# Age-related changes in the control of mediolateral dynamic stability during volitional and reactive stepping

by

Jonathan Craig Singer

A thesis

presented to the University of Waterloo

in fulfilment of the

thesis requirement for the degree of

Doctor of Philosophy

in

Kinesiology

Waterloo, Ontario, Canada, 2012

## AUTHOR'S DECLARATION

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

**Jonathan C. Singer**

# ABSTRACT

The high incidence of falls and fall-related injuries among Canadians over the age of 65 continues to be a key public health issue. As the current proportion of individuals within this cohort of the population is predicted to double by the year 2031, the absolute number of individuals experiencing falls, fall-related injuries and subsequent hospitalization will increase dramatically. While a fall in any direction can lead to injury and reduced quality of life, lateral falls have been shown to be prevalent and can be particularly devastating because of the increased probability of hip fracture. Forward stepping tasks, whether initiated volitionally or by external perturbation, pose a challenge to stability, as they require the precise regulation of the spatial and temporal characteristics of the whole body centre of mass (COM) in relation to a changing base of support (BOS). Despite our understanding of both proactive and reactive mechanisms for balance control at movement initiation during such stepping tasks, there appears to be little understanding or consensus regarding the origins of age-related decline in mediolateral stability, which can manifest during the restabilisation phase, at movement termination. From this, the global objective of this thesis was to develop further understanding regarding such age-related differences in mediolateral dynamic stability control during the restabilisation phase of forward stepping. Notwithstanding the well documented differences between volitional and perturbation-evoked stepping until the time of foot-contact, we have proposed the control of the COM during the restabilisation phase of such stepping tasks to be a central determinant of age-related differences in mediolateral dynamic stability, common to both forms of stepping. We quantified the COM kinematics during the restabilisation phase and calculated the magnitude of incongruity between the peak and

final, stable, COM position, in addition to the intertrial variability of this incongruity. Further, we analysed the orientation of the net ground reaction force (GRF) with respect to the COM, which allowed us to draw conclusions regarding the mechanisms that may be responsible for the age-related differences in the COM kinematics. To vary the challenge to control, we included conditions in which individuals were required to step with altered step width. In addition, we attempted to probe the extent and means by which individuals could alter the dynamics of stepping over time, with trial repetition. In general, we found that overshoots of the final COM position were common to all forms of stepping and may serve the functional role of simplifying reactive control during the restabilisation phase. The magnitude and intertrial variability of incongruity, however, were greater among the older adults during all forms of stepping. We believe such increased COM incongruity is likely indicative of greater instability within this group, which may be associated with the increased time required to reorient the net GRF in a manner necessary to oppose the total body angular momentum that developed during the swing phase. Particularly interesting was the use of proactive strategies by older adults, which may have the potential to offset instability that arises due to difficulty with reactive control during the restabilisation phase. The present work provides support for previous studies, which have suggested that the control of mediolateral stability may be particularly challenging for older adults. Further, our work provides evidence that the challenges associated with mediolateral stability control have important links to the restabilisation phase and are common to both volitional and reactive stepping. This work highlights the need to further explore the control of mediolateral stability and develop therapeutic interventions to reduce such incidence of instability among older adults.

# ACKNOWLEDGEMENTS

Although it is customary for the student's name to stand alone on the title page of the thesis, this work could not have been completed without the assistance and support of many people. Firstly, I would like to thank my supervisors, Dr. Steve Prentice and Dr. Bill McIlroy, for their patience in allowing me to complete this work and for their vocal belief in my ability at times when I was least confident. I am very fortunate to have had the opportunity to work with Dr. Prentice, who offered me a considerable amount of latitude to forge my own path, yet was always available to offer solutions when I ran into road blocks. I am grateful to have had the privilege to work with Dr. McIlroy, who has established an exceptional research training environment in which I was exposed to new ways to approach research questions, which continually challenged me to improve my scientific thinking and reasoning.

I would like to thank my examining committee, Drs. Andrew Laing and John McPhee for their insight and thought provoking questions. Dr. Stacey Acker was very kind in her willingness to serve as a proxy examiner, but went further to read the entire document and provide her own recommendations, despite the extremely limited timeframe. I owe many thanks to my external examiner, Dr. Joyce Fung, for her unique perspective and her suggestions to strengthen this work and subsequent manuscripts.

I am also very fortunate for the interactions I have had with Drs. Jack Callaghan, Jim Frank, Aftab Patla and Nancy Theberge throughout the course of my degree. All of these individuals were always available to offer advice and support at times when it was needed most. I would like to express my utmost appreciation to Ruth Gooding, who consistently went above and beyond her responsibilities as Graduate Coordinator and expressed genuine concern for the wellbeing of each student within the department. I am very thankful for the expert technical and computer support provided by Wendell Prime, Jeff Rice and Craig McDonald.

Many thanks are extended to Rupesh Patel and Kira Pattison for their assistance with data collection. Despite the occasional monotony of laboratory work, both of these individuals demonstrated a high degree of professionalism when working with research participants. I am indebted to all the participants who donated their time to take part in the studies within this thesis. I am especially grateful to the individuals within my sample of older adults. Many of these participants came to the lab because they believed they would learn about balance control. It was I, however, who left the lab with greater knowledge, as these individuals shared their wisdom, experience and perspective on life.

I have met a number of remarkable graduate students during the time I have spent at UW, too numerous to list individually. I thank you all for your friendship. I will attempt to express my gratitude to each of you in person, rather than on paper.

## DEDICATION

This thesis is dedicated to my parents, for their quiet encouragement and wisdom; to my brother, for leading by example; and to Natasha, for being here during the many highs and lows, for always believing in me and for her personal and professional sacrifices, which have allowed me to pursue my academic goals.

# TABLE OF CONTENTS

# LIST OF FIGURES

# LIST OF TABLES

# CHAPTER 1

# Introduction

## 1.1 GENERAL OVERVIEW

The ability to habitually stand and walk upright on two legs is characteristic of the human lineage (Vaughan, 2003; Sockol, Raichlen & Pontzer, 2007). This distinction, however, introduces inherent instability, as two thirds of the body mass is located two thirds of the body height above the ground (Winter, Patla, Frank & Walt, 1990). As a result, the control of upright stability is non-trivial, requiring continuous muscle activation and modulation to compensate for both internal and external perturbations. To further complicate this control, intersegmental coupling allows inertial forces to be transmitted across segments, such that a muscle can accelerate segments far removed from the joint or joints it spans (Zajac, 1993). Despite this, while balance control does require attentional resources (Lajoie, Teasdale, Bard & Fleury, 1993; Shumway-Cook & Woollacott, 2000; Woollacott & Shumway-Cook, 2002), upright stability can be maintained in healthy systems with little conscious effort. Problems do, however, arise with age or various medical conditions, whereby the ability to maintain dynamic stability is challenged and the risk of falls increases (Tinetti, Speechley & Ginter, 1988; Nevitt, Cummings, Kidd & Black, 1989; O'Loughlin, Robitaille, Boivin & Suissa, 1993; McIlroy & Maki, 1996; Maki & McIlroy, 2006). Such challenges to stability control in these special populations have prompted significant scientific investigation.

Dynamic tasks, such as gait initiation or termination, rapid limb withdrawal, or reactive stepping in response to external perturbations, pose a challenge to stability because they require the precise regulation of the spatial and temporal characteristics of the centre of mass (COM) in relation to a changing base of support (BOS). A particularly interesting feature of both volitional and perturbation-evoked forward stepping is the development of mediolateral instability, which is orthogonal to the intended or induced direction of motion. Such instability is not directly related to the internal or external perturbation itself, but arises as an indirect consequence of the shift from bipedal to unipedal stance. As a result of the change in BOS configuration during the initial aspect of the swing phase, the line of action of the gravitational force acting at the COM falls lateral to the ankle joint of the stance limb, resulting in an external moment that tends to accelerate the COM away from the stance limb. In volitional movements, such mediolateral instability is pre-emptively countered, to a varying extent, by mediolateral anticipatory postural adjustments (APA), which serve to accelerate the COM toward the stance limb before the initiation of swing phase, thereby reducing the external gravitational moment and subsequent acceleration toward the contralateral side (Fig. 1-1a). In compensatory stepping reactions, however, this APA is often absent or insufficient to have a marked effect on the COM kinematics (McIlroy & Maki, 1993b; McIlroy & Maki, 1999; Rogers et al., 2001), which leads to an increased external gravitational moment in the frontal plane at the onset of stepping (Fig. 1-1b) and may complicate subsequent mediolateral dynamic stability control during the restabilisation phase, after foot-contact (Fig. 1-2).

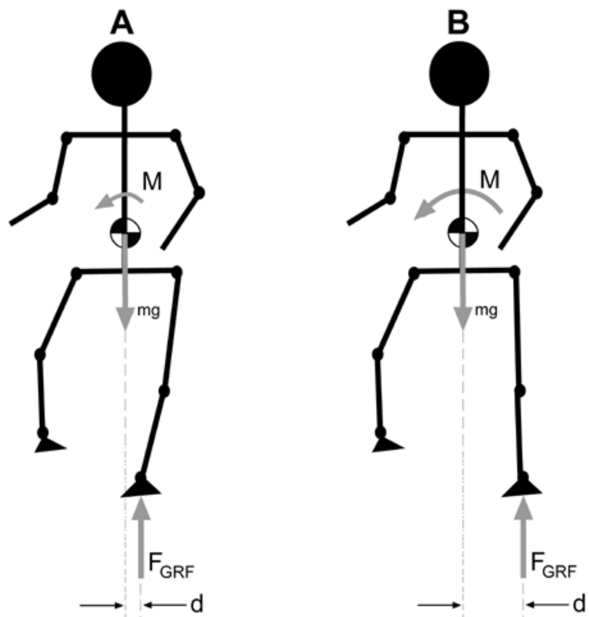


**Figure 1-1.** Frontal plane configuration of the body at the onset of stepping during a volitional (panel A, left) and perturbation-evoked step (panel B, right). The frontal plane external gravitational moment (M = mg * d) is typically smaller during volitional stepping because of the influence of the ML APA, which tends to move the line of action of the gravitational force (mg) closer to the supporting ankle.

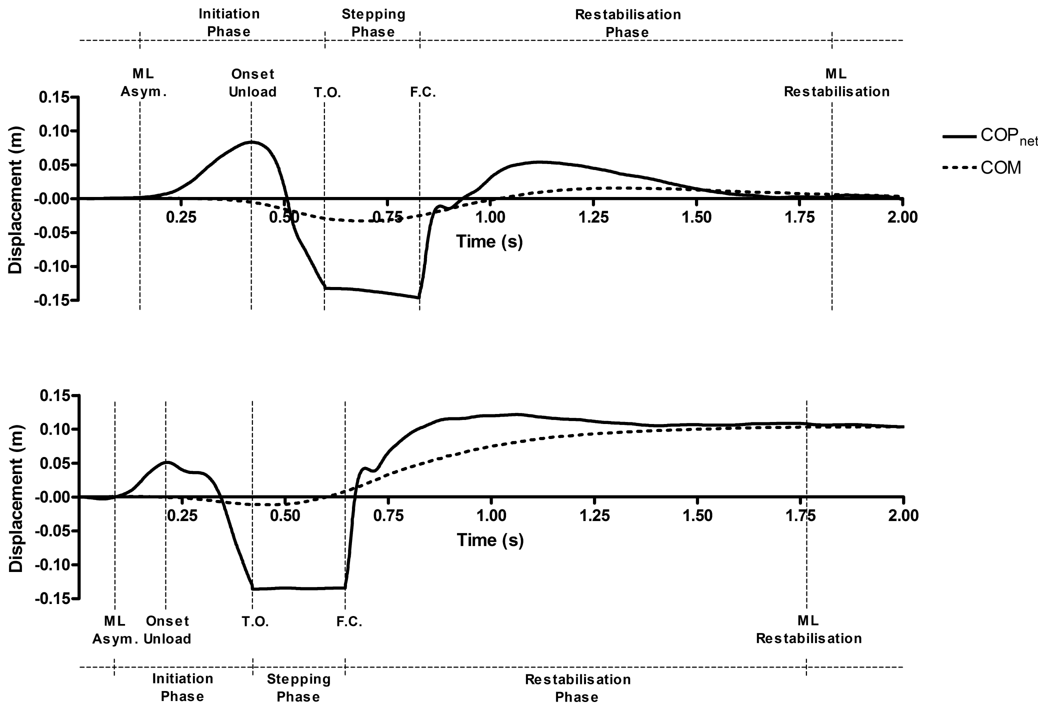


**Figure 1-2.** Displacement of the net COP (solid line) and COM (dotted line) for a representative trial within each the volitional (top) and perturbation-evoked (bottom) stepping conditions (Studies 2 and 3). Data begin at cable-release (perturbation-evoked) or presentation of auditory tone (volitional). ML Asym. = onset of ML asymmetry; Onset unload = onset of ML unloading; T.O. = Toe-off; F.C. = Foot-contact; ML Restabilisation = point of ML restabilisation. Positive values indicate displacement toward stepping limb; negative values indicate displacement toward stance limb.

There has been a substantial body of work regarding age-related changes in balance control during quiet standing (Fernie, Gryfe, Holliday & Llewellyn, 1982; Shumway-Cook, Woollacott, Kerns & Baldwin, 1997), perturbed standing (Lin, Woollacott & Jensen, 2004) gait initiation (Chang & Krebs, 1999) and steady-state gait (Winter et al., 1990), with inferences regarding a relationship to falls risk. In addition, considerable research exists on the control of anteroposterior stability upon anteroposterior perturbation to standing balance (Do, Breniere & Brenguier, 1982; McIlroy & Maki, 1996; Thelen et al., 2000; Wojcik, Thelen, Schultz, Ashton-Miller & Alexander, 2001; Hsiao-Wecksler & Robinovitch, 2007; Karamanidis, Arampatzis & Mademli, 2008). Taken together, this work has revealed that older adults are not able to recover from as large a maximal perturbation magnitude as are younger adults (Thelen, Wojcik, Schultz, Ashton-Miller & Alexander, 1997; Wojcik, Thelen, Schultz, Ashton-Miller & Alexander, 1999). Such differences have been largely attributed to reductions in swing limb velocity, step length, or sagittal plane lower limb intersegmental moments at foot-contact (Hsiao-Wecksler & Robinovitch, 2007; Karamanidis, Arampatzis & Mademli, 2008).

In contrast, studies employing a sub-maximal perturbation magnitude have not found age-related differences in the initial temporospatial parameters of the response (McIlroy & Maki, 1996; Thelen et al., 1997; Rogers, Hedman, Johnson, Cain & Hanke, 2001; Rogers & Mille, 2003). This work has, however, found older adults to have greater mediolateral instability at the time of foot-contact, as evidenced by greater lateral step placement (Rogers et al., 2001; Schulz, Ashton-Miller & Alexander, 2005) or greater number of laterally directed steps (McIlroy & Maki, 1996), which suggests that the development of mediolateral instability occurs during the stepping phase or restabilisation phase after foot-contact. Similar evidence of mediolateral instability has been found during unplanned and unplanned gait termination (O'Kane, McGibbon & Krebs, 2003; Tirosh & Sparrow, 2005).

Despite our understanding of the proactive and reactive mechanisms for balance control at movement initiation, such as anticipatory postural adjustments in volitional stepping or early automatic postural responses in reactive stepping, respectively, there appears to be little understanding or consensus regarding the origins of age-related mediolateral instability that can occur, and manifest, at movement termination during such stepping tasks (McIlroy & Maki, 1996; O'Kane, McGibbon & Krebs, 2003; Tirosh & Sparrow, 2005). Notwithstanding the well documented differences between volitional and perturbation-evoked stepping until the time of foot-contact, we believe that the challenges for effectively controlling and arresting the lateral progression of the COM within the base of support have important links to the restabilisation phase, and are common to both forms of stepping.

From this, it is proposed that research with a more direct focus on mediolateral stability control, specifically at foot contact during volitional and reactive stepping, will lead to a more complete understanding of the challenges faced by older adults in the maintenance of dynamic stability. Further, we hope that insight gained during such single step responses can be used to direct subsequent study into mediolateral dynamic stability during steady-state gait. The current work sets out to provide new insight to help understand the factors that influence age-related changes in dynamic stability control during stepping.

## 1.2 RELEVANCE AND SIGNIFICANCE

The high incidence of falls and fall-related injuries among Canadians over the age of 65 continues to be a key public health issue. Although this cohort represents approximately 14% of the Canadian population (as of 2006), they accounted for 51% of all unintentional fall cases requiring hospitalization during the 2007 fiscal year (Canadian Institute of Health Information, 2010) and totalled 46% of health care costs for falls in 2004 (SMARTRISK, 2009).

Many authors have suggested that measures of mediolateral stability may be good predictors of falls (Maki, Holliday & Topper, 1994; Lord, Rogers, Howland & Fitzpatrick, 1999; Brauer, Burns & Galley, 2000; Hilliard et al., 2008; Schrager, Kelly, Price, Ferrucci, Shumway-Cook, 2008). While a fall in any direction can lead to injury, increased fear of falling, reduction in physical activity levels and quality of life, lateral falls have been shown to be prevalent (Maki et al., 1994) and can be particularly devastating because of the increased probability of impact to the lateral aspect of the pelvis or leg, which increases the likelihood of hip fracture and subsequent hospitalisation (Nevitt & Cummings, 1993; Hayes et al., 1996; Robinovitch, Inkster, Maurer, Warnick, 2003).

Both the rate of fall related hospitalization and the length of stay per case have been found to be positively related with age. This trend rises sharply when examining the subset of individuals above the age of 85. With this hospitalization comes the risk of comorbidity and subsequent reduction in physical activity levels, which may give rise to increased fall risk and reduced quality of life (Nevitt et al., 1989; Bloem, Steijns & Smits-Engelsman, 2003). Unfortunately, as the current proportion of the population over the age of 65 is predicted to double by 2031 (Statistics Canada, 2007), the absolute number of individuals experiencing falls, fall-related injuries and subsequent hospitalization will increase dramatically. As a consequence, without a better understanding of the factors responsible for falls and subsequent evidence-based interventions, so too will the social and economic burdens on Canadian society.

## 1.3 RESEARCH OBJECTIVES

The global objective of the following studies was to elucidate the factors responsible for age-related decline in the control of mediolateral dynamic stability, which are potentially linked to fall risk. To this end, this work focussed on the recovery of upright dynamic stability on, and after, foot contact when stepping was evoked either volitionally or reactively by applied whole-body perturbation.

As a foundation for subsequent studies, the first study in this series sought to better understand the control of the centre of mass during voluntary stepping, by quantifying the trajectory of the COM and the intertrial variability during the restabilisation phase of self-initiated single steps. The intention of this study was to characterise the COM kinematics in young adults under ideal conditions, in which there was an opportunity to plan and self-initiate the movement. The second study in this series sought to extend our previous work of voluntary stepping in young adults to address mediolateral stability control in healthy older adults, again under conditions where there was the opportunity to pre-plan appropriate movement parameters for maximal stability. As it was possible that older adults would mask difficulty in stability control by altering their movement speed, we included rapid-stepping trials initiated in response to an auditory cue. Again, we quantified COM kinematics to reveal age-related differences in incongruity magnitude and intertrial variability. Further, we analysed the timing and magnitude of the waveform representing the difference between the inclination angles of the net ground reaction force and COP-COM, to aid in the clarification of the underlying mechanisms responsible for any age-related differences in the abovementioned kinematic variables. The third study extended the aforementioned methodology to understand the mechanisms responsible for previously reported age-related declines in mediolateral dynamic stability during reactive stepping, by examining the restabilisation phase following a cable-release perturbation to standing balance.

# CHAPTER 2

# Background Literature

## 2.1 FALL RISK

Early investigations attempting to uncover a principal cause of falls in community-dwelling older adults have instead revealed a complex interaction of intrinsic and extrinsic factors, which combine to influence fall risk. Intrinsic factors include both biological and behavioural mechanisms such as reduced muscle strength, physical fitness, visual, vestibular or cognitive impairments, fear of falling, depression and falls history (Tinetti et al., 1988; Campbell, Borrie & Spears, 1989; Maki et al., 1994; Graafmans et al., 1996; Lord & Dayhew, 2001). Extrinsic factors tend to compound the effect of any intrinsic components and have been found to include the use of assistive devices (Bateni & Maki, 2005) and the physical environment of both the home and community. Interestingly, investigations have proposed that, with specific interventions, the impact of many of these risk factors can be diminished – especially those of an extrinsic nature. Due to the complexity of intrinsic factors, however, a large fraction of research has been aimed at understanding how these elements relate to fall risk. More specifically, much emphasis has been placed on uncovering the role of the balance control system.

Several clinical tests of balance have been used, with varying success, to predict fall risk in older adults. For example the ‘functional reach’ test is a clinical measure of balance, tested for reliability and construct validity, which has been suggested to reflect the limits of stability (Duncan, Weiner, Chandler & Studenski, 1990). A subsequent study by Duncan, Studenski, Chandler and Prescott (1992) evaluated this test in its utility as a tool for identifying older adults at risk for recurrent falls. While these authors indicated that the test demonstrated predictive validity, other authors (Jonsson, Henriksson &

Hirschfeld, 2002) have questioned the construct validity of this tool, as it relates to an individual's stability limits, as they found a low correlation between centre of pressure (COP) displacement measures and reach distance and argue that an individual's choice of movement pattern may be more highly correlated to reach distance.

The Berg Balance Scale (BBS) has also been evaluated in its predictive validity for falls, in comparison to older adults' self-reported fall rate. Shumway-Cook, Baldwin, Polissar and Gruber (1997) have reported that this instrument demonstrated high sensitivity and specificity in dichotomously classifying fall risk. In contrast, Bogle Thorbahn and Newton (1996) performed a similar retrospective study and found the BBS to have sufficient specificity to classify non-fallers, but low sensitivity in identifying individuals prone to falling. Interestingly, Muir, Berg, Chesworth and Speechley (2008) have asserted that the BBS was developed as a clinical tool to evaluate functional balance in older adults and, as such, was not intended to provide a dichotomous, predictive, measure of fall risk. Because of its wide use as a predictive tool, these authors performed a prospective evaluation of the predictive validity of the BBS and found this tool to have insufficient sensitivity to justify its use as a dichotomous scale to predict fall risk (Muir et al., 2008). While the functional reach test, BBS and similar clinical tests of balance can provide an indication of global balance capabilities, the resolution of such instruments in detecting small changes in balance performance is not known. In addition, although clinical tests are generally intended to evaluate balance during functional tasks, it is not clear which elements of balance control are being probed. While clinical scales of balance performance may reduce the need for time and equipment resources and have a clear role as evaluative tools in the assessment of a patient's general functional balance capability, quantitative experimental analyses may provide more detailed information regarding the underlying mechanisms governing balance control.

## 2.2 STATIC- AND PERTURBED-STANCE BALANCE CONTROL

An idealized rigid body at rest will remain passively stable when subjected to a transient unbalancing force, so long as the line of action of the gravitational force, acting at the COM, does not exceed the geometric limits of the BOS. In contrast, humans are comprised of a system of linked segments, which is inherently unstable during standing. This is largely due to the numerous degrees of freedom and continuous internal sources of perturbation, such haemodynamics and respiration (Conforto, Schmid, Camomilla, D'Alessio & Cappozzo 2001; Schmid, Conforto, Bibbo & D'Alessio, 2004). The understanding of stability control in a human system is further complicated by the fact that the relationship between the COM and BOS is not only governed by constraints related to BOS geometry, but also by the functional limits of the BOS, which may be related to the capacity of the sensory and motor systems or fear of falling (King, Judge & Wolfson, 1994; Binda, Culham & Brouwer, 2003; Holbein-Jenny, McDermott, Shaw & Demchak, 2007). As a result of such factors, the maintenance of upright standing stability requires the continuous generation and modulation of muscular forces to oppose internal and external destabilising forces - essentially, necessitating active control of the COM.

This concept has formed the foundation for numerous studies seeking to understand how humans maintain standing, or 'static', balance. Traditionally, researchers focussed on global measures of balance, such as the location of the net centre of pressure ($COP_{net}$) in both mediolateral and anteroposterior directions, using a single force plate. While measurement of the time-varying displacement of the $COP_{net}$ has, at times, been termed ‘postural sway’, Winter (1995) asserts that this term describes the movement of the COM, rather than the COP. Nevertheless, while the COP and COM are independent signals, displacement of the COP reflects the generation of active muscular forces to control the position of the COM and, as such, can be used to make inferences about postural control. Winter (1995) further highlights the fact that authors who have incorrectly interpreted the COP displacement as ‘postural sway’ likely overlooked the earlier work of Murray, Seirig and Scholz (1967), who may have been the first to concurrently quantify both the COP and COM and identify the interaction between these two variables. Moreover, Murray et al. (1967) may have also been the first to indicate that the variations in the position of the COP were much larger than those of the COM and were indicative of muscular contractions, which, during quiet standing, served to accelerate the COM to regulate its position within the area of the BOS.

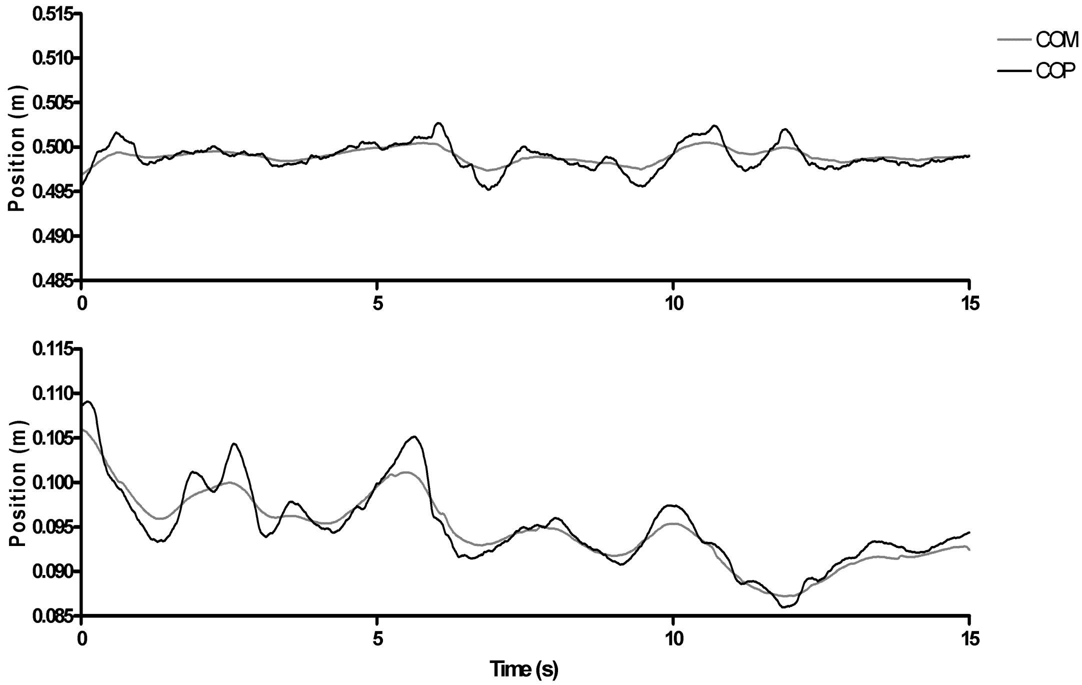


**Figure 2-1.** Centre of mass (grey) and centre of pressure (black) trajectories for mediolateral (top) and anteroposterior (bottom) directions. Fifteen seconds of data are presented from one participant, performing the 60 second quiet standing trial required for studies 2 and 3 from this thesis. Positions of the centre of mass and centre of pressure are plotted with respect to the origin of the global coordinate system.

Early studies of age-related differences in postural control have focussed on the measurement of the COP as a global indicator of the spontaneous postural activity that occurs during standing balance. It is believed that the COP signal contains information pertaining to the underlying control response to COM displacement. As such, an increased magnitude of displacement-based measures of the COP, such as peak-to-peak amplitude, root-mean-square deviation, ‘sway’ area or path length, have been presumed to reflect an impaired ability to control the COM, as greater COP displacement implies increased active muscular involvement in COM control – presumably in response to increased COM displacement. Indeed, older adults have been found to have increased COP displacement (Era & Heikkinen, 1985) and velocity (Fernie et al., 1982; Prieto, Myklebust, Hoffmann, Lovett & Myklebust, 1996), however, Patla, Frank and Winter (1990) and Maki and McIlroy (1996) emphasize that an increase in COP displacement and/or velocity, in isolation, may not be sufficient to deduce deterioration of the postural control system or instability. As such, a more complete understanding of the capacity of the postural control system may be gained from concurrent analysis of both the control (COP) and controlled (COM) variables. To further emphasise this point, Patla et al. (1990) and Hof, Gazendam and Sinke (2005) have proposed hypothetical experiments – involving either a mannequin or broom handle balanced on end on a force platform – during which the measured time-varying COP displacement would clearly be nil, but these systems could not be assumed to be stable. Accordingly, studies in humans have revealed that displacement-based COP measures can be reduced as a function of neurological disorder (Horak, Nutt & Nashner, 1992), or perceived threat to stability (Carpenter, Adkin, Brawley & Frank, 2006; Laufer, Barak & Chemel, 2006). Further, to provide a more complete understanding of the underlying control, displacement-based COP measures could also be supplemented with frequency-based measures, such as the mean (or median) power frequency of the COP or COM-COP error signal, which could provide information regarding the stiffness of the muscles governing COM control.

In contrast to quiet standing, falls are more likely to result from self-imposed internal or unexpected external perturbations, which disrupt the relationship between the COM and BOS. Consequently, early studies of perturbed standing balance emerged as a simple means to safely probe the balance control system. Nashner (1977) has observed that, in response to an unexpected platform translation, postural responses were functionally organized, whereby there was generally an orderly sequence of muscle activation from distal to proximal. This distal-to-proximal temporal sequencing of postural muscle activation has also been elicited in an anticipatory manner, prior to the onset of focal muscle activity, when young adults were asked to exert force through the hand upon a manipulandum, while maintaining a standing position (Cordo & Nashner, 1982).

In addition to this distal-to-proximal response, which has been termed an ‘ankle strategy’ when elicited in response to external postural perturbation, Horak and Nashner (1986) also observed another strategy, termed a ‘hip strategy’, in which abdominal/lumbar and anterior/posterior thigh musculature were activated in response to unexpected posterior/anterior platform translations, respectively. Relative to the ‘ankle strategy’, this functional coupling of muscle activation was believed to occur in response to platform translations of higher displacement or velocity or in situations where the generation of ankle moments is ineffective in altering the acceleration of the COM, such as when standing on narrow or compliant surfaces. While ‘ankle’ and ‘hip’ strategies can occur as distinct responses, they have been observed to most often emerge in various combinations (Horak & Nashner, 1986).

Age-related changes in muscle onset timing, organization, antagonist co-activation and even the choice of ‘strategy’ have been noted by some authors examining external perturbations to stance (Woollacott & Shumway-Cook, 1990; Lin & Woollacott, 2002). More specifically, these authors have found that older adults can display a delayed onset of the ankle musculature and an increased likelihood of a proximal-to-distal organization of muscle activation, ultimately leading to the emergence of a ‘hip strategy’ under conditions that would otherwise evoke an ‘ankle strategy’ in young adults. Studies examining anticipatory control in response to focal arm movements have also been performed in samples of older adults, in which increased onset latencies in postural muscles, relative to the onset of focal muscle activation, have been observed (Stelmach, Populin & Muller, 1990; Rogers, Kukulka & Soderberg, 1992). While such evidence of disordered control likely results in instability, Patla et al. (1990) suggest that the ineffectiveness of these altered responses can only be inferred unless supported by kinetic analyses and measurement of the COM.

## 2.3 COMPENSATORY STEPPING RESPONSES

Historically, ‘fixed-support’ strategies have been believed to lie on a continuum, where ‘ankle’ and ‘hip’ strategies are used in response to small and large amplitude perturbations, respectively. Stepping, or ‘change-in-support’, strategies were generally assumed to lie at the far end of the continuum and were employed only when the vertical projection of the COM travelled outside of the BOS (Horak & Nashner, 1986).

While the maintenance of ‘static’ upright stability requires the maintenance of the position of the COM within the BOS, this is not a necessary condition for the maintenance of ‘dynamic’ postural control. Murray et al. (1967) observed, during jumping and sit-to-stand tasks, that the COM was positioned well outside the BOS and presented this as evidence that “should dispel the concept that the line of gravity must constantly remain within the supporting area to avoid falling”(p. 837). This reasoning has been formalized by Pai and Patton (1997), who have indicated that isolated measurement of the position of the COM relative to the BOS is insufficient to infer stability. Rather, measurement of both COM position and velocity, in relation to the BOS, are required. Using a reductionist approach, these authors used a two-segment sagittal plane model to demonstrate that stability can be maintained even if the position of the COM is outside the BOS, given that the velocity is directed toward the BOS. Conversely, it may be impossible to maintain stability when the COM is within the BOS, if the velocity is of sufficient magnitude and is directed away from the BOS. This work has given rise to a dynamic model of stability, based on both the position and velocity of the COM within the BOS, which has been better able to predict the need for stepping after postural perturbation (Pai, Rogers, Patton, Cain & Hanke, 1998; Pai, Maki, Iqbal, McIlroy & Perry, 2000). This research has provided support for the earlier work of McIlroy and Maki (1993c), who proposed that change-in-support reactions may not be at the end of a continuum of responses to postural perturbation, but may occur even when the COM is well inside the BOS (Pai et al., 1998; Pai et al., 2000; Mille, et al., 2003). Accordingly, the predominance of previously reported fixed-support strategies may have been the result of instructions given to the participants to regain balance without the use of a step (McIlroy and Maki, 1993c). In addition, this concept reveals the possibility that experiments examining age-related differences in postural control using ‘fixed-support’ strategies may lack ecological validity. Given findings that healthy older adults are more likely, relative to young adults, to initiate a stepping response at a given level of perturbation (Jensen, Brown & Woollacott, 2001), previously reported age-related effects may have occurred because older adults were required to use a non-preferred strategy for restabilisation. Investigations of stepping behaviour, whether driven volitionally or by perturbation, may provide better insight into the age-related mechanisms responsible for balance dyscontrol and falls.

### *2.3.1 Early Automatic Postural Responses*

Although compensatory stepping reactions to postural perturbation are more common than previously believed, early automatic postural responses, such as the generation of ankle moments, typically precede the onset of stepping. This can present a conflict between responses, as stepping would require the generation of an ankle moment, resulting in COP movement in a direction opposite to that which would be elicited by the early automatic response. In an attempt to understand this conflict, McIlroy and Maki (1993a) performed a study using forward platform translations to determine the influence of prior planning of a step on the characteristics of the early postural responses (viz. magnitude of tibialis anterior EMG activity). These authors found that the automatic postural responses were always elicited, with the same onset latency, irrespective of whether subjects initiated a stepping response. Steps that were pre-planned, however, exhibited a reduced magnitude of tibialis anterior activity when compared to either feet-in-place- or unplanned stepping responses. Taken together, the authors suggest that the early responses are either immutable, but can be attenuated as a function task demands, or may serve a functional benefit by reducing the COM velocity prior to stepping.

Further support for both the immutability of the early automatic postural responses and their potential functional role has been provided by Weerdesteyn, Laing and Robinovitch

(2008), who asked subjects to either recover their balance or fall onto a mattress, after release from a backward lean. Given minimal differences in tibialis anterior onset latency, yet significantly reduced amplitude of EMG activity in ‘fall’ trials, the authors suggest that these two tasks require the activation of the same motor program but with down-regulated gain in the ‘fall’ trials. Similar results regarding the consistency of the timing of early automatic postural responses have also been observed in response to release from varying forward lean angles (Do et al., 1982), with and without constraints on the use of stepping strategies (McIlroy & Maki, 1993c) and when participants are repeatedly exposed to the same perturbation magnitude (McIlroy & Maki, 1995).

### *2.3.2 Onset of Step Initiation*

Despite the general consensus regarding consistent onset latencies of the early automatic postural responses, there is some disparity between studies examining the onset timing of stepping. These discrepancies, however, may be largely due to differences in perturbation method (i.e. moving platform or cable-release), recovery method (i.e. number of steps) and variation in the means by which step onset was identified.

A study by Do et al. (1982), in which participants were released from an initial forward lean and instructed to recover balance by walking, found that the time to stepping limb toe-off (denoting the onset of the stepping phase) was not affected by the magnitude of the lean angle. This is somewhat contradictory to the results of McIlroy and Maki (1993c) who found that the time to onset of intra-limb vertical force asymmetry (denoting the onset of the step initiation phase) was reduced with increasing magnitude of platform perturbation. While the aforementioned measures used to identify the onset of stepping are slightly different, it is perhaps more important that the study by Do et al. (1982) featured a predictable direction of perturbation and allowed subjects to pre-plan multiple consecutive alternating steps of self-selected length as a means to recover balance. As a result, subjects may not have been forced to generate a rapid-onset first step, since the forward COM velocity could be controlled in subsequent steps.

Supporting evidence for this notion has been presented in a subsequent study by Do, Schneider and Chong (1999) using a similar methodology to their previous study. Specifically, relative to self-selected step length trials, participants reduced the time to stepping limb toe-off when they were forced to take an initial step of reduced length. This suggests that the reduction in the latency of step onset could be related to the risk of falling, since a short step would require a more rapid onset of response to bring the stepping foot ahead of the COM, for the generation of an external moment to reverse the polarity of the COM acceleration. Similar findings regarding an inverse relationship between time to stepping limb toe-off and perturbation magnitude have been reported within studies by Thelen et al. (1997) and Thelen et al. (2000), both using the cable-release method.

