Voluntary Postural Sway Dynamics in Young and Older Adults

Murray G. Tucker
Bachelor of Exercise Science (Hons)

School of Physiotherapy and Exercise Science,
Griffith Health, Griffith University, Gold Coast, Australia

Submitted in fulfilment of the requirements of the degree of Doctor of Philosophy, May 2010
Voluntary Postural Sway Dynamics in Young and Older Adults
ABSTRACT

The general objective of this thesis was to examine the postural responses of young and older adults during reactive and self-paced voluntary postural sway tasks. To achieve this objective, three specific aims were addressed: (1) to determine if differences exist in the reaction time of voluntary postural sway movements between young adults, and low and high falls-risk older adults, (2) to determine if differences exist in coordination during the performance of voluntary postural sway movements between young adults, and low and high falls-risk older adults, and (3) to determine the combination of voluntary postural sway tasks, sway directions, and balance measures that best predict the falls-risk and falls-history status of community-dwelling older adults. The falls-risk and falls-histories of the older adults were determined based on an assessment of sensorimotor and balance function by the Physiological Profile Assessment (PPA), and self-reported 12 months history of falls, respectively.

Four experiments were conducted to examine the performance of voluntary postural sway movements in the anterior-posterior (AP) and medial-lateral (ML) directions. The specific voluntary postural sway tasks were maximum static leans, maximum voluntary postural sway, continuous voluntary postural sway, rapid initiation of voluntary postural sway, rapid termination of voluntary postural sway, and rapid orthogonal switches of voluntary postural sway between the AP and ML directions. Measures of task performance included reaction time, the amplitudes of the centre of pressure (COP) and centre of mass (COM) in the AP and ML directions, the level of coupling between motions of the COP, trunk, and head, and the separation distance between the COP and COM (COP-COM) in the AP and ML directions. The age, falls-risk, and falls-history groups were compared for differences in these measures, and the predictive capacity of these measures for group status were evaluated.
Reaction time during voluntary postural sway in young and older adults

High falls-risk older adults had slower reaction times during the initiation, termination, and orthogonal switch tasks in the AP and ML directions compared with the young adults and low falls-risk older adults. Low falls-risk older adults had slower reaction times during rapid initiation of sway and rapid orthogonal switches of voluntary postural sway for all response directions compared with the young, except for rapid termination of voluntary postural sway. These findings indicate that older adults with more severe age-related declines in postural control physiology have universally slowed voluntary postural sway reaction times, whereas older adults with more subtle age-related physiological declines have slower reaction times during more challenging voluntary postural sway tasks. Multiple fallers had slower orthogonal switch reaction times, AP sway initiation reaction time, and ML sway termination reaction time compared with non-fallers. The significant associations of reaction time with ageing, PPA score, and falls-history indicate that voluntary postural sway reaction times are a composite measure of postural control function that are strongly related to an increased risk of falling among community-dwelling older adults.

Coordination during voluntary postural sway in young and older adults

Older adults had tighter temporal coupling between COP, trunk, and head motions and reduced AP COP amplitudes during voluntary postural sway and orthogonal transitions compared with young adults. The low and high falls-risk older adults exhibited reduced COP-COM separation in the direction of voluntary postural sway for the sway initiation, continuous voluntary sway, and orthogonal switch tasks compared with the young, except during ML continuous voluntary sway for the low falls-risk older adults. These differences in coordination were due to age-related deterioration in postural control. Older adults had a reduced capacity to coordinate movements of the COP with respect to the COM, which limited their capacity to rapidly generate and correct body momentum to stabilise the voluntary postural sway. High falls-risk older adults had reduced ML COP-COM separation and increased ML COM amplitude during ML continuous voluntary sway and AP to ML orthogonal transitions compared with the young and low falls-risk older adults. The results indicate that the high falls-risk older
adults had poor control of their ML COM trajectory that resulted in lateral instability during ML continuous voluntary sway, which was presumably due to deterioration of hip abductor-adductor and/or ankle inverter-everter neuromuscular control.

**Prediction of falls-risk and falls-history using voluntary postural sway measures**

Reaction time and coordination measures of voluntary postural sway performance were significantly predictive of age, falls-risk, and falls-history statuses. Reaction time and AP COP-COM separation during initiation of AP sway predicted the young adults and low falls-risk older adults with 76% sensitivity and 72% specificity. High falls-risk older adults and young adults were predicted with 100% sensitivity and 88% specificity by reaction time and AP COP-COM separation during AP sway initiation and reaction time during ML sway termination. The high and low falls-risk older adults were predicted with 73% sensitivity and 88% specificity by slower AP and ML termination reaction times and reduced ML COP-COM separation during ML continuous voluntary sway. Similar to the predictive accuracy of the PPA and age, multiple fallers and non-multiple fallers were predicted with 80% sensitivity and 100% specificity by ML sway termination reaction time, age, and AP sway initiation reaction time. Collectively, these findings indicate that the reaction time and coordination measures assessed during reactive voluntary postural sway tasks provide useful information about the functional status of the postural control system. Termination reaction times were the strongest predictors of high versus low falls-risk and multiple falls versus non-multiple falls, and therefore an important outcome of the current studies is that a reduced capacity to terminate voluntary postural sway movements is strongly associated with an increased susceptibility to falls.

In conclusion, the current findings provide new insights into the slowing of postural reaction responses and postural instability that occur with ageing and increased falls-risk among older adults. Older adults with greater age-related declines in postural control physiology have particularly slowed postural reaction times and altered postural coordination responses that place them at increased risk of balance loss during dynamic postural movements.
STATEMENT OF ORIGINALITY

This work has not been previously been submitted for a degree or diploma in any university. To the best of my knowledge and belief, the thesis contains no material previously published or written by another person except where due reference is made in the thesis itself. All experiments reported in this thesis complied with the Declaration of Helsinki and the guidelines of the Griffith University Human Research Ethics Committee (GU protocol number: PES/04/05/HREC).

__________________________________________

Murray Tucker, 11\textsuperscript{th} May 2010
The following publications directly support this thesis:

**Publications in peer-reviewed international journals:**


**Conference proceedings:**


# Table of Contents

Abstract .................................................................................................................................................. i
Statement Of Originality ....................................................................................................................... iv
Publications ........................................................................................................................................... v
Table Of Contents ............................................................................................................................... vii
Acknowledgements ............................................................................................................................. ix
Preamble .................................................................................................................................................. xi

General Introduction ............................................................................................................................ 1
  1.1 Background .................................................................................................................................. 1
  1.2 Statement of the Problems ............................................................................................................ 6
  1.3 Significance of the Thesis ............................................................................................................ 7
  1.4 Thesis Aims and Hypotheses ....................................................................................................... 8
  1.5 Definition of Terms ..................................................................................................................... 9
  1.6 Thesis Organisation ................................................................................................................... 11

Literature Review ................................................................................................................................. 14
  2.1 Falls in Older Adults ................................................................................................................... 14
  2.2 Postural Control ......................................................................................................................... 28
  2.3 Postural Control Physiology ....................................................................................................... 52
  2.4 Task Specific Postural Control in Relation to Ageing and Falls ................................................. 62
  2.5 Voluntary Postural Sway and Leaning Movements ..................................................................... 77
  2.6 Summary of Major Findings ....................................................................................................... 85

Experiment One ................................................................................................................................... 89
  3.1 Introduction ................................................................................................................................. 89
  3.2 Methods ...................................................................................................................................... 91
  3.3 Results ......................................................................................................................................... 96
  3.4 Discussion ................................................................................................................................... 101
  3.5 Conclusion .................................................................................................................................. 103

Experiment Two .................................................................................................................................. 105
  4.1 Introduction .................................................................................................................................. 105
  4.2 Methods ....................................................................................................................................... 108
ACKNOWLEDGEMENTS

This thesis represents the culmination to date of many years of undergraduate and postgraduate work. During this period I have been assisted, encouraged, and often inspired by a number of extraordinary individuals.

Above all, I would like to thank the LORD for His amazing grace and for giving me a second chance to do well in life. Never in my wildest dreams did I think I would be capable of achieving a doctorate. Yet, the LORD has forgiven me and has placed me with perfect positioning and timing to help me to achieve my highest aspirations.

Ecclesiastes 9: 11 (New Living Translation)

. . . The fastest runner doesn’t always win the race, and the strongest warrior doesn’t always win the battle. The wise sometimes go hungry, and the skilful are not necessarily wealthy. And those who are educated don’t necessarily lead successful lives. It is all decided by chance, by being in the right place at the right time.

I would like to extend my sincere appreciation to my supervisors Dr. Rod Barrett, Dr. Steven Morrison, and Dr. Justin Kavanagh. I thank all of you for helping me to grow as a researcher, for continually stretching the limits of my understanding, and for teaching me the rigour of scientific investigation.

Rod, I thank you for your delightfully friendly manner, enduring patience, and caring attitude. You have taught me a logical approach to scientific research and also that great patience ultimately leads to better quality work. On many occasions I believed I had done my best work, but you showed me that you can always improve your work if you give it another try. I am especially grateful for the high priority you gave our research, your timely feedback, and that your door was always open to talk irrespective of your busy schedule.
Steve, I thank you for encouraging me to undertake Honours when I approached you in second year Exercise Science, and also for your continued encouragement throughout candidature. In addition to the research, I am grateful that you mentored me in drinking coffee, giving presentations and lectures, attending conferences, teaching, and the importance of maintaining an appropriate balance between work and recreation. The conferences we attended together were very enjoyable.

Justin, I wholeheartedly thank you for your continued dedication to developing my research skills and self-confidence since the early stages of Honours. I don’t think there is any way I could ever repay you for what you’ve done for me over the past five years. The general support that you have given me has well and truly exceeded the job description of a supervisor. Your empathy, encouragement, and advice during some of the more frustrating moments of my candidature were invaluable. My visits to your office and the ensuing conversations about music, academia, sport, and life’s dilemmas were always an enjoyable part of my day.

I would like to thank my parents Glen and Lois and my sister Wendy for their ongoing love, support, and guidance during my University studies. Without your help, I know that this degree would’ve been so much harder for me to complete. It has been a great source of strength to know that you’re always behind me, backing me up in my pursuits, and that you’re willing to do whatever you can to help me to achieve my goals. I must also thank our family friends, Drs. Graham and Teresa Chataway for encouraging me as an emerging scholar and also for allowing me to use your study environment.

I would also like to extend my gratitude to Stuart Gendle for his technical assistance during data collection, and to all of the participants who donated their time to my research. Thanks must also go to Dr. Chris Carty who assisted me in preparing the participants for motion analysis. I’m thankful for the support given by the School of Physiotherapy and Exercise Science staff members and postgraduate students. I would also like to thank my friends from church and school for their encouragement.
PREAMBLE

“For in all the world there are no people so piteous and forlorn as those who are forced to eat the bitter bread of dependency in their old age, and find how steep are the stairs of another man’s house. Wherever they go they know themselves unwelcome. Wherever they are, they feel themselves a burden. There is no humiliation of the spirit they are not forced to endure.”

Dorothy Dix, 1926

A particularly vivid memory comes to mind when introducing the topic of this thesis which I would like to share with the reader. The memory concerns an encounter that I had with an elderly lady who had suffered severe declines in her ability to move and to balance her body. Her mobility and balance limitations were, and still are, incredibly profound to me. Similar to the observations of Dorothy Dix, I had experienced first-hand how dependent some elderly people can become upon others to perform relatively common activities that most of us take for granted. My belief is that we can all learn some useful lessons about the nature of life, ageing, and balance control from thoughtful considerations of the movement-related events surrounding us in daily life. Such considerations emphasise the critical need to preserve the quality of life and independence of older people, and also raise important questions about how a person’s balance should be measured to prevent movement-related problems among older people.

