most natural walking styles, possibly leading to slowing down and associated changes in spatio-temporal parameters and other gait variables.

Reduction in step velocity due to lower step length, prolonged double support time and larger step width are the main effects due to ageing (Whittle, 2007). These ageing effects are not only due to age-associated functional declines in gait performance, but also safety adaptations (Nagano et al., 2012; Chamberlin et al., 2005; Shkuratova et al., 2004). When walking with controlled widths, older adults were therefore hypothesised to accentuate these typical ageing effects as they would possibly require more intentional control of the CoM to be located within the Safety Zone during width controlled walking conditions. For detailed investigation of ageing and step width effects during relatively vulnerable single support, kinetics of both the swing initiating and terminating events, toe-off and heel contact respectively, were first analysed.

Both toe-off and heel contact define the swing phase with ground reaction force (GRF) kinetic inputs. Slower gait was considered to be achieved by smaller GRF inputs at toe-off and heel contact, possibly detected in the older population especially during width controlled walking (Marasovic et al., 2009; Simmonds et al., 2012). GRF kinetic characteristics were examined for assessing slipping risk at the two RCoF peaks (Figure

2.7.3), heel anterior slipping after heel contact and forefoot posterior slipping due to push-off force, respectively (Chang et al., 2012). Lower horizontal GRF relatively decreases RCoF. Older adults were, therefore, hypothesised to elicit less risk of slipping at toe-off and heel contact. In relation to anterior heel slipping, smaller foot contact angle and larger heel contact angle were advantageous in reducing RCoF1 following heel contact (Brady et al., 2000; McGorry et al., 2008). Such gait characteristics were accordingly anticipated in the older adults especially during width controlled walking conditions if older adults' gait is less prone to slipping

To assess tripping and ML balance during single support, minimum foot clearance (MFC) and minimum lateral margin (MLM) were examined. Shorter MFC_h increases the chance of unexpected swing foot contact, but ageing alone has never been reported as decreasing MFC_h. From previous studies (Nagano et al., 2011; Wass et al., 2005), increased MFC_h in the non-dominant limb relative to the dominant limb was anticipated to emerge when older adults walked under more challenging walking conditions i.e. in different step width conditions. Timing of MFC (MFC_t) has been reported to be around halfway (≈ 50%) into the swing phase (Winter, 1991), but no previous investigations have concerned the effects of ageing, limb dominance or step width on MFC_t. For safety adaptation, older adults may alter MFC_t to avoid swing foot to be located next to stance foot at MFC.

While tripping and slipping cause AP balance perturbation, MLM can be considered the key event for understanding lateral balance loss because it represents the shortest lateral margin from the CoM to the stance foot (MLM_d). Little research has examined MLM before (e.g. Lugade et al., 2011), but in the current project we are focusing intensively on the period between toe-off and MLM to understand lateral balance perturbation. In addition to examined distance (MLM_d) and timing (MLM_t) of MLM, lateral CoP displacement and mean medial acceleration were examined in this period. Safe MLM characteristics were considered as greater MLM_d, less CoP displacement and greater medial acceleration (section 2.6). In terms of timing, having MLM_t and MFC_t simultaneously in the short time frame could increase the severity of balance loss during the mid-swing phase in case of balance perturbation. Having MLM_d distinct from MFC_t may therefore, secure the mid-swing phase period. If ageing impairs ML balance control, especially when their step width is manipulated, both conservative gait adaptations and intrinsic age-associated gait changes were expected to appear in older adults' gait. Conservative gait adaptations are hypothesised to appear as a result of attempting to secure ML balance, while greater variability in the parameters above or lower limb asymmetry could reflect functional impairments in ML balance control due to ageing (Montero-Odasso et al., 2011; Sadeghi et al., 2000; Sadeghi, 2003).

7.2 Results

A 2×3×2 (age × width x limb) repeated measures mixed model Analysis of Variance (ANOVA) was applied to show how ageing, step width and limb dominance would influence spatio-temporal gait descriptions and various balance hazards at toe-off, heel contact, MFC and MLM, described by mean ± S.D.

7.2.1 Spatio-temporal gait variables

Figure 7.2.1 displays step width effects on step parameters. The three step width conditions were statistically distinguished ($F (2, 53) = 272.0$, $p < .01$) as preferred (9.9cm), narrow (5.4cm) and wide (14.8cm), and width control closely matched the target 50%. Age-related increase in step width was by 1.3cm but statistically not differentiated. Step width variability in preferred width, narrow and wide walking was 3.0cm, 2.2cm and 2.1cm, respectively, such that SD reduced when step width was controlled ($F (2, 53) = 13.8$, $p < .01$).

Older adults walked with lower step velocity than their young counterparts ($F (1, 54) = 37.6$, $p < .01$). Step velocity was fastest in preferred width, followed by wide and slowest in narrow walking but these effects were observed only in older adults (age x width: $F (2, 53) = 18.7$, $p < .01$). A significant positive correlation between step length and step velocity ($r = .809$, $p < .01$) suggested interdependency between step velocity and length. Older adults walked with significantly shorter steps than young adults by 11.7cm ($F (1, 54) = 30.7$, $p < .01$). Older participants' longest steps in preferred width, shorten by about 4cm in wide and in narrow walking then was a further 4cm reduction, but step width did not affect young adults' step length (age x width: $F (2, 53) = 10.0$, $p < .01$).

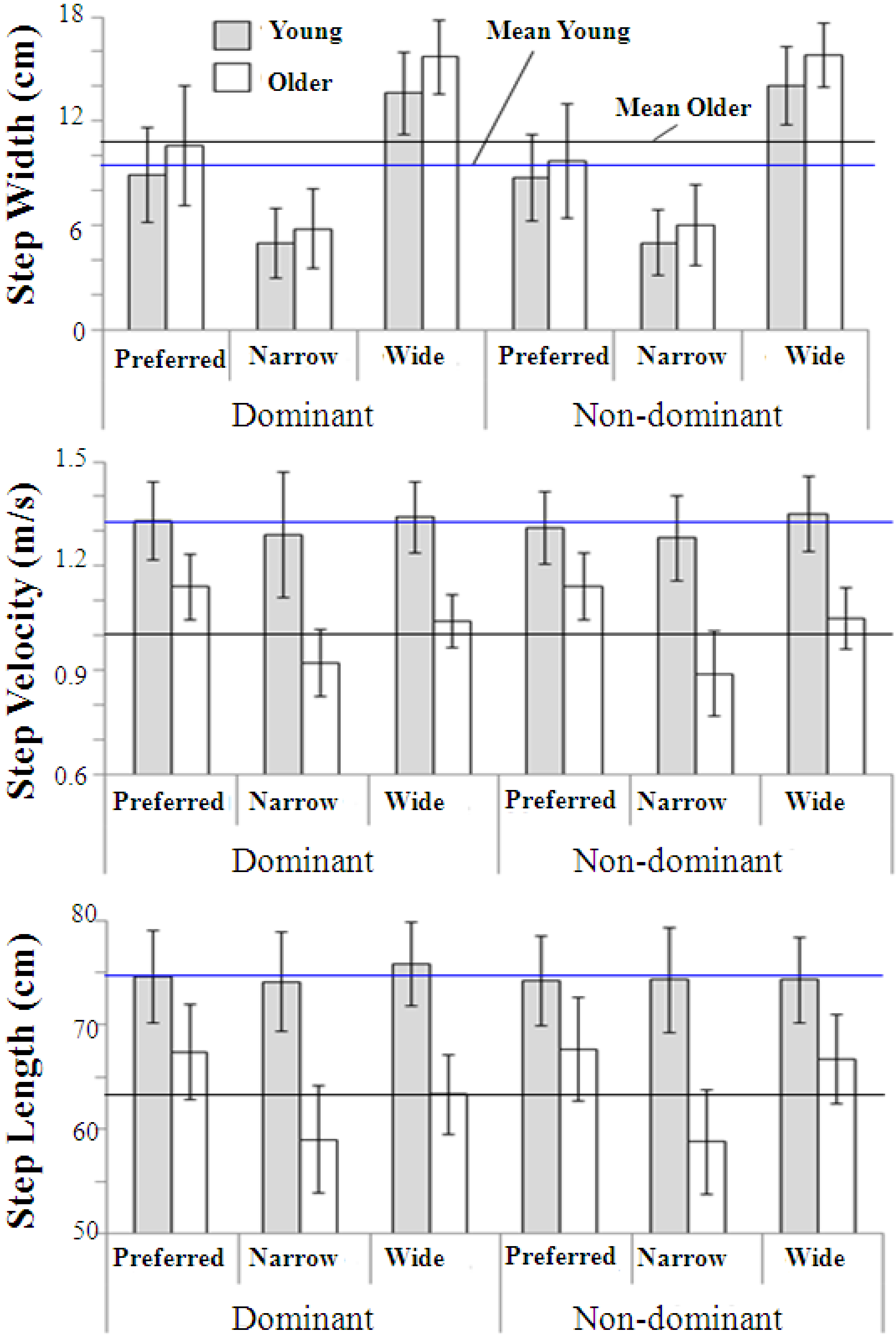

Figure 7.2.1. Ageing effects on step width, step velocity and step length under preferred, narrow and wide walking conditions.

Figure 7.2.2 summarises the findings of ageing and step width effects on double support time. Older adults had longer double support time by 0.05s than young adults (F (1, 54) = 21.7, p <.01) and was negatively correlated with step length (r = -.746, p <.01) and step velocity (r = -.772, p <.01), indicating prolonged double support time when step velocity and length were reduced. Double support time was longest in narrow walking (0.17s) compared to wide (0.15s) and 0.14s at preferred walking (F (2, 53) = 9.1, p <.01). This effect was, again, absent in the young group but seen only in older adults (age x width: F (2, 53) = 7.0, p <.01).

In the percentage analysis indicating the proportion of double support relative to step time, the same trend was observed. Ageing was a factor to increase double support time by 4% (F (1, 54) = 43.6, p < .01). A width effect was obtained (F (2, 53) = 5.7, p < .01) for narrow walking eliciting longest double support time percentage followed by wide and least in preferred width walking, but this effect was confirmed only in the older group (age x width: F (2, 53) = 4.3, p < .05).

Variability of double support time was higher in the older group (HO: 0.03s vs. HY: 0.02s) but this was not significant (F (1, 54) = 4.0, p = .053). While young adults maintained variability of double support time across the three walking conditions, the older group had higher SD in narrow walking (0.05s), highlighted by an age x width interaction (F (2, 53) = 7.1, p < .01).

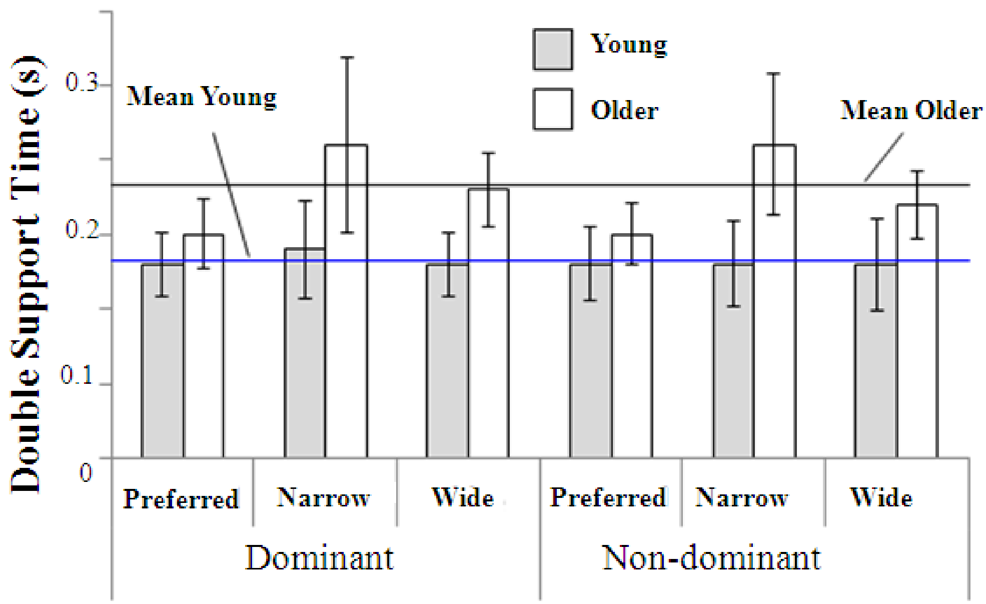

Figure 7.2.2 Ageing effects on double support time under preferred, narrow and wide walking tasks.

7.2.2 Push-off GRF Kinetics Prior to Toe-off at RCoF2

The results of ground reaction forces (GRF) kinetics measured at RCoF2, the negative RCoF peak prior to toe-off (but not peak GRF), are summarised in Figure 7.2.3 below.

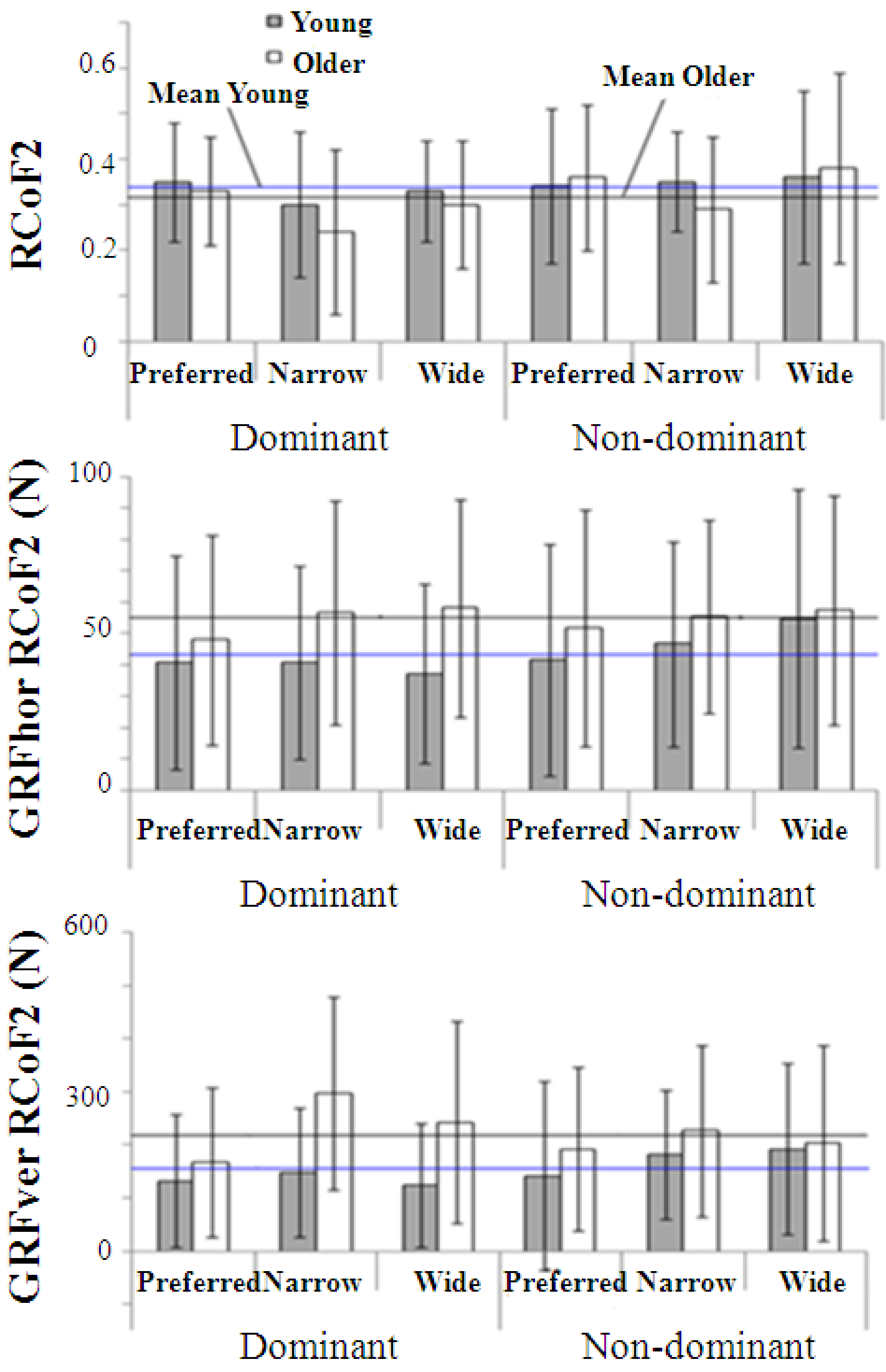

Figure 7.2.3 Ageing effects on RCoF and GRF characteristics at RCoF2 prior to toe-off under preferred, narrow and wide walking tasks.

RCoF2 was lower in the older group ($F(1, 54) = 9.4$, $p < .01$) and narrow walking of 0.29 ($F(2, 53) = 11.2$, $p < .01$) compared to wide (0.32) and preferred width walking (0.34). Lowest RCoF2 (0.26) was, therefore, identified when older adults walked with narrow steps (age x width: $F(2, 53) = 4.4$, $p < .05$).

Horizontal push-off force was greater in the older group by approximately 7N ($F(1, 54) = 13.5$, $p < .01$) or 13.3% after body mass normalisation. A width effect ($F(2, 53) = 4.5$, $p < .05$) was due to horizontal push-off force being lowest in preferred width walking (42N) compared to narrow (48N) and wide walking (48N). Width effects specific to the young group was non-dominant horizontal push-off GRF stronger than the dominant limb by 5N ($F(1, 54) = 6.0$, $p < .05$) during wide walking ($F(1, 54) = 3.6$, $p < .05$).

Vertical GRF was 70N (or 30% in normalised value) greater among older adults at RCoF2 ($F(1, 53) = 60.3$, $p < .01$). In relation to step width effects, greatest vertical GRF of 214N was seen in narrow walking, followed by 191N during wide and 158N during preferred width walking ($F(2, 53) = 14.4$, $p < .01$). No asymmetry was obtained for vertical GRF but young adults had greater vertical GRF in their non-dominant limb by 37N than the dominant side, while the dominant side was greater for older adults by 30N (age x limb: $F(1, 54) = 13.7$, $p < .01$). Older adults' dominant limb showed greater vertical GRF in width controlled walking (age x limb x width: $F(2, 53) = 5.9$, $p < .01$).

7.2.3 GRF Kinetics Immediately Following Heel Contact at RCoF1

Ground reaction force (GRF) characteristics in relation to heel contact were assessed at RCoF1 (Figure 7.2.4). Overall, identical RCoF1 were found between the two age groups (Young 0.21 vs. Older 0.20). Step widths affected RCoF1 ($F(2, 53) = 3.3$, $p < .05$) but the effect was seen only for older adults in response to different walking widths (Normal 0.23 vs. Narrower 0.17 vs. Wider 0.19), supported by an age x width interaction ($F(2, 53) = 10.7$, $p < .01$).

Horizontal GRF decreased with age by 24N ($F(1, 54) = 46.6$, $p < .01$), which is 33% reduction after body mass normalisation. Both narrow and wide walking accompanied lower horizontal GRF than preferred width ($F(2, 53) = 4.5$, $p < .05$). Non-dominant horizontal GRF was about 10N greater than the dominant side ($F(1, 54) = 8.1$, $p < .01$). The young group had greater vertical GRF by 90N (23.7%) ($F(1, 54) = 33.0$, $p < .01$). Vertical GRF was greatest in preferred (473N) compared to 450N in narrow and 423N in wide walking ($F(2, 53) = 3.9$, $p < .05$). This effect appeared only in the young group (age x width: $F(2, 53) = 7.2$, $p < .01$), while the older group demonstrated greatest vertical GRF in wide walking (417N).

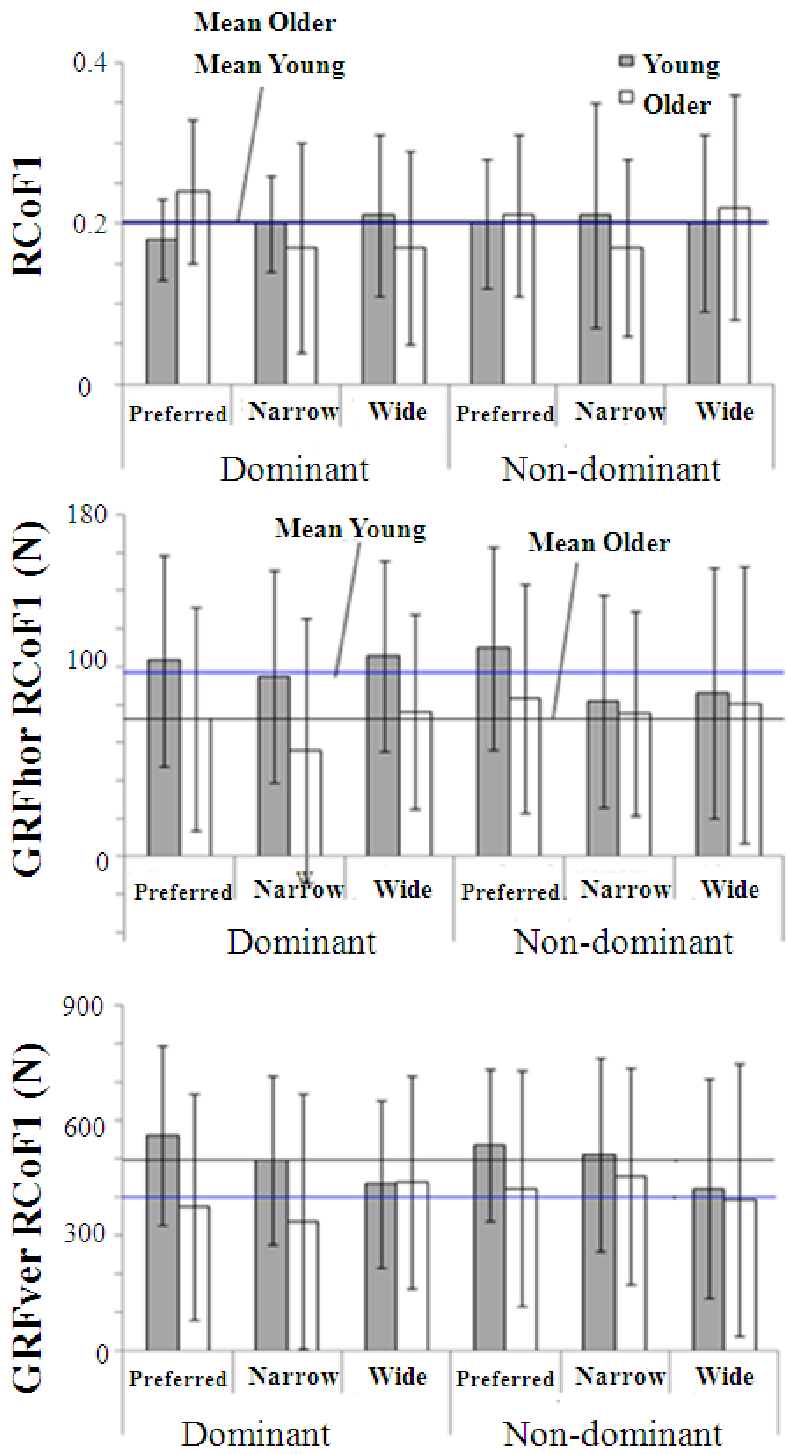

Figure 7.2.4 Ageing effects on RCoF and GRF characteristics at RCoF1 following heel contact under preferred, narrow and wide walking tasks.

7.2.4 Heel Contact Dynamics

Figure 7.2.5 shows heel and foot contact angles at heel contact indicating the slipping potential for anterior heel slipping (Brady et al., 2000; McGorry et al., 2008; 2010).

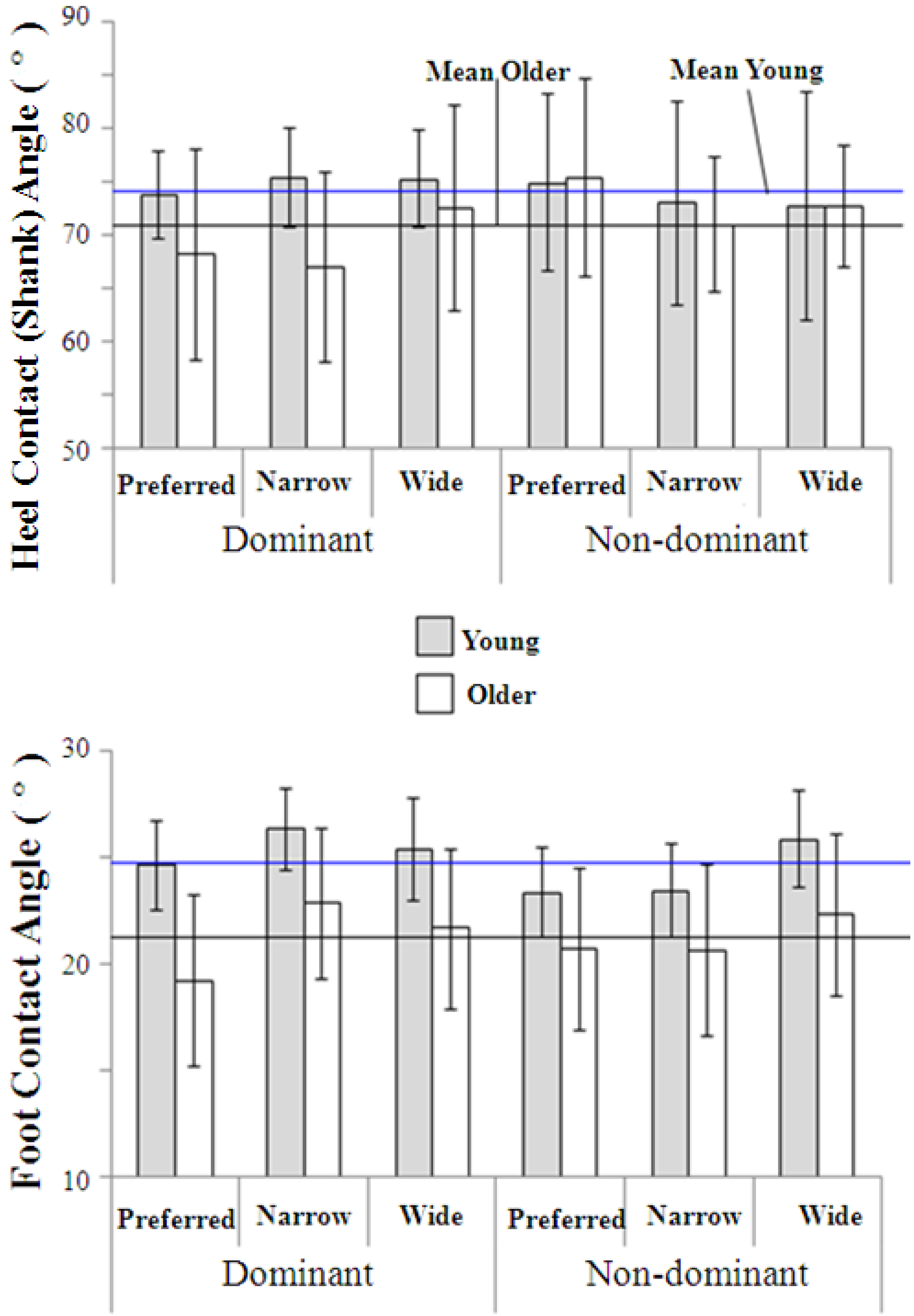

Figure 7.2.5 Ageing effects on heel contact angle and foot contact angle under preferred, narrow and wide walking tasks.

Older adults showed lower heel contact angle ($F (1, 54) = 3.9$, $p < .05$) but the difference was seen only in the dominant limb by 6° but not with the non-dominant limb (age x limb: $F (1, 54) = 4.4$, $p < .05$). Foot contact angle was smaller by 3.5° in the older group ($F (1, 54) = 13.0$, $p < .01$). Foot contact angle was generally smaller in preferred width than width controlled walking conditions but this effect was not significant ($F (2, 53) = 2.5$, $p = .082$).

7.2.5 MFC

Figure 7.2.6 below illustrates how age, limb dominance and step widths influence MFC control. Non-dominant MFC_h was higher than the dominant side for the older group (age x limb: $F (1, 54) = 14.7$, $p < .01$). In width controlled walking conditions, MFC_h increased relative to preferred width walking ($F (2, 53) = 6.3$, $p < .05$). For intra-subject variability, highest IQR was identified in narrow walking ($F (2, 53) = 7.1$, $p < .05$).

Older adults showed MFC_t approximately 3.7% earlier than young adults ($F (1, 54) = 31.8$, $p < .01$). Compared to preferred step width walking, both narrow and wide walking elicited earlier MFC by approximately 5% ($F (2, 53) = 16.1$, $p < .01$) but this effect was more clearly seen in young adults (age x width: $F (2, 53) = 6.4$, $p < .01$). Young adults showed greater MFC_t in the non-dominant swing limb while the opposite was seen in older adults (age x limb: ($F (1, 54) = 10.6$, $p < .01$).

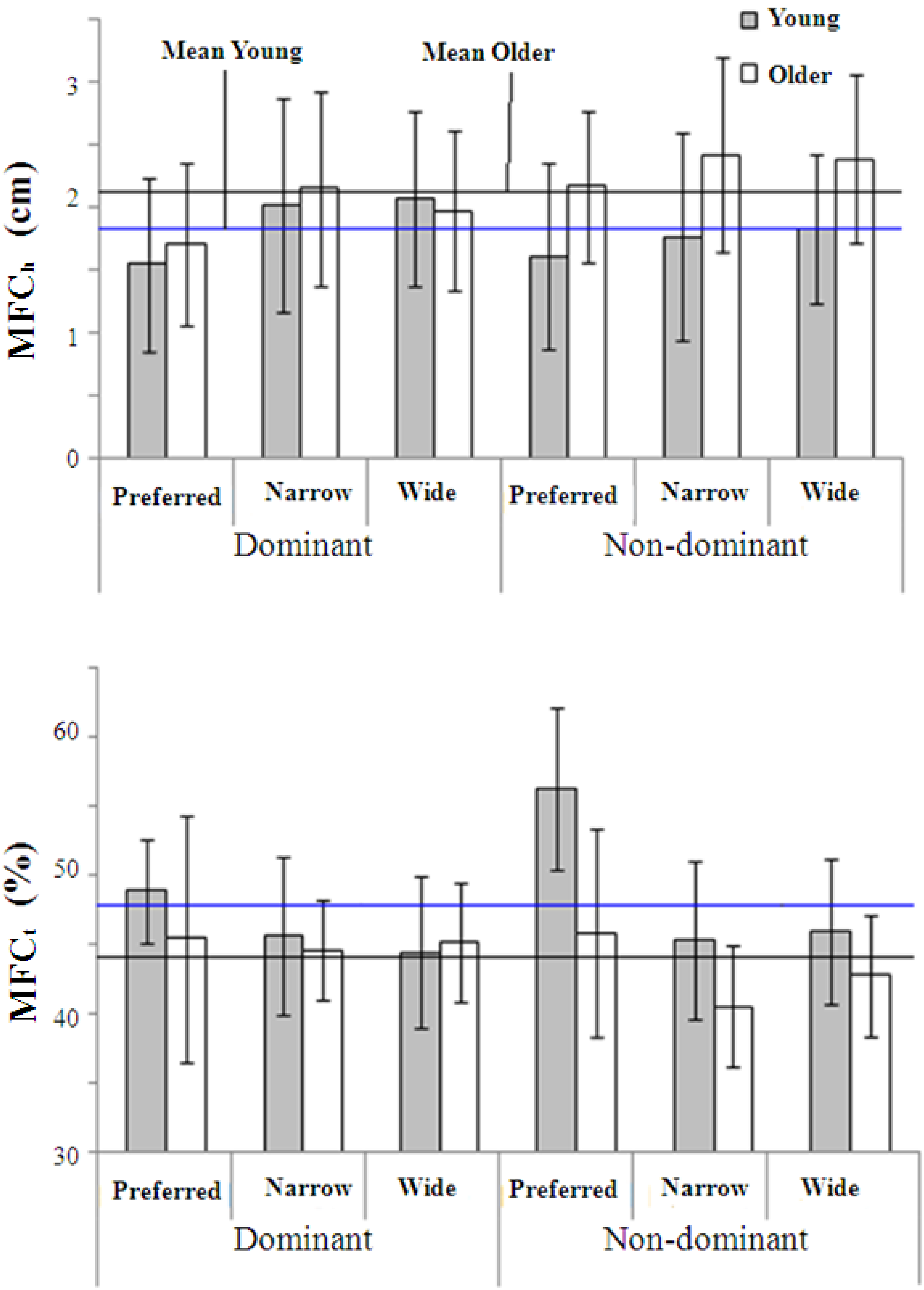

Figure 7.2.6 Ageing effects on MFC characteristics (mean ±IQR) in preferred width, narrow and wide walking conditions. Error bar indicating intra-subject variability of MFC.

7.2.6 MLM

The investigation of critical mid-swing phase event in relation to lateral balance loss, minimum lateral margin (MLM), is examined for spatial (MLM_d) and temporal (MLM_t) data as presented in Figure 7.2.7.

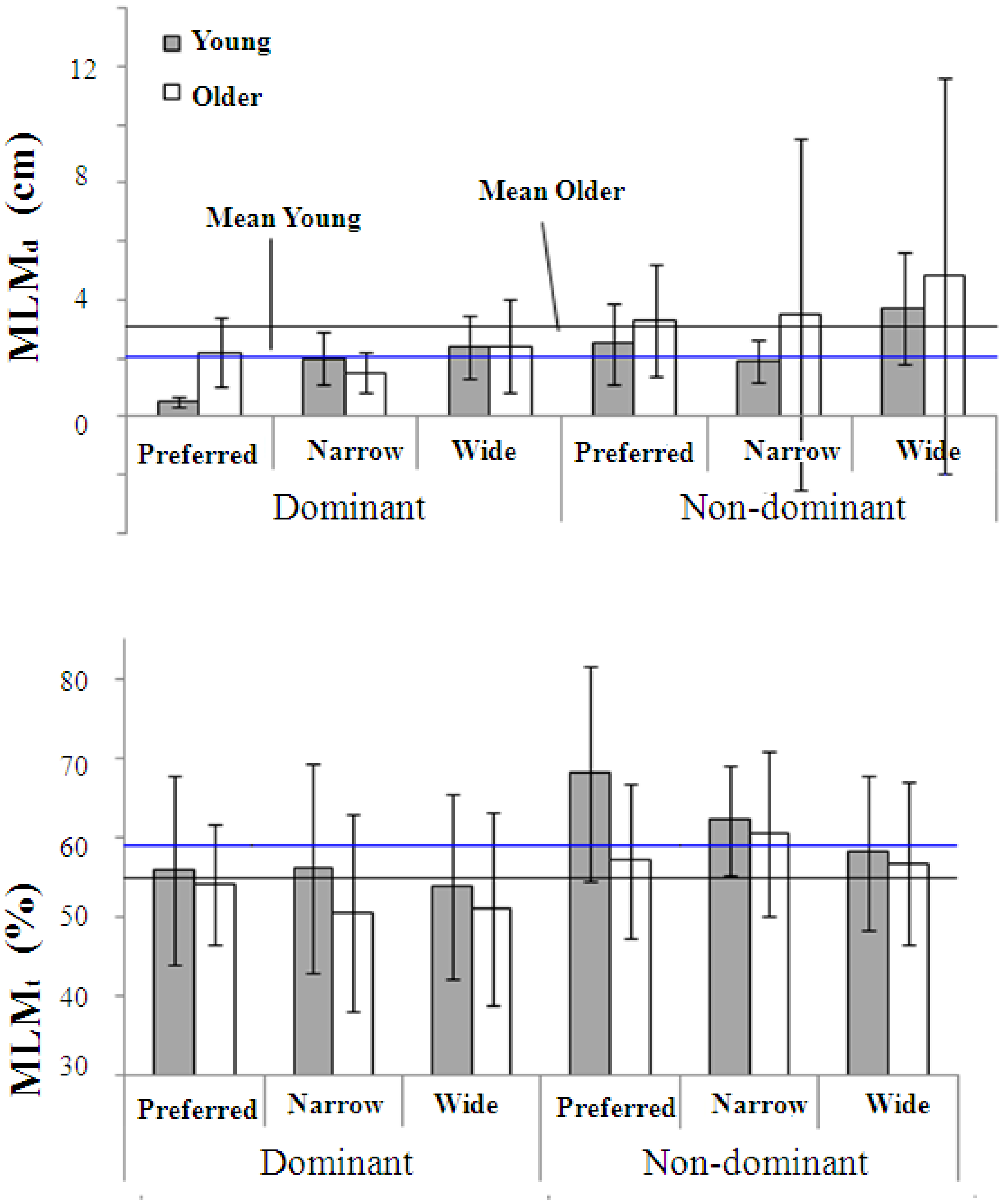

Figure 7.2.7 Ageing effects on MLM characteristics under preferred width, narrow and wide walking.