Interestingly, both McIlroy and Maki (1993c) and Rogers, Hain, Hanke and Janssen (1996) suggest the trigger for the rapid onset of stepping, and its coupling to perturbation magnitude, could be based on sensory information relating to platform acceleration, since displacement- or velocity-based variables would not peak until well after the onset of stepping. This view has been indirectly supported by studies examining the attenuation and facilitation of plantar foot surface cutaneous sensation (Maki, Perry, Norrie & McIlroy, 1999; Perry, McIlroy & Maki, 2000), which have found increased and reduced latency in the onset of step initiation, respectively, which is likely related to the ability to detect the onset of platform acceleration. Similarly, Jensen et al. (2001) suggest that stepping is triggered as a function of the platform-induced accelerations, as they are propagated up the linked system of segments. For cable-release experiments, however, the necessity for a more rapid onset of response with increasing lean angle could be related to the opportunity to form a detailed internal representation of the initial positions of the COM and BOS during the interval before cable-release. The ultimate trigger for the onset of instability and, hence, the necessity for response initiation could be related to sensory information conveying a change in (or removal of) pressure from the cutaneous surface beneath the harness used to restrain the subject in the initial lean position.

### *2.3.3 Swing Phase: Step Length and Step Time*

In theory, a wide array of combinations of step length and step time could be used to recover balance from a given magnitude of perturbation, ranging from short duration steps of minimal length to long duration steps of maximal length. A short length step response could be completed in a short time, which could allow for the execution of additional steps, should the initial step prove to be ineffective (Maki & McIlroy, 1999). Moreover, the execution of multiple-steps may reduce the biomechanical demands of each step, such that each step demands a smaller joint range of motion, peak swing phase moments and AP impulse (King, Luchies, Stylianou, Schiffman & Thelen, 2005). Larger step lengths, however, can maximize the stability margins associated with the initial step by extending the BOS length well beyond the position of the COM at the time of foot-contact, which would allow ample distance and time to decelerate the COM before reaching the stability limits (Maki & McIlroy, 1999). In general, studies have found that an increase in perturbation magnitude is associated with an increased step length and reduced step time (Do et al., 1982; Thelen et al., 1997; Hsiao & Robinovitch, 1999; Wojcik et al., 1999).

Using displacement- and velocity-based estimates of stability margins, Maki and McIlroy (1999), indicated the existence of a trade-off between the rapidity of response execution and anteroposterior stability when executing stepping responses. Briefly, reductions in the latency of step contact, marked by reductions in step length, were predicted to result in reduced stability. Conversely, increases in step contact latency, and subsequent increases in step length and swing duration, were predicted to result in increased stability. Within anatomical constraints, step length may largely determine the degree of anteroposterior restabilisation afforded by the step, as it is the determinant of the distance over which the COM can be decelerated before reaching the physical limits of the BOS and is related to the magnitude of the restorative moment created by the vertical component of the ground reaction force, since step length would dictate the length of the moment arm. Comparing their model predictions to experimental data (McIlroy & Maki, 1996), these authors concluded that stability took precedence in single step responses, whereas reductions in the latency of step contact, at the expense of stability, was favoured in responses consisting of multiple anterior steps.

Wu, Ji, Jin and Pai (2007) extended their previous work and modelling approach (Pai et al., 2000) to examine the minimal step length required to regain stability following a forward perturbation. These authors revealed that a greater anterior COM position or velocity at the instant of toe-off required an increase in the minimal step length necessary for balance recovery. Moreover, the model predicted an inverse relationship between ankle strength and step length, such that an inability to produce a large magnitude ankle moment required a corresponding increase in the minimal step length for balance recovery. It should be noted, however, that the authors used a four-segment sagittal plane model and did not include the contribution of hip or knee moments or the effect of swing time. When matching their initial conditions and anthropometric data to experimental data from McIlroy and Maki (1996), Wu et al. (2007) noted that the experimentally derived minimal step length for single-step responses was longer than the predicted value. Conversely, the experimentally derived minimal step length was shorter than the predicted value when multiple anterior steps were executed. This may indicate that single step responses reflect an attempt to maximize the stability margin at the expense of increased biomechanical demand, while multiple anterior step responses may occur because of reduced strength, an attempt to reduce peak loads, or an inability to generate rapid leg movement required to execute a step of increased length. Thus, it appears that multiple anterior step responses may emerge because of actual instability during the first step (Maki & McIlroy, 1999), which may be linked to either a pre-planned strategy or an inability to meet the biomechanical demands of the task.

### *2.3.4 Age-Related Effects*

A number of studies examining age-related differences in compensatory stepping behaviour have focussed on differences in the ability to recover anteroposterior stability as a function of perturbation magnitude. Studies of both men (Thelen et al., 1997) and women (Wojcik et al., 1999) have found that the maximum lean angle from which older adults were able to recover was significantly smaller than that of younger adults. These authors have suggested this difference to be primarily associated with age-related declines in the speed with which the lower extremity segments could be moved, rather than with issues associated with response initiation, such as reaction time or time of step onset. This theory has been supported by the authors' subsequent work, which has observed no age-related difference in the onset latency of muscular activity (Thelen et al., 2000).

Hsiao-Wecksler and Robinovitch (2007) have used a slightly different approach, whereby step length was manipulated and its effect on the maximum recoverable lean angle was noted. As with previous studies, there was a positive relationship between step length and the maximum lean angle. Interestingly, at a given step length, younger women were able to recover balance at a greater lean angle, largely due to an increased swing limb speed and ability to generate sagittal plane lower limb moments at step contact. These results correspond well with the work of Karamanidis et al. (2008) who suggested that the mechanism responsible for the age-related reduction in maximum recoverable lean angle was the inability of older adults to effectively reduce their anteroposterior COM velocity, resulting from lower muscle strength and consequent reductions in anteroposterior braking force.

Studies that have tested balance recovery abilities in response to non-maximal anteroposterior perturbation have generally failed to find significant age-related differences in many temporospatial parameters of the response. McIlroy and Maki (1996) have noted that older adults less frequently employ a mediolateral APA before stepping, but exhibit no differences in the onset of swing limb unloading, foot-off time, swing time, or foot contact time, relative to younger adults. Despite the ability of the younger adults to execute a mediolateral APA before stepping, which could reveal an ability to more rapidly discriminate the onset of perturbation, there were no differences in step length, step width, AP or ML COM displacement or velocity at the time of foot contact. This is consistent with later research, which has indicated that the mediolateral APAs that may emerge during compensatory step initiation have little influence on the subsequent mediolateral COM dynamics (McIlroy & Maki 1999; Rogers et al., 2001).

Despite the similarity in many initial movement parameters, older adults have been found to use a greater number of steps to recover balance (McIlroy & Maki, 1996; Schulz et al., 2005), which is believed to be indicative of fall risk (Maki et al., 2001), since each step, if not executed correctly, presents an additional opportunity for instability. In addition to the number of steps required for restabilisation, older adults have also been found to either place the initial step more laterally (Rogers et al., 2001; Schulz et al., 2005) or to include a lateral component in the subsequent steps (McIlroy & Maki, 1996), despite the fact that the primary instability triggered by these perturbations was in the sagittal plane. Notwithstanding the differences in the manifestation of mediolateral instability, which may be related to the predictability of the perturbation and the ability to pre-plan a response, both results suggest that older adults may have trouble regulating mediolateral dynamic stability when responding to anteroposterior perturbations. When considering this in concert with the temporospatial and kinematic similarities in the initial movement, it appears likely that mediolateral instability may arise from problems with on-line control of the swing phase or with the control of applied forces on, and after, foot-contact.

### 2.4 SUMMARY

Although there is a considerable body of literature concerning the risk factors for falls, the identification of individuals at risk for falling and the potential underlying mechanisms associated with both balance control and dyscontrol, the prevalence of falls among older adults continues to be a significant public health concern. Most falls have been found to occur during dynamic activity, whether self-initiated or evoked by external perturbation. A number of studies have stressed the association between mediolateral stability, falls and subsequent hip fracture. Interestingly, little attention has been given to the identification of mechanisms responsible for the observed age-related differences in the ability to recover mediolateral dynamic stability at movement termination, either during self-initiated or perturbation-evoked stepping. A better understanding of such age-related differences could lead to improved evidence-based interventions, with the goal of reducing the incidence of falls and fall-related injuries in the population.

# CHAPTER 3

## Dynamic Stability Control during Volitional Stepping: A Focus on the Restabilisation Phase at Movement Termination

## 3.1 OVERVIEW

This work sought to advance the understanding of dynamic stability control during stepping. The specific intention was to better understand the control of the centre of mass (COM) during voluntary stepping, by characterizing its trajectory and intertrial variability. Young participants (n=10) performed five different stepping tasks to vary the challenge to COM control: 1) preferred step, 2) long step 3) wide step, 4) long and wide step and 5) rapid step. The trajectory of the total body COM during the restabilisation phase was assessed by quantifying the magnitude of incongruity between the peak and final COM position. The intertrial variability of incongruity and the extent to which incongruity was reduced with trial repetition were also evaluated. Interestingly, incongruity was typical during preferred stepping, with a strong bias toward overshoot. In the frontal plane, the magnitude of incongruity and the incidence of overshoot were greater in trials with increased step width. The variability of incongruity did not vary by condition nor was there evidence of adaptive changes. Together, these results suggest that overshoots may represent a strategy linked to gait initiation or to the simplification of reactive control during the restabilisation phase. Further insight into these mechanisms will be gained by examining the kinetic determinants of dynamic stability control.

## 3.2 INTRODUCTION

The regulation of the relationship between the centre of mass (COM) and base of support (BOS) is a complex control problem, which is essential for the maintenance of upright stability. Various pathological conditions or advancing age, however, can affect the ability to maintain dynamic stability, thereby increasing the risk of falls (McIlroy & Maki, 1996; Maki & McIlroy, 2006). The challenges to dynamic stability are manifest in tasks such as voluntary gait initiation (Breniere & Do, 1991; Brunt et al., 1991; Jian, Winter, Ishac & Gilchrist, 1993; Elble, Moody, Leffler & Sinha, 1994; Halliday, Winter, Frank, Patla & Prince, 1998), termination (Jian et al., 1993; O’Kane et al., 2003), turning (Cao, Ashton-Miller, Schultz & Alexander, 1997) and perturbation-evoked stepping (McIlroy & Maki, 1993c; Rogers et al., 2001) – all of which have been studied extensively.

The control of stepping involves several important phases: initial preparation, step initiation, limb unloading, swing phase, followed by foot-contact and restabilisation. Few studies have focussed specifically on the restabilisation phase of movement, which occurs subsequent to foot contact. This phase is particularly important for the maintenance of dynamic stability, as it may have the most direct influence on the kinematics of the COM after movement initiation. Challenges to control during the restabilisation phase may be evident from the occurrence of multiple step responses when individuals attempt to regain balance by stepping (Luchies, Alexander, Schultz & Ashton-Miller, 1994; McIlroy & Maki, 1996; Maki et al., 1999; Perry et al., 2000). Similarly, older adults have been found to require additional steps during unplanned gait termination (Tirosh & Sparrow, 2004), which may arise from difficulty in regulating the position and velocity of the COM within the BOS after foot contact.

We suggest that the capacity for effectively regulating the kinematics of the COM during the period subsequent to foot-contact to be a central determinant of dynamic stability during both voluntary and reactive stepping. This initial study is focussed on the kinematics of the COM during the restabilisation phase of a voluntarily-initiated single step.

The primary hypothesis was that when participants stepped with self-selected step length and width, there would be little incidence of incongruity between the peak COM position and the final, stable, COM position, when examined in either the anteroposterior (AP) or mediolateral (ML) direction (Fig. 3-1). Operationally, during the restabilisation phase, we expected that the peak COM position would remain within a 95% confidence band around the mean final COM position.

In contrast, it was anticipated that increasing and constraining step length or width would increase the challenges in stability control after foot contact. Correspondingly, we hypothesized that we would observe an increase in incongruity magnitude, an increased proportion of trials in which the COM overshot its final position, and greater intertrial variability of incongruity magnitude. Lastly, with practice during non-preferred stepping conditions, we hypothesized we would observe a reduction in the corresponding AP and ML incongruity over the course of consecutive trials, as individuals became familiar with the movement dynamics during the restabilisation phase. We view this initial work examining dynamic stability control during voluntary stepping in a sample of healthy young adults to be an important precursor to subsequent studies focussed on age-related or disordered control.

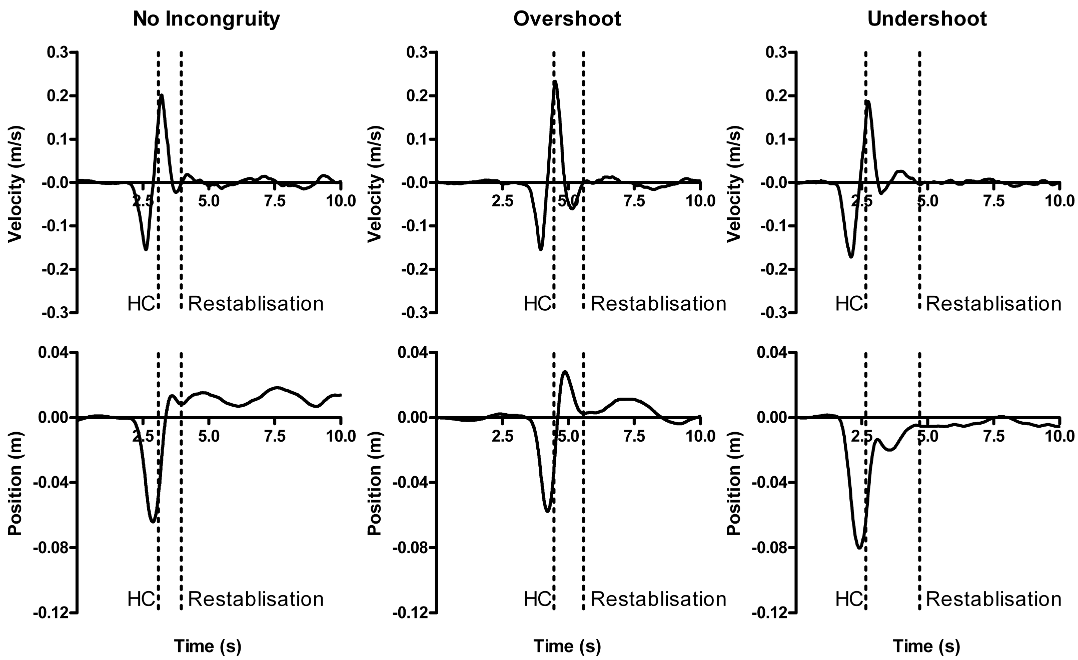


**Figure 3-1.** Representative centre of mass (COM) velocity-time (top) and position-time (bottom) waveforms depicting the three possible incongruity forms: no incongruity (left), overshoot (centre) and undershoot (right). Restabilisation signifies the point of restabilisation. The restabilisation phase occurs between heel-contact (HC) and the point of restabilisation.

## 3.3 METHODS

### *3.3.1 Participants*

Ten healthy young male participants (age 24.1 (2.9) years), without balance impairment or history of falls, were recruited from the University population. Male participants were recruited based on ease of anatomical landmark determination and marker placement for the upper body. There is no current evidence that we should expect a difference in stability control between healthy young males and females (Hsiao-Wecksler, 2008).

### *3.3.2 Instrumentation and Set-Up*

Six Vicon MX-3+ cameras (Vicon Motion Systems, Los Angeles, CA) were used to record kinematic data (64 Hz). Four force platforms (Advanced Mechanical Technology, Inc., Watertown, MA.), embedded in the laboratory floor in a rectangular array, were used to measure the reaction forces and moments (512 Hz).

Retroreflective calibration markers, of 1 cm diameter, were placed on the participant over anatomical landmarks similar to those described by Hamill and Selbie (2004) for the lower limbs and pelvis. Additional calibration markers were placed bilaterally on the upper body, to define local coordinate systems for the trunk, head, upper and lower arms and hands. Rigid clusters containing four markers, placed on the sacrum and trunk, and bilaterally on the feet, legs, thighs, upper and lower arms were used to track the position and orientation of each respective segment.

### *3.3.3 Protocol*

Participants took part in four different task conditions, which required a single voluntary step with the preferred leg. Ten consecutive trials were collected in each condition:

1. Preferred AP step length/width (AP and ML preferred) (PREF1);
2. **Increased** AP step length (ML preferred) (AP);
3. I**ncreased** ML step width (AP preferred) (ML);
4. **Increased** AP step length, **increased** ML step width (AP&ML);

Conditions with preferred step length and width were performed as the first trial block (PREF1) and again as the last trial block (PREF2) to assess long-term adaptive changes. The order of the remaining three task conditions (AP, ML and AP&ML) was randomized across subjects. An additional block was conducted in which participants were instructed to step as “rapidly as possible” with preferred step length and width (RAPID). This was included after the completion of all other task conditions to avoid task instruction carryover that may influence speed of stepping in the other task conditions.

Due to constraints on force plate positioning, an absolute target point was prescribed (rather than standardized across subjects), which maximized step length and/or width. Two lengths of adhesive tape were placed on the force platform, parallel and/or perpendicular to the sagittal plane. Average step lengths were increased to 0.73 m; average step widths were increased to approximately 0.50 m, depending on the initial stance width.

Participants began by standing with their feet side-by-side, shoulder-width apart, on separate force platforms. The initial stance width and foot position was standardized within participants. After an auditory command, participants initiated a single step with their preferred leg and, upon landing, remained in a stable position until the end of the trial (approximately 5 seconds). To counter the possibility of anticipating the auditory command, the intervals at which the next command was given were varied.

### *3.3.4 Data Analysis*

The lower extremity was modelled as a rigid system of independently tracked segments. Segment masses were estimated using Dempster's segment parameters and segment COM positions were estimated using the geometrical model proposed by Hanavan (1964) (cited in Robertson et al., 2004). The total body COM was calculated as a weighted average of all body segments, where each segment was weighted according to its mass proportion.

The COM restabilisation point was defined as the time point after the first zero-crossing, at which the COM velocity waveform entered and remained inside an amplitude bandwidth bordered by +/- two standard deviations of the mean velocity during the last two seconds of the trial. The incongruity magnitude was assessed by calculating the local maximum COM position after foot contact and subtracting the mean of the stable region of the waveform, bound by a two second window beginning at the restabilisation point. The trial was considered to contain overshoot or undershoot if it exceeded upper or lower boundaries defined by the 95% confidence interval of the stable region. These calculations were carried out independently for AP and ML directions (Fig. 3-1).

### *3.3.5 Statistical Analyses*

To assess the first hypothesis, the percentage of trials in which there was no incongruity was calculated for each participant for the PREF1 condition. These values were entered into a one-sample t-test, to determine if trials without incongruity occurred at greater than a 50% occurrence rate.

To assess the second hypothesis, the mean absolute incongruity magnitude and the between trial variability were computed for both the AP and ML directions for each subject in each condition. Separate one-way repeated-measures ANOVAs were run on mean and variability measures for AP and ML data. Subsequent comparisons were made using Bonferroni-adjusted paired-samples t-tests (p=0.01). Measures of effect size were calculated using the formula for Cohen's d, using the original standard deviations. Cochran's Q test was used to determine if a larger proportion of trials containing overshoot occurred in conditions with unfamiliar step task conditions. Subsequent pair-wise tests were run with adjusted Wilcoxon signed ranks tests. Effect size, r, was calculated by dividing the z-score by the square root of the total number of observations.

To address the third hypothesis, trial-to-trial adaptations within each condition were assessed using separate ANOVAs, with 'trial' as the within-subjects factor. Planned polynomial contrasts were performed to test for the presence of linear, quadratic and cubic trends in the data. The potential for long-term adaptation was assessed in the comparison between PREF1 and PREF2 included in the analyses used to evaluate the second hypothesis.

## 3.4 RESULTS

### *3.4.1 Incongruity Magnitude – Trials with Preferred Step Characteristics*

Contrary to our first hypothesis, less than 30% of trials contained no incongruity between the peak and final COM position ($t(9) = -7.72$, $p < 0.001$, $d = 2.44$ [AP]; $t(9) = -4.13$, $p =$

0.003, $d$ = -1.30 [ML]). Interestingly, overshoots of the final COM position were most prevalent, occurring in 77% (AP) and 68% (ML) of all trials in the PREF1 condition. Undershoots occurred in 8% (AP) and 4% (ML) of these trials. The average magnitude of incongruity was 0.019 m ($SD$ = 0.019 m) in the AP direction and 0.013 m ($SD$ = 0.012) in the ML direction (Fig. 3-2).

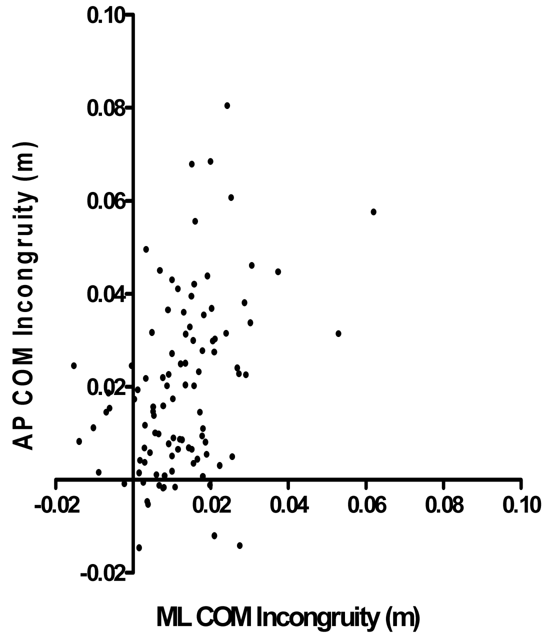


**Figure 3-2.** Centre of mass (COM) incongruity magnitude (AP vs. ML) for all participant data within the PREF1 condition. Positive values represent overshoot; negative values represent undershoot. Note: data points represent individual trials.

### *3.4.2 Task Differences*

Across task conditions, statistically significant differences in absolute incongruity magnitude emerged only in the mediolateral direction, $F(5,45) = 10.14$, $p < 0.001$, $\eta_p^2 = 0.530$. Adjusted t-tests indicated that there was an increased absolute magnitude of incongruity in ML trials, as compared to the PREF1 trials, $t(9) = 4.42$, $p = 0.002$, $d = 1.51$ (Fig. 3-3a). There was also increased ML absolute incongruity in the AP&ML trials relative to the PREF1 trials, but this difference was not statistically significant at the alpha level set for subsequent comparisons, $t(9) = 2.64$, $p = 0.027$, $d = 1.06$. There were no statistically significant differences in the standard deviations of the incongruity magnitude between conditions (Fig. 3-3b).

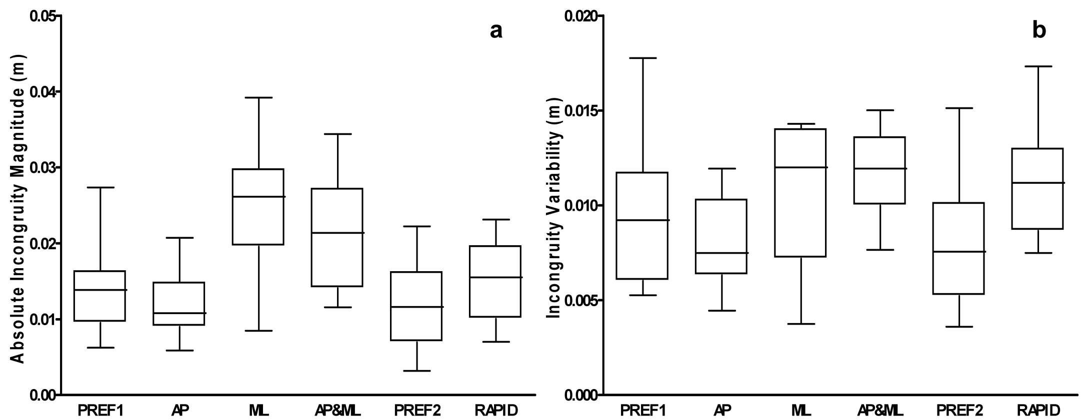


**Figure 3-3.** Absolute centre of mass (COM) incongruity magnitude (a) and variability of incongruity (b). Central box represents the lower to upper quartiles, with midline at the median. Whiskers extend to minimum and maximum values. Task conditions include: preferred stepping (PREF1 and PREF2); long step (AP); wide step (ML); long and wide step (AP&ML); rapid stepping (RAPID).

Significant differences in the proportion of trials with overshoot, again, emerged only in the mediolateral direction, $Q(5) = 29.62$, $p < 0.001$ (Fig. 3-4). The ML condition had an increased proportion of trials with overshoot, as compared to the PREF1, $Z = 2.96$, $p = 0.003$, $r = 0.21$. There were no differences between conditions when examining data in the anteroposterior direction.

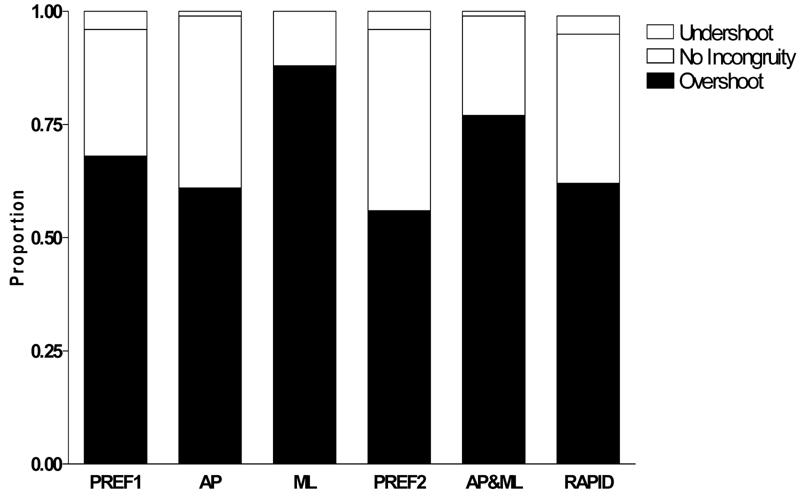

**Figure 3-4.** Average proportion of trials in each task condition containing each of the three forms of centre of mass incongruity (overshoot, no incongruity, undershoot). Task conditions include: preferred stepping (PREF1 and PREF2); long step (AP); wide step (ML); long and wide step (AP&ML); rapid stepping (RAPID).

### *3.4.3 Trial-to-Trial and Long-Term Adaptations*

When comparing across trials within each separate condition, no linear, quadratic or cubic trends were found for absolute incongruity in either the ML or AP direction for any condition (Fig. 3-5). Similarly, there were no long-term adaptive changes from the PREF1 to PREF2 condition when examining the magnitude, intertrial variability or proportion of trials with overshoot (Fig. 3-3, 3-4).

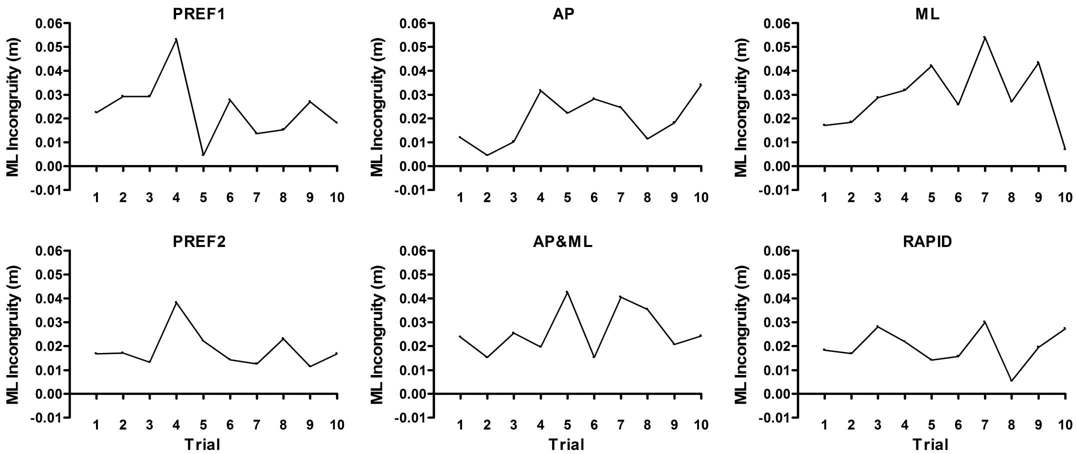


**Figure 3-5.** Variations in ML incongruity magnitude over the course of ten trials for each condition. Data are from one representative subject. Task conditions include: preferred stepping (PREF1 and PREF2); long step (AP); wide step (ML); long and wide step (AP&ML); rapid stepping (RAPID).

### *3.4.4 Secondary Analyses*

It was possible that COM incongruity was a product of improper planning or execution of the initial movement characteristics: ML APA amplitude, instantaneous COM velocity at foot-off, step length, width or time or the instantaneous COM velocity at foot contact. As such, we performed a correlation analysis and determined that such associations did not exist for any condition, in any direction. Similarly, COM incongruity could have been related to anthropometric parameters, such as body mass, height or leg length. Correlation analyses again failed to reveal such a relationship between anthropometrics and COM incongruity for any condition in any direction. Additionally, a paired-samples t-test was used to determine if the difference in increased absolute incongruity between the PREF1 and ML conditions persisted when the peak ML COM position was expressed a function of BOS width. No difference was found between PREF1 and ML conditions when peak ML COM position was normalized to step-width $t(9) = 0.805$, $p = 0.442$, $d = 0.23$.

## 3.5 DISCUSSION

The present work focussed on the restabilisation phase of stepping that occurs subsequent to foot contact, as difficulties in dynamic stability control during stepping may arise during this movement phase (Luchies, et al., 1994; McIlroy & Maki, 1996; Tirosh & Sparrow, 2004). Surprisingly, the data did not support our first hypothesis, as only a small percentage of trials occurred without incongruity. The data did, however, partially support our second hypothesis, as both the magnitude of ML incongruity and the incidence of ML overshoot were significantly larger in trials with increased step width. Interestingly, the variability of these incongruities was not larger in conditions with non-preferred step placement. Lastly, there was no reduction in the incongruity magnitude, variability of incongruity or in the proportion of trials with overshoot in the PREF2 trials relative to PREF1. Similarly, the magnitude of incongruity was not reduced over the course of repeated trials within any condition.

The question emerges as to why there is incongruity under conditions of well-learned, voluntary stepping with self-selected step length and width. The observed AP and ML incongruity during the restabilisation phase is likely unrelated to errors in planning/executing the prior phases of stepping or to the damping afforded by the musculoskeletal system, as incongruity was not correlated to any initial movement characteristic or to the instantaneous velocity of the COM at foot contact, respectively. Further, the considerable bias toward overshoots in this healthy, young sample suggests that, contrary to our initial belief, this form of COM incongruity may not be “error” whatsoever, but may exist to serve a functional role.

One possible explanation is that overshoots could represent a strategy linked to gait initiation and the regulation of momentum during steady-state gait. If the gait cycle were to be continued after the first step, anterior progression of the COM toward the stepping limb – expressed as AP overshoot – would moderate the additional mechanical energy input required to propel the COM forward and upward over the stance limb for the subsequent step (Donelan, Kram & Kuo, 2002). In the ML direction, a sinusoidal COM trajectory that approaches the medial border of the stepping foot would reduce the frontal plane gravitational moment about the supporting foot and the subsequent acceleration toward the forthcoming swing limb, which could modify the mechanical energy requirements for medially redirecting the COM during the successive step (Donelan, Kram & Kuo, 2001).

Alternatively, overshoots may serve as a strategy to simplify reactive control after foot contact, whereby a greater anterior or lateral progression of the COM could place emphasis on the stepping limb to achieve restabilisation. In theory, difficulties in COM control, resulting in a larger than expected forward or lateral COM excursion after foot contact, would only require an increase in applied force by the stepping limb and/or the initiation of an additional forward or lateral step to regain stability – typical features of stepping responses to anterior postural perturbation among older adults (McIlroy & Maki, 1996). In contrast, large AP or ML undershoots would necessitate the generation of additional propulsive force by the support limb, a subsequent posterior or medial step, or potentially a step with the support limb, which could be more challenging if increased interlimb interaction is required.

When stepping with non-preferred step placement, there was no change in the variability of incongruity across conditions nor was there any reduction in incongruity magnitude over repeated trials, as would be expected if these novel conditions presented challenges to dynamic stability control. Together, this evidence provides additional support for the notion that COM overshoot in this sample of healthy, young participants did not result from errors in anticipatory or reactive control. As inability to modify step placement would heighten the importance of reactive control on foot contact, these data may suggest that, during a single self-initiated step, foot placement and COM kinematics may be pre-planned and may not be modified online in response to the COM kinematics (Lyon & Day, 1997). This differs from steady-state gait, whereby lateral stability has been suggested to be regulated, in part, by variations in ML foot placement (Bauby & Kuo, 2000; Donelan, Shipman, Kram & Kuo, 2004). There is continued uncertainty, however, concerning the extent to which ML foot placement is used to preserve ML stability during reactively-generated forward stepping evoked by postural perturbation (McIlroy & Maki, 1996; Rogers et al., 2001).

While the magnitude of ML overshoot did increase in trials with increased step width, these differences did not persist when the incongruity was referenced to the BOS, which suggests that there may be an attempt to scale the peak ML COM displacement to step width. This supports the hypothesis that overshoot may be an attempt to take advantage of the passive dynamics, with implications for either gait initiation or dynamic stability control during the restabilisation phase of movement termination. With the challenges faced by older adults in the maintenance of ML dynamic stability (McIlroy & Maki, 1996; Lord et al., 1999; Rogers et al., 2001; Rogers & Mille, 2003), particular interest lies in understanding the role of ML overshoot in simplifying ML stability control during forward stepping. ML overshoot could be a means to simplify balance control under the specific task requirement that individuals terminate stepping with a single step, resulting in a forward stance configuration. Further insight into this matter will be gained by examining the relationship between the COM and BOS in older adults during a variety of stepping conditions, both volitional and reactive. Relative to younger adults, we propose that older adults will exhibit a larger peak COM displacement toward the stance limb and greater variability of incongruity, which may be brought about by difficulties in regulating the magnitude, direction or timing of applied force during the restabilisation phase. For such future studies, we suggest increasing the sample size to benefit the external validity of the results.

# CHAPTER 4

## Age-Related Differences in the Control of Mediolateral Dynamic Stability during the Restabilisation Phase of Volitional Stepping

## 4.1 OVERVIEW

Our previous work (Singer, Prentice & McIlroy, 2012) has suggested that COM incongruity during volitional stepping may arise as a means to simplify reactive control during the restabilisation phase. The purpose of the present study was to extend this work to examine age-related differences in the control of mediolateral dynamic stability during volitional stepping. Healthy young (n=20) and older (n=20) participants performed voluntarily-initiated single-step trials at their preferred speed and step placement. In attempt to reduce the influence of anticipatory control prior to step-onset, we included a condition of rapid-stepping, cued by an auditory tone. To further increase the challenge for mediolateral stability control, we also included a condition of cued rapid-stepping with narrow step width. The magnitude of incongruity between the peak and final COM position was quantified along with the intertrial variability of incongruity. To aid in the determination of the mechanisms responsible for age-related differences in the kinematic variables, the timing and magnitude of the waveform representing the difference between the inclination angles of the net ground reaction force and COP-COM were analysed. As with our previous work, the current results revealed that overshoots of the final COM position were common across all stepping conditions, however, COM incongruity was greater among older adults. Older adults also exhibited greater variability of incongruity. Overall, results from the present study suggest that while COM overshoot may aid in the simplification of reactive control, increased overshoot among older adults may arise from difficulty in reactive control during the restabilisation phase. We propose that COM incongruity during volitional stepping is likely governed by reactive control during the restabilisation phase, but can be offset to some degree by anticipatory postural control prior to step-onset, which serves to minimize ML instability until stepping foot-contact. More specifically, age-related differences in COM incongruity may be primarily linked to the time lag in active reorientation of the ground reaction force after foot-contact. Further study of COM incongruity and its underlying mechanisms during perturbation-evoked, reactive, stepping will be helpful in clarifying the role of the restabilisation phase in the absence of anticipatory control.

## 4.2 INTRODUCTION

Falls among older adults continue to present a major public health concern in Canada (Canadian Institute of Health Information, 2010). Despite substantial scientific study, our understanding of specific mechanisms related to the age-related increase in fall risk has been complicated by the both complexity of the balance control system and the multifactorial nature of falls. There is general consensus, however, that falls emerge from a complex interaction of variables pertaining to both the individual and the environment. Nevertheless, the occurrence of a fall ultimately results from an internal or external perturbation to the balance control system, coupled with subsequent inability of the individual to generate an effective response to restore static or dynamic stability (Maki & McIlroy, 1996).

Of particular importance for older adults is the control of mediolateral stability, as declines during both gait and compensatory stepping have been associated with fall risk in older adults (Maki et al., 1994; Lord et al., 1999; Brauer et al., 2000; Hilliard et al., 2008; Schrager et al., 2008). Moreover, falls in a lateral direction have been associated with an increased probability of hip fracture (Nevitt & Cummings, 1993; Hayes et al., 1996; Robinovitch et al., 2003), which can lead to reduced independence and higher risk of mortality. Transitions during movement or between movement states present a particular problem for stability control, as these tasks require the generation of appropriately directed, timed and scaled forces to move the COM outside the BOS (e.g. gait initiation, turning), to decelerate and return the COM to a position within the BOS (e.g. sit-to-stand, gait termination, compensatory stepping responses) or to regulate the position and velocity of the COM with respect to a changing BOS (e.g. step-to-step transitions during gait).