The context of the anecdote is found in the weekend adventures of me and my friends who are a group of young adults brought together from a common interest in surfing. It started off just like any other typical weekend for us. Without fail, the imperative questions asked on the Friday afternoon preceding the weekend by this keen group of surfers were: Is there going to be surf tomorrow? If so, where are we going to surf, and what time are we meeting? The forecast for this particular Saturday was a small south-
east swell with early light winds and afternoon sea breezes. These are staple surf conditions for southeast Queensland, and we knew just how to handle them, so with minimal deliberation we decided to surf at Cudgen beach just south of the Queensland border to take advantage of its smaller crowds and openness to south swell. As always, we hit the surf as close as possible to the crack of dawn to relish in the primest conditions, and the session was filled with much hooting, fun, laughter and quality rides. Experiences such as this are highly characteristic of the exuberance of youth. The vitality and sense of freedom that can be obtained from the ability to interact with nature and friends on such a high level is an enriching factor in any young person’s life.

It was during our traditional post-surf breakfast splurge at the Kingscliff Bakery that I was confronted with the balance and mobility problems endured by some older people. I watched as a four-wheel drive vehicle suddenly pulled up outside the bakery and the left-rear door slowly crept open to reveal an elderly lady. She appeared quite frail and I estimated her age to be about 85 years. Her frailness was visually evident in the whitish, bony, and fragile frame of her body. I continued to watch with intrigue as she attempted to exit the vehicle. Her facial expressions changed from the depiction of a sombre demeanour to one of consternation as her best efforts to exit the vehicle became unsuccessful. She appeared too weak to perform the subtasks required to safely transfer her body from a seated position in the car to a standing position on the adjacent pavement. Even if she was somehow able to rotate her body towards the door opening and initiate a downwards step onto the pavement, she would’ve almost certainly lost her balance and fell onto the ground. After a few extra moments of time, a fellow passenger I assumed was a close relative came to her assistance. He helped her out of her seat and supported her body weight as she stepped off the door ledge onto the pavement. Based on her low level of mobility and balance function, it was obvious to me that she experienced great difficulty in performing many other activities of daily living.

My next thoughts contemplated the striking differences in her balancing capabilities compared to the balancing capabilities of myself and my friends. How was it that we
could excel in a highly challenging postural activity whereas she struggled in her basic everyday activities? It is not uncommon to witness surfers gracefully gliding over the top of a wave’s falling lip on the brink of catastrophe, and yet completely in control of the combined movements of their posture and the surfboard underneath their feet. Riding through the barrel of a wave requires continual weight adjustments within a confined space to avoid the momentum of the lip despite perturbation from the spitting and hissing of sand, foam, and water in one’s face. In contrast, at a very basic level of postural control, the elderly lady could not control her body position to execute a simple balance transfer despite the availability of objects to grasp onto for support. Although we could withstand sensory and mechanical perturbations of our posture to maintain balance on our surfboards, a slight nudge to the lady’s sternum probably would’ve led to an unrecoverable loss of balance. We could exploit nature for all its worth, whereas the environment for the lady was a continual burden and a source of great difficulty and despair. These extreme differences in balancing capabilities and the resultant effect on quality of life were truly profound.

I then wondered at what point she began to lose normal balance function. When did she become unable to perform a relatively simple balance task that most individuals perform without much conscious effort? What are the age-associated changes in postural control which could lead to such a drastic deterioration in her balance compared to the days of her youth? Were the changes slow and insidious, or did some type of disastrous neurological event associated with the processes of disease occur? Of course my questions regarding this specific lady will remain unanswered, but some general insights may be gained from my observations of other older people. Some of the typical movement-related problems I have seen in people of advanced age include unintentional falls, dependence on a gait assistive device, muscle weakness, fear of falling, restriction of activities, slowness of movement, loss of independence, and difficulty in performing simple movement tasks such as leaning, bending down, walking, and stair negotiation. Any one of these conditions would render an older person unable to participate in busy life so that they must remain a spectator on the sideline. One of the biggest problems of older age is the loneliness from losing touch
with friends and acquaintances. Many older people don’t venture out of their home a
great deal because of the overwhelming movement challenges associated with
unfamiliar or unpredictable environments. They mostly stay within their place of
dwelling, and that’s it.

Any person who has contemplated these issues and also concerns for the welfare of
older people would be aware of the important need to prevent the occurrence of
mobility and balance problems amongst the elderly. To undertake such a task, one is
faced with the initial problem of how to accurately measure a person’s mobility and
balance. Firstly, although mobility and balance are often coupled together when
describing movement, they are separate entities. For example, a mobile person can
move freely throughout the environment and to do so they must be able to balance
their body. Therefore, balance is a prerequisite for effective mobility. So how do we
measure balance then? If we knew the answer to this question, the problem of falling
by older people would’ve been solved a long time ago. There are countless aspects of
balance which can be measured because there are a vast number of physiological
elements involved in the generation of balance responses. One could potentially
measure various indicators of anatomical structure, brain activity, neuromuscular
activation, sensory acuity, muscle strength or power, speed of movement, movements
or non-movements of the body or its segments, rotations of the head and eyes, or
measures relating to force production.

Similar to the ‘bird-egg’ problem which was pondered by philosophers such as Plato
and his pupil Aristotle, there is also a causality dilemma between an individual’s level
of balance and the movement patterns they adopt. For instance, older people are
known to have different movement patterns compared with young people. Does this
reflect that the older person became unstable and so they adopted a new movement
pattern to achieve a similar degree of stability to that of the young, and therefore they
are no longer unstable? Or does the simple fact that they adopted a different
movement pattern indicate that the person is in some way unstable even though this
new pattern may facilitate stability? Questions such as these are difficult to answer
without a truthful, all encompassing measure of postural stability. However, even if researchers and clinicians were able to derive a veridical measure of balance, it would not apply universally to all individuals, tasks, and environments because balance depends on interactions between these factors. Therefore, the most practical approaches to screening the susceptibility of older adults to falls are attempts to recreate the task and environment situations in which falls are most likely to occur, or the development of simple balance tests that generalise to such situations. In this thesis, the latter approach was adopted in the hope it would add another very small piece to the entire puzzle of age-associated changes in balance and falls-risk.

CHAPTER 1

GENERAL INTRODUCTION

1.1 Background

In an average person’s lifetime, they will perform a great number of different movement tasks within a variety of different environments, each presenting unique challenges to the regulation of posture (Newell, 1986, Shumway-Cook and Woollacott, 2001). For the most part, humans are able to successfully maintain their posture to achieve the overall movement goal in spite of these unique challenges. It may be questioned how this remarkable feat of human movement control is accomplished. The postural control system underlies the ability of humans to maintain the body in a specific position or to progress through a deliberate movement without a loss of balance (Melvill Jones, 2000). The postural control system of humans is highly developed and represents the coordinated actions of the sensory, cognitive, motor, neuromuscular, and musculoskeletal systems (Horak, 2006, Stelmach and Worringham, 1985, Grabiner and Enoka, 1995). The sensory systems, namely the visual, vestibular and somatosensory senses, provide continually updated feedback regarding the position and motion of the body with respect to the environment (Shumway-Cook and Woollacott, 2001). Sensory information is processed within the central nervous system (CNS) so that an appropriate postural response can be selected, if one is required (Johansson and Magnusson, 1991). Postural responses are then executed through the neuromuscular system in which muscular strength is ultimately used to adjust posture and to maintain balance (Nashner and McCollum, 1985). Ideally, the postural response is implemented quickly so that balance is preserved, and with no disruption to the movement task (Stelmach and Worringham, 1985, Grabiner and Enoka, 1995).
Chapter One

The normal ageing process is associated with deterioration of the postural control system. Age-related degenerative changes in the sensory, cognitive, motor, neuromuscular, and musculoskeletal systems reduce the capacity of older adults to regulate their posture and balance (Woollacott, 1993, Lord and Sturnieks, 2005, Horak et al., 1989). As older adults are a relatively heterogeneous group, some older individuals experience only small age-related declines in postural control compared with their younger counterparts (Lord et al., 2007). In contrast, some older adults experience marked loss of function despite appearing healthy with no obvious clinical signs or symptoms of disease (Lord et al., 2007). Older adults with disease-related impairments to postural control such as stroke or Parkinson’s disease often experience a drastic loss of function and thus have a high susceptibility to balance loss during everyday activities (Sturnieks et al., 2008b). With regards to the detection of stimuli, the sensory systems undergo extensive deterioration with ageing, and this has a negative effect on the abilities of older adults to monitor the position and motion of the body and to detect environmental stimuli (Horak et al., 1989). Response selection processes are also negatively influenced by ageing, most notably the length of time required to select an appropriate response (Welford, 1977). Further age-related decline occurs in the execution of postural responses. In particular, older adults tend to have reduced leg muscle strength (Frontera et al., 2000) and exhibit differences in muscle activation patterns and response strategies (Horak et al., 1997, Manchester et al., 1989) compared with young adults. The ultimate outcome of these age-related changes is the generation of postural responses that are less effective in maintaining postural stability (Lord et al., 2007, Horak, 2006).

One important aspect of a person’s postural control is their ability to rapidly react to maintain their balance (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Reaction time is the time required to initiate a movement as rapidly as possible to a suddenly presented stimulus (Schmidt and Wrisberg, 2004). Fast reaction times are important during postural tasks because there is often a limited amount of time in which to generate an effective movement response to a disturbance of balance (Lajoie and Gallagher, 2004). If an appropriate postural response is not initiated before the
critical time boundary, balance becomes unrecoverable and a fall will occur (Stelmach and Worringham, 1985). It has been well-established that ageing results in a slowing of reaction time over the course of the adult lifespan (Cerella, 1985, Welford, 1977, Welford, 1984). However, age-related changes in reaction time have been typically assessed under task conditions in which participants reacted with the upper limbs from a static position whilst seated in a chair. While such measures provide valuable insight into the slowing of cognitive processing speed with ageing, the reaction times obtained under these task conditions may not generalise to situations in which standing postural control must be maintained (Weerdesteyn et al., 2004, Lajoie et al., 1993, Lajoie et al., 1996). A requirement to maintain an upright stance and to react while the body is in motion may influence the slowing of reaction time with ageing (Weerdesteyn et al., 2004, Lajoie et al., 1993, Lajoie et al., 1996). Examination of reaction time under these posturally challenging conditions may also provide a greater understanding of the mechanisms underlying falls by older adults (Lord and Fitzpatrick, 2001).

Although older people are most likely to lose their balance whilst walking (Bradley and Pointer, 2009, Berg et al., 1997), any assessment of reaction time and balance changes during locomotion are difficult to accurately measure. An alternative method of examining the dynamics of postural responses (e.g., speed, stability, and coordination properties) is to measure the abilities of young and older adults to perform reactive and self-paced voluntary postural sway movements (Morrison et al., 2007). Voluntary postural sway is a balance and coordination task in which individuals are required to rhythmically oscillate the body in the anterior-posterior (AP) or medial-lateral (ML) directions. Voluntary postural sway movements can also be performed as a reaction time task when the task requirement is to sway as rapidly as possible in the direction specified by a suddenly presented stimulus. Importantly, voluntary postural sway significantly challenges the maintenance of standing postural control because of the large and intentional sways of the body towards the limits of stability (Winter, 1995). This is likely to place a high level of demand upon the sensory (Kavounoudias et al., 1998), cognitive (Lajoie et al., 1996) and neuromuscular (Melzer et al., 2009a) systems
to ensure that standing balance is adequately maintained. Voluntary postural sway movements therefore represent a convenient model for examining age-related deficits in postural control because of the simplicity of assessment and the potential to systematically manipulate the aspects of the task.