MLM_d was not differentiated between older and young adults. Non-dominant MLM_d was found to be 1.5cm greater than the dominant side ($F (1, 54) = 21.9$, $p < .01$). Greatest MLM_d was found in wide (3.4cm) followed by narrow (2.2cm) and preferred walking (2.1cm), supported by the width effect ($F (2, 54) = 8.7$, $p < .01$). Less consistent MLM_d control was found in the older group ($F (1, 53) = 35.5$, $p < .01$) due to increased MLM_d variability (by 0.8cm) in response to narrow and wide (age x width: ($F (2, 53) = 6.1$, $p < .01$).

Older adults experienced MLM earlier in swing by 4% than young adults ($F (1, 54) = 8.4$, $p < .01$). Walking with wide steps caused lower MLM_t ($F (2, 53) = 4.0$, $p < .05$). MLM_t in the non-dominant limb was 7% greater than the dominant limb ($F (1, 54) = 29.6$, $p < .01$).

7.2.7 Lateral CoP Displacement and Mean Medial CoM Acceleration

To characterise the ML balance of a particular gait style, lateral CoP displacement and average medial velocity on CoM were recorded between toe-off and MLM (Figure 7.2.8). No ageing effect was discovered for lateral CoP displacement from toe-off to MLM. CoP displacement in the dominant limb was 0.9cm greater than the non-dominant limb ($F (1, 54) = 21.9$, $p < .01$). SD of lateral CoP displacement was significantly higher in the older group ($F (2, 53) = 4.5$, $p < .05$) and in the non-dominant stance foot ($F (1, 54) = 6.5$, $p < .05$). An age x limb ($F (1, 54) = 6.5$, $p < .05$) and age x width interactions ($F (2, 53) = 3.8$, $p < .05$) were consistently supporting older adults increasing non-dominant SD during narrow and wide walking with no such effects for their young counterparts.

Linear medial CoM acceleration was influenced by step width and highest acceleration observed in wide and lowest in narrow step walking ($F (2, 53) = 70.8$, $p < .01$). Older adults, however, had highest medial acceleration in preferred width walking (age x width: $F (2, 53) = 7.5$, $p < .01$). No age or limb effects were observed for either mean or SD of medial CoM acceleration.

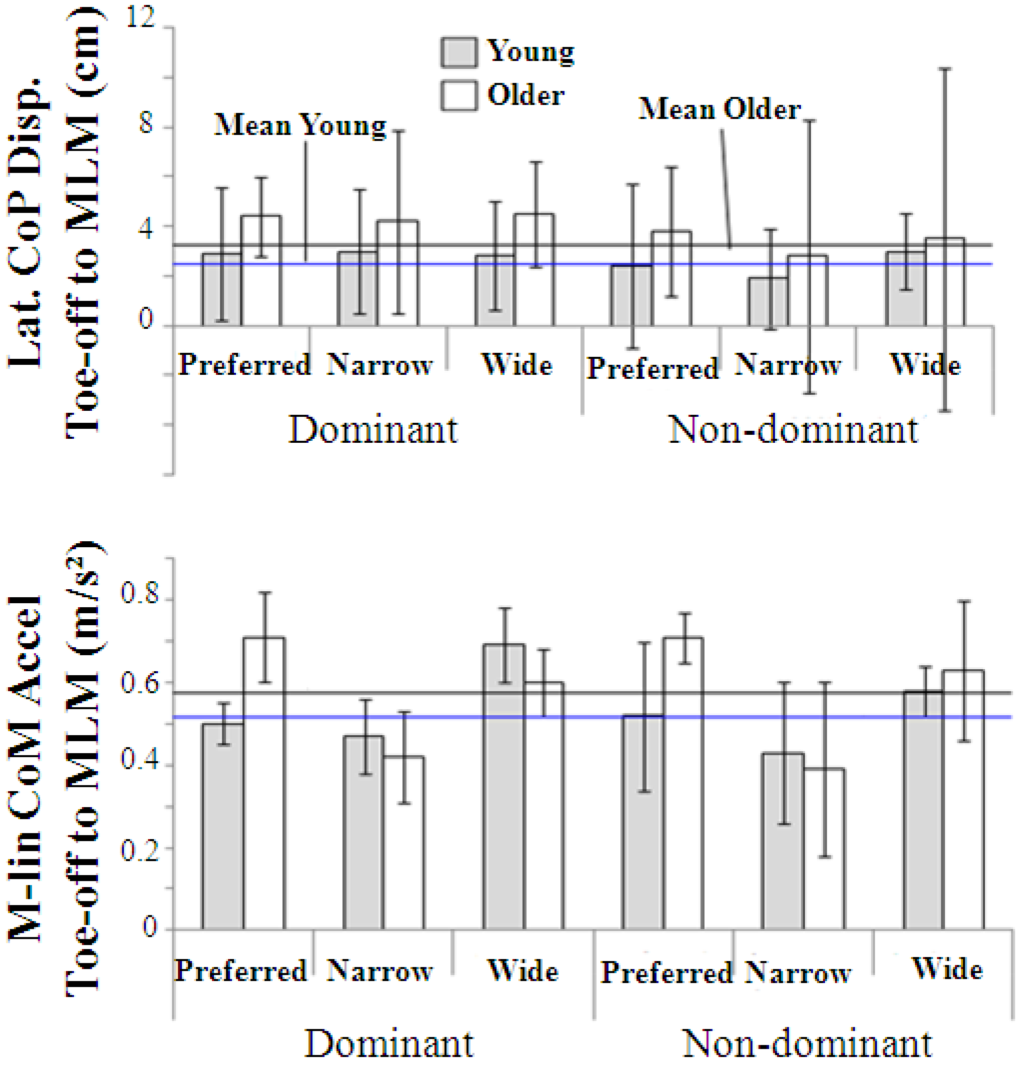

Figure 7.2.8 Ageing effects on lateral CoP displacement (top) and mean medial linear CoM acceleration (bottom). From toe-off to MLM under preferred width, narrow and wide walking conditions.

7.3 Chapter Summary

Ageing has been widely recognised as a factor for increased risk of falls largely due to balance loss, and older adults tend to adopt wider steps to secure balance (Hurt et al., 2010; Pandy et al., 2010; Whittle, 2007; Ko et al., 2007). To explore age-related effects on balance control, effects of increased and decreased step width on balance during gait were investigated. Effects of width were visible only in older adults' gait but not their younger counterpart. Both controlled width walking conditions were, however, discovered to reduce step width variability probably because lines aided the foot targeting task.

As in previous research, ageing effects on spatio-temporal parameters (Winter, 1991; Whittle, 2007) were consistently observed such as reduced step velocity due to shorter step length and prolonged double support time. These typical ageing effects were most accentuated in narrow walking, followed by wide and least in preferred width walking. These findings confirm previous studies (Nagano et al., 2012; Chamberlin et al., 2005; Shkuratova et al., 2004) stating that ageing effects on gait are accentuated in challenging conditions. During narrow walking, older adults demonstrated remarkably high double support time variability, possibly reflecting balance adjustment.

Ageing and step width effects on balance were examined at four swing phase events. Swing phase begins with toe-off and is terminated by heel contact. The common characteristics of these two events are ground reaction forces (GRF) generation due to braking and push-off, associated with anterior and posterior slipping, respectively. Older adults revealed lower negative peak of require coefficient of friction (RCoF2), safer against posterior forefoot slipping at push-off phase before toe-off. Contrary to expectation, both horizontal and vertical GRF components at RCoF2 were greater in older adults, especially during narrow walking. In contrast, GRF kinetics immediately following heel contact showed the opposite trend as young adults showed greater GRF in both horizontal and vertical components. Older adults reduced RCoF1 in width controlled walking while in preferred width walking, older adults showed higher RCoF1 especially following dominant heel contact. The higher risk of slipping in the dominant heel contact of older adults was further supported by smaller heel contact angle (Brady et al., 2000; McGorry et al., 2008; 2010). As hypothesised, lower foot contact angle was characterised in the older group, the adaptation to reduce RCoF1. Thus, despite increased risk of heel anterior slipping in the dominant limb of older adults, flatter foot contact indicated by lower foot contact angle in older adults was considered to assist lower RCoF1.

For both age groups, walking with different step widths caused increase in MFC_h. Obtained results for MFC data confirmed expected outcomes, in that ageing was not associated with reduced MFC_h, consistent with previous studies (e.g. Begg et al., 2007). In width controlled walking, non-dominant MFC_h was significantly higher in older adults, similar to previous findings of older adults elevating non-dominant MFC_h in response to treadmill walking (Nagano et al., 2011). Variability of MFC_h (IQR) increased in response to narrow walking, indicating less consistent MFC_h control when walking width narrower width.

At another mid-swing event, MLM, effects of ageing on ML balance were examined. Step width tends to increase with ageing to preserve dynamic balance especially in ML direction (Ko et al., 2007). For this reason, MLM characteristics were thought to be influenced by step width, therefore ageing. Ageing effects were not, however, detected in the examined parameters for ML balance but age-related impairment in consistent control of MLM_d and lateral CoP displacement especially in the non-dominant limb during controlled width walking conditions were characterised in greater variability.

CHAPTER 8

Ageing and Step Width Effects on the Safety Zone

8.1 Overview

A Safety Zone was defined by the positions of the toes and heels in the transverse (x-y) plane relative to the CoM. Consistency of the centre of mass (CoM) control within the Safety Zone was assessed by step-to-step variation in the distance between the CoM and foot in AP and ML directions. Variability of the Safety Zone across multiple gait cycles can be due to either the highly variable CoM or foot control. Variability in step length and step widths is due to foot control. Consistency in spatio-temporal parameters with low variability in the Safety Zone boundaries indicates that the CoM control is steady. Balance loss direction is defined by which boundary of the Safety Zone, the CoM is likely to cross.

Ageing often induces conservative gait adaptations to secure balance (Bock and Beurskens, 2010; Dunlap et al., 2012; Hollman et al., 2007; Ko et al., 2007; Nagano et al., 2012; Nordin et al., 2012; Schrager et al., 2008). It was therefore predicted that older adults would demonstrate balance securing adaptations to preserve the Safety Zone especially during narrow or wide walking. In addition to age-specific adaptations to the Safety Zone, age-associated impairment in the balance control was expected to be more evident during narrow step walking. The more variable CoM position relative to both feet was generally expected to reflect age-related impairment in balance controlling ability. Because ageing is a factor to reduce the skill to separate the CoM from both feet in the AP direction (Lugade et al., 2011), shorter in AP length but larger in ML extent were expected in the older adults' Safety Zone compared to the young due to age-related increase in step width and reduction in step length (Whittle, 2007).

The older adults' non-dominant limb is more vulnerable than dominant limb for reduced strength and joint range of motion (Perry et al., 2007), possibly causing asymmetry of the Safety Zone. Age-associated gait asymmetry may assign gait securing to the non-dominant limb (Nagano et al., 2011; 2012), and older adults were, therefore, expected to show limb-specific adaptations when gait was challenged by wide and most importantly narrow step walking.

8.2 Results

A 2×3×2 (age × width x limb) repeated measures mixed model Analysis of Variance (ANOVA) was applied to show how ageing, step width and limb dominance would influence the Safety Zone characteristics described by both mean and standard deviation (SD) characteristics of anterior-posterior (AP) and medio-lateral (ML) distances of toes and heels from the CoM.

8.2.1 Toe-off

At toe off, the CoM was directed anterior slightly toward the lead-foot. During narrow walking in older adults, however, the CoM was directed to the lateral border of the lead non-dominant foot. As visualised in Figure 8.2.1, the older adults' non-dominant foot showed toeing-in during narrow step and, to a lesser degree, in wide step walking.

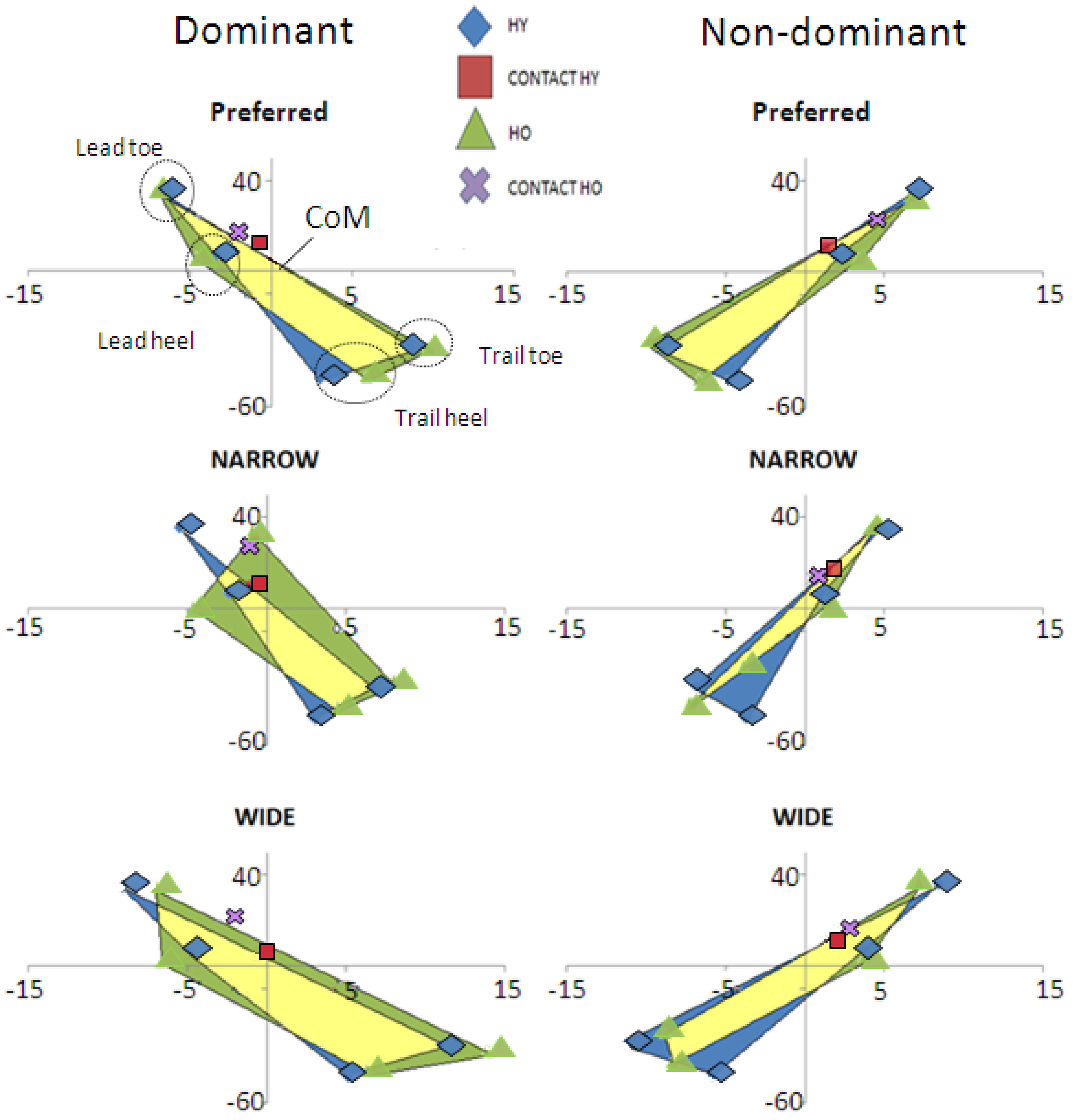

Figure 8.2.1 Safety Zone at toe-off. HY = healthy young adults; HO = healthy older adults; CONTACT = balance loss initiation point based on the CoM velocity. Safety Zone, yellow = common to HY and HO; blue = young adults; green = older adults. X axis (cm), positive (> 0) = direction to the dominant and negative (< 0) to the non-dominant foot. Y axis (cm) = anterior-posterior. Locations relative to the CoM (0, 0).

➢ <u>Toeing-in of the Non-dominant Foot</u>:

'Toeing-in'of the feet was observed in older adults' narrow and wide walking, either due to a more medially placed toe or increased lateral heel location, or both. Toeing-in was not seen in young adults.

Lead foot: The older group showed a more medial lead-toe by 1.2cm ($F(1, 54) = 6.4$, $p < .05$) despite the same ML heel position. The lead toe was found more medial in the non-dominant foot than the dominant limb by 1.8cm ($F(1, 54) = 9.3$, $p < .01$) but not the heel, reflecting toeing-in of the lead non-dominant foot, as seen in Figure 8.1.1. Non-dominant toe ing-in was, however, found only for older adults when walking with narrow steps and also less strongly in the wide condition (age x limb x width: $F(2, 53) = 4.8$, $p < .05$).

Trail (push-off) foot: Non-dominant toeing-in was characterised in the older group's trail foot due to a 1.2cm more medial toe (age x limb: $F(1, 54) = 5.1$, $p < .05$) and 2.2cm more lateral heel (age x limb: $F(1, 54) = 6.0$, $p < .05$).

- <u>Higher CoM Variability in Older Adults:</u>

Lead foot: ML CoM control within the lead foot's area of the Safety Zone was significantly more variable in the older group (>0.5cm); toe: $F(1, 54) = 58.7$, $p < .01$; heel: $F(1, 54) = 20.0$, $p < .01$. When the non-dominant foot was leading, ML control became considerably more variable than when the dominant limb was leading the gait cycle (heel: $F(1, 54) = 6.3$, $p < .05$). More variable AP CoM control within the lead foot region of the Safety Zone was found in the older group (age x limb: $F(1, 54) = 14.9$, $p <.01$), especially relative to the non-dominant heel location in narrow walking (age x limb x width: $F(2, 53) = 7.5$, $p < .01$).

Trail (push-off) foot: In the trail foot area of the Safety Zone older adults also had more variable ML CoM control (>0.4cm); toe: $F(1, 54) = 38.4$, $p < .01$, heel: $F(1, 54) = 22.6$, $p < .01$. ML CoM control relative to the toe was more variable (SD=0.8cm) when the dominant trail foot was trailing ($F(2, 53) = 125.2$, $p < .01$), with SD=0.5cm at the heel ($F(2, 53) = 25.4$, $p < .01$).

- <u>Reduction in Variability During Width-controlled Walking</u>

Lead foot: In both controlled width conditions, ML CoM location within the lead foot Safety Zone area became more consistent relative to preferred width walking (toe: $F(2, 53) = 7.5$, $p < .01$), but the effect was only for the non-dominant lead foot (toe: limb x width: $F(2, 53) = 14.3$, $p < .01$, heel: $F(2, 53) = 21.5$, $p < .01$).

Trail (push-off) foot: Only older adults reduced ML CoM variability during controlled width walking while young adults showed more consistent variability across the three walking widths (age x width: toe: $F(2, 53) = 15.8$, $p < .01$, heel: $F(2, 53) = 11.7$, $p < .01$).

- Less AP Foot Separation in Older Adults

Lead foot: Older adults showed less separation of the CoM from the AP location of lead foot (toe: $F(1, 54) = 6.3$, $p < .05$, heel: $F(1, 54) = 7.6$, $p < .01$).

Trail (push-off) foot: Posterior separation of the trail foot from the CoM was smaller in the older group (toe: $F(1, 54) = 4.4$, $p < .05$, heel: $F(1, 54) = 11.7$, $p < .05$).

8.2.2 MFC

Figure 8.2.2 visualises the Safety Zone characteristics in the three walking conditions at MFC. CoM movement was always anterior at MFC independent of age, limb dominance, or step width. Despite, however, the consitent CoM anterior motion, balance loss will be *lateral* to the non-dominant stance foot when older adults walk with narrow steps. Toeing-in of the non-dominant limb in narrow and wide walking can be observed in the older groups data.

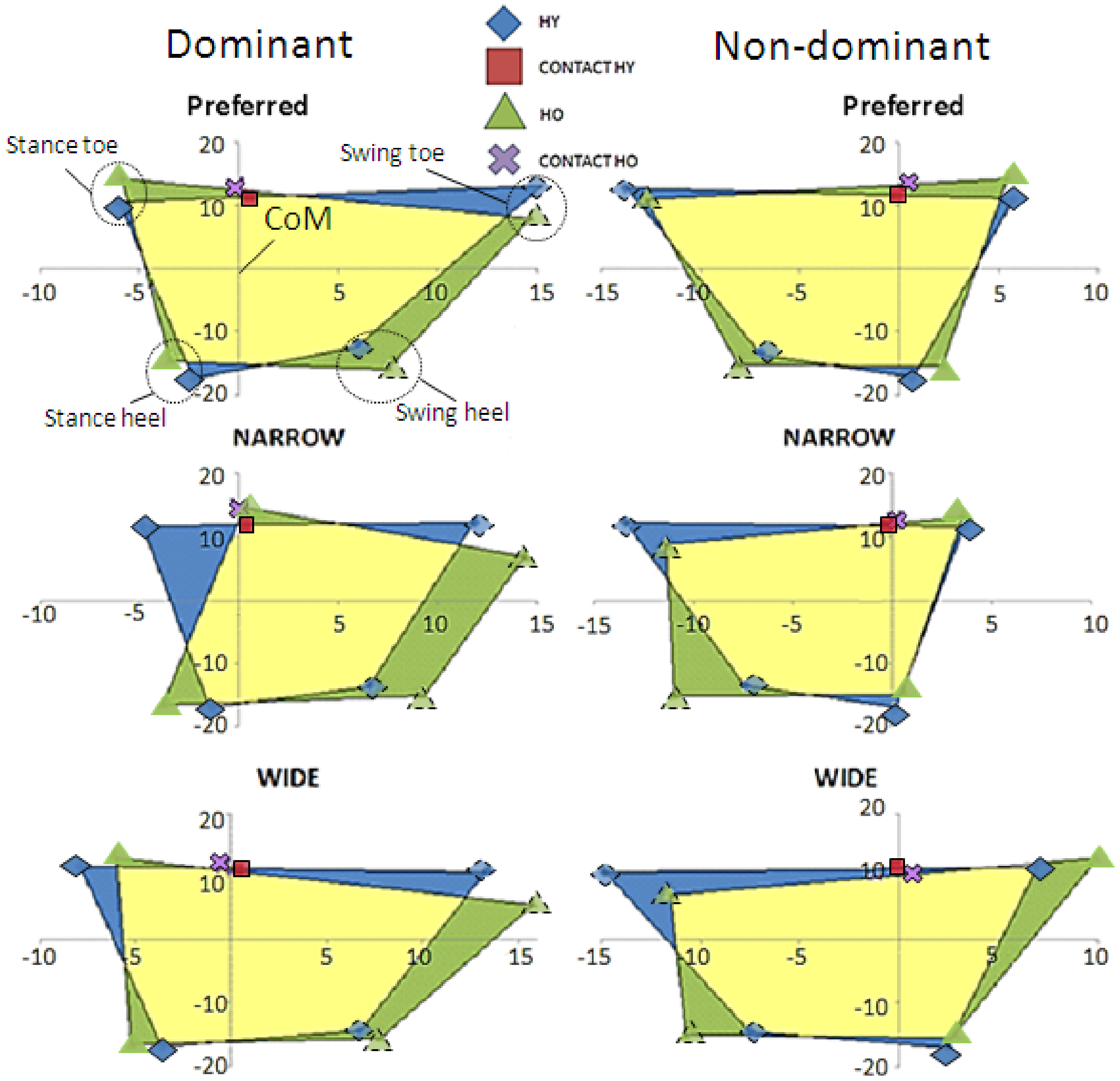

Figure 8.2.2 Safety Zone at MFC. HY = healthy young adults; HO = healthy older adults; CONTACT = point of balance loss initiation based on the CoM velocity. Safety Zone, yellow = common to HY and HO; blue = only for younger adults; green = only for older adults. X axis (cm), positive (> 0) = direction to the dominant and negative (0>) to the non-dominant foot. Y axis (cm) = anterior-posterior. Locations relative to the CoM (0, 0).

➢ Toeing-in of the Non-dominant Foot:

Swing Foot: Older adults' non-dominant swing foot showed toeing-in due to a more medial swing toe (age x limb: $F(1, 54) = 6.9$, $p < .05$). In addition, the swing foot's heel of the older group was more lateral during width controlled walking conditions than unconstrained walking (limb x width: $F(2, 53) = 7.8$, $p < .01$; age x width: $F(2, 53) = 11.1$, $p < .01$).

Stance Foot: Older adults' stance toe was more medial than the young but not significantly ($F(1, 54) = 3.8$, $p = .060$). Toe location in the older group's narrow walking was, however, significantly more medial than the young (age x width: $F(2, 53) = 52.1$, $p < .01$). The non-dominant stance heel was more lateral to the CoM by 1.8cm compared to the dominant stance heel ($F(1, 54) = 9.1$, $p < .01$).

➢ Higher CoM Variability in Older Adults:

Swing Foot: ML CoM movement on the swing foot's side of the Safety Zone was more variable in the older group (0.3cm), toe: $F(1, 54) = 22.5$, $p < .01$, heel $F(1, 54) = 21.9$, $p < .01$ and also AP (toe $F(1, 54) = 18.0$, $p < .01$; heel $F(1, 54) = 24.6$, $p < .01$).

Stance Foot: Older adults had more variable ML CoM control within the Safety Zone on the stance foot side (toe $F(1, 54) = 20.2$, $p < .01$; heel $F(1, 54) = 24.1$, $p < .01$). ML CoM control variability was greater in the non-dominant stance foot (toe $F(1, 54) = 14.3$, $p < .01$; $F(1, 54) = 14.6$, $p < .01$). AP CoM control was also more variable in the older group (toe: $F(1, 54) = 52.5$, $p < .01$, heel: $F(1, 54) = 24.6$, $p < .01$).

➢ Reduction in CoM Variability During Width-controlled Walking

Swing Foot: Variability of the ML CoM control reduced during width controlled walking but this effect was observed only in the non-dominant limb (limb x width: toe: $F(2, 53) = 9.3$, $p < .01$, heel: $F(2, 53) = 7.8$, $p < .01$).

Stance Foot: In width controlled walking conditions, variability of the ML CoM control reduced in older adults (age x width: toe: $F(2, 53) = 4.2$, $p < .05$, heel $F(2, 53) = 5.0$, $p < .05$).

- <u>Lower AP Foot Separation in Older Adults</u>

Swing Foot: More posterior swing heel from the CoM was found in the older group ($F(1, 54) = 5.6$, $p < .05$).

Stance Foot: Young adults showed more posterior stance foot from the CoM by approximately 3cm than the older group (toe $F(1, 54) = 24.5$, $p < .01$; heel $F(1, 54) = 16.2$, $p < .01$).

8.2.3 MLM

Figure 8.2.3 describes the Safety Zone at MLM. The ML CoM velocity at MLM is theoretically close to zero and accordingly, the CoM movement was anterior. Toeing-in was characterised most clearly in the older adults' non-dominant stance foot during narrow walking with greater risk of lateral balance loss at non-dominant MLM. The non-dominant swing foot's toeing-in did not affect AP margins to balance loss.

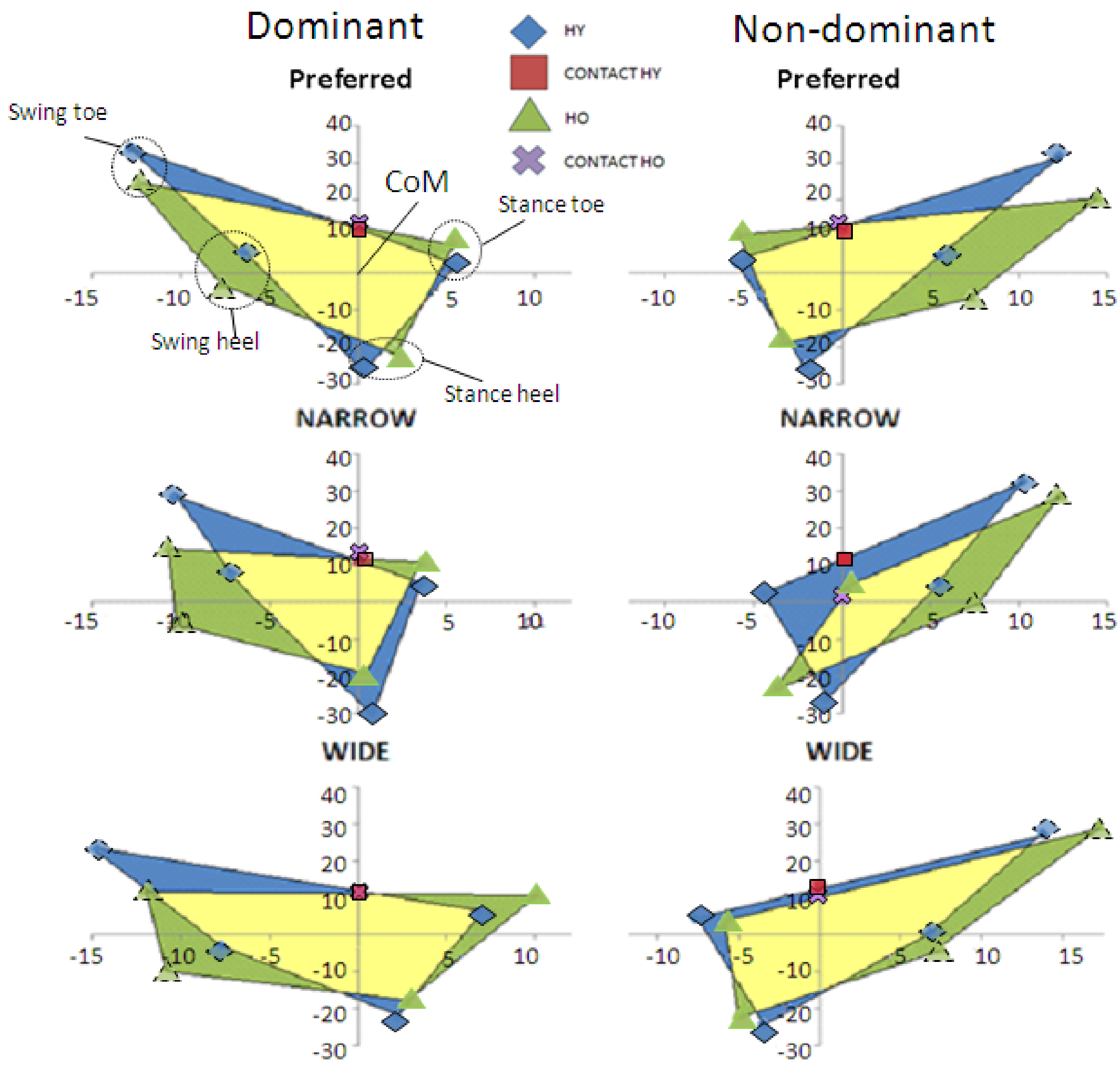

Figure 8.2.3 Safety Zone at MLM. HY = healthy young adults; HO = healthy older adults; CONTACT = the point of balance loss initition based on the CoM velocity. Safety Zone, yellow = common to HY and HO; blue = only for younger adults; green = only for older adults. X axis (cm), positive (> 0) = direction to the dominant and negative (0>) to the non-dominant foot. Y axis (cm) = anterior-posterior. Locations relative to the CoM (0, 0).

➢ <u>'Toeing-in' of the Non-dominant Foot</u> :

Swing Foot: Swing foot toeing-in was characterised in the older group as a more lateral heel (1.8cm); age $F (1, 54) = 10.1$, $p < .01$. Swing foot lateral deviation from the CoM was 1.3cm more lateral in the non-dominant swing heel compared to the dominant side ($F (1, 54) = 5.7$, $p < .05$). More lateral swing heel was detected in both the width-controlled walking tasks ($F (2, 53) = 4.8$, $p < .05$). The toe was most medial in narrow walking ($F (2, 53) = 30.6$, $p < .01$),.

Stance Foot: ML distance between the stance heel and the CoM was MLM_d (Figure 7.7). Toeing-in was seen most clearly in narrow walking due to the older adults' non-dominant medial stance toe (age x limb x width: $F (2, 53) = 4.3$, $p < .05$). The ML CoM-stance toe distance was shortest in narrow and longest in wide walking (toe: $F (2, 53) = 63.6$, $p < .01$) with no step width effects on the stance heel.

➢ <u>Higher CoM Variability in Older Adults:</u>

Swing Foot: More variable ML CoM in the swing foot's Safety Zone region was seen in the older group (toe $F (1, 54) = 8.6$, $p < .01$; heel $F (1, 54) = 28.2$, $p < .01$) and more accentuated at dominant MLM compared to non-dominant MLM (toe $F (1, 54) = 17.8$, $p < .01$; heel $F (1, 54) = 23.4$, $p < .01$). In AP, CoM control was more variable in the older group (5.0cm) at the toe ($F (1, 54) = 17.1$, $p < .01$) and at heel (5.4cm); $F (1, 54) = 53.4$, $p < .01$.

Stance Foot: Older adults showed higher variability in ML CoM control relative to the stance foot (toe $F (1, 54) = 20.3$, $p < .01$; heel $F (1, 54) = 29.2$, $p < .01$). This effect was particularly noticable in the non-dominant stance foot which was 0.2cm more variable than the dominant side at the toe ($F (1, 54) = 5.6$, $p < .05$) but the same trend was not seen reliably at the stance heel. More variable ML CoM location relative to the non-dominant stance foot was found in the older group (age x limb interaction: toe: $F (1, 54) = 13.6$, $p < .01$, heel: $F (1, 54) = 4.6$, $p < .05$).

➢ <u>Effects of Width Controlled Walking on ML CoM Variability</u>

Swing Foot: Older adults increased variability of the ML CoM position in response to narrow walking (age x width: toe $F (2, 53) = 9.0$, $p < .01$; heel $F (2, 53) = 15.1$, $p < .01$). This increase in variability was more clearly observed on the non-dominant swing foot side (toe $F (1, 54) = 5.0$, $p < .05$; heel $F (1, 54) = 2.7$, $p = .087$).

Stance Foot: Wide walking assisted in more consistent ML CoM in the stance foot's Safety Zone area (toe: ($F (2, 53) = 3.7$, $p < .05$) with a similar but non-significant result for the stance heel.

- Less AP Foot Separation in Older Adults

Swing Foot: In the AP axis, the older adults showed a more posterior swing foot location relative to the CoM (toe $F (1, 54) = 4.5$, $p < .05$; heel $F (1, 54) = 8.1$, $p < .01$). The dominant swing toe was 3.5cm more anterior than the non-dominant swing toe at D-MLM ($F (1, 54) = 6.1$, $p < .05$). This limb effect was highlighted in both of the width controlled conditions compared to preferred width walking (limb x width: toe $F (2, 53) = 4.5$, $p < .05$; heel $F (2, 53) = 4.2$, $p < .05$).

Stance Foot: Older adults showed a more anterior stance foot location relative to the CoM than the young group (toe $F (1, 54) = 24.2$, $p < .01$; heel $F (1, 54) = 21.4$, $p < .01$).