Research exploring age-related differences in the control of dynamic stability during stepping tasks, such as unplanned gait termination, have found that older adults tend to require an increased number forward steps to regain stability (Cao, Schultz, Ashton-Miller & Alexander, 1998; Tirosh & Sparrow, 2004). Stability modelling algorithms (Pai & Patton, 1997) employed by Tirosh and Sparrow (2004), along with subsequent electromyographic studies (Tirosh & Sparrow, 2005), have provided evidence to support the suggestion that the additional forward steps employed by older adults may actually aid in the preservation of mediolateral stability. Similar ideas have been put forth by researchers examining perturbation-evoked anterior stepping reactions, who have also reported lateral instability subsequent to foot contact (McIlroy & Maki, 1996).

Particularly interesting, and common to both voluntary and reactive stepping, is that lateral instability occurs subsequent to the initial sagittal plane instability. While the exact source of such instability is still somewhat unclear, the lack of age-related differences in initial temporospatial and kinematic movement parameters have led some researchers to propose that mediolateral instability may arise from events that occur during the swing phase or at foot-contact (McIlroy & Maki, 1996; Rogers et al., 2001; Rogers & Mille, 2003).

Notwithstanding this suggestion, few studies have focussed specifically on the control of mediolateral stability during the restabilisation phase, after foot-contact. O'Kane et al. (2003) have performed one such study, which examined the kinetic energy of the head-arms-trunk segment during planned gait termination in two patient populations, relative to healthy controls. Individuals with cerebellar damage or bilateral vestibular hypofunction (BVH) were noted to have an excessive conversion of forward to lateral kinetic energy. A study from the same laboratory, comparing steady-state gait in older adults with BVH to age-matched healthy controls, provides complementary information regarding mediolateral stability control (Kaya, Krebs & Riley, 1998). The authors noted that both groups reduced sagittal plane momentum by reducing forward gait velocity, but BVH patients had particular difficulty controlling frontal plane momentum during gait.

More recent research has continued to highlight the importance of the regulation of angular momentum by the central nervous system for stability control during walking. Specifically, dynamic stability appears to be enhanced by directing the resultant ground reaction force at or near the total body centre of mass, thereby minimizing changes in angular momentum (Kaya et al., 1998; Neptune & McGowan, 2011). Further, differences in the ability to regulate angular momentum have been suggested to have important implications with respect to dynamic stability and falls during walking (Simoneau & Krebs, 2000) and recovery from a trip (Pijnappels, Bobbert & van Dieen, 2004). While the majority of such studies have focussed on momentum and stability in the sagittal plane, age-related differences in mediolateral dynamic stability could arise from an altered ability to regulate frontal plane momentum via the direction of the resultant ground reaction force vector relative to the COM.

Our previous work (Singer, Prentice & McIlroy, 2012) aimed to characterise the trajectory of the COM during self-initiated single stepping within a sample of healthy young adults, to reveal the magnitude and trial-to-trial variability of incongruity between the peak and final COM position during the restabilisation phase. Contrary to our hypotheses, COM incongruity was apparent during trials in which step placement was self-selected. The magnitude of incongruity increased in trials with increased step width, while the variability of incongruity did not change across stepping conditions, which consisted of combinations of increased step length and width. Further, frontal plane incongruity was biased toward overshoots of the final COM position. Together, these results suggested that COM incongruity in this sample of young participants was unlikely the result of difficulty with anticipatory or reactive control, but may represent an attempt to simplify balance control on foot contact.

Despite the conclusions of this study, it remains to be determined exactly how these variables relate to age-related decline in stability control. From this, the purpose of the current study was to extend our previous work of voluntary stepping in young adults to address mediolateral stability control after foot contact in healthy older adults when there was the opportunity to pre-plan appropriate movement parameters for maximal stability. More specifically, we sought to quantify COM kinematics in a sample of healthy older adults to determine whether the magnitude and variability of COM incongruity were reduced, to provide an increased margin of safety, or increased, which may indicate a deficit in control. We operated under the premise that the greatest opportunity for mediolateral dynamic stability control during stepping occurs during double support – either by anticipatory control prior to step-onset or by reactive control during the restabilisation phase, after foot-contact. As such, we included a condition of cued, rapid stepping with self-selected step placement, which we believed would shift the burden of ML stability control to the restabilisation phase, by reducing the potential for anticipatory ML stabilisation during the step-initiation phase. In addition, we included a condition of cued, rapid stepping with narrow step width, which, in addition to placing emphasis on the restabilisation phase for ML stability control as previously described, we believed would pose a further challenge for ML restabilisation after foot-contact due to imposed constraints on BOS width. In light of previous research highlighting the importance of the regulation of angular momentum for stability control, we analysed the timing and magnitude of the waveform representing the difference between the inclination angles of the net ground reaction force and COP-COM, to aid in the clarification of the underlying mechanisms responsible for any age-related differences in the abovementioned kinematic variables.

Three specific hypotheses were tested in the present study:

1) Incongruity magnitude (Fig. 4-1):

a) We hypothesized that we would find no difference between younger and older adults in the magnitude of COM incongruity, when age-groups were compared at their preferred stepping speed. This was based on our previous work in young adults, which suggested that the bias toward overshoot of the final COM position was representative of a strategy to simplify COM control after foot contact. We believed this would arise because older adults would likely elongate the duration over which the step was executed (i.e. alter movement velocity), as a means to compensate for difficulties in controlling dynamic stability (Dingwell & Marin, 2006).

b) In contrast, during rapid speed stepping with preferred step placement, we believed that older adults would not utilize this simplification strategy, but would attempt to constrain the COM within a smaller area after foot contact. Specifically, we believed that, on average, older adults would exhibit an incongruity magnitude near zero (i.e. neither overshoot nor undershoot), which would be significantly less than their own preferred speed stepping trials and relative to younger adults performing rapid speed stepping. Under this condition we did not expect to find differences among younger adults, relative to their preferred stepping condition.

c) Lastly, we believed we would observe similar reductions in mediolateral COM incongruity among older adults during trials in which step width was reduced. Given our previous work, which suggested that overshoot was scaled with step width as a means to simplify reactive control, we believed younger adults would reduce incongruity magnitude in trials in which step width was reduced. Further, we believed that under such conditions, age-related differences would persist.

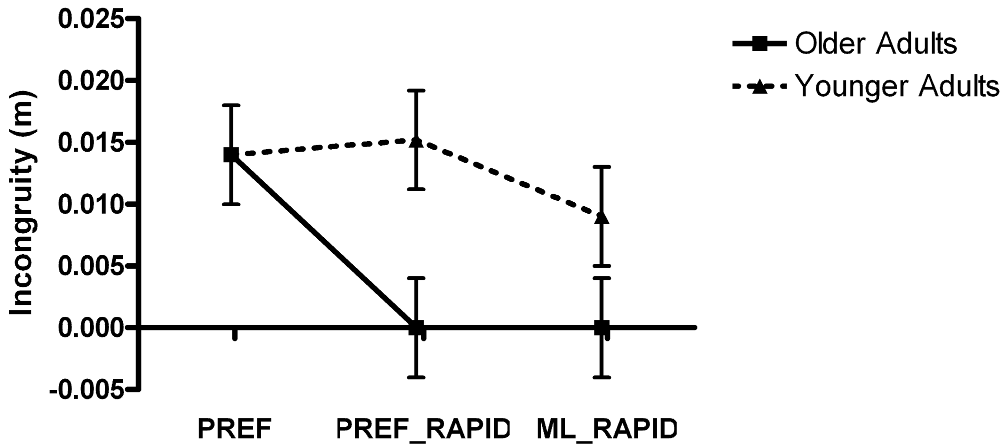


**Figure 4-1.** Hypothetical mediolateral incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Actual data from younger adults, for PREF and PRER_RAPID conditions, is from Singer, et al. (2012). Error bars represent the 95% confidence interval for the mean, observed in the abovementioned study. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

2) Trial-to-trial variability (Fig. 4-2):

a) As we believed that older adults would increase the time over which the step was executed, to offset difficulty with dynamic stability control during the restabilisation phase, we hypothesized that we would find no age-related differences in trial-to-trial variability at participants' preferred stepping speed.

b) In contrast, during rapid stepping, it was hypothesized that increased instability in older adults would be revealed by increased trial-to-trial variability of incongruity relative to their own preferred stepping trials and to younger adults performing rapid speed stepping.

c) Lastly, we anticipated that such age- and task-related differences hypothesized to occur for rapid stepping with preferred step width would be greater under conditions requiring a narrow step width.

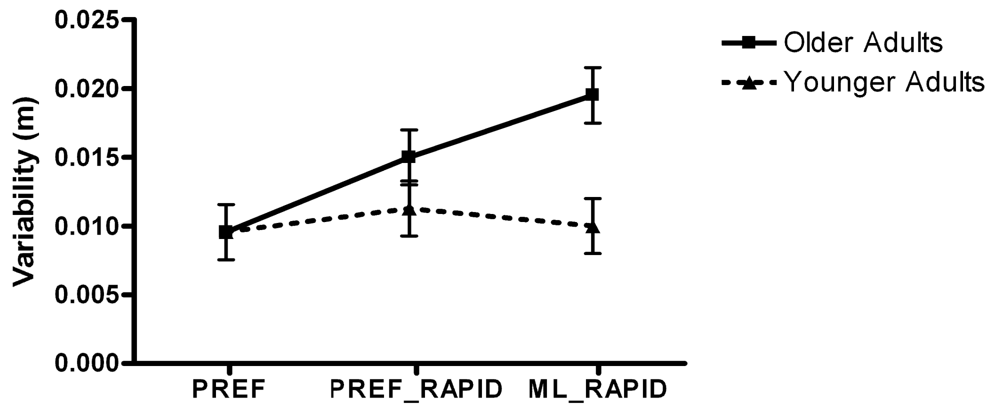


**Figure 4-2.** Hypothetical trial-to-trial variability of incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Actual data from younger adults, for PREF and PRER_RAPID conditions, is from Singer, et al. (2012). Error bars represent the 95% confidence interval for the mean, observed in the abovementioned study. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

3) Root-mean-square deviation (RMSD) of the GRF inclination angle relative to the COM (Fig. 4-3):

a) We hypothesized that, during preferred speed stepping, there would be no differences between age groups in the orientation of the frontal plane GRF vector with respect to the COM during the period from heel-contact to the point of restabilisation.

b) During rapid speed stepping, we believed that older adults would exhibit increased RMSD of the GRF inclination angle, when compared to their preferred speed stepping trials and to younger adults performing rapid speed stepping.

c) Lastly, we believed that that the abovementioned age- and task-related differences hypothesized to occur for rapid stepping with preferred step width would be greater when tested during conditions requiring a narrow step width.

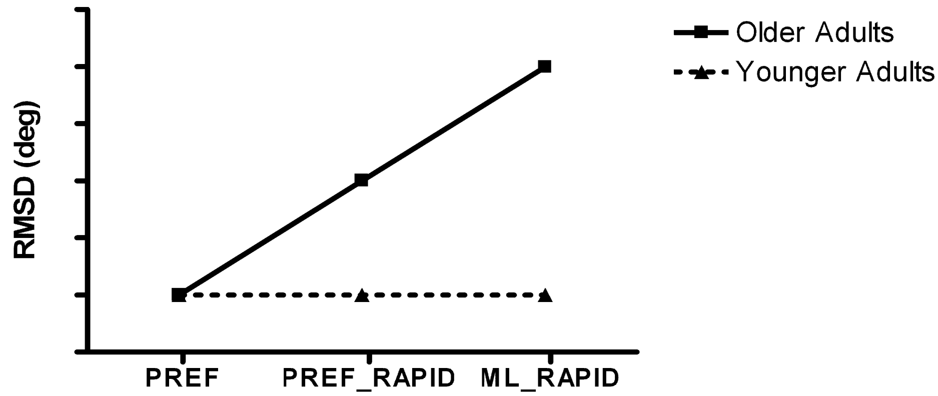


**Figure 4-3.** Hypothetical frontal plane RMSD of the net GRF inclination angle, with respect to the inclination angle of the line from COP to COM, for older (solid line) and younger adults (dotted line) adults, averaged within condition. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID). Please note that error bars and vertical axis scale are not shown, as this data is completely hypothetical and not based on previous work.

## 4.3 METHODS

### *4.3.1 Participants*

Twenty healthy young (age 24 (5) years) and twenty community-dwelling healthy older adults (age 71 (5) years) were recruited for inclusion within the study (10 male, 10 female per age-group). Participants were free from anatomical, neurological, or cognitive impairments. All participants were able to stand and walk unaided, had no previous history of falls and were not using psychoactive medications at the time of the study (specifically for pain control or management of mental or emotional disorder).

### *4.3.2 Instrumentation and Set-Up*

Six Vicon MX-3+ cameras (Vicon Motion Systems, Los Angeles, CA) were used to record kinematic data (100 Hz). Four force platforms (Advanced Mechanical Technology, Inc., Watertown, MA.), arranged in a rectangular array and embedded in the laboratory floor, were used to measure the reaction forces and moments (2000 Hz). In conjunction with a differential amplifier (Bortec Biomedical, Calgary, AB), disposable, self-adhesive Ag/AgCl electrodes were placed bilaterally on the tibialis anterior and soleus and were used to record electromyographic activity from these sites (2000 Hz). All motion capture data and analogue-to-digital converted signals were synchronously recorded by Vicon Nexus software (Vicon Motion Systems, Los Angeles, USA).

Retroreflective calibration markers, of 1 cm diameter, were placed on the lower limbs of the participant at anatomically relevant locations, in a similar fashion to the marker set used by the National Institute of Health and described by Hamill and Selbie (2004). To define segment endpoints for the pelvis, trunk, head and upper limbs, additional calibration markers were placed bilaterally on the iliac crests, anterior and posterior superior iliac spines, acromioclavicular joints, anterior to the external auditory meatus, greater and lesser tubercles of the humerus, medial and lateral epicondyles of the humerus, radial and ulnar styloid processes and the head of the $3^{rd}$ metacarpal. Rigid clusters containing four markers, placed on the sacrum and trunk, and bilaterally on the feet, legs and thighs were used to determine the 3D kinematics of each respective segment during the experimental trials. Positive x-,y- and z-axes for the laboratory coordinate system were oriented laterally to the right side of the participant, anteriorly and upward, respectively.

### *4.3.3 Protocol*

A standing reference trial was collected prior to the collection of the experimental trials. Participants were asked to stand in a neutral position, roughly aligned to the laboratory coordinate system. These data were used to determine the segment endpoints, segment embedded local coordinate systems, as well as the transformation matrices between the local and global coordinate systems for each segment. Following collection of the standing reference trial, markers used solely for calibration were removed.

Participants were asked to stand with their feet side-by-side on separate force platforms. Foot position was standardized across all participants to a width of 0.17 m, with an offset of 14 degrees between the longitudinal axis of the foot and the anteroposterior axis of the laboratory coordinate system, as per McIlroy and Maki (1997). Adhesive foam weatherstripping was placed on the force platform, along the medial and posterior borders of each foot, to ensure consistent foot placement between trials. Two quiet standing trials, of 60-second duration, were collected to obtain kinematic and electromyographic variables necessary for post-processing. The first trial was collected with the feet in the abovementioned standardized side-by-side configuration. For the second trial, the participant was asked to start with their feet in the standardized position, take a single step forward and remain in this forward-stance configuration for the duration of the trial.

Participants took part in three different task conditions, which required a single voluntary step with the preferred leg. Ten experimental trials were performed within each experimental condition.

It was possible that changes in incongruity magnitude that may occur with task familiarity – and the potential for a differential effect by age – may help to inform us about the role or cause of such incongruity. To address this, the ten experimental trials in each condition were split into two blocks of five trials. The effect of 'time' was incorporated into the omnibus ANOVA model used to address each dependent variable, as described below. The order of all blocks of trials was randomized across subjects in each group.

The following experimental conditions were performed:

1. **Preferred** step length/width and speed (PREF);
2. **Preferred** step length/width, **rapid** speed stepping (PREF_RAPID);
3. **Reduced** step width, **rapid** speed stepping (preferred step length) (ML_RAPID);

Participants were asked to begin each trial by standing with their feet side-by-side on separate force platforms, using the abovementioned standardized foot position. To ensure symmetrical mediolateral weight distribution, the vertical force under each foot was monitored before cue onset. In addition, the anterior position of the COP under each foot was monitored, while the bilateral electromyographic activity of the soleus and tibialis anterior was monitored and recorded to ensure minimal pre-perturbation activity. As a reference, we attempted to ensure that the level of activity observed during the experimental trials did not exceed that which was recorded during quiet upright standing. During the PREF condition, participants were presented with an auditory tone, which signified the beginning of the trial. At any point thereafter, they were to initiate a single step with their preferred leg at a self-selected speed and, upon landing, come to rest and remain in the final, stable position until the end of the trial. Participants were asked to remain in this position for approximately ten seconds after completing the step, to allow sufficient data for post-processing. In the remaining two conditions (PREF_RAPID and ML_RAPID), participants were to minimize the time between the presentation of the auditory tone and step-initiation in addition to executing the step as rapidly as possible. Additionally, during the ML_RAPID condition, participants were asked to step onto a length of adhesive tape placed on the force platform, 5 cm lateral (toward the stepping limb side) to the median plane. To counter the possibility of participants anticipating the timing of the auditory command, the intervals at which the auditory cue was presented were varied.

### *4.3.4 Data Analysis*

Force platform data were lowpass filtered using a zero-lag, fourth-order, Butterworth filter with a cut-off frequency of 15 Hz. As the Butterworth filter is underdamped, there was a possibility using such a filter could induce artefact in individual marker trajectories during rapid transitions, which could lead to artefact in the computation of the total body COM position. Marker data were lowpass filtered using a zero-lag, twentieth-order critically damped filter, with a cut-off frequency of 6Hz, which should prevent marker over/undershoot as a result of filter artefact and also provide an equivalent roll-off to that of the Butterworth filter (Robertson & Dowling, 2003). A linear envelope (LE-EMG) was calculated from raw EMG data, after initial removal of dc-bias, full-wave rectification and lowpass filtering using a second-order Butterworth filter, with a cut-off frequency of 3 Hz.

The entire body was modelled as a rigid system of independently tracked segments. For the older adults, segment masses were estimated using Dempster's segment parameters and segment centre of mass positions were estimated using the geometrical model proposed by Hanavan (1964) (cited in Robertson et al., 2004). Segments of the appendicular skeleton were modelled as conical frusta; the pelvis and trunk were modelled as elliptical cylinders and the head as a sphere. As Dempster's segment parameters are not representative of younger adults, segment masses and centres of mass were estimated for this group using segmental mass proportions and relative segmental centre of mass locations from a sample of young, male and female undergraduate students, initially reported by Zatsiorsky, Seluyanov and Chugunova (1990) and subsequently modified by de Leva (1996). The trunk segment was modelled as a hybrid of the upper- and middle-trunk segments, defined by de Leva (1996). The proximal and distal endpoints for the trunk segment for the present study were the iliac crests and acromia, respectively, and the lengths of upper and middle trunk segments for each subject, as defined by de Leva (1996), were not known. Average segment lengths of a sample of males and females from Zatsiorsky et al. (1990) (as cited in de Leva, 1996), were used to calculate the lengths of the upper- and middle-trunk segments, as proportions of the total trunk length. These scale factors, in conjunction with the relative segmental COM locations and measured total trunk length (from marker coordinates), were used to compute the hybrid trunk COM location, which was assumed to lie on the longitudinal axis running from the midpoint between iliac crest and acromial markers, respectively. The total body centre of mass position was calculated as a weighted average of all body segments, where each segment was weighted according to its mass proportion.

The COM restabilisation point was defined as the time point at which the COM velocity waveform entered and remained within an amplitude bandwidth bordered by +/- two standard deviations of the mean velocity during the initial quiet standing trial with forward-stance configuration. Velocity, rather than position, was chosen for determination of the COM restabilisation point, as this allowed for calculations that were less affected by low frequency oscillations or drift of the COM position-time waveform after restabilisation.

COM incongruity was defined as the difference between the local maximum COM position after foot contact and the mean of the stable region of the waveform, which was a two second window beginning at the restabilisation point. Overshoot was defined as positive incongruity; undershoot was defined negative incongruity. Trial-to-trial variability was assessed by computing the standard deviation of the mean COM incongruity over each block of 5 repeated experimental trials for each condition.

The ground reaction forces from all force platforms were combined to yield a single force vector. The frontal plane inclination angle of this net force was calculated, with respect to the x-axis (ML) of the global coordinate system. The frontal plane inclination angle of the line joining the COP and COM was also determined, relative to the x-axis (ML) of the global coordinate system. In theory, if the inclination angle of the GRF is equal to the inclination angle of the COP-COM, there should be no external moment about the COM caused by the net GRF and, hence, no change in angular momentum (Fig. 4-4). To provide a single metric for characterization of this compound kinetic variable, the RMSD of the frontal plane GRF inclination angle with respect to the frontal plane COP-COM inclination angle was computed, for the interval from foot-contact to the restabilisation point.

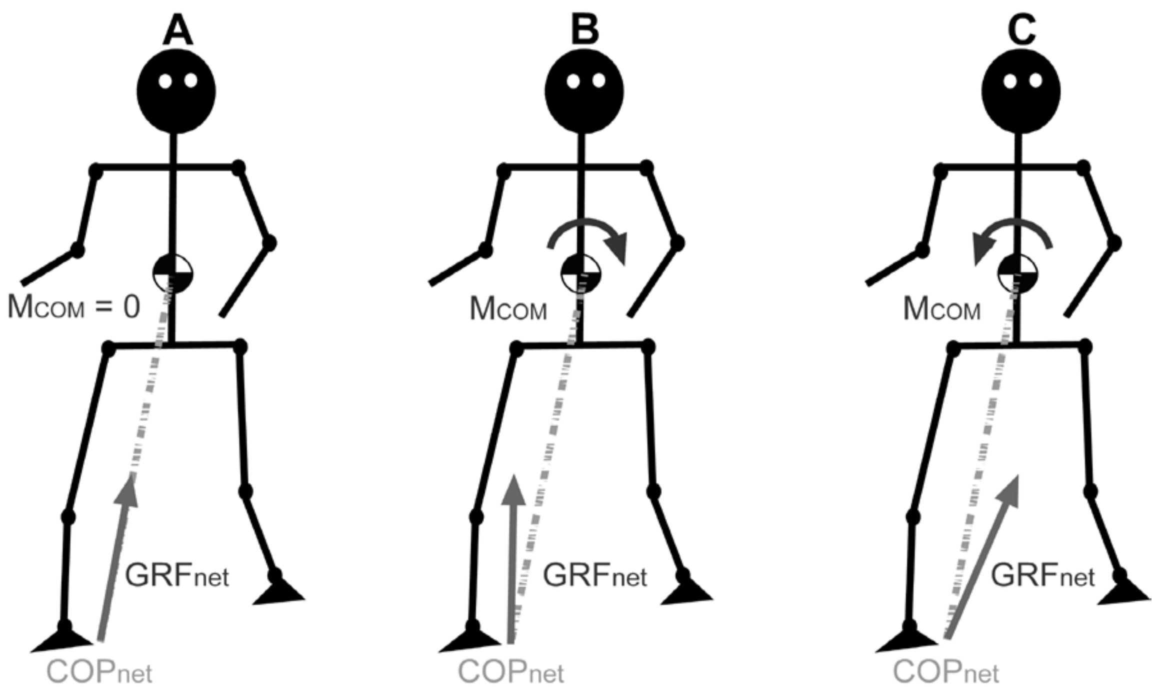


**Figure 4-4.** Orientation of the net ground reaction force with respect to the centre of mass during the restabilisation phase (double support), after foot-contact. Panel A: the inclination angle of the net ground reaction force (arrow) is coincident with the COP-COM inclination angle (dashed line), resulting in zero external moment about the centre of mass; Panel B: the net ground reaction force is oriented such that a clockwise external moment results, in this case toward the stance limb; Panel C: the net ground reaction force is oriented such that an anticlockwise moment results, in this case toward the stepping limb

### 4.3.5 Statistical Analyses

The statistical analyses are described in reference to each of the three hypotheses:

1. COM incongruity values for each trial were averaged within-subject for each block of 5 trials and were subsequently analysed with a three-factor ANOVA, with repeated measures [2 within group factors (step – 3 levels)(time – 2 levels); 1 between group factor (age – 2 levels)]. Sphericity was evaluated using Mauchly's test. Violations of the sphericity assumption were corrected using the Greenhouse-Geisser method, unless otherwise indicated. If the omnibus ANOVA revealed a significant main effect of step or significant interaction effects, follow-up independent and paired samples t-tests were used, as appropriate, to localize the differences. Measures of effect size were estimated using the formula for Cohen's d, using the means and standard deviations for each group. To counter the effect of alpha-inflation due to multiple subsequent comparisons, we employed a Bonferroni correction whereby the alpha level for a family of follow-up analyses was divided by the number of comparisons within that family.

2. Standard deviations, representing within-subject trial-to-trial variability, were computed for each subject in each age group and were analysed with a three-factor ANOVA, with repeated measures. Sphericity was evaluated using Mauchly's test. Violations of the sphericity assumption were corrected using the Greenhouse-Geisser method, unless otherwise indicated. Main effects of step or interaction effects were analysed with Bonferroni-corrected independent and

paired samples t-tests. When appropriate, measures of effect size were calculated using the formula for Cohen's d, as above.

3. RMSD values were averaged within-subject and were analysed with a three-factor ANOVA, with repeated measures. Sphericity was evaluated using Mauchly's test. Violations of the sphericity assumption were corrected using the Greenhouse-Geisser method, unless otherwise indicated. Main effects of step or interaction effects were analysed with Bonferroni-corrected independent and paired samples t-tests. When appropriate, measures of effect size were calculated using the formula for Cohen's d, as above.

### *4.3.6 Secondary Analyses*

It remained possible that differences in the initial temporospatial parameters of the stepping response between young and older adults may have existed and may have partially explained any potential differences in the variables of interest. To expose this possibility, secondary analyses were performed to assess age-related differences in the AP COP position with respect to the vertical projection of the COM during the interval before cue presentation, ML APA amplitude, lateral COM displacement prior to step-onset, onset of ML asymmetry and unloading, step length, step width, step time, peak AP swing foot velocity, AP and ML COM velocity at foot-contact, and the position of the COM with respect to the lateral aspect of the BOS at the instant of foot-contact.

Additionally, as noted previously, while we hypothesized that the direction of the resultant ground reaction force exerts a large influence on an individual's ability to regain stability after foot contact, the timing of such force application may be of equal importance. The timing of key time points of the waveform representing the difference between the inclination angles of the COP-COM and the net GRF vector were analyzed, in addition to the magnitude at these time points.

## 4.4 RESULTS

### *4.4.1 Incongruity Magnitude – Effect of age, step condition, and trial repetition*

A main effect of age indicated that older participants had a greater magnitude of incongruity than the younger participants, $F(1,38) = 4.87$, $p = 0.033$, $\eta_p^2 = 0.11$. There was also a main effect of step condition, $F(2,76) = 4.24$, $p = 0.018$, $\eta_p^2 = 0.10$. Follow-up paired-samples t-tests indicated that there was no difference between PREF_RAPID and ML_RAPID trials, while there was greater COM incongruity in PREF_RAPID relative to PREF trials, $t(39) = 2.83$, $p = 0.007$, d = 0.45 (Fig. 4-5).

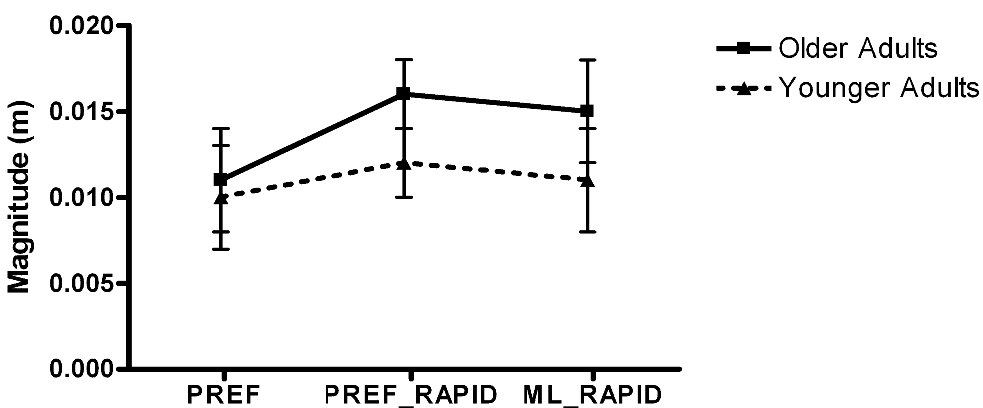


**Figure 4-5.** Mediolateral incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Positive values indicate overshoot. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

### *4.4.2 Trial-to-Trial Variability of Incongruity - Effect of age, step condition, and trial repetition*

A main effect of age indicated that older adults exhibited greater trial-to-trial variability of incongruity relative to the young, $F(1,38) = 12.11$, $p = 0.001$, $\eta_p^2 = 0.24$. There was also a main effect of step condition, $F(1.71,64.94) = 3.67$, $p = 0.030$, $\eta_p^2 = 0.09$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that there was no difference between PREF_RAPID and ML_RAPID trials, while there was greater trial-to-trial variability of incongruity in PREF_RAPID relative to PREF trials, $t(39) = 3.24$, = 0.002, $d = 0.61$ (Fig 4-6).

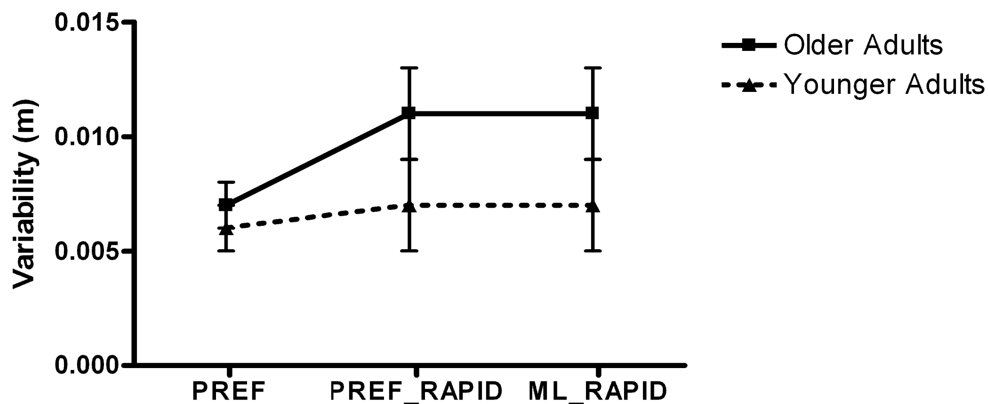


**Figure 4-6.** Trial-to-trial variability of mediolateral incongruity for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

### *4.4.3 RMSD of the Ground Reaction Force Inclination Angle with Respect to the COM – Effect of age, step condition, and trial repetition*

A main effect of age revealed that young participants had increased RMSD relative to the older participants, $F(1,38) = 7.12$, $p = 0.011$, $\eta_p^2 = 0.16$, which was qualified by an interaction between age and time, $F(1,38) = 6.81$, $p = 0.013$, $\eta_p^2 = 0.15$ (Fig. 4-7). Follow-up independent samples t-tests revealed that younger adults had greater RMSD than the older adults during the first block of trials, $t(38) = 2.94$, $p = 0.006$, $d = 0.93$. While this trend persisted during the second block of trials, the difference was not significant at the alpha level for follow-up analyses, $t(38) = 2.15$, $p = 0.038$, $d = 0.68$. Paired-samples t-tests did not reveal alterations with practice within older adult group. While there was a reduction in the RMSD within the younger adults group during the second block of trials, relative to the first, this difference was not significant at the alpha level for follow-up analyses, $t(19) = 2.58$, $p = 0.019$, $d = 0.29$. The omnibus ANOVA revealed a main effect of step condition, $F(1.48,56.23) = 31.03$, $p < 0.001$, $\eta_p^2 = 0.45$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that there was no difference between PREF_RAPID and ML_RAPID conditions, while RMSD was greater in the PREF_RAPID condition than in the PREF condition, $t(39) = 7.33$, $p < 0.001$, $d = 0.94$ (Fig. 4-8).

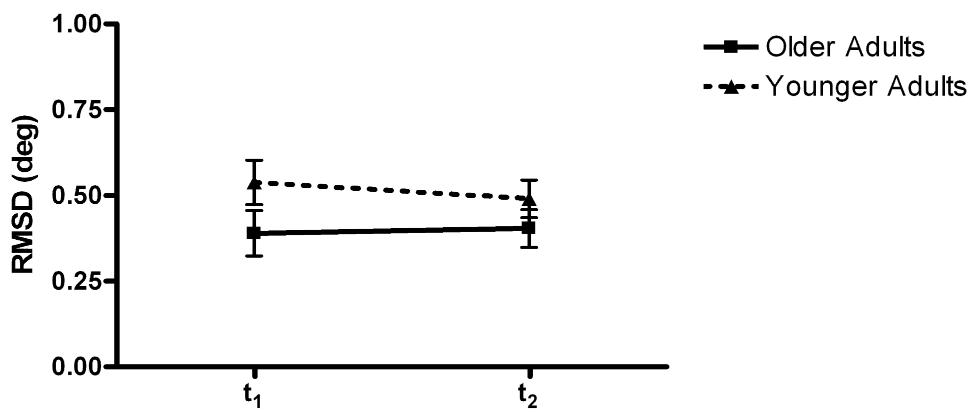


**Figure 4-7.** Root-mean-square deviation of the inclination angle of the net ground reaction force with respect to the inclination angle formed by a line joining the centre of pressure and centre of mass, averaged within blocks of repeated trials. Data are depicted for older (solid line) and younger adults (dotted line), during the first ($t_1$) and second ($t_2$) time blocks. Error bars represent the 95% confidence interval for the mean.

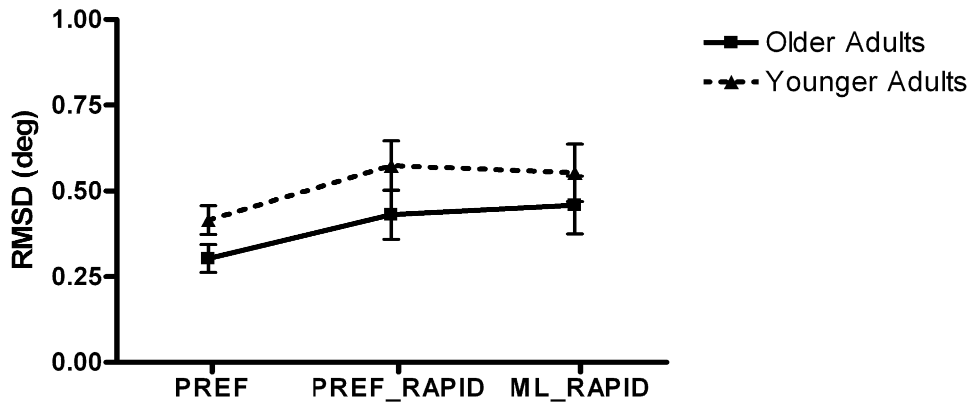


**Figure 4-8.** Root-mean-square deviation of the inclination angle of the net ground reaction force with respect to the inclination angle formed by a line joining the centre of pressure and centre of mass. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

### *4.4.4 Secondary Analyses*

#### *4.4.4.1 Initial Conditions*

Analysis of the AP distance between the vertical projection of the COM and the position of the COP during the pre-cue interval revealed main effects of age, $F(1,38) = 22.92$, $p < 0.001$, $\eta_p^2 = 0.39$, and step condition, $F(1.42,54.06) = 4.53$, $p = 0.026$, $\eta_p^2 = 0.11$ (sphericity not assumed), which were qualified by an interaction between age and step condition, $F(1.41,54.06) = 7.72$, $p = 0.003$, $\eta_p^2 = 0.17$ (sphericity not assumed). Follow-up independent-samples t-tests provided information consistent with the main effect of age. Specifically, the older group exhibited a more posterior COM position relative to the COP for the PREF_RAPID, $t(38) = -5.13$, $p < 0.001$, $d = 1.62$, ML_RAPID, $t(38) = -5.10$, $p < 0.001$, $d = 1.61$, and PREF conditions, $t(28.93) = -3.91$, $p = 0.001$, $d = 1.31$ (equal variances not assumed). Paired-samples t-tests, performed separately for each age group, did not reveal a statistically significant differential trend by age group, across step conditions.

*4.4.4.2 Initial Movement Parameters*

When examining the mediolateral anticipatory postural adjustment (ML APA), there was no significant main effect of age. There was, however, a main effect of step condition, $F(1.45,54.94) = 109.98$, $p < 0.001$, $\eta_p^2 = 0.74$. Follow up paired-samples t-tests indicated that the ML APA was larger in the ML_RAPID condition relative to the PREF_RAPID, $t(39) = 6.34$, $p < 0.001$, $d = 0.57$, and PREF conditions, $t(39) = 12.72$, $p < 0.001$, $d = 2.41$, and in the PREF_RAPID condition relative to the PREF condition, $t(39) = 9.20$, $p < 0.001$, $d = 1.82$. There was also a main effect of time, as there was a larger ML APA amplitude in the second block of trials relative to the first, $F(1,36) = 9.81$, $p = 0.003$, $\eta_p^2 = .21$.