Research examining the dynamics of voluntary postural sway actions for healthy younger and older adults is currently limited. The majority of previous studies have not examined age-related differences in the characteristics of continuous voluntary postural sway, but have rather examined the abilities of young and older adults to lean maximally in multiple directions (Murray et al., 1975, King et al., 1994, Blaszczyk et al., 1994). These studies almost invariably found an age-related reduction in maximum static lean amplitudes, a result which is argued to demonstrate that ageing reduces the capacity to shift the body maximally within the base of support area (Blaszczyk et al., 1994, Murray et al., 1975). However, other investigations have not reported similar associations with prospectively measured falls among older adults, which suggests that maximum leaning tests are poor predictors of falls (Boulgarides et al., 2003, Brauer et al., 2000). Two other common tasks that involve voluntary postural sway movements are the Limits of Stability and Rhythmic Weight Shifts tests (Cheng et al., 2004, Clark et al., 1997). In these tests, participants are required to lean/sway towards targets positioned within their base of support area as accurately as possible whilst guided by visual feedback. Task performance can be assessed using measures of reaction time, movement time, path accuracy of sway, and maximum lean amplitude. During the Limits of Stability and Rhythmic Weight Shifts tasks, ageing and balance impairments related to falling have been associated with slower sway responses (Nitz et al., 2003, Liaw et al., 2009, Ben Achour Lebib et al., 2006, Hageman et al., 1995) and reduced path accuracy of sway with respect to the target direction of movement (Delbaere et al., 2006a, Liaw et al., 2009, Hageman et al., 1995). Taken together, the aforementioned research indicates that ageing and an increased risk of falling are significantly associated with a reduced capacity to perform voluntary postural sway movements.
A number of important questions regarding the performance of voluntary postural sway movements remain unanswered. In particular, issues relating to the control mechanisms employed to regulate postural stability during voluntary postural sway have not been explored. Such information may provide valuable insights as to why particular groups of older adults are more susceptible to falls compared with others. In addition, the motions of the body’s segments during voluntary postural sway actions have not been examined. Therefore, the inter-segmental dynamics involved in maintaining postural stability during voluntary postural sway are unknown. It is also unclear whether the reaction time, stability, and coordination of voluntary postural sway movements changes due to ageing or falls-related conditions because no single study has examined these issues in detail. For example, it is currently unknown whether the reaction time of voluntary postural sway movements is slower in fallers compared with non-fallers and whether such reaction time measures are useful markers of falls-risk. The findings of previous research are also limited because many investigations of voluntary postural sway or leaning actions have averaged the response directions to obtain an overall index of performance, or have examined motion in a single plane only (see for example, Lord and Fitzpatrick, 2001, Stelmach et al., 1989, Nitz et al., 2003, Borah et al., 2007, Delbaere et al., 2006b). A limitation of this approach is that important scientific and clinical information may be lost, particularly as the AP and ML directions of postural sway are separately and independently regulated by the postural control system (Winter, 1995, Winter et al., 1996, Winter et al., 1993). It is also important to examine the AP and ML directions of voluntary postural sway separately because a growing body of research indicates that older adults with falls and balance problems experience particular declines in ML postural stability rather than AP postural stability (Piirtola and Era, 2006, Rogers and Mille, 2003). Given that ageing and falls are associated with deterioration of voluntary postural sway in a single plane (Delbaere et al., 2006b, Liaw et al., 2009), older adults may experience particular difficulty to coordinate a rapid orthogonal transition of voluntary postural sway between the AP and ML directions. Therefore, there is a current need for studies examining voluntary postural sway in the AP and ML
directions of movement separately, and orthogonal transitions of voluntary postural sway between the AP and ML directions in different age and falls-risk groups.

1.2 Statement of the Problems

A number of studies have compared the speed, accuracy, and maximum amplitude of voluntary postural sway or leaning actions between young and older adults and between fallers and non-fallers (Nitz et al., 2003, Delbaere et al., 2006a, Liaw et al., 2009, Ben Achour Lebib et al., 2006, Hageman et al., 1995). However, the findings of these studies are limited because reaction time and coordination measures of voluntary postural sway performance were not examined in detail. In particular, it is unknown whether slower voluntary postural sway reaction times are associated with declines in sensorimotor function and increased falls-risk among older adults. In addition, voluntary postural sway has been typically measured using a single variable called the centre of pressure (COP) which is related to overall body movement. Therefore, the coordination between body segments and postural parameters during voluntary postural sway, and any changes in this coordination with ageing and falls-risk is also unknown.

Only a few studies have examined the capacity of voluntary postural sway measures to accurately differentiate between older adults with differing levels of balance function. The main finding of these studies was that the amplitude of maximum static lean is a poor predictor of falls in healthy older adults (Boulgarides et al., 2003, Brauer et al., 2000). Although there is some suggestion that speed and accuracy measures of voluntary postural sway can significantly predict different classifications of falls-status (Delbaere et al., 2006a, Delbaere et al., 2006b), the combinations of voluntary postural sway tasks, movement directions, and performance measures that form the best predictive models for ageing and falls-risk effects on voluntary postural sway is unknown.

Older adults form a relatively heterogeneous group with some individuals experiencing greater rates of decline in their postural control compared with others
(Lord et al., 2007, Shumway-Cook and Woollacott, 2001). As previous investigations of voluntary postural sway have examined young adults compared to relatively heterogeneous groups of older adults, a more focussed analysis of age-related differences in voluntary postural sway movements may be beneficial. The division of healthy older adults into separate groups based on an assessment of sensorimotor function is likely to provide a more comprehensive understanding of the differences in voluntary postural sway performance that can be expected between adults with differing ages and levels of balance impairment.

1.3 Significance of the Thesis

As postural control underlies the ability to successfully perform movements, it is important to develop an increased understanding of the postural control system and its operational mechanisms. Any attempt to advance the understanding of postural control is a difficult undertaking because the human postural control system is remarkably complex (Horak, 2006). For example, the generation of any postural response is achieved via the coordination of a vast number of elements of the human system such as neurons, muscles, and joints (Bernstein, 1967). In addition, a common view is that the coordination of these elements and the generated postural responses are uniquely shaped by factors relating to the characteristics of the individual (e.g., anthropometry, age, or disease), the movement task, and the environment in which the task is performed (Shumway-Cook and Woollacott, 2001, Newell, 1986). Therefore, empirical observations obtained under one set of conditions using a particular group of participants may not necessarily generalise to different sets of conditions or participant samples. Although a considerable amount of scientific progress has been made in postural control research, many important research questions remain unanswered. In particular, a greater understanding of the most important variables involved in the maintenance of balance and the variables that should be targeted in preventing falls is currently required. Examination of age-related differences in postural control and differences in postural control between older adult fallers and non-fallers is likely to provide insights into the key factors associated with postural instability.
Another issue of concern to researchers and the wider community alike are falls and mobility problems among older people. Approximately one third of community-dwelling older adults aged 65 years and above fall each year with higher rates occurring in the particularly old those living in aged-care institutions (Rubenstein and Josephson, 2002, Hill et al., 2004, Lord et al., 2007). Although most falls do not result in serious injury, they can still result in psychological problems, reduced mobility, activity restriction, social isolation, depression, and reduced independence (Bradley and Pointer, 2009, National Institute for Clinical Excellence, 2004). Many older adults restrict their activities following a fall, which over time leads to a reduction in their mobility and physical fitness, and eventually proceeds into a downward spiral of frailty (Arfken et al., 1994, Delbaere et al., 2004). Frail individuals typically have reduced muscle strength, fatigue easily, are physically inactive, exhibit a slow and unsteady walking pattern, and have an increased risk and fear of falling (Arfken et al., 1994, Delbaere et al., 2004). Frail older adults are also at an increased risk of disability, morbidity, and mortality (Fried et al., 2001). Falls also represent the leading cause of injury-related death for older adults with 2,645 deaths recorded in Australia during 2004-05 (Henley and Harrison, 2009). The economic cost associated with falls is considerable as their total cost was estimated at $AUD 406.6 million during 1993-94 (Mathers and Penm, 1999) and 566 million during 2003-04 (Bradley and Harrison, 2007). Therefore, it is clear that falls have profound physical, social, and economic consequences and that these events must be predicted and prevented.

1.4 Thesis Aims and Hypotheses

The general aim of this thesis was to examine the postural responses of young and older adults during reactive and self-paced voluntary postural sway tasks.

The specific aims of this thesis were to:

(1) Determine if differences exist in the reaction time of voluntary postural sway movements between young adults, low falls-risk older adults, and high falls-risk older adults.
(2) Determine if differences exist in coordination during the performance of voluntary postural sway movements between young adults, low falls-risk older adults, and high falls-risk older adults.

(3) Determine which combination of voluntary postural sway tasks, sway directions, and balance measures best predict falls-risk (high falls-risk versus low falls-risk) and falls-history status (multiple fallers versus non-multiple fallers) of older adults.

The specific hypotheses in relation to these aims were that:

(1) Voluntary postural sway reaction times would be slower for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults.

(2) Coordination during the performance of voluntary postural sway movements would be altered for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults.

(3) Reaction time and coordination measures of voluntary postural sway tasks would accurately predict the group status of the high and low falls-risk older adults and the multiple fallers and non-multiple fallers.

1.5 Definition of Terms

For the purpose of this thesis, operational definitions for key terms are as follows:

Voluntary Postural Sway

Voluntary postural sway is a movement task requiring rhythmical oscillations of the body in the AP or ML directions at a self-selected frequency. Voluntary postural sway in the AP direction involves oscillation of the body primarily about the ankle joint with
minimal rotations at the hip and knee joints. Voluntary postural sway in the ML direction involves sequential load-unload movements between the legs in the frontal plane. Voluntary postural sway movements are performed with arms hanging relaxed alongside the trunk and with little to no movement of the feet. Orthogonal transitions represent a switch of voluntary postural sway from the AP to ML direction or from the ML to AP direction.

Reaction time

Reaction time is the time required to initiate a specific movement as rapidly as possible. It was measured as the period of time between the presentation of an unanticipated auditory cue to the beginning of the movement response to the cue (Schmidt and Lee, 1999). Voluntary postural sway reaction times represent a rapid voluntary sway movement, initiated as fast as possible, in the direction indicated by an auditory cue whilst standing balance is maintained.

Coordination

Coordination is generally defined as the relationship between movements of the body’s elements (Kurz and Stergiou, 2004). Two primary measures of postural coordination were used in this thesis. The first measure was the strength and direction of temporal coupling between motions of the COP, trunk, and head. The second measure was the separation distance between movements of the COP and the centre of mass (COM) in the AP and ML directions.

Falls-Risk

A person’s falls-risk represents their susceptibility to experiencing an unintentional fall during activities of daily living. The two primary measures of falls-risk used in this thesis were Physiological Profile Assessment (PPA) falls-risk score and 12 months history of falls.
General Introduction

Postural Control System

The human postural control system represents the coordinated actions of the sensory, cognitive, motor, neuromuscular, and musculoskeletal physiological systems. These systems act cooperatively to regulate the body’s position in space so that the body’s position facilitates the movement goals of the individual.

1.6 Thesis Organisation

Chapter 1 is a general introduction that presents background information on posture, balance, and voluntary postural sway movements. The problems to be addressed in this thesis are outlined in relation to previous literature, which is followed by a statement of aims, hypotheses, operational definitions, and the significance of the thesis.

Chapter 2 reviews the literature pertinent to this thesis, opening with a review of falls and common falls-risk assessments. A description of posture and the postural control system is then presented, which is followed by a review of the biomechanics of standing postural control and the measurement of standing postural stability. The decline in postural control physiology due to ageing and in older adult fallers compared with non-fallers is then summarised. Changes in the performance of common movement tasks with ageing and increased falls-risk among older adults are reviewed, which is followed by a detailed summary of voluntary postural sway investigations involving healthy young and older adults. The literature review concludes with a summary of findings.

Chapter 3 is the first experimental investigation of this thesis. The study examined differences between younger and older men in reaction time and the coordination of postural responses during rapid initiation and orthogonal switches of voluntary postural sway.

Chapter 4 is the second experimental investigation of this thesis. The study examined differences between young adults, and low and high falls-risk older adults in reaction
time and postural stability during voluntary postural sway and orthogonal switches of voluntary postural sway.

Chapter 5 is the third experimental investigation of this thesis. The study examined differences between young adults, and low and high falls-risk older adults in reaction time and postural stability during continuous voluntary postural sway, and rapid initiation and termination of voluntary postural sway. The capacity of the task measures to discriminate between the groups was evaluated.

Chapter 6 is the fourth experimental investigation of this thesis. Voluntary postural sway measures similar to those used in Chapters 3, 4, and 5 were evaluated for their capacity to differentiate between and to accurately predict the falls-history status of older adults.