8.2.4 Heel Contact

Figure 8.2.4 illustrates that older adults had a small anterior margin to balance loss at dominant heel contact in narrow walking. In contrast, at non-dominant heel contact, lead foot toeing-in extended the anterior margin, clearly seen in the narrow and wide walking conditions below (Figure 8.2.4).

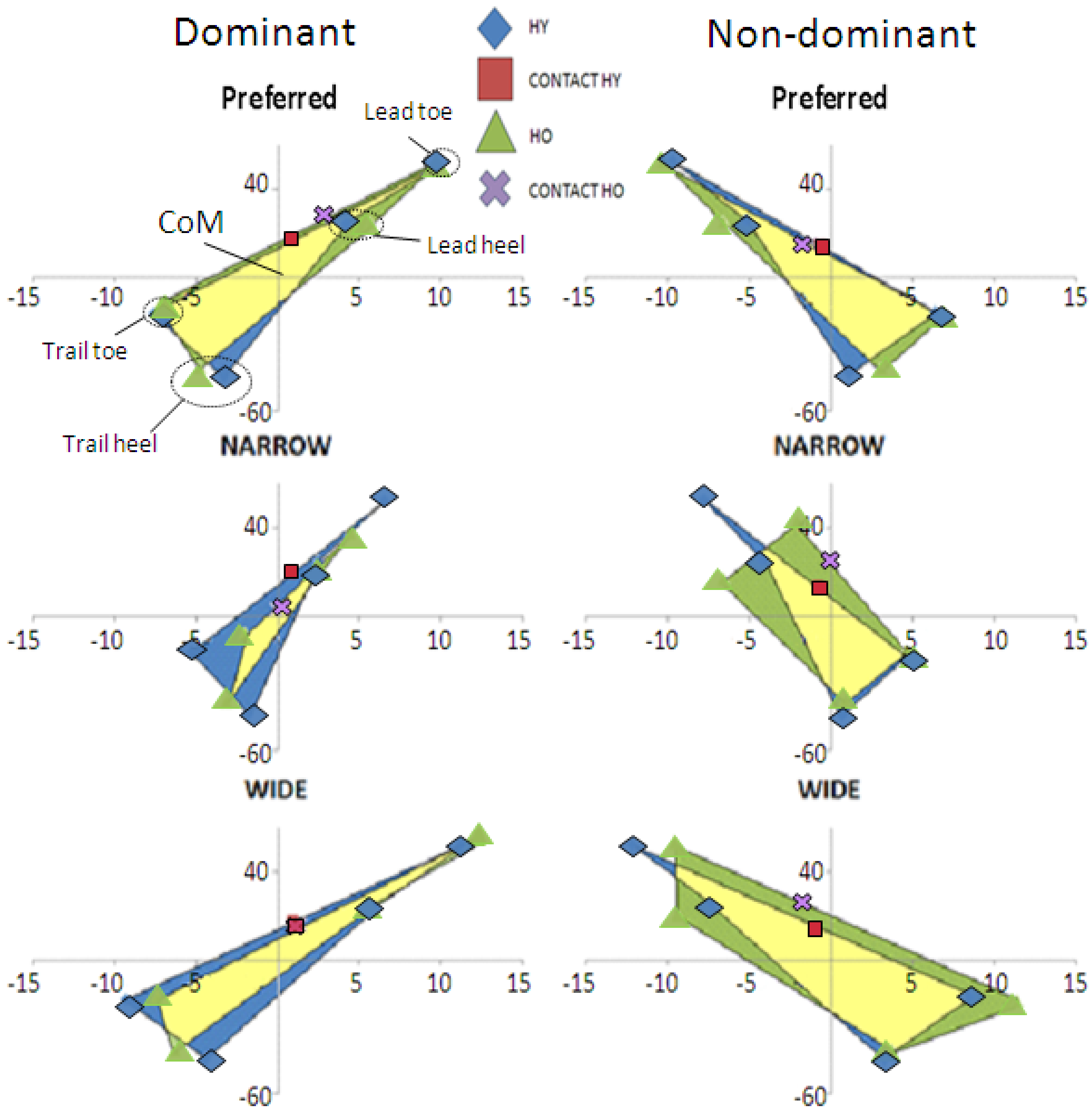

Figure 8.2.4 Safety Zone at heel contact. HY = healthy young adults; HO = healthy older adults; CONTACT = the point of balance loss initition based on CoM velocity. Safety Zone, yellow = common to HY and HO; blue = only for younger adults; green = only for older adults. X axis (cm), positive (> 0) = direction to the dominant and negative (0>) to the non-dominant foot. Y axis (cm) = anterior-posterior. Locations relative to the CoM (0, 0).

- <u>'Toeing-in' of the Non-dominant foot</u> :

Lead Foot: Toe position was most medial in narrow walking (4.9cm) followed by wide (9.8cm) and preferred width walking (10.2cm); $F(2, 53) = 40.1$, $p < .01$, confirming toeing-in during width-controlled walking. Lead heel location was, in contrast, consistent with the step width manipulations, more lateral in wide and medial in narrow ($F(2, 53) = 35.9$, $p < .01$). The non-dominant lead heel was more laterally displaced from the CoM compared to the dominant lead heel (1.8cm); $F(1, 54) = 7.0$, $p < .05$, accounting for lead foot toeing-in only for the non-dominant lead contact foot. As clearly seen in Figure 8.4.1, non-dominant lead foot toeing-in during width controlled walking conditions was unique to the older group (age x width: $F(2, 53) = 3.7$, $p < .05$).

Trail Foot: The older group showed a 0.9cm more medial trail toe ($F(1, 54) = 4.4$, $p < .05$) despite no differences in trail heel. Trail foot was most lateral in wide and least in narrow walking (toe: $F(2, 53) = 71.5$, $p < .01$, heel: $F(2, 53) = 13.5$, $p < .01$). Toeing-in was characterised in the older group's medial non-dominant trail toe, more in narrow and less in wide walking (age x width: $F(2, 53) = 8.6$, $p < .01$).

- <u>Higher CoM Variability in Older Adults:</u>

Lead Foot: Older adults showed higher variability in ML CoM of the lead foot (>0.3cm) at toe ($F(1, 54) = 19.7$, $p < .01$) and (>0.6cm) at the heel ($F(1, 54) = 21.9$, $p < .01$). The effect was more pronounced at non-dominant lead foot contact (age x limb: toe $F(1, 54) = 9.4$, $p < .01$; heel $F(1, 54) = 10.7$, $p < .01$).

Trail Foot: More variable ML CoM in the trail foot's Safety Zone was detected in the older group by (>0.4cm); toe: $F(1, 54) = 26.3$, $p < .01$, heel: $F(1, 54) = 34.1$, $p < .01$.

- <u>Effects of Width Controlled Walking on the ML CoM Variability</u>

Lead Foot: During width controlled walking ML CoM control of the lead foot's side became less variable at toe ($F(2, 53) = 10.1$, $p < .01$) but not significantly with lead heel. This effect was observed only in the older group (age x width: $F(2, 53) = 15.0$, $p < .01$).

Trail Foot: During width controlled walking reduced variability was seen in ML CoM control in the trail foot side of the Safety Zone at toe ($F(2, 53) = 6.5$, $p < .01$) but not at heel.

Only older adults showed this effect (age x width: toe: $F(2, 53) = 17.0$, $p < .01$, heel: $F(2, 53) = 13.5$, $p < .01$).

- <u>Less AP Foot Separation in Older Adults</u>

Lead Foot: The older group had reduced anterior lead foot relative to the CoM at toe ($F(1, 54) = 6.2$, $p < .05$) but not significantly different at heel ($F(1, 54) = 3.6$, $p = .065$).

Trail Foot: Posterior separation of the trail foot from the CoM was smaller (6cm); toe $F(1, 54) = 45.6$, $p < .01$; heel $F(1, 54) = 20.9$, $p < .01$.

8.3 Chapter Summary

The four most notable ageing effects on the Safety Zone are 1) non-dominant foot's toeing-in during width controlled walking, 2) less consistent ML CoM control 3) more posterior swing foot location from the CoM at MFC and MLM and 4) less AP separation of both feet relative to the CoM.

Older adults' non-dominant *lead* foot's toeing-in extended the anterior boundary of the Safety Zone at toe-off and heel contact. When in *trail* foot, however, toeing-in reduced the AP Safety Zone. Non-dominant foot's toeing-in appeared only when older adults walked with controlled width most clearly in narrow and less in wide walking.

Although step width variability was not differentiated between the age groups (Chapter 8), the more variable ML CoM position within the Safety Zone was characterised in the older group, possibly reflecting age-related loss of ML balance control.

Swing foot location at MFC and MLM were found to be more posterior in older adults. Consequently, swing foot at MLM was relatively more parallel to stance foot, causing the shorter anterior boundary of the Safety Zone. Consistent with the previous finding (Lugade et al., 2011), older adults showed smaller AP separation of both feet from the CoM at toe-off and heel contact possibly reflected in reduced step length in older adults.

Width controlled walking reduced ML CoM control especially in the older adults' non-dominant stance foot. Walking on lines can be therefore considered to assist the more consistent ML Safety Zone control.

CHAPTER 9

Gait Data Computation for Balance Simulation

9.1 Overview

One application of the Safety Zone Model is in devising methods for balance loss simulation due to any perturbations to either CoM or foot, as outlined in section 3.3. This chapter presents balance loss simulation using the experimental data obtained from both young and older adults during preferred step width walking. All the balance loss simulations at the examined swing phase events required segment mass of CoM and for this reason, the transverse pelvis CoM was used for simulation as proposed by Kingma et al. (1995). Except for lateral balance loss simulation where perturbation is modelled directly into CoM, it was necessary to estimate how perturbation of the foot would affect CoM movement. Following the inverted pendulum model, this concept can be described by expressing perturbation as kinetic inputs on foot that generates net force transferred to CoM.

9.2 Results

The leg length and segment mass of the foot and pelvis for the young and older groups were obtained by Visual 3D modelling (Table 9.2.1).

Table 9.2.1 Average segment mass of foot and pelvis and lower limb length of the two age groups. Lower limb length = from lateral malleolus to greater trochanter

		Young	**Older**
1	Foot Segment Mass (kg)	1.10	1.11
2	Pelvis Segment Mass (kg)	10.7	10.9
3	Leg Length (m)	0.81	0.77

9.2.1 Simulation of Anterior Balance Loss due to Posterior Forefoot Slipping

At toe-off, if push-off force exceeds the friction force available, the forefoot will slip in the posterior direction. A simulated slippery walking surface was defined as CoF = 0.10 (Nagano et al., 2013) to compute the friction force based on ground reaction force (GRF) characteristics at RCoF2 (**4** in Table 9.2.2). As the horizontal GRF component at RCoF2 exceeds friction force, net posterior force can be divided by foot segment mass (**1** in Table

9.2.1) to obtain posterior toe acceleration (**5**). Anterior CoM acceleration (**6**) is based on net posterior force divided by pelvis segment mass (**2**), and with anterior CoM velocity (**7**), movement simulation (detailed in Figure 3.4.2) can estimate the time taken for CoM to reach the posterior boundary of safety zone (**8**).

Table 9.2.2 *Forefoot Posterior Slipping Simulation*

		Young		Older	
		Dom	**Non**	**Dom**	**Non**
4	Friction (N) at Cof = 0.10	77.4	76.3	84.6	77.4
5	Posterior toe acceleration (m/s^2)	23.2	20.6	23.5	27.1
6	Anterior CoM acceleration (m/s^2)	2.39	2.12	2.39	2.76
7	Anterior CoM velocity (m/s)	1.36	1.34	1.04	1.05
8	ART due to forefoot posterior slipping (s)	0.03	0.03	0.04	0.04

9.2.2 Simulation of Posterior Balance Loss due to Anterior Heel Slipping

Ground reaction forces (GRF) data used for the simulation of anterior heel slipping are taken from ghe first peak of required coefficient of friction (RCoF1) during preferred width walking. The vertical component of GRF and a simulated slippery walking surface of CoF = 0.10 (Nagano et al., 2013) give friction force (**9** in Table 9.2.3). Anterior heel acceleration (**11**) is based on net positive force created by horizontal GRF subtracted by friction force and (**9**) divided by foot segment mass (**2** in Table 9.2.1). Net horizontal force divided by pelvis segment mass computes CoM acceleration due to slipping. Integrating original velocity state of heel (**10**) and the CoM (**13**), the coordinates of both heels define the boundary and posterior balance loss initiation due to heel anterior slipping, and ART (**14**) can be calculated.

Table 9.2.3 *Anterior Heel Slipping Simulation*

		Young		Older	
		Dom	**Non**	**Dom**	**Non**
9	Friction (N) at CoF = 0.10	56.0	53.5	37.6	42.3
10	Heel contact velocity (m/s)	0.60	0.56	0.42	0.34
11	Anterior heel acceleration (m/s^2)	93.9	99.9	65.2	74.9
12	Acceleration on CoM due to slipping (m/s^2)	9.65	10.27	6.64	5.87
13	Anterior CoM velocity (m/s)	1.56	1.53	1.35	1.36
14	ART due to heel anterior slipping (s)	0.11	0.41	0.13	0.20

9.2.3 Simulation of Lateral Balance Loss at MLM

Lateral balance loss simulation was different from tripping and slipping in that a kinetic description of balance disturbance could not be estimated from the obtained gait data. In this case, the approach was to compute the minimum lateral force required for balance loss in the lateral direction. The simulation results revealed that young adults were unlikely to experience lateral balance loss due to unexpected lateral balance perturbation at MLM. This is primarily because the stance toe was not located sufficiently anterior to the CoM, approximately only 2cm. It was estimated that the CoM would pass the AP location of the stance toe within 0.02s and there would be no risk of lateral balance loss. Lateral forces >200,000N at dominant MLM and >100,000N at non-dominant MLM would cause the CoM to cross the toe-heel line of the stance foot for the young group. In the older group, more anterior stance toes (dominant: 8.9cm and non-dominant: 10.8cm) relative to the CoM and lower CoM horizontal velocity would result in 0.08s and 0.11s, respectively to lateral balance loss.

9.2.4 Simulation of Anterior Balance Loss due to Tripping

ART for tripping-related anterior balance loss at MFC has been illustrated in Figure 3.4.4. Foot segment mass (**1**) and toe acceleration (**15** in Table 9.2.4) determine Estimated Force for tripping at MFC (**16**). Impact forces due to tripping are transferred as resultant linear CoM acceleration (**18**), estimated by dividing estimated force (**16**) by the pelvis segment mass (**2**). The linear component of acceleration (**18**) coupled with CoM linear velocity (**17**) and distance to anterior balance loss (**19**) calculated ART due to tripping (**20**).

Table 9.2.4 *Tripping Simulation*

		Young		**Older**	
		Dom	**Non**	**Dom**	**Non**
15	MFC toe acceleration (m/s²)	1.37	1.42	1.42	0.69
16	Estimated Force (N)	1.51	1.56	1.56	0.76
17	CoM linear velocity at MFC (m/s)	1.30	1.32	0.92	0.89
18	Linear CoM acceleration due to tripping (m/s²)	0.14	0.14	0.14	0.07
19	Distance to Forward Balance Loss (cm)	11.6	11.7	13.0	11.7
20	ART due to Tripping (s)	0.09	0.09	0.14	0.13

9.3 Chapter Summary

This chapter applies established balance loss simulation methods (Chapter 3) into the empirical gait data. For slipping and tripping simulations, balance simulation has to be conducted first in the sagittal plane to obtain acceleration on a foot and the CoM. Then, the transverse simulation was possible to compute available response time (ART), time taken for the CoM to be dislocated from the Safety Zone. The sagittal plane simulation is based on ‘force couple’ as illustrated in Figure 3.4.1. While balance can be described by the coordinates and velocity of both feet relative to the CoM, balance disturbance can be described as kinetic inputs, therefore acceleration given to change balance status. The Safety Zone characteristics have been described in the previous Chapter 8, but here, the major differences between the age groups for ART calculation have been attributed to anterior velocity of the CoM and a foot. While older adults were considered to require longer ART due to impaired reaction speed, no such evidence has been observed. Simulation of lateral balance loss does not have rationales to determine the adequate kinetic inputs as balance disturbance. Inverted pendulum based force couple methods were not applicable, and therefore, the simulation is for computing time for the CoM to possibly cross the lateral boundary of the Safety Zone and also the minimum required lateral force to cause balance loss lateral to the stance foot was computed. The results imply that minimum lateral margin (MLM) was a risky event for lateral balance loss only for the older population.

CHAPTER 10

Effects of Insole on Balance Loss and Injuries Risks

10.1 Overview

Due to lower cost, easy engagement and wide applicability, shoe-insoles deserve research attention to advance the possibility of insole interventions to prevent falls and lower limb injuries associated with walking. The current study tested two ankle joint motion support functions, 2.2º dorsiflexion and 4.5º eversion, with the aim of incorporating these features in shoe-insoles to reduce potential balance loss and impact at loading. Despite change to ankle orientation, the fundamental gait pattern as described by spatio-temporal parameters was not anticipated to change, based on previous findings (Silver-Thorn et al., 2012).

The first primary function of the insole was to elevate foot-ground clearance by increasing dorsiflexion at MFC. In addition to the MFC-height ($MFC_{h)}$ median and IQR, full analysis of the MFC_h distribution pattern was conducted. Although increased MFC_h was hypothesised due to dorsiflexion support, it was unknown as to how the insole would affect the MFC_h distribution. The insole eversion function was designed to reduce lateral balance loss possibly by increasing MLM_d, reducing lateral CoP displacement and increasing mean medial CoM acceleration.

Ankle dorsiflexion and eversion have not been previously reported to be advantageous in reducing the risk of anterior heel slipping. One concern, however, remained as dorsiflexion support at heel contact, increasing foot contact angle, has been considered to disadvantageous in attaining lower RCoF (Brady et al., 2000; McGorry et al., 2008; 2010). Dorsiflexion and eversion are ankle joint motions that may promote energy efficient loading to enhance shock absorption during walking. Furthermore, dorsiflexion triggers knee flexion, necessary for impact reduction at the knee joint after heel contact (Nakajima et al., 2009; Radzimski et al., 2012). In summary, it can be hypothesised that the insole's effect on impact absorption would lead to greater recovery rate, prolonged time to foot flat, reduced knee adduction moment, increased dorsiflexion and greater knee flexion. The insole would cause only minor changes in gait (e.g. less than 1cm increase in MFC_h), which could, however, possibly reduce the risk of balance loss and lower limb injuries by correcting sub-optimal gait features.

10.2 Results

A 2×2×2 (age × insole x limb) repeated measures mixed model Analysis of Variance (ANOVA) was applied to specifically examine the insole effects on spatio-temporal parameters, minimum foot clearance (MFC), minimum lateral margin (MLM) characteristics and a series of kinematic and kinetic variables during the loading response. Interaction effects are obtained when the insole effects are specific to the age groups or either limb.

10.2.1 Spatio-temporal Parameters

Figure 10.2.1 illustrates the insole effects on spatio-temporal parameters.

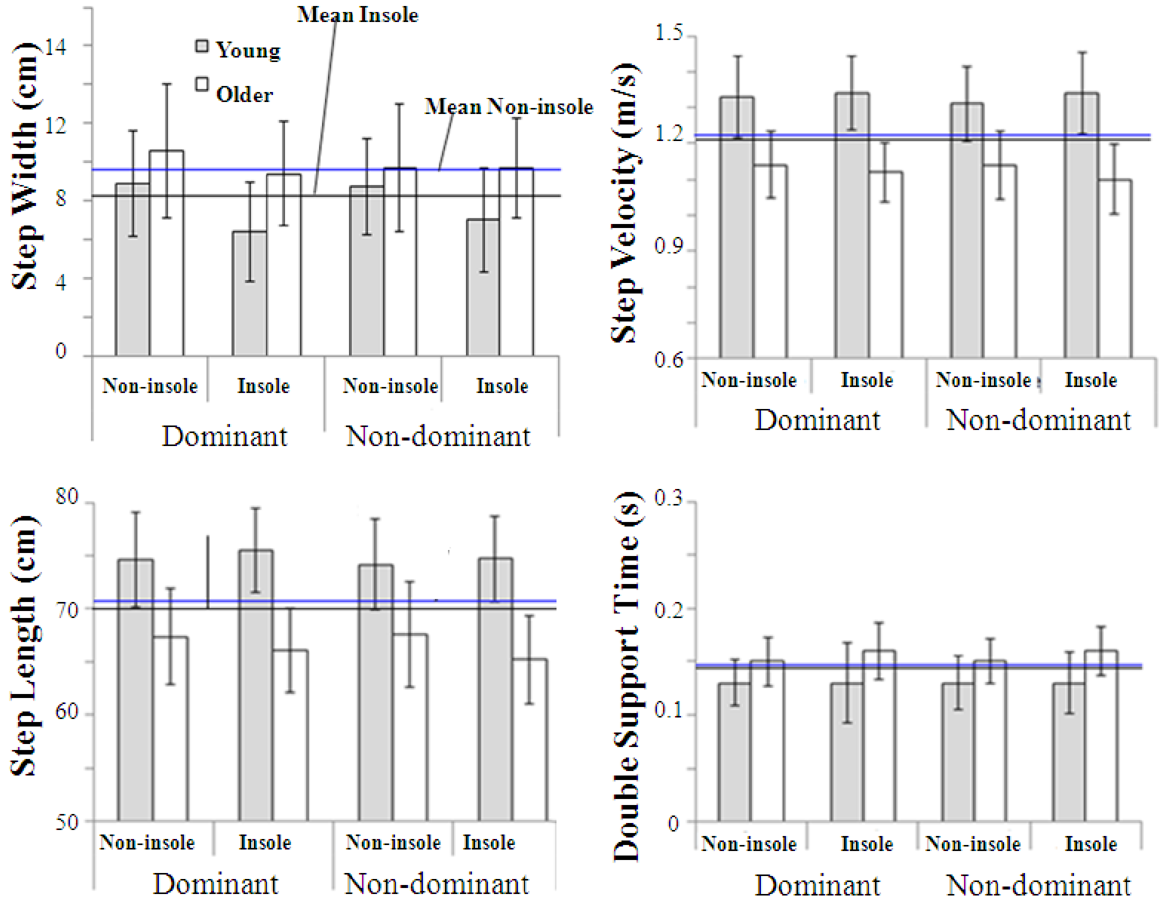

Figure 10.2.1 Effects of insole on step width, length and velocity and double support time.

Step width reduced in the insole condition, but no statistical difference was identified. Overall, the insole did not have any significant effects on mean ± SD descriptions of spatio-temporal parameters.

10.2.2 Effects of the Insole on Minimum Foot Clearance (MFC)

The insole significantly increased median MFC_h (F (1, 54) = 96.7, p <.01) most noticably in older adults' dominant MFC_h, with an increase up to .43cm (F (1, 54) = 10.7, p <.01).

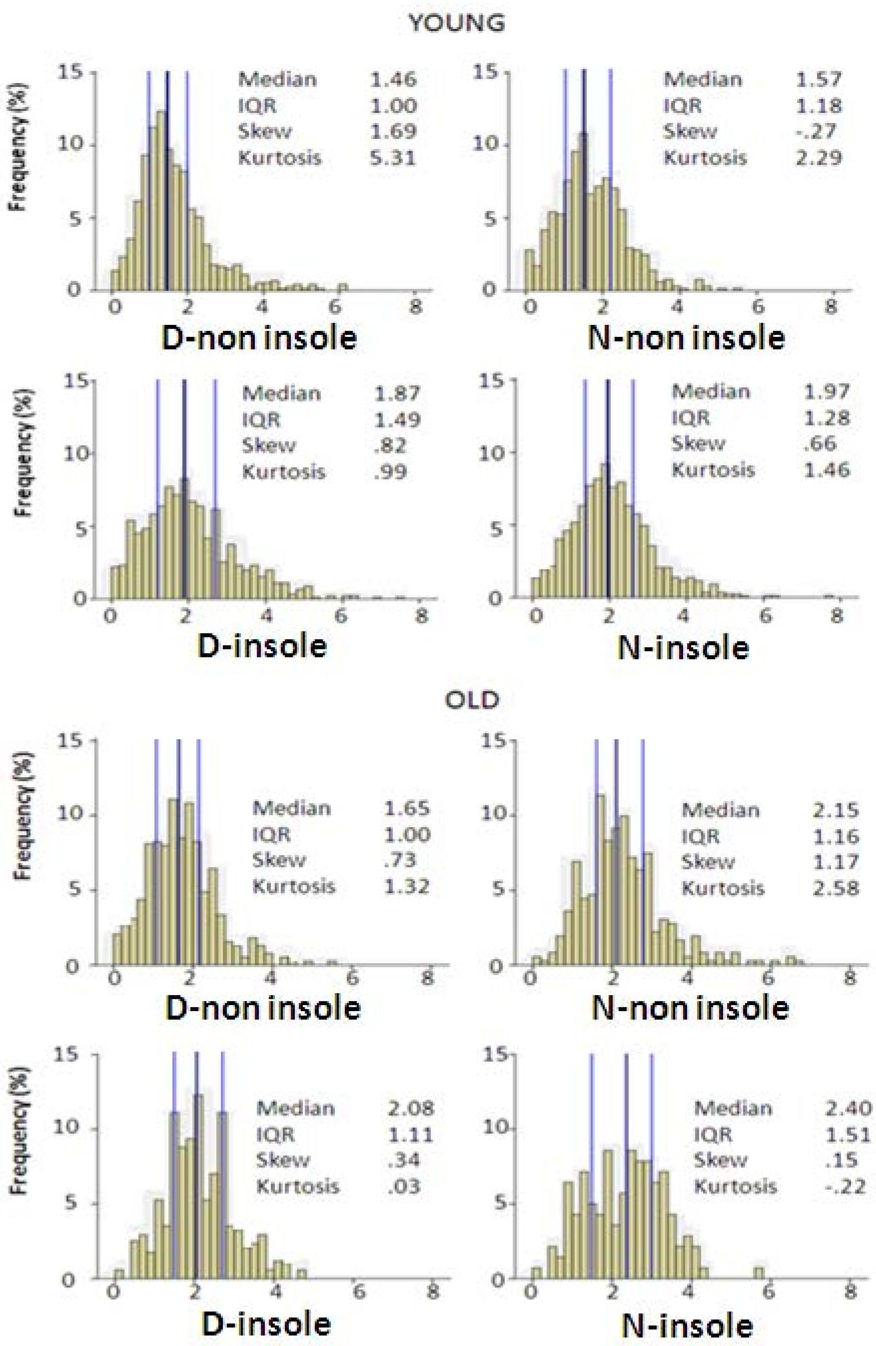

Figure 10.2.2 MFC_h histograms comparison between young and older adults; dominant and non-dominant limbs; and normal walking and insole conditions. Blue lines: left, 25 percentile; middle (bold) median = 50 percentile; and right, 75 percentile.

If increase in MFC_h is due to the shift of the whole MFC_h distribution pattern rather than just an increase in median MFC_h, IQR zone should move to the positive direction. Such effect was observed for dominant MFC_h of the older adults and for non-dominant MFC_h of the young adults to a lesser extent. In contrast, the dominant foot of young adults showed similar lower (25%) percentile despite elevated clearance if compared by median or 75 percentile, resulting in the less kurtotic histogram. In the non-dominant foot of the older adults, higher median and 75% MFC were seen when wearing insole, but the 25% percentile decreased, causing higher variation (IQR) and negative kurtosis.

Figure 10.2.3 illustrates that timing of MFC (MFC_t) was not affected by the insole.

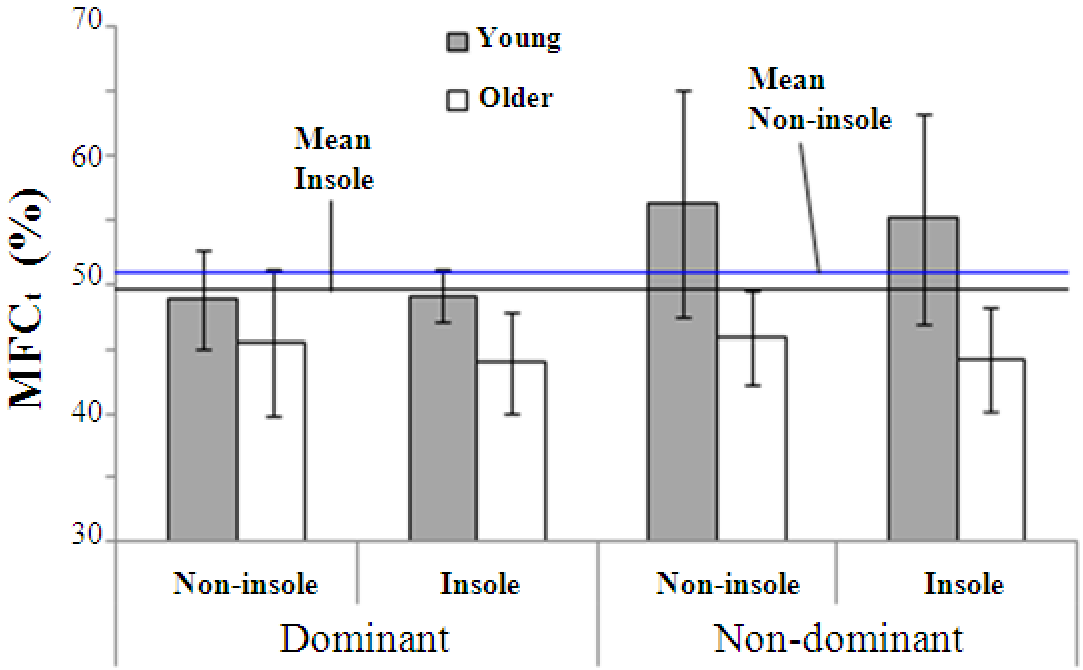

Figure 10.2.3 Effects of insole on MFC_t

In summary, the insole elevated MFC_h with wider distribution with no substantial changes in MFC_t. MFC_h control is a consequence of multiple joint-movements and dorsiflexion support of insole was expected to increase dorsiflexion status of ankle at MFC and possibly associated knee flexion (Figure 10.2.4). Wearing insole increased dorsiflexion status of ankle by 0.7° ($F(1, 54) = 4.7$, $p < .05$). Greater knee flexion was also seen ($F(1, 54) = 10.9$, $p < .01$) except the older adults' dominant knee (age x footwear x limb: $F(1, 54) = 5.7$, $p < .05$). Relative to the dominant limb, the non-dominant limb received a greater knee flexing effect of the shoe insole ($F(1, 54) = 5.7$, $p < .05$).

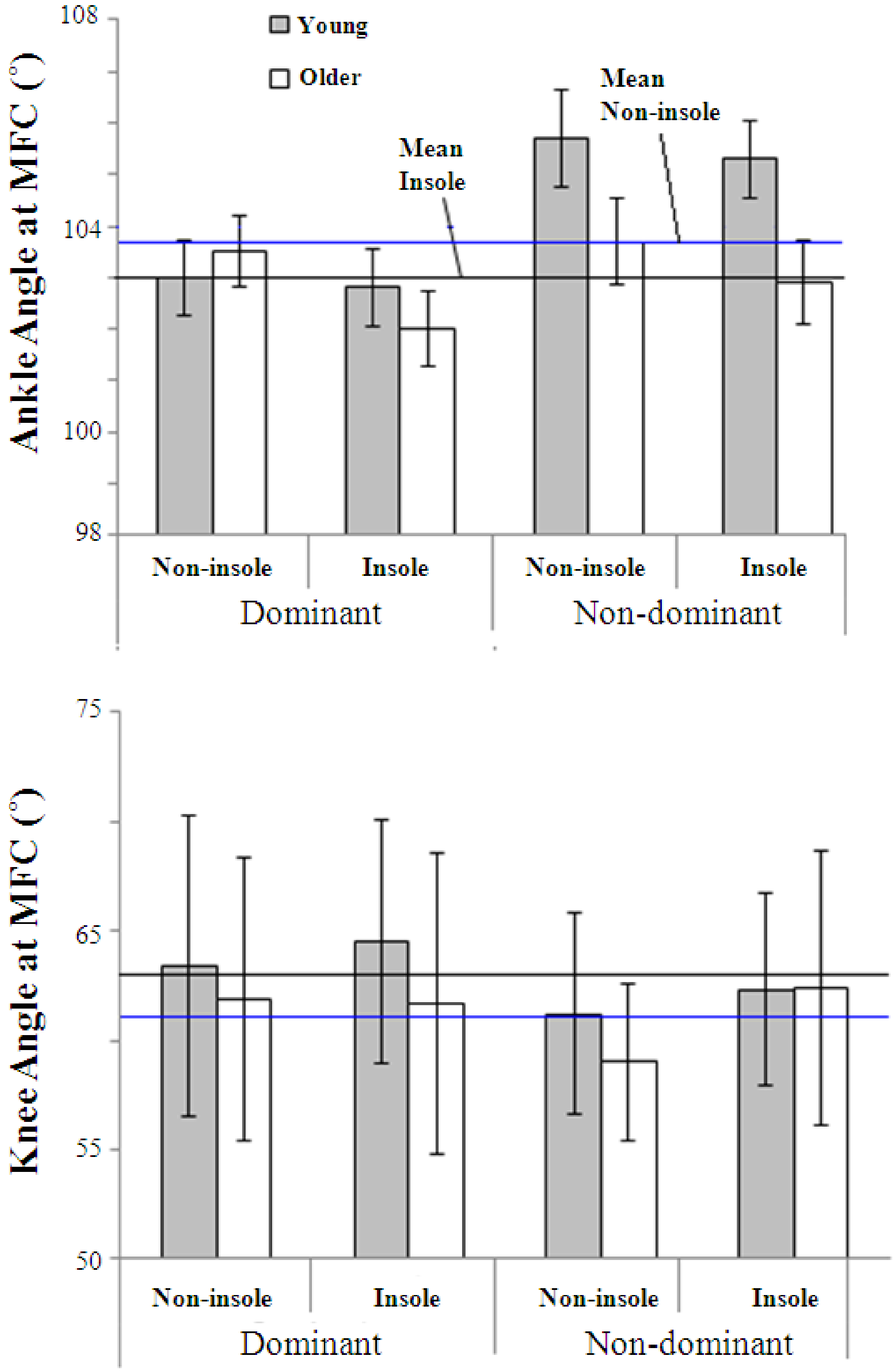

Figure 10.2.4 Insole effects on knee and ankle angles at MFC.

10.2.3 Effects of the Insole on ML Balance

MLM_d increased ($F(1, 54) = 20.2$, $p < .01$) and lateral CoP displacement decreased ($F = 8.4$, $p < .01$) when wearing the insole, but medial acceleration on CoM from toe-off to contralateral MLM did not change due to the insole.