Despite the lack of a significant main effect of age on the ML APA amplitude, there was a main effect of age on the influence of the ML APA on the peak lateral COM displacement of the COM toward the stance limb, $F(1,38) = 4.17$, $p = 0.048$, $\eta_p^2 = 0.10$, as the older participants exhibited a greater lateral COM displacement. Additionally, there was a main effect of step condition, $F(1.49,56.85) = 125.73$, $p < 0.001$, $\eta_p^2 = 0.77$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that the lateral deviation was greatest in the PREF condition relative to the PREF_RAPID, $t(39) = 13.42$, $p < 0.001$, $d = 2.85$, and ML_RAPID conditions, $t(39) = 9.40$, $p < 0.001$, $d = 1.91$, and also in the ML_RAPID condition relative to the PREF_RAPID condition, $t(39) = 7.51$, $p < 0.001$, $d = 1.12$ (Fig. 4-9).

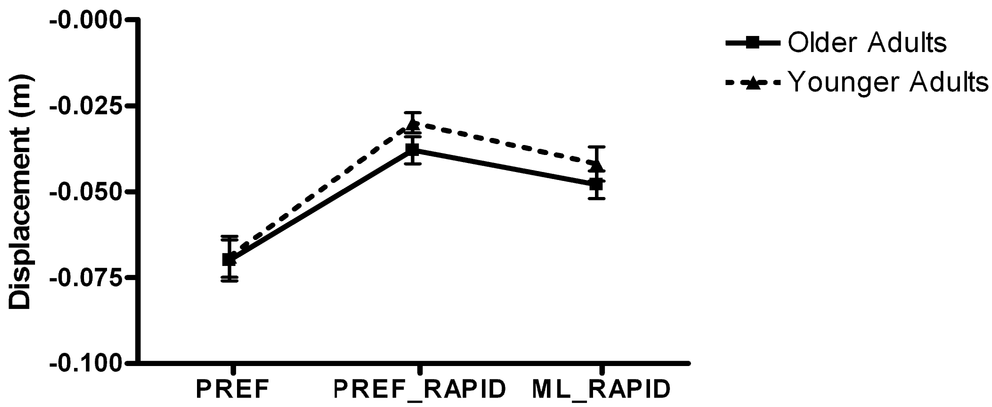


**Figure 4-9.** Peak mediolateral displacement of the centre of mass, referenced to the initial COM position prior to movement initiation. Negative values indicate displacement along the negative x-axis of the global coordinate system, toward the stance limb. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

Consistent with the task requirements, there was a main effect of step condition when examining the time to onset of ML asymmetry, $F(1.04,39.62) = 25.45$, $p < 0.001$, $\eta_p^2 = 0.40$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that there was no difference between PREF_RAPID and ML_RAPID conditions, while the time to onset of ML asymmetry was longer in the PREF condition relative to both the PREF_RAPID and ML_RAPID conditions, $t(39) = 5.04$, $p < 0.001$, $d = 1.07$ (t-values identical for both comparisons). This main effect of step condition persisted when examining the time to onset of ML unloading, $F(1.16,43.81) = 57.26$, $p < 0.001$, $\eta_p^2 = 0.60$ (sphericity not assumed). Follow-up paired-samples t-tests revealed that there was no difference between PREF_RAPID and ML_RAPID conditions, while the time to onset of ML unloading was greater in the PREF condition than in either the PREF_RAPID or ML_RAPID conditions, $t(39) = 7.87$, $p < 0.001$, $d = 1.70$, $t(39) = 7.18$, $p < 0.001$, $d = 1.62$, respectively.

To confirm that both age groups were performing the stepping conditions in accordance with the requirement that step width be reduced in the ML_RAPID condition, we performed an analysis of both step length and step width. As expected, there were no main effects of age or interactions when examining step length. There was a main effect of step condition, $F(1.43,54.48) = 6.23$, $p = 0.003$, $\eta_p^2 = 0.14$ (sphericity not assumed). Follow-up paired-samples t-tests revealed no differences when comparing PREF_RAPID to ML_RAPID or PREF conditions. Step length in the ML_RAPID condition was, however, greater than that in the PREF condition, $t(39) = 2.98$, $p = 0.005$, $d = 0.46$. Consistent with the task conditions, there was only a main effect of step condition when examining step width, $F(2,76) = 52.94$, $p < 0.001$, $\eta_p^2 = 0.58$. Follow-up paired-samples t-tests indicated that step width was reduced in the ML_RAPID condition, as compared to either the PREF_RAPID, $t(39) = -9.29$, $p < 0.001$, $d = 2.13$, or PREF conditions, $t(39) = -5.93$, $p < 0.001$, $d = 1.36$. Interestingly, step width was greater in the PREF_RAPID condition, as compared to the PREF condition, $t(39) = 4.25$, $p < 0.001$, $d = 0.63$. There was a main effect of time, $F(1,38) = 4.67$, $p = 0.037$, $\eta_p^2 = 0.11$, whereby individuals further reduced step width in the second block of trials, irrespective of step condition. Analysis of the swing foot path length did not reveal significant main effects of age or interactions. There was a main effect of step condition, $F(1.43,54.29) = 9.70$, $p = 0.001$, $\eta_p^2 = 0.20$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that swing foot path length was greater in the ML_RAPID condition than either the PREF_RAPID, $t(39) = 3.31$, $p = 0.002$, $d = 0.26$, or PREF conditions, $t(39) = 3.72$, $p = 0.001$, $d = 0.58$.

To evaluate if individuals were performing the tasks in accordance with the instructions to step as rapidly as possible, we evaluated the peak anteroposterior (AP) swing foot velocity during the swing phase. This analysis revealed main effects of age, $F(1,38) = 16.34$, $p < 0.001$, $\eta_p^2 = 0.30$, and of step condition, $F(1.48,56.24) = 259.37$, $p < 0.001$, $\eta_p^2 = 0.87$ (sphericity not assumed), which were qualified by an interaction between age and step condition, $F(1.48,56.24) = 4.96$, $p = 0.02$, $\eta_p^2 = .12$ (sphericity not assumed). In accordance with the task demands, follow-up paired-samples t-tests revealed that both the older and younger adults increased the anteroposterior swing foot velocity in the PREF_RAPID and ML_RAPID conditions relative to the PREF condition, $t(19) = 14.53$, $p < 0.001$, $d = 2.60$, $t(19) = 14.12$, $p < 0.001$, $d = 2.75$ (older group), $t(19) = 10.57$, $p < 0.001$, $d = 2.63$, $t(19) = 12.36$, $p < 0.001$, $d = 2.97$ (younger group), respectively. While the younger adults further increased swing limb velocity in the ML_RAPID condition, as compared to the PREF_RAPID condition, $t(19) = 3.04$, $p = 0.007$, $d = 0.35$, there was no significant difference between these two conditions among the older adults. Further, follow-up independent-samples t-tests indicated that the younger group had greater anteroposterior swing foot velocity during both PREF_RAPID, $t(38) = 3.61$, $p = 0.001$, $d = 1.14$, and ML_RAPID conditions, $t(38) = 4.52$, $p < 0.001$, $d = 1.43$. Differences between age groups for the PREF condition, however, were not significant at the post hoc alpha level, $t(38) = 2.26$, $p = 0.030$, $d = 0.71$. Interestingly, the main effect of age in AP swing limb velocity did not persist when analysing swing time. There was a main effect of step condition, $F(1.29,49.11) = 187.84$, $p < 0.001$, $\eta_p^2 = 0.83$ (sphericity not assumed). Follow-up paired-samples t-tests revealed no difference between PREF_RAPID and ML_RAPID conditions, while swing time was reduced in both PREF_RAPID, $t(39) = -$

14.30, $p < 0.001$, $d = 2.87$, and ML_RAPID conditions, $t(39) = -14.74$, $p < 0.001$, $d = 2.85$, relative to the PREF condition.

Given the age-related differences in the peak lateral deviation of the COM as a result of the ML APA, we performed an analysis of the AP and ML COM velocity at foot-contact to determine if these differences persisted after the swing phase. Analysis of the AP COM velocity at foot-contact revealed a main effect of age, $F(1,38) = 15.22$, $p < 0.001$, $\eta_p^2 = 0.29$, as the older adults had a reduced AP COM velocity across all step conditions. There was also a main effect of step condition, $F(1.19,45.41) = 26.19$, $p < 0.001$, $\eta_p^2 = 0.41$. Follow-up paired-samples t-tests indicated that there was no significant difference between PREF_RAPID and ML_RAPID conditions, while AP COM velocity was greater in both the PREF_RAPID and ML_RAPID conditions relative to the PREF condition, $t(39) = 5.36$, $p < 0.001$, $d = 0.88$, and $t(39) = 5.33$, $p < 0.001$, $d = 0.85$, respectively. The omnibus ANOVA for the analysis of the ML COM velocity at foot-contact revealed a main effect of age, $F(1,38) = 4.58$, $p = 0.039$, $\eta_p^2 = 0.11$, with older adults having a reduced magnitude. There was a main effect of time, $F(1,38) = 16.66$, $p < 0.001$, $\eta_p^2 = 0.29$, revealing a reduction in the second block of trials. Lastly, there was a main effect of step condition, $F(1.72,65.32) = 48.52$, $p < 0.001$, $\eta_p^2 = 0.56$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that the ML COM velocity at foot-contact was reduced in the ML_RAPID condition relative to both the PREF_RAPID, $t(39) = -8.10$, $p < 0.001$, $d = 1.12$, and PREF conditions, $t(39) = -7.97$, $p < 0.001$, $d = 1.66$. There was also reduced ML COM velocity at foot-contact within the PREF_RAPID condition relative to the PREF condition, $t(39) = 2.63$, $p = 0.012$, $d = 0.42$ (Fig. 4-10).

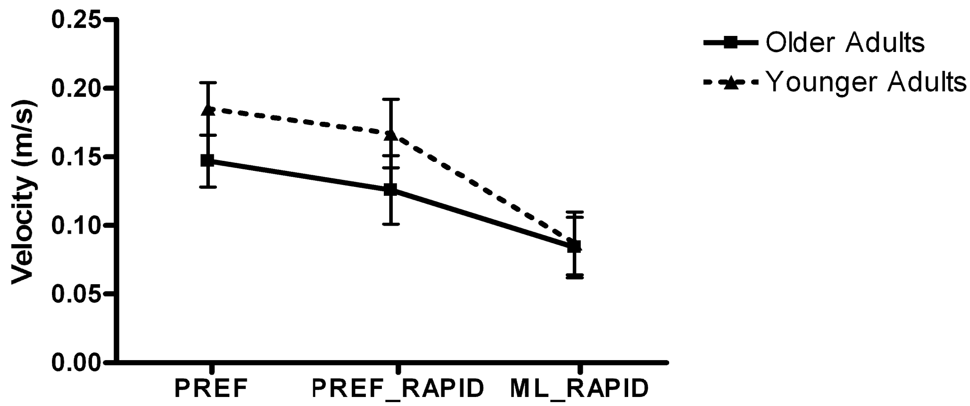


**Figure 4-10.** Mediolateral instantaneous velocity of the centre of mass at the time of foot-contact. Positive values indicate velocity consistent with the positive x-axis of the global coordinate system, toward the swing limb. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

With the abovementioned age-related differences during both the step-initiation and stepping phases, it was necessary to put the age-related differences in COM incongruity in the context of the ML position of the COM with respect to the lateral border of the BOS (i.e. the stepping limb) at the instant of foot-contact. Analysis revealed a main effect of age, $F(1,38) = 5.69$, $p = 0.022$, $\eta_p^2 = 0.13$, as older adults had a greater mediolateral distance between the COM and stepping foot. There was also a main effect of step condition, $F(2,76) = 85.33$, $p < 0.001$, $\eta_p^2 = 0.69$. Follow up paired samples t-tests indicated that the distance between the COM and stepping foot was greatest in the PREF condition relative to the PREF_RAPID condition, $t(39) = 4.02$, $p < 0.001$, $d = 0.67$, and in the PREF_RAPID condition relative to the ML_RAPID condition, $t(39) = 8.45$, $p < 0.001$, $d = 1.46$ (Fig 4-11).

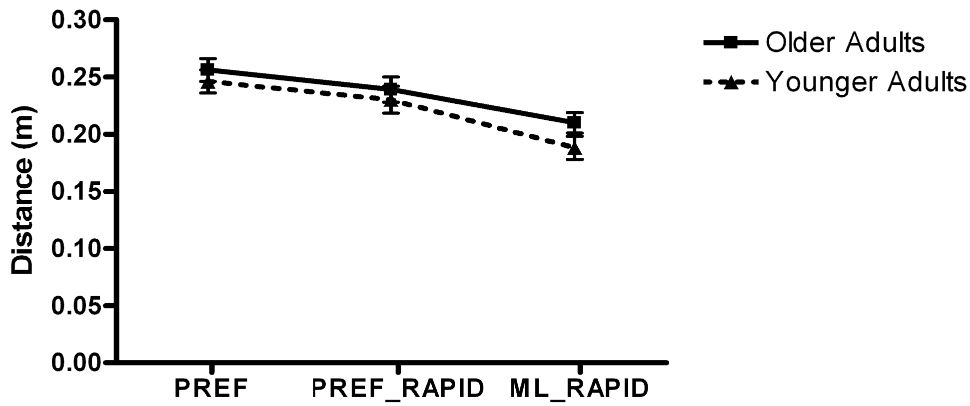


**Figure 4-11.** Mediolateral distance between the centre of mass and lateral aspect of base of support at the instant of foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

*4.4.4.3 Kinetics*

As the examination of the RMSD of the GRF inclination angle relative to the COP-COM inclination angle revealed an age-related difference in a direction opposite to our hypotheses, we analysed the mean and variability in the amplitude and timing (with respect to foot-contact) of what were believed to be three key peaks in the waveform representing the divergence between the GRF inclination angle and the COP-COM inclination angle (Fig 4-12).

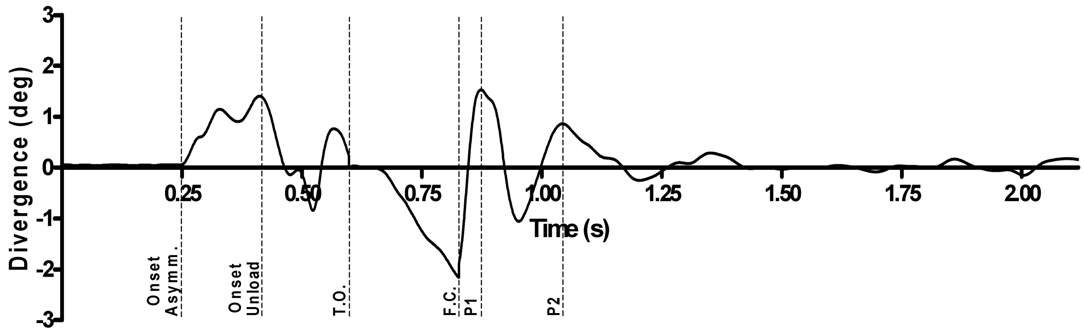


**Figure 4-12.** Representative trial from one subject (right-footed stepping), displaying the divergence between the GRF inclination angle and the COP-COM inclination angle. Positive values indicate a net GRF orientation that would tend to cause angular acceleration toward the stance limb side. Data begin at cue onset and end at the ML restabilisation point. ML Asymm. = onset of ML asymmetry; Onset Unload = onset of swing limb unloading; T.O. = toe-off; F.C. = foot-contact; P1 = peak of the first positive (passive) phase, immediately following foot-contact; P2 = peak of the second positive (active) phase, following foot-contact.

Examination of this waveform revealed a consistent pattern with three distinct peaks at, and after, foot contact. We proposed that the first negative peak at foot contact signifies a net GRF vector orientation that would result in an external moment and angular acceleration toward the swing limb (right-footed stepping: positive moment, anticlockwise about an anterior axis with origin at the whole body COM; left-footed stepping: negative moment, clockwise about an anterior axis with origin at the whole body COM); the magnitude of this peak is likely influenced the by the anticipatory postural adjustments and swing phase dynamics. The subsequent two positive peaks (P1 and P2) denote a net GRF orientation that would result in an external moment and angular acceleration toward the side of the stance limb (right-footed stepping: negative moment, clockwise about an anterior axis with origin at the whole body COM; left-footed stepping: positive moment, anticlockwise about an anterior axis with origin at the whole body COM), thereby countering the angular momentum that would develop during the swing phase. We propose the magnitude and timing of the first positive peak (P1) to be a consequence of stepping limb stiffness on foot-contact, since reduced limb compliance would increase the rate of axial loading and would more rapidly increase the ratio between the axial and mediolateral GRF components, thereby resulting in an earlier peak. We also propose that stepping limb stiffness and, hence, the timing of this peak, are pre-planned parameters, since the average time to peak (0.061 s) is too brief to be modulated by afferent sensory information upon foot-contact. The magnitude and timing of the second positive peak is, however, likely modulated by sensory information regarding the state of the COM at the onset, and throughout, the restabilisation phase.

Analysis of the amplitude of first negative peak, at heel-contact revealed main effects of age, $F(1,38) = 9.01$, $p = 0.005$, $\eta_p^2 = 0.19$ – the magnitude was reduced (less negative) among older adults. There was also a main effect of step condition, $F(1.48, 56.20) = 29.47$, $p < 0.001$, $\eta_p^2 = 0.44$ (sphericity not assumed). Follow-up paired-samples t-tests revealed that the amplitude at heel-contact was larger (more negative) in the PREF_RAPID condition than the ML_RAPID, $t(39) = 4.32$, $p < 0.001$, $d = 0.48$, or PREF conditions, $t(39) = 7.87$, $p < 0.001$, $d = 1.08$, and was also larger in the ML_RAPID condition than in the PREF condition, $t(39) = 3.30$, $p = 0.002$, $d = 0.62$ (Fig. 4-13). Examination of the variability revealed no main effects of age. There was, however, a main effect of step condition, $F(2,76) = 14.81$, $p < 0.001$, $\eta_p^2 = 0.28$. Follow-up paired-samples t-tests indicated that there was no significant difference between PREF_RAPID and ML_RAPID conditions, while trial-to-trial variability was greater in both PREF_RAPID and ML_RAPID conditions than the PREF condition, $t(39) = 4.98$, $p < 0.001$, $d = 0.93$, and $t(39) = 4.00$, $p < 0.001$, $d = .82$, respectively. There was also a main effect of time, $F(1,38) = 6.70$, $p = 0.014$, $\eta_p^2 = 0.15$, as the trial-to-trial variability was reduced in the second block of trials.

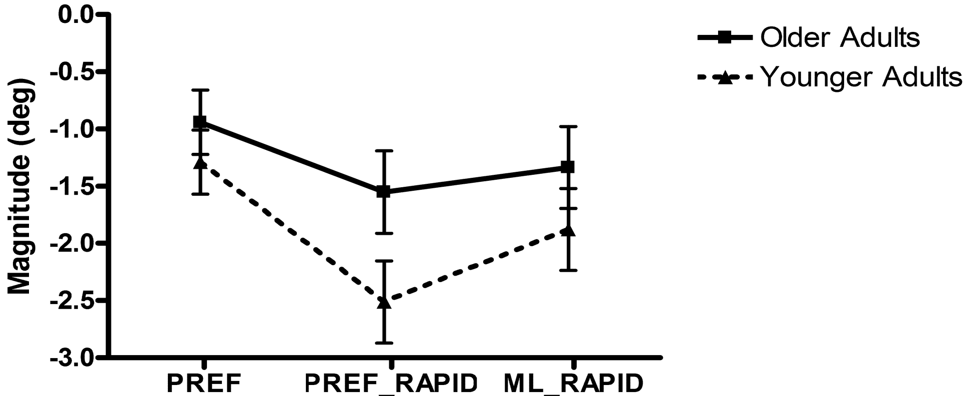


**Figure 4-13.** Magnitude of divergence between the inclination angles of the net ground reaction force and the COP-COM at the instant of foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

Evaluation of the magnitude of the first positive peak after foot-contact revealed no interactions or main effects of age or time. There was a main effect of step condition, $F(1.37, 52.09) = 37.76$, $p < 0.001$, $\eta_p^2 = 0.50$ (sphericity not assumed). Subsequent paired-samples t-tests indicated that there was no significant difference between PREF_RAPID and ML_RAPID conditions. The magnitude of this peak was larger in both PREF_RAPID and ML_RAPID conditions as compared to the PREF condition, $t(39) = 6.23$, $p < 0.001$, $d = 0.88$, and $t(39) = 6.87$, $p < 0.001$, $d = 0.99$, respectively (Fig. 4-14). Similarly, analysis of the trial-to-trial variability in the magnitude of this peak revealed only a main effect of step, $F(2,76) = 18.94$, $p < 0.001$, $\eta_p^2 = 0.33$. Again, there was no difference between the PREF_RAPID and ML_RAPID conditions, while trial-to-trial variability was larger in both PREF_RAPID and ML_RAPID as compared to the

PREF condition, $t(39) = 5.62$, $p < 0.001$, $d = 1.14$ and $t(39) = 4.886$, $p < 0.001$, $d = 0.89$, respectively.

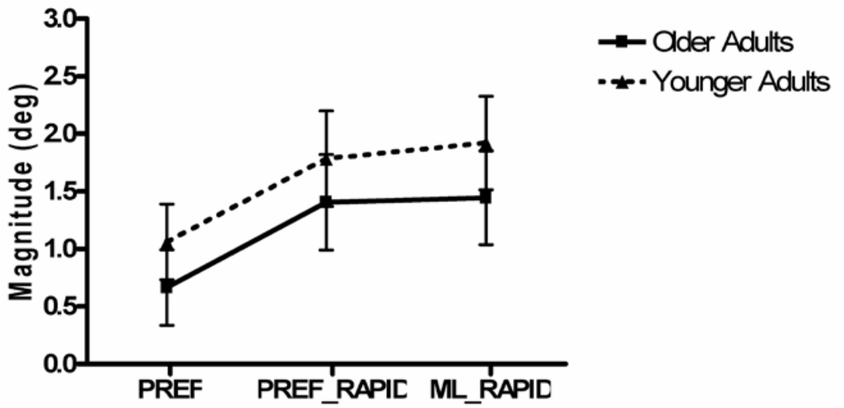


**Figure 4-14.** Magnitude of divergence between the inclination angles of the net ground reaction force and the COP-COM at the first positive peak following foot-contact (P1). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

Evaluation of the time from foot-contact to P1 revealed a main effect of step condition, $F(1.44, 54.67) = 5.85$, $p = 0.010$, $\eta_p^2 = 0.13$ (sphericity not assumed), which was qualified by an interaction between age and step condition, $F(1.44, 54.67) = 5.09$, $p = 0.017$, $\eta_p^2 = 0.12$. Follow-up independent-samples t-tests did not reveal age-related differences by step condition. Paired-samples t-tests did not reveal effects of step condition within the younger adult group, however an increased time to peak was observed in the PREF condition relative to both the PREF_RAPID, $t(19) = 3.21$, $p = 0.005$, $d = 0.71$, and ML_RAPID conditions, $t(19) = 3.29$, $p = 0.004$, $d = 0.89$, within the older adult group (Fig. 4-15). Analysis of the trial-to-trial variability revealed a main effect of step, $F(2.76) = 6.93$, $p = 0.002$, $\eta_p^2 = 0.15$. Follow-up paired-samples t-tests indicated that there was no difference between PREF_RAPID and ML_RAPID conditions, while variability was higher in the PREF condition than in both the

PREF_RAPID, $t(39) = 2.95$, $p = 0.005$, $d = 0.59$, and the ML_RAPID conditions, $t(39) = 3.14$, $p = 0.003$, $d = 0.61$.

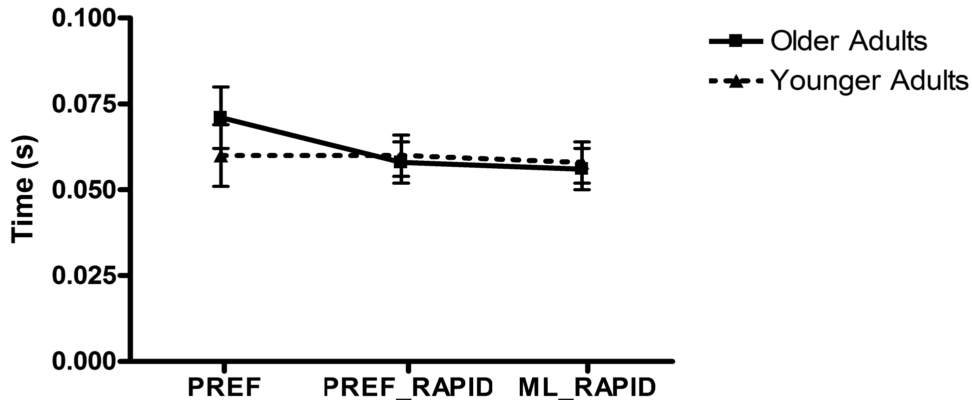


**Figure 4-15.** Timing of the first peak divergence (P1) between the inclination angles of the net ground reaction force and the COP-COM following foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

Analysis of the second positive peak after foot-contact (P2) revealed a main effect of age, $F(1,38) = 14.44$, $p < 0.001$, $\eta_p^2 = 0.28$, with the younger participants exhibiting a greater magnitude. There was a main effect of step, $F(1.69, 64.21) = 13.49$, $p < 0.001$, $\eta_p^2 = 0.26$ (sphericity not assumed). Paired-samples t-tests indicated that there was no difference between the rapid stepping conditions, while the magnitude of P2 was greater in PREF_RAPID, $t(39) = 3.66$, $p = 0.001$, $d = 0.48$, and ML_RAPID, $t(39) = 4.37$, $p < 0.001$, $d = 0.66$, relative to the PREF condition (Fig. 4-16). A main effect of time indicated that reductions in magnitude occurred in the second block of trials, relative to the first, $F(1,38) = 4.78$, $p = 0.035$, $\eta_p^2 = 0.11$. There was a main effect of step condition for the trial-to-trial variability of the P2 peak, $F(1.66, 63.03) = 8.51$, $p = 0.001$, $\eta_p^2 = 0.18$ (sphericity not assumed). Follow-up paired-samples t-tests indicated that there was no significant difference between the PREF_RAPID and ML_RAPID conditions, while the

PREF_RAPID, $t(39) = 2.74$, $p = 0.009$, $d = 0.58$, and ML_RAPID conditions, $t(39) = 3.69$, $p = 0.001$, $d = 0.86$, had larger variability than that which occurred within the PREF condition. There was also a main effect of time, $F(1,38) = 6.40$, $p = 0.016$, $\eta_p^2 = 0.14$, which was qualified by an interaction between age and time, $F(1,38) = 6.54$, $p = 0.015$, $\eta_p^2 = 0.15$. Follow-up independent-samples t-tests did not reveal significant effect of age at either block of trials. Paired samples t-tests indicated that there were no significant differences between the first and second block of trials among the older participants. The younger participants, however, reduced the trial-to-trial variability in the second block of trials, $t(19) = 3.18$, $p = 0.005$, $d = 0.56$.

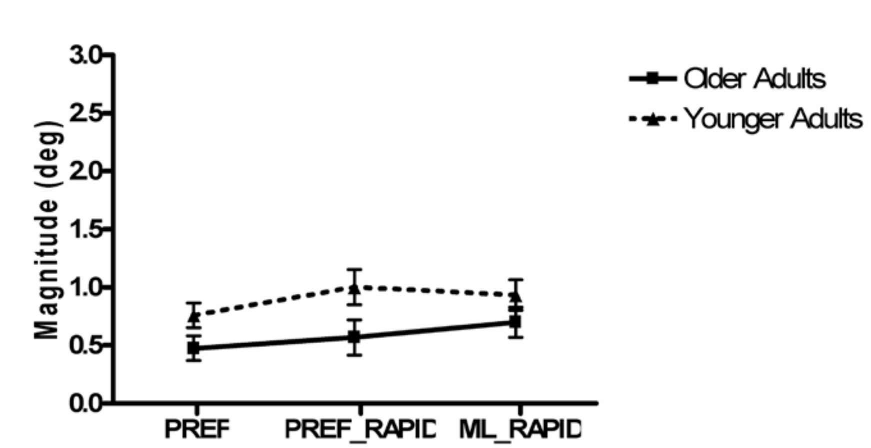


**Figure 4-16.** Magnitude of divergence between the inclination angles of the net ground reaction force and the COP-COM at the second positive peak following foot-contact (P2). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

Examination of the time from foot-contact to P2 revealed a main effect of age, $F(1,38) = 41.15$, $p < 0.001$, $\eta_p^2 = 0.52$, as older adults had a longer time to peak than did the younger group. There was also a main effect of step condition, $F(1.40,53.33) = 10.38$, $p = .001$, $\eta_p^2 = 0.22$. Follow-up paired-samples t-tests indicated that there was no difference between the PREF_RAPID and ML_RAPID conditions, while the time to the P2 peak was longer in the PREF condition than in either the PREF_RAPID or ML_RAPID conditions, $t(39) = 3.47$, $p = 0.001$, $d = 0.57$, $t(39) = 3.33$, $p = 0.002$, $d = 0.57$, respectively (Fig. 4-17). Analysis of the trial-to-trial variability of the time to P2 peak also revealed a main effect of age, $F(1,38) = 30.62$, $p < 0.001$, $\eta_p^2 = 0.45$. Older adults had greater trial to trial variability in the time to peak than did the young (Fig. 4-18).

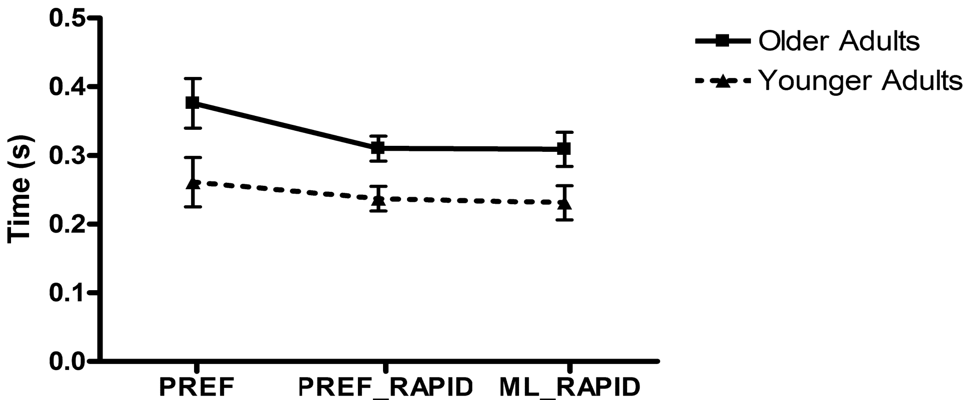


**Figure 4-17.** Timing of the second peak divergence (P2) between the inclination angles of the net ground reaction force and the COP-COM following foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

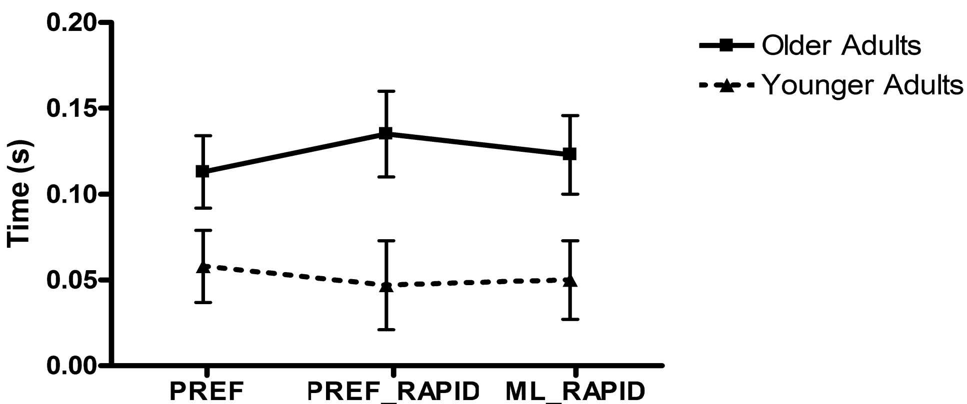


**Figure 4-18.** Trial-to-trial variability in the timing of the second peak divergence (P2) between the inclination angles of the net ground reaction force and the COP-COM following foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

## 4.5 DISCUSSION

The purpose of the present study was to extend our previous work in young adults to examine age-related differences in the control of mediolateral dynamic stability during volitional stepping. As with our previous work, the current results revealed that overshoots were common across all stepping conditions. Differences by age-group and step condition, however, were in direct opposition to our hypotheses, as incongruity was greater among older adults, particularly within the two rapid stepping conditions. Analyses of trial-to-trial variability were consistent with our hypotheses. Older adults exhibited greater variability of incongruity and this variability was increased in the two rapid stepping conditions. Results concerning the RMSD of the GRF inclination angle relative to the COP-COM inclination angle were consistent with our hypotheses that RMSD would be larger in the rapid stepping conditions. Contrary to our hypothesis, however, the younger adults exhibited increased RMSD of the GRF inclination angle in relation to the older adults.

### *4.5.1 Initial Conditions and Initial Movement Parameters*

Although the goal of the current study was to examine the restabilisation phase of stepping, we felt it would be imprudent to attempt to do so in isolation, without analyses of the prior phases of the movement. In contrast to reactive stepping, volitionally initiated stepping affords the potential for anticipatory control of mediolateral stability, prior to step onset. Lyon and Day (1997) have proposed that the state of the COM at toe-off largely influences the COM kinematics at foot-contact, as the COM falls laterally under the influence of gravity during the stepping phase. Further, forward simulations of sagittal plane balance control have suggested that the precise selection of foot-contact position, consistent with the forward COM velocity, can restore static stability simply by the passive conversion of kinetic- to potential energy (Wight, Kubica & Wang, 2008; Millard, Wight, McPhee, Kubica & Wang, 2009). Although it is not known if the latter mechanism applies to frontal plane stability control, these studies would suggest that the execution and scaling of anticipatory postural adjustments prior to step onset may impact subsequent mediolateral stability and the necessity for active regulation of applied forces for stability control during the restabilisation phase. From this, age-related differences in anticipatory control could either mask or exacerbate instability during the restabilisation phase, perhaps independent of the capacity for reactive control.

Within each age group, the ML APA appeared to be scaled in proportion to the potential for forthcoming mediolateral instability during the restabilisation phase; however the effect of the APA on the lateral displacement of the COM was reduced within the rapid stepping conditions. Interestingly, while there were no age-related differences in the mediolateral COP excursion, the older adults exhibited a larger lateral displacement of the COM toward the stance limb, which may have resulted from increased linear or angular impulse generation during the step initiation phase. Regardless of the means by which this was achieved, increased lateral COM displacement prior to step-onset would minimize the gravitational moment about the stance limb and the potential for subsequent COM acceleration toward the stepping limb during the stepping phase, which may have the effect of simplifying mediolateral stability control during the restabilisation phase, if step placement is chosen accordingly. Consistent with the abovementioned observation, we found that the mediolateral COM velocity at the instant of foot contact was reduced across all conditions among the older adults. In addition, upon foot-contact, the older adults exhibited an increased distance between the COM and lateral border of the BOS. These alterations in initial movement parameters among older adults could be viewed as planned strategies, beneficial for offsetting mediolateral instability that may emerge during the restabilisation phase.

### *4.5.2 Incongruity Magnitude, Variability and their Potential Origins*

Consistent with our previous work in younger adults, we found that overshoots of the final COM position were common during the restabilisation phase. We had previously proposed that overshoots were unrelated to COM velocity and, as such, were not a simply a consequence of the underdamped nature of the musculoskeletal system, but were a reflection of the underlying control (Singer et al., 2012). This appears to be supported by the present data, as the largest magnitude of COM incongruity occurred within conditions with the lowest mediolateral COM velocity at foot-contact. In contrast to our hypotheses, however, the older adults exhibited greater COM incongruity than did the young, which was most pronounced within the two rapid stepping conditions. We propose that such increased overshoot among older adults and within the rapid stepping conditions could arise due to one of three means: (a) increased reliance on reactive control during the restabilisation phase, due to minimization of the potential for anticipatory postural control; (b) simplification of reactive control by reduction of the frontal plane gravitational moment about the stepping limb; (c) exploitation of passive mechanical energy conversion, for minimization of joint loading. Each of these options is considered for discussion in the following text.

We believe the most plausible explanation, due to its consistency with the kinetic data, is that COM incongruity during volitional stepping is likely governed by reactive control during the restabilisation phase, but can be offset to some degree by anticipatory postural control prior to step-onset, which serves to minimize the development of ML instability during the stepping phase, until stepping foot-contact. Even when movement speed and step placement are self-selected and there is the opportunity for anticipatory control, overshoots may be the typical response in a healthy system. Under such ideal conditions, overshoots may arise because of the length of time required, after foot contact, to detect the state of the COM and actively generate an appropriate response to arrest its forward and lateral progression, rather than as a result of the absence of anticipatory control or poorly scaled active force modulation during the restabilisation phase. In contrast, the task of rapid stepping appears to truncate the expression of the ML APA, which potentially shifts the burden of ML stability control to the restabilisation phase. Under these conditions, the time required to detect instability and generate active force to decelerate the COM is compounded by the necessity to precisely regulate the direction of applied force on foot contact to achieve stability, which may potentially lead to increased COM incongruity.