Chapter 7 is a general discussion which presents a summary and synthesis of the experimental findings and the conclusions of the thesis.
2.1 Falls in Older Adults

2.1.1 Epidemiology of falls

A fall is commonly defined as “an event that results in a person coming to rest unintentionally on the ground or some lower level, and not the result of a violent blow, overwhelming hazard, loss of consciousness, sudden onset paralysis, or an epileptic seizure” (Gibson et al., 1987). This definition is appropriate to register falls that result from an unexpected loss of balance, however some studies have broadened the definition of falls to include those resulting from major intrinsic events such as stroke or syncope (Lord et al., 2007). Despite the different definitions of falls, many studies have reported similar rates of falls in populations of older adults, particularly in individuals older than 65 years (Lord et al., 2007). Retrospective studies that randomly selected large numbers of participants from the Australian population have found that approximately one in three older adults fall each year (Lord et al., 1993, Lord et al., 1994a, Dolinis et al., 1997, Gill et al., 2005), and that one in ten older adults fall on multiple occasions (Lord et al., 1993, Lord et al., 1994a). Internationally, similar rates of falling have been reported in New Zealand (Campbell et al., 1989), United Kingdom (Prudham and Evans, 1981, Blake et al., 1988), United States of America (Tinetti et al., 1988), and Canada (O’Loughlin et al., 1993). Falls rates are also known to increase substantially with age, with approximately one in two community-living individuals aged 80 years and above falling at least once per year (Tinetti et al., 1988, Lord et al., 1994a, Lord et al., 1993, O’Loughlin et al., 1993). The prevalence of falls for residents of aged-care institutions is also considerably higher compared with older adults living independently within the wider community (Yip and Cumming,
1994, Rubenstein et al., 1994, Lipsitz et al., 1991, Lord et al., 1991a), suggesting that frail and less mobile adults are particularly susceptible to falls.

The impact of falls is compounded by the increased susceptibility of older people to injury, burdensome sequelae, and worldwide increases in the number of people older than 65 years (Department of Ageing and Life Course, 2007). Falls are a significant cause of hip and wrist fracture, fear of falling, restriction of activities, loss of mobility and quality of life, social embarrassment, admission to aged-care institutions, and injury-related death (Lord et al., 2007, Department of Health and Ageing, 2005). In a study of 50 community-living older adults who were hospitalised due to a fall, 27% of these individuals had three or more additional falls, were readmitted to hospital, transferred to a nursing home, or died in the following 12 months (Lord et al., 1992). Falls are the leading cause of injury-related death for older adults with 2,645 deaths recorded in Australia during 2004-05 (Henley and Harrison, 2009). Bradley and Pointer (2009) reported that 66,784 older adults were admitted to Australian hospitals due to a fall in 2005-06, which represented 68% of all injury-related hospitalisations for this group. It was additionally found that falls injury rates increased exponentially after 65 years, falls injury rates were higher for females compared with males, and that most falls were due to slipping, tripping, or stumbling (Bradley and Pointer, 2009).

Demographic change in Australia is predicted to result in an unprecedented two-fold increase in the proportion of individuals older than 65 years over the next 40 years (Møller, 2003). This increased proportion of older people will increase the total number of falls per annum and the demand for healthcare services. As a result, economic projections indicate that falls will cost Australians approximately $AUD 1.375 billion per annum by 2051 unless effective falls prevention initiatives are implemented (Møller, 2003).

2.1.2 Falls-risk factors

In the last 30 years, there have been 1000’s of scientific studies on falls by older people and the risk factors associated with falls (Lord et al., 2007, Close, 2005a). In general, most of these studies have focussed on three main areas which include
examining age-related differences in tasks involving the regulation of posture, identifying the specific risk factors that are strongly predictive of falls by older adults, and determining which kinds of interventions can successfully prevent falls. Overall, this body of research has reported four main findings: (1) falls are not a normal consequence of ageing despite the increased frequency of falls in older age (i.e., not all older adults experience falls), (2) falls are caused by multifactorial and diverse processes, (3) multifactorial and individualised intervention strategies are the most effective in preventing falls, and (4) the types of falls-risk factors and the effectiveness of interventions depends on whether the setting is within the community, an aged-care institution, or a hospital (Todd and Skelton, 2004, Hill et al., 2004, Gillespie et al., 2009, McClure et al., 2005, Kellog International Work Group, 1987, National Institute for Clinical Excellence, 2004).

At present, the general consensus is that further research needs to be performed in the area of falls prevention, especially in the development of assessments that can predict an individual’s likelihood of falling in the future (i.e., their falls-risk) (Todd and Skelton, 2004, Hill et al., 2004, Gillespie et al., 2009, Department of Ageing and Life Course, 2007). However, this is not a straightforward task because over 400 potential risk factors have been identified with falls (Gillespie, 1998). Systematic reviews of the falls literature have summarised the main risk factors that predispose older adults to falls (Table 2.1; National Institute for Clinical Excellence, 2004, Hill et al., 2004, Rubenstein and Josephson, 1996). These risk factors can be broadly classified into those factors that are intrinsic or extrinsic to the individual, and the individual’s exposure to risk of falling based on the tasks they undertake and the behaviours they adopt (Todd and Skelton, 2004). Research has shown that the risk factors for falling have an additive effect, and therefore the more risk factors an individual has the greater their likelihood of falling (Cwikel et al., 1998, Robbins et al., 1989, Tinetti et al., 1988, Tinetti et al., 1986, Nevitt et al., 1989).

The falls-risk factors with the strongest and most consistent associations with falls are intrinsic to the individual (Lord et al., 2007, Lamb et al., 2008, Rubenstein and
Josephson, 1996, Perell et al., 2001, Hill et al., 2004, National Institute for Clinical Excellence, 2004). Intrinsic risk factors affect the functioning of the postural control system, which has the task of regulating the posture and balance of the body (see section 2.2). Any factor that has an adverse effect on a person’s postural control system increases their susceptibility to a loss of balance. For example, ageing and certain diseases can result in deterioration of the sensory, cognitive, motor, neuromuscular, and musculoskeletal systems. This deterioration in postural control physiology produces deficits in balance and gait performance, thereby increasing susceptibility to falls during everyday activities (Lord and Sturnieks, 2005, Sturnieks et al., 2008b, Shumway-Cook and Woollacott, 2001, Romero and Stelmach, 2003). The functioning of postural control physiology is also negatively affected by the side-effects of some medications. In particular, medications which directly alter the activity of the central nervous system (CNS) such as benzodiazepines have been strongly linked with falls after adjustment for other confounding factors (Leipzig et al., 1999a).

Table 2.1
Summary of important risk factors for falls in older people.

<table>
<thead>
<tr>
<th>Intrinsic Risk Factors</th>
</tr>
</thead>
<tbody>
<tr>
<td>Past history of falls</td>
</tr>
<tr>
<td>Advanced age</td>
</tr>
<tr>
<td>Female sex</td>
</tr>
<tr>
<td>Medications acting on the CNS that slow reaction time or cause postural hypotension</td>
</tr>
<tr>
<td>Deterioration of sensory, cognitive, motor, neuromuscular, and musculoskeletal physiology</td>
</tr>
<tr>
<td>Diseases (e.g., Parkinson’s disease, cognitive impairment, and stroke)</td>
</tr>
<tr>
<td>Sedentary lifestyle</td>
</tr>
<tr>
<td>Psychological status (e.g., fear of falling, attitudes)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Extrinsic Risk Factors</th>
</tr>
</thead>
<tbody>
<tr>
<td>Environmental hazards</td>
</tr>
<tr>
<td>Inappropriate footwear, clothing, or gait assistive device</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Task Risk Factors</th>
</tr>
</thead>
<tbody>
<tr>
<td>Balance requirements of the task</td>
</tr>
<tr>
<td>Risky behaviour (e.g., unsafe movements, rushing, dual tasking)</td>
</tr>
</tbody>
</table>
The combined effects of four or more medications, which is referred to as polypharmacy, is also regarded as a prominent falls-risk factor amongst older people (Zeiere et al., 2006, Leipzig et al., 1999b). Female sex is also a risk factor for falling because older women have higher rates of falling compared with older men (Stevens and Sogolow, 2005, Campbell et al., 1990b). It has been suggested that older women have increased falls-risk because of reduced muscle strength and a greater reliance on visual information for balance control compared with older men (Lord et al., 1994a). Unsurprisingly, a recent history of falling is also a risk factor for subsequent falls (Tinetti et al., 1988, Nevitt et al., 1989). This association may be related to the continued influence of the risk factors that resulted in the initial fall, but may also be due to new risk factors caused by the fall such as injury, reduced mobility, and fear of falling (Kiel et al., 1991, Scheffer et al., 2008, Tinetti and Williams, 1998). Lack of physical activity is an important falls-risk factor because physical activity can prevent age-related losses of strength, flexibility, and mobility, which all play a role in balance control (Gillespie et al., 2009, Lord et al., 2007). However, high doses of exercise and participation in activities that greatly exceed the balance capabilities of the individual may increase exposure to falls-risk situations and thus actually facilitate falls (Stel et al., 2003a, Graafmans et al., 2003, Peeters et al., 2009).

Extrinsic risk factors generally refer to those factors that are a part of the environment. Although extrinsic risk factors have weaker associations with falls compared with intrinsic risk factors, a number of environmental hazards have been consistently identified as contributing to falls (Lord et al., 2006, Lord et al., 2007). Some relatively common environmental hazards include poor lighting, obstacles, irregular or slippery terrain, raised floor edges, and an absence of handrails or objects to grasp (Connell, 1996b). A number of studies found that older adults who were physically active were more likely to have a fall involving an environmental hazard compared with older adults who are frail (Northridge et al., 1995, Speechley and Tinetti, 1991). Perhaps this association is due to a less cautious approach to movement by the active older adults or alternatively a greater level of exposure to falls-risk situations. There is evidence to suggest that the level of exposure to
environmental hazards is associated with falls because most falls occur during periods of maximum activity throughout the day (Downton and Andrews, 1991, Campbell et al., 1990a). Other extrinsic risk factors for falls include inappropriate clothing and footwear (Menant et al., 2008), unsuitable or incorrectly used gait assistive devices (Dean and Ross, 1993), and the use of multifocal spectacles that can visually blur hazards (Lord et al., 2002a).

Most falls by older adults occur during walking rather than during sitting or standing (Bradley and Pointer, 2009, Norton et al., 1997, Berg et al., 1997, Cali and Kiel, 1995). This suggests that falls are most likely to occur during tasks that place a high level of demand upon posture and balance control. Therefore, an important consideration for determining an individual’s falls-risk during a given activity is whether their balance ability is adequate with respect to the balance demands of the task (Shumway-Cook and Woollacott, 2001, Lord et al., 2007). For example, if an individual has a high functional level of balance, it is likely that they will be able to avoid falls and recover from episodes of balance loss during their activities of daily living (Lord et al., 2007, Close, 2005b). Other movement activities that can be challenging for older people include transfers from a lying or sitting position to a standing position, dual tasking (e.g., walking and talking at the same time), carrying loads, climbing ladders, and standing on chairs to reach high objects (Hill et al., 1999, Lord et al., 1993). The risk of falling during these hazardous activities may be further enhanced by dangerous behaviours such as rushing, inattentiveness, unsafe movements beyond the limits of stability, or continuing to perform a task despite difficulty, discomfort, or fatigue (Connell, 1996b).

Whilst the identification of separate risk factors for falls is a fundamental step towards the development and implementation of falls-risk assessment tools and intervention strategies, falls are multifactorial events that result from interactions between risk factors. Newell (1986) proposes a paradigm whereby falls result from the interaction between the balance limitations of the individual (intrinsic factors), the demands of the particular movement task they are performing (task factors), and the properties of
the environment in which the task is performed (extrinsic factors). This theory of movement control has been referred to as a ‘systems’ approach by other researchers because of the requirement for dynamic interplay between physiological systems of postural control (Horak et al., 1997, Horak et al., 1989, Shumway-Cook and Woollacott, 2001, Woollacott, 1993). Given this complex interaction between intrinsic, extrinsic and task factors, and also the transient nature of particular combinations of falls-risk factors, falls by older people are not easily predicted or prevented (Kannus et al., 2005, Huxham et al., 2001, Horak, 2006).