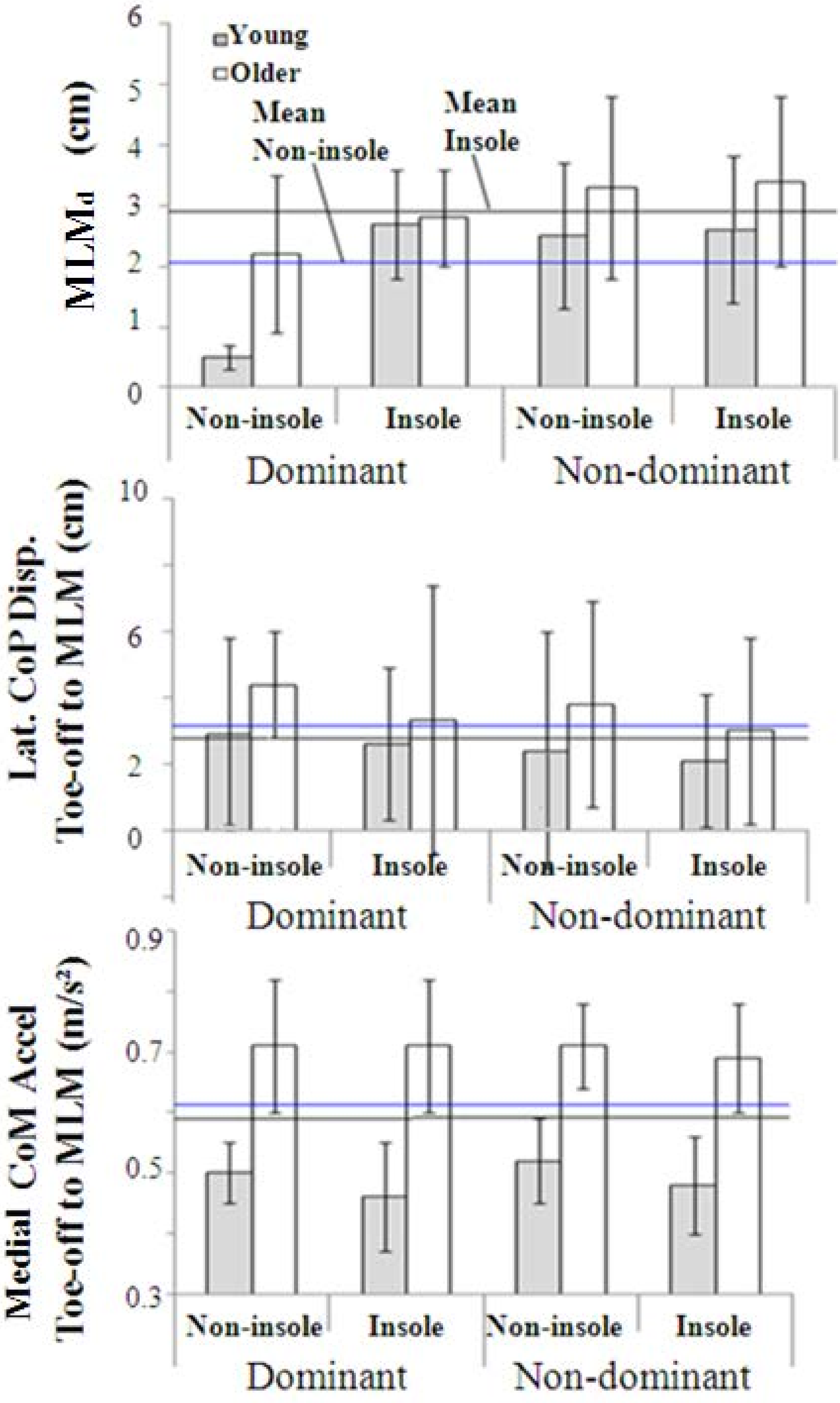

Figure 10.2.5 Effects of Insole on ML Balance Control

As described in Figure 10.2.3, the insole delayed MFC_t ($F (1, 54) = 21.3$, $p < .01$) but the delay was more clearly seen in young adults (age x insole: $F (1, 54) = 10.5$, $p < .01$). The dominant limb indicated a stronger insole effect by increasing MLM_t further (limb x insole: interaction as seen in Figure 10.2.6 ($F (1, 54) = 11.4$, $p < .01$).

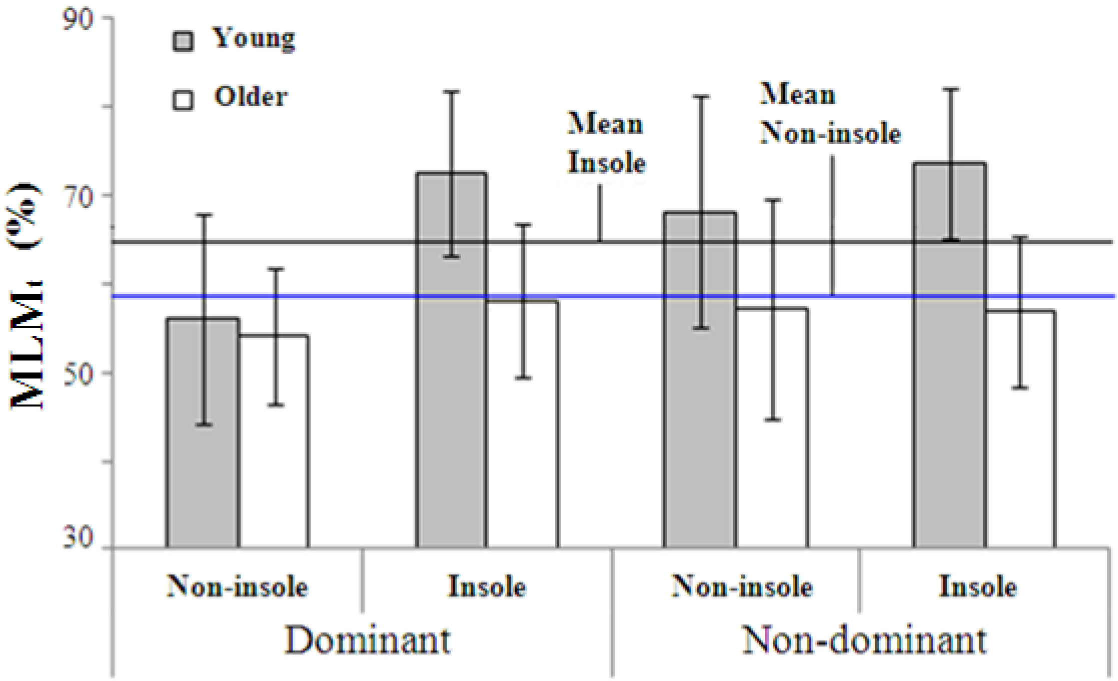

Figure 10.2.6 Insole *effects on MLM_t*

10.2.4 Effects of the Insole on Heel Contact Characteristics for Slipping Risk

Slipping is reflected in heel contact kinematics as greater heel contact angle and lower foot contact angle may protect against anterior heel slipping (e.g. McGorry et al., 2008). As shown in Figure 10.2.7 the insole increased foot contact angle ($F (1, 53) = 5.9$, $p < .05$) but this increase was more clearly seen in the older subjects by more than 2° (age x insole: $F (1, 53) = 5.4$, $p < .05$). An increase in heel contact angle due to the insole was seen only in the older adults' dominant limb (age x insole x limb: $F (1, 53) = 6.4$, $p<.05$).

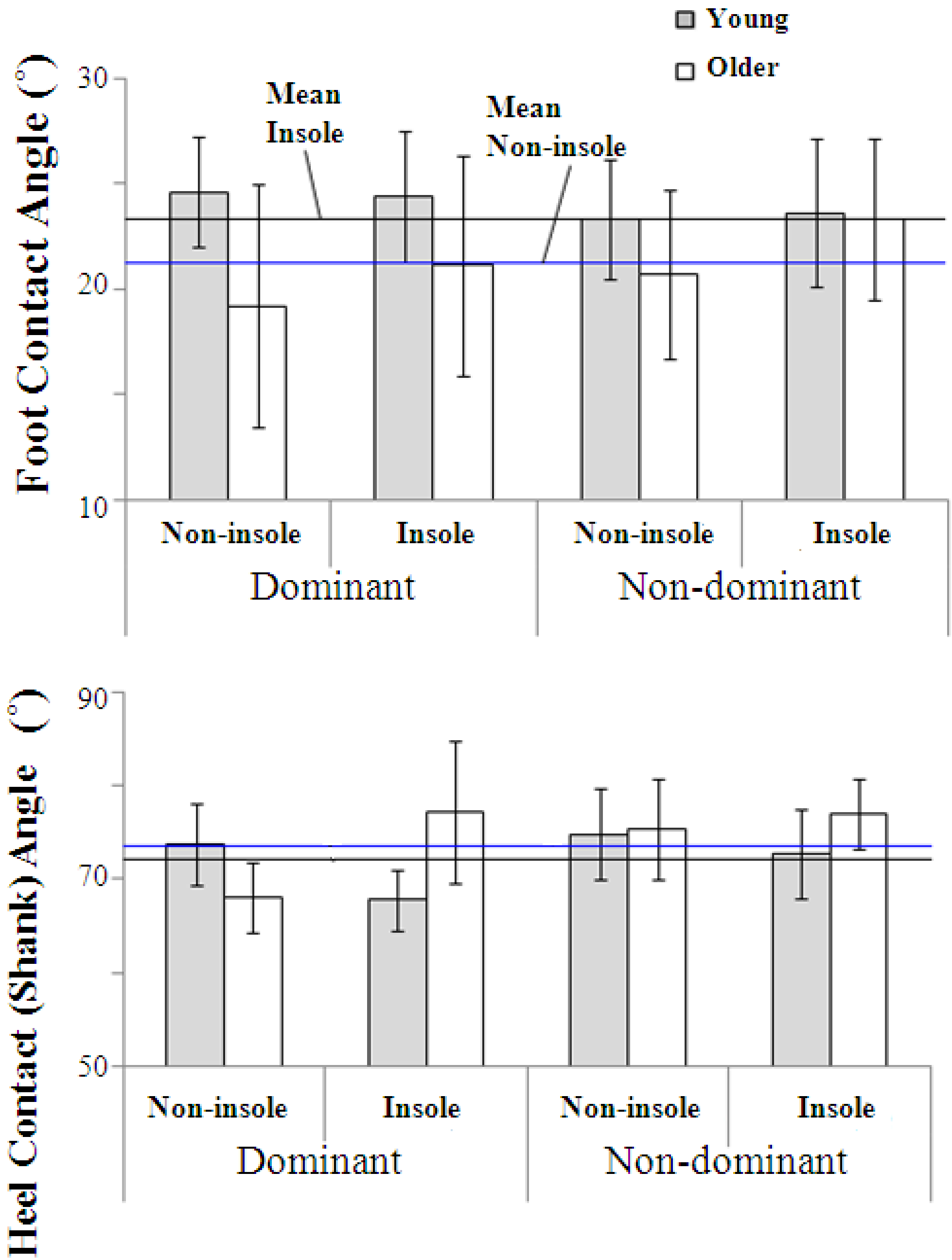

Figure 10.2.7 Insole effects on foot contact angle and heel contact angle associated with anterior heel slipping

10.2.5 Insole Effects on Loading Response

The transition from swing to stance phase always accompanies impact at heel contact and ankle dorsiflexion and knee flexion are necessary for shock absorption at this event.

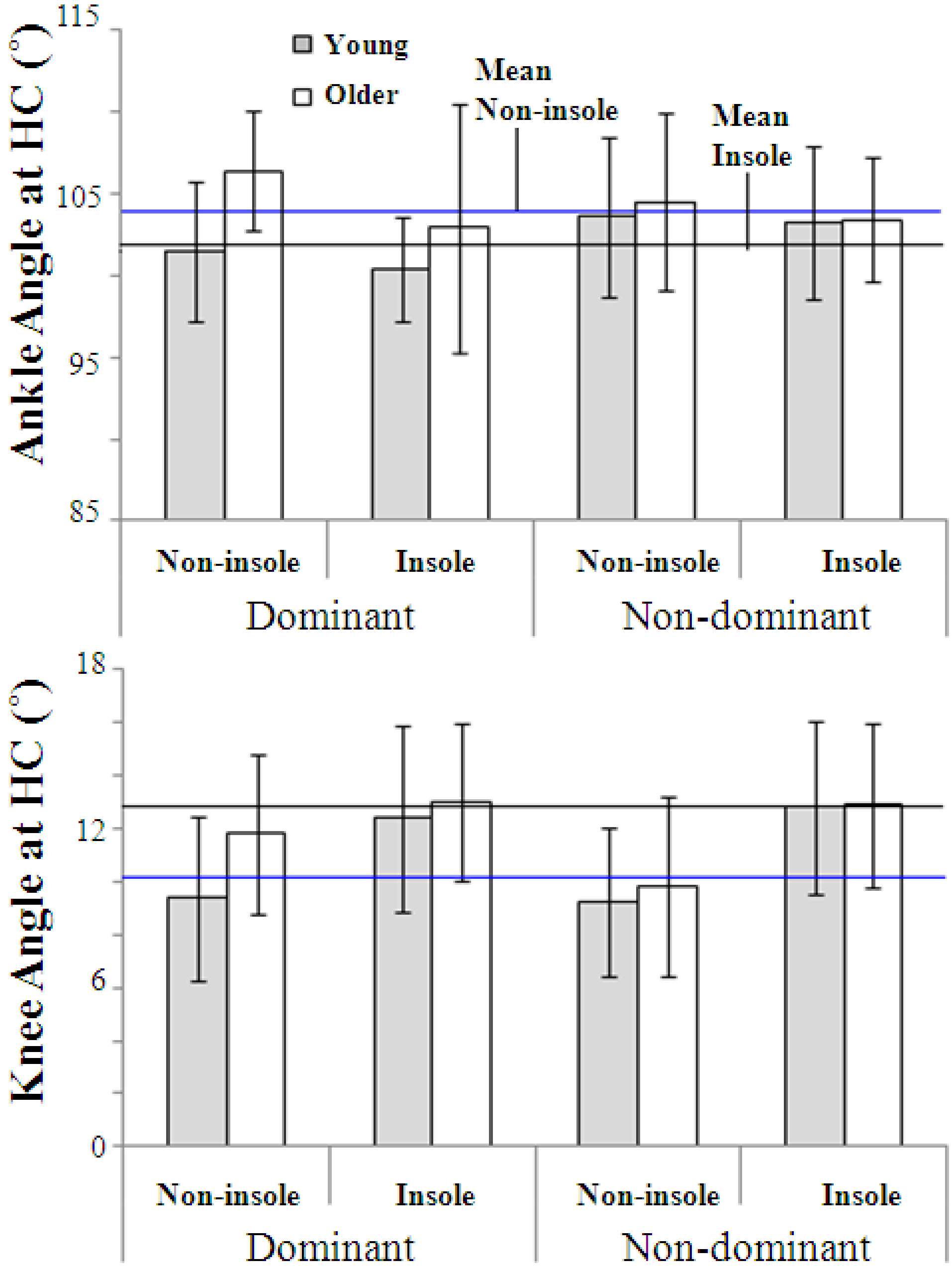

Figure 10.2.8 Insole Effects on Lower Limb Joint Angles at Heel Contact

Figure 10.2.8 demonstrates that the insole supporting dorsiflexion of 1.5° (F (1, 54) = 17.0, p <.01) and knee flexion of 2.7° (F (1, 54) = 54.2, p <.01) at heel contact were seen but only for the younger group (age x insole: F (1, 54) = 8.7, p<.01).

The effects of the insole on impact reduction are summarised in Figure 10.2.9 below.

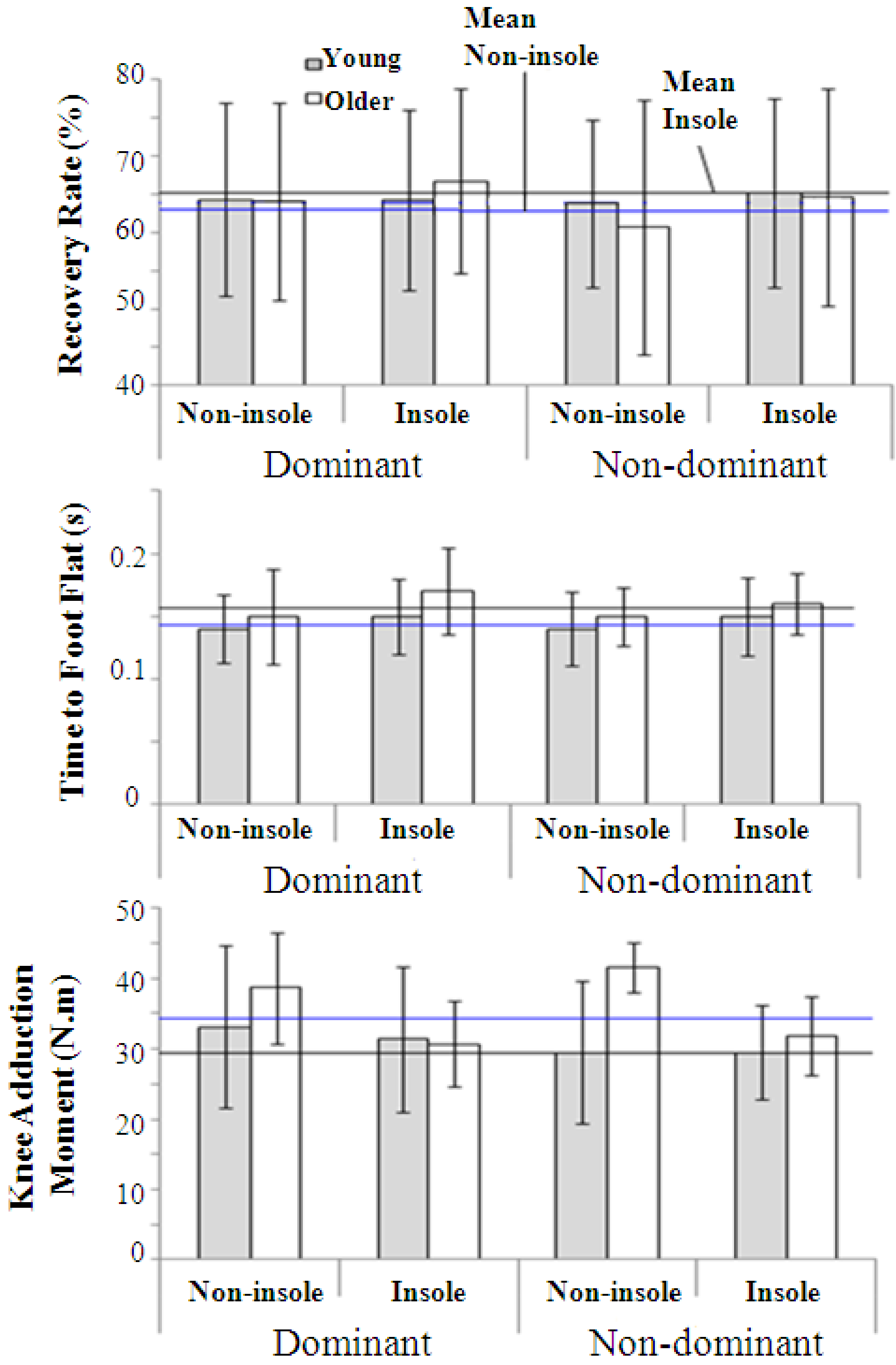

Figure 10.2.9 Insole Effects on Loading Response

An increased recovery rate of 2% was confirmed in insole walking (F (1, 54) = 4.1, p<.05). The insole also prolonged time to foot flat by .015s (F (1, 54) = 39.0, p<.01). Reduction in knee adduction moment was also seen (F (1, 54) = 18.0, p<.01), especially in older adults (age x insole: F (1, 54) = 12.3, p <.01).

10.3 Chapter Summary

Comparisons between non-insole and insole walking revealed no differences in spatio-temporal parameters. The insole increased MFC_h but also led to greater variability and lower kurtosis in the MFC_h distribution. Greater MFC_h was due to increased ankle dorsiflexion and knee flexion by the insole. As expected, the insole increased MLM_d and decreased lateral CoP displacement, suggesting that the insole was effective in preserving ML balance. Older adults increased foot contact angle due to ankle dorsiflexion support, suggesting an increased risk of anterior heel slipping. The insole's ability to reduce shock at heel contact was confirmed by greater recovery rate, prolonged time to foot flat and reduced knee adduction moment. These effects were accompanied by dorsiflexion and knee flexion support at heel contact.

Chapter 11.

Discussion

The first broad aim of the Thesis was to investigate ageing effects on balance control during walking by examining four critical swing phase events including toe-off, minimum foot clearance (MFC), minimum lateral margin (MLM) and heel contact. Step width effects on balance control were examined by employing ±50% width walking as experimental conditions. In addition to extending established concepts of required coefficient of friction (RCoF), MFC and MLM, a Safety Zone Model was developed to further characterise dynamic balance especially at four swing phase events in two principal planes of motion (i.e., horizontal and sagittal). A further component of the current study was to use the Safety Zone Model to simulate balance loss. As detailed in Chapter 3, balance perturbation was described as kinetic inputs to the CoM and associated changes in kinematics were estimated to simulate change in the CoM position and the Safety Zone.

In addition to balance-related gait biomechanics, three applications of the Thesis are discussed in this chapter. First, the shoe insole was tested to determine whether ankle joint support would modify gait to maintain balance and reduce knee joint moments at heel contact. Second, the experimental conditions of width-controlled walking revealed positive effects on balance control and the application of these findings to gait training is discussed. Third, the Safety Zone Model's application to clinical settings is considered with respect to assessing balance and informing clinical decision making in older adults and other gait-impaired populations.

11.1 Effects of Ageing and Step Width Manipulation on Gait Characteristics

Older adults often fall during locomotion due to age-related impairments to balance. The gait characteristics of older adults have been described using spatio-temporal parameters in this study and were consistent with the previous research (Hollman et al., 2007; Macellari et al., 1999; Nagano et al., 2012; Whittle, 2007; Winter, 1991), including slower step velocity due to 11.7cm shorter step length and prolonged double support. Increased double support by 4% during width controlled walking was confirmed when normalised to step time, confirming a qualitative increase regardless of step time duration. These ageing effects appeared most clearly in narrow width walking and were also seen in wide walking compared to preferred width walking, consistent with the previous study by Schrager et al. (2008). While the older group showed 1.3cm greater step width than the young, the difference was not however statistically significant. As also previously implied (Nagano et al., 2012), ageing-associated increase in step-width tends to be minimum in the least obstructed overground walking environment, especially for the highly healthy older individuals recruited as part of the current research. Changes to preferred step width were, however, found to negatively affect balance even for the healthy older population.

Maintaining the CoM within the Safety Zone was considered more difficult in narrow walking where the distance between the two lateral boundaries reduced (step width: preferred = 9.9cm vs. narrow = 5.4cm). Slowing down with prolonging double support time was, therefore, considered to reflect an attempt of the precise CoM control within the Safety Zone. Wide walking (step width = 14.8cm) was expected to provide functional advantage in the extended ML Safety Zone, but step velocity and associated step length reduced compared to preferred width walking possibly because any alteration to preferred walking styles can affect forward progression. While the greater ML Safety Zone secures balance, shorter step length reduced the anterior boundary, therefore increasing anterior balance loss.

Another possibility accounting for typical cautious gait adaptations in spatio-temporal parameters can be due to unfamiliarity of walking with controlled width. As previously discussed, typical ageing effects on spatio-temporal parameters tend to emerge in challenging walking conditions (Bock and Beurskeus 2010; Dunlap et al., 2012; Hollman et al., 2007; Ko et al., 2007; Nagano et al., 2012; Nordin et al., 2010; Schrager et al., 2008). For older adults, narrow walking was therefore, most challenging, followed by wide walking, with preferred width walking the least. Gait adaptations due to step width manipulation can be attributed not only to changes in functional requirement specific to controlled step width but also to challenges due to unfamiliarity that could trigger preparatory responses against perceived hazards (section 2.5; Menz et al., 2007; Donoghue et al., 2012; Prince et al., 1997). Young adults were free from any changes in spatio-temporal parameters in response to different step widths. Accordingly, step width manipulations influenced only older adults, implying walking with decreased or increased widths does not require *functional* adaptations to spatio-temporal parameters for healthy young individuals, but could also be secondary to perception of a challenging walking environment for a population with impaired adaptability. This may lead to a fear of falling that affects gait patterns.

When comparing the two width-controlled walking conditions, typical ageing effects were more evident under narrow than wide walking. Increased double support time variability in older adults' gait was, however, unique in narrow walking possibly due to more attention demanding walking requirement and also secondary to slower walking speed (Dubost et al., 2006) suggesting reduced automaticity to maintain consistent step timing (Callisaya et al., 2010). Once rhythmic gait automaticity is impaired, unexpected gait perturbations may occur more frequently (Hausdorff et al., 2001). Loss of automaticity in consecutive gait cycles can negatively affect efficient use of mechanical energy during the loading response, from heel contact to toe-off (Kuo, 2007). Less energy efficient gait could lead to not only fatigue but also greater impacts transferred to the lower limb joints, including the ankles and knees (Ventura et al., 2011). If automaticity is lost and energy costs at loading increase in narrower step walking, it could also eventually cause development of lower limb joints' pain.

As illustrated in Figure 6.2.3, narrower walking increased the risk of lateral balance loss due to shorter distance to the lateral boundary, as with the previous studies (Ko et al., 2007; Åberg et al., 2010). Minor balance restoration is usually performed at each step, as reflected in step-to-step variation in step parameters (Wezenberg et al., 2011). During width controlled narrow walking, balance cannot be restored by taking a recovery step. The role of double support time then becomes to ensure balance during step-to-step transitions (Kuo, 2007) before we move to the next single support phase. Increased variability in double support time could be therefore interpreted as more active minor balance recovery attempt during double support when recovery steps are not possible.

In summary, young adults did not show step width effects on any gait parameters while older adults changed spatio-temporal parameters, with narrow walking most pronounced. Functional adaptations to the Safety Zone by increased and decreased step width and possible onset of fear of falling due to more challenging walking conditions may have accounted for these effects in older adults. Increased double support time variability in older adults' narrow walking may reflect continual balance adjustments, which disturb normal gait, do not require a recovery step but at the cost of gait automaticity.

11.2 Ageing and Step Width Effects on Balance at Four Swing Phase Events

Greater step width has been recognised as a gait adaptation to secure balance, while other typical ageing effects, such as reduction in step length and associated gait velocity, have been reported to impair stability (Åberg et al., 2010; Menz et al., 2007; Ko et al., 2007; Whittle, 2007). The current study is the first to investigate decreased and increased step widths effects on dynamic balance and related key swing phase events. This section discusses age-associated step widths effects with the aim of further understanding balance control in older adults. Two RCoF peaks, before toe-off and after heel contact, were examined to characterise the likelihood of heel anterior and forefoot posterior slipping under different step widths. In contrast, kinematics of the mid-swing events, minimum foot clearance (MFC) and minimum lateral margin (MLM), were investigated to evaluate gait patterns in terms of anterior and lateral balance loss during single support.

11.2.1 Toe-off (RCoF2) and Heel Contact (RCoF1)

The required coefficient (RCoF) peaks and vertical ground reaction force (GRF) data (Figure 7.2.3 and Figure 7.2.4) were comparable to Chang et al. (2008), who reported vertical GRF of approximately 650N and 100N at the first and second RCoF peaks (Figure 2.7.3), respectively. Obtained RCoF values at the two peaks were also similar to previous studies (e.g. Chang et al., 2008; Lockhart et al., 2009). It is important to emphasise again that the examined GRF characteristics were at the RCoF peaks where the risk of slipping was considered maximal, but were different from the GRF peaks where approximately 110% of body weight was seen (e.g. Taylor et al., 2005).

Older adults showed lower RCoF2, especially in narrow walking, suggesting a reduced risk of forefoot posterior slipping. Lower RCoF2 can be attained by proportionally lower horizontal GRF and therefore, higher vertical GRF (Leclercq, 1999). Weaker horizontal push-off GRF was found to be linked to slower step velocity and shorter step length (Marasovic et al., 2009; Simmonds et al., 2012). Contrary to the expectation lower horizontal GRF at RCoF2 was not identified in older adults. It was interesting that older adults had greater GRF at the negative RCoF peak where posterior forefoot slipping is most likely, but RCoF2 does not match peak push-off forces. Despite greater GRF in the older group at RCoF2, the increase was more prominent in the vertical component by 70N, accounting for overall reduction in RCoF2 (Figure 7.2.3).

Swing phase termination at heel contact is also associated with slipping due to heel anterior sliding when the surface coefficient of friction (CoF) is lower than RCoF1 (Cham and Redfern, 2002; Fong et al., 2009; Li et al., 2004; Lockhart et al., 2003; Figure 2.7.3). Ageing did not affect RCoF1 despite flatter foot contact in older adults, consistent with previous research (Perry et al., 2007; Brady et al., 2000; McGorry et al., 2008; 2010). Nevertheless, flatter foot contact does not facilitate efficient toe-off because reduced dorsiflexion limits the Achilles Tendon's capacity to store impact forces as elastic energy (Lichtwark and Wilson, 2006; Lidtke et al., 2010; Silver-Thorn et al., 2011; Ventura et al., 2011).

From an applied perspective, dorsiflexion support at heel contact provided by the shoe insole in the current project should enhance the energy efficiency of loading and minimise forces transferred to lower limb joints. As increased foot contact angle encourages efficient loading, prevention of anterior heel slipping should rely more on external adaptations, such as anti-slip material on the heel (Aschan et al., 2005; Kim et al., 2001). Flatter foot contact cannot be recommended as a safety adaptation due to potential development of lower limb joint osteoarthritis (Moden et al., 2009).

Heel contact and toe-off kinetics revealed that older adults showed lower GRF than young adults at RCoF1 immediately following heel contact, while contrary to expectation, greater ground reaction forces (GRF) were confirmed in older adults at RCoF2. Older adults may prefer softer landing, reflected in lower GRF at RCoF1 but this may be due to slower walking (Hsiang and Chang, 2002; Keller et al., 1996). Reduced impact at heel contact inhibits energy transfer to assist toe-off, possibly requiring increased push-off forces (Michel and Do, 2002; Miller and Verstraete, 1999; Savelberg et al., 2009; Vanderpool et al., 2008; Ventura et al., 2011). From the obtained GRF characteristics at RCoF1 and RCoF2, young adults generated greater GRF kinetic inputs at heel contact, and the impact was transferred through the loading response to initiate toe-off, requiring less additional push-off force toward toe-off. Schimitz et al. (2009) used electromyography (EMG) to confirm increased muscle co-activation prior to push-off in older participants supporting the idea that older adults have less efficient loading responses and, consequently, greater push-off force is demanded. It is important to emphasise again that the kinetic data presented here were for the two RCoF peaks which do not correspond precisely with previous reports of push off and braking (Figure 2.7.3). Implications from these findings to improve older adults' walking will be to enhance efficient loading during stance, measured by GRF characteristics with recovery rate.

11.2.2 Minimum Foot Clearance (MFC) and Minimum Lateral Margin (MLM)

The mid-swing phase events, MFC and MLM, have been linked to balance loss. Previous research on the timing, rather than magnitude, of mid-swing phase events is scarce and the only reports are that MFC_t is at approximately 50% of the swing phase (Begg et al., 2007; Levinger et al., 2012; Nagano et al., 2011; Winter, 1991). No examination of task or limb dominance effects on MFC_t has been conducted in the previous studies. For MLM_t, there are no previous findings. MFC usually precedes MLM but if there is high variability in MLM_t (i.e. variability of MLM_t exceeding 20%), this may not be the case. In older adults, these events generally occur earlier than for young adults. Figure 11.2.1 illustrates four consecutive right limb swing phases of a single older participant. The right swing foot's location is highly variable at MLM but MFC is relatively consistent, for example, MLM1 is close to heel contact while MLM2 follows shortly after MFC.

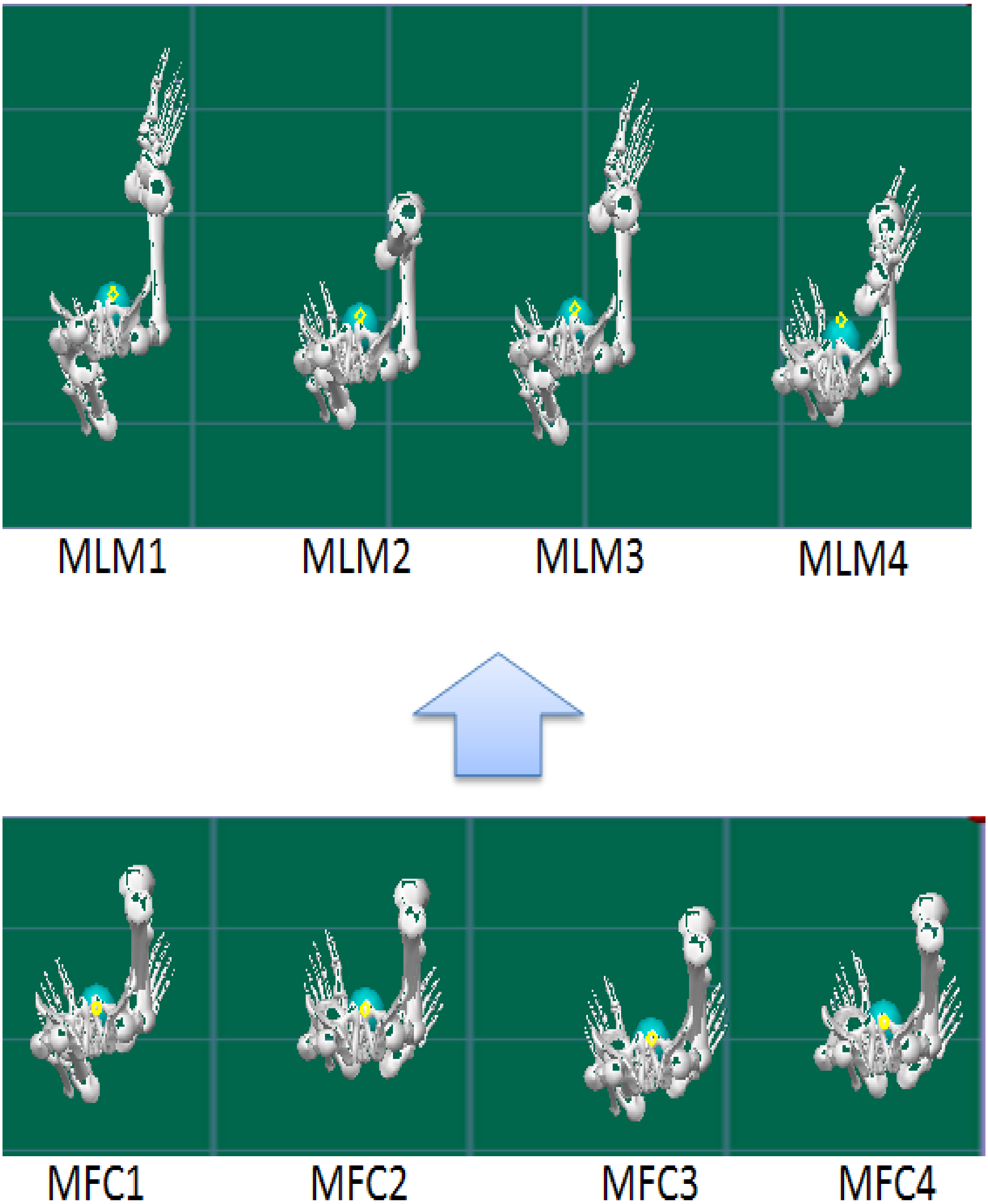

Figure 11.2.1 Timing of MFC (bottom) and MLM (top) in the transverse plane for four consecutive right foot swing phases. Data for a single older participant during preferred-width walking.

One direction for future research into swing phase control is to more comprehensively characterise the timing and variability of swing phase events from toe-off (0%) to heel contact (100%). These may include the first and second maximum ground clearance, Mx1 and Mx2 (Figure 2.2.4) and the timing of the contralateral stance limb's events: foot-flat, heel off and shank-vertical (Whittle, 2007). As with double support time variability, consistent timing of gait events across multiple gait cycles enhances gait automaticity with fewer gait perturbations (Frenkel-Toledo et al., 2005; Hausdorff, 2005). Treadmill walking is challenging for older individuals (Wass et al., 2005), but with adequate practice, it can be an ideal gait training method to learn gait patterns with minimised variability.

For tripping prevention MFC_h is required to be higher than the walking surface (Winter, 1991). While ageing increases the risk of falls, and tripping is a major cause (Blake et al., 1988; Berg et al., 1997), in the present experiment, consistent with previous studies (Begg et al., 2007; Barrett et al., 2010; Mills et al., 2008; Nagano et al., 2011; Sparrow et al., 2008; Winter, 1991) ageing did not affect MFC_h. Older adults were, however, found to have higher MFC_h in the non-dominant swing foot than on the dominant side (Nagano et al., 2011). Increased non-dominant MFC_h in the wide and narrow conditions, as seen in this research, can be interpreted as a safety adaptation because balance loss due to tripping by the non-dominant limb is more difficult to recover (Nagano et al., 2011; Perry et al., 2007; Pijnappels et al., 2008; Neptune et al., 2001; Winter, 1980).