In relation to the present work, we propose that the ideal control within either age group occurred when stepping with self selected speed and step placement. Under these conditions, we observed the largest lateral displacement of the COM during the step-initiation phase and largest lateral distance between the COM and BOS at the point of foot contact, which, we believe, may facilitate subsequent stability control during the restabilisation phase. Correspondingly, participants exhibited the smallest (least negative) divergence between the GRF and COP-COM inclination angles at the instant of foot contact. Such a reduction would indicate a greater tendency for the line of action of the net GRF to pass near the COM throughout swing phase, which may potentially result in a smaller moment about the COM and reduced generation of angular momentum toward the swing limb side. As such, the magnitude of this divergence may be a marker of the extent of lateral instability at the onset of the restabilisation phase. During the restabilisation phase, both age groups exhibited lesser magnitudes of the P1 and P2 peaks relative to the rapid stepping trials. We have suggested that the magnitudes of P1 and P2 may be modulated in response to the state of the COM during the restabilisation phase, by proactive and reactive mechanisms, respectively. As such, smaller magnitudes of these variables may indicate that there was a reduced necessity to counteract the instability that developed during the stepping phase. Despite this best possible pairing between the initial movement parameters and those occurring during the restabilisation phase, COM incongruity was nonetheless apparent. We believe that, under these conditions, COM incongruity may be related to the time delay, after foot-contact, in reorienting the ground reaction force to counter lateral instability arising within the swing phase (i.e. time to P2 peak).

During the rapid stepping conditions, however, we observed an increase in COM incongruity among both age groups. During the step-initiation phase, there was a reduction in the lateral displacement of the COM, which likely led to reduced dynamic stability at the instant of foot contact, as signified by the larger divergence (more negative) between the GRF and COP-COM inclination angles. Such alterations in the initial movement parameters associated with rapid stepping likely heightened the importance of the restabilisation phase for ML stability control. This is consistent with the increased magnitude of P1 and P2 peaks, which may serve to generate ‘restabilising’ moments about the COM. This is also in accordance with the increased RMSD of the GRF inclination angle, which may suggest that it is the regulation, rather than minimization, of angular momentum during the restabilisation phase that is important for stability control. Together, these results imply that the magnitude of incongruity and consequent instability may be dictated not only by the time lag to the reorientation of the ground reaction force, but by proper scaling of the magnitude of the restabilising moment about the COM.

Among the older adults, we observed both an increased magnitude and trial-to-trial variability of COM overshoot. Interestingly, across all step conditions, increased incongruity magnitude was apparent among older adults in spite of the abovementioned alterations in initial movement parameters, which may serve to maximize stability at the onset of the restabilisation phase, as previously described. Given the potential that older adults were more dynamically stable at the onset of the restabilisation phase, yet exhibited an increased magnitude of overshoot, we suggest that the origin of such instability most likely arises at some point during the restabilisation phase, which is consistent with previous research (McIlroy & Maki, 1996; Rogers et al., 2001; Rogers & Mille, 2003). We believe that the age-related differences in the magnitude of the P2 peak were unlikely the explicit cause of the age-related differences in incongruity magnitude. Specifically, although this peak was reduced among older adults, it may have been scaled correctly to the actual amount of instability that developed during the stepping-phase, which was signified by a reduced divergence between GRF and COP-COM inclination angles at foot-contact among older adults. Instead, we believe that the age-related increases in the timing and variability in timing of the P2 peak were most likely related to the increased magnitude and trial-to-trial variability of overshoot among the older adults. If P2 does represent an active response to the state of the COM at the onset of the restabilisation phase, as we propose, the increased time to peak could signify difficulties in detecting the position/velocity of the COM, slowing of conduction velocity (Inglis, Horak, Shupert & Jones-Rycewicz, 1994), impairment of central processing (Horak, Henry & Shumway-Cook, 1997), multisensory integration (Horak, Shupert & Mirka, 1989), or reduced rate of force development (Chang, Mercer, Giuliani & Sloane, 2005) due to muscle atrophy, slowing of contractile properties or impaired coordination (Barry, Riek & Carson, 2005), all of which have been previously documented. This assertion is given some support by the fact that overshoots were not reduced with trial repetition and, as such, may be due to such biological factors, rather than the result of inappropriate movement planning, which could potentially be altered with practice to augment stability.

Although there was no significant interaction of age and step condition, age-related differences in incongruity magnitude were particularly apparent within the two rapid stepping conditions (Fig. 4-5). It is possible that the task of rapid stepping exposes underlying challenges with stability control not evident during preferred stepping, given the associated minimization of the potential for anticipatory ML stabilisation prior to step-onset and the emphasis on reactive control for ML stabilisation.

What remains unclear, however, is whether the delay in the time to the reorientation of the ground reaction force was intended, as a means to facilitate overshoot, given the increased time to peak during the PREF condition relative to the rapid stepping conditions. In our previous work, we had suggested that overshoot of the final COM position may be associated with the maintenance of steady-state gait following step initiation and may aid in the simplification of reactive control. Specifically, minimization of the frontal plane gravitational moment about the supporting foot could be achieved by allowing the vertical projection of the COM to progress laterally to a point nearly coincident with ankle joint centre of the stepping limb. In such a case, the acceleration of the COM toward the contralateral side during the forthcoming step would be reduced. If paired with appropriate step placement, this could be an effective method to simplify stability control during the forthcoming step (Lyon and Day, 1997, Wight et al., 2008; Millard et al., 2009).

In a similar manner, the increased magnitude of overshoot among the older adults could represent an attempt to preserve mediolateral stability in the event an additional step was needed to arrest the forward COM velocity. This is consistent with research indicating that during both gait termination (Cao et al., 1998; Tirosh & Sparrow, 2004) and anterior compensatory stepping responses (McIlroy & Maki, 1996; Schulz et al., 2005), older adults are more likely to employ multiple forward steps than are younger adults. In effect, mediolateral COM overshoot would play an analogous role to the ML APA executed during the step-initiation phase. When comparing across stepping conditions, the increased overshoot observed during the two rapid stepping tasks, relative to the PREF condition, may also act as a safeguard against ML instability with subsequent forward steps, given the increased anterior COM velocity associated with rapid stepping. While it is reasonable to assume this mechanism may serve to simplify reactive control among older adults, who may lack sufficient strength to arrest the COM within a single step, it seems unlikely that this mechanism would explain the increased overshoot among the younger adults, who should not require multiple forward steps to regain stability.

In a similar context, it is possible that overshoot may be a means to avoid high peak forces and joint loading by exploiting the passive conversion of kinetic to potential energy as the COM progresses toward the stance limb. In addition, equivalent deceleration impulses could be generated, without high peak forces, by increasing the time over which the COM velocity is arrested. This, of course, would require there to be a sufficient distance between the COM and the lateral stability limits over which to achieve this deceleration, which was observed among the older adults.

Although it may be reasonable to assume that such a strategy may be employed by older adults, who may either lack the strength to generate high peak forces or are attempting to moderate joint loading, allowing increased lateral COM displacement could compromise stability if the COM exceeded the lateral stability limits. Moreover, if the increased lateral COM displacement before step-onset and the reduced ML COM velocity are, in fact, proactive strategies to counter the potential for instability during the restabilisation phase, the act of deliberately allowing the COM to travel closer to the lateral BOS limits would likely be counterproductive to this objective. Further, as with the previously outlined strategy for simplification of ML stability control, this theory does not account for the increased incongruity among the young adults during the rapid stepping conditions, given that this group should not lack sufficient strength nor be attempting to limit the magnitude of joint loads that are incurred during a single voluntary step.

### *4.5.3 Overall Implications*

Overall, results from the present study suggest that while COM overshoot may aid in the simplification of reactive control, increased overshoot among older adults may arise from difficulty in reactive control during the restabilisation phase. This is given some support by the increased trial-to-trial variability of incongruity among the older adults, which suggests that COM incongruity may represent some degree of disordered control. We propose that age-related differences in COM incongruity are primarily linked to the time lag in active reorientation of the ground reaction force after foot-contact. Rapid stepping appears to minimize the potential for anticipatory ML stabilisation prior to step-onset, which may shift the burden of ML dynamic stability control to the restabilisation phase.

Under these conditions, increased overshoot arose because of the combined effects of the time required to detect instability and reorient the ground reaction force and the necessity to precisely regulate the magnitude of divergence of the ground reaction force (i.e. moment), relative to the COM, during the restabilisation phase. We propose the larger magnitude of overshoot among the older adults during rapid stepping may signify increased sensitivity to minimization of the potential for anticipatory ML stabilisation as a means to offset subsequent instability during the restabilisation phase.

Interestingly we did not observe within-group differences in incongruity magnitude or variability when comparing the two rapid stepping conditions. While stepping with reduced step width did not increase the magnitude of overshoot, it does pose a challenge to the maintenance of stability, as a narrower step in association with an equal magnitude of overshoot would tend to reduce the mediolateral distance between the COM and lateral stability limits. This is particularly relevant when viewed in concert with the increased trial-to-trial variability among the older adults and during the rapid stepping tasks. Even if overshoot is a mechanism to simplify mediolateral stability control, increased overshoot coupled with increased variability may lead to lateral instability if the COM exceeds the lateral stability limits.

The overall aim of the present study was complicated, to some degree, by age-related differences in the initial movement parameters. Future research using perturbation-evoked reactive stepping, which would reduce the potential for anticipatory control equally across age groups, may provide a clearer picture of age-related differences in stability control during the restabilisation phase. Under such conditions, further increases in COM overshoot among older adults may support the idea that instability arises during the restabilisation phase. Moreover, by removing the influence of age-related differences in the initial movement parameters, we may be able to more clearly discern the origins of ML instability during the restabilisation phase.

# CHAPTER 5

## Age-Related Differences in the Control of Mediolateral Dynamic Stability during the Restabilisation Phase of Stepping Evoked by Postural Perturbation

## 5.1 OVERVIEW

The control of mediolateral dynamic stability has been found to be particularly problematic for older adults during stepping, which is particularly relevant to falls and fall-related injury. There is a need for better understanding of the origins of age-related mediolateral instability, with a specific emphasis on movement termination (restabilisation). The purpose of the current study was to advance our understanding regarding age-related changes in the control of mediolateral dynamic stability during the restabilisation phase of reactive stepping, evoked by postural perturbation. Healthy young (n=20) and older (n=20) participants took part in two experimental conditions. Participants were asked to regain their balance using a single step, after being unexpectedly released from a forward leaning position. In one condition, there were no constraints on step placement. We sought to further explore the effects of age by including an additional condition in which individuals were asked to step with reduced step width, which we believed would pose an increased challenge to the maintenance of stability. Secondary to this aim, we attempted to probe the extent and means by which individuals could alter the dynamics of the compensatory stepping response over time, with trial repetition. As such, trials in each condition were presented in blocks of 5 repeated trials. Blocks of trials were randomly presented to participants. The magnitude of incongruity between the peak and final COM position was quantified along with the intertrial variability of incongruity. To aid in the determination of the origins of age-related differences in the kinematic variables, the timing and magnitude of the waveform representing the difference between the inclination angles of the net ground reaction force and COP-COM were analysed. Results revealed that overshoots of the final COM position were common among both age groups in all stepping conditions, but were subsequently reduced with trial repetition. Older adults, however, exhibited an increased magnitude of overshoot relative to the younger adults. Older adults also exhibited greater trial-to-trial variability of COM incongruity than the younger adults, which was also reduced with trial repetition. Examination of the kinetic data in concert with these kinematic variables provided support for previous work, which has suggested that overshoot may be a means to simplify the control of mediolateral dynamic stability during the restabilisation phase of stepping, should additional steps become necessary for balance restoration. Increased overshoot and intertrial variability in COM incongruity exhibited by older adults, however, was unlikely to be advantageous for stability control and is proposed to be a consequence of age-related differences in the timing and variability in timing of the reorientation of the net ground reaction force with respect to the COM. Together these results suggest that older adults may have particular difficulty with reactive control during the restabilisation phase. Reductions in overshoot and intertrial variability, specifically among the older adults, are proposed to occur primarily through the alteration of pre-planned movement parameters and swing phase dynamics. This work highlights the importance of exploring the restabilisation phase of stepping, which complements previous work on step initiation.

## 5.2 INTRODUCTION

Compensatory stepping reactions are essential responses for recovering stability following postural perturbation, as an increase in the area defining the base of support (BOS) affords a larger potential for restabilisation. Contrary to initial research indicating that these reactions lie at the end of a continuum of responses available to regain stability following a perturbation (Horak & Nashner, 1986), subsequent research has indicated that compensatory stepping may be the preferred response in the absence of experimental restrictions to the use of fixed-support strategies (McIlroy & Maki, 1993c). This assertion has been given support from studies employing stability modelling (Pai et al., 1998), which have revealed that stepping may be initiated well before the centre of mass (COM) reaches the limits of the BOS (Mille et al., 2003). Further, experimental studies of older adults have revealed an increased prevalence of stepping following an external perturbation to standing balance, perhaps due to either a diminished ability to attenuate accelerations induced by the perturbation or reduced stability limits (Pai et al., 1998; Jensen et al., 2001).

A large number of studies have been conducted to shed light on such age-related declines in sagittal plane balance recovery ability (Cao et al., 1998; Chu, Tang, Chen & Cheng, 2009; Menant, Steele, Menz, Munro & Lord, 2009; Carty, Mills & Barrett, 2011). Some degree of ambiguity remains, however, likely due to the multiple modalities for inducing sagittal plane instability and the differences between studies in the specific characteristics of the sample participants. For example, age related decline in volitionally generated lower limb joint moments have been found by some researchers (Wojick et al., 2001) to be poor predictors of the maximum lean angle from which participants could successfully recover AP stability via stepping. Conversely, others (Karamanidis et al., 2008) have found a reduction in both ankle and knee maximal voluntary isometric strength and margin of stability (AP COM-foot distance) in older adults, concluding that muscle strength may be responsible for the inability of older adults to effectively increase the BOS length to successfully arrest the anteroposterior COM velocity. Despite the discrepancy, lower limb volitional strength measures in isolation may not be adequate indicators of available strength or the capacity to rapidly move the limbs, given the time-sensitive nature of balance recovery responses. Moreover, it seems likely that the inability to recover stability may be more greatly influenced by a large number of small neuromuscular deficits, rather than decrements in a single faculty alone (Hsiao et al., 1999). Notwithstanding the disagreement, there appears to be a general consensus that the inability for older adults to successfully regain stability is independent of many of the initial movement parameters, such as reaction time (Thelen et al., 1997), muscle onset latency (Thelen et al., 2000), foot-off time, foot-contact time and step placement (McIlroy & Maki, 1996).

Although there has been extensive research devoted to understanding sagittal plane restabilisation, the current body of literature has not fully addressed the particular difficulty that older adults face in the control of mediolateral stability during forward stepping, which is particularly relevant to fall-related injury (Maki et al., 1994; Lord et al., 1999; Brauer et al., 2000; Hilliard et al., 2008; Schrager et al., 2008). While compensatory stepping reactions often lack the mediolateral anticipatory postural adjustments (ML APAs) that typically precede volitional stepping (McIlroy & Maki, 1993b), subsequent work has found these APAs, when present during reactive stepping, to be insufficient to affect subsequent COM dynamics (McIlroy & Maki, 1999). Even with the striking similarity between age groups until the time of foot contact, as previously mentioned, older adults have been found to have an increased instability at the time of foot-contact (McIlroy & Maki, 1996; Rogers et al., 2001; Schulz et al., 2005), which has been related to fall risk (Maki et al., 1994; Lord et al., 1999; Brauer et al., 2000; Hilliard et al., 2008; Schrager et al., 2008).

Although multiple stepping responses are clear markers of instability in older adults (McIlroy and Maki, 1996), the underlying origins of mediolateral instability have received extremely little consideration and, hence, remain somewhat elusive. Our previous study of volitional stepping in younger adults has examined frontal plane COM displacement during the restabilisation phase and suggested that incongruity between the peak and final COM position after foot-contact – specifically overshoot of the final position toward the stepping limb side – may play a functional role in simplifying mediolateral balance control (Singer et al., 2012). More specifically, this strategy may place emphasis on the stepping limb to achieve restabilisation, while reducing the necessity for interlimb coordination. Further, allowing the COM to move closer to the stepping foot could simplify mediolateral stability control for subsequent steps, should the initial step be insufficient to arrest the forward COM velocity. Such a reduction in the mediolateral distance between the COM and support limb, during the single-support phase of a subsequent step, would reduce the frontal plane gravitational moment and subsequent acceleration toward the side of the forthcoming stepping limb.

Despite the potential for simplification of mediolateral stability control, our subsequent work (study 2) has found an increased magnitude of incongruity among older adults, relative to younger adults, during rapid speed stepping. Rapid stepping reduced the extent of anticipatory mediolateral stabilisation prior to step-onset, which likely shifted the burden of ML dynamic stability control to the restabilisation phase. We believe the increased incongruity among the older adults during such rapid stepping conditions to be indicative of difficulty with reactive control during the restabilisation phase of stepping. More specifically, older adults exhibited an increased time to reorient the net ground reaction force with respect to the COM during the 'active' phase of the restabilisation phase, which may delay the generation of angular impulse necessary for frontal plane angular deceleration during double support.

Similarly, regulation of whole-body angular momentum is one factor that has been suggested to be an important determinant of dynamic stability during walking (Simoneau & Krebs, 2000; Herr & Popovic, 2008) and balance recovery following a trip (Pijnappels et al., 2004). It has been suggested that the stance limb plays a role in sagittal plane balance recovery during stepping, by generating intersegmental moments appropriate for regulating the direction of the GRF with respect to the COM (i.e. angular momentum) (Pijnappels et al., 2004). While the stance limb no doubt aids in sagittal plane restabilisation, the complete recovery of mediolateral stability likely occurs after stepping-limb foot contact, given that there may be insufficient time during single support for the stance limb to generate forces of large enough magnitude to appreciably alter the mediolateral COM velocity. More likely, it is the double limb support phase in which external forces and moments are generated with sufficient magnitude and duration to play a more considerable role in balance recovery. Along these lines, Herr and Popovic (2008) have suggested that, during level walking, the net ground reaction force and centre of pressure trajectory are regulated throughout the gait cycle in such a way as to minimize the net external moment about the total body COM. From this, it is proposed that the generation of a net ground reaction force with a line of action through the COM may have an important relationship with the control of mediolateral COM displacement during compensatory stepping. In addition, age-related differences in the ability or execution of such control may lead to the increased instability often observed during compensatory stepping.

From this, the purpose of the present work was to advance our understanding of mediolateral dynamic stability control during compensatory stepping, throughout the restabilisation phase – a phase of the movement that may represent the earliest opportunity to substantially alter ML COM trajectory and velocity. More specifically, the aim of this study was to examine age-related differences in the control of mediolateral dynamic stability during the restabilisation phase, following a cable-release perturbation to standing balance.

Four specific hypotheses were tested in the present study:

1) Incongruity magnitude (Fig. 5-1):

a) We hypothesized that older adults would exhibit greater incongruity magnitude (i.e. overshoot) than younger adults, when stepping without restrictions on foot placement. This is consistent with our previous research (study 2), in which we observed an increased magnitude of overshoot during rapid speed stepping due to the reliance on reactive control during the restabilisation phase.

b) We also hypothesized that, among the older adults, incongruity magnitude would increase when individuals were forced to step with a narrow step width, as step responses with constrained step width would reduce the potential to move the COP lateral to the COM to generate a net moment about the COM, while also reducing the potential to generate a mediolateral force component with sufficient magnitude to arrest the lateral linear COM velocity. We did not expect to find differences between step conditions within the young adults.

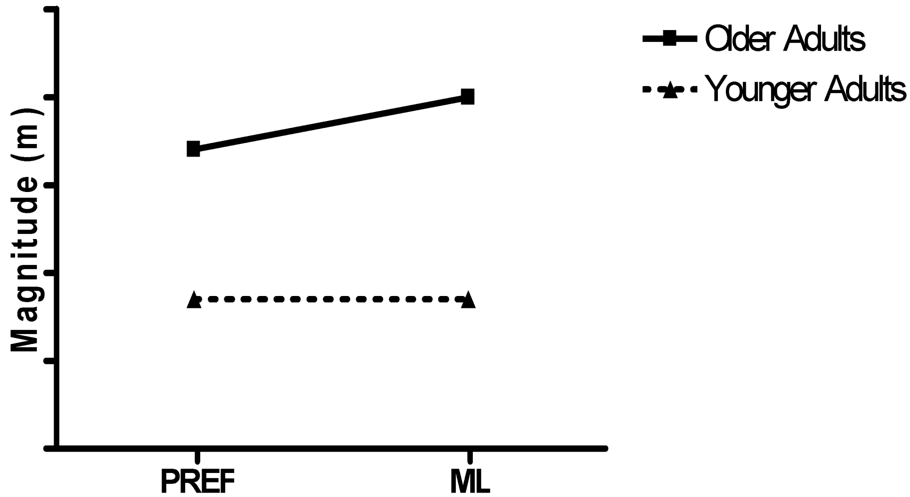


**Figure 5-1.** Hypothetical mediolateral incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Task conditions include: preferred step placement (PREF); narrow step width (ML). Please note that error bars and vertical axis scale are not shown, as this data is completely hypothetical and not based on previous work.

2) Trial-to-trial variability (Fig. 5-2):

a) We hypothesized that we would observe a greater amount of trial-to-trial variability among the older adults, when step width was not constrained.

b) We believed that we would observe further increases in trial-to-trial variability among the older adults when forced to step with a narrow step width, in relation to both their own preferred-width stepping trials and trials among the younger adults. We did not expect to find differences between step conditions within the young adults.

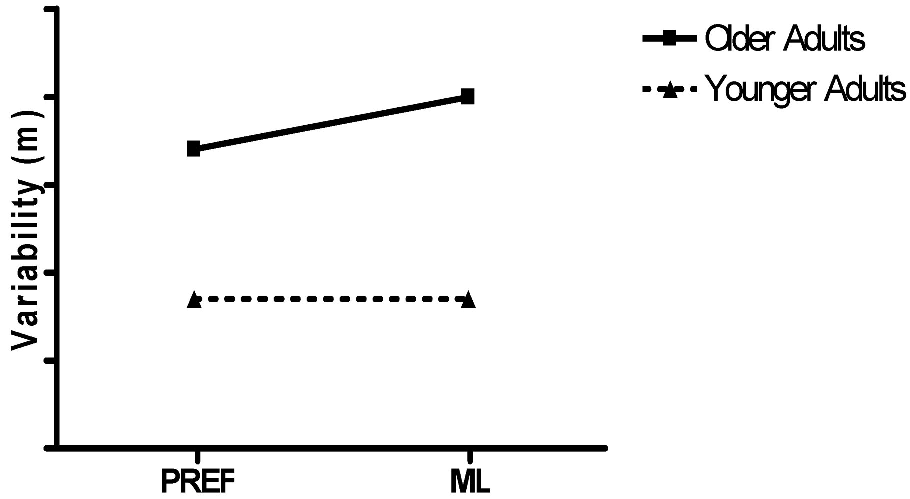


**Figure 5-2.** Hypothetical trial-to-trial variability of incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Task conditions include: preferred step placement (PREF); narrow step width (ML). Please note that error bars and vertical axis scale are not shown, as this data is completely hypothetical and not based on previous work.

3) RMSD of the GRF inclination angle relative to the COM (Fig. 5-3):

a) In our previous work (study 2), we observed an increased RMSD among younger adults. Such differences, however, were believed to occur because younger adults made fewer modifications to the initial movement parameters to offset subsequent instability during the restabilisation phase, leading to a greater necessity to regulate total body angular momentum during the restabilisation phase. We anticipate that the cable-release perturbations employed within the current study will nullify the potential to make modifications to the initial movement parameters prior to step-contact. As such, we believe that there will be no difference in RMSD between age-groups during trials with preferred step placement, as both will be required to equally regulate the orientation of the net GRF on foot contact to achieve restabilisation.

b) Further, without the potential to offset subsequent instability by modifications of the initial movement parameters, as we observed during our previous work of volitional stepping (study 2), we expected to observe equal reductions in RMSD within each group when stepping with reduced step width, relative to trials with preferred step placement.

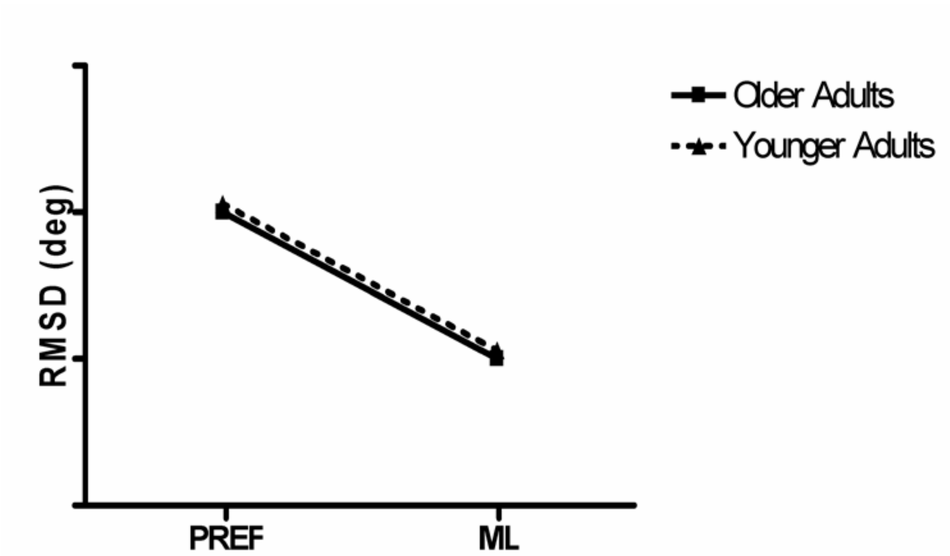


**Figure 5-3.** Hypothetical frontal plane RMSD of the actual GRF inclination angle, with respect to the inclination angle of the line from COP to COM, for older (solid line) and younger adults (dotted line) adults, averaged within condition. Task conditions include: preferred step placement (PREF); narrow step width (ML). Please note that error bars and vertical axis scale are not shown, as this data is completely hypothetical and not based on previous work.

4) Time to 'active' reorientation of the net GRF (time to P2 peak) (Fig. 5-4):

a) We hypothesize that older adults will exhibit a longer time to achieve the peak divergence of the net GRF inclination angle with respect to the COP-COM inclination angle (P2 peak) than younger adults when stepping with preferred step placement.

b) We believe that the differences in the timing of the P2 peak will be unaffected by step placement. Specifically, we hypothesize that there will be no change in timing within either group with constraints on step placement. Further, we believe that the differences between age groups observed in the condition with preferred step placement, will persist when step width is reduced.

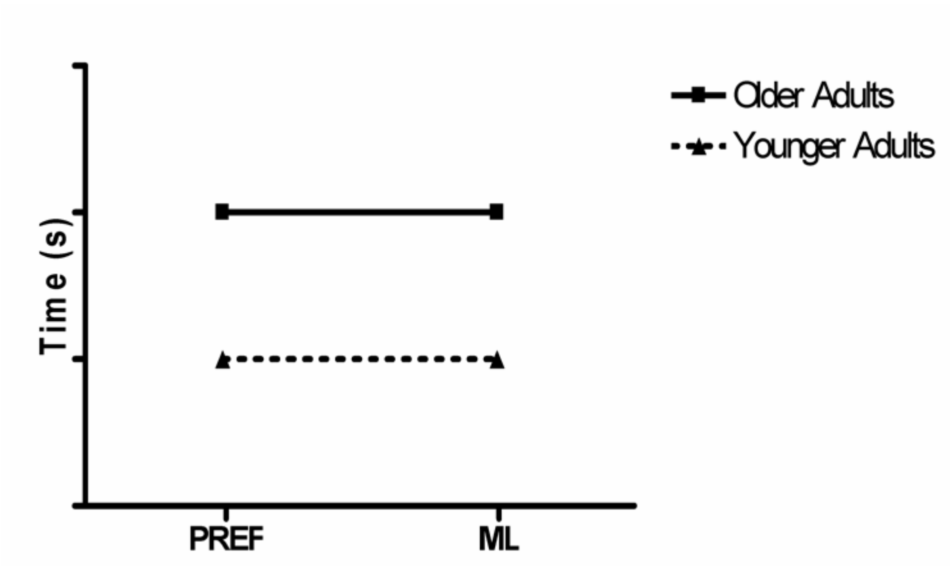


**Figure 5-4.** Hypothetical timing of the peak divergence between the inclination angles of the net ground reaction force and the COP-COM at the second positive peak following foot-contact (P2). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Task conditions include: preferred step placement (PREF); narrow step width (ML). Please note that error bars and vertical axis scale are not shown, as this data is completely hypothetical and not based on previous work.

## 5.3 METHODS

### *5.3.1 Participants*

Twenty healthy young (age 24 (5) years) and twenty community-dwelling healthy older adults (age 71 (5) years) were recruited for inclusion within the study. Participants were free from anatomical, neurological, or cognitive impairments. All participants were able to stand and walk unaided, had no previous history of falls and were not using psychoactive medications (specifically for pain control or management of mental or emotional disorder).

### *5.3.2 Instrumentation and Set-Up*

Six Vicon MX-3+ cameras (Vicon Motion Systems, Los Angeles, CA) were used to record kinematic data (100 Hz). Four force platforms (Advanced Mechanical Technology, Inc., Watertown, MA.), arranged in a rectangular array and embedded in the laboratory floor, were used to measure the reaction forces and moments (2000 Hz). Participants were anchored to a rigid frame, via an adjustable cable and chest harness, which was in series with a force transducer. In conjunction with a differential amplifier (Bortec Biomedical, Calgary, AB), disposable, self-adhesive Ag/AgCl electrodes were placed bilaterally on the tibialis anterior and soleus and were used to record electromyographic activity from these sites (2000 Hz). All motion capture data and analogue-to-digital converted signals were synchronously recorded by Vicon Nexus software (Vicon Motion Systems, Los Angeles, USA).

Retroreflective calibration markers, of 1 cm diameter, were placed on the subject over anatomically relevant locations in a similar fashion to the marker set used by the National Institute of Health and described by Hamill and Selbie (2004). Additional calibration markers were placed bilaterally on the iliac crests, anterior and posterior superior iliac spines, acromioclavicular joints, anterior to the external auditory meatus, greater and lesser tubercles of the humerus, medial and lateral epicondyles of the humerus, radial and ulnar styloid processes and the head of the 3$^{rd}$ metacarpal. Rigid clusters containing four markers, placed on the sacrum and trunk, and bilaterally on the feet, legs and thighs were used to determine the 3D kinematics of each respective segment during the experimental trials. Positive x-,y- and z-axes for the laboratory coordinate system were oriented laterally to the right side of the participant, anteriorly and upward, respectively.

### *5.3.3 Protocol*

A standing reference trial was collected prior to the collection of the experimental trials. Participants were asked to stand in a neutral position, roughly aligned to the laboratory coordinate system. These data were used to determine the segment endpoints, the segment embedded local coordinate system, as well as the transformation matrices between the local and global coordinate systems for each segment. Following collection of the standing reference trial, markers used solely for calibration were removed.

Participants were asked to stand with their feet side-by-side on separate force platforms. Foot position was standardized across all participants to a width of 0.17 m, with an offset of 14 degrees between the longitudinal axis of the foot and the anteroposterior axis of the laboratory coordinate system, as per McIlroy and Maki (1997). Adhesive foam weatherstripping was placed on the force platform, along the medial and posterior borders of each foot, to ensure consistent foot placement between trials. Two quiet standing trials, of 60-second duration, were collected to obtain kinematic and electromyographic variables necessary for post-processing. The first trial was collected with the feet in the abovementioned standardized side-by-side configuration. For the second trial, the participant was asked to start with their feet in the standardized position, take a single step forward and hold this forward-stance configuration for the duration of the trial.

For experimental trials, an initial forward lean was established, standardized to a cable load of 10% body weight, and was monitored during the trial. During pilot testing, this lean magnitude was found to evoke stepping in 100% of trials in young adults while also failing to evoke multiple forward steps in older adults. To ensure symmetrical mediolateral weight distribution, the vertical force under each foot was monitored before cable release. The anterior position of the COP under each foot was also monitored. Electromyographic activity was monitored and recorded bilaterally from the soleus and tibialis anterior, to ensure minimal pre-perturbation activity. As a reference, we attempted to ensure that the level of activity observed during the experimental trials did not exceed that which was recorded during quiet upright standing. This ensured that the participant had fully committed to the use of the cable and harness to maintain the forward lean. To discourage anticipation of cable release onset, the cable was released at random intervals after the participant had adopted the forward lean and the initial conditions had been established. Furthermore, 1/3 of experimental trials (5 trials) consisted of 'catch trials' in which no perturbation was administered. These trials were randomly presented throughout the course of testing. Participants were asked to regain their balance following the perturbation by using a single step, if necessary. Further, participants were told that, regardless of whether a step was needed, they should remain in the final position for approximately 10 seconds once they regained stability. This allowed sufficient data for examination of the restabilisation phase of stepping.

Participants were asked to take part in two experimental conditions. The first condition placed no restriction on step placement (PREF). The second condition was used to further explore the potential for age-related differences in mediolateral stability control, by requiring participants to regain balance by using a step of narrow width (ML). Participants were asked to step onto a length of adhesive tape placed on the force platform, parallel to, and 5 cm lateral to the median plane (toward the stepping limb side).

We believed it was possible that reductions in incongruity magnitude may occur with task familiarity, as individuals learned to take advantage of the predictable features of the perturbation to augment stability control. Such changes, and the potential for a differential effect by age, may have helped to inform us about the role or cause of COM incongruity. To address this, the ten experimental trials in each condition were split into two blocks of five trials (PREF1, PREF2, ML1, ML2). The effect of 'time' was incorporated into the omnibus ANOVA model used to address each dependent variable, as described below. The presentation of these four blocks of trials was randomized for each participant. Subsequent analyses were limited to successful single-step balance recovery responses.

### *5.3.4 Data Analysis*

Force platform data were lowpass filtered using a zero-lag, fourth-order, Butterworth filter with a cut-off frequency of 50 Hz. As the Butterworth filter is underdamped, there was a possibility using such a filter could induce artefact in individual marker trajectories during rapid transitions, which could lead to subsequent artefact in the computation of the total body COM position. Marker data were lowpass filtered using a zero-lag, twentieth-order critically damped filter, with a cut-off frequency of 6Hz, which should prevent marker over/undershoot as a result of filter artefact and also provide an equivalent roll-off to that of the Butterworth filter (Robertson & Dowling, 2003). A linear envelope (LE-EMG) was calculated from raw EMG data, after initial removal of dc-bias, full-wave rectification and lowpass filtering using a second-order Butterworth filter, with a cut-off frequency of 3 Hz.

The entire body was modelled as a rigid system of independently tracked segments. For the older adults, segment masses were estimated using Dempster’s segment parameters and segment centre of mass positions were estimated using the geometrical model proposed by Hanavan (1964) (cited in Robertson et al., 2004). Segments of the appendicular skeleton were modelled as conical frusta; the pelvis and trunk were modelled as elliptical cylinders and the head as a sphere. As Dempster’s segment parameters are not representative of younger adults, segment masses and centres of mass were estimated for this group using segmental mass proportions and relative segmental centre of mass locations from a sample of young, male and female undergraduate students, initially reported by Zatsiorsky, Seluyanov and Chugunova (1990) and subsequently modified by de Leva (1996). The trunk segment was modelled as a hybrid of the upper- and middle-trunk segments, defined by de Leva (1996). The proximal and distal endpoints for the trunk segment for the present study were the iliac crests and acromia, respectively, and the lengths of upper and middle trunk segments for each subject, as defined by de Leva (1996), were not known. Average segment lengths of a sample of males and females from Zatsiorsky, Seluyanov and Chugunova (1990) (as cited in de Leva, 1996), were used to calculate the lengths of the upper- and middle-trunk segments, as proportions of the total trunk length. These scale factors, in conjunction with the relative segmental COM locations and measured total trunk length (from marker coordinates), were used to compute the hybrid trunk COM location, which was assumed to lie on the longitudinal axis running from the midpoint between iliac crest and acromial markers, respectively. The total body centre of mass position was calculated as a weighted average of all body segment COM locations, where each segment was weighted according to its mass proportion.

The COM restabilisation point was defined as the time point at which the COM velocity waveform entered and remained inside an amplitude bandwidth bordered by +/- two standard deviations of the mean velocity during the initial quiet standing trial with forward-stance configuration. Velocity, rather than position, was chosen for determination of the COM restabilisation point, as this allowed for calculations that were less affected by low frequency oscillations or drift of the COM position-time waveform after restabilisation.

COM incongruity was defined as the difference between the local maximum COM position after foot contact and the mean of the stable region of the waveform, which was a two second window beginning at the restabilisation point. Overshoot was defined as positive incongruity; undershoot was defined negative incongruity. Trial-to-trial variability was assessed by computing the standard deviation of the mean COM incongruity over experimental trials in each block of 5 repeated trials for each condition.

The ground reaction forces from all force platforms were combined to yield a single force vector. The frontal plane inclination angle of this net force was calculated, with respect to the x-axis (ML) of the global coordinate system. The frontal plane inclination angle of the line joining the COP and COM was also determined, relative to the x-axis (ML) of the global coordinate system. In theory, if the inclination angle of the net GRF is coincident with the inclination angle of the COP-COM, there should be no external moment about the COM caused by the net GRF and, hence, no change in frontal plane angular momentum. To provide a single metric for characterization of this compound kinetic variable, the RMSD of the frontal plane GRF inclination angle with respect to the frontal plane COP-COM inclination angle was computed, for the interval from foot-contact to the restabilisation point.