Although it is unlikely that all falls by older people could ever be prevented (Lord et al., 2007), there are currently some important gaps in knowledge which may aid efforts to prevent falls in the future. It is known that specific risk factors and their cumulative effects increase susceptibility to falls, however the manner by which these factors interact to result in an episode of balance loss is both rarely examined and poorly understood. Studies are also required that identify the causative factors in isolated fall events, and which distinguish between these causative factors and other factors which may be associated with poor balance but did not contribute to the fall. This view is consistent with that of Horak (2006), who emphasises that an individual who has physiological impairments associated with falling may still be able to avoid falls by using compensatory movement strategies to facilitate their balance. Moreover, the identification of factors actually causing falls would improve the accuracy of falls prediction and would help to develop more focussed falls prevention strategies. Although studies of age-related differences in postural control have been beneficial in understanding falls, there is also a lack of understanding of why some groups of healthy older adults are particularly susceptible to falls compared with others.

2.1.3 Falls-risk assessments

A key element in the prevention of falls is the need to identify individuals that are likely to fall in the future. Following the identification of high falls-risk individuals, a multifactorial intervention programme can be created, which may include
components such as education, balance training, and home hazard modification (Lin et al., 2007, Tinetti et al., 1994, Jensen et al., 2004, Sze et al., 2008). The general premise of falls-risk assessment is to screen for a number of different risk factors that are significantly associated with falls. However, given the wide variety of falls-risk factors, subpopulations of older adults, functional balance tasks, testing apparatus, and methods of measurement, falls-risk assessment tools are diverse and vary greatly in their strengths, weaknesses, and their predictive capabilities for falls (Rogers et al., 2003a, Perell et al., 2001, Scott et al., 2007). As such, no falls-risk assessment tool has perfect construct validity. Some of the most widely used assessment tools, which have also undergone rigorous scientific evaluation, include the Physiological Profile Assessment, the Berg Balance Scale, the Performance Oriented Mobility Assessment, the Timed Up and Go, and the Functional Reach Test. A summary of studies examining the predictive capabilities of these falls-risk tests is displayed in Table 2.2.

The Physiological Profile Assessment (PPA) is a highly refined falls-risk assessment tool which has been developed from the physiological and balance profiles of over 2,000 adults (Lord et al., 2003). The PPA has a strong predictive capability for falls as it can differentiate between multiple fallers and non-multiple fallers with 75% accuracy or greater in both institutional and community settings (Lord et al., 2003). The PPA differs philosophically from the other falls-risk assessments displayed in Table 2.2 because it does not involve any test of the ability to perform a common everyday movement. Instead, the PPA involves a series of tests of visual and peripheral sensation, leg muscle strength, reaction time, and standing and leaning balance which are each scored objectively on a continuous measurement scale (for detailed information on the individual tests see Lord et al., 2003, Lord et al., 2007). The guiding principle of the PPA is that accumulated deficits or impairments in the physiological systems of postural control will result in a reduced ability to maintain balance during everyday activities. Therefore, an advantage of the PPA is that it provides quantitative information about the potential causes of instability so that a targeted and individualised intervention strategy can be developed (Whitney et al., 2005). Once a physiological profile is formed, the test scores are weighted and combined using
Table 2.2
Summary of studies examining the predictive capabilities of a selection of well-established falls-risk assessment tools.

<table>
<thead>
<tr>
<th>Name of Instrument</th>
<th>Authors</th>
<th>Sample Size</th>
<th>Setting</th>
<th>Items</th>
<th>Range of Scores</th>
<th>Cut-off Score</th>
<th>Sensitivity</th>
<th>Specificity</th>
<th>Overall Accuracy</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Physiological Profile Assessment</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><em>Test development:</em></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lord et al. (1991a)</td>
<td>95</td>
<td>I</td>
<td>13</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>79</td>
</tr>
<tr>
<td>Lord et al. (1994a)</td>
<td>1,762</td>
<td>C</td>
<td>9</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Lord et al. (1994b)</td>
<td>414</td>
<td>C</td>
<td>11</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>75</td>
</tr>
<tr>
<td>Lord et al. (1996)</td>
<td>66</td>
<td>I</td>
<td>9</td>
<td>—</td>
<td>—</td>
<td>86</td>
<td>86</td>
<td>86</td>
<td></td>
</tr>
<tr>
<td><em>Compiled results:</em></td>
<td></td>
<td>&gt; 2,000</td>
<td>C/I</td>
<td>5-18</td>
<td>-2.0 to 4.0*</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>75</td>
</tr>
<tr>
<td><strong>Berg Balance Scale</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><em>Test development:</em></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Berg et al. (1992, 1989)</td>
<td>113</td>
<td>C/H/I</td>
<td>14</td>
<td>0 to 56†</td>
<td>49</td>
<td>77</td>
<td>86</td>
<td>—</td>
<td></td>
</tr>
<tr>
<td>Bogle Thorbahn et al. (1996)</td>
<td>66</td>
<td>C</td>
<td>14</td>
<td>0 to 56†</td>
<td>45</td>
<td>53</td>
<td>96</td>
<td>85</td>
<td></td>
</tr>
<tr>
<td>Shumway-Cook et al. (1997)</td>
<td>44</td>
<td>C</td>
<td>14</td>
<td>0 to 56†</td>
<td>49</td>
<td>77</td>
<td>86</td>
<td>82</td>
<td></td>
</tr>
<tr>
<td>Brauer et al. (2000)</td>
<td>100</td>
<td>C</td>
<td>14</td>
<td>0 to 56†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
<td></td>
</tr>
<tr>
<td>Boulgarides et al. (2003)</td>
<td>99</td>
<td>C</td>
<td>14</td>
<td>0 to 56†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
<td></td>
</tr>
<tr>
<td>Lajoie et al. (2004)</td>
<td>125</td>
<td>C/I</td>
<td>14</td>
<td>0 to 56†</td>
<td>46</td>
<td>83</td>
<td>93</td>
<td>89</td>
<td></td>
</tr>
<tr>
<td>Melzer et al. (2007)</td>
<td>100</td>
<td>C</td>
<td>14</td>
<td>0 to 56†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
<td></td>
</tr>
<tr>
<td><strong>Performance Oriented Mobility Assessment</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><em>Test development:</em></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Tinetti (1986)</td>
<td>79</td>
<td>I</td>
<td>17</td>
<td>0 to 28†</td>
<td>—</td>
<td>80</td>
<td>74</td>
<td>—</td>
<td></td>
</tr>
<tr>
<td>Tinetti et al. (1986)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><em>Subsequent studies:</em></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Raiche et al. (2000)</td>
<td>225</td>
<td>C</td>
<td>24</td>
<td>0 to 40†</td>
<td>36</td>
<td>70</td>
<td>52</td>
<td>59</td>
<td></td>
</tr>
<tr>
<td>Verghese et al. (2002)</td>
<td>60</td>
<td>C</td>
<td>9</td>
<td>0 to 16†</td>
<td>10</td>
<td>62</td>
<td>70</td>
<td>68</td>
<td></td>
</tr>
<tr>
<td>Chiu et al. (2003)</td>
<td>44</td>
<td>H</td>
<td>14</td>
<td>0 to 24†</td>
<td>17</td>
<td>96</td>
<td>96</td>
<td>96</td>
<td></td>
</tr>
<tr>
<td>Name of Instrument</td>
<td>Authors</td>
<td>Sample Size</td>
<td>Setting</td>
<td>Items</td>
<td>Range of Scores</td>
<td>Cut-off Score</td>
<td>Sensitivity</td>
<td>Specificity</td>
<td>Overall Accuracy</td>
</tr>
<tr>
<td>------------------------------------</td>
<td>-------------------------------</td>
<td>-------------</td>
<td>---------</td>
<td>-------</td>
<td>-----------------</td>
<td>---------------</td>
<td>-------------</td>
<td>-------------</td>
<td>------------------</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Murphy et al. (2003)</td>
<td>45</td>
<td>C</td>
<td>9</td>
<td>0 to 16†</td>
<td>12</td>
<td>55</td>
<td>97</td>
<td>87</td>
</tr>
<tr>
<td></td>
<td>Lin et al. (2004)</td>
<td>1,200</td>
<td>C</td>
<td>13</td>
<td>0 to 26†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td></td>
<td>Faber et al. (2006)</td>
<td>81</td>
<td>I</td>
<td>17</td>
<td>0 to 28†</td>
<td>19</td>
<td>64</td>
<td>66</td>
<td>65</td>
</tr>
<tr>
<td></td>
<td>Michel-Pellegrino et al.</td>
<td>19</td>
<td>I</td>
<td>17</td>
<td>0 to 28†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td><strong>Timed Up and Go</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Test development:</strong></td>
<td>Podsiadlo et al. (1991)</td>
<td>60</td>
<td>H</td>
<td>1</td>
<td>cont. scale*</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td><strong>Subsequent studies:</strong></td>
<td>Shumway-Cook et al. (2000)</td>
<td>30</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>14</td>
<td>87</td>
<td>87</td>
<td>87</td>
</tr>
<tr>
<td></td>
<td>Dite et al. (2002)</td>
<td>81</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>13</td>
<td>89</td>
<td>67</td>
<td>78</td>
</tr>
<tr>
<td></td>
<td>Boulgarides et al. (2003)</td>
<td>99</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td></td>
<td>Lindsay et al. (2004)</td>
<td>160</td>
<td>H</td>
<td>1</td>
<td>cont. scale*</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td></td>
<td>Thomas et al. (2005)</td>
<td>30</td>
<td>H</td>
<td>1</td>
<td>cont. scale*</td>
<td>32.6</td>
<td>75</td>
<td>67</td>
<td>73</td>
</tr>
<tr>
<td></td>
<td>Arnold et al. (2007)</td>
<td>106</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>11</td>
<td>67</td>
<td>55</td>
<td>60</td>
</tr>
<tr>
<td></td>
<td>Melzer et al. (2007)</td>
<td>100</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td></td>
<td>Thrane et al. (2007)</td>
<td>974</td>
<td>C</td>
<td>1</td>
<td>cont. scale*</td>
<td>12</td>
<td>44</td>
<td>58</td>
<td>52</td>
</tr>
<tr>
<td></td>
<td>Haines et al. (2008)</td>
<td>1,373</td>
<td>H</td>
<td>1</td>
<td>cont. scale*</td>
<td>30</td>
<td>80</td>
<td>22</td>
<td>31</td>
</tr>
<tr>
<td></td>
<td>Nordin et al. (2008)</td>
<td>183</td>
<td>I</td>
<td>1</td>
<td>cont. scale*</td>
<td>25</td>
<td>62</td>
<td>62</td>
<td>62</td>
</tr>
<tr>
<td><strong>Functional Reach Test</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Test development:</strong></td>
<td>Duncan et al. (1990, 1992b)</td>
<td>345</td>
<td>C</td>
<td>1</td>
<td>cont. scale†</td>
<td>6</td>
<td>SP</td>
<td>SP</td>
<td>SP</td>
</tr>
<tr>
<td><strong>Subsequent studies:</strong></td>
<td>Thapa et al. (1996)</td>
<td>303</td>
<td>I</td>
<td>1</td>
<td>cont. scale†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
<tr>
<td></td>
<td>Eagle et al. (1999)</td>
<td>98</td>
<td>H</td>
<td>1</td>
<td>cont. scale†</td>
<td>6</td>
<td>76%</td>
<td>34%</td>
<td>47%</td>
</tr>
<tr>
<td></td>
<td>Brauer et al. (2000)</td>
<td>100</td>
<td>C</td>
<td>1</td>
<td>cont. scale†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
</tr>
</tbody>
</table>
Table 2.2. Continued.