In addition to safety adaptations, elevated MFC_h in controlled step width conditions can also be attributed to functional imperatives. While older adults showed safety adaptations in spatio-temporal parameters, as discussed, young adults did not change spatio-temporal parameters in response to the experimental conditions. Nevertheless, increased MFC_h was observed in both age groups. From an energetics perspective (Sparrow, 2000), higher MFC_h in controlled step walking is possibly achieved at low cost by utilising the surplus energy associated with more consistent step width control (Chou et al., 1997; Lu et al., 2012; Nagano et al., 2010).

Energy inputs via GRF from every step cycle need to be expended mainly through loading response after heel contact for step-to-step transition of the inverted pendulum cycle (Kuo, 2007), but when step width and the ML Safety Zone were controlled to be consistent, achieving less variability than natural walking, surplus energy was possibly yielded by restricting mechanical energy expenditure in ML stepping or the CoM movement. Higher swing foot clearance could accordingly reflect mechanical energy dissipation also implied by Lu et al., (2012), therefore consistently seen in both the older and young groups regardless of safety adaptations. This mechanics can be described as conversion of kinetic energy into potential energy of a swing foot (Nagano et al., 2010).

Further examination of width effects on MFC_h suggests that narrow walking caused higher median MFC_h but also increased variability in the older population and therefore, increased tripping risk (Begg et al., 2007; Barrett et al., 2010; Mills et al., 2008). In contrast, wide step walking elevated MFC_h without higher variability, suggesting that tripping risk was reduced. If gait adaptations due to fear of falling and mechanical energy utilisation discussed earlier (i.e. reducing ML movement of both the CoM and swing foot) account for increased MFC_h in width-controlled conditions, gait training procedures could be devised incorporating target lines with increased or preferred step width. With adequate practice, line-walking may elevate MFC_h and improve balance. The potential for line walking in gait training is discussed further in section 11.5.2.

Except for Perry et al. (2008) and Lugade et al. (2011) minimum lateral margin (MLM) characteristics have been investigated by a few researchers. The current research explored effects of ageing, step widths and limb dominance on MLM and provided new insights into lateral balance control during the swing phase by characterising *spatial* (MLM_d) and *temporal* (MLM_t) information. In addition lateral CoP displacement and the medial CoM acceleration between toe-off and MLM were analysed. Although older adults were expected to show safety adaptations to MLM including greater MLM_d, smaller lateral CoP and the higher medial CoM acceleration, no ageing effects were found to support these hypotheses.

When comparing the two limbs for ML balance control at MLM, the dominant limb tends to take larger burden in ML balance control compared to the non-dominant limb, as reflected in lower MLM_d and greater lateral CoP displacement. Findings suggest that the dominant limb participates more into ML balance control or the vulnerable non-dominant limb tends to avoid excessive load (Sadeghi et al., 2000). Despite asymmetrical gait deterioration previously reported for older adults' non-dominant lower limbs (Erja, 2008; Perry et al., 2007), this asymmetrical feature in ML balance control at MLM was observed regardless of the age groups. Evidence for age-related asymmetrical decline in ML balance was, however, characterised in higher step-to-step variability in MLM_d and lateral CoP displacement in the older group's non-dominant limb during controlled width walking conditions.

As shown in earlier work (Ko et al., 2007; Wezenberg et al., 2011), more variable MLM_d and lateral CoP displacement in the older adults' non-dominant limb during narrow and wide walking is considered disadvantageous in maintaining ML stability. Increased CoP variability in controlled-step conditions was due to an ankle strategy to compensate minor ML balance fluctuations (Vanderpool et al., 2008; Wezenberg et al., 2011). During preferred width walking, higher step width variability was observed, that appears to reflect minor balance adjustments by changing the Safety Zone characteristics (Mille et al., 2005). While adaptations to both the CoM movement and the Safety Zone characteristics are available in controlling balance during unconstrained walking, with step width restrictions, fluctuating balance must be restored primarily by controlling the CoM movement because changes to the ML Safety Zone were limited to the experimental requirements.

Centre of pressure (CoP) control is a primary contribution to the ML CoM movement, and as discussed above, more active non-dominant ankle movement was possibly reflected in increased CoP and MLM_d variability (Bus et al., 2009; Hans et al., 1999; Rietdyk et al., 1999). Given this asymmetrical feature, ML balance in older adults may be more difficult for the non-dominant stance limb, as also suggested in other previous kinetic and kinematic data highlighting the non-dominant limb's weakness in older adults (Perry et al., 2007; Nagano et al., 2011; 2012; Yogev et al., 2007). As increased variability in CoP and MLM_d was seen both in narrow and wide walking conditions, age-related impairment in ML balance was observed equally with decreased and increased step widths.

As further discussed in Section 11.5.2 below, it is interesting to note that despite increased variability in MLM_d and ML CoP displacement in the older adults' non-dominant limb, both step width and the ML CoM position were more consistent when walking with controlled step widths, probably due to the lines assisting foot targeting. There are, accordingly, positive effects on ML balance when walking with controlled step width. Again from the perspective of energetics (Sparrow, 2000), more consistent step width and ML CoM control during width controlled walking can be considered less energy demanding due to reducing energy expended in ML balance control. Increased variability in MLM_d and ML CoP control seen in older adults' non-dominant limb could therefore reflect the expenditure of surplus energy, generated from minimised ML CoM and foot movement variability (Donelan et al., 2004). Excessive mechanical energy will not necessarily improve gait safety as it is difficult to predict how excessive energy is utilised and affects gait. Strategies to reduce variability in MLM_d and CoP displacement could be achieved by dispersing impact forces at heel contact, using footwear, shoe-insole, ankle or knee braces designs. To absorb surplus energy from the non-dominant limb's CoP and MLM_d variability, incorporation of a spring in, for example, an ankle brace or a shoe could effectively absorb impact as elastic energy, possibly reducing variability (Barton, 1994, Kim et al., 2013).

11.3 Safety Zone Analysis

Step width effects at critical events revealed how balance is maintained in young and older adults. These events were also incorporated into the Safety Zone Model to overcome limitations of traditional balance assessments using only the CoM and the BoS during double support (Figure 2.6.1). Safety Zone analysis revealed three fundamental characteristics of swing phase control in older adults, which will be discussed in separte sections below: Non-Dominant Toeing-In, ML Safety Zone control, AP Safety Zone control.

11.3.1 Non-Dominant Toeing-In

When walking with narrow or wide steps, older adults showed non-dominant toeing-in, also seen in a previous study (Menz et al., 2005). Toeing-in is adaptive in increasing the anterior Safety Zone margin at toe-off and heel contact (Figure 8.2.1 and Figure 8.2.4) when the non-dominant foot is leading. When however, the trail foot was non-dominmant, the Safety Zone shrank. Older adults' balance can, therefore, be considered more vulnerable when the non-dominant limb is trailing. At minimum foot clearance (MFC) and minimum lateral margin (MLM), non-dominant stance foot toeing-in tended to increase the possibility of balance loss in the lateral direction as CoM movement was directed toward the lateral toe-heel boundary of the non-dominant stance foot. It has already been well documented that age-related functional impairment is more prominent in the non-dominant limb (Perry et al., 2007) and gait asymmetry is due to older adults using their non-dominant limb to secure gait and the dominant limb to serve progression (Sadeghi et al., 2001; Nagano et al., 2011; 2012). The non-dominant limb toeing-in phenonenon further supports these findings because in the present study it is shown to be a safety adaptation associated with assigning a gait securing task to the non-dominant limb. If this is the case, older adults may perceive a higher risk of stability loss when the non-dominant limb is leading and as a consequence toe-in to increase the Safety Zone anterior margin and engender a safer gait pattern. When, however, the non-dominant limb is trailing, toeing-in reduces the Safety Zone and, interestingly, *increases* balance loss risk despite increased confidence associated with the non-dominant limb serving as the trail foot. From the Safety Zone analysis it is concluded that the highest risk phases of the gait cycle for older adults are from late swing to heel contact during the non-dominant stance phase.

11.3.2 Medio-Lateral Safety Zone Control

As described in the previous chapter 5 about ‘Research Questions and Hypotheses’, older adults’ gait characteristics reflect safety adaptations and functional impairments. While ‘toeing-in’ appears to be a safety adaptation, age-associated functional decline in gait parameters is generally reflected in greater variability (Hausdorff, 2001; 2005). While step width variability was not differentiated between the age groups, ML distance between the CoM and both feet was more variable in older individuals. Older adults, therefore, showed less consistent ML CoM control than the young participants but with *similarly stable ML Safety Zone control*. Even when foot control is well maintained, therefore, older adults show less consistent ML CoM control, detected as the evidence for age-related loss of ML stability.

11.3.3 Anterior-Posterior Safety Zone Control

Reduced step length and associated walking speed with ageing were due to earlier and more posterior mid-swing events. Older adults revealed a more posterior swing foot location compared to young adults at MFC and MLM, with the timing of the swing phase events, MFC_t and MLM_t, earlier in the older group. Older adults accordingly seem to hasten these events toward relatively posterior locations. As a consequence, the anterior Safety Zone margin at MFC and MLM was considerably reduced in older adults. Lugade et al. (2011) also pointed out that ageing and fear of falling were associated with more conservative AP separation of the CoM from both feet at toe-off and heel contact. Following Lugade et al. (2011), conservative AP separation of CoM reflects lack of balance control.

11.3.4 Future Research

In future research into the Safety Zone Model the centroid description, the centre coordinate of the Safety Zone determined by the average of the coordinates of both toes and heels, could be investigated as method for balance description (Figure 11.3.1).

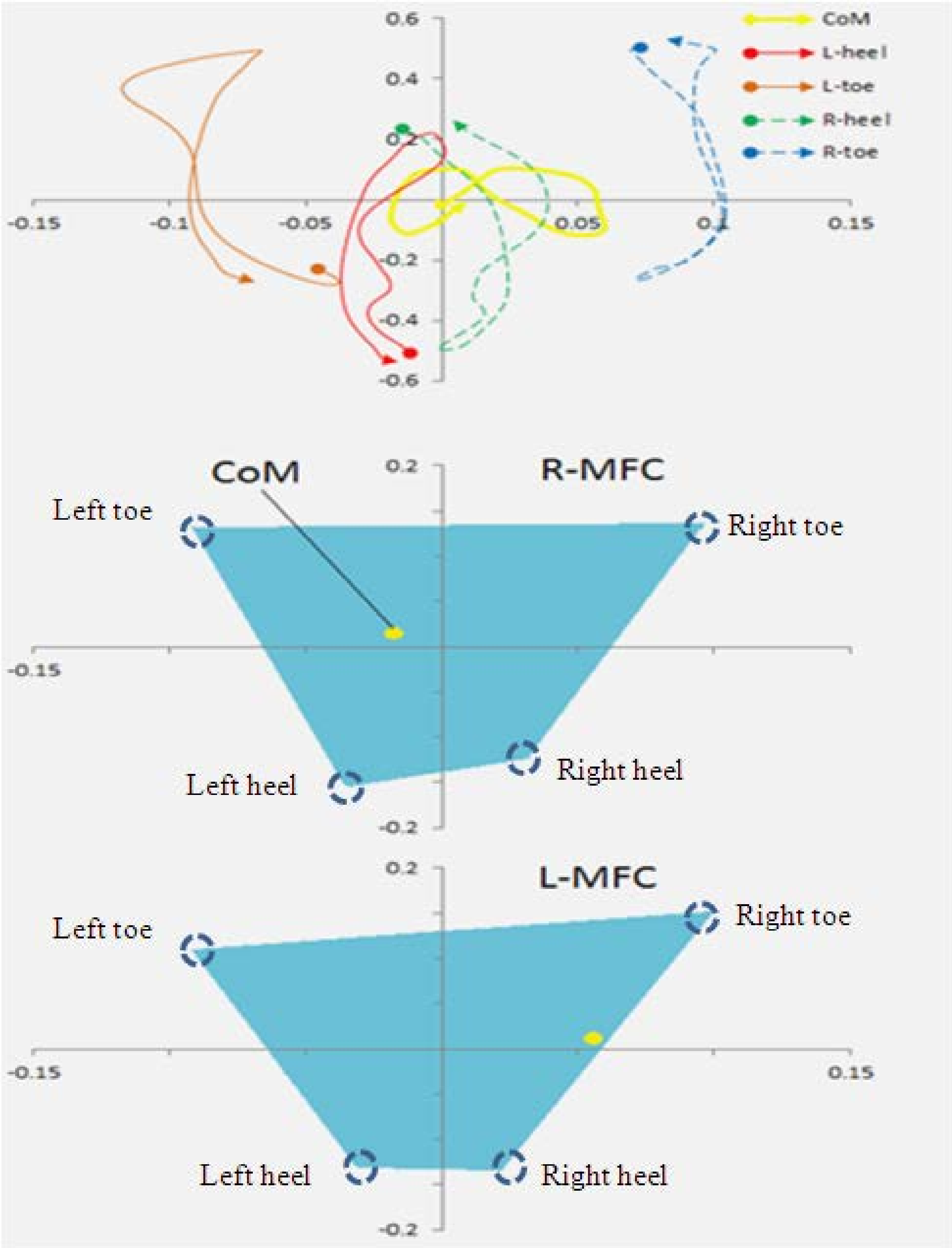

Figure 11.3.1 Centroid description of dynamic balance using safety zone. (top) Right gait cycle, right heel contact to right heel contact. X & Y axes: ML and AP distance from the centroid. (middle) Right MFC and (bottom) Left MFC Data taken from one walkthrough trial of single young subject.

While dynamic balance description using the CoM as the origin of coordinate (0, 0) is particularly useful in balance loss simulation for ART computation, the centroid description can also visualise dynamic balance, effective in observing balance relative to the entire Safety Zone area and possibly providing new balance description methods. Top panel of Figure 11.3.1 describes the CoM is deviated to the right in general relative to the centroid during the right foot gait cycle. As shown in both middle and bottom panels of Figure 11.3.1, some asymmetrical features of dynamic balance can be better illustrated by the centroid description. These two descriptions illustrate the Safety Zone shapes are relatively symmetrical between the right and left swing foot MFC, but the location of the CoM is closer to the centroid during right MFC whereas it is closer to the lateral boundary of the Safety Zone, stance foot, during left MFC. Despite advantages of centering the CoM to characterise margins of the Safety Zone, the centroid description can be expected as another useful presentation of the Safety Zone analysis.

11.4 Balance Loss Simulation

A further application of the Safety Zone Model devised in the current research is balance loss simulation, in which balance disturbances are modelled as kinetic inputs to predict their effects on the CoM displacement within the Safety Zone. Using the IP model to estimate the CoM acceleration due to foot perturbation, various balance disturbances can be simulated. One advantage of balance simulation using the current Safety Zone Model is its applicability to any event within the gait cycle. We recall from Chapter 3, that when a net force perturbs the foot the CoM is displaced in the opposite direction. The CoM acceleration at that time can be calculated by dividing this force by pelvis segment mass.

Available response time (ART) has conventionally been calculated as the time required for the CoM to cross the anterior Safety Zone boundary. Lugade et al. (2011), for example, computed ART at heel contact based on distance to balance loss divided by walking velocity. At heel contact anterior balance loss is, however, unlikely and such models emphasising only the anterior Safety Zone boundary (Lugade et al., 2011) do not account for posterior or lateral balance disturbances. By using foot-ground kinetics foot acceleration and associated CoM acceleration due to slipping can be modelled (Figure 3.4.1). dominant heel contact on a walking surface of CoF = 0.10 by the older group was predicted to cause *posterior* balance loss in 0.13s. Without a simulation of slipping, however, the CoM was predicted to cross the anterior boundary in 0.23s, which is unlike a gait disturbance at heel contact (Smeesters et al., 2001). By modelling a variety of balance disturbances as kinetic inputs to the CoM or the foot, the ART computation can be applied to any gait cycle event.

11.4.1 Simulation of Anterior and Posterior Slipping

Foot slipping was considered to accelerate the pelvis in a direction opposite to the slipping foot movement. Simulation of both anterior and posterior balance loss due to slipping commonly identified the horizontal centre of mass (CoM) velocity as critical in determining ART. As reflected in greater step velocity in the young group, the horizontal CoM velocity at toe-off and heel contact was also higher. For anterior heel slipping, the faster anterior CoM velocity was advantageous in providing longer time to reach the posterior boundary of the Safety Zone to pass the CoM and initiate posterior balance loss (You et al., 2001). For posterior toe slipping, in contrast, the anterior boundary moved more posterior and combined with the increased horizontal CoM, ART to anterior balance loss was reduced.

The balance loss simulation following anterior heel slipping was effective in assessing slipping risk. While gait experiments incorporating slippery surfaces using oil, soap or water as contaminant are essential in characterising slipping (Cham and Redfern, 2002; Fong et al., 2009; Li et al., 2004; Lockhart et al., 2003), the problem of anticipatory responses toward a potentially slippery surface has been acknowledged as a factor influencing the results (Heiden et al., 2006). For this reason, simulation using non-slipping walking trials could be helpful in evaluating dynamic balance control in response to a *potential* slip.

Table 9.2.3 presents that older adults had lower friction and the slower CoM velocity compared to young adults at heel contact consistent with Lockhart et al. (2007), disadvantageous in maintaining the CoM velocity sufficiently fast enough to prolong available response time (ART) to posterior balance loss as suggested in the previous study (You et al., 2001). Lower heel contact velocity and acceleration on both heel and the CoM were, in contrast, also identified in the older group, useful in prolonging ART. Overall similar ART between the age groups was found for simulated dominant heel anterior slipping but the older group revealed shorter ART of 0.20s compared to the young of 0.41s for non-dominant anterior heel slipping. Older adults are considered to require longer ART due to age-related declines in reaction speed and muscular strength necessary for decelerating the slipping foot, accelerating the CoM and maintaining the vertical height of the CoM (Ratcliff et al., 2001; Der and Deary, 2006; Haber et al., 2008; Liu and Lockhart, 2009).

For older adults, 90% of hip fractures are attributed to posterior falling (Cummings et al., 1985; Hayes et al. 1993) and one in every five cases eventually result in death (Deprey, 2009; Leibson et al., 2002). Safety Zone analysis for ART due to anterior slipping is, therefore, a new approach to further understand the mechanism of posterior balance loss, which is capable of assessing dynamic balance at heel contact.

Previous studies have not been dedicated into posterior forefoot slipping at the negative RCoF peak prior to toe-off due to the commonly granted speculation that balance can be maintained even if a toe slips in the posterior direction (Strandberg and Lanshammar, 1981; Leclercq, 1999). ART computation here, however, supports the view of Myung (2003) that forefoot posterior slipping deserves caution. Successful compensatory reactions to gait instability must take place within response times lower than the predicted ART (Der and Deary, 2006). From this perspective it is important to note that ART for anterior balance loss as a consequence of posterior toe slipping was considerably less than anterior heel slipping (0.03s-0.05s). The Safety Zone Model presented here, however, does not include any lead stance limb contribution to posterior force generation that may assist in increasing ART. Although it is accordingly not conclusive to decide whether anterior or posterior slipping can be more hazardous among older adults, the Safety Zone simulation revealed potential dangers of posterior forefoot slipping during the push-off phase prior to toe-off especially among vulnerable older adults. If turning motions are accompanied, forefoot posterior slipping will increase the balance loss hazard (Nagano et al., 2013).

In terms of the direction of walking, the CoM and foot slipping movements were not completely parallel, eliciting slight ML velocity. For the accurate description, angles of push-off or braking forces were incorporated into ART calculation. Yet, less than 0.01s difference was accounted by including ML component. A simplified method for ART computation can be, therefore, based on the assumption that the CoM and slipped foot travel straight in the AP direction without negligible ML movement. Simulation of slipping during turning motion (Nagano et al., 2013) will, however, require inclusion of ML movement of the CoM and slipping foot.

Influence of the slipping foot on the CoM movement based on the IP model was insightful in which net force on the slipping foot can be simply divided by the pelvis mass to estimate acceleration on the CoM, according to the torque convention (force couple) and its linear component. The Safety Zone simulation method devised here could be applied to characterising effects of ageing, pathologies, gait tasks and anti-slipping devices on AP balance loss risks due to slipping.

11.4.2 Lateral Balance Loss Simulation

In contrast to tripping or slipping, lateral balance loss simulation was not predicted on external forces. Simulation was, accordingly, based on the minimum force necessary to cause lateral balance loss as observed in CoM dislocation lateral to the stance toe-heel boundary at minimum lateral margin (MLM). The results indicated that young adults were resistant against ML balance perturbation because within 0.02s after MLM, the CoM would pass the AP stance toe location and therefore, the CoM would not cross the Safety Zone lateral boundary. The lateral force required to cause lateral balance loss at MLM within such a short time frame is equivalent to impact by physical collision of an individual with body mass of the young group (75.2kg) moving at 48.3m/s and 32.5m/s at dominant and non-dominant MLM, respectively. From these assumptions, it can be concluded that lateral balance loss is unlikely at MLM for young adults due to the lateral force without any posterior component. Applying the same description for the older group, collision of a person with body mass of 76.7kg, the older group's average, should be greater than 2.15m/s and 1.36m/s to induce lateral balance loss at dominant and non-dominant MLM, respectively. This can be a more realistic hazard and collision with a heavier object or lateral perturbation that also contains posterior force could further increase lateral balance loss at MLM for older adults. This is primarily attributed to the prolonged time limit for the CoM to pass the AP stance toe location for 0.08s and 0.11s for dominant and non-dominant MLM, respectively.

As Hilliard et al. (2008) stated, ML balance controlling ability is predictive of future falls among older adults, and MLM was confirmed as the event associated with lateral balance perturbation for older adults. Young adults were, in contrast, unlikely to experience the CoM dislocation lateral to the Safety Zone. If MLM is not associated with lateral balance loss for young adults, other events such as heel anterior slipping can be, for example, one of the causes of lateral falls (Smeesters et al., 2001). Particularly when turning motions are involved while slipping (Nagano et al., 2013; Taylor et al., 2005) or recovery is attempted involving a complex and case-specific action (Lockhart et al., 2007), balance loss can take place in the lateral direction.

11.4.3 Tripping Simulation

Simulation of ART for tripping uses swing toe acceleration and foot segment mass to estimate reaction force working on a tripped foot. The current simulation is for a fixed object, but it is also possible to estimate ART for tripping on an ‘unfixed’ object by considering the law of conservation of momentum. It was first considered that torque created by tripping would cause angular acceleration on the CoM, but when ART was computed based on the CoM angular movement, ART was not considered valid (1.65s-5.16s). For ART calculation while tripping, therefore, effects of torque should be limited to the linear motion probably because the IP model is only for stance limb but not applicable to swing limb movement. Same as slipping simulation, reaction forces on a tripped foot and the CoM movement from tripping can be optimised only to the AP axis affecting less than 0.01s. For example, at dominant MFC of the older group, the CoM was travelling 0.9º toward the non-dominant limb. Optimisation of the CoM travelling only anterior underestimates the distance to the anterior balance loss by 0.6mm. The error of ART computation by this optimisation was less than 0.001s, considered negligible in assessing reaction for balance recovery.

Based on the obtained results, young adults showed slightly lower ART by 0.04s-0.05s, but the current ART computation method for tripping does not account for any stance limb pushing off task to provide extra ART (Neptune et al., 2001; Pijnappels et al., 2005; 2008; Winter, 1980). Older adults were characterised for weaker plantarflexors and knee extensors (Perry et al., 2007; Winter, 1991), therefore considered to have less ability to push off the walking surface in the case of tripping and provide longer period for recovering from anterior balance loss. To further develop the Safety Zone Model to predict tripping-related anterior balance loss, effects of stance limb kinetics on balance recovery should be investigated. Stance limb can possibly decelerate the CoM in addition to the previously reported CoM lifting effects (Neptune et al., 2001; Pijnappels et al., 2005; 2008; Winter, 1980), vertical CoM acceleration and associated velocity.

Another method to estimate reaction forces by a simulated trip depends on the momentum-impulse principle. Despite the difficulty in reasonably simulating *time* of reaction forces working on a tripped foot, this approach does not have to use variable acceleration data. The current model relies on highly variable swing toe acceleration data, and when acceleration is negative, force calculation also indicates negative, considered not representing a collision of a trip. In future studies, however, it will be interesting if a cutoff frequency of low-pass filtering is found to successfully remove noise and visualise a consistent pattern in acceleration data (e.g. Lockhart et al., 2007), which could possibly further improve the accuracy of ART computation.

Available response time (ART) computational method for tripping has the capacity to examine *severity* of tripping. While the majority of previous research on minimum foot clearance (MFC) and tripping focused only on *frequency* of tripping (Begg et al., 2007; Winter, 1991), no previous studies have estimated impact forces due to tripping and their effects on balance. Use of the Safety Zone Model on tripping biomechanics is, therefore, expected to develop further understanding of practical tripping-related events and falls.

11.4.4 Future Research into Balance Loss Simulation

A limitation to the ART computation employed here is due to using the inverted pendulum (IP) model to characterise single limb support (Kuo, 2007) because the stance limb is not a rigid segment and knee joint motion is excluded. Furthermore, ART only indicated the time limit for the centre of mass (CoM) to be dislocated from the transverse Safety Zone when no recovery attempt was made. As discussed above, in the event of balance loss the stance limb's push-off action can accelerate the CoM vertically to provide extra time (Neptune et al., 2001; Pijnappels et al., 2005; 2008; Winter, 1980). If vertical displacement of the CoM can be maintained, balance loss will not cause a fall. It is, therefore, possible to characterise balance loss leading to a fall as having two components: first, CoM dislocation beyond the Safety Zone and, second, vertical CoM height lost due to 'body collapsing'. For example, even if the CoM is maintained within the Safety Zone, a fall may result due to the CoM reaching the critically low height.

11.5 Falls Prevention and Gait Rehabilitation

The ultimate objectives of gait research in relation to falls among older adults are to provide insights in establishing effective intervention strategies for falls prevention and rehabilitation programs due to injuries. Using the results of the current study three approaches were evaluated: 1) possibility of the tested shoe-insole for reduction in falls injuries during locomotion, 2) treadmill walking with guided parallel lines for balance training, and 3) effective clinical decision making utilising the optimised Safety Zone Model.

Outdoor walking provides exercise and social interaction, therefore it is recommended for older individuals to maintain their healthy lifestyles (WHO, 2013). As emphasised throughout the Thesis, however, the falls risk must be minimised and any suboptimal gait features should be modified to prevent joint arthritis as a consequence of long-term engagement in outdoor walking. This is possible by enhancing mechanical energy efficiency during the loading response (Cavagna et al., 1976; Collett et al., 2007; Vereecke et al., 2006; Ortega and Farley, 2003; Detremblur et al., 2005; Schepens et al., 2004). Effective footwear interventions for safer walking encourage voluntary engagement in outdoor walking. Shoe-insole modification is relatively cheaper in manufacturing costs and applicable to various types of shoes. The obtained results for the tested shoe-insole are evaluated in this section for its possibility to be incorporated into footwear intervention.

Establishment of effective gait training is important for falls prevention and also for rehabilitation from falls-related injuries. One goal to achieve safer gait is to reduce step-to-step variability and gait asymmetry (Hausdorff, 2005). The line walking conditions reduced gait cycle parameter variability and ML CoM variability within the Safety Zone. Treadmill walking provides highly predictable sensory input and accordingly, a steady state of gait pattern is easier to attain. Line walking on the treadmill therefore deserves attention in future studies as a gait training technique.

The Safety Zone Model can also be applied in clinical settings for dynamic balance assessment. Marker placement is necessary only at toes and heels in addition to modelling of the pelvis segment by a minimum of three markers. Safety Zone analysis can be utilised to characterise potential hazards, asymmetry and deficiency in balance control. Prescription of adequate gait training or identification of caution-demanding balance disturbances specific to individuals' gait patterns can be inspected by the Safety Zone assessment. .

11.5.1 Shoe Insole Interventions

The current study examined the insole effects on ankle dorsiflexion and eversion support. First, 2.2° dorsiflexion support was provided by the insole during static standing (Figure 4.4.1), but at MFC, 0.7° ankle dorsiflexion increase was confirmed. According to foot clearance sensitivity (Moosabhoy and Gard, 2006), 0.7° ankle dorsiflexion would elevate MFC_h by 0.21cm. In addition to dorsiflexion, knee flexion is another joint motion that can increase MFC_h, and ankle dorsiflexion is known to trigger knee flexion (Fong et al., 2011).

Insole increased knee flexion by 1.3° at MFC was achieved, which is considered to add another 0.06cm to MFC_h. The estimated increase in MFC_h by added dorsiflexion and knee flexion increase due to wearing the insole was therefore 0.27cm (Moosabhoy and Gard, 2006). The experimental results of the current research identified elevation of MFC_h of 0.36cm during insole walking. Other joint motions such as hip flexion or lower limb abduction at MFC are also considered to control MFC_h, and such joint motions might be influenced by the insole.

Higher MFC_h is necessary to prevent swing foot from unexpected contact of the swing foot with an obstacle, and as expected, the insole's dorsiflexion support was confirmed in increased MFC_h. While in this sense, the insole can reduce tripping risk, wider MFC_h distribution, reflected in higher IQR and lower kurtosis (Figure 10.2.2), was also found. Without overcoming these negative features in terms of consistent MFC control, therefore, the tested insole cannot be concluded as successfully reducing tripping risk (Begg et al., 2007). In addition to increased familiarisation to the insole, stronger attachment between the foot and the insole could also reduce MFC_h variability.

Despite wider distribution of MFC_h dataset in insole walking, variability in both ankle and knee angles at MFC did not increase while wearing the insole, implying that consistent lower limb joint control was not impaired by the insole. From qualitative feedback from the participants, some reported a foot separated from the insole due to the inclined insole surface. This might be a cause of MTP movement less synchronised with the foot segment motion. Addition of medial arch support or a semi-custom moulded surface based on heating and freezing moulding technology (Crabtree et al., 2009; section 4.3.2) can provide greater contact area and stronger attachment between the foot and the insole, while increased contact area can be also useful in pressure distribution (Tsung et al., 2004).

Despite higher MFC_h variability, insole effects on ML balance and improved shock absorption are the positive outcomes of the tested insole. Eversion support increased MLM_d and reduced lateral CoP displacement, meaning improved ML balance and effective in inversion sprain prevention. Incorporating eversion support into a shoe insole to relieve knee osteoarthritis has been already widely recognised as 'lateral wedge insole' that has been reported to reduce knee adduction moment (Rafiaee and Karimi, 2012; Kerrigan et al., 2002; Nakajima et al., 2009; Toda and Tsukimura, 2004) despite other research reporting no significant effects (Bennell et al., 2011; Reilly et al., 2006). The current research findings suggest that ankle eversion combined with dorsiflexion is effective in reducing knee adduction moment and promoting energy efficient loading.

Ankle dorsiflexion and eversion promoted mechanically energy efficient loading (section 4.3.1) as shown in a 2% increase in mechanical energy recovery. More energy efficient gait can utilise impact at heel contact to activate toe-off (Han et al., 1999; Silver-Thorn et al., 2011). The shoe-insole reduced impact forces at heel contact and knee adduction moment decreased by more than 20% compared to the no-insole control condition. In summary, knee flexion and ankle dorsiflexion at heel contact and prolonged time from heel contact to foot flat were attained by the insole, accounting for energy efficient loading and reduced impact on knees.

The tested insole has demonstrated a variety of positive effects on gait despite higher variability in MFC_h. In collaboration with the footwear industries, shoe-insole intervention can provide a more practical intervention technique for older adults in addition to cost effectiveness and accessibility compared to exercise intervention at rehabilitation facilities. Addition of medial-arch support and use of semi-moulding technology for inner material of shoe insole can further improve pressure distribution and possibly reduce MFC_h variability.

11.5.2 Gait Training Using Line Walking

Step width control in the current experimental condition was designed to influence balance control. Walking on two lines was revealed to decrease variability of various gait parameters including step width and the ML CoM movement relative to both feet. Furthermore, high symmetry in spatio-temporal parameters was generally attained while walking on two lines. Narrow walking and also wide width walking to a lesser extent, however, also disturbed gait patterns of the older group such that typical caution related adaptations on spatio-temporal parameters were found such as slower gait velocity due to shorter step length (e.g. Bock and Beurskeus 2010; Dunlap et al., 2012; Hollman et al., 2007; 2011; Ko et al., 2007; Nagano et al., 2012; Nordin et al., 2010; Schrager et al., 2008). More variable double support, MLM_d and lateral CoP displacement were also evident during width controlled (line) walking.

If line walking is conducted with optimum width, benefits of line walking without having adverse gait adaptations could possibly be expected. One particular concern is the possibility of the toeing-in effect of the non-dominant foot when older adults walked in width controlled conditions. If toeing-in is a result of walking with unfamiliar widths, line walking with optimum step width may not accompany such effect. It could be, however, also possible that a foot targeting was more demanding for older adults (Berg and Murdock, 2011) and non-dominant toeing-in could be a strategy to secure balance.

To maximise the learning effects of steady gait control, treadmill walking can be recommended (Herman et al., 2007; Lay et al., 2002; Schmidt, 1991) despite difficulty in familiarising to motor-driven walking environment for older adults (Wass et al., 2005). Positive results of treadmill training for older adults' gait were already confirmed on various spatio-temporal parameters (Oh-Park et al., 2011). Step length, width and time are expected to show more consistency once familiarisation is completed and gait achieves steady state condition (Hockey et al., 2005). It will be, therefore, interesting to conduct preferred width line walking on the treadmill for the purpose of gait training. A potential issue can be altered vision by looking down on the two lines to control precise foot placement.

While Parkinson's patients improved balance during line walking on the treadmill by looking down over the treadmill belt surface (Almeida and Bhatt, 2012), Buckley et al. (2011) reported that healthy young individuals did not rely on real-time visual feedback on the surface environment for foot placement. Instead, feedforward visual information on the surface condition is processed previously to determine later foot placement. To provide normal gait vision, attention should be maintained in the walking direction. In treadmill line walking the problem of vision being directed downward can be overcome using a projection screen in front of the patient to show real-time foot movement relative to the guidance lines.

Same as general precautious of gait testing involving older adults, highly protective walking environment should be provided by treadmill for the purpose of gait training of older adults. Further research may deserve attention on inventing treadmills for self-gait training among the older population. For advanced users, random induced balance disturbance can be imposed by, for example, suddenly accelerating the anterior belt movement to induce anterior slipping, which has been reported as an effective training technique for anterior heel slipping (Pai and Bhatt, 2007). Effectiveness of real-time feedback of MFC_h during treadmill training to improve foot clearance has been already reported in both healthy young and older adults (Tirosh et al, 2012; Begg et al., 2012). With further advancement, real-time feedback on balance based on the Safety Zone could also be incorporated for treadmill-based gait and balance training.

11.5.3 Gait Diagnostics and Clinical Decision Making

Safety Zone analysis employed in the current study can be also used in clinical gait assessment. The advantage of the Safety Zone analysis is its visual presentation by plotting as x-y graph. First, its shape relative to the CoM indicates in which boundary the CoM tends to cross over or what type of disturbance a particular walking pattern is more vulnerable to. For example, dominant heel contact of older adults in narrow walking showed very small posterior margin, suggesting increased danger of posterior balance loss in the case of anterior heel slipping. It could be then suggested that anti-slippery material applied especially on the heel part of the dominant foot shoe could be useful. For gait with small lateral margin, inclination of insole could be recommended to improve ML balance of gait.