Examination of this waveform revealed a characteristic pattern (Fig. 5-5) with three distinct peaks at, and after, foot contact. We proposed that the first negative peak at foot contact signifies a net GRF vector orientation that would result in an external moment and angular acceleration toward the swing limb (right-footed stepping: positive moment, anticlockwise about an anterior axis with origin at the whole body COM; left-footed stepping: negative moment, clockwise about an anterior axis with origin at the whole body COM); the magnitude of this peak is likely influenced the by the anticipatory postural adjustments and swing phase dynamics. The subsequent two positive peaks (P1 and P2) denote a net GRF orientation that would result in an external moment and angular acceleration toward the side of the stance limb (right-footed stepping: negative moment, clockwise about an anterior axis with origin at the whole body COM; left-footed stepping: positive moment, anticlockwise about an anterior axis with origin at the whole body COM), thereby countering the angular momentum that would develop during the swing phase. We propose the magnitude and timing of the first positive peak (P1) to be a consequence of stepping limb stiffness on foot-contact, since reduced limb compliance would increase the rate of axial loading and would more rapidly increase the ratio between the axial and mediolateral GRF components, thereby resulting in an earlier peak. We also propose stepping limb stiffness and, hence, the timing of this peak, to be pre-planned parameters, since the average time to peak (0.045 s) is too brief to be modulated by afferent sensory information upon foot-contact. The magnitude and timing of the second positive peak is, however, likely modulated by sensory information regarding the state of the COM at the onset, and throughout, the restabilisation phase.

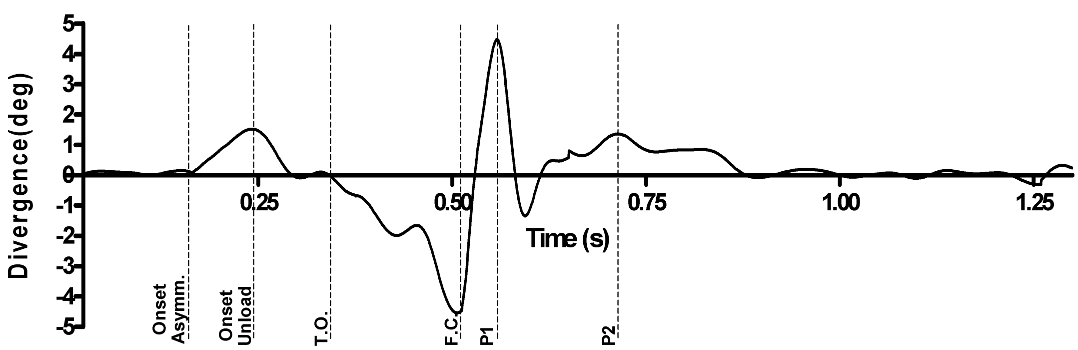


**Figure 5-5.** Representative trial from one subject (right-footed stepping), displaying the divergence between the GRF inclination angle and the COP-COM inclination angle. Positive values indicate a net GRF orientation that would tend to cause angular acceleration toward the stance limb side. Data begin at the instant of cable release and end at the ML restabilisation point. Onset Asymm. = onset of ML asymmetry; Onset Unload = onset of swing limb unloading; T.O. = toe-off; F.C. = foot-contact; P1 = peak of the first positive (passive) phase, immediately following foot-contact; P2 = peak of the second positive (active) phase, following foot-contact.

### *5.3.5 Statistical Analyses*

The statistical analyses are described in reference to each of the four dependent variables:

1. COM incongruity values for each trial were averaged within-subject for each block of 5 trials and were subsequently analysed with a three-factor ANOVA, with repeated measures [2 within group factors (step – 2 levels)(time – 2 levels); 1 between group factor (age – 2 levels)]. If the omnibus ANOVA revealed a significant interaction effect, follow-up independent and paired samples t-tests were used, as appropriate, to localize the differences. Measures of effect size were estimated using the formula for Cohen's d, using the means and standard deviations for each group. To counter the effect of alpha-inflation due to multiple subsequent comparisons, we employed a Bonferroni correction whereby the alpha level for a family of follow-up analyses was divided by the number of comparisons within that family ($p = 0.0125$, for evaluation of interaction effects).

2. Standard deviations, representing within-subject trial-to-trial variability, were computed for each subject in each group and were analysed with three-factor ANOVA, with repeated measures [2 within group factors (step – 2 levels)(time – 2 levels); 1 between group factor (age – 2 levels)]. Interaction effects were analysed with Bonferroni-corrected independent and paired samples t-tests ($p = 0.0125$). When appropriate, measures of effect size were calculated using the formula for Cohen's d, as above.

3. RMSD values were averaged within-subject and were analysed with three-factor ANOVA, with repeated measures [2 within group factors (step – 2 levels)(time – 2 levels); 1 between group factor (age – 2 levels)]. Interaction effects were analysed with Bonferroni-corrected independent and paired samples t-tests ($p$ = 0.0125). When appropriate, measures of effect size were calculated using the formula for Cohen's d, as above.

4. Values representing the time from foot-contact to the P2 peak were averaged within-subject and were analysed with a three-factor ANOVA [2 within group factors (step – 2 levels)(time – 2 levels); 1 between group factor (age – 2 levels)]. Interaction effects were analysed with Bonferroni-corrected independent and paired samples t-tests ($p$ = 0.0125). When appropriate, measures of effect size were calculated using the formula for Cohen's d, as above.

### *5.3.6 Secondary Analyses*

Although bilateral EMG activities of the medial and lateral gastrocnemius were monitored during the period preceding cable release, we compared the mean of the LE-EMG during each experimental trial against that which was obtained during quiet upright standing trial. We assumed equivalence in EMG activation if the mean LE-EMG of the perturbation trial did not exceed two standard deviations above of the mean of the quiet upright standing trial. This analysis was performed on a trial-by-trial basis for each experimental trial.

It remained possible that, despite previous work indicating similarity between young and older adults in the initial temporospatial parameters of the stepping response (i.e. until foot contact), differences between young and older adults may have existed and may have partially explained any potential differences in the variables of interest. To expose this possibility, post-hoc tests were performed to assess between-group differences in the AP distance between the COP and vertical projection of the COM at the instant before cable-release, ML APA amplitude (if present), effect of the ML APA on the peak lateral displacement of the COM prior to step onset, time to onset of ML asymmetry and unloading, step time, length and width, peak AP swing foot velocity, in addition to the AP and ML COM velocity at the instant before foot-contact.

As noted previously, we hypothesized that the direction of the resultant ground reaction force itself exerts a large influence on an individual's ability to regain stability after foot contact. From our previous work (study 2), we believe the magnitude of the divergence between the GRF and COP-COM inclination angles and the ability to reorient the net ground reaction force at the appropriate time may be of equal importance. The timing and magnitude of key time points of the waveform representing the difference between the inclination angles of the COP-COM and the net GRF vector were analyzed.

## 5.4 RESULTS

### *5.4.1 Incongruity Magnitude – Effect of age, step condition, and trial repetition*

The omnibus ANOVA for the analysis of incongruity magnitude revealed a main effect of age, $F(1,38) = 22.10$, $p < 0.001$, $\eta_p^2 = 0.37$, with older adults exhibiting an increased incongruity magnitude (Fig. 5-6). There was also a main effect of time, $F(1,36) = 21.59$, $p < 0.001$, $\eta_p^2 = 0.36$, whereby the incongruity magnitude was reduced in the second block of repeated trials, relative to the first (Fig. 5-7). There was no difference between preferred or reduced step width conditions.

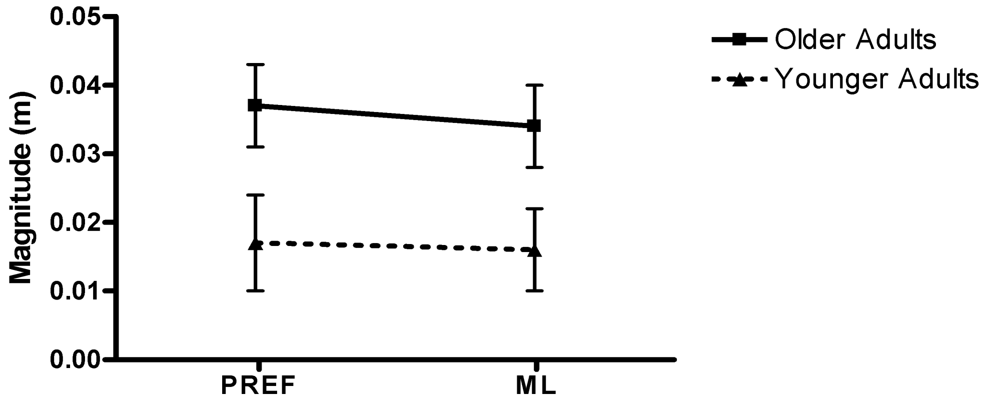


**Figure 5-6.** Mediolateral incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Positive values indicate overshoot. Task conditions include: preferred step placement (PREF); narrow step width (ML).

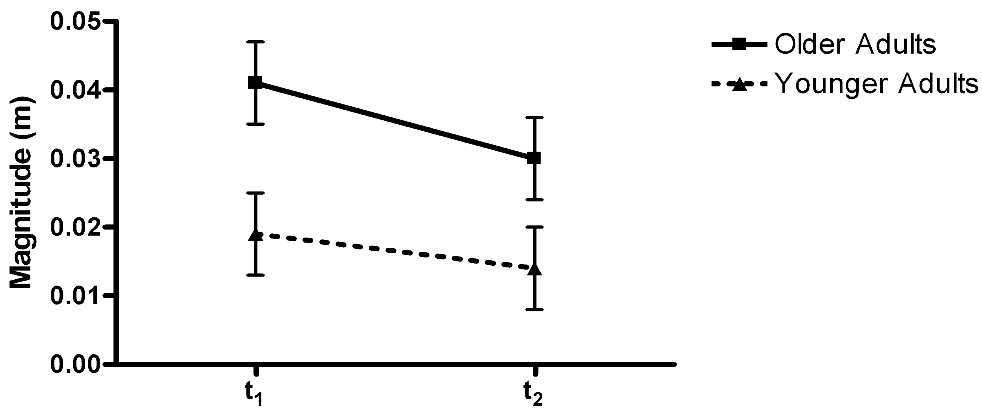


**Figure 5-7.** Mediolateral incongruity magnitude for older (solid line) and younger adults (dotted line), averaged within first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean. Positive values indicate overshoot.

### *5.4.2 Trial-to-Trial Variability of Incongruity - Effect of age, step condition, and trial repetition*

As hypothesized, a main effect of age indicated that older adults had greater trial-to-trial variability of incongruity than did the young adults, $F(1,38) = 26.49$, $p < 0.001$, $\eta_p^2 = 0.41$ (Fig. 5-8). Additionally, a main effect of time indicated that there was greater trial-to-trial variability of incongruity in the first block of trials relative to the second, $F(1,38) = 24.20$, $p < 0.001$, $\eta_p^2 = 0.39$. These main effects were qualified by an interaction between age and time, $F(1,38) = 9.02$, $p = 0.005$, $\eta_p^2 = 0.19$. Follow-up independent- and paired-samples t-tests revealed that while older adults had greater variability than the young during both the first, $t(38) = 5.06$, $p < 0.001$, $d = 1.60$, and second, $t(38) = 2.92$, $p = 0.006$, $d = 0.92$, time blocks, only the older adults exhibited a reduction in trial-to-trial variability from the first to the second group of trials, $t(19)=4.832$, $p < 0.001$, $d = 1.35$ (Fig. 5-9). There were no main effects of step condition.

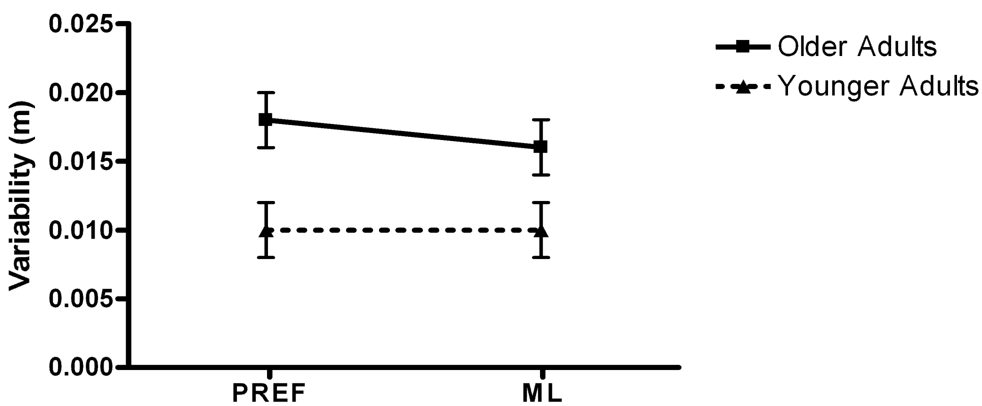


**Figure 5-8.** Trial-to-trial variability of mediolateral incongruity for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred step placement (PREF); narrow step width (ML).

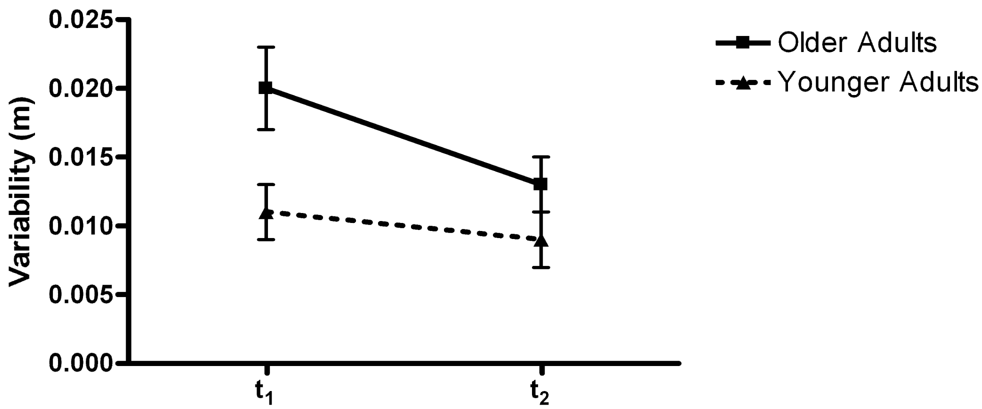


**Figure 5-9.** Trial-to-trial variability of mediolateral incongruity for older (solid line) and younger adults (dotted line), averaged within first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean. Positive values indicate overshoot.

### *5.4.3 RMSD Ground Reaction Force Inclination Angle with Respect to the COM – Effect of age, step condition, and trial repetition*

There were no age-related differences in the direction of the net GRF vector with respect to the COM, as was quantified by examining the RMSD of the difference between the GRF inclination angle and the COP-COM inclination angle. There was no effect of step condition, however there was a main effect of time, $F(1,38) = 4.31$, $p = 0.045$, $\eta_p^2 = 0.10$, as the second block of trials had a reduced RMSD with respect to the first block (Fig. 5-10).

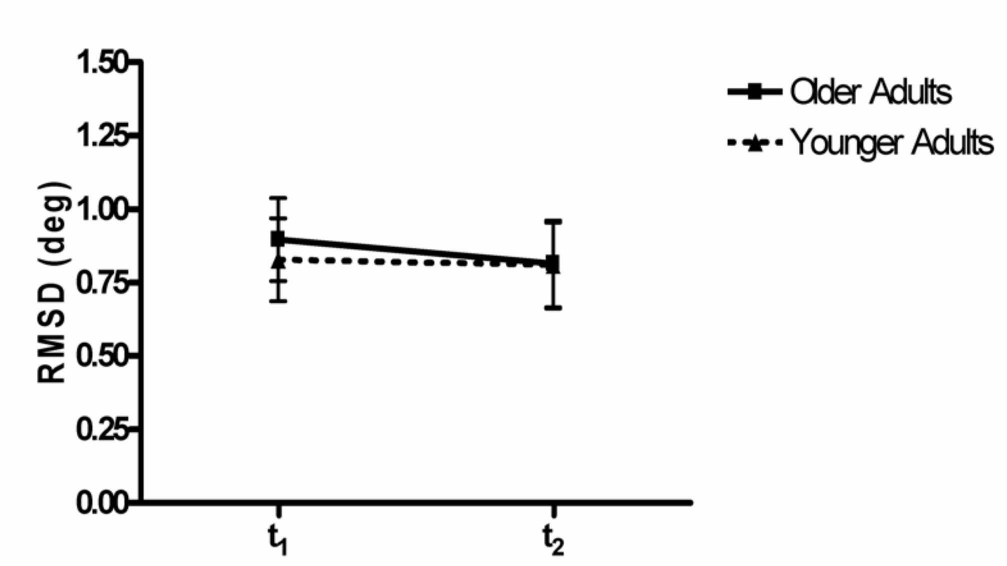


**Figure 5-10.** Root-mean-square deviation of the inclination angle of the net ground reaction force with respect to the inclination angle formed by a line joining the centre of pressure and centre of mass. Data are depicted for older (solid line) and younger adults (dotted line), averaged within the first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean.

### *5.4.4 Timing of 'Active' Reorientation of the Net Ground Reaction Force (P2) – Effect of age, step condition, and trial repetition*

Examination of the time from foot-contact to P2 revealed no effects of step condition or time. There was a main effect of age, $F(1,38) = 9.69$, $p = 0.004$, $\eta_p^2 = 0.20$, with older adults exhibiting a longer time to peak than did the younger adults (Fig. 5-11).

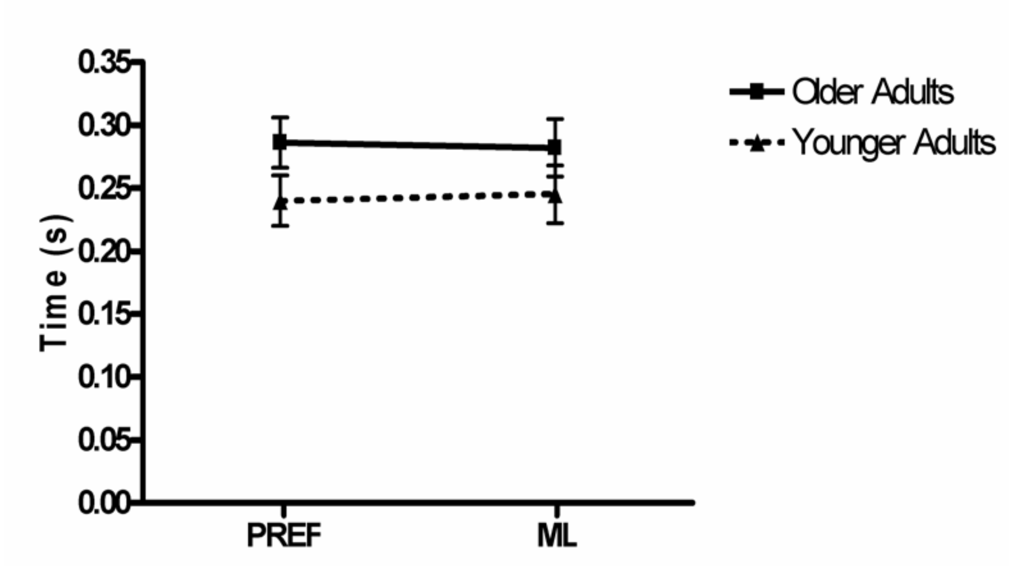


**Figure 5-11.** Timing of the second peak divergence between the inclination angles of the net ground reaction force and the COP-COM following foot-contact (P2). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

### *5.4.5 Secondary Analyses*

#### *5.4.5.1 Initial Conditions*

Independent-samples t-tests did not reveal significant between-group differences in body mass, height, or lean angle in either the PREF or ML condition (Table 5-1). Nevertheless, the (non-significant) between-group difference in body mass, in isolation, would not alter the initial lean angle and the initial destabilizing moment about the ankle upon cable release, as the pre-perturbation cable load was standardized to body weight. Although there was particular attention paid to ensuring equivalence between groups in pre-perturbation parameters, analysis of the pre-perturbation LE-EMG magnitudes revealed that five of the twenty older adults performed trials in which the pre-perturbation LE-EMG magnitude exceeded two standard-deviations from the mean LE-EMG magnitude during the quiet standing trial performed before the experimental trials. This accounted for 18.5% of trials in the PREF condition and 17% of trials in the ML condition. In contrast, only two younger adults performed such trials, which accounted for only 2.5% of trials in each the PREF and ML conditions. Correspondingly, a three-factor ANOVA used to analyse the AP distance between the vertical projection of the COM and the position of the COP during the pre-perturbation interval revealed a main effect of age, $F(1,38) = 11.19$, $p = 0.002$, $\eta_p^2 = 0.23$, with older adults having a reduced AP COM-COP distance by 0.011 m, on average.

Table 5-1. Participant and lean-angle characteristics.

| | Younger Adults | | Older Adults | | |
|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | *p** |
| Mass (kg) | 68.11 | 10.31 | 76.98 | 16.76 | 0.051 |
| Height (m) | 1.71 | 0.081 | 1.70 | 0.050 | 0.763 |
| $\theta_{lean}$ (deg) [PREF] | 10.11 | 1.11 | 9.86 | 1.41 | 0.541 |
| $\theta_{lean}$ (deg) [ML] | 10.16 | 1.31 | 10.01 | 1.43 | 0.724 |

* *p*-values are reported for the independent-samples t-test, comparing the mean of each variable between age-groups.

*5.4.5.2 Initial Movement Parameters*

There were main effects of step condition, $F(1,38) = 70.52$, $p < 0.001$, $\eta_p^2 = 0.65$, and of age, $F(1,38) = 7.58$, $p = 0.009$, $\eta_p^2 = 0.17$, when examining the mediolateral anticipatory postural adjustment (ML APA), with increased amplitudes in the ML step condition and in the younger adults, respectively. These main effects were qualified by a step condition by age interaction, $F(1,38) = 11.15$, $p = 0.002$, $\eta_p^2 = 0.23$. Follow-up paired-samples t-tests indicated that both the younger, $t(19) = 6.61$, $p < 0.001$, $d = 0.92$, and older groups, $t(19) = 5.51$, $p < 0.001$, $d = 0.61$, increased the amplitude of the ML APA during the ML stepping condition. The relative increase, however, was greater for the younger adults, as the independent-samples t-tests revealed a difference within the ML stepping condition, $t(38) = 3.32$, $p = 0.002$, $d = 1.05$ (Fig. 5-12).

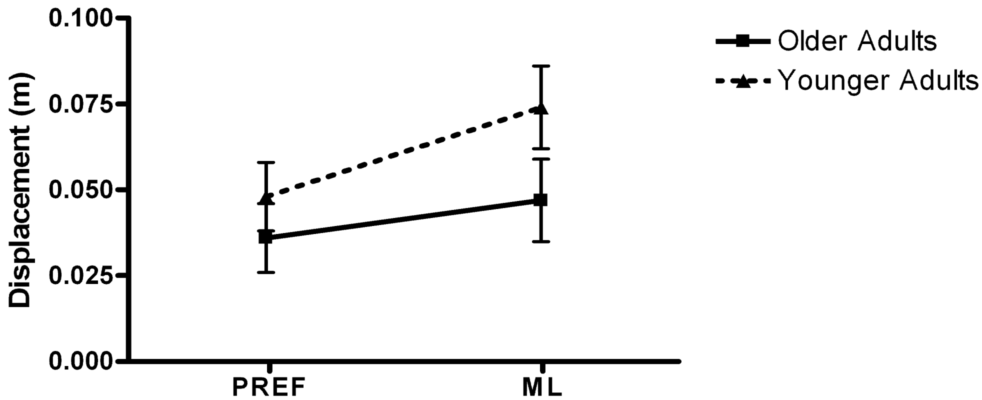


**Figure 5-12.** Mediolateral displacement of the centre of pressure prior to step-onset, referenced to the initial centre of pressure position prior to step initiation. Positive values indicate displacement along the positive x-axis of the global coordinate system, toward the swing limb. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

The age-related differences in ML APA amplitude did not result in a main effect of age when examining the effect of the ML APA on the ML COM trajectory prior to step initiation. There was, however, a main effect of step condition, $F(1,38) = 57.61$, $p < 0.001$, $\eta_p^2 = 0.60$, as all participants exhibited increased COM movement toward the stance-limb before step initiation in the ML step condition (Fig. 5-13). There was also a main effect of time, $F(1,38) = 10.55$, $p = 0.002$, $\eta_p^2 = 0.22$, with increases noted in the second block of trials, irrespective of age or step condition (Fig. 5-14).

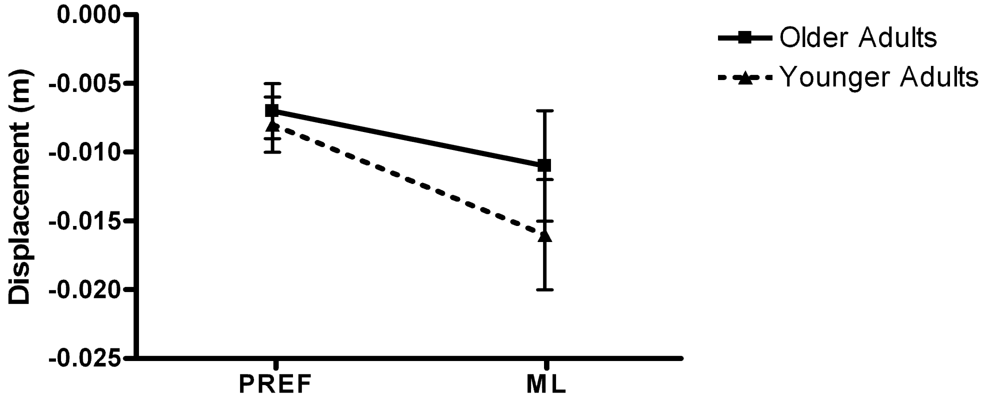


**Figure 5-13.** Mediolateral displacement of the centre of mass prior to step-onset, referenced to the initial COM position prior to movement initiation. Negative values indicate displacement along the negative x-axis of the global coordinate system, toward the stance limb. Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

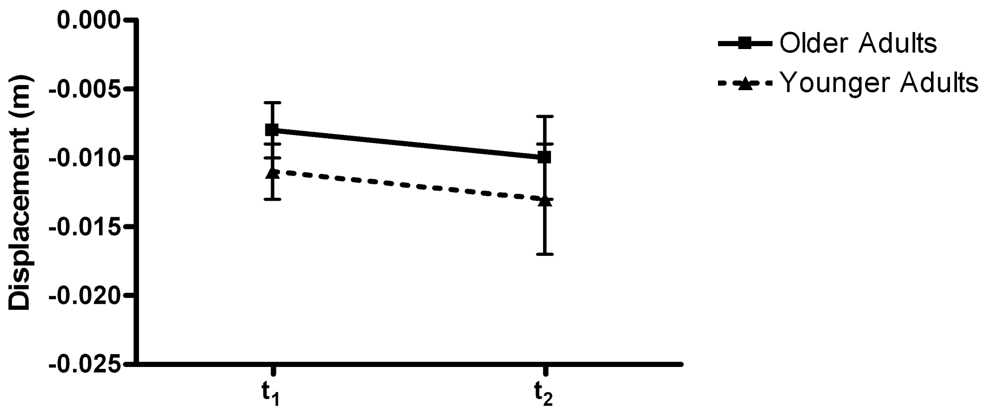


**Figure 5-14.** Mediolateral displacement of the centre of mass prior to step-onset, referenced to the initial COM position prior to movement initiation. Negative values indicate displacement along the negative x-axis of the global coordinate system, toward the stance limb. Data are depicted for older (solid line) and younger adults (dotted line), averaged within the first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean.

There were no interactions or main effects of age, step condition or time when examining the time to onset of ML asymmetry. The time to onset of ML unloading was delayed in the second block of trials, relative to the first, irrespective of age or step condition, $F(1,36) = 13.25$, $p = 0.001$, $\eta_p^2 = 0.26$.

There was no main effect of age or step condition on step length. There was a main effect of time, $F(1,386) = 12.60$, $p = 0.001$, $\eta_p^2 = 0.25$, with increased step lengths observed in the second block of trials, irrespective of step condition. Similarly, there was also no main effect of age on step width. A main effect of time was noted, $F(1,38) = 6.53$, $p = 0.015$, $\eta_p^2 = 0.15$, with individuals reducing step width in the second block of trials, relative to the first. There was also a main effect of step condition on step width, which emerged as a direct consequence of the reduced step width required by the experimental protocol, $F(1,38) = 155.60$, $p < 0.001$, $\eta_p^2 = 0.80$.

There were no main effects of age or time on the peak AP velocity of the swing foot. There was a main effect of step condition, $F(1,38) = 9.34$, $p = 0.004$, $\eta_p^2 = 0.20$, as there was a greater AP foot velocity observed in the ML step condition. There were no main effects or interactions when examining step time. Correspondingly, there were no main effects or interactions when examining swing foot path length.

Despite the small number of age-related differences in initial movement parameters up to the point of foot-contact, there were no main effects of age when examining either the AP or ML COM velocity just prior to foot-contact. All participants exhibited a reduced lateral (towards the stepping limb) COM velocity at foot-contact in the ML stepping condition, $F(1,38) = 130.95$, $p < 0.001$ $\eta_p^2 = 0.78$. There was also a main effect of time for both the AP and ML COM velocity prior to foot-contact, with an increased AP magnitude, $F(1,38) = 5.51$, $p = 0.024$, $\eta_p^2 = 0.13$, and reduced ML magnitude, $F(1,38) = 4.36$, $p = 0.044$, $\eta_p^2 = 0.10$ , in the second block of trials, relative to the first.

*5.4.5.3 Kinetics*

As the RMSD of the net GRF inclination angle relative to the COP-COM inclination angle did not explain the observed age-related differences in COM incongruity magnitude, we analysed the mean and variability in the amplitude and timing (with respect to foot-contact) of what were believed to be three key peaks in the waveform representing the divergence between the GRF inclination angle and the COP-COM inclination angle (Fig. 5-5).

There were no main effects of age, step condition or time when examining the amplitude of the first negative peak, at foot-contact. There was a significant age by time interaction effect, $F(1,38) = 8.62$, $p = 0.006$, $\eta_p^2 = 0.19$. Follow up paired-samples t-tests indicated that only the older adults reduced the divergence between two inclination angles from the first block to the second block of trials, $t(19) = -3.85$, $p = 0.001$, $d = 0.43$, which also resulted in a reduced magnitude of divergence relative to the young adults during the second block of trials, $t(38) = -2.58$, $p = 0.014$, $d = 0.82$ (Fig. 5-15). There were no age-related effects when examining the variability in the magnitude of this peak.

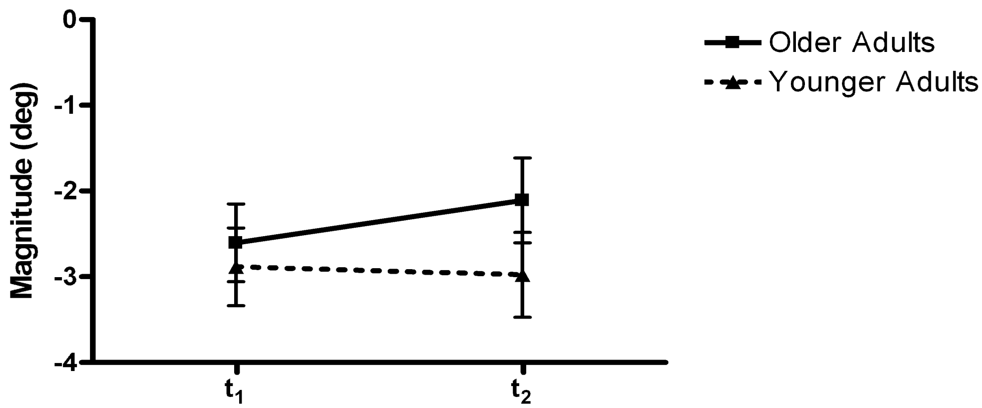


**Figure 5-15.** Magnitude of divergence between the inclination angles of the net ground reaction force and the COP-COM at the instant of foot-contact. Data are depicted for older (solid line) and younger adults (dotted line), averaged within the first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean.

Evaluation of the magnitude of the first positive peak after foot-contact revealed no interactions or main effects of age or time. There was a main effect of step condition, $F(1,38) = 23.13$, $p < 0.001$, $\eta_p^2 = 0.38$, as this peak had a greater magnitude in the ML step condition. There were also no effects of age, step condition or time when evaluating the variability in the magnitude of this peak.

Evaluation of the time from foot-contact to P1 revealed that there was a main effect of age, $F(1,38) = 5.50$, $p = 0.024$, $\eta_p^2 = 0.13$, that was qualified by interactions between age and step condition, $F(1,38) = 5.83$, $p = .021$, $\eta_p^2 = 0.13$, and age and time, $F(1,38) = 4.89$, $p = 0.033$, $\eta_p^2 = 0.11$. Follow-up independent-samples tests for the interaction between age and step condition indicated that there was no difference between the younger and older adults during the PREF stepping condition, while the older adults had reduced time to peak during ML stepping, $t(38) = 3.23$, $p = 0.003$, $d = 1.02$. Paired-samples t-tests did not reveal differences by step condition within either age group (Fig. 5-16). Further, independent-samples t-tests for the interaction between age and time revealed no difference between the younger and older adults during the first block of trials, while the older adults exhibited a reduced time to peak during the second block of trials, $t(38) = 3.04$, $p = 0.004$, $d = 0.96$. Paired-samples t-tests did not reveal differences by time block within either age group (Fig. 5-17). There were no effects of age, step condition or time when examining the variability in timing of this peak.

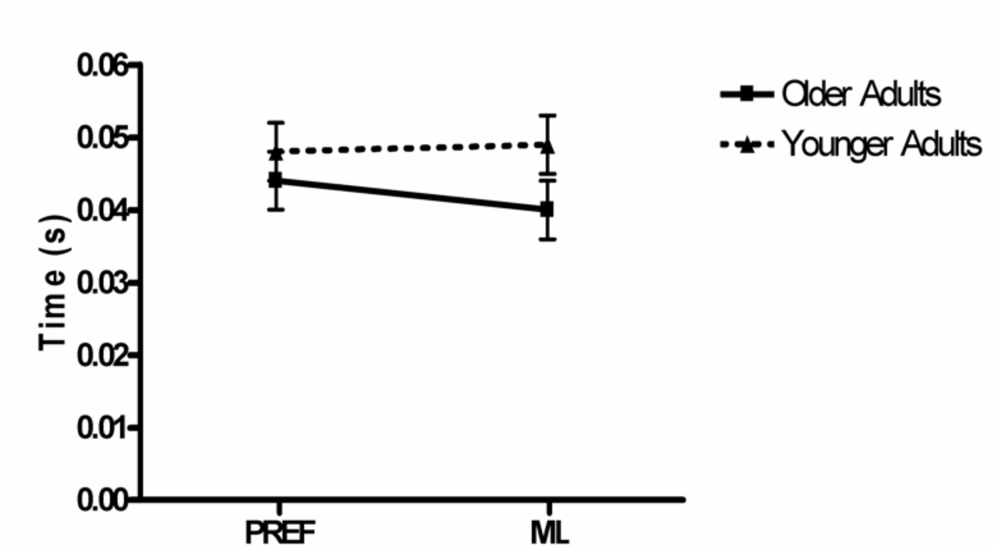


**Figure 5-16.** Timing of the first peak divergence between the inclination angles of the net ground reaction force and the COP-COM following foot-contact (P1). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

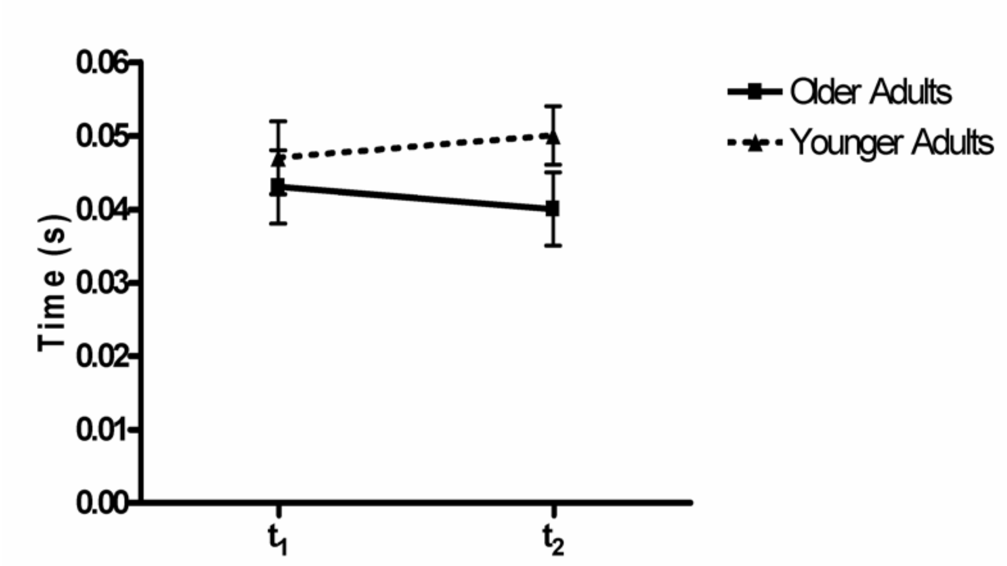


**Figure 5-17.** Timing of the first peak divergence between the inclination angles of the net ground reaction force and the COP-COM following foot-contact (P1). Data are depicted for older (solid line) and younger adults (dotted line), averaged within the first ($t_1$) and second ($t_2$) blocks of repeated trials. Error bars represent the 95% confidence interval for the mean.

Analyses of the magnitude of the second positive peak after foot-contact (P2) only revealed a main effect of time, $F(1,38) = 37.28$, $p < 0.001$, $\eta_p^2 = 0.50$, as all participants had a reduced the magnitude of this peak in the second block of trials, relative to the first. There were no other main effects or interactions in either amplitude or variability.

Examination of the variability time from foot-contact to P2 revealed a main effect of age, $F(1,38) = 7.14$, $p = 0.001$, $\eta_p^2 = 0.16$, as older adults had greater variability in the timing of this peak than did the younger adults (Fig. 5-18).