<table>
<thead>
<tr>
<th>Name of Instrument</th>
<th>Authors</th>
<th>Sample Size</th>
<th>Setting</th>
<th>Items</th>
<th>Setting Items</th>
<th>Range of Scores</th>
<th>Cut-off Score</th>
<th>Sensitivity</th>
<th>Specificity</th>
<th>Overall Accuracy</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Murphy et al. (2003)</td>
<td></td>
<td>45</td>
<td>C</td>
<td>1</td>
<td>cont. scale†</td>
<td>8</td>
<td>73%</td>
<td>88%</td>
<td>84%</td>
<td></td>
</tr>
<tr>
<td>Lin et al. (2004)</td>
<td></td>
<td>1,200</td>
<td>C</td>
<td>1</td>
<td>cont. scale†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
<td></td>
</tr>
<tr>
<td>Thomas et al. (2005)</td>
<td></td>
<td>30</td>
<td>H</td>
<td>1</td>
<td>cont. scale†</td>
<td>—</td>
<td>NSP</td>
<td>NSP</td>
<td>NSP</td>
<td></td>
</tr>
<tr>
<td>Haines et al. (2008)</td>
<td></td>
<td>1,373</td>
<td>H</td>
<td>1</td>
<td>cont. scale†</td>
<td>1.57</td>
<td>70%</td>
<td>43%</td>
<td>47%</td>
<td></td>
</tr>
</tbody>
</table>

*High score represents high falls-risk.
†High score represents low falls-risk.

Setting: H = hospital inpatient or falls clinic; I = Institutional aged-care; C = community-dwelling.
cont. scale = continuous measurement scale; NSP = not a significant predictor of falls; SP = Significant predictor of falls calculated with odds ratios.

Units of measurement: The Physiological Profile Assessment, Berg Balance Scale, and Performance Oriented Mobility Assessment provide a unitless score.
The Timed Up and Go is measured in seconds and the Functional Reach Test is measured in inches.

In the study of Duncan et al. (1992b), the odds of being a multiple faller was 4.02 for a functional reach score of less than or equal to six inches.
discriminant function analysis to produce a standardised falls-risk score (Lord et al., 2003, Lord et al., 2007). A falls-risk score of one or greater is associated with a 60.7% risk of multiple falls, whereas a falls-risk score of less than one is associated with an 11.6% risk of multiple falls (St George et al., 2007).

The Berg Balance Scale (BBS) is arguably the most popular and widely used falls-risk assessment tool in clinical settings. The BBS was initially developed as a simple clinical indicator of balance function in stroke patients and older adults (Berg et al., 1989, Berg et al., 1995, Berg et al., 1992). The testing procedure involves qualitative measurement of the ability to maintain balance during tasks such as sitting, standing, transfers, reaching, leaning over, turning, and stepping. Overall, the predictive capacity of the BBS is currently unclear because the results have not been consistent across studies and settings. Two retrospective studies (Lajoie and Gallagher, 2004, Shumway-Cook et al., 1997) reported similar sensitivities (77% to 83%) and specificities (86% to 93%) compared with the initial validation study (sensitivity: 77%; specificity: 86%; Berg et al., 1992). In contrast, three well conducted prospective studies in healthy older adults found that the BBS was a poor predictor of falls (Bogle Thorbahn and Newton, 1996, Boulgarides et al., 2003, Brauer et al., 2000). Conradsson and colleagues (2007) found that a change of 8 BBS points (14% of scale magnitude) was required to demonstrate a meaningful change in the balance function of institutionalised older adults. This suggests that the BBS has low sensitivity to detect small changes in balance function, which may limit its effectiveness when used on highly functional older adults with less severe balance deficits (Garland et al., 1997, Rose et al., 2006).

The Performance Oriented Mobility Assessment (POMA) also involves a qualitative assessment of balance and mobility function during basic everyday tasks. The original purpose of the POMA was to screen the falls-risk of frail older adults dwelling within nursing homes (Tinetti and Ginter, 1988, Tinetti et al., 1986), but it is also used widely in community settings (Murphy et al., 2003, Raiche et al., 2000, Verghese et al., 2002, Cho et al., 2004, Mecagni et al., 2000). Given that the POMA is divided into separate
balance and gait subscales, different versions of the POMA exist (Faber et al., 2006, Tinetti, 1986). The balance subscale includes sitting, sit-to-stand transfers, standing, external perturbation, and turning tasks. The gait subscale involves observational analysis of gait initiation and the characteristics of straight-line walking. Overall, empirical evidence indicates that the POMA is an accurate predictor of fallers and non-fallers in older adults with chronic disabilities (sensitivity: 80% to 96%; specificity: 74% to 96%; Chiu et al., 2003, Tinetti et al., 1986). The POMA has also been reported to be significantly associated with falls in frail and community-dwelling older adults, but with lower sensitivity and specificity compared to individuals with disease-related impairments (Faber et al., 2006, Raiche et al., 2000, Verghese et al., 2002, Thomas and Lane, 2005). The findings of some studies suggest that the POMA may not be a good predictor of falls in healthy and physically active older adults (Lin et al., 2004, Michel-Pellegrino et al., 2007). Baloh and colleagues (2003) found that age-related changes in visual, vestibular, somatosensory, and auditory function in 59 normal healthy older adults over 8 to 10 years were weakly associated with Tinetti balance and gait function. Taken together, the literature suggests that the POMA can predict falls resulting from severe mobility impairments, but the scale lacks the sensitivity to detect small but significant changes in postural stability in individuals without predominant balance problems.

The Timed Up and Go (TUG) test is a modified version of the Get Up and Go test reported by Mathias and colleagues (1986). The TUG is a quantitative test as it measures the total time required to stand up from a chair, walk 3 m, turn around, walk back to the chair, and sit down again (Podsiadlo and Richardson, 1991). As older adults with reduced postural stability and muscle strength are known to move slower (Callisaya et al., 2009), a longer time to complete the TUG indicates increased falls-risk. In the initial study of the TUG (Podsiadlo and Richardson, 1991), the test demonstrated high inter-rater and intra-rater reliabilities and was significantly associated with the BBS and gait speed in a group of 60 frail older adults. Subsequent reports of the sensitivities and specificities of the TUG for falls-status have been particularly inconsistent across studies. The test appears to have limited predictive
value in community settings as most studies found no significant association with falls or had low overall classification accuracy (Arnold and Faulkner, 2007, Thrane et al., 2007, Boulgarides et al., 2003, Melzer et al., 2007). Evidence suggests that the TUG is most beneficial for use on frail or unwell older adults. Although one hospital study found no significant association between the TUG and falls (Lindsay et al., 2004), two other studies reported high sensitivity (Haines et al., 2008, Thomas and Lane, 2005). Another two hospital studies found that inability to perform the TUG was predictive of falls (Large et al., 2006, Salgado et al., 1994). Therefore, the TUG may be best suited as a quick preliminary test of falls-risk in hospital settings.

In the Functional Reach Test (FRT), participants are required to reach as far forward as possible without stepping or losing their balance. The FRT is scored quantitatively as the difference between the outstretched arm length and the maximum forward reach distance attained. Duncan et al (1992b) examined the predictive validity of the FRT for falls in a 6 month prospective study of 217 older male veterans. Logistic regression analyses revealed that individuals who were unable to reach had the greatest odds of recurrent falls, followed by individuals that could not reach further than 6 inches without losing their balance. However, the majority of experimental evidence accumulated since this initial study indicates that the FRT is a poor predictor of falls (Wallmann, 2001, Cho and Kamen, 1998, Thomas and Lane, 2005, Brauer et al., 2000, Lin et al., 2004, Thapa et al., 1996). Perhaps the inconsistency of findings is due to the fact that participants are permitted to select their own reaching strategy, which is known to influence the reach distance attained (Liao and Lin, 2008). Eagle and colleagues (1999) found that the FRT had 76% sensitivity and 34% specificity for falls in 98 hospital inpatients, and that predictive accuracy was not superior to clinical judgement. Murphy and colleagues (2003) found that the FRT predicted a high proportion of community-dwelling fallers and non-fallers, but was not a significant and independent discriminator for falls status compared with other clinical balance tests. In one prospective cohort study of 1,200 older adults, it was found that fallers actually had increased reach amplitudes compared with non-fallers (Lin et al., 2004).
Taken together, these findings indicate that caution should be used when using information from the FRT to assess balance impairment (Wallmann, 2001).

In summary, many different assessments are being used to screen the falls-risk of older adults. Although the reliabilities of some of the most popular and widely used falls-risk assessments have been high-to-excellent, their predictive capabilities for falls have been inconsistent. Studies performed after the initial validation studies have found much lower and often poor accuracies of the tests to predict fallers. To some extent these disparate findings may reflect methodological and participant differences between studies, but may also reflect that most instruments are subjectively assessed on an ordinal measurement scale, prone to ceiling effects, and insensitive to changes in balance function. Most balance assessment tools appear to be more suitable to screen older adults that are frail or have disease-related impairments. Therefore, there is a current need for quantitative and unbiased falls-risk assessment tools with high sensitivity and specificity for predicting falls. The PPA papers to be a premier falls-risk screening tool as it overcomes the limitations of many other assessments and has consistently predicted falls with high accuracy across different participant settings (Lord et al., 2007, Lord et al., 2003).

2.2 Postural Control

2.2.1 Posture, orientation, and stability

A key factor contributing to falls and other mobility problems among older adults is deterioration of postural control (Fernie et al., 1982, Gu et al., 1996, Maki et al., 2000). Posture refers to the positioning of body segments in relation to other body segments, the positioning of the body with respect to the environment, or the alignment of the body with respect to gravity (Melvill Jones, 2000). The process of maintaining the appropriate relationships between the body segments, and between the body and the environment for a given task is called postural orientation (Shumway-Cook and Woollacott, 2001). A separate, yet critically important feature of an individual’s posture is the regulation of stability during movements. Postural
stability, which is often generally referred to as balance, is the ability to maintain posture in equilibrium (Shumway-Cook and Woollacott, 2001). Static postural equilibrium is the capacity to maintain the body in a stationary position, whereas dynamic postural equilibrium is the capacity to progress through an intended movement without failure (Horak et al., 1997, Johansson and Magnusson, 1991, Melvill Jones, 2000). Although postural orientation and postural stability are two distinct goals, they are both achieved via continual adjustments of posture which are implemented by the postural control system.

2.2.2 The postural control system

The postural control system represents a complex integration of the sensory, cognitive, motor, neuromuscular, and musculoskeletal systems (Horak, 2006, Shumway-Cook and Woollacott, 2001, Carr and Sheperd, 1998). These systems act cooperatively to regulate posture during voluntary movements, and are critical for preserving stability when perturbations are applied to the body (Huxham et al., 2001, Horak et al., 1997, Balasubramaniam and Wing, 2002). Postural perturbations refer to sensory and mechanical disturbances that have a negative influence on posture. Some common postural perturbations include gravity, environmental hazards, distorted or weakened sensory information (e.g., dim versus full light), mechanical forces associated with voluntary movements, and even the small forces transmitted through the body associated with respiratory and cardiac events (Horak et al., 1997). To counteract these perturbations, the postural control system uses anticipatory and compensatory postural adjustments. Anticipatory postural adjustments occur immediately prior to voluntary movements that are considered to represent a threat to stability based on prior experience (Massion, 1992). They are triggered via interoceptive processes when the decision is made to perform the voluntary action (Massion, 1992). In contrast, compensatory postural adjustments occur following externally applied disturbances to the body (Horak et al., 1997, Maki and McIlroy, 1997, Maki and McIlroy, 2006, Maki et al., 2003). External perturbations give rise to sensory feedback from the visual, vestibular, and somatosensory systems. This sensory information is weighted, integrated, and interpreted within the CNS via high
level (perceptual) processes so that an appropriate compensatory response is selected and then executed (Horak, 2006, Horak et al., 1989, Shumway-Cook and Woollacott, 2001).