Visual presentation is useful in identifying asymmetry in balance control ability. Once asymmetry was found, further asymmetry examination on spatio-temporal parameters, kinetics at RCoF peaks, MFC and MLM characteristics can be conducted. Examination of joint kinematics and kinetics may assist in identifying which muscles should be strengthened to attain more symmetrical balance control. Use of electromyography (EMG) will also be effective in confirming where exactly kinetic asymmetry arises from (i.e., which muscles contribute). Symmetrical balance control can be set as a goal of gait training and constructive feedback on training methods is expected to be provided.

Variability analysis on the Safety Zone is essential in identifying impaired balance control. If the ML CoM control is diagnosed as highly variable relative to the non-dominant limb, balance training should especially aim to improve the non-dominant side of the Safety Zone. In case degeneration of balance is considered difficult to fully regain due to traumatic injuries, other gait adaptations can be devised for those special populations. Modification of gait to secure altered gait due to dysfunction should be designed to provide sufficient margins to the Safety Zone boundaries. Alteration of pathologic gait can be, therefore, experimented by the Safety Zone examination.

Similar to X-ray for suspicion of fractures, the Safety Zone assessment can be possibly placed as the first diagnostic tool in clinical gait assessment. Optimised marker setups are individual markers on toes and heels of both feet and modelling of the pelvis segment. Foot markers can be mounted permanently on a shoe to minimise time necessary for marker setup. Pelvis modelling can be efficient if a set of three markers is installed on a belt using the virtual marker function in the Optotrak system.

11.6 Conclusion

Gait can be described as a task requiring the centre of mass (CoM) to be transported in the desired direction while remaining within the Safety Zone. The ability to maintain the CoM within the Safety Zone requires continuous foot placement, adaptations reflected in spatio-temporal gait parameters. The current research employed step width manipulations to examine how spatio-temporal gait parameters related to medio-lateral (ML) balance influence the Safety Zone. Ageing effects cause different gait adaptations in response to increased and decreased step width.

Young adults did not show notable differences in overall gait parameters except distance between the two ML boundaries as a result of controlled step width and associated minimum lateral margin (MLM) and lateral centre of pressure (CoP) characteristics. Narrow walking for older adults was identified as the cause of reduced walking speed accompanied by shorter step length and longer double support time. These gait adaptations in spatio-temporal parameters successfully maintained the CoM within the Safety Zone during narrow walking with a shorter lateral boundary. Slowing is likely to be helpful for the precise CoM control but associated reduction in step length reduces AP margin. ‘Toeing-in’ detected as the notable asymmetrical feature specific to the non-dominant foot was found to lengthen the AP Safety Zone when the lead-foot was non-dominant. This adaptation is, however, at the cost of the reduced Safety Zone when the toeing-in non-dominant foot is trailing. The later dominant swing phase particularly from MLM to heel contact was especially identified as the period of the Safety Zone vulnerable to balance disturbance for older adults during narrow walking.

Wide walking accompanied similar but less effects on spatio-temporal parameters for older adults. Reduction in step length leads to the smaller AP Safety Zone, compensated by non-dominant toeing-in to extend the anterior boundary during the non-dominant swing phase. Non-dominant toeing-in was, again, confirmed as the negative balance factor during the dominant swing phase. The functional advantage of enlarging step width especially for ML balance has been biomechanically described as the greater ML distance of the Safety Zone with greater MLM_d and shorter ML CoP displacement. Both width-controlled walking conditions identified increased variability of MLM_d and ML CoP distance in the non-dominant foot, indicating the less consistent ML balance. Width-control made energy dissipation more difficult for the non-dominant stance foot, reported as another evidence of more impairment in the non-dominant limb for the older group.

Width control led to lower RCoF and greater MFC_h in general, positive adaptations for slipping and tripping. While reduced step velocity was considered to contribute for lower RCoF, mechanical energy dissipation and safety adaptation account for increased MFC_h in width-controlled walking. From this point, gait training can possibly incorporate line walking with preferred step width to maximise ML stability although possible increase in variability of the older adults' non-dominant foot needs to be attended.

To identify gait features that may cause falls, balance definition of the CoM within the Safety Zone and effects of each step to change the CoM relative to the Safety Zone need to be understood. Examined swing events were associated with balance perturbation that can possibly cause the CoM to be dislocated from the Safety Zone. Simulation methods were, therefore, devised by describing balance perturbation as kinetic inputs into foot and associated acceleration into pelvis segment. Except for lateral balance perturbation at MLM, inverted pendulum model was used to convert net force on the foot into kinetic inputs acting on the CoM. By dividing the net force on the perturbed foot by pelvis segment mass, acceleration of the CoM can be estimated. Estimated equivalent movement of the CoM and its effect on the Safety Zone boundary was used to calculate ART.

By describing human gait as successful transport of the CoM within the Safety Zone controlled by each step, ageing effects on gait to understand falls risks have been thus advanced throughout the Thesis. While separating gait into different phases and events were useful in characterising specific balance perturbations, it is also important to view gait and balance as the inseparable concept that interacts with each other to attain safe walking, with which ageing effects on increased falls incidence can be investigated in more details.

Another purpose of the Thesis was to test the possibility of incorporating dorsiflexion and eversion support into the shoe-insole to prevent tripping, improve ML balance, and avoid injuries such as inversion sprain and lower limb joints' osteoarthritis. Although the insole improved ML balance and reduced injury risks, elevated but more variable MFC_h was not conclusive to reduce tripping risk. In addition to more familiarisation in walking with the insole, increased contact area between the foot and the insole could minimise MFC_h variability, possibly achieved by taking advantage of, for example, semi-custom moulding technology and medial-arch support. The insole intervention for falls prevention seems quite promising in terms of its effectiveness in improving lower limb biomechanics as well as its cost effectiveness and simplicity. Further research should be undertaken to investigate the potential positive impact of such intervention for falls prevention and maintaining healthy lifestyles for older adults.

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Appendix I: Advertisement for older adults' participants into the research.

http://www.fiftyplusnews.com/home/.maitos/fiftyplus/fiftyplusnews.com/wp-content/uploads/2010/10/Fifty_Plus_Oct10_issue_web.pdf

Appendix II: Ethics

VICTORIA UNIVERSITY
HUMAN RESEARCH ETHICS COMMITTEE

Application for Approval of Project Involving Human Participants in Victoria University

REGISTER NUMBER (*office use only*) : HRETH ________

INFORMATION FOR APPLICANTS

1. Applicants are advised to follow the *Guidelines for Applications* prior to submitting *Application for Approval*. Applicants are to forward a) Twelve (12) hard copy applications (including one original copy)* with any accompanying documentation to your Faculty Ethics Officer and b) an electronic application to your Faculty Ethics Officer. Note: *Non Minimum Risk applications* may be forwarded directly to the Secretary, Victoria University Human Research Ethics Committee (researchethics@vu.edu.au).
2. A *Consent Form for Participants Involved in Research* template and *Information for Participants* template is also available on-line.
3. The above documents are located at: http://research.vu.edu.au/hrec.php

* *Applications to be considered at the Faculty of Business & Law need to submit one original hard copy application.*

YOU ARE REMINDED THAT THIS PROJECT MUST NOT COMMENCE WITHOUT PRIOR WRITTEN APPROVAL

FROM THE APPROPRIATE HUMAN RESEARCH ETHICS COMMITTEE.

Please Note:

- Ethics approval will not be finalised until electronic & hard copy applications and copies of all necessary materials have been received by the Secretary of the relevant Human Research Ethics Committee.
- This application form is included in the Human Research Register. If your project includes information of a commercial or patentable nature, this information should be sent separately and marked as confidential.
- If an institution other than Victoria University is to be involved in the project, please provide this information and evidence of ethics approval from the other institution with this application.
- Research projects undertaken by individuals who are not staff members of VU that require access to a cohort of VU staff or students for research must be 'sponsored' by a member of VU staff who will take responsibility for all interactions with the University and the HREC. A copy of the approved project and approval letter must be forwarded to the Ethics & Biosafety Administration Group.
- If sufficient space is not available on the form for your answer/s, please attach additional page/s.
- Ensure all questions are appropriately answered and the hardcopy application is authorised by appropriate staff (Applications will not be processed without the appropriate authorisation).
- To avoid unnecessary delays, please ensure your full application (both hard copies and soft copy application) has been received by the relevant Human Research Ethics Committee submission date. Refer to University/Faculty *Committee Meeting Dates* at http://research.vu.edu.au/hrec.php

University & Faculty Forwarding Details:

Victoria University Human Research Ethics	Faculty Human Research Ethics Contacts
Send electronic applications to: E-mail address: researchethics@vu.edu.au	Send electronic applications to: Faculty of Arts Education & Human Development: AEHDEthics@vu.edu.au Faculty of Business & Law: BLEthics@vu.edu.au Faculty of Health Engineering & Science: HESEthics@vu.edu.au
Hard copy applications to be delivered to: Ethics Secretary, Victoria University Human Research Ethics Committee Office for Research Victoria University PO Box 14428, Melbourne VIC 8001 Or deliver in person to the Ethics & Biosafety Administration Group located within the Office for Research at Building C, Room 302, Footscray Park campus.	Hard copy applications to be delivered: Faculty Ethics Officer *Nominated* Faculty Human Research Ethics Committee Office for Research Victoria University PO Box 14428, Melbourne VIC 8001 Or deliver in person to the Ethics & Biosafety Administration Group located within the Office for Research at Building C, Room 302, Footscray Park campus.
For Further Information: Web: http://research.vu.edu.au/hrec.php	Telephone: 9919 4148 or your Faculty Ethics Officer

I attach a proposal for a project involving human participants for the purposes specified on the attached sheets.

Note: The Human Research Ethics Committee normally grants approval for periods of up to two years, subject to annual review. Consideration will be given to granting approval for a longer period in certain circumstances. Applications for extension of approval should be lodged prior to expiry of existing approval.

1. Project Title:

Understanding the Ageing Effects on Gait Control Dynamics and Falling Risks

2. Principal Investigator/s:
(Projects to be undertaken by students should list the Supervisor as the Principal Investigator)

Title	*First Name*	*Surname*	*School/Centre*	*Phone Numb*	*Mobile Number*	*VU E-Mail Address*
A/P	Rezaul	Begg	School of Sport and Exercise Science City Flinders Campus	0399191116		Rezaul.Begg@vu.edu.au

***3. (a)* Associate Investigator/s and/or Co-Investigator/s:**
(Please insert additional lines & information if there is more than one)

Title	*First Name*	*Surname*	*School/Centre*	*Phone Numbe*	*Mobile Numbe*	*E-Mail Address*
Dr.	Tony	Sparrow	School of Sport and Exercise Science City Flinders Campus	0399191183		Tony.Sparrow@vu.edu.au

3. (b) VU Sponsor:

*(For a*pplications for research involving participants from individuals who are not staff members of VU. *Please refer to declaration page for further details and signature)*

Title	First Name	Surname	School/Centre	Phone Numbe	Mobile Numbe	VU E-Mail Address

***4.* Student Project**

(Please insert additional lines & information if required)

4.1. Is the application part of a student project? Yes √ No ☐

4.2. If YES, select the appropriate tick box:

PHD √ Masters by Research ☐ Honours ☐ Postgraduate Coursework ☐ Undergraduate (not honours) ☐

- Has this research project been approved by the Postgraduate Research Committee?
  Yes √ No ☐

Student details

Title	First Name	Surname	School/Centre	Student Numb	Phone Numb	VU E-Mail Address
Mr	Hanatsu	Nagano	School of Sport and Exercise Science City Flinders Campu	3694502	0413102799	hanatsu.nagano@live.vu.e

- Is the student currently enrolled at Victoria University?
  Yes √ No ☐

---

5. Type of Project:
(please select Yes or No to the following questions)

5.1. Type of Program

(a) Is application for a higher degree program?

Yes √ No ☐

(b) Is this application for a pilot program of a higher degree?

Yes ☐ No √

[If yes, please note that a second application will be required for the full program]

(c) *Is application for an honours program of an undergraduate degree?*

Yes ☐ No √

*If yes, please indicate semester dates:*______________________________________

(d) *Other student project? Please specify* ____

5.2. Funded Program

(a) *Is application for a funded research program?*

Yes ☐ No √ ☐

*If yes, please indicate source of funding:*____________________________________

____

(b) *Do you require ethical approval prior to funding being granted?*

Yes ☐ No √

If yes, ***attach*** *any necessary forms to be completed by the Ethics Committee and indicate* ***grant closing date****.*

Date:_______________

5.3. Intrusiveness of Project

(please select Yes or No to the following questions)

a) Uses physically intrusive techniques

Yes √ No □

b) Causes discomfort in participants beyond normal levels of inconvenience

Yes √ No □

c) Examines potentially sensitive or contentious areas

Yes □ No √

d) Uses therapeutic techniques

Yes □ No √

e) Seeks disclosure of information which may be prejudicial to participants

Yes □ No √

f) Uses ionising radiation

Yes □ No √

g) Uses of personal information obtained from a Commonwealth department or agency

Yes □ No √

I. If YES, and the project is <u>not</u> medical research, does the research meet the Guidelines under Section 95 of the Privacy Act 1988?

Yes □ No □

II. If YES, and the project is medical research (including epidemiological research) does the research meet the Guidelines under Section 95A of

the Privacy Act 1988?

Yes ☐ No ☐

h) Clinical trial

Yes ☐ No √

(A clinical trial is a study involving humans to find out whether an intervention, including treatments or diagnostic procedures, which it is believed may improve a person's health, actually does so. A clinical trial can involve testing a drug, a surgical or other therapeutic or preventive procedure, or a *therapeutic, preventive or diagnostic device or service. Any intervention, including so-called "natural" therapies and other forms of complementary medicine, can be tested in this way).*

i) Research focuses on Aboriginal and/or Torres Strait Islander Peoples

Yes ☐ No √

- If YES, does the project involve health research?

  Yes ☐ No ☐

j) Involves potentially vulnerable groups (eg children, people in dependent/unequal relationships, highly dependent on medical care, cognitive impairment or intellectual disability, may be involved in illegal activities)

Yes ☐ No √

If YES, please provide additional detail: ________________

k) Involves deception or covert observation

Yes ☐ No √

If YES, please provide additional rationale: ________________

Note: If you have ticked "YES" to any of the items g to k, please forward your ethics application to the Secretary, Victoria University Human Research Ethics Committee (VUHREC). Note that Faculty HREC submission deadlines differ to that of the VUHREC, and this may impact on your project's timelines.

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6. Aim of project:

(In brief terms, state the aims and the expected benefits of the project in no more than 250 words)

The aim of the proposed research is to study the gait pattern of young and older individuals to investigate the effects of ageing on gait control mechanisms, and assess any risk associated with three major types of falls: tripping, sideways balance loss and slipping. The research will focus on a number of biomechanical parameters, including: 1) shoe end-point control and foot clearance (tripping risk); 2) body centre of mass movement (linked to sideways balance loss), and 3) foot-ground reaction forces during heel contact (slipping risk). Gait testing will be conducted under three different walking conditions: 1) normal walking, 2) walking with a wide step width (normal + 50%), and 3) walking with a narrow step width (normal - 50%).

The results will reveal how the width of the walking base of support can affect the walking patterns in young and older adults, and influence any associated risks of sustaining a fall. Gait recordings will also be done while wearing a custom-made shoe-insole to evaluate the effectiveness of a footwear intervention to minimise the risk of tripping during walking.

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7. Plain language statement of project:

(Provide a brief summary of the project [not more than 2 pages] outlining the broad aims, background, key questions, research design/approach and the participants in the project. Include a theoretical background or context of the research. If there are multiple participant groups or interventions/phases, please specify relevant information for each. Please make sure implications associated with multiple groups/phases is addressed throughout the application. It is recognised that in some areas of research, it may be appropriate that this statement is repeated elsewhere in this application form, and that it may comprise part of your response to questions 6, 8, 15, 16 and 17. This section is to be stated in simple language and any terms or jargon must be accompanied by explanation).

Falls during walking among older adults (over 65 years) is a major healthcare issue in Australia and worldwide due to its high frequency, high medical cost, and high injury rate (Hill et al., 1999; Keskin et al., 2008; Stalenhoef et al., 2002). Among the factors that are responsible for falls, tripping, sideway balance loss, and slipping have been identified as the main causes of falls, and they account for about 74% of all falls (Sherrington and Menz, 2003). Falls prevention therefore requires strategies for minimizing tripping, sideway balance loss, and slipping. Past studies have identified a strong link between minimum foot clearance (MFC) data and the risk of tripping; medio-lateral body centre of mass motion and sideways balance loss; and, ground reaction force characteristics with the risk of slipping (Begg et al., 2007; Leclerzq, 1999; Lord et al., 2007; Whittle, 2007). Investigating these gait parameters in both young and older adults will allow us to better understand how the risks of these three types of fall are influenced by ageing. Width of the base of support can affect the dynamic balance and older adults are known to increase their step width to stabilise their medio-lateral (sideways) balance (Lord et al., 2007; Whittle, 2007). However, there is controversy in the literature as to the effectiveness of such adaptation in assisting older adults to avoid a fall (Schrager et al., 2008). The current study will investigate how the base of support can influence walking patterns.

The proposed research will compare three gait parameters (MFC, centre of mass, and foot-ground reaction force) in both young and older adults. For an accurate estimation of body centre of mass the whole body motion will be recorded. Gait testing will be conducted on a 10m laboratory walkway under three walking conditions. For each of these conditions, the participants will be asked to complete about 50 to 100 trials, to obtain approximately 100 gait cycles' data to develop accurate histograms and assess risks. The 3 conditions will include: 1) walking with normal step width, 2) walking with a wide step width (normal + 50%), and 3) walking with a narrow step width (normal - 50%). Participants' natural step width will be obtained during their preferred normal walking.

A further motivation of this study is to test whether a shoe-insole can be used as an intervention technique to reduce the risk of sustaining a fall. Results from past studies (e.g., Begg et al., 2007; Moosabhoy et al., 2006; Nakajima et al., 2009; Schrager et al., 2008) implicated that a shoe-insole with certain specific features (e.g., increased dorsiflexion) may be useful to reduce the risk of foot contact with obstacles.

Participants will include 30 healthy young (aged 18-35 years) and 30 healthy older adults (aged 65+ years). The young participants will be volunteers from the Victoria University community. The older participants will be recruited from a subject pool maintained by the Biomechanics Unit. All participants will be active and be able to walk for at least 30 min continuously. Individuals affected by conditions that might directly affect their gait (such as any neurological, musculoskeletal, cardiovascular, or respiratory disorder; rheumatoid arthritis, or diabetes) will be excluded. Participants will be also excluded if they have experienced a fall within 2 years. Prior to testing each participant will be asked to complete a consent form (Attachment A) and 'general health survey' questionnaire (Attachment C) to assess whether potential participants are free from the conditions described above. Visual acuity (< 6/12; Nicklason et al., 2002), contrast sensitivity (Melbourne edge test < 6/15; Lord et al., 2001) and Timed Up & Go (> 13.5 secs; van Iersel et al., 2008) tests will be also conducted to screen participants.

Statistical analyses will be undertaken to examine the descriptive parameters of MFC, centre of mass, and ground reaction forces. Age, limb, footwear, and walking condition effects will be examined.

References:

Begg, R., Best, R., Dell'Oro, L. & Taylor,S. (2007) Minimum foot clearance during walking: Strategies for the minimisation of trip-related falls. *Gait & Posture*, 25: 191-198.

Hill, K., Schwarz, J., Flicker, L., & Carroll, S. (1999) Falls among healthy, community-dwelling, older women: a prospective study of frequency, circumstances, consequences and prediction accuracy. *Australian and New Zealand Journal of Public Health,* 23: 41-48

Keskin, D., Borman, P., Ersöz, M., Kurtaran, A., Bodur, H., & Akyüz, M. (2008) The risk factors related to falling in elderly females. *Geriatric Nursing,* 29: 58-63.

Lord, S., Sherrington, S., Menz, Hylton., Close, J. (2007) Falls in older people: Risk factors and strategies for prevention. *Cambridge University Press*, Cambridge.

Lord, SR., & Daybew, J. (2001) Visual Risk Factors for Falls in Older People. *JAGS,* 49: 508-515.

Lesclercq, S. (1999) The prevention of slipping accidents: a review and discussion of work related to the methodology of measuring slip resistance. *Safety Science*, 95-125.

Moosabhoy, MA., & Gard, SA. (2006) Methodology for determining the sensitivity of swing leg toe clearance and leg length to swing leg joint angles during gait. *Gait & Posture,* doi: 10. 1016/j.gaitpost.2005. 12.004.

Nakajima, K., Kakihana, W., Nakagawa, T., Mitomi, H., Hikita, A., Suzuki, R., Akai, M., Iwaya, T., Nakamura, K., and Fukui, N. (2009) Addition of an arch support improves the biomechanical effect of a laterally wedged insole. Gait & Posture 29, 208-213.

Nicklason, F. (2002) Falls and gait disorders and their relation to drug therapy in the elderly. Royal Hobart Hospital and Falls Injury Prevention Clinic.

Schrager, MA., Kelly, VE., Price, R., Ferrucci, L., Shumway-Cook, A. (2008) The effects of age on medio-lateral stability during normal and narrow base walking. *Gait & Posture* 28, 466-471.

Sherrington, C., Lord, SR., & Finch CF. (2004) Phyysical activity interventions to prevent falls among older people: update of the evidence. *Journal of Science and Medicine in sport* 7 (1): Supplement: 43-51.

Stalenhoef, PA., Diederiks, JPM., Knottnerus, JA., Kester, ADM., & Crebolder, HFJM. (2002) A risk model for the prediction of recurrent falls in community-dwelling elderly: A prospective cohort study. *Journal of Clinical Epidemiology,* 55: 1088-1094.

van Iersel, MB., Munneke, M., Esselink, RAJ., Benraad, CEM., & Olde Rikkert., MGM. (2008) Gait velocity and the timed-up-and-go test were sensitive to changes in mobility in frail elderly patients. *Journal of Clinical Epidemiology,* 62: 186-191.

Whittle, M. (2007) Gait analysis: an introduction. 4th edition. Butterworth-Heinemann Elsevier.

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8. Nature of research, including methodology and a list of all procedures to be used on human participants. Please include a statistical power analysis statement if applicable.

The participants will include 30 healthy young (aged 18-35 years recruited from Victoria University community) and 30 healthy older adults (>65 years, recruited from the VU Biomechanics database and through advertisement in the local newspaper). The sample size for the proposed project has been calculated based on results from our recent work of minimum toe clearance data involving young and elderly individuals (Begg et al., 2007). Individuals presenting to participate will not be included if, following a health status questionnaire (Attachment C), they have any condition that might directly affect their mobility and balance, such as a neurological, musculoskeletal, cardiovascular, or respiratory disorder, rheumatoid arthritis, or diabetes.

All testing methods will be explained to the participants prior to the actual testing session. In this information session, the participants will be made aware that they can withdraw from testing at any time without explanation. To maintain confidentiality participants' names will not be published. Prior to commencing the testing procedure all participants will be required to fill in a consent form based on Victoria University standards (Attachment A). They will also complete a general health questionnaire (Attachment C).

Prior to the walking test, a ‘visual acuity’ test (vision), ‘contrast sensitivity’ test (vision), and three ‘time up & go’ tests (mobility and balance function) will be undertaken. The gait tests will examine swing foot trajectory data, centre of mass motion, and ground reaction forces. The subjects will have IRED (small markers) placed as rigid bodies on the segment landmarks as illustrated in Figure 1. Additionally, markers will be attached to upper body on the following locations: back of the neck and upper arms.

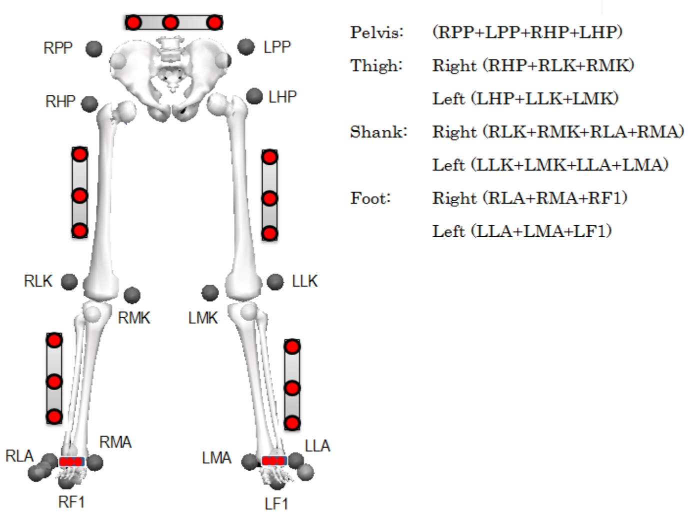

Figure 1: Pelvis and lower limb marker setup: rigid body markers are displayed as groups of 3 located on the segments, while imaginary marker registrations of joint centres (Arnold et al., 2007; Delp et al., 2007; Liu et al., 2008; Xiao et al., 2008)

Surface Electromyography (EMG) data will be recorded from the following muscles: hamstrings, biceps femoris long head, rectus femoris, gastrocnemius, and anterior tibialis as per standard biomechanical procedures (Arnold et al., 2007).

First, normal walking test will be conducted; followed by larger (+50%) and narrower (-50%) step width walking. Here, subjects will be asked to walk along a 10 meter walkway at their comfortable walking speed. At the middle of the walkway, the Optotrak motion system will record approximately two to three strides from each limb and the two forceplatforms (AMTI) will record ground reaction force when each foot is in contact with the walking surface. This measurement will be repeated to collect approximately 100 gait cycles data for each walking condition (50-100 walking trials); this is necessary to obtain reliable histograms to assess the risk of falls (Best and Begg, 2008). The entire experiment will be repeated with the ‘shoe-insole’ condition. Considering that walking trial is performed only in one direction, participants will be required to walk back to the starting point after each walkthrough trial. The total distance each participant will be asked to walk in a single experimental day will be therefore estimated to be between 2km and 4km (10m walkway x 50-100 trials x 2 footwear conditions x 2 two-way path).

The testing will be spread over three days with about 2 hours/day to avoid any fatigue effects and to ensure adequate rest during the testing sessions.

Statistical analyses using SPSS software will be undertaken to examine the descriptive parameters of MFC, centre of mass, and ground reaction force at heel contact from the gait tests with the aim of determining differences between age groups, walking conditions (normal, larger (+50%), and narrower (-50%) step width), and the two shoe-insole conditions. Dependent variables to be measured will be foot clearance (e.g., MFC), centre of mass acceleration, and ground reaction force kinetics, in addition to common time-distance features (e.g., walking velocity, cadence, step length, step width, swing time, and double support time). Further data analysis will be undertaken using OpenSim software (Arnold et al., 2007) to derive joint kinetics (moments and power) as well as obtaining simulation results from the developed biomechanical models.

References:

Arnold, AS., Schwartz, MH., Thelen, DG, & Delp, SL. (2007) Contributions of muscles to terminal-swing knee motions vary with walking speed. *Journal of Biomechanics*, 40: 3660-3671.

Begg, R., Best, R., Dell'Oro, L. & Taylor,S. (2007) Minimum foot clearance during walking: Strategies for the minimisation of trip-related falls. *Gait & Posture,* 25: 191-198.

Best, R., & Begg, R. (2008) A method for calculating the probability of tripping while walking. *Journal of Biomechanics* 41, 1147-1151.

Delp, SL., Anderson, FC., Arnold, AS., Loan, P., Habib, A., John, CT., Guendelman, E., & Thelen, DG. (2007) OpenSim: Open-source software to create and analyse dynamic simulations of movement. *IEEE Transactions on Miomedical Engineering,* 54: 11.

Liu, MQ., Anderson, FC., Schwartz, MH., & Delp, SL. (2008) Muscle contributions to support and progression over a range of walking speeds. *Journal of Biomechanics*, 41: 3243-3252.

Xiao, M., & Higginson, JS. (2008) Muscle function may depend on model selection in forward simulation of normal walking. *Journal of Biomechanics,* 41: 3236-3242.

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9. Description of those techniques which are considered by the profession to be established and accepted. Please give details of support for their application.

(If, in the course of your research, procedures are significantly varied from those stated here, the Human Research Ethics Committee must be informed).

Refer to section 8

All biomechanical procedures are routinely used as part of a number of projects undertaken within the biomechanics unit. This project protocol is identical to previously approved ethics applications (e.g., HRETH: 06/260; HRETH.FHD.032/02; HRETH08/75; HRETH 08/193)**.**

10. Proposed start and end date of project:

(Note: new research projects may not commence prior to approval by the Human Research Ethics Committee).

Proposed start date:	(01/05/2010)	**Proposed end date:**	(01/05/2012)

11. Details of participants:

Name of Phase/Group	**Young Adults**	**Older Adults**	
Number of Participants	**30**	**30**	
Type	**Male**	**Male**	
Age Range	**18-35 years**	**65+ years**	

12. Source of participants

(specify for each group/phase if relevant), and means by which participants are to be recruited)

Younger subjects will be recruited by advertisements placed on notice boards throughout Victoria University. This is a commonly used and successful procedure adopted by the Biomechanics research group. Recruitment posters have been attached to this document (Attachment D).

Older subjects will be recruited from a subject pool maintained by the 'Biomechanics Unit'. These older adults are regular participants as volunteers in gait projects involving similar research protocols. All interested persons will follow a process before approval of participation is provided. First, subjects will complete an in person phone interview where background details will be gathered. Provided initial inclusion requirements have been met, a mail out to the participant will contain further information about the project (Attachment B) and re-iterate the exclusion criteria listed in section 7. A consent form (Attachment A) will be included in the mail out to notify the participant of the details of the study conditions.

On the testing day, 'General Health Survey Questionnaire' (Attachment C) will be given to each participant to confirm whether they will meet the participation criteria including; 1) living independently, 2) being able to perform routine daily activities, 3) being free of any known cognitive, orthopaedic or neurological abnormalities, 4) being able to walk for at least 30 minutes continuously, and 5) experiencing no falls for the past two years.

Prior to gait experiment, a series of pre-tests will be then conducted for the further screening. Both young and older volunteers will be excluded if they exceeded 12 s on a 'timed up and go test' or score less than 20 on a visual contrast sensitivity test ('Melbourne Edge Test').

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13. Is there any payment of participants proposed: Yes ☐ No √

- If yes, state the amount:
- Provide rationale for payment and the amount:
- Describe the financial controls applicable to the program:

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14. Premises on which project is to be conducted:

If using an institution/s other than Victoria University, attach a copy of documents giving approval to use participants or premises in the relevant institution/s.

Victoria University, 300 Flinders Street,

Biomechanics Laboratory, Basement

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15. Dealing with potential risks (for each phase/group where applicable):

*(a) Indicate any **physical risks** connected with the proposed procedures*

The physical risks associated with gait testing are minimal and are no greater than the risk involved while undertaking normal daily walking.

*(b) Indicate any **psychological risks** connected with the proposed procedures*

There are ***no psychological risks*** identified.

*(c) Indicate any **social risks** connected with the proposed procedures*

There are **no social risks** anticipated by the investigators.

*(d) Indicate any **legal risks** connected with the proposed procedures*

There are **no legal risks** anticipated by the investigators.

*(e) Indicate if there are any **other risks** connected with the proposed procedures*

N/A

(f) Management of the potential risks identified above- indicate how each of these potential risks will be minimised and/or managed if they occur (if risks have <u>not</u> been identified in 15 a – e, go to item 16).

(i) how risks are to be minimised:

Gait testing will be conducted on a 10m non-slippery and unobstructed walkway at participants' preferred walking speed. Participants will be excluded from the research if they do not fulfil the requirements detailed in section 7. They will be free of any pathological conditions and will be capable of walking for at least 30 min continuously. Adequate rest will be provided during testing sessions as outlined in section 8. The walking conditions do not require any challenging environment but requires some adjustments to their step width. For example, from Nagano's honours study, it was found that older adults walked with a mean of 8cm step width. If similar results are found in this study, it would mean a range of 4cm (narrow) to 12cm (wide) step widths for the older adults. Shoe-insoles used in the experiment will change the ankle angle by a maximum of 3-4° only, and these will be manufactured by a professional orthotists. In addition, a general health survey will confirm that the participants are in good health. Thus, the risks associated with gait testing are not expected to be more than those associated with their daily activities.

(ii) how adverse events would be managed if they were to occur:

A researcher will stand by, and in the event of any unsteadiness, testing will be stopped and participation reviewed. Telephones are located within the facility if further medical attention is needed, and the subjects will be escorted to a convenient hospital.

(g) *If you consider there to be no potential risks, explain fully why no potential risks have been identified.*

N/A

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16. If you consider the participants to be 'at risk', give your assessment of how the potential benefits to the participants or contributions to the general body of knowledge would outweigh the risks.

The potential risk associated with the project is minimal as it only involves normal everyday walking activities. The project aims to identify how ageing will change the walking performance and what types of falls older adults will be prone to. Therefore, contributions by this study will be very important in understanding falling risks and also to developing a falls prevention strategy. In addition, the effectiveness of 'shoe-insoles' will be tested; this will be potentially applied to reduce the risk of falls in the older population.

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17. Informed Consent (If materials are to be distributed in languages other than English, a copy of non-English version and a letter from an independent person verifying accuracy of content is required):

(a) *As part of the informed consent process, it is necessary to provide information to participants prior to obtaining consent.* ***Please attach a copy of your 'Information to Participants Involved in Research' Letter*** *[See http://research.vu.edu.au/hrec.php for a template] with information about your research that you intend to give to potential participants. This needs to:*

- *state briefly the aims, procedures involved and the nature of the project, including a clear indication of any potential risks associated with this project;*
- *if you consider participants to be 'at risk' (see Question 16), state exactly what the researcher will communicate to the participant (this must be stated in clear and concise language) in order to obtain informed consent. This must be in a written format that is given to the participant particularly for this purpose; and*
- *be written in language which may readily be understood by members of the general public, with explanation of any technical terms.*

See attachment B

(b) *Please **attach a copy of your <u>Consent form</u>** [See <u>http://research.vu.edu.au/hrec.php</u> for a template consent form.]*

See attachment A

(c) *State the process you will use to obtain documentation of informed consent hereunder…*

(It is essential to clearly detail the steps involved in obtaining informed consent. It is recommended that a procedure or flow chart be attached as an appendix commencing from the recruitment stage to consent taking into consideration issues such as communications and awareness of recruitment, provision for considering participation, etc.)

The aims, procedures and the potential risks associated with the project will be explained to the participants by Hanatsu Nagano (student investigator). They will also be given a copy of the 'Informed Consent' form (Attachment A) and 'Information about the project' to read (Attachment B). After going through the details, they will have the option to participate in the study or if they choose to, they will be able to withdraw from the study

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18. Confidentiality:

(a) Describe the procedures you will adopt to ensure confidentiality.

All data extracted will be identified using a code. Consent and questionnaire forms will be locked in a filing cabinet.

(b) Indicate who will be responsible for the security of confidential data, including consent forms, collected in the course of the research. (Note: the Principal Investigator should be nominated as the responsible person in this section. An alternative person may be nominated with clear justification)

Associate Prof Rezaul Begg

(c) Indicate the period for which the data will be held. (Data must be held for at least 5 years post-publication. Please refer to section 3.2 of the University's **Code of Conduct for Research, 1995***).*

5 years.

(d) *Name all people who will be granted access to the data and the reason for the access. People identified are required to maintain all aspects of confidentiality.*

A/P Rezaul Begg, Dr Tony Sparrow and Mr Hanatsu Nagano will have access to the data for analysis.

---

19. Privacy:

(a) *Does this project involve the use of personal information obtained from a Commonwealth department or agency?*

Yes ☐ No √

If YES you may need to comply with the requirements of the Privacy Act 1988.

Under the Commonwealth Privacy Act 1988 disclosure of personal information by Commonwealth agencies is not permitted except in a number of circumstances specified in Information Privacy Principle (IPP) II. These include consent by the individual concerned. Where consent has not been given, and where none of the other circumstances specified in IPP II apply, additional guidelines for consideration of the project application and for conduct of research apply. Note that the Act does not apply to publicly available material (such as electoral rolls).