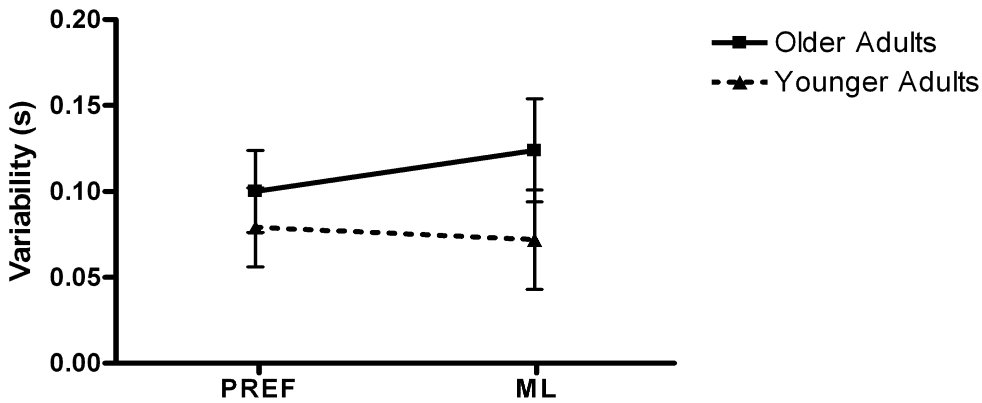


**Figure 5-18.** Trial-to-trial variability in the timing of second peak divergence between the inclination angles of the net ground reaction force and the COP-COM following foot-contact (P2). Data are depicted for older (solid line) and younger adults (dotted line), averaged within condition. Error bars represent the 95% confidence interval for the mean. Task conditions include: preferred speed stepping with preferred step placement (PREF); rapid speed stepping with preferred step placement (PREF_RAPID); rapid speed stepping with narrow step width (ML_RAPID).

## 5.5 DISCUSSION

The purpose of the current study was to examine age-related differences in the control of mediolateral dynamic stability during the restabilisation phase of perturbation-evoked stepping. We sought to further explore the effects of age by including trials in which individuals stepped with reduced step width, which we believed would pose an increased challenge to the maintenance of stability. Secondarily, we attempted to probe the extent and means by which individuals could alter the dynamics of the compensatory stepping response over time, with trial repetition. Results revealed that overshoots of the final COM position were common among both age groups in all stepping conditions, but were subsequently reduced with trial repetition. Older adults, however, exhibited an increased magnitude of overshoot relative to the younger adults. Older adults also exhibited greater trial-to-trial variability of COM incongruity than the younger adults, which was also reduced with trial repetition. Examination of the kinetic data in concert with these kinematic variables provided support for previous work, which has suggested that overshoot may be a means to simplify the control of mediolateral dynamic stability during the restabilisation phase of stepping. Increased overshoot and intertrial variability in COM incongruity exhibited by older adults, however, is proposed to be a consequence of age-related differences in the timing and scaling of the ground reaction force, which may be indicative of difficulties with reactive control during the restabilisation phase. Reductions in overshoot and intertrial variability, specifically among the older adults, are proposed to occur primarily through the alteration of pre-planned movement parameters and swing phase dynamics.

### *5.5.1 Initial Conditions and Initial Movement Parameters*

Despite the utmost attempts to ensure the equivalence of initial conditions before perturbation onset, results of the secondary analyses indicated that there were group differences in these variables. Specifically, the older adults exhibited increased soleus LE-EMG amplitude during the period before perturbation onset, which likely resulted in the reduced AP distance between the COM and COP (1.1 cm, on average, between groups). Furthermore, the younger participants exhibited a larger amplitude APA, but this did not result in group differences in peak lateral COM displacement toward the stance limb. There were no age-related differences in the time to onset of ML asymmetry, time to onset of ML unloading, step length, step width or step time. Despite the abovementioned differences in a small number of initial movement parameters, there were no age-related differences in the AP or ML COM velocity at the instant of foot contact. As a result, we believe that subsequent analyses of age-related differences in dynamic stability control during the restabilisation phase would not be confounded by group differences in initial movement parameters.

### *5.5.2 Incongruity Magnitude*

As per our previous work (Singer et al., 2012), we found that overshoots were predominant among both age groups. During volitional stepping, overshoot during the restabilisation phase has been suggested to play an important role in the simplification of mediolateral dynamic stability control, in that it may limit the frontal plane gravitational moment about the stepping limb and the subsequent acceleration. Similarly, during steady-state gait, such a reduction of the frontal plane gravitational moment may have important implications for simplifying mediolateral stability control during single limb support and step-to-step transitions (Donelan et al., 2002; Donelan et al., 2004). This strategy may also be beneficial during compensatory stepping, in the event that the forward COM velocity cannot be arrested within the first step and an additional step with the stance limb becomes necessary to achieve stability. Such multiple alternating step responses have been shown to be common among older adults (Luchies et al., 1994; McIlroy and Maki, 1996; Maki & McIlroy, 1999; Rogers et al., 2001; King et al., 2005; Schulz et al., 2005) and were also employed by older adults in early trials of the present study. From this, the larger magnitude of overshoot among the older adults may have emerged to ensure mediolateral stability, given the increased potential for sagittal plane instability and consequent multiple alternating step responses. Subsequent reductions in overshoot with practice may suggest that this strategy was modified once individuals learned that the perturbation magnitude was unlikely to result in sagittal plane instability subsequent to the first step.

Alternatively, while overshoot could be generally advantageous, the increased overshoot exhibited by the older adults may be maladaptive, as the COM would be displaced to a greater extent toward the lateral stability limits. This is especially true when considering the trials with reduced step width, since step width reduction, in concurrence with the equality of the magnitude of overshoot between stepping conditions, would place the COM even closer to the support limb at the point of peak lateral COM displacement. This could be considered somewhat beneficial, as it would considerably reduce the frontal plane gravitational moment in the event of subsequent steps. In contrast, the reduction of step width, per se, would have little bearing on the necessity for such additional forward steps, assuming equality of step length and step time, as the anterior perturbation magnitude was equivalent across step conditions. Given this, such a strategy to further limit the frontal plane gravitational moment when stepping with reduced step width would have little practical value. Previously reported reductions in stability limits among older adults (Pai et al., 1998; Jensen et al., 2001) – either real or perceived – would suggest that the most prudent strategy for the maintenance of mediolateral stability would be to maintain the COM at a safe distance from the lateral border of the BOS. Further, both age groups reduced the magnitude of overshoot with practice: the older group exhibited nearly twice the reduction as the young, while the young reduced the magnitude to values approaching those observed in our previous work on volitional stepping (study 1). This was unlikely to have occurred if the increased overshoot exhibited by the older adults was advantageous for dynamic stability control. From this, we propose that the initial increased magnitude of overshoot was unintentional, but that overshoot in general, and the reduced magnitude exhibited in later trials, may have a practical function.

### *5.5.3 Origins of Increased Incongruity Magnitude: Reactive Control*

Of particular interest, when exploring the origin of increased mediolateral COM incongruity, was the increased time to P2 exhibited among older adults. While it is possible that these individuals deliberately increased the time to peak, which may permit greater overshoot and aid in mediolateral stability control, as previously outlined, we do not believe this to be the case. Rather, if P2 does represent an active response to the state of the COM during the restabilisation phase, as proposed, the increased time to peak could signify difficulties in sensing the position/velocity of the COM, slowing of conduction velocity (Inglis et al., 1994), impairment of central processing (Horak et al., 1997), multisensory integration (Horak et al., 1989), or reduced rate of force development (Chang et al., 2005) due to muscle atrophy, slowing of contractile properties or impaired coordination (Barry et al., 2005), all of which have been previously documented.

Such difficulty in stability control among older adults was also signalled by the increased trial-to-trial variability in incongruity magnitude. While early research attributed movement variability to random equipment or physiological noise, more recent research is beginning to suggest that variability represents the underlying neural control and may result from a lack of coordination between the components of the control system (Brach, Berlin, VanSwearingen, Newman & Studenski, 2005; McGibbon, Krebs & Wagenaar, 2005; Hausdorf, 2005; Hausdorf, 2007). For the current study, we believed that increased intertrial variability in COM incongruity would result from difficulty with reactive control during the restabilisation phase. The finding of increased intertrial variability in the time to the P2 peak among older adults provides support for this hypothesis and for previous studies, which have suggested that challenges to mediolateral stability control arise after stepping foot contact (Pai et al., 1998; Maki and McIlroy, 1999).

### *5.5.4 Reductions in Incongruity and Variability: Augmented Movement Planning*

Because of the potential difficulty with reactive control during the restabilisation phase, practice-related reductions in COM incongruity and intertrial variability among older adults may have arisen from augmented movement planning and reduced reliance on reactive control during the restabilisation phase. Such a suggestion is given support by the kinetic data, as both age groups also exhibited practice-related reductions in the magnitude of the second peak (P2) after foot contact in addition to the RMSD. If P2 is indeed modulated in response to the state of the COM during the restabilisation phase, this finding may indicate that there was a reduced need to counteract the angular momentum that developed during swing phase.

Such a reduction may have arisen from either of two mechanisms. Firstly, individuals may have become more adept at directing the GRF at the COM with trial repetition, as was evidenced by the reduction of RMSD values. Alternatively, it is more likely the practice-related reduction in COM overshoot occurred because individuals became familiar with the predictable features of the perturbation and relied less on reactive control during the restabilisation phase. Specifically, the older adults exhibited practice-related reductions in the magnitude of divergence between GRF- and COP-COM inclination angles at the instant of foot-contact, in addition to a reduction in the time to P1 peak – variables that we have proposed to be modified by the mediolateral anticipatory postural adjustment and swing phase dynamics and limb stiffness on foot-contact, respectively. This is consistent with studies of standing balance and downward stepping, in which increased coactivation and limb stiffness were thought to represent strategies employed by older adults to compensate for impaired neuromotor function (Hortobagyi & DeVita, 2000; Benjuya, Melzer & Kaplanski, 2004; Cenciarini, Loughlin, Sparto & Redfern, 2009; Cenciarini, Loughlin, Sparto & Redfern, 2010). As these variables are regulated by events that occur before foot-contact, is possible that the observed changes represent a proactive strategy in attempt to offset instability during the restabilisation phase, which may be linked to the increased time to achieve the second 'active' peak divergence between the net GRF and COP-COM inclination angles (P2). While these results suggest that practice-related improvements in older adults may arise from the alteration of pre-planned movement parameters, it remains possible that such improvements were also gained by the enhancement of sensory-based feedback control or sensory re-weighting, as proposed by Mansfield, Peters, Liu and Maki (2010). Future research should attempt to disentangle the origins of practice-related augmentations in stability control, perhaps by determining whether improvements acquired at one perturbation magnitude are transferable to other magnitudes.

### *5.5.5 Overall Implications*

In general, the results of this study suggest that while overshoot may help to simplify mediolateral stability control during the restabilisation phase, the increased overshoot observed among the older adults and in early trials may arise because of difficulties in reactive control after foot contact. While stepping with reduced step width did not increase the magnitude of overshoot, it does pose a challenge to the maintenance of stability, as a narrower step in association with an equal magnitude of overshoot would tend to reduce the mediolateral distance between the COM and lateral stability limits. While reductions in COM incongruity and variability with trial repetition may occur because of augmentation of reactive control, the results of the current study suggest that such reductions were brought about predominantly by alterations in pre-planned movement parameters.

The increased trial-to-trial variability among older adults does have particularly important implications with respect to the potential for the development of mediolateral instability after foot contact. This is especially apparent when viewed in concert with the abovementioned increased magnitude of overshoot relative to the younger adults, and the relationship between the COM and BOS in trials with reduced step width. Despite its utility as a mechanism for simplification of mediolateral stability control, increased overshoot coupled with increased variability may occasionally lead to compensatory stepping responses in which the COM exceeds the lateral stability limits. Such lateral instability after foot contact is consistent with previous research examining older adults (Maki et al., 1994; McIlroy & Maki, 1996; Lord et al., 1999; Rogers et al., 2001; Rogers & Mille, 2003; Tirosh & Sparrow, 2004; Schulz et al., 2005; Tirosh & Sparrow, 2005). As instability and falls are intermittent events, measures of variability may provide a more sensitive measure of stability and fall risk. Future studies should be directed toward examining this potential.

# CHAPTER 6

# General Discussion

## 6.1 SUMMARY OF RESEARCH FINDINGS

The global objective of this thesis was to develop further understanding regarding mediolateral stability control during the restabilisation phase of stepping. More specifically, we sought to uncover age-related differences in COM kinematics and applied force generation during the restabilisation phase of both volitional and reactive stepping, which may elucidate factors associated with instability and fall risk. Despite the differences between the two forms of stepping, we believed that the challenges for effectively controlling and arresting the lateral progression of the COM within the base of support may have important links to the restabilisation phase and would be common to both volitional and reactive stepping.

To date, much of the research exploring voluntary gait- or step-termination in healthy adults has focussed on the sequence of lower limb muscle activation (Hase and Stein, 1998; Bishop, Brunt, Pathare & Patel, 2002; Bishop, Brunt, Pathare & Patel, 2004; Chu et al., 2009), ground reaction forces (Jaeger & Vanitchatchavan, 1992; Crenna, Cuong & Breniere, 2001) or the relationship between the COP and COM necessary to arrest the forward progression of the COM (Jian et al., 1993; Oates, Patla, Frank & Greig, 2005). Further, studies of age related differences in the ability to control the COM, during either voluntary or reactive stepping, have restricted analysis to either the sagittal plane (Cao et al., 1998; Chu et al., 2009; Menant et al., 2009; Carty et al., 2011) or to the initial phase of movement, until the time of foot contact (McIlroy & Maki, 1996). Given the evidence suggesting that control of mediolateral stability may be particularly problematic for older adults (Maki et al., 1994; McIlroy et al., 1996; Lord et al., 1999; Rogers et al., 2001;

Rogers & Mille, 2003; Tirosh & Sparrow, 2004; Schulz et al., 2005; Tirosh & Sparrow, 2005) and those with balance disorders (Kaya et al., 1998; O'Kane et al., 2003), coupled with the suggestion that instability may arise independently of many of the initial movement parameters prior to foot-contact (McIlroy & Maki, 1996; Thelen et al., 1997; Thelen et al., 2000; Rogers et al., 2001), we proposed that the control of the COM during the restabilisation phase of stepping was likely a central determinant of age-related differences in mediolateral dynamic stability.

We initiated this line of inquiry by quantifying the kinematics of the COM during the restabilisation phase of a voluntarily initiated-single step in a sample of healthy young adults. At the outset of this initial work, we believed that we would observe no COM incongruity during conditions where stepping was executed with preferred speed and step placement, as individuals had the opportunity to plan movement parameters for maximal stability. We believed that stepping with non-preferred step placement would increase the challenges associated with COM control and would lead to increased mediolateral incongruity and trial-to-trial variability. We also believed that, when afforded the opportunity to practice under non-preferred stepping conditions, we would observe subsequent reductions in incongruity, as individuals became familiar with the necessary movement dynamics during the restabilisation phase to achieve stability without under- or overshoot. Contrary to our hypotheses, mediolateral COM incongruity was common among all trials, with a bias toward overshoot. Non-preferred step placement did lead to increased incongruity, however there was neither a change in trial-to-trial variability nor were there reductions in COM incongruity with trial repetition. Together, these results suggested that overshoots likely served the functional role of simplifying reactive control during the restabilisation phase.

With this knowledge, we sought to determine how the abovementioned kinematic variables were associated with age-related declines in stability control. Further, we analysed the orientation of the net ground reaction force vector with respect to the COM to help clarify the mechanisms responsible for any age-related differences in the COM kinematics. We initially believed that effective control of the COM would be achieved by directing the line of action of the net ground reaction force through the COM to minimize the net external moment about the COM, and consequent changes in angular momentum. In this regard, it was possible that under- and overshoot of the final COM position may be linked to the polarity of the net external moment about the COM. Contrary to our hypotheses, we found that the magnitude of overshoot and the trial-to-trial variability of incongruity were greater among the older adults. This was particularly true during rapid stepping conditions, which minimized the potential for anticipatory ML stabilisation prior to step-onset and likely shifted the burden of COM control to the restabilisation phase. While increased COM incongruity could be a means to simplify reactive control, as previously proposed, we believed that such age-related differences were likely related to a reduced divergence between the inclination angles of the GRF and the COP-COM during the restabilisation phase and/or to the increased time to ‘actively’ reorient the GRF, which was necessary to oppose the angular momentum about the COM that developed during the swing phase. Such mechanisms would increase the time necessary to arrest the lateral progression of the COM as it approached the lateral stability limits, which could account for the greater overshoot. Particularly interesting was the fact that older adults appeared to make alterations to the initial movement parameters, which may represent an attempt to offset subsequent mediolateral instability that may emerge after foot-contact. Despite this, increased incongruity did manifest within this group during the restabilisation phase.

When the potential for mediolateral anticipatory postural adjustments was further minimized during perturbation-evoked stepping, we observed further increases in both COM incongruity and trial-to-trial variability among the older adults, as hypothesized. With trial repetition, however, we observed a reduction in incongruity within both groups. Such reductions among the older adults occurred in company with a reduction in the divergence of the net GRF inclination angle with respect to the COM at foot-contact and an earlier time-to-peak divergence immediately following foot-contact. As these variables may be related to the ML APA, swing phase dynamics and limb stiffness at foot-contact, it is possible that reductions in incongruity with trial repetition arose from a proactive strategy to offset difficulties with ML stability control during the restabilisation phase, as individuals learned the predictable features of the perturbation.

## 6.2 SIGNIFICANCE

The present work provides support for previous studies, which have suggested that the control of mediolateral stability may be especially challenging for older adults (Maki et al., 1994; McIlroy & Maki, 1996; Lord et al., 1999; Rogers et al., 2001; Rogers & Mille, 2003; Tirosh & Sparrow, 2004; Schulz et al., 2005; Tirosh & Sparrow, 2005; Perry, Radtke, McIlroy, Fernie & Maki, 2008). Further, our work provides evidence that the challenges associated with ML stability control have important links to the restabilisation phase and are common to both volitional and reactive stepping. We suggest that the orientation of the net ground reaction force with respect to the COM to be an important variable for effective control of the COM during the restabilisation phase. Specifically, the deviation of the line of action of the net ground reaction force with respect to the COM serves to generate an external moment about the COM to reverse the polarity of the angular momentum, which developed during the swing phase. As such, and contrary to our hypotheses, we suggest that it is likely the regulation, rather than minimization, of changes to angular momentum that is highly important in effecting restabilisation. This is in contrast to the "zero moment" control scheme for dynamic stability in the robotics literature (Popovic, Goswami & Herr, 2005), but is consistent with experimental studies quantifying angular momentum in humans, as angular momentum is manipulated to achieve a particular state of motion (Pijnappels et al., 2004; Herr & Popovic, 2008; Mathiyakom, McNitt-Gray, 2008; Pijnappels, Kingma, Wezenberg, Reurink & van Dieen, 2010). Of chief importance, then, is the ability to direct angular momentum to zero, when necessary.

The particular difficulty older adults face during restabilisation from either internal or external perturbation appears not to be centred on the magnitude of the deviation of the line of action of the net GRF with respect to the COM (i.e. external moment), but with the time required to carry out such net GRF reorientation. As such, healthy older adults appear to make balance corrections of appropriate magnitude to restabilise, but do so with inappropriate timing to limit the lateral progression of the COM. The increased overshoot exhibited by older adults during reactive stepping, relative to younger adults and to volitional stepping trials, suggests that the ability to effectively time the components of the restabilisation response becomes especially important during the time-sensitive nature of recovery from an external perturbation.

Despite the importance of reactive control during the restabilisation phase, one cannot overlook the role of anticipatory control or proactive strategies, which, given prior knowledge regarding the nature of the postural disturbance, would alter the state of the COM at foot-contact to augment mediolateral stability. The modification of proactive strategies with task exposure has been well documented, and has been observed to occur during slip-related loss of balance (Pavol & Pai, 2002; Pavol, Runtz, Edwards & Pai, 2002; Pai, Wening, Runtz, Iqbal & Pavol, 2003) and platform perturbations (Laessoe & Voigt, 2008). In regard to the present work, we observed the use of proactive strategies for mediolateral stability control among older adults during both volitional and reactive stepping. During volitional stepping, older adults more greatly increased the displacement of the COM toward the stance limb during step-initiation, which reduced the lateral COM velocity and increased the distance between the COM and lateral BOS limits at foot-contact. During reactive stepping, older adults may have modified swing phase dynamics and increased stepping limb stiffness at foot-contact, which may have served to limit changes in angular momentum during the swing phase and caused an earlier onset of the external moment about the COM, respectively. Both of these mechanisms may have the potential to offset instability which may stem from difficulty with reactive control during the restabilisation phase. It is important to note, however, that while increased muscle coactivation and limb stiffness may be a common strategy among older adults to compensate for altered neuromotor function (Hortobagyi & DeVita, 2000; Benjuya et al., 2004; Cenciarini et al., 2010), such a strategy may impair the ability to make subsequent corrective responses should the initial step be insufficient to restore stability (Huang & Ahmed, 2010). From this, it is possible that increased limb stiffness to bring about an earlier onset of the P1 peak, may in fact hamper the ability to make subsequent modifications to the orientation of the net GRF (viz., timing of P2 peak). Moreover, the previously reported association between multiple step responses and fall risk (McIlroy & Maki, 1996; Maki et al., 2001; Schulz et al., 2005) may in fact stem from such increased limb stiffness, in that coactivation may alter the amplitude and timing characteristics of the P2 peak, while also impeding upon the ability to effectively reposition the limb for subsequent steps, should the initial step be insufficient to restore stability.

## 6.3 IMPLICATIONS FOR IMPROVING STABILITY CONTROL

Given the potential ramifications of falls among older adults, there has been considerable work toward developing interventions effective in reducing instability and falls (Bieryla, Madigan & Nussbaum, 2007; Mansfield, Peters, Liu & Maki, 2007; Perry et al., 2008; Mansfield et al., 2010). While generalized exercise programs have potential to attain such an end (Sherrington et al., 2008; Sherrington, Tiedemann, Fairhall, Close & Lord, 2011), interventions that affect the specific components underlying balance dyscontrol may have greater effectiveness. Recently, perturbation-based training programs (Mansfield et al., 2007; Mansfield et al., 2010) have demonstrated some success in reducing the frequency of specific features of stepping reactions believed to be associated with fall risk, such as multiple step reactions (McIlroy & Maki, 1996, Schulz et al., 2005), laterally-directed steps in response to anterior perturbation (Rogers et al., 2001; Schulz et al., 2005), and foot collisions during lateral perturbations (Maki, Edmondstone & McIlroy, 2000).

What remains somewhat unclear, however, are the specific biological components of the stepping reactions that are affected by training, although enhanced feedback from the plantar surface of the foot or sensory re-weighting have been suggested (Mansfield et al., 2010). If the timing of the active peak reorientation of the net GRF with respect to the COM is indeed a principal factor governing effective mediolateral stabilisation, it is possible that afferent somatosensory information, specifically from the plantar foot surface, may be augmented or better integrated with practice. Indeed, Perry et al. (2008) have found that older adults using a balance-enhancing insole, designed to facilitate foot-sole sensation, exhibited increased lateral stability during gait (quantified as a increase in the distance between the COM and lateral BOS at the time of peak lateral COM displacement) and reduced incidence of falls. It is also a possibility that the timing of the net GRF reorientation is a function of age-related differences in muscle contractile properties, resulting in a reduced rate of force development. From this, interventions that seek to increase the ability to generate lower limb power may be effective in reducing age-related mediolateral instability (Skelton, Kennedy & Rutherford, 2002; Rogers and Mille, 2003). Further, while there are differences between volitional and reactive stepping until the time of foot-contact (McIlroy & Maki, 1993a; McIlroy & Maki, 1993b; McIlroy & Maki, 1995; Maki & McIlroy, 1997; Maki & McIlroy, 2005), there appears to be similarity in the challenges associated with mediolateral stability control during the restabilisation phase of both forms of stepping. As such, it is possible that rapid step training could provide a similar benefit to the aforementioned perturbation-based training programs, provided stepping was executed as rapidly as possible. A distinct advantage of rapid step training would be the ability to perform such a program in a typical clinical environment without complex or expensive equipment to deliver balance perturbations.

## 6.4 LIMITATIONS

It is important to note the limitations that should be considered when interpreting the preceding outcomes. Firstly, a number of sources of error exist, which are inherent to current three-dimensional motion analysis techniques. Photogrammetric error (i.e. random error affecting accuracy of individual marker reconstruction) (Thornton, Morrissey & Couts, 1998), errors in model calibration and anatomical landmark determination (Della Croce, Cappozzo, Kerrigan & Luchetti, 1997), each affect the accuracy of the segment endpoints and anatomically-based segmental coordinate systems (Ramakrishnan & Kadaba, 1991) in addition to the location of segment centres of mass. As the total body centre of mass is the weighted average of individual segment centres of mass, it is sensitive to positional errors related to any of the individual segments. Although tracking markers were placed on rigid clusters to eliminate relative marker movement, photogrammetric error may itself induce relative error, which may affect the validity of the rigid body assumption. Absolute marker movement, related to the degree to which the rigid clusters represent the motion of the underlying skeletal structures, may have an influence on the calculation of the position of individual segment centres of mass and, hence, on the total body centre of mass. Correct determination of the total body centre of mass is also dependent on the anthropometric data used to characterize the individual segment mass proportions, geometry and relative centre of mass position as a proportion of segment length. Further, error in defining the position and orientation of each force plate within the kinematic reference frame may affect the calculation of compound variables derived from kinematic and kinetic data.

With the abovementioned sources of error in mind, extreme care was taken to minimize their influence before embarking upon data collection. To minimize the influence of photogrammetric error, Vicon cameras were positioned such that their optical axes intersected at an angle greater than 30 degrees and were placed as close to the participant as possible while retaining the ability to view all retroreflective markers. Random error, due to variation in marker centroid estimation, was reduced by filtering individual marker trajectories. Marker placement was performed by the same investigator (JS) for each participant, which would eliminate concerns regarding inter-examiner precision in landmark definition (Della Croce, Cappozzo & Kerrigan, 1999). Absolute marker movement has been suggested to induce the largest errors for rotations about the longitudinal axis of a segment. Such error was further minimized by adhering to the recommendations of Manal, McClay, Stanhope, Richards and Galinat (2000), who have suggested the use of large, rigid clusters of 4 markers, located distally on the segment of interest and held in place with elastic wrapping. To reduce error induced by the choice of anatomical data set, we used anthropometric data specific to each age-group and sex, when possible (Dempster, 1955; Hanavan, 1964, as cited in Robertson et al., 2004; de

Leva, 1996). Lastly, the spatial synchronization of kinematic and kinetic reference frames was quantified during pilot testing by examining the position of the total body COM and the COP during 60 seconds of quiet standing, under the assumption that over the course of the trial the average positions of the COM and COP would be equal.

From a methodological perspective, we chose to standardise the cable force for the tether-release protocol to 10% of the participant's body weight, as this magnitude had been found to evoke stepping in 100% of trials in young adults (Singer, Prentice & McIlroy, 2010), while also failing to evoke multiple forward steps. Such a method serves to standardize the magnitude of the anterior perturbation, at the instant of cable release, due to the net external moment about the ankle joint (i.e. the difference in moments caused by the gravitational force acting at the COM and the net GRF acting at the net COP), thereby accounting for between-trial variations in the position of the COP. Between-subject differences in body mass alone (i.e. assuming equal net COP position) have no influence on the anteriorly-directed angular acceleration upon cable release, as the moment caused by the gravitational force acting at the COM scales in proportion to the mass moment of inertia about a mediolateral axis passing through the ankle joint. Between-subject variations in body height, however, will have an effect on the angular acceleration, as changes in body height will alter the mass moment of inertia to a greater extent than the moment caused by the gravitational force acting at the COM (due to body height induced changes in the length of the moment arm of the gravitational force). Interestingly, there were no between-group differences in body height (Table 5-1). As a result, it is unlikely that the observed between-group differences were a result of differences in perturbation magnitude. Despite the potential such confounds of between-group differences in body height on perturbation magnitude, the present method of standardisation was chosen over one in which we would titrate the perturbation magnitude for each participant to a level that would be scarcely large enough to evoke a forward step (i.e. standardizing relative to the individual), as doing so could expose participants to a differing number of trials before testing began. Additionally, on account of time efficiency while participants were in the lab, the chosen method was also favoured over one in which we would estimate, or precisely calculate, the vertical position of the COM.

There are some limitations inherent to cable-release perturbations over other forms of perturbation. While the timing of perturbation onset can be altered and catch trials included in attempt to reduce predictability, both the direction and the relative magnitude can be deduced by the participant before perturbation onset. As such, it is possible that the conclusions reached during our examination of perturbation-evoked stepping, specifically those regarding the role of limb stiffness at foot-contact in reorienting the net GRF (P1) and the subsequent effect on the P2 peak may be related, in part, to age-related differences in arousal state, central set or fear.

We chose to standardize initial foot placement, as per McIlroy and Maki (1997). While this may have forced some participants to adopt an unnatural stance width, we felt it was a necessary compromise to ensure that the potential differences in the choice of stance width did not bias the effect of the ML APA in displacing the COM toward the stance limb prior to step-onset.

Participants were asked to “try to use only a single step, if necessary” to regain their balance. We acknowledge that without constraints on the number of steps, it is possible that some individuals, especially some older adults, would prefer the use of multiple steps to restore stability (McIlroy & Maki, 1996; Hsiao & Robinovitch, 1998: Hsiao & Robinovitch, 2001). While there is some debate as to whether multiple step responses are a planned strategy for restabilisation following an external perturbation (Luchies et al., 1994; McIlroy & Maki, 1996; Rogers et al., 2001; Maki & McIlroy, 2006), we believe it is the characteristics of initial step that have the largest role. The function of the initial step for arresting the COM may be particularly important if each successive step is considered to be an opportunity for the development of further instability, if not scaled correctly. In addition, there are instances where environmental constraints would restrict the response to a single step. Ultimately, the approach utilised within this series of studies allowed us to quantify the biomechanical variables that may be associated with age-related instability during the initial step, without the confounding influence of individual differences in the number of steps taken to restore stability.

Arm movement was not restricted during any of the tasks. Although not quantified, the older adults were observed to exhibit larger arm abduction after step initiation than did the young. It is possible that such arm movement during the stepping phase, and the corresponding increase in the moment of inertia about the anteroposterior axis, could moderate frontal plane angular acceleration caused by the gravitational force, which may simplify stability control during the restabilisation phase. Had arm use been restricted, it is possible that older adults would have exhibited either a larger magnitude of divergence of the line of action of the net GRF with respect to the COM, or an increased magnitude of overshoot. Such an outcome, however, would likely only further amplify the observed age-related differences.

We quantified both kinematic and kinetic variables in a lab-fixed coordinate system, rather than a body-fixed coordinate system (e.g. a local coordinate system aligned with the mean foot progression angle during the restabilisation phase). While such lab-centric calculations for the total body centre of mass are common practice, large axial rotations of the body could invalidate the anatomical meaning of ‘lateral’ when calculated with reference to the frontal plane of a lab-fixed coordinate system. Nevertheless, in the context of balance control, lateral instability has referred to that which occurs in direction orthogonal to the initial perturbation (McIlroy & Maki, 1996; Huang & Ahmed, 2011). As such, we believe the conclusions drawn from the preceding studies continue to have value, as anterior perturbations were aligned with the AP axis of the lab.

Although we quantified the divergence of the frontal plane inclination angles of the net GRF and COP-COM, we did not quantify the net external moment about the COM. We initially chose to quantify the divergence between the two inclination angles because we believed ML instability to be the result of a control challenge rather than limitations in the ability to generate muscle force. As such, we were specifically interested in how the vertical and ML components of the net GRF were modulated to direct the line of action through the total body COM during the restabilisation phase. As we found that the line of action of the net GRF was not directed through the COM, it was possible that our interpretations concerning the effect of the P1 and P2 peaks on the total body angular acceleration could be incorrect if the magnitude of the net frontal plane GRF at these instances was small. We quantified the frontal plane external moment about the COM in a subset of participants. The waveform representing the net external moment has the same shape as the divergence between the inclination angles of the net GRF and COP-COM, but its amplitude is modulated by the magnitudes of the net GRF and the COP-COM position vector. As a result, we believe our conclusions would be consistent with those arising from an analysis of the net external moment.

## 6.5 FUTURE DIRECTIONS

The results of this thesis form the foundation for several lines of inquiry which could further progress our current understanding of age-related differences in mediolateral dynamic stability control. In the near future, it will be necessary to perform additional work utilizing electromyography and inverse dynamics, coupled with the quantification of the COM kinematics, to more deeply probe the purpose and underlying mechanisms governing the deviation of the net GRF line of action with respect to the COM. Specifically, such work would elucidate the role of anticipatory control in regulating the timing of the P1 peak; the role of muscle activation, subsequent to foot contact, in altering the timing of the P2 peak; whether such timing is affected by the magnitude of preparatory muscle activity during the swing phase; the role of instability subsequent to foot-contact for modulating the amplitude of the P2 peak. Following this, work should be directed at determining the influence of variables at the level of individual joints, which are associated with ML instability.

In the longer term, we are also interested in exploring a number of avenues that diverge from the present work. Firstly, the studies within this thesis examined the orientation of the net GRF line of action, given the selected foot placement. An alternative approach would be to use techniques similar to Wight et al. (2008) and Millard et al. (2009) to determine if differences in actual, as opposed to optimal, foot placement can account for age-related differences in stability control. In addition, despite the suggestion that ML stability control has important links to the restabilisation phase, the influence of the swing phase cannot be overlooked. Future work employing segmental mechanical energy analysis will help to elucidate age-related differences in energy transfer to the trunk at the end of the swing-phase, which may influence mediolateral stability. Finally, we hope to apply the techniques utilized within this thesis to the analysis of steady-state gait. Through prospective studies, we hope to explore the relationship to fall risk.

## 6.6 CONCLUSIONS

This thesis sought to develop further understanding regarding age-related differences in mediolateral stability control during the restabilisation phase of volitional and reactive stepping. Greater ML COM incongruity was observed among older adults, independent of the means by which stepping was initiated. We believe such increased COM incongruity is likely indicative of greater instability, which may be associated with the timing of reorientation of the net GRF to reverse the polarity of angular acceleration about the COM. This work highlights the need to further explore the control of lateral stability and develop therapeutic interventions to reduce the incidence of lateral instability among older adults.

## REFERENCES

Barry, B. K., Riek, S., & Carson, R. G. (2005). Muscle coordination during rapid force production by young and older adults. J Gerontol A Biol Sci Med Sci, 60(2), 232-240.

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: benefits, demands, and adverse consequences. Arch Phys Med Rehabil, 86(1), 134-145.

Bauby, C. E., & Kuo, A. D. (2000). Active control of lateral balance in human walking. J Biomech, 33(11), 1433-1440.

Benjuya, N., Melzer, I., & Kaplanski, J. (2004). Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing. J Gerontol A Biol Sci Med Sci, 59(2), 166-171.

Bieryla, K. A., Madigan, M. L., & Nussbaum, M. A. (2007). Practicing recovery from a simulated trip improves recovery kinematics after an actual trip. Gait Posture, 26(2), 208-213.

Binda, S. M., Culham, E. G., & Brouwer, B. (2003). Balance, muscle strength, and fear of falling in older adults. Exp Aging Res, 29(2), 205-219.

Bishop, M., Brunt, D., Pathare, N., & Patel, B. (2002). The interaction between leading and trailing limbs during stopping in humans. Neurosci Lett, 323(1), 1-4.

Bishop, M., Brunt, D., Pathare, N., & Patel, B. (2004). The effect of velocity on the strategies used during gait termination. Gait Posture, 20(2), 134-139.

Bloem, B. R., Steijns, J. A., & Smits-Engelsman, B. C. (2003). An update on falls. Curr Opin Neurol, 16(1), 15-26.

Bogle Thorbahn, L. D., & Newton, R. A. (1996). Use of the Berg Balance Test to predict falls in elderly persons. Phys Ther, 76(6), 576-583; discussion 584-575.

Brach, J. S., Berlin, J., VanSwearingen, J., Newman, A., & Studenski, S. (2005). Too much or too little step width variability is associated with a fall history only in older persons who walk at or near normal gait speed. J Am Geriatr Soc, 53(4), S133-S134.

Brauer, S. G., Burns, Y. R., & Galley, P. (2000). A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers. J Gerontol A Biol Sci Med Sci, 55(8), M469-476.

Breniere, Y., & Do, M. C. (1991). Control of gait initiation. J Mot Behav, 23(4), 235-240.

Brunt, D., Lafferty, M. J., McKeon, A., Goode, B., Mulhausen, C., & Polk, P. (1991). Invariant characteristics of gait initiation. Am J Phys Med Rehabil, 70(4), 206-212.

Campbell, A. J., Borrie, M. J., & Spears, G. F. (1989). Risk factors for falls in a community-based prospective study of people 70 years and older. J Gerontol, 44(4), M112-117.

Canadian Institute of Health Information (2010). National Trauma Registry 2009 Report: Major Injury in Canada (Includes 2007–2008 Data). Ottawa: CIHI.

Cao, C., Ashton-Miller, J.A., Schultz, A.B., Alexander, N.B. (1997). Abilities to turn suddenly while walking: effects of age, gender, and available response time. J Gerontol A Biol Sci Med Sci, 52A(2), M88-M93.

Cao, C., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (1998). Sudden turns and stops while walking: kinematic sources of age and gender differences. Gait Posture, 7(1), 45-52.