Apart from the physiological components that are involved in the generation of postural responses, there are also many different tasks, environments, and intra-individual factors which may influence postural orientation and stability. Some of the main factors that are known to have an influence on posture are displayed in Figure 2.1. Based on this conceptual diagram, it is evident that a large number of factors may combine at any particular moment to influence posture. Despite the complexity of the postural control problem, a healthy individual is remarkably versatile in maintaining their orientation and stability. Two features of the postural control system that enable this versatility are its redundancy and adaptability (Carr and Sheperd, 1998, Newell, 1986). One example of this redundancy and adaptability is when the sensory information used to regulate posture suddenly changes or becomes unreliable. Research suggests that under these circumstances the CNS automatically adjusts the relative importance of sensory modalities, a process which is referred to as multisensory reweighting (Jeka et al., 2010). Sensory reweighting was examined in studies by Corna et al. (1999) and Buchanan and Horak (1999). In both studies participants were required to maintain their balance with and without their vision during stance on a sinusoidally translating platform. When vision was available during intermediate to high platform translation frequencies (0.3 to 1 Hz), participants fixed their head and trunk in space and reacted to the mechanical perturbations by allowing their legs to oscillate with the platform. When vision was withdrawn, participants produced a different postural coordination pattern that was characterised by large and slow swaying movements of the head and trunk. While the findings of both studies indicate that visual information was essential for stabilisation of the upper body in space during the platform translations, different explanations were provided for the reweighting of sensory information when visual feedback was removed. Corna and colleagues (1999) suggested that participants re-weighted their postural control in favour of the somatosensory and vestibular systems because they provided more
Figure 2.1. Conceptual diagram of factors that may influence an individual’s postural stability and postural orientation.
accurate information regarding the movements of the body induced by the translating platform. In contrast, Buchanan and Horak (1999) suggested that participants primarily relied upon proprioceptive rather than vestibular information as humans with profound loss of vestibular function are able to produce normal responses to surface translations (Horak et al., 1994).

A massive number of physiological and biomechanical elements may be recruited by the CNS to control movement. These elements are often referred to as ‘degrees of freedom’ and exist at many different levels of analysis such as the cell, tissue, system, and body segment levels. For example, sensory receptors, neurons, neural pathways, motor units, muscles, joints, and axes of joint movement represent some degrees of freedom from different levels of analysis which must be coordinated in time and space for an individual to be able to maintain their postural orientation and stability (Gielen et al., 1998). The degrees of freedom are often referred to as redundant because their total number often greatly exceeds the minimum number required to perform a given postural task (Turvey, 1990). It has been suggested that this redundancy is beneficial to humans, as it affords the postural control system the ability to overcome the novel challenges to stability inherent in the changing intra-individual, task, and environment demands (Newell, 1986). Redundancy is also incredibly beneficial for degenerative conditions such as ageing and disease where degrees of freedom are often damaged and lost (Vaillancourt and Newell, 2002). Under these circumstances, the redundancy enables older adults to generate compensatory responses that help to preserve their sensorimotor and balance capabilities (Reeves et al., 2009, Raghavan et al., 2010).

Large numbers of degrees of freedom presents an incredibly complex problem for researchers trying to understand how they are regulated during movement. Bernstein (1967) was the first to suggest that the task of coordinating movement is one of ‘mastering’ the degrees of freedom. Bernstein’s ideas were developed from a kinematic analysis of a skilled Blacksmith’s hammer swinging actions. He observed that despite variation in the movements of the individual joints, the strike-point of the hammer was relatively invariant. These results suggested that the movement
repetitions were not necessarily a linear summation of the same individual degrees of freedom, and that degrees of freedom were newly coordinated during each repetition to achieve the task goal. The question of how the motor system recruits and reduces redundant degrees of freedom during movement still motivates a growing body of contemporary postural control research. Vereijken and colleagues (1992) measured how the joint degrees of freedom were coordinated in five young men who performed slalom-like movements on a ski simulator. In the initial stages of motor learning, individuals adopted rigid postural and movement strategies as demonstrated by low variability and range of joint angles of the lower limbs and torso and high correlations between joint motions. Over seven days of practice, the angular movements of the joints significantly increased as postural movements became more actively involved in the performance of the task. These findings suggest that in the early stages of motor learning, individuals form tight couplings between the joints, thereby ‘freezing’ these degrees of freedom, which helps to eliminate redundancy and reduces the complexity of the required postural control (Bernstein, 1967). With further practice, the initial restrictions placed on the joint degrees of freedom are gradually released by the postural control system and the joint motions become coordinated in a manner that facilitates task performance (Bernstein, 1967).

Although the aforementioned study of Vereijken et al (1992) provide some insight into the coordination of degrees of freedom during postural tasks, scientists are currently dealing with the seemingly overwhelming problem of piecing together how the degrees of freedom of the human system are regulated. It is currently unfeasible to measure the activity of large numbers of individual degrees of freedom from various levels of analysis in a meaningful way. Even if it was possible to measure large numbers of postural degrees of freedom from one person or a group of people, the findings may not be generalisable to others because of individual/population differences in physiology, biomechanics, and behaviour. Scientists are limited to reductionist methods and must select only a few variables that are believed to encapsulate the activity of interest. In addition, most studies tend to measure outcome variables from only one or two levels of analysis, and rarely are attempts
made to link the activities of micro and macro degrees of freedom. As scientists are limited by reductionist methods, understanding of postural control is formed by piecing together the findings of separate studies from various laboratories. Unfortunately, the different participant samples, measurement techniques, outcome measures, and experimental designs of studies are a significant limitation of this approach and often hinder synthesis of findings.

Despite the challenges currently facing postural control researchers, the future holds promise for advancement of understanding of the mechanisms of postural control. Not only does technological advancement continue to revolutionise scientific measurement and analysis, the postural control literature is growing at an unprecedented rate and is easily distributed and accessed via online databases. Such technological advances provide the capability for the development of new and innovative experimental methodologies and greatly enhance the sharing of information between scientists. Examples of some newly published studies of postural control are presented in Table 2.3. These studies were accessed from a literature search using the online database Scopus with the following search terms identified in the titles, keywords, or abstracts of potentially relevant documents: (posture OR “postural control”) AND (balance OR stability OR orientation). This search revealed a total of 10,591 matching documents on the 31/12/2010. The 723 matching documents published in 2010 or available ahead of print were then screened for relevance and a selection of prominent studies examining postural control issues were reviewed.

The growing body of literature examining the role of cognition in postural systems is one example of how recent scientific research has revolutionised current understanding of postural control. Traditionally, postural control was viewed as an automatic process governed by neural reflexes (Gurfinkel et al., 1974, Sherrington, 1906), suggesting that postural control required minimal or no cognitive input. However, more recent investigations have found that when participants performed an attention-demanding task concurrently with a postural task, the performance of the
Table 2.3
Selection of recent peer-reviewed studies of human postural control.

<table>
<thead>
<tr>
<th>Study</th>
<th>Aims</th>
<th>Summary of Findings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lion et al. (2010)</td>
<td>The effect of exercise induced dehydration on postural sensorimotor strategies (n = 10 young sportsmen).</td>
<td>Post- versus pre-exercise, postural sway was increased during normal stance and when vestibular cues were the only reliable sensory source, suggesting that dehydration disrupted vestibular function.</td>
</tr>
<tr>
<td>Bisson et al. (2010)</td>
<td>The effect of ankle and hip muscle fatigue on postural sway and attentional demand during unipedal stance (n = 28 young adults).</td>
<td>Fatigue increased postural sway in the AP and ML directions. Reaction time did not change with fatigue, suggesting that muscle fatigue did not increase the attention necessary for balance control.</td>
</tr>
<tr>
<td>Carpenter et al. (2010)</td>
<td>To elucidate the mechanisms of standing balance control by artificially reducing postural sway (n = 47 young adults).</td>
<td>Ground reaction forces, reflecting postural control activity, were increased when body sway was artificially minimised without participant awareness, suggesting an exploratory role of postural sway.</td>
</tr>
<tr>
<td>Oude Nijhuis et al.</td>
<td>To examine whether first trial reactions to platform perturbations were similar to the acoustic startle reflex (n = 8 young adults).</td>
<td>Mechanical and acoustic stimuli elicited a clear first trial reaction with larger segmental rotations and muscle activations compared with later trials. Upper body muscle responses were similar for the mechanical and acoustic startle stimuli, suggesting a common neural origin.</td>
</tr>
<tr>
<td>Yiou et al. (2010)</td>
<td>The effect of ML fear of falling on anticipatory postural control of ML stability (n = 10 young adults).</td>
<td>For high versus low fear of falling, the ML shift in ground reaction forces was slower, suggesting that the APAs used to regulate ML stability were modified to attenuate the risk of a sideways fall.</td>
</tr>
<tr>
<td>Mochizuki et al. (2010)</td>
<td>To examine cortical activity in relation to perturbation size and predictability during balance recovery reactions (n = 12 young adults).</td>
<td>Cortical activity was scaled to perturbation size when the perturbation size was known. When perturbation size was unknown, cortical activity preceding the perturbation was heightened, presumably as a safety mechanism to preserve stability.</td>
</tr>
<tr>
<td>Study</td>
<td>Aims</td>
<td>Main Findings</td>
</tr>
<tr>
<td>------------------------</td>
<td>----------------------------------------------------------------------</td>
<td>---------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------</td>
</tr>
<tr>
<td>Stins et al. (2010)</td>
<td>To identify similarities/differences in postural control during attention distraction and increased postural anxiety (n = 18 young adults).</td>
<td>Attention distraction versus increased anxiety resulted in smaller amplitude and higher frequency of postural sway. Increased anxiety resulted in smaller time lags between postural sway and calf muscle activity, suggesting tighter anticipatory control of stability.</td>
</tr>
<tr>
<td>Sozzi et al. (2010)</td>
<td>To estimate the time interval from unexpected changes in visual information to the onset of postural control changes (n = 13 young adults).</td>
<td>With the addition or removal of vision, postural sway changed within 2 s and steady state was recovered around 3 s. Changes in muscle activity preceded sway changes irrespective of the visual shift direction.</td>
</tr>
<tr>
<td>Santos et al. (2010)</td>
<td>The contribution of APAs to compensatory postural responses during predictable and unpredictable perturbations (n = 8 young adults).</td>
<td>When perturbations were predictable, anticipatory activation was observed in the leg and trunk muscles, which reduced the amplitude of subsequent compensatory responses. Coordination between anticipatory and compensatory postural control improves stability.</td>
</tr>
<tr>
<td>Bonnet et al. (2010)</td>
<td>The effect of proximity, location, and visibility of surrounding objects on stance control (n = 12 young adults).</td>
<td>Postural sway amplitude decreased when objects were closest, but only when the object was visible, suggesting an effect of enhanced visual feedback. Participants also leaned away from the closest placed objects regardless of visibility, presumably to increase postural safety margin.</td>
</tr>
<tr>
<td>Hall et al. (2010)</td>
<td>The effect of novel and familiar postural supports on APAs and whether postural strategies were refined by practice (n = 12 young adults).</td>
<td>Muscle activity was recruited to enhance postural stability, irrespective of whether the postural support was familiar or novel. Postural strategies adapted slower for novel than familiar supports.</td>
</tr>
<tr>
<td>Reynolds et al. (2010)</td>
<td>The effect of stance width and vision on the ability to voluntarily reduce postural sway (n = 14 young adults).</td>
<td>Participants who swayed most when relaxed had the greatest capacity to voluntarily reduce sway. Increased difficulty of the standing task was associated with a reduced ability to minimise sway.</td>
</tr>
</tbody>
</table>

Abbreviations: APA = anticipatory postural adjustment; AP = anterior-posterior; ML = medial-lateral.
postural task or the cognitive task deteriorated due to competition for attentional resources (Barra et al., 2006, Müller et al., 2004, Melzer et al., 2001). A study by Bisson et al (2010) built upon previous investigations by examining whether fatigue of the ankle and hip muscles influenced the attention required to maintain postural stability during unipedal stance tasks. Auditory reaction time was used as an indicator of the attentional resources used by the postural control system. It was found that postural sway significantly increased following fatigue, however reaction time was not slower post-fatigue, suggesting that fatigue did not increase the attentional requirement necessary for maintaining postural stability. This null effect of fatigue on attentional demands was contrary to the findings of a similar study by Vuillerme, Forestier, and Nougier (2002). The different fatigue methodologies may account for this discrepancy, as fatigue in Bisson’s study was induced by concentric muscle contractions on an isokinetic dynamometer until force output dropped below 50% of maximum for three consecutive sets. In contrast, Vuillerme’s fatigue protocol may have been more severe because participants stood on their tiptoes in plantar flexion and received strong verbal encouragement until maximal exhaustion. Participants in the study of Vuillerme et al (2002) also had visual feedback removed during the standing tasks which increases the attentional requirement for maintaining balance (Teasdale et al., 1993).