If a Commonwealth agency (for instance, the Australian Bureau of Statistics, Commonwealth Government departments, Australian Electoral Commission, most Repatriation Hospitals) is involved in the collection, storage, security, access, amendment, use or disclosure of personal information for a research project investigators must ensure that the project complies with the requirements of the Act.

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20. Conflict of interest

Is there a conflict of interest between any of the researchers and potential participants in the research (i.e due to a relationship between researcher and participant population)?

Yes ☐ No √

If yes, provide details and ensure that the conflict is identified and addressed in Section 15.

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21. Research in other countries.

Is any part of the program to be conducted in another country?

Yes ☐ No √

If yes, please provide information about any relevant legal or regulatory requirements and any ethical review processes in that other country.

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22 Is approval required for data collection from other organisations? If so, please provide information of consent process (attach evidence of approval/s)

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23. Collaborative program

Does the program involve collaboration with another institution?

Yes ☐ No √

If YES, please describe the arrangements with the other institution/s for managing the program including, if appropriate, confidentiality, intellectual property, ethics and safety clearances, reporting to appropriate agencies and the dissemination of research findings.

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24 Other relevant comments (including information that you deem necessary to inform the HREC that may impact on the project)

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25. Application Review Check list

A completed and signed *Application Review Check List* must be submitted with all applications. A copy may be downloaded from the Victoria University Human research ethics webpage at: http://research.vu.edu.au/hrec.php

Important: Attach Application Review Form on cover

Has the Principal Investigator completed and signed the Application Review Form?

Yes √ No ☐

Is the Application Review Form attached with a hard copy of this application?

Yes √ No ☐

VICTORIA UNIVERSITY | A NEW SCHOOL OF THOUGHT

DECLARATION FORM

I, the undersigned, have read the current NH&MRC Statement on Human Experimentation and the relevant Supplementary Notes to this Statement, or Code of Ethics for the Australian Psychological Society, (or *) and accept responsibility for the conduct of the experimental and research procedures detailed above in accordance with the principles contained in the Statement and any other condition laid down by the Human Research Ethics Committee.

Principal Investigator (1) Print Name: A/Prof Rezaul Begg

Signature ______________________________ Date _________________

Principal Investigator (2) Print Name: ____________________________________

Signature ______________________________ Date _________________

Associate Investigator ** Print Name: Dr Tony Sparrow

Signature ______________________________ Date _________________

VU Sponsor ***

Print Name: ____________________________________

Signature ______________________________ Date _________________

<u>Student/s Details</u> (*If the project is to be undertaken by a student, please provide details):*

Name: Hanatsu Nagano

Signature ______________________________ Date _________________

<u>Co-Investigator</u> Print Name: _____________________________

Signature ______________________________ Date _________________

I, the undersigned, understand that the above person/s have read the current NH&MRC Statement on Human Experimentation and the relevant Supplementary Notes to this Statement, or Code of Ethics for the Australian Psychological Society, (or *) and that responsibility is accepted by the above person(s) and by this Department for the conduct of the experimental and research procedures detailed above in accordance with the principles contained in the Statement and any other condition laid down by the University Human Research Ethics Committee and fully support the project undertaken within the Department and Faculty.

<u>Head of Department</u>

Print Name: ____________________________

Signature ______________________________ Date _________________

* If NHMRC Statement or APS Code are not appropriate to your project, please identify your professional code of ethics under which this project would operate.

** The Associate Investigator will assume responsibility for the project in the absence of the Principal Investigator.

*** Applications for research involving participants from individuals who are not staff members of VU and who require access to the cohort of VU staff or students to undertake their research. Such research proposals are to be 'sponsored' by a member of staff, who would be required to take responsibility for all interactions with the University and the HREC in relation to ethics issues and their management.

Appendix III **<u>Consent Form</u>**

CONSENT FORM

FOR PARTICIPANTS

INVOLVED IN RESEARCH

INFORMATION TO PARTICIPANTS

We are pleased to invite you to be a part of a study involving gait control mechanisms and the risk of falls. The aim of this research is to further our understanding of how older adults may differ from younger persons when walking. You are asked to participate in the testing procedures outlined below.

- This research has been approved by the Victoria University Human Research Ethics Committee.
- The physical risks associated with the procedures are minimal.
- The testing area will be kept private with access limited only to the researchers.
- All data will be kept confidential and only the researchers will have access to the data files.
- Please be advised that although you are volunteering for this research, you are free to withdraw at anytime.

CERTIFICATION BY SUBJECT

I, ____________________________________

of ____ certify that I am above 18 years old and that I am voluntarily giving my consent to participate in the research entitled: **Understanding the Ageing Effects on Gait Control Dynamics and Falling Risks,** being conducted at Victoria University by: **A/Prof Rezaul Begg (supervisor), Dr. Tony Sparrow (co-supervisor), and Mr Hanatsu Nagano (student researcher).**

I certify that the objectives of the research, together with any risks to me associated with the procedures listed hereunder, have been fully explained to me by **Mr Hanatsu Nagano.** I freely consent to participating in these procedures.

PROCEDURES

- The tester will take your body height and weight measurements
- Attachment of light and small plastic shells to shoes using Velcro straps. These shells have small ‘diode’ markers that are tracked by a 3D camera system. Small diodes are also attached to joint landmarks on the upper body (neck, back, and hip) and right/left upper/lower limbs. The markers are connected to a small control box with wire cables. The control box will be attached to a waist belt. The markers are powered by low voltage batteries and will be fastened along the outside of the shoe using adhesive tape, safety pins and plastic clips. These wires will not affect walking. The research will be conducted on three different days, and the total distance each participant will be asked to walk in a single experimental day will be 2-4km (10m walkway x 30trials x 2 footwear conditions x 2 two-way path).

Day 1

- Approximately 100 walking trials over ground walking along a 10 meter walkway at preferred normal walking speed. This will take approximately 20-30 minutes.
- Rest for 10-15 minutes.
- Another 100 trials with the ‘shoe-insoles’

Day 2

- 100 trials of over ground walking along a 10 meter walkway at preferred walking speed along lines that represent normal step width + 50%. This will take approximately 30 minutes.
- Rest for 10-15 minutes.
- Repeat another 100 trials with the ‘shoe-insoles’

Day 3

- 100 trials of over ground walking along a 10 meter walkway at your normal walking speed with the narrower step width (i.e., Normal -50% step width obtained during Day 1). This will take approximately 30 minutes.
- Rest for 10-15 minutes.
- Repeat another 100 trials with the ‘shoe-insoles’

I certify that the objectives of the study, together with any risks and safeguards associated with the procedures listed hereunder to be carried out in the research, have been fully explained to me by: Hanatsu Nagano. I certify that I have had the opportunity to have any questions answered and that I understand that I can withdraw from this research 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:________________________________

Date: ________________

Any queries about your participation in this project may be directed to the researcher

Prof Rezaul Begg (ph 03 9919 1116) or Mr Hanatsu Nagano (ph 03 9919 1128). If you have any queries or complaints about the way you have been treated, you may contact the Secretary, Victoria University Human Research Ethics Committee, Victoria University, PO Box 14428, Melbourne, VIC, 8001 phone (03) 9919 4781.

Appendix IV **Information to participants involved in research**

INFORMATION

TO PARTICIPANTS

INVOLVED IN RESEARCH

You are invited to participate

You are invited to participate in a research project entitled "**Understanding the Ageing Effects on Gait Control Dynamics and Falling Risks**."

This project is being conducted by A/Prof Rezaul Begg, Dr Tony Sparrow and Hanatsu Nagano from the School of Sport and Exercise Science at Victoria University.

Project explanation

The aim of the proposed research is to examine the gait pattern of both young and older individuals to examine the effects of ageing on gait control mechanism and associated risk of the three types of falls: tripping, sideways balance loss and slipping. To understand the gait pattern and assess the risk of falls, the research will focus on biomechanical investigation such as the toe (shoe end) clearance (tripping risk), body centre of mass motion (balance risk), and the forces applied to the ground during walking (slipping risk). Gait testing will be conducted under three walking conditions: normal walking, walking with wide step width (normal + 50%), and narrow step width (normal - 50%) walking. Another aim of this current study is the use of shoe-insoles to evaluate the effectiveness of the footwear intervention to minimise the risk of falls.

The participants will be 30 healthy young volunteers aged 18-35 years and 30 healthy older volunteers aged 65 or above. The study will exclude anyone with conditions that might directly influence their walking and balance such as neurological, musculoskeletal, cardiovascular, respiratory disorder, rheumatoid arthritis, or diabetes. All research methods will be explained prior to testing and participants will be able to stop testing at any time without need for explanation. To abide by confidentiality participants' names will not be published. Prior to commencing all participants will be required to fill in a consent form, based on Victoria University standards. Participants will also complete a general health questionnaire (Attachment C).

What will I be asked to do?

1 – All participants will be equipped with small markers attached on the important joint landmarks including the upper body (neck, back, hip) and right/left upper/lower limbs to record the motion of the body.

2 – Participants will be first asked to walk naturally at their preferred walking speed with the standard shoe-insoles on a 10m walkway repeatedly for a maximum of 100 trials. Participants will be asked to repeat the same task wearing 'shoe-insoles'.

3 – In the following week, participants will be asked to complete the second walking task, walking with the larger (normal +50%) step width for a maximum of 100 trials. Participants will be asked to repeat the same task wearing 'shoe-insoles'.

4 – In the third week, the participants will be asked to complete the final walking task, walking with a narrow (normal-50%) step width, for a maximum of 100 trials. Participants will be asked to repeat the same task wearing 'shoe-insoles'.

The total distance a participant will be asked to walk in a single experimental day will be 2-4 km (10m walkway x 30 trials x 2 footwear conditions x 2 two way path). During the experiment, the participants will be allowed to rest anytime and if required withdraw from the experiment.

What will I gain from participating?

- Making a contribution to gait research.
- Understanding more about gait biomechanics and risks of falls.

How will the information I give be used?

The information obtained through the experiment will be confidential and used only for the purpose of this study, which will include the research paper, thesis, posters, presentation, etc. The private information that can identify the individual (such as name, address, or contact information) will not be reported.

What are the potential risks of participating in this project?

There are no foreseen risks regarding to this experiment because the procedures involve only normal everyday activities and there are no invasive physiological or medical research techniques. If a participant feels uncomfortable during the test, the test will be stopped immediately and the participant will be given the option to withdraw. Telephones are located within the facility if medical attention is needed, and the participants will be escorted to a convenient hospital.

How will this project be conducted?

Participants will be asked to walk on a flat, unobstructed, lighted, and non-slippery 10m laboratory walkway repeatedly. All the participants will have IRED (small markers) placed on their upper body (head, neck, back, and hip) and right/left upper/lower limb segments for collecting data allowing gait to be characterized. Gait biomechanics will be examined by measuring swing foot trajectory, body centre of mass, and ground reaction force. It will be ensured that participants understand they can take rest any time and they can also withdraw from the experiment any time. The rest will be provided for participants when changing the walking task (shoe-insole).

Who is conducting the study?

School of Sports and Exercise Science, City Flinders Campus, Biomechanics Laboratory

Principal Investigator

A/Prof Rezaul Begg

Phone: 0399191116

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Student Investigator

Mr. Hanatsu Nagano

Phone: 0399191128

E-mail: hanatsu.nagano@live.vu.edu.au

Associate Investigator

Dr Tony Sparrow

Phone: 0399191183

E-mail: Tony.Sparrow@vu.edu.au

Any queries about your participation in this project may be directed to the Principal Researcher listed above. If you have any queries or complaints about the way you have been treated, you may contact the Secretary, Victoria University Human Research Ethics Committee, Victoria University, PO Box 14428, Melbourne, VIC, 8001 phone (03) 9919 4781.

Appendix V <u>General health survey questionnaire</u>

General Health Survey Questionnaire

Thank you for filling out this questionnaire.

1. Please fill out as much as you can.

2. When you are finished, please leave it with the investigator.

Section .01STATEMENT OF CONFIDENTIALITY

All information that would permit identification of investigators or their participants will be regarded as strictly confidential, will be used only for the purpose of operating and evaluating the study, and will not be disclosed or released for any other purposes without prior consent, except as required by law.

Please circle the correct answer to the following questions. Please use the space provided to add any additional information you believe is required.

1. In general, would you say your health is?

b. Excellent
c. Very good
d. Good
e. Fair
f. Poor

2. Have you previously fallen, or tripped, in the past 12 months?

Yes No (If no please skip to question 4)

If yes please provide date/s and description of fall/s and ________________________
____
____
____
____
____
____

3. Did the fall result in physical injury?

Yes No

If yes please describe injury and length of hospitalization and rehab if required?_____
____
____
____
____
____

4. Would you say you can walk comfortably without stopping for?

a. Less than 10 minutes
b. 10 to 30 minutes
c. 30 minutes to one hour
d. Greater than one hour

5. Do you live independently and require no aid for walking?

Yes No

6. Have you a history of orthopedic problems?

Yes No

If yes please explain
____

____
____
____

7. Do you suffer from any muscle or skeletal problems that you know of?

Yes No

If yes please explain____
____
____

8. Have you ever or do you currently suffer from any heart or respiratory problems that you know of?

Yes No

If yes please explain____
____
____
____

9. Would you say your balance while both standing and walking is?

Good Poor

10. Do you currently take any psychotropic / antipsychotic medications?

Yes No

If yes, how long have you been using that drug for?

____
____

Appendix VI Thank you letter and the report for older participants

Dear

Thank you for participating into our recent research entitled “Understanding the Ageing Effects on Gait Control Dynamics and Falling Risks.” Your participation was a precious contribution to advance the research area on “falls issues.”

Due to the tight testing schedule, it took a long time to analyse your data. Finally, we finished assessing your walking data, and we would like to give some of the basic outcomes to you. However, I need to emphasise that this information should not be used for clinical decision making. Should you have any concern with your gait, it is advised that you should go to see the medical doctor or qualified clinicians.

Victoria University cannot take any legal responsibility regarding the given data.

Although we have provided the basic information about gait terminology and guidance about how to interpret the data, please feel free to contact Hans at 03 9919 1128 for any further question.

Again, thank you very much for the contribution.

Best Regards,

Hanatsu Nagano

<u>Gait Data</u>

This page explains about the gait data that quantitatively describe your walking pattern.

1. Walking Velocity: walking speed (m/s)
2. Step Length: (anterior-posterior) distance of one step (cm)
3. Swing Time: the interval (s) between toe off and heel contact
4. Double Support Time: the time (s) when both feet are on the walking surface
5. Step Width: the medial (sideway) distance between the left and right foot segment centre of mass at heel contact (cm)

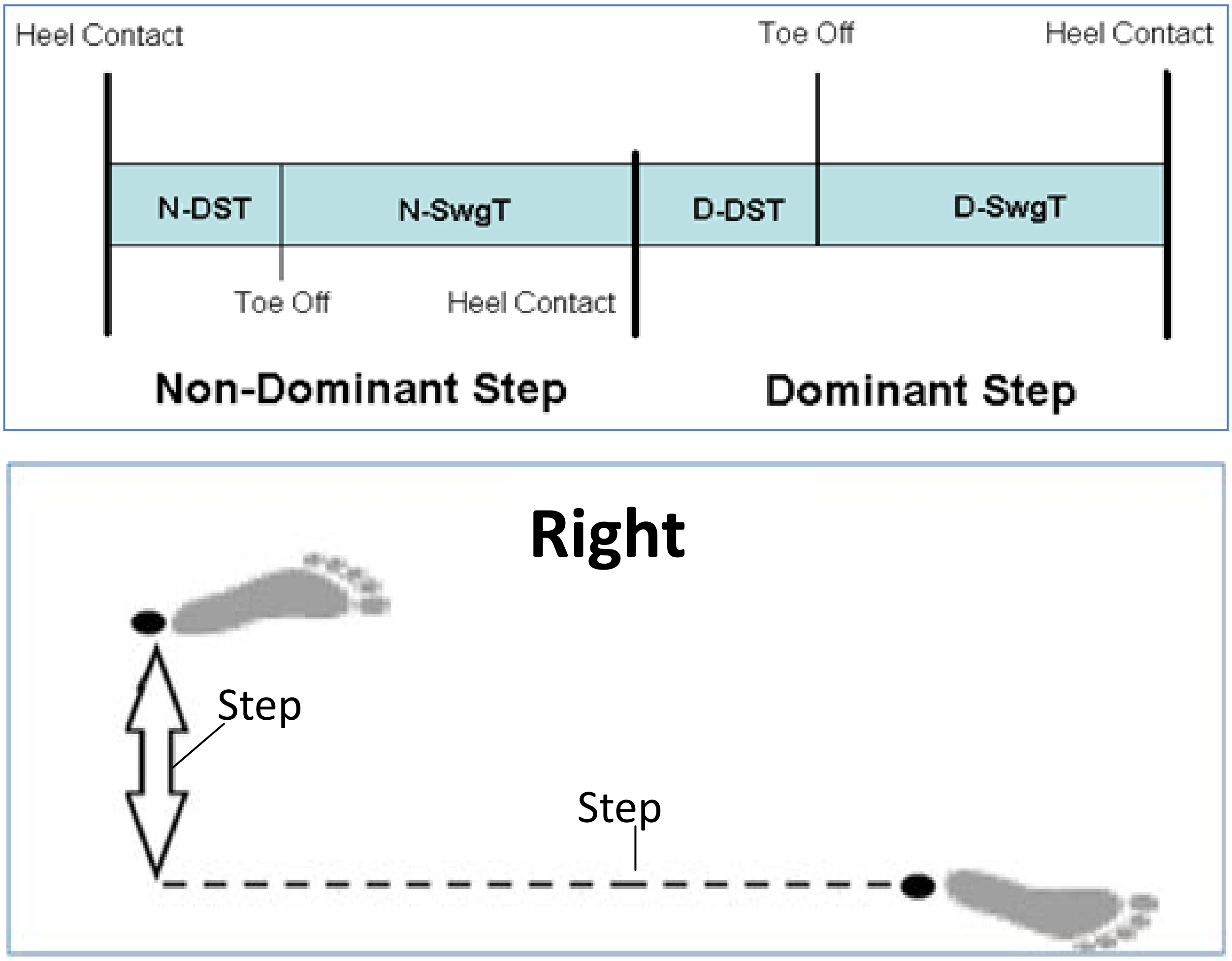


Figure 1 The illustration of examined gait parameters of both the dominant (D) step and non-dominant (N) step; double support time (DST); swing time (SwgT).

Results

The following Figure 2 compares your walking data with the young and older data.

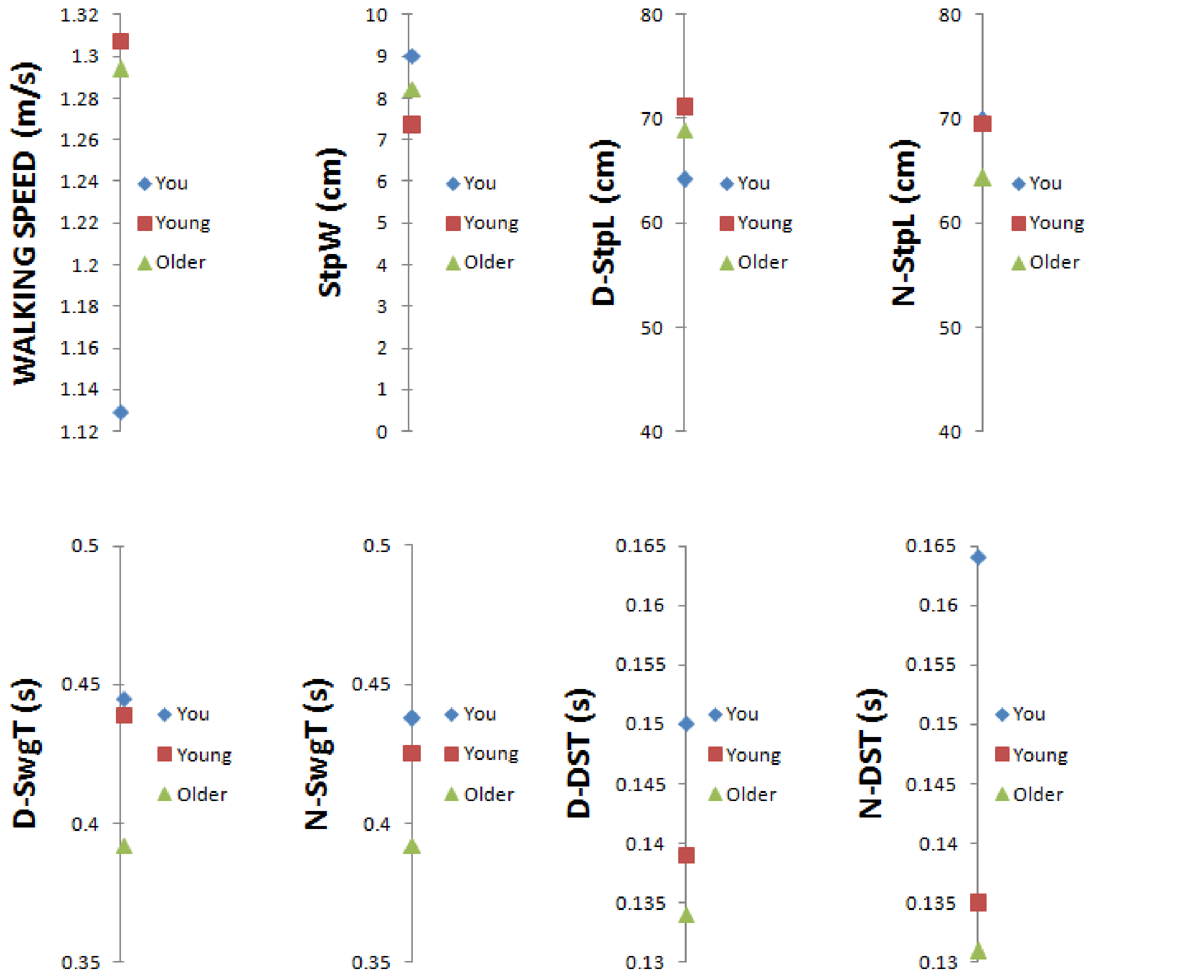
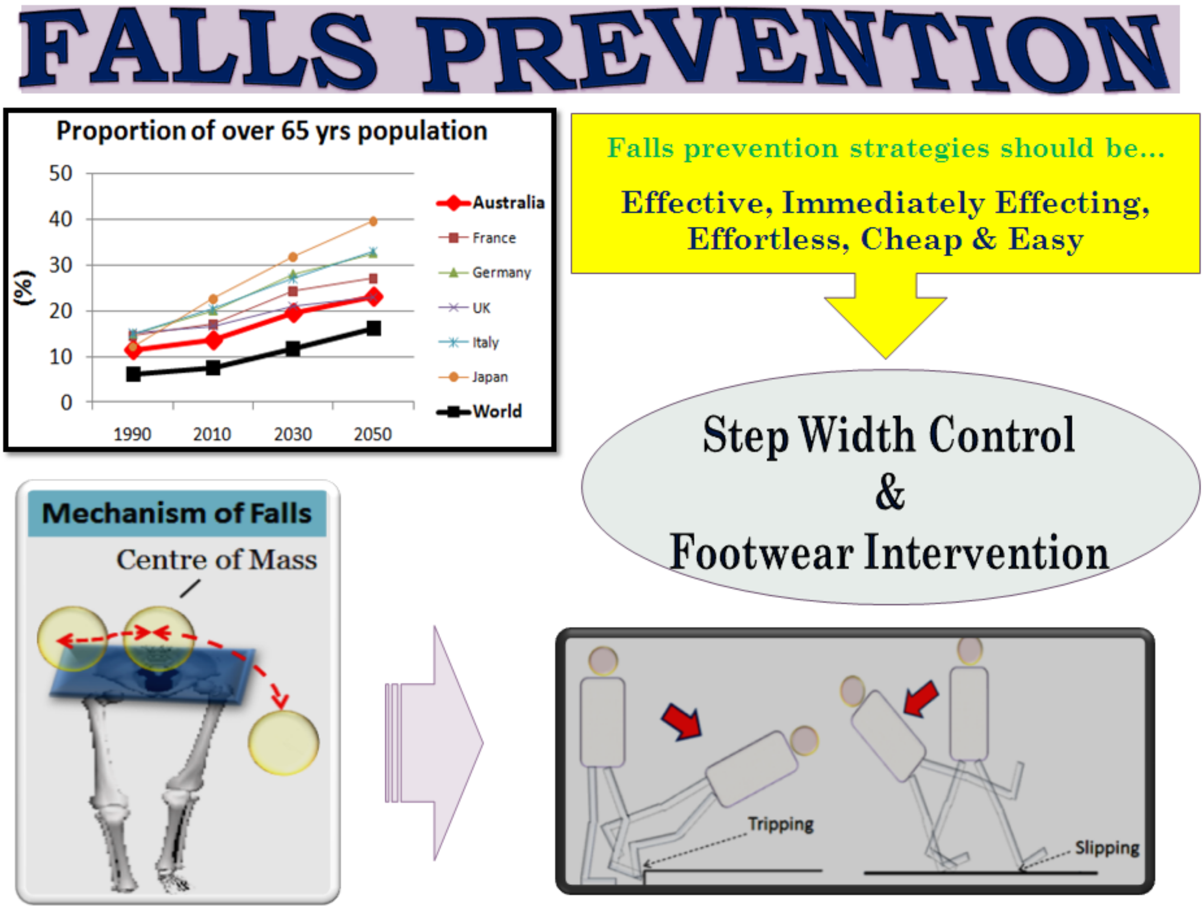


Figure 2: You = your data; Young = the young adults' data collected in honours study; Older = the older adults' data collected in honours study. (step width = StpW; dominant limb parameter = D-; non-dominant limb parameter = N-; step length = StpL; swing time = SwgT; double support time = DST)

If there are further questions, please feel free to contact me on 0399191128 or E-mail me on hanatsu.nagano@live.vu.edu.au

Thank you very much for the contribution.

Appendix VII **Poster for presentation**



Appendix VIII CANDIDATURE PROPOSAL: SUMMARY OF PROJECT

Title: Understanding Gait Control Dynamics: Ageing Effects on Falling Risks

Introduction:

Falls among the older (> 65 years) population is a major public heath issue due to its frequent occurrence and enormous associated cost (>3billion$ Pa in Australia; Australian Bureau of Statistics, 2006). It has been reported that at least 33% of older adults fall every year and more than 50% fell again in the subsequent year (Anderson, 2008, Hill et al., 1999; Keskin et al., 2008; Stalenhoef et al., 2002). Epidemiological studies have shown that the risk of falling increases with ageing and the environmental conditions in which they occur have been identified. It has been shown that in Australia 74% of falls were due to tripping, slipping and balance loss during locomotion (Berg et al., 1997). This trend is expected to further advance in a more rapid pace due to the ongoing ageing of the society in Australia and other developed countries (Moller et al., 2005). Research efforts are, therefore, urgently needed towards better understanding of the causes of falls and the establishment of falls prevention strategies.

As the largest number of falls has been reported during outdoor level walking (Lord et al., 2007), it is important to characterise the walking patterns that increase the risk of falls. Since the early studies undertaken by Winter (1991), gait has been commonly described by many quantitative parameters (e.g., time-distance, kinematics and kinetics) and often these parameters are compared between young and older populations to identify how ageing alters walking patterns that might be linked to increased risk of sustaining falls. Age-related gait pattern changes such as shorter step length, longer double support duration, or larger step width have been identified (Lord et al., 2007; Whittle, 2007), however these parameters do not provide the sensitivity needed and a more detailed analysis of locomotion characteristics is required to better understand the mechanisms of falling.

The increased falls risks in older adults may be associated with small changes in walking mechanics that cause slipping-, tripping- or balance loss-induced destabilisations. Tripping perturbations can be defined biomechanically as when the swing limb (usually the lowest part of the shoe or foot) forcefully contacts either the supporting surface or objects on it sufficient to destabilise the walker. Research undertaken at VU Biomechanics has shown how statistical properties of the minimum foot-ground clearance distribution (i.e., mean, median, skewness, kurtosis, and inter quartile range) contribute to the probability of foot contact with obstacles that could cause trip-related falls (Begg et al. 2007; Best and Begg, 2008). Slipping is due to unexpected loss of traction, again in such a way that the individual is significantly destabilized. Slipping can occur during either single support, when one foot is in contact with the surface or, less frequently, during double support when both feet are on the ground. Balance loss not due to tripping or slipping is characterized by significant displacement of the centre of mass relative to the base of support provided by the feet. In all three of these balance-disturbing behaviours falls appear to be due to older adults' musculoskeletal abilities being over-challenged and need to be investigated.

Besides advanced data analysis techniques, improvement in an experimental design is also essential in future gait studies to understand gait control techniques and the associated risk of falls while walking. For example, many of the past studies have examined only single limb biomechanics despite the fact that the two lower limb motions are asymmetrical (e.g., Nagano, 2008; Nagano et al., 2009; Sadeghi, 2000; Seeley et al., 2007) suggesting differing risk factors for the two limbs. Further investigation is required to explain the reason for gait asymmetry in young and ageing populations as gait asymmetry has been linked to increased falling incidences (Hill et al., 1999). It is quite obvious that the precise description of human gait is only possible through a 3D bilateral analysis of the whole body motion. Quantitative gait descriptions such as time-distance and toe clearance parameters are determined by foot/shoe mounted markers, but it is important to note that foot trajectory is the fine endpoint control task that involves contributions from a number of lower limb segments and joints (Winter, 1991). This proposed study aims to investigate bilateral whole body movement in 3 dimensions combined with advanced data analysis procedures to provide a better insight into the human gait control processes. The capacity to effectively control both kinematic and kinetic (force-related) dimensions of walking is fundamental to avoid tripping, slipping and balance loss. To evaluate the likelihood of falling due to tripping, gait kinematics will be measured by recording foot trajectories during the swing phase of the gait tasks using a 3D motion analysis system. To determine the risk of falling due to slipping the dynamic distribution of overground forces during the stance phase of the experimental tasks will be measured using force platforms. Finally, the risk of falling due to balance loss will be determined by using the 3D kinematic data to measure the position of the centre of mass relative to the supporting base when walking.

A further motivation of this current study is the use of shoe-insoles in the gait experiment to evaluate the effectiveness and possibility of the footwear intervention to minimise the risk of falls. Although various intervention strategies such as resistance and balance training, home-modification, or yoga has been proposed as possible falls prevention strategies (Lord et al., 2007; Yardley et al., 2008), none of these strategies has proved to be highly successful. For example, older adults were found to be reluctant to take up exercise intervention despite its effectiveness in preventing falls (Yardley et al., 2008). Considering the aspect of easy engagement, the ideal falls prevention strategies should provide effortless immediate effect that reasonably reduces the risk of falls (Mitty et al., 2007; Yardley et al., 2008). For this reason, footwear intervention can be another possible falls prevention strategy and it is a reasonable approach to control the very sensitive gait pattern where the change of only 1cm of toe clearance or 1˚ of heel-contact angle can determine whether a person will fall or not (Begg et al., 2007; Leclercq, 1999). In summary, there is a substantial literature to suggest that footwear intervention such as shoe-insoles may be helpful to modify the foot orientation, however their precise effect on specific musculoskeletal biomechanics needs to be quantified.

The aim of the project is to first determine how age-related changes to musculoskeletal abilities affect gait biomechanics and associated falls risks. In the first phase of the project the relationship between age-related changes in gait biomechanics will be determined in healthy older adults. In the second phase the effectiveness of footwear intervention will be tested to compensate ageing effects on neuromuscular abilities and, as a consequence, modify gait biomechanics to reduce the risk of falling. The proposed project will meet these requirements by addressing the following research questions.

(i) How do ageing and gender influence musculoskeletal biomechanics during level walking and when the balance is challenged? These gait biomechanics variables will be measured in gender-matched groups of young and older participants.

(ii) What is the association between ageing and the risk of tripping, slipping and balance loss when level walking and walking on restricted step width (medio-lateral balance)? To address this question these associations will be demonstrated as a function of age and gender.

(iii) How do footwear modifications such as shoe-insoles affect the risks of tripping, slipping, and balance loss in older adults? The effect of standard shoe and shoe-insoles will be determined by testing older participants wearing shoes with and without shoe-insoles proposed in the literature.

Explanation of Terminology for Basic Gait Parameters:

The fundamental description of human gait has been expressed in a variety of time-distance parameters including step length/time, stance time, swing time, double support time, and step width (Figure 1 top). Stance phase is determined as the period when the foot is in contact with the walking surface and this foot is referred as stance foot while swing phase is when the foot is traveling through the air to make a forward progression and this foot is referred as swing foot. Tripping and balance loss occur more frequently during swing phase (Lord et al., 2007) since only single limb effectively supports the bodily balance. Accordingly, swing toe trajectory is important to be investigated. Figure 1 (bottom) illustrates the major toe clearance events in normal walking.

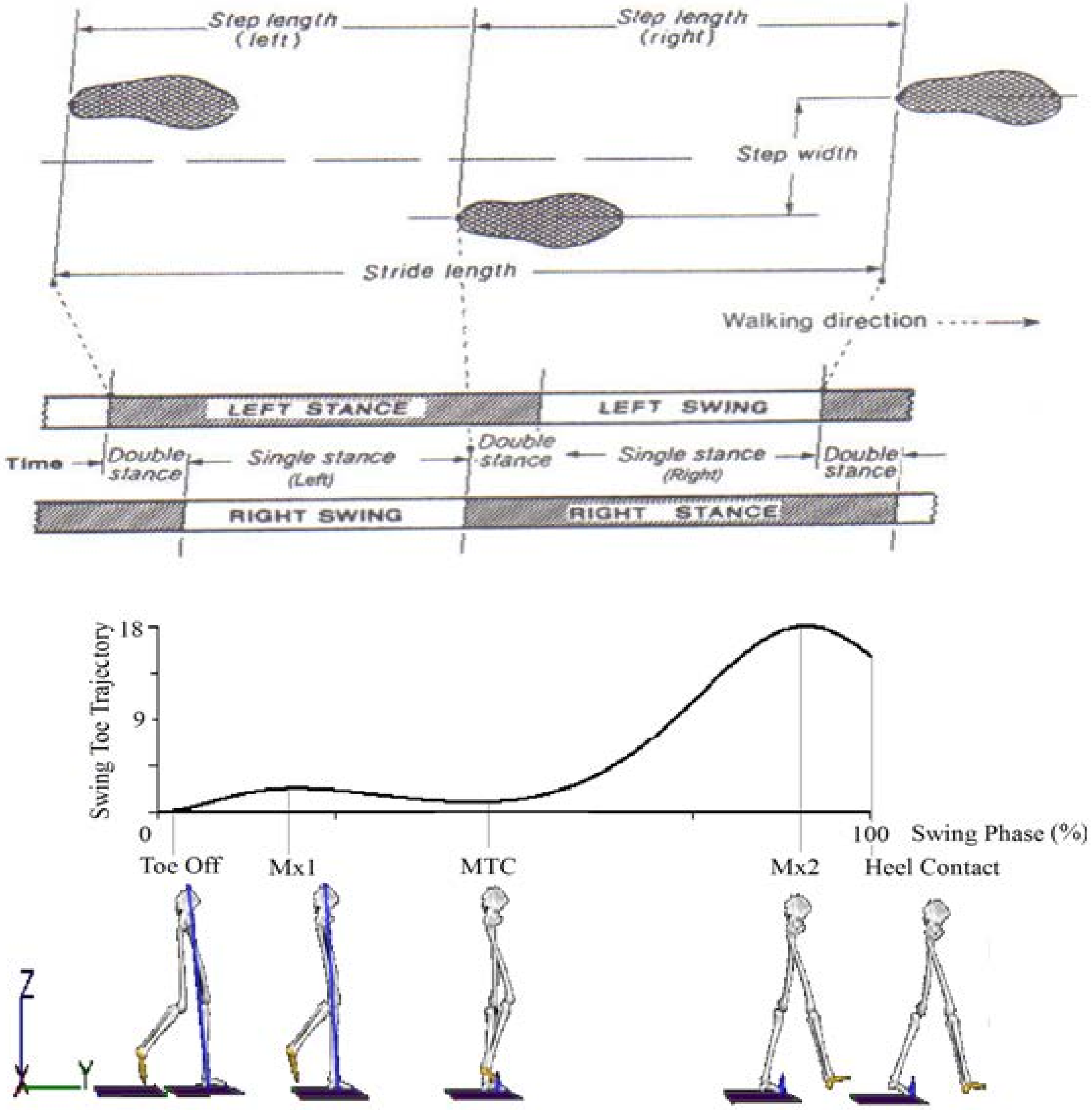


Figure 1: **Top: Description of time-distance parameters (adapted from Rose et al., 2006). Bottom: The swing toe clearance events in normal walking; swing foot indicated with the yellow colour.**

Contribution to knowledge:

The proposed gait study will be one of the most complete gait analyses. First, instead of single lower limb motion analysis, both the dominant and non-dominant lower limbs will be investigated, which will help to advance our understanding of the functional differences between the two lower limbs, extended from the notion of 'functional asymmetry' (Nagano et al., 2009; Sadeghi et al., 2000; Seeley et al., 2008). Second, advanced statistical and signal processing approaches (similar to Begg et al., 2007; Best and Begg 2008; mean, S.D., median, IQR, skewness, and kurtosis) will be examined for all the relevant gait parameters (time-distance, toe trajectory, joint kinematics, joint/muscle kinetics, foot-ground reaction forces, and centre of mass excursions). From an individual's minimum foot clearance distribution the probability of foot contact with an obstacle (i.e., tripping risk) can be determined (Best and Begg 2008). This statistical approach is the first to show strong potential for identification of individuals at risk of trip-related fall using foot trajectory data. These analysis techniques are powerful in providing the high sensitivity required to demonstrate the effect on falls risk of very small changes to gait biomechanics. Third, the proposed study will investigate the whole bodily motions and their linkages to gait control dynamics. Correlation analysis will be done to reveal how each joint motion and kinetics (moments, power) contribute to the motion of the foot and its control over the walking surface. Finally, this study will test the effectiveness of the falls-preventing footwear proposed in the literature through a detailed biomechanical analysis. Most previous studies used participant survey questionnaire to determine the effectiveness of shoes in reducing falls. As described in the previous section, foot motion characteristics during key events (e.g., mid-swing phase, heel contact event) and the body centre of mass motion in medio-lateral direction are the potential key biomechanical indicators of the risk of falls (Begg et al., 2007; Leclercq, 1999; Lee et al., 2006; Mills et al., 2008; Schrager et al., 2008). This study will show for the first time how effective the falls-preventing footwear can be in influencing these biomechanical variables and hence the risk of falls.