Carpenter, M. G., Adkin, A. L., Brawley, L. R., & Frank, J. S. (2006). Postural, physiological and psychological reactions to challenging balance: does age make a difference? Age Ageing, 35(3), 298-303.

Carty, C. P., Mills, P., & Barrett, R. (2011). Recovery from forward loss of balance in young and older adults using the stepping strategy. Gait Posture, 33(2), 261-267.

Cenciarini, M., Loughlin, P. J., Sparto, P. J., & Redfern, M. S. (2009). Medial-Lateral Postural Control in Older Adults Exhibits Increased Stiffness and Damping. Embc: 2009 Annual International Conference of the Ieee Engineering in Medicine and Biology Society, Vols 1-20, 7006-7009.

Cenciarini, M., Loughlin, P. J., Sparto, P. J., & Redfern, M. S. (2010). Stiffness and damping in postural control increase with age. IEEE Trans Biomed Eng, 57(2), 267-275.

Chang, H., & Krebs, D. E. (1999). Dynamic balance control in elders: gait initiation assessment as a screening tool. Arch Phys Med Rehabil, 80(5), 490-494.

Chang, S. H. J., Mercer, V. S., Giuliani, C. A., & Sloane, P. D. (2005). Relationship between hip abductor rate of force development and mediolateral stability in older adults. Arch Phys Med Rehabil, 86(9), 1843-1850.

Chu, Y. H., Tang, P. F., Chen, H. Y., & Cheng, C. H. (2009). Altered muscle activation characteristics associated with single volitional forward stepping in middle-aged adults. Clin Biomech, 24(9), 735-743.

Conforto, S., Schmid, M., Camomilla, V., D'Alessio, T., & Cappozzo, A. (2001). Hemodynamics as a possible internal mechanical disturbance to balance. Gait Posture, 14(1), 28-35.

Cordo, P. J., & Nashner, L. M. (1982). Properties of postural adjustments associated with rapid arm movements. J Neurophysiol, 47(2), 287-302.

Crenna, P., Cuong, D. M., & Breniere, Y. (2001). Motor programmes for the termination of gait in humans: organisation and velocity-dependent adaptation. J Physiol, 537(Pt 3), 1059-1072.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. J Biomech, 29(9), 1223-1230.

della Croce, U., Cappozzo, A., Kerrigan, D. C., & Luchetti, L. (1997). Bone Position and Orientation Errors: Pelvis and Lower Limb Anatomical Landmark Identification Reliability Gait Posture, 5(2), 156-157.

della Croce, U., Cappozzo, A., & Kerrigan, D. C. (1999). Pelvis and lower limb anatomical landmark calibration precision and its propagation to bone geometry and joint angles. Med Biol Eng Comput, 37(2), 155-161.

Dingwell, J. B., & Marin, L. C. (2006). Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. J Biomech, 39(3), 444-452.

Do, M. C., Breniere, Y., & Brenguier, P. (1982). A biomechanical study of balance recovery during the fall forward. J Biomech, 15(12), 933-939.

Do, M. C., Schneider, C., & Chong, R. K. (1999). Factors influencing the quick onset of stepping following postural perturbation. J Biomech, 32(8), 795-802.

Donelan, J. M., Kram, R., & Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. Proc R Soc B, 268(1480), 1985-1992.

Donelan, J. M., Kram, R., & Kuo, A. D. (2002). Mechanical work for step-to-step transitions is a major determinant of the metabolic cost of human walking. J Exp Biol, 205(Pt 23), 3717-3727.

Donelan, J. M., Shipman, D. W., Kram, R., & Kuo, A. D. (2004). Mechanical and metabolic requirements for active lateral stabilization in human walking. J Biomech, 37(6), 827-835.

Duncan, P. W., Studenski, S., Chandler, J., & Prescott, B. (1992). Functional reach: predictive validity in a sample of elderly male veterans. J Gerontol, 47(3), M93-98.

Duncan, P. W., Weiner, D. K., Chandler, J., & Studenski, S. (1990). Functional reach: a new clinical measure of balance. J Gerontol, 45(6), M192-197.

Elble, R. J., Moody, C., Leffler, K., & Sinha, R. (1994). The initiation of normal walking. Mov Disord, 9(2), 139-146.

Era, P., & Heikkinen, E. (1985). Postural sway during standing and unexpected disturbance of balance in random samples of men of different ages. J Gerontol, 40(3), 287-295.

Fernie, G. R., Gryfe, C. I., Holliday, P. J., & Llewellyn, A. (1982). The relationship of postural sway in standing to the incidence of falls in geriatric subjects. Age Ageing, 11(1), 11-16.

Graafmans, W. C., Ooms, M. E., Hofstee, H. M., Bezemer, P. D., Bouter, L. M., & Lips, P. (1996). Falls in the elderly: a prospective study of risk factors and risk profiles. Am J Epidemiol, 143(11), 1129-1136.

Halliday, S. E., Winter, D. A., Frank, J. S., Patla, A. E., & Prince, F. (1998). The initiation of gait in young, elderly, and Parkinson's disease subjects. Gait Posture, 8(1), 8-14.

Hamill, J., & Selbie, W. S. (2004). Three Dimensional Kinetics. In D. G. E. Robertson, G. E. Caldwell, J. Hamill, G. Kamen & S. N. Whittlesey (Eds.), Research methods in biomechanics (pp. 145-160). Champaign, Ill. ; Leeds, U.K.: Human Kinetics.

Hase, K., & Stein, R. B. (1998). Analysis of rapid stopping during human walking. J Neurophysiol, 80(1), 255-261.

Hausdorff, J. M. (2005). Gait variability: methods, modeling and meaning. J Neuroeng Rehabil, 2, 19.

Hausdorff, J. M. (2007). Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking. Hum Mov Sci, 26(4), 555-589.

Hayes, W. C., Myers, E. R., Robinovitch, S. N., Van Den Kroonenberg, A., Courtney, A. C., & McMahon, T. A. (1996). Etiology and prevention of age-related hip fractures. Bone, 18(1 Suppl), 77S-86S.

Herr, H., & Popovic, M. (2008). Angular momentum in human walking. J Exp Biol, 211(4), 467-481.

Hilliard, M. J., Martinez, K. M., Janssen, I., Edwards, B., Mille, M.-L., Zhang, Y., et al. (2008). Lateral balance factors predict future falls in community-living older adults. Arch Phys Med Rehabil, 89(9), 1708-1713.

Hof, A. L., Gazendam, M. G., & Sinke, W. E. (2005). The condition for dynamic stability. J Biomech, 38(1), 1-8.

Holbein-Jenny, M. A., McDermott, K., Shaw, C., & Demchak, J. (2007). Validity of functional stability limits as a measure of balance in adults aged 23–73 years. Ergonomics. 50(5), 631-646.

Horak, F. B., Henry, S. M., & ShumwayCook, A. (1997). Postural perturbations: New insights for treatment of balance disorders. Phys Ther, 77(5), 517-533.

Horak, F. B., & Nashner, L. M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. J Neurophysiol, 55(6), 1369-1381.

Horak, F. B., Nutt, J. G., & Nashner, L. M. (1992). Postural inflexibility in parkinsonian subjects. J Neurol Sci, 111(1), 46-58.

Horak, F. B., Shupert, C. L., & Mirka, A. (1989). Components of Postural Dyscontrol in the Elderly - a Review. Neurobiol Aging, 10(6), 727-738.

Hortobagyi, T., & DeVita, P. (2000). Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging. J Electromyogr Kinesiol, 10(2), 117-126.

Hsiao, E. T., & Robinovitch, S. N. (1998). Common protective movements govern unexpected falls from standing height. J Biomech, 31(1), 1-9.

Hsiao, E. T., & Robinovitch, S. N. (1999). Biomechanical influences on balance recovery by stepping. J Biomech, 32(10), 1099-1106.

Hsiao, E. T., & Robinovitch, S. N. (2001). Elderly subjects' ability to recover balance with a single backward step associates with body configuration at step contact. J Gerontol A Biol Sci Med Sci, 56(1), M42-M47.

Hsiao-Wecksler, E. T. (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology, 18(2), 179-187.

Hsiao-Wecksler, E. T., & Robinovitch, S. N. (2007). The effect of step length on young and elderly women's ability to recover balance. Clin Biomech (Bristol, Avon), 22(5), 574-580.

Huang, H. J., & Ahmed, A. A. (2011). Tradeoff between stability and maneuverability during whole-body movements. PLoS One, 6(7), e21815.

Inglis, J. T., Horak, F. B., Shupert, C. L., & Jones-Rycewicz, C. (1994). The Importance of Somatosensory Information in Triggering and Scaling Automatic Postural Responses in Humans. Exp Brain Res, 101(1), 159-164.

Jaeger, R. J., & Vanitchatchavan, P. (1992). Ground reaction forces during termination of human gait. J Biomech, 25(10), 1233-1236.

Jensen, J. L., Brown, L. A., & Woollacott, M. H. (2001). Compensatory stepping: the biomechanics of a preferred response among older adults. Exp Aging Res, 27(4), 361-376.

Jian, Y., Winter, D. A., Ishac, M. G., & Gilchrist, L. (1993). Trajectory of the body COG and COP during initiation and termination of gait. Gait Post, 1(1), 9-22.

Jonsson, E., Henriksson, M., & Hirschfeld, H. (2002). Does the functional reach test reflect stability limits in elderly people. J Rehabil Med, 35(1), 26-30.

Karamanidis, K., Arampatzis, A., & Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. J Electromyogr Kinesiol, 18(6), 980-989.

Kaya, B. K., Krebs, D. E., & Riley, P. O. (1998). Dynamic stability in elders: momentum control in locomotor ADL. J Gerontol A Biol Sci Med Sci, 53(2), M126-134.

King, G. W., Luchies, C. W., Stylianou, A. P., Schiffman, J. M., & Thelen, D. G. (2005). Effects of step length on stepping responses used to arrest a forward fall. Gait Posture, 22(3), 219-224.

King, M. B., Judge, J. O., & Wolfson, L. (1994). Functional base of support decreases with age. J Gerontol, 49(6), M258-M263.

Laessoe, U., & Voigt, M. (2008). Anticipatory postural control strategies related to predictive perturbations. Gait Posture, 28(1), 62-68.

Lajoie, Y., Teasdale, N., Bard, C., & Fleury, M. (1993). Attentional demands for static and dynamic equilibrium. Exp Brain Res, 97(1), 139-144.

Laufer, Y., Barak, Y., & Chemel, I. (2006). Age-related differences in the effect of a perceived threat to stability on postural control. J Gerontol A Biol Sci Med Sci, 61(5), 500-504.

Lin, S. I., & Woollacott, M. H. (2002). Postural muscle responses following changing balance threats in young, stable older, and unstable older adults. J Mot Behav, 34(1), 37-44.

Lin, S. I., Woollacott, M. H., & Jensen, J. L. (2004). Postural response in older adults with different levels of functional balance capacity. Aging Clin Exp Res, 16(5), 369-374.

Lord, S. R., & Dayhew, J. (2001). Visual risk factors for falls in older people. J Am Geriatr Soc, 49(5), 508-515.

Lord, S. R., Rogers, M. W., Howland, A., & Fitzpatrick, R. (1999). Lateral stability, sensorimotor function and falls in older people. J Am Geriatr Soc, 47(9), 1077-1081.

Luchies, C. W., Alexander, N. B., Schultz, A. B., & Ashton-Miller, J. (1994). Stepping responses of young and old adults to postural disturbances: kinematics. J Am Geriatr Soc, 42(5), 506-512.

Lyon, I. N., & Day, B. L. (1997). Control of frontal plane body motion in human stepping. Experimental brain research Experimentelle Hirnforschung Experimentation cerebrale, 115(2), 345-356.

Maki, B. E., Edmondstone, M. A., & McIlroy, W. E. (2000). Age-related differences in laterally directed compensatory stepping behavior. J Gerontol A Biol Sci Med Sci, 55(5), M270-M277.

Maki, B. E., Edmondstone, M. A., Perry, S. D., Heung, E., Quant, S., & McIlroy, W. E. (2001). Control of rapid limb movements for balance recovery: do age-related changes predict falling risk? Paper presented at the International Society for Posture and Gait Research.

Maki, B. E., Holliday, P. J., & Topper, A. K. (1994). A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. J Gerontol, 49(2), M72-84.

Maki, B. E., & McIlroy, W. E. (1996). Postural control in the older adult. Clin Geriatr Med, 12(4), 635-658.

Maki, B. E., & McIlroy, W. E. (1997). The role of limb movements in maintaining upright stance: The "change-in-support" strategy. Phys Ther, 77(5), 488-507.

Maki, B. E., & McIlroy, W. E. (1999). The control of foot placement during compensatory stepping reactions: does speed of response take precedence over stability? IEEE Trans Rehabil Eng, 7(1), 80-90.

Maki, B. E., & McIlroy, W. E. (2005). Change-in-support balance reactions in older persons: An emerging research area of clinical importance. Neurologic Clin, 23(3), 751.

Maki, B. E., & McIlroy, W. E. (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. Age Ageing, 35 Suppl 2, ii12-ii18.

Maki, B. E., Perry, S. D., Norrie, R. G., & McIlroy, W. E. (1999). Effect of facilitation of sensation from plantar foot-surface boundaries on postural stabilization in young and older adults. J Gerontol A Biol Sci Med Sci, 54(6), M281-287.

Manal, K., McClay, I., Stanhope, S., Richards, J., & Galinat, B. (2000). Comparison of surface mounted markers and attachment methods in estimating tibial rotations during walking: an in vivo study. Gait Posture, 11(1), 38-45.

Mansfield, A., Peters, A. L., Liu, B. A., & Maki, B. E. (2007). A perturbation-based balance training program for older adults: study protocol for a randomised controlled trial. BMC geriatr, 7, 12.

Mansfield, A., Peters, A. L., Liu, B. A., & Maki, B. E. (2010). Effect of a perturbation-based balance training program on compensatory stepping and grasping reactions in older adults: a randomized controlled trial. Phys Ther, 90(4), 476-491.

Mathiyakom, W., & McNitt-Gray, J. L. (2008). Regulation of angular impulse during fall recovery. J Rehabil Res Dev, 45(8), 1237-1247.

McGibbon, C. A., Krebs, D. E., & Wagenaar, R. (2005). Stepping stability: effects of sensory perturbation. J Neuroeng Rehabil, 2, 9.

McIlroy, W. E., & Maki, B. E. (1993a). Changes in early 'automatic' postural responses associated with the prior-planning and execution of a compensatory step. Brain Res, 631(2), 203-211.

McIlroy, W. E., & Maki, B. E. (1993b). Do anticipatory postural adjustments precede compensatory stepping reactions evoked by perturbation? Neurosci Lett, 164(1-2), 199-202.

McIlroy, W. E., & Maki, B. E. (1993c). Task constraints on foot movement and the incidence of compensatory stepping following perturbation of upright stance. Brain Res, 616(1-2), 30-38.

McIlroy, W. E., & Maki, B. E. (1995). Adaptive changes to compensatory stepping responses. Gait Post, 3(1), 43-50.

McIlroy, W. E., & Maki, B. E. (1996). Age-related changes in compensatory stepping in response to unpredictable perturbations. J Gerontol A Biol Sci Med Sci, 51(6), M289-296.

McIlroy, W. E., & Maki, B. E. (1997). Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. Clin Biomech (Bristol, Avon), 12(1), 66-70.

McIlroy, W. E., & Maki, B. E. (1999). The control of lateral stability during rapid stepping reactions evoked by antero-posterior perturbation: does anticipatory control play a role? Gait Posture, 9(3), 190-198.

Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2009). Rapid gait termination: Effects of age, walking surfaces and footwear characteristics. Gait Posture, 30(1), 65-70.

Millard, M., Wight, D., McPhee, J., Kubica, E., & Wang, D. (2009). Human foot placement and balance in the sagittal plane. J Biomech Eng, 131(12), 121001.

Mille, M. L., Rogers, M. W., Martinez, K., Hedman, L. D., Johnson, M. E., Lord, S. R., et al. (2003). Thresholds for inducing protective stepping responses to external perturbations of human standing. J Neurophysiol, 90(2), 666-674.

Muir, S. W., Berg, K., Chesworth, B., & Speechley, M. (2008). Use of the Berg Balance Scale for predicting multiple falls in community-dwelling elderly people: a prospective study. Phys Ther, 88(4), 449-459.

Murray, M. P., Seireg, A., & Scholz, R. C. (1967). Center of gravity, center of pressure, and supportive forces during human activities. J Appl Physiol, 23(6), 831-838.

Nashner, L. M. (1977). Fixed patterns of rapid postural responses among leg muscles during stance. Exp Brain Res, 30(1), 13-24.

Neptune, R. R., & McGowan, C. P. (2011). Muscle contributions to whole-body sagittal plane angular momentum during walking. J Biomech, 44(1), 6-12.

Nevitt, M. C., & Cummings, S. R. (1993). Type of fall and risk of hip and wrist fractures: the study of osteoporotic fractures. The Study of Osteoporotic Fractures Research Group. J Am Geriatr Soc, 41(11), 1226-1234.

Nevitt, M. C., Cummings, S. R., Kidd, S., & Black, D. (1989). Risk factors for recurrent nonsyncopal falls. A prospective study. JAMA, 261(18), 2663-2668.

Oates, A. R., Patla, A. E., Frank, J. S., & Greig, M. A. (2005). Control of dynamic stability during gait termination on a slippery surface. J Neurophysiol, 93(1), 64-70.

O'Kane, F. W., McGibbon, C. A., & Krebs, D. E. (2003). Kinetic analysis of planned gait termination in healthy subjects and patients with balance disorders. Gait Posture, 17(2), 170-179.

O'Loughlin, J. L., Robitaille, Y., Boivin, J. F., & Suissa, S. (1993). Incidence of and risk factors for falls and injurious falls among the community-dwelling elderly. Am J Epidemiol, 137(3), 342-354.

Pai, Y. C., Maki, B. E., Iqbal, K., McIlroy, W. E., & Perry, S. D. (2000). Thresholds for step initiation induced by support-surface translation: a dynamic center-of-mass model provides much better prediction than a static model. J Biomech, 33(3), 387-392.

Pai, Y. C., & Patton, J. (1997). Center of mass velocity-position predictions for balance control. J Biomech, 30(4), 347-354.

Pai, Y. C., Rogers, M. W., Patton, J., Cain, T. D., & Hanke, T. A. (1998). Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults. J Biomech, 31(12), 1111-1118.

Pai, Y. C., Wening, J. D., Runtz, E. F., Iqbal, K., & Pavol, M. J. (2003). Role of feedforward control of movement stability in reducing slip-related balance loss and falls among older adults. J Neurophysiol, 90(2), 755-762.

Patla, A., Frank, J., & Winter, D. (1990). Assessment of balance control in the elderly: major issues. Physiotherapy Canada, 42(2), 89-97.

Pavol, M. J., & Pai, Y. C. (2002). Feedforward adaptations are used to compensate for a potential loss of balance. Exp Brain Res, 145(4), 528-538.

Pavol, M. J., Runtz, E. F., Edwards, B. J., & Pai, Y. C. (2002). Age influences the outcome of a slipping perturbation during initial but not repeated exposures. J Gerontol A Biol Sci Med Sci, 57(8), M496-M503.

Perry, S. D., McIlroy, W. E., & Maki, B. E. (2000). The role of plantar cutaneous mechanoreceptors in the control of compensatory stepping reactions evoked by unpredictable, multi-directional perturbation. Brain Res, 877(2), 401-406.

Perry, S. D., Radtke, A., McIlroy, W. E., Fernie, G. R., & Maki, B. E. (2008). Efficacy and effectiveness of a balance-enhancing insole. J Gerontol A Biol Sci Med Sci, 63(6), 595-602.

Pijnappels, M., Bobbert, M. F., & van Dieen, J. H. (2004). Contribution of the support limb in control of angular momentum after tripping. J Biomech, 37(12), 1811-1818.

Pijnappels, M., Kingma, I., Wezenberg, D., Reurink, G., & van Dieen, J. H. (2010). Armed against falls: the contribution of arm movements to balance recovery after tripping. Exp Brain Res, 201(4), 689-699.

Popovic, M. B., Goswami, A., & Herr, H. (2005). Ground reference points in legged locomotion: Definitions, biological trajectories and control implications. Int J Robot Res, 24(12), 1013-1032.

Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., & Myklebust, B. M. (1996). Measures of postural steadiness: differences between healthy young and elderly adults. IEEE Trans Biomed Eng, 43(9), 956-966.

Ramakrishnan, H. K., & Kadaba, M. P. (1991). On the Estimation of Joint Kinematics during Gait. J Biomech, 24(10), 969-977.

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. N. (2004). Research methods in biomechanics. Champaign, Ill. ; Leeds, U.K.: Human Kinetics.

Robertson, D. G., & Dowling, J. J. (2003). Design and responses of Butterworth and critically damped digital filters. J Electromyogr Kinesiol, 13(6), 569-573.

Robinovitch, S. N., Inkster, L., Maurer, J., & Warnick, B. (2003). Strategies for avoiding hip impact during sideways falls. J Bone Miner Res, 18(7), 1267-1273.

Rogers, M. W., Hain, T. C., Hanke, T. A., & Janssen, I. (1996). Stimulus parameters and inertial load: effects on the incidence of protective stepping responses in healthy human subjects. Arch Phys Med Rehabil, 77(4), 363-368.

Rogers, M. W., Hedman, L. D., Johnson, M. E., Cain, T. D., & Hanke, T. A. (2001). Lateral stability during forward-induced stepping for dynamic balance recovery in young and older adults. J Gerontol A Biol Sci Med Sci, 56(9), M589-594.

Rogers, M. W., Kukulka, C. G., & Soderberg, G. L. (1992). Age-related changes in postural responses preceding rapid self-paced and reaction time arm movements. J Gerontol, 47(5), M159-165.

Rogers, M. W., & Mille, M.-L. (2003). Lateral stability and falls in older people. Exerc Sport Sci Rev, 31(4), 182-187.

Schmid, M., Conforto, S., Bibbo, D., & D'Alessio, T. (2004). Respiration and postural sway: detection of phase synchronizations and interactions. Hum Mov Sci, 23(2), 105-119.

Schrager, M. A., Kelly, V. E., 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(3), 466-471.

Schulz, B. W., Ashton-Miller, J. A., & Alexander, N. B. (2005). Compensatory stepping in response to waist pulls in balance-impaired and unimpaired women. Gait Posture, 22(3), 198-209.

Sherrington, C., Tiedemann, A., Fairhall, N., Close, J. C. T., & Lord, S. R. (2011). Exercise to prevent falls in older adults: an updated meta-analysis and best practice recommendations. New South Wales public health bulletin, 22(3-4), 78-83.

Sherrington, C., Whitney, J. C., Lord, S. R., Herbert, R. D., Cumming, R. G., & Close, J. C. T. (2008). Effective Exercise for the Prevention of Falls: A Systematic Review and Meta-Analysis. J Am Geriatr Soc, 56(12), 2234-2243.

Shumway-Cook, A., Baldwin, M., Polissar, N. L., & Gruber, W. (1997). Predicting the probability for falls in community-dwelling older adults. Phys Ther, 77(8), 812-819.

Shumway-Cook, A., & Woollacott, M. (2000). Attentional demands and postural control: the effect of sensory context. J Gerontol A Biol Sci Med Sci, 55(1), M10-16.

Shumway-Cook, A., Woollacott, M., Kerns, K. A., & Baldwin, M. (1997). The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Gerontol A Biol Sci Med Sci, 52(4), M232-240.

Simoneau, G., & Krebs, D. (2000). Whole-body angular momentum during gait: a preliminary study of non-fallers and frequent fallers. J Appl Biomech, 16, 1-13.

Singer, J. C., Prentice, S. D., & McIlroy, W. E. (2010). Evaluation of the stepping limb centre of pressure on foot contact during volitional and perturbation-evoked stepping. Paper presented at the XVI Conference of the Canadian Society for Biomechanics.

Singer, J. C., Prentice, S. D., & McIlroy, W. E. (2012). Dynamic stability control during volitional stepping: A focus on the restabilisation phase at movement termination. Gait Posture, 25, 106-110.

Skelton, D. A., Kennedy, J., & Rutherford, O. M. (2002). Explosive power and asymmetry in leg muscle function in frequent fallers and non-fallers aged over 65. Age Ageing, 31(2), 119-125.

SMARTRISK (2009). The Economic Burden of Injury in Canada. (2009). Toronto: SMARTRISK.

Sockol, M. D., Raichlen, D. A., & Pontzer, H. (2007). Chimpanzee locomotor energetics and the origin of human bipedalism. Proc Natl Acad Sci U S A, 104(30), 12265-12269.

Statistics Canada (2007). Portrait of the Canadian Population in 2006, by age and sex. (Catalogue no. 97-551-XIE). Ottawa: Statistics Canada.

Stelmach, G. E., Populin, L., & Muller, F. (1990). Postural muscle onset and voluntary movement in the elderly. Neurosci Lett, 117(1-2), 188-193.

Thelen, D. G., Muriuki, M., James, J., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (2000). Muscle activities used by young and old adults when stepping to regain balance during a forward fall. J Electromyogr Kinesiol, 10(2), 93-101.

Thelen, D. G., Wojcik, L. A., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (1997). Age differences in using a rapid step to regain balance during a forward fall. J Gerontol A Biol Sci Med Sci, 52(1), M8-13.

Thornton, M. J., Morrissey, M. C., & Coutts, F. J. (1998). Some effects of camera placement on the accuracy of the Kinemetrix three-dimensional motion analysis system. Clin Biomech, 13(6), 452-454.

Tinetti, M. E., Speechley, M., & Ginter, S. F. (1988). Risk factors for falls among elderly persons living in the community. N Engl J Med, 319(26), 1701-1707.

Tirosh, O., & Sparrow, W. A. (2004). Gait termination in young and older adults: effects of stopping stimulus probability and stimulus delay. Gait Posture, 19(3), 243-251.

Tirosh, O., & Sparrow, W. A. (2005). Age and walking speed effects on muscle recruitment in gait termination. Gait Posture, 21(3), 279-288.

Vaughan, C. L., Langerak, N. G., & O'Malley, M. J. (2003). Neuromaturation of human locomotion revealed by non-dimensional scaling. Exp Brain Res, 153(1), 123-127.

Weerdesteyn, V., Laing, A. C., & Robinovitch, S. N. (2008). Automated postural responses are modified in a functional manner by instruction. Exp Brain Res, 186(4), 571-580.

Wight, D. L., Kubica, E. G., & Wang, D. W. L. (2008). Introduction of the Foot Placement Estimator: A Dynamic Measure of Balance for Bipedal Robotics. J Comput Nonlinear Dynam, 3(1).

Winter, D. A. (1995). Human balance and posture control during standing and walking. Gait Post, 3, 193-214.

Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. Phys Ther, 70(6), 340-347.

Wojcik, L. A., Thelen, D. G., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (1999). Age and gender differences in single-step recovery from a forward fall. J Gerontol A Biol Sci Med Sci, 54(1), M44-50.

Wojcik, L. A., Thelen, D. G., Schultz, A. B., Ashton-Miller, J. A., & Alexander, N. B. (2001). Age and gender differences in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall. J Biomech, 34(1), 67-73.

Woollacott, M. H., & Shumway-Cook, A. (1990). Changes in posture control across the life span--a systems approach. Phys Ther, 70(12), 799-807.

Woollacott, M., & Shumway-Cook, A. (2002). Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture, 16(1), 1-14.

Wu, M., Ji, L., Jin, D., & Pai, Y. C. (2007). Minimal step length necessary for recovery of forward balance loss with a single step. J Biomech, 40(7), 1559-1566.

Zajac, F. E. (1993). Muscle coordination of movement: a perspective. J Biomech, 26 Suppl 1, 109-124.

# APPENDIX A: ELSEVIER COPYRIGHT LICENSE

**ELSEVIER LICENSE**
**TERMS AND CONDITIONS**

Mar 20, 2012

---

This is a License Agreement between Jonathan C Singer ("You") and Elsevier ("Elsevier") provided by Copyright Clearance Center ("CCC"). The license consists of your order details, the terms and conditions provided by Elsevier, and the payment terms and conditions.

**All payments must be made in full to CCC. For payment instructions, please see information listed at the bottom of this form.**

| | |
|---|---|
| Supplier | Elsevier Limited<br>The Boulevard,Langford Lane<br>Kidlington,Oxford,OX5 1GB,UK |
| Registered Company Number | 1982084 |
| Customer name | Jonathan C Singer |
| Customer address | 200 University Avenue West |
| | Waterloo, ON N2L 3G1 |
| License number | 2873170573953 |
| License date | Mar 20, 2012 |
| Licensed content publisher | Elsevier |
| Licensed content publication | Gait & Posture |
| Licensed content title | Dynamic stability control during volitional stepping: A focus on the restabilisation phase at movement termination |
| Licensed content author | Jonathan C. Singer,Stephen D. Prentice,William E. McIlroy |
| Licensed content date | January 2012 |
| Licensed content volume number | 35 |
| Licensed content issue number | 1 |
| Number of pages | 5 |
| Start Page | 106 |
| End Page | 110 |
| Type of Use | reuse in a thesis/dissertation |
| Portion | full article |
| Format | both print and electronic |
| Are you the author of this Elsevier article? | Yes |
| Will you be translating? | No |
| Order reference number | |
| Title of your thesis/dissertation | Age-related changes in the control of mediolateral dynamic stability during volitional and reactive stepping |
| Expected completion date | Mar 2012 |

| | |
|---|---|
| Estimated size (number of pages) | 200 |
| Elsevier VAT number | GB 494 6272 12 |
| Permissions price | 0.00 USD |
| VAT/Local Sales Tax | 0.0 USD / 0.0 GBP |
| Total | 0.00 USD |
| Terms and Conditions | |

## INTRODUCTION

1. The publisher for this copyrighted material is Elsevier. By clicking "accept" in connection with completing this licensing transaction, you agree that the following terms and conditions apply to this transaction (along with the Billing and Payment terms and conditions established by Copyright Clearance Center, Inc. ("CCC"), at the time that you opened your Rightslink account and that are available at any time at http://myaccount.copyright.com).

## GENERAL TERMS

2. Elsevier hereby grants you permission to reproduce the aforementioned material subject to the terms and conditions indicated.

3. Acknowledgement: If any part of the material to be used (for example, figures) has appeared in our publication with credit or acknowledgement to another source, permission must also be sought from that source. If such permission is not obtained then that material may not be included in your publication/copies. Suitable acknowledgement to the source must be made, either as a footnote or in a reference list at the end of your publication, as follows:

"Reprinted from Publication title, Vol /edition number, Author(s), Title of article / title of chapter, Pages No., Copyright (Year), with permission from Elsevier [OR APPLICABLE SOCIETY COPYRIGHT OWNER]." Also Lancet special credit - "Reprinted from The Lancet, Vol. number, Author(s), Title of article, Pages No., Copyright (Year), with permission from Elsevier."

4. Reproduction of this material is confined to the purpose and/or media for which permission is hereby given.

5. Altering/Modifying Material: Not Permitted. However figures and illustrations may be altered/adapted minimally to serve your work. Any other abbreviations, additions, deletions and/or any other alterations shall be made only with prior written authorization of Elsevier Ltd. (Please contact Elsevier at permissions@elsevier.com)

6. If the permission fee for the requested use of our material is waived in this instance, please be advised that your future requests for Elsevier materials may attract a fee.

7. Reservation of Rights: Publisher reserves all rights not specifically granted in the combination of (i) the license details provided by you and accepted in the course of this licensing transaction, (ii) these terms and conditions and (iii) CCC's Billing and Payment terms and conditions.

8. License Contingent Upon Payment: While you may exercise the rights licensed

immediately upon issuance of the license at the end of the licensing process for the transaction, provided that you have disclosed complete and accurate details of your proposed use, no license is finally effective unless and until full payment is received from you (either by publisher or by CCC) as provided in CCC's Billing and Payment terms and conditions. If full payment is not received on a timely basis, then any license preliminarily granted shall be deemed automatically revoked and shall be void as if never granted. Further, in the event that you breach any of these terms and conditions or any of CCC's Billing and Payment terms and conditions, the license is automatically revoked and shall be void as if never granted. Use of materials as described in a revoked license, as well as any use of the materials beyond the scope of an unrevoked license, may constitute copyright infringement and publisher reserves the right to take any and all action to protect its copyright in the materials.

9. Warranties: Publisher makes no representations or warranties with respect to the licensed material.

10. Indemnity: You hereby indemnify and agree to hold harmless publisher and CCC, and their respective officers, directors, employees and agents, from and against any and all claims arising out of your use of the licensed material other than as specifically authorized pursuant to this license.

11. No Transfer of License: This license is personal to you and may not be sublicensed, assigned, or transferred by you to any other person without publisher's written permission.

12. No Amendment Except in Writing: This license may not be amended except in a writing signed by both parties (or, in the case of publisher, by CCC on publisher's behalf).

13. Objection to Contrary Terms: Publisher hereby objects to any terms contained in any purchase order, acknowledgment, check endorsement or other writing prepared by you, which terms are inconsistent with these terms and conditions or CCC's Billing and Payment terms and conditions. These terms and conditions, together with CCC's Billing and Payment terms and conditions (which are incorporated herein), comprise the entire agreement between you and publisher (and CCC) concerning this licensing transaction. In the event of any conflict between your obligations established by these terms and conditions and those established by CCC's Billing and Payment terms and conditions, these terms and conditions shall control.

14. Revocation: Elsevier or Copyright Clearance Center may deny the permissions described in this License at their sole discretion, for any reason or no reason, with a full refund payable to you. Notice of such denial will be made using the contact information provided by you. Failure to receive such notice will not alter or invalidate the denial. In no event will Elsevier or Copyright Clearance Center be responsible or liable for any costs, expenses or damage incurred by you as a result of a denial of your permission request, other than a refund of the amount(s) paid by you to Elsevier and/or Copyright Clearance Center for denied permissions.

**LIMITED LICENSE**

The following terms and conditions apply only to specific license types:

15. **Translation**: This permission is granted for non-exclusive world **English** rights only unless your license was granted for translation rights. If you licensed translation rights you

may only translate this content into the languages you requested. A professional translator must perform all translations and reproduce the content word for word preserving the integrity of the article. If this license is to re-use 1 or 2 figures then permission is granted for non-exclusive world rights in all languages.

16. **Website**: The following terms and conditions apply to electronic reserve and author websites:
**Electronic reserve**: If licensed material is to be posted to website, the web site is to be password-protected and made available only to bona fide students registered on a relevant course if:
This license was made in connection with a course,
This permission is granted for 1 year only. You may obtain a license for future website posting,
All content posted to the web site must maintain the copyright information line on the bottom of each image,
A hyper-text must be included to the Homepage of the journal from which you are licensing at http://www.sciencedirect.com/science/journal/xxxxx or the Elsevier homepage for books at http://www.elsevier.com , and
Central Storage: This license does not include permission for a scanned version of the material to be stored in a central repository such as that provided by Heron/XanEdu.

17. **Author website** for journals with the following additional clauses:

All content posted to the web site must maintain the copyright information line on the bottom of each image, and
he permission granted is limited to the personal version of your paper. You are not allowed to download and post the published electronic version of your article (whether PDF or HTML, proof or final version), nor may you scan the printed edition to create an electronic version,
A hyper-text must be included to the Homepage of the journal from which you are licensing at http://www.sciencedirect.com/science/journal/xxxxx , As part of our normal production process, you will receive an e-mail notice when your article appears on Elsevier's online service ScienceDirect (www.sciencedirect.com). That e-mail will include the article's Digital Object Identifier (DOI). This number provides the electronic link to the published article and should be included in the posting of your personal version. We ask that you wait until you receive this e-mail and have the DOI to do any posting.
Central Storage: This license does not include permission for a scanned version of the material to be stored in a central repository such as that provided by Heron/XanEdu.

18. **Author website** for books with the following additional clauses:
Authors are permitted to place a brief summary of their work online only.
A hyper-text must be included to the Elsevier homepage at http://www.elsevier.com

All content posted to the web site must maintain the copyright information line on the bottom of each image
You are not allowed to download and post the published electronic version of your chapter, nor may you scan the printed edition to create an electronic version.
Central Storage: This license does not include permission for a scanned version of the material to be stored in a central repository such as that provided by Heron/XanEdu.

19. **Website** (regular and for author): A hyper-text must be included to the Homepage of the journal from which you are licensing at http://www.sciencedirect.com/science/journal/xxxxx. or for books to the Elsevier homepage at http://www.elsevier.com

20. **Thesis/Dissertation**: If your license is for use in a thesis/dissertation your thesis may be submitted to your institution in either print or electronic form. Should your thesis be published commercially, please reapply for permission. These requirements include permission for the Library and Archives of Canada to supply single copies, on demand, of the complete thesis and include permission for UMI to supply single copies, on demand, of the complete thesis. Should your thesis be published commercially, please reapply for permission.

21. **Other Conditions**:

**v1.6**

**If you would like to pay for this license now, please remit this license along with your payment made payable to "COPYRIGHT CLEARANCE CENTER" otherwise you will be invoiced within 48 hours of the license date. Payment should be in the form of a check or money order referencing your account number and this invoice number RLNK500743226.**
**Once you receive your invoice for this order, you may pay your invoice by credit card. Please follow instructions provided at that time.**

**Make Payment To:**
**Copyright Clearance Center**
**Dept 001**
**P.O. Box 843006**
**Boston, MA 02284-3006**

**For suggestions or comments regarding this order, contact RightsLink Customer Support: customercare@copyright.com or +1-877-622-5543 (toll free in the US) or +1-978-646-2777.**

**Gratis licenses (referencing $0 in the Total field) are free. Please retain this printable license for your reference. No payment is required.**