Cognitive factors associated with balance control were also investigated in a novel study by Mochizuki and colleagues (2010). Using electroencephalography (EEG), they measured the cortical activities of 12 young adults before and after the release of a tether that held them in a forward leaning position. Participants recovered their balance using an ankle rotation strategy for small perturbations and a compensatory step for large perturbations. To manipulate perturbation predictability, the large and small perturbation sizes were presented sequentially (predictable condition) or randomly (unpredictable condition). It was found that pre- and post-perturbation cortical activities were independently generated, and that pre-perturbation activity was scaled to the size of predictable perturbations but was significantly increased for unpredictable perturbations. The authors suggested that participants compensated
for the unpredictability by increasing their level of cortical activity to prepare the postural system for the worst case scenario (i.e., the larger perturbation). The findings of the study were also consistent with the view that pre-perturbation cortical activity reflected the preparedness of the postural system based on environment context and prior experience, whereas post-perturbation cortical activity reflected an adjustment of the initial balance recovery strategy once the precise characteristics of the perturbation were known (e.g., timing, size, and direction). A complicating factor associated with studies examining brain-related activity is that some brain regions can have an inhibitory effect upon other areas. Therefore, significant increases in generalised measures of brain activity may not necessarily reflect greater stimulation of the motor apparatus. Apart from this limitation, the study was experimentally robust as special care was taken to limit the influence of potentially biasing factors on the results such as foot positioning, load on the tether prior to release, and anticipatory responses by participants.

Fear of falling is another type of cognitive factor that can influence balance control. In the study of You and colleagues (2010), the influence of fear of falling on anticipatory postural control was investigated. Ten young adults were required to perform rapid leg flexion movements in response to an auditory cue whilst standing with their foot aligned laterally with the edge of a heightened platform. Previous studies have demonstrated that immediately prior to rapid voluntary leg flexion actions, the postural control system performs an anticipatory shift of body weight towards the stance limb to ensure that the voluntary action does not lead to a loss of balance (Mouchino et al., 1990, Rogers and Pai, 1990). These anticipatory weight shifts are referred to as anticipatory postural adjustments (APA). It was the primary interest of You and colleagues (2010) to determine how APAs were influenced by the increased postural threat when the APA required a shift of body weight towards the edge of the platform. The platform was raised to heights of 6 or 66 cm off the ground to induce low and high fear of falling respectively, which were confirmed by fear of falling and coping efficacy questionnaires. It was found that the ML forces generated during APAs were reduced in amplitude but longer in duration for high compared with the low
surface height. These findings suggest that with increased fear of falling, participants executed a more cautious APA prior to foot liftoff which attenuated the risk of a ML fall off the platform edge. Given that many older adults experience fear of falling (Arfken et al., 1994) and ML instability (Rogers and Mille, 2003), it would be interesting to determine the nature of older adults’ postural responses under these experimental conditions. Future studies should also reduce the potential for undesirable preliminary movements, such as shifting a proportion of body weight onto the stance limb prior to the signal to move. One method to counteract this unwanted anticipation would be to use a choice reaction time task so that participants have to move with either the left or right leg.

The adaptability of APAs with respect to environmental conditions was examined by Hall and colleagues (2010). The aim of this study was to determine if the postural control system would use multiple external supports to stabilise the body in anticipation of a perturbation, and whether the use of external supports was dependent upon familiarity. The authors hypothesised that any muscle with the potential to contribute to postural stability, irrespective of its familiarity with providing whole-body postural support, would be recruited in advance of the voluntary action. The four different conditions of external support were unsupported, bilateral handgrip, bite plate, and combined handgrip and bite plate. Testing involved the collection of ground reaction forces, external support reaction forces, and electromyographic (EMG) activity of the limb, trunk, and jaw muscles from 12 healthy young adults while they performed randomised left and right rapid leg lift trials in response to a visual cue. The findings confirmed the authors’ hypothesis as the APAs were adapted to incorporate multiple external supports irrespective of support familiarity. The use of multiple supports also attenuated the normal anticipatory shift in forces between the legs prior to liftoff, and the EMG data demonstrated that muscle activity only increased for the conditions that the muscles could contribute to postural stability. The handgrip and bite strategies were also scaled differently during the single and multiple support conditions, suggesting that the CNS adapted all aspects of the multisegment responses and did not simply add together the single
support strategies during the multiple support condition. Overall, this study emphasises the flexibility/adaptability of the CNS to organise postural strategies to meet the demands of postural stability during both familiar and novel situations.

Anticipatory and compensatory responses are often used interactively by the CNS when responding to perturbations. For example, when an APA is insufficient to maintain equilibrium it may be necessary to further adjust posture with a compensatory response after the effect of the perturbation is accurately known (for a review of compensatory responses see section 2.4.2). However, anticipatory and compensatory motor strategies have been typically examined separately and therefore the nature of any interaction between them requires more research. Santos et al (2010) investigated the relationships between anticipatory and compensatory postural responses by delivering anterior-directed perturbations to the shoulder level of eight participants when the perturbations were predictable (eyes open) or unpredictable (eyes closed). Auditory cues associated with the perturbation were masked by requiring participants to listen to music through earphones. Participants received the perturbation impacts on their hands with their arms extended whilst the EMG activity of the leg and trunk muscles were recorded. The findings revealed that during unpredictable perturbations, no anticipatory muscle activity was observed, which was followed by large compensatory activity after perturbation impact. When perturbations were predictable, strong anticipatory action was seen in all recorded muscles which subsequently reduced the level of compensatory correction required to maintain stability. These findings confirm the link between anticipatory and compensatory strategies and suggest that equilibrium maintenance may be enhanced in individuals with balance impairments by improving their anticipatory responses prior to mechanical perturbations. This study also had a number of limitations, one of which conceded by the authors was the small sample size. In addition, as the outstretched arms were the first point of contact with the perturbation, an analysis of arm and shoulder muscle responses would have provided a more complete description of the motor strategies employed. Vision is also known to have a strong influence on postural control, and therefore future studies should keep visual
feedback identical between the predictable and unpredictable conditions. One suitable method would be to have the perturbation striking participants posteriorly rather than anteriorly, with a visual cue to signal the oncoming perturbation in the predictable condition only.

Carpenter and colleagues (2010) performed a well designed and carefully executed study to elucidate biomechanical and neural mechanisms of upright stance. Using a novel apparatus, the investigators were able to experimentally separate the sway motions of the body from neural responses to the sway (for a review of postural sway see section 2.2.3). Participants were required to stand upright with their back firmly braced against a rigid board with adjustable straps fastened around the head, shoulders, chest, hips, and lower legs. The board was linked into a closed-loop pulley system that allowed participants to experience normal postural sway oscillations about the ankle joint similar to an inverted pendulum. During 75 s periods interspersed within 6 minute trials, the experimenters discretely applied a brake to the pulley-system which locked the board (and thus sway) in place in the sagittal plane. Special care was taken to keep the participants naive to the locking mechanism by eliminating auditory cues via headphones and eliminating visual cues via blinders. The 47 young adults recruited for the study and were randomised into experimental and control groups. The control group performed the standing tasks with and without attachment to the apparatus so that the effect of the apparatus itself on postural sway could be determined. The results revealed that the apparatus was highly effective in minimising postural sway and that participants were unaware of any manipulation of postural sway during the experiment. The main finding of the study was that the ground reaction forces were generated independently of body sway. That is, when sway was experimentally minimised, the ground reaction forces in anterior-posterior (AP) and medial-lateral (ML) directions were significantly increased by the postural control system. A traditional view is that ground reaction forces during upright stance are synchronised with the amount of body sway because such forces are necessary for correcting the sway. However, the results of this study challenge traditional views of standing balance control and suggest that postural sway might
have an exploratory role to ensure that the postural control system receives continually updated information from multiple sensory inputs (Patla et al., 1990a).

Recent studies have demonstrated that with increased conscious effort, individuals were able to reduce the amplitude of their standing postural sway (Mitra and Fraizer, 2004, Loram et al., 2001). A study by Reynolds (2010) built upon this previous work by examining whether the capacity to voluntarily reduce sway was dependent upon the difficulty of the standing task. Fourteen young adults were required to stand either relaxed or completely still during 1 minute trials. For the relaxed condition, participants were instructed to stand completely relaxed and let their thoughts wander freely, but not to deliberately produce increased sway. For the still condition, participants were instructed to concentrate fully on their body movement and to minimise their sway as far as possible. The sway was recorded by measures of trunk motion and cocontraction of the tibialis anterior and medial gastrocnemius muscles. Task difficulty was increased by removing vision and by reducing stance width for the following foot positions: 15 cm apart (wide), together with medial malleoli touching (narrow), and right foot immediately in front of the left foot (tandem). Still compared with relaxed sway resulted in significantly reduced trunk sway velocity, which was mediated via an attenuation of peak trunk frequency. With increased difficulty of the standing task, there was a corresponding reduction in the ability of participants to reduce their sway velocity. Increased cocontraction of the leg muscles was significantly associated with reduction of sway, however, these associations were not observed in individual participant data, suggesting that cocontraction was not a successful strategy for voluntarily reducing sway. As the reduced stance width was designed to challenge ML stability, the findings of this study may have been more insightful if the changes in postural sway were examined separately for the AP and ML directions. With reduced stance width, participants may have still been able to voluntarily reduce their AP sway despite a reduced capacity to voluntarily reduce ML sway. Overall, the findings of this study raise important theoretical questions similar to Carpenter et al (2010) about the functional role of postural sway. Given that
postural sway can be voluntarily reduced, the findings suggest that sway reduction may not be a primary goal of the postural control system.

Assessment of postural responses to sudden changes in sensory information can provide insights into the sensory reweighting mechanisms of postural control, and can also help to predict the transient sensory situations in which older adults are susceptible to falls. However, most postural control investigations measure steady-state postural responses with the different sensory conditions examined in separate trials. Such an approach may ignore potentially important features of postural behaviour linked to transient sensory events. Sozzi and colleagues (2010) performed a study to quantify the response time and the nature of adaption in stance control following sudden shifts in visual information using electronically controlled shutter spectacles. Ground reaction forces and leg muscle EMG were acquired from 13 young adults during 400 s standing trials. The results showed that transitions from full vision to occluded vision resulted in significant increases in ground reaction force variability and shifts in the average standing position, and vice versa for transitions from occluded vision to full vision. These changes in postural activity generally occurred within 2 s. Interestingly, the recovery time of the ground reaction forces, and thus postural activity, was significantly shorter when vision was removed compared to when vision was added. On the basis of these findings, the authors suggested that when vision was suddenly added, more time was required to reset the sensorimotor integration process to regain control of postural sway. It is likely that the increased sensory redundancy associated with the sudden availability of vision also required a more complex computation by the CNS to compare the sensory inputs and reweight their contributions in favour of the most stable reference frame. These results may not be readily generalised to everyday situations involving sudden changes in visual illumination (e.g., shifts between well-lit and dim environments) because the shutter spectacles ensured that the eyes remained completely illuminated when the lenses were closed. Overall, the findings of this study provide a new and interesting perspective on the coupling between vision, postural reference, muscle activity, and the response time of the sensory weighting and integration process. Given that this
study has established the normal time course required for visual sensory reweighting, the findings may be useful for indentifying sensory integration deficits in older adults and in persons with diseases affecting postural control.

In summary, a review of the current literature revealed that postural control is influenced by a variety of individual, task, and environment factors. Complex interactions between these factors are met by a complex postural control system consisting of massive physiological and biomechanical degrees of freedom. These degrees of freedom enable the postural control system the redundancy and adaptability it needs to continually meet the demands of orientation and stability in both familiar and novel situations, but also present an incredibly difficult challenge to researchers attempting to understand the operational mechanisms of control. Emerging studies of postural control continue to provide new and interesting insights into all aspects of postural control, particularly the role of sensory, cognitive, and neuromuscular systems, sensory integration and weighting processes, anticipatory and compensatory responses to postural perturbations, and the regulation of postural sway oscillations. Despite the potential usefulness of voluntary postural sway tasks for