Significance of the study:

Successful completion of the project will lead to two major outcomes. First is the public health outcome of reducing costs. Increased falls in older adults and the associated mortality, morbidity, and disability is a major concern for Australian healthcare (Moller, 2005). Advancing age is associated with increased rates of bone fracture; of all fractures 50% occur at the hip and 25% of those individuals so affected die within the following year. Despite a substantial research effort to find effective interventions for reducing falls in older people there is no evidence at a national level that falls related hospitalisation rates have reduced over recent years (Bradley and Harrison 2007). The financial cost of falls has already been emphasised and it has been estimated that each 1% reduction in falls can save Australia $32million each year in direct medical costs (Gillespie et al. 2003). Establishment of falls prevention strategies will benefit many other countries as well (e.g. 1% reduction could save about A$3,000 million and 250,000 older adults in Japan; statistics taken from Japanese statistics bureau, 2008). A major outcome of this project will be the contribution to falls prevention through the testing and evaluation of 'safer shoes'. There is considerable potential for reducing falls simply by influencing choice of footwear, an intervention that is highly cost effective, easily monitored in terms of compliance, requires no mental or physical effort, and cannot harm or encumber the older person.

Timeline:

Period;

March 2009-February 2010 Literature Review/Candidature Proposal, Ethics, Subject recruitment, pilot test

March 2010 - February 2011 Gait testing, data analysis

March 2011 – August 2011 Re-testing older adults who have fallen, validation, advanced data analysis,

September 2011-February 2012 Thesis write up

Budget and Facilities:

Facilities and Equipment:

Available: Optotrak, Visual 3D, Force plates

Required	**Budget**
Standard shoes (6 x $250) and shoe-insoles (~$500)	$2,000
Markers ($30×20)	$600
Refreshments	$100
Advertisement in local newspaper	$500
Taxi vouchers (taxi $50×30; 2 lab visits)	$1,500
Total	**$4,500**

Literature Review:

Along with the development of medical science and decline of the birth rate, the proportion of older adults relative to the entire national population has been increasing, known as ‘ageing society.’ As this tendency progresses, there has emerged a variety of new social concerns related to the older population. The significance of falls among older adults is one of the largest healthcare issues for many developed countries due to its high frequency, injury rate, and cost. In addition, this tendency of 'ageing society' is predicted to further advance in the future (Moller et al., 2005). Therefore, the urgent establishment of preventive strategies is essential in the current society. As a number of falls incidence increases with age, early gait studies began with differentiating the walking pattern between young and older adults. The earlier gait studies (Whittle, 2007; Winter, 1991) reported that age-related physiological (e.g. loss of strength and reaction speed) and psychological changes (e.g. higher fear of falling) cause older adults to walk in different manners such as slower walking speed, larger step width, shorter step length (Perry et al., 2007; Whittle, 2007). However, more specific examinations on each type of falls were essential to establish strategies for falls prevention.

Types and causes of falls

Falls can occur in any directions and categorised separately, depending on its direct causes. The three main factors that lead people to fall have been identified as tripping, slipping, and sideway balance loss (Smeesters et al., 2001), which accounts for approximately up to 70 % of the entire falls issues in normal level walking (Figure 2).

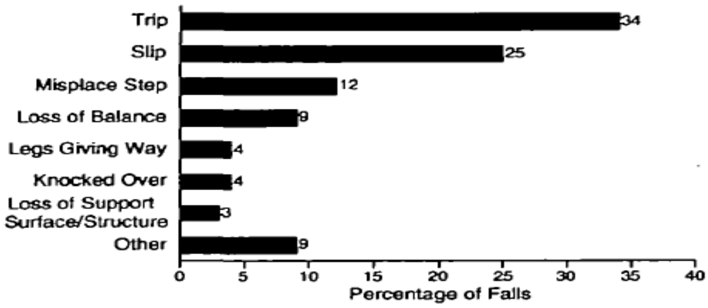


Figure 3 Causes of falls (adapted from Berg et al., 1997).

As briefly illustrated in the previous section 'Explanation of Terminology,' tripping only occurs during the swing phase (Lord et al., 2007), and vertical toe clearance was investigated by the past studies (Begg et al., 2007; Mills et al., 2008; Sparrow et al., 2008; Winter, 1991). These studies identified minimum toe clearance (MTC) as the most risky point where the largest number of falls occurrence can be attributed because of the low toe height (1-2 cm), maximum forward velocity of the foot segment (4.6 m/s; Winter, 1991), and the sagittally aligned position of the two feet (Winter, 1991). Following to tripping, slipping is the second largest cause of falls, which occupies about a quarter of the total falls occurrence and often causes hip fracture as a consequence (Lecercq, 1999). Initiation of slipping depends mostly on the heel dynamics between a shoe and the walking surface

(Leclerzq, 1999). Sideway falls due to the balance loss occurs when the body fails to maintain COM within the range of medio-lateral supporting base (Wilson et al., 2006). Older adults seem to adjust their step width to be larger to minimise the sideway falls (Whittle, 2007), but it still seems ineffective (Schrager et al., 2008) as a number of falls is much larger in the aged population. Therefore, other strategies to stabilize COM in medio-lateral direction are necessary.

Future directions of gait studies and falls prevention strategies

One of the future directions of falls prevention studies can accordingly be extended from the further inspection of these specific biomechanical viewpoints and their relations with falls occurrence, which should be based on more complete and accurate examination of human locomotion. For example, past studies (Nagano, 2008; Nagano et al., 2009; Sadeghi, 2000; Seeley et al., 2008) found the evidence of gait asymmetry, suggesting bi-lateral limb motion analysis should be conducted for the precise expression of human gait. It is also important to look at the whole body motion to fully explain how each joint motion is coordinated to determine the control of the walking task rather than looking at a few joint segments. As a summary, the whole body motion should be monitored bilaterally in 3D coordinate system for the future gait studies, which will be also necessary to develop the effective and complete falls prevention strategy.

Unfortunately, none of the previously proposed falls prevention strategies can be considered as completely successful. For example, balance training and resistance training have been reported to effectively reduce the risk of falls (Sturnieks et al., 2009; Sherrington et al., 2004), but Yardley et al. (2008) found that nearly 60% of older adults would not voluntarily participate in group exercise sessions for the purpose of falls prevention mainly due to the lack of motivation and excessive cost. Other passive strategies such as simple home modification has been found effective and financially affordable (Salkeld et al., 2000), but the effect is only limited within the inside of the house and does not encourage older adults to walk outside. Considering these aspects together, falls prevention strategies should be effective, easy to take up, and help older adults walk outside rather than fulfilling a few of these requirements.

Footwear intervention to minimise the risk of falls

For these reasons, footwear intervention could be the ideal strategy to reduce the risk of falls. Menant et al. (2008; 2009) conducted the experiments to examine several shoe features including sole hardness, heel height and shape, and tread pattern in terms of effects on human gait pattern. By changing only single shoe feature, many gait parameters including walking velocity, step length, step width, double support time, MTC, and heel contact velocity were differentiated, confirming that the footwear can change the walking pattern.

Footwear is not only limited to shoes, but shoe-insoles can be another possible footwear intervention to reduce the risk of falls as foot is in the direct contact with shoe-insoles. Compared to shoes as footwear intervention, use of insoles mainly has the two advantages: cheaper manufacturing cost and wide applicability into different shoes. Maki et al. (2008) proved the effectiveness of ‘sole sensors’ as the falls prevention shoe insoles. The fundamental mechanism of ‘sole sensors’ is to provide more tactile sensation on the bottom of feet, and enhance sensory feedback system. There have been other past studies about shoe insoles, looking at the pressure distribution (Bus et al., 2004) and knee adduction moment reduction (Nakajima et al., 2009). In conclusion, it will be interesting if the understanding of human gait is deepened and utilised in the footwear intervention to reduce the number of falls incidence.

Research Design:

The proposed project will employ groups of young and older adults, and detailed gait experiments to address the research questions and aims of the project.

Participants and Exclusion Criteria:

Two groups of participants will be recruited; 3 size will be maximized but it is expected that there will be at least 30 young healthy adults (age range: 20 years to 35 years) and 30 healthy elderly individuals (age range of 65 to 85 years). There will be equal number of male and female participants in each age group. The sample size for the proposed project has been calculated based on results from our recent work involving young and elderly individuals and minimum toe clearance data (Begg, Best et al. 2007). The young adults will be recruited from the VU academic community. The older group will be recruited from the VU Biomechanics subject pool and through advertisement in the local newspaper. Questionnaires for screening purposes have been developed and are used routinely in gait studies (Hill et al., 2004). Exclusion criteria include; (i) history of serious injuries and/or the falls within the past two years (as the falls history affects the walking performance; Lord et al., 2007; Whittle, 2007), (ii) visual acuity (< 6/12; Nicklason et al., 2002) and/or contrast (Melbourne edge test <6/15; Lord et al., 2001), (iii) Time Up&Go test (13.5 secs <; van Iersel et al., 2008). Height, weight, shoe size, and the limb dominance (defined as the limb used to kick a ball more comfortably Sadeghi et al., 2003; Seeley et al., 2008) will be used. Participants will be asked to complete informed consent procedures consistent with the VU Human Research Ethics Committee guidelines.

Experimental Setting;

- Each walking test will be conducted in the two footwear conditions: standard shoes (defined by Menant et al., 2008) with and without falls-preventing shoe-insoles.
- Imaginary markers registration is described in Figure 4 (similar landmark definitions; Labbe et

al., 2008; Matsas et al., 2000; Protopapadaki et al., 2006; Wass et al., 2004) and upper body landmarks include wrist, elbow, shoulder, and head. Optotrak motion capture system (NDI, Canada; 240Hz) will be used to track these registered real and imaginary markers.

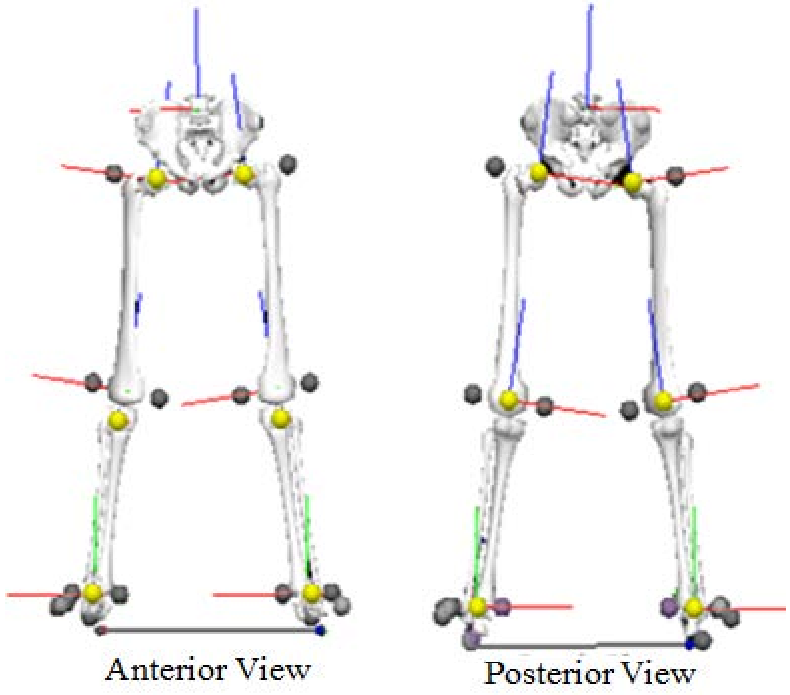


Figure 4: **Image of imarinary markers registration. Hip (lateral surface on both anterior and posterior iliac spine), knee (medial & lateral knee and patella), and foot (medial & lateral ankles, 2nd & 5th metatarsal phalanges, minimum toe point, and heel). Yellow markers indicate the joint centres and gray markers indicate registered imaginary markers.**

- Visual 3D software (C motion, U.S.A) will be used to extract biomechanical variables of interest such as swing phase time-distance & toe clearance gait parameters, joint angular changes, moments, power.

Testing Procedure;

- In each walking condition, participants will be asked to walk on a 10m walkway repeatedly for a minimum of 40 times with the different footwear conditions. Two AMTI force platforms will be located on the walkway to record synchronized foot-ground reaction forces during walking.
- Testing conditions will include;
  (i) Normal walking at preferred walking speed;
  (ii) Walking with restricted step widths: Walking along two lines indicative of narrow (5 cm) and wide (30 cm) base of support to determine the effect of step-width on walking pattern (Schrager et al., 2008).

Data Analysis;

Advanced statistical methods and modelling procedure for non-normal distributions of foot ground clearance data will be applied to determine the precise probability of tripping (Best and Begg 2008). These will also be adapted to ground-reaction force and centre of mass displacement data. These statistical methods are highly advantageous to the proposed project in providing considerably more detailed information concerning the effects of gait biomechanics on falls risks. From an individual's minimum foot clearance distribution the probability of foot contact with an obstacle (i.e. tripping risk) can be determined (Best and Begg 2008). This statistical approach is the first to show strong potential for identification of individuals at risk of trip-related fall using foot trajectory data. These analysis techniques are powerful in providing the high sensitivity required to demonstrate the effect on falls risk of very small changes to gait biomechanics.

- Major dependent variables will include time-distance parameters (walking velocity, cadence, step length, step width, swing time, and double support time), toe clearance parameters (Mx1, MTC, and Mx2), foot-ground reaction data, joint kinematics/kinetics, and Centre of mass displacement, described by the following statistics: mean/median, S.D./IQR, and skewness/kurtosis.
- A 2 x 2 x 2 x 3 (age x limb x footwear x condition) repeated measures mixed analysis of variance (ANOVA) design will be applied to all the examined variables to determine main effects and any interactions (significant p-value = .05) for each experimental conditions. Equivalent non-parametric tests will be undertaken for non-normal distributions.

Approaches to research questions;

<u>*Research Question 1: How does ageing influence the walking patterns?*</u>

All the age effects and interactions with limb or condition effects on gait parameters and tripping, slipping and balance loss will be determined and reported.

<u>*Research Question 2: Does limb dominance influence walking pattern?*</u>

All the limb effects on gait parameters and interactions with age, footwear, or condition effects will be reported.

<u>*Research Question 3: If there are differences identified [from Research Question 1 & 2], where do these differences come from and how are they affected?*</u>

Correlation analysis will be undertaken (using both linear and non-linear approaches) between the key dependent variables (e.g., Toe clearance, foot contact velocity) and the various joint kinematics/kinetics to evaluate how various joints/segments/muscles gait control.

<u>*Research Question 4: How does 'restriction of step width' influence medio-lateral balance?*</u>

Effects of step width changes on COM displacement characteristics will be determined to find out how step width changes influence medio-lateral balance.

<u>*Research Question 5: How effective is the falls preventing footwear?*</u>

Falls risk assessment (tripping, slipping and sideway balance loss) will be undertaken both *with* and *without* footwear conditions (i.e., standard shoes with and without the shoe-insoles).

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Appendix IX Computation for ART due to Forefoot Posterior Slipping

$$\frac{T(y) - V(cy) \times t - 0.5 \times A(cy) \times t^2}{T(x) - V(cx) \times t - 0.5 \times A(cx) \times t^2} = \frac{V(cy) \times t + 0.5 \times A(cy) \times t^2 - S(y) - 0.5 \times A(fy) \times t^2}{V(cx) \times t + 0.5 \times A(cx) \times t^2 - S(x) - 0.5 \times A(fx) \times t^2}$$

Where T(y) = stance toe AP coordinate, V(cy) = CoM AP velocity, A(cy) = CoM AP acceleration, T(x) = stance toe ML coordinate, V(cx) = CoM ML velocity, A(cx) = CoM ML acceleration, S(y) = a slipping foot's AP coordinate, A(fy) = a slipping foot's AP acceleration, S(x) = a slipping foot's ML coordinate, A(fx) = a slipping foot's ML acceleration

*all coordinates are at anterior balance loss due to forefoot posterior slipping

*for the following computations, T(y) = A, V(cy) = B, A(cy) = C, T(x) = D, V(cx) = E, A(cx) = F, S(y) = G, A(fy) = H, S(x) = I, A(fx) = J

$$\boldsymbol{\frac{A - Bt - 0.5Ct^2}{D - Et - 0.5Ft^2} = \frac{Bt + 0.5Ct^2 - G - 0.5Ht^2}{Et + 0.5Ft^2 - I - 0.5Jt^2}}$$

$$\boldsymbol{(CJ - FH)t^4 + (2BJ - 2EH)t^3 + (2A(F - J) + 2CI - 2D(C - H) - 2FG)t^2 + 4(AE + BI - EG - BD)t - 4AI + 4DG = 0}$$

$$\boldsymbol{at^4 + bt^3 + ct^2 + dt + e = 0}$$

*for the following computations, a = CJ-FH, b = 2BJ-2EH, c = 2AF-2AJ, d = 4(AE+BI-EJ-BD), e = -4AI+4DG

$$\boldsymbol{t = \frac{\pm 2u_n\sqrt{u_n} \pm \sqrt{\left(2u_n\sqrt{u_n}\right)^2 - 8u_n\left(u_n\left(-\frac{3b^3}{8a^3} + \frac{c}{a}\right) + u_n^2 \pm \sqrt{u_n}\left(\frac{b^3}{8a^3} - \frac{bc}{2a^2} + \frac{d}{a}\right)\right)}}{4u_n} - \frac{b}{4a}}$$

$$\boldsymbol{where\ u_n = -\frac{2}{3}\left(-\frac{3b^3}{8a^3} + \frac{c}{a}\right) + \sqrt[3]{\alpha_+} + \sqrt[3]{\alpha_-}, -\frac{2}{3}\left(-\frac{3b^3}{8a^3} + \frac{c}{a}\right) + \omega\sqrt[3]{\alpha_+} + \omega^2\sqrt[3]{\alpha_-}, -\frac{2}{3}\left(-\frac{3b^3}{8a^3} + \frac{c}{a}\right) + \omega^2\sqrt[3]{\alpha_+} + \omega\sqrt[3]{\alpha_-}}$$

$$\boldsymbol{\alpha_+ = \frac{-\eta 1 + \sqrt{\eta 1^2 - 4\eta 0}}{2}, \alpha_- = \frac{-\eta 1 - \sqrt{\eta 1^2 - 4\eta 0}}{2}, \omega = -\frac{1}{2} + \frac{\sqrt{3}i}{2}, \omega^2 = -\frac{1}{2} - \frac{\sqrt{3}i}{2}}$$

$$\eta 1=\frac{16\left(-\frac{3b^3}{8a^3}+\frac{\mathrm{c}}{\mathrm{a}}\right)^3}{27}-\frac{2\left(-\frac{3b^3}{8a^3}+\frac{\mathrm{c}}{\mathrm{a}}\right)\left(\left(-\frac{3b^3}{8a^3}+\frac{\mathrm{c}}{\mathrm{a}}\right)^2-4\left(-\frac{3b^4}{256a^4}+\frac{b^2c}{16a^3}-\frac{bd}{4a^2}+\frac{e}{a}\right)\right)}{3}-\left(\frac{b^3}{8a^3}-\frac{bc}{2a^2}+\frac{d}{a}\right)^2$$

$$\eta 0=-\frac{1}{27}\left(-\frac{4\left(-\frac{3b^3}{8a^3}+\frac{\mathrm{c}}{\mathrm{a}}\right)^2}{3}+\left(-\frac{3b^3}{8a^3}+\frac{\mathrm{c}}{\mathrm{a}}\right)^2-4\left(-\frac{3b^4}{256a^4}+\frac{b^2c}{16a^3}-\frac{bd}{4a^2}+\frac{e}{a}\right)\right)^3$$

$$at^4+bt^3+ct^2+dt+e=t^4+\frac{b}{a}t^3+\frac{c}{a}t^2+\frac{d}{a}t+\frac{e}{a}=t^4+A3t^3+A2t^2+A1t+A0=0$$

$$(A3=\frac{b}{a},A2=\frac{\mathrm{c}}{\mathrm{a}},A1=\frac{d}{a},A0=\frac{e}{a})$$

$$L=t+\frac{A3}{4}$$

$$\begin{aligned}t^4+\frac{b}{a}t^3+\frac{c}{a}t^2+\frac{d}{a}t+\frac{e}{a}&\\ &=L^4+\left(-\frac{3}{8}A3^3+A2\right)L^2+\left(\frac{1}{8}A3^3-\frac{1}{2}A3A2+A1\right)L-\frac{3}{256}A3^4+\frac{1}{16}A3^2A2\\ &-\frac{1}{4}A3A1+A0=L^4+B2L^2+B1L+B0=0\end{aligned}$$

$$\begin{aligned}(B2=-\frac{3}{8}A3^3+A2,B1=\frac{1}{8}A3^3-\frac{1}{2}A3A2+A1,B0&\\ =-\frac{3}{256}A3^4+\frac{1}{16}A3^2A2-\frac{1}{4}A3A1+A0)\end{aligned}$$

$$L^4+B2L^2+B1L+B0=0$$

$$L^4+B2L^2=-B1L-B0$$

$$L^4+B2L^2+uL^2=-B1L-B0+uL^2$$

$$(L^2+\frac{1}{2}(B2+u))^2-\frac{1}{4}(B2+u)^2=u\left(L-\frac{B1}{2u}\right)^2-\frac{1}{4u}B1^2-B0$$

$$(L^2+\frac{1}{2}(B2+u))^2=u\left(L-\frac{B1}{2u}\right)^2+\frac{1}{4}(B2+u)^2-\frac{1}{4u}B1^2-B0=u\left(L-\frac{B1}{2u}\right)^2+F(u)$$

$$\boldsymbol{F(u) = \frac{1}{4}(B2+u)^2 - \frac{1}{4u}B1^2 - B0}$$

$$\boldsymbol{F(u) = 0}$$

$$\boldsymbol{F(u) = \frac{1}{4}(B2+u)^2 - \frac{1}{4u}B1^2 - B0 = 0}$$

$$\boldsymbol{u^3 + 2B2u^2 + \left(B2^2 - 4B0\right)u - B1^2 = 0}$$

$$\boldsymbol{u^3 + 2B2u^2 + \left(B2^2 - 4B0\right)u - B1^2 = u^3 + U2u^2 + U1u + U0 = 0}$$

$$\boldsymbol{(U2 = 2B2, U1 = B2^2 - 4B0, U0 = -B1^2)}$$

$$\boldsymbol{v = u + \frac{U2}{3}}$$

$$\boldsymbol{u = v - \frac{U2}{3}}$$

$$\boldsymbol{u^3 + U2u^2 + U1u + U0 = v^3 + \left(-\frac{U2^2}{3} + U1\right)v + \frac{2U2^3}{27} - \frac{U2U1}{3} + U0 = v^3 + V1v + V0 = 0}$$

$$\boldsymbol{(V1 = -\frac{U2^2}{3} + U1, V0 = \frac{2U2^3}{27} - \frac{U2U1}{3} + U0)}$$

$$\boldsymbol{m + n = v, mn = -\frac{V1}{3}}$$

$$\boldsymbol{m^3 + n^3 = (m+n)^3 - 3mn(m+n) = -V0}$$

$$\boldsymbol{m^3n^3 = \left(-\frac{V1}{3}\right)^3}$$

$$\boldsymbol{\left(\alpha - m^3\right)\left(\alpha - n^3\right) = \alpha^2 - (m^3+n^3)\alpha + m^3n^3 = \alpha^2 + V0\alpha + \left(-\frac{V1}{3}\right)^3 = \alpha^2 + \eta1\alpha + \eta0 = 0}$$

$$\boldsymbol{\eta1 = Vo, \eta0 = \left(-\frac{V1}{3}\right)^3}$$

$$\boldsymbol{\alpha = \frac{-\eta1 \pm \sqrt{\eta1^2 - 4\eta0}}{2}}$$

$$\boldsymbol{m^3 = \frac{-\eta1 + \sqrt{\eta1^2 - 4\eta0}}{2}(= \alpha_+), n^3 = \frac{-\eta1 - \sqrt{\eta1^2 - 4\eta0}}{2}(= \alpha_-)}$$

$$m = \sqrt[3]{\alpha_+}, \omega\sqrt[3]{\alpha_+}, \omega^2\sqrt[3]{\alpha_+}$$

$$n = \sqrt[3]{\alpha_-}, \omega\sqrt[3]{\alpha_-}, \omega^2\sqrt[3]{\alpha_-}$$

$$\omega = -\frac{1}{2} + \frac{\sqrt{3}i}{2}, \omega^2 = -\frac{1}{2} - \frac{\sqrt{3}i}{2}$$

$$mn = -\frac{V1}{3}$$

$$(m, n) = \left(\sqrt[3]{\alpha_+}, \sqrt[3]{\alpha_-}\right), \left(\omega\sqrt[3]{\alpha_+}, \omega^2\sqrt[3]{\alpha_-}\right), \left(\omega^2\sqrt[3]{\alpha_+}, \omega\sqrt[3]{\alpha_-}\right)$$

$$v = m + n$$

$$v = \sqrt[3]{\alpha_+} + \sqrt[3]{\alpha_-}, \omega\sqrt[3]{\alpha_+} + \omega^2\sqrt[3]{\alpha_-}, \omega^2\sqrt[3]{\alpha_+} + \omega\sqrt[3]{\alpha_-}$$

$$u = v - \frac{U2}{3}$$

$$u = -\frac{U2}{3} + \sqrt[3]{\alpha_+} + \sqrt[3]{\alpha_-}, -\frac{U2}{3} + \omega\sqrt[3]{\alpha_+} + \omega^2\sqrt[3]{\alpha_-}, -\frac{U2}{3} + \omega^2\sqrt[3]{\alpha_+} + \omega\sqrt[3]{\alpha_-}$$

$$u = u1, u2, u3$$

$$u_n = u1, u2, u3$$

$$\left(L^2 + \frac{1}{2}(B2 + u_n)\right)^2 = u_n\left(L - \frac{B1}{2u_n}\right)^2 + F(u_n) = u_n\left(L - \frac{B1}{2u_n}\right)^2$$

$$\because F(u_n) = 0$$

$$L^2 + \frac{1}{2}(B2 + u_n) = \pm\sqrt{u_n}\left(L - \frac{B1}{2u_n}\right)$$

$$2u_nL^2 \mp 2u_n\sqrt{u_n}L + B2u_n + u_n^2 \pm \sqrt{u_n}B1 = 0$$

$$2u_nL^2 \mp 2u_n\sqrt{u_n}L + B2u_n + u_n^2 \pm \sqrt{u_n}B1 = C2L^2 + C1L + C0 = 0$$

$$C2 = 2u_n, C1 = \mp 2u_n\sqrt{u_n}, C0 = B2u_n + u_n^2 \pm \sqrt{u_n}B1$$

$$L = \frac{-C1 \pm \sqrt{C1^2 - 4C2C0}}{2C2}$$

$$L = L1, L2, L3, L4$$

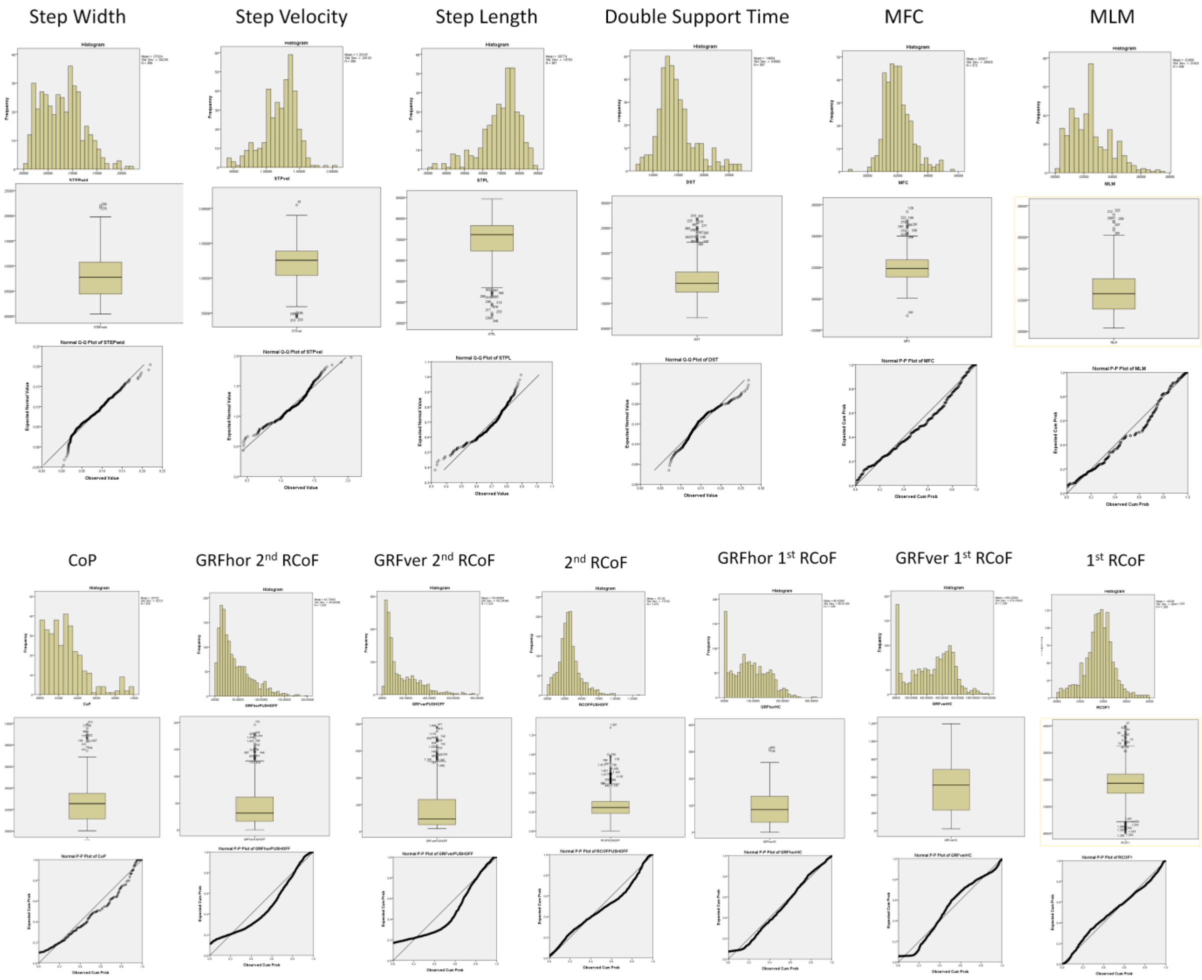
$$L = t + \frac{A3}{4}$$

$$\boldsymbol{t = L - \frac{A3}{4}}$$

$$\boldsymbol{t = L1 - \frac{A3}{4}, L2 - \frac{A3}{4}, L3 - \frac{A3}{4}, L4 - \frac{A3}{4}}$$

Appendix X Normality of the experimentally obtained fundamental position-time & GRF data.

Normality of the experimentally obtained fundamental position-time & GRF data was confirmed using histogram descriptions, Q-Q plot and the stem-and-leaf plot.



Appendix XI Publications

Publications based on the current thesis

Conference **-**Refereed full papers

Nagano, H., Sparrow, W.A., and Begg, R. 2012. Ageing effects on the mechanical energy cost of walking. In proceedings of the *World Congress 2012: medical physics and biomedical engineering.* Beijing, China, 26-31 May 2012. 168-171.

Nagano, H., Begg, R., Sparrow, W.A. 2013. Computation method for available response time dur to tripping at minimum foot clearance. In proceedings of the *IEEE Engineering in Medicine and Biology Society*. Osaka, Japan, 3-7 July, 2013.

Nagano, H., Begg, R., Sparrow, W.A. 2013. Ageing effects on medio-lateral balance during walking with increased and decreased step width. In proceedings of the *IEEE Engineering in Medicine and Biology Society*. Osaka, Japan, 3-7 July, 2013.

Other publications during the PhD period

Journals

Nagano, H., Sparrow, W.A., and Begg, R.K. 2013. Biomechanical characteristics of slipping during unconstrained walking, turning, gait initiation and termination. *Ergonomics*. ***In press***

Nagano, H., Begg, R., Sparrow, W., and Taylor, S. 2012. A comparison of treadmill and overground walking effects on step cycle variability and asymmetry in young and older individuals. *Journal of Applied Biomechanics*. ***In press***

Nagano, H., Begg, R., Sparrow, W., and Taylor, S. 2011. Ageing and limb dominance effects on foot-ground clearance during treadmill and overground walking. *Clinical Biomechanics*, 26(9): 962-968.

Conference **-**Refereed full papers

Nagano, H., Begg, R., and Sparrow, W.A. 2010. Controlling swing foot centre of mass and toe trajectory to minimise tripping risk. In proceedings of the *32nd International Conference of the IEEE Engineering in Medicine and Biology Society (EMBS), IEEE*. Buenos Ires, Argentina, 31 August-4 September 2010. 4854-4857.

Nagano, H., Begg, R., and Sparrow, W.A. 2010. Modeling foot trajectory control during walking. In proceedings of the *International Conference on Modelling and Simulation 2010* (MS'10). Prague, Czech Republic. 22-25 June 2010. 313-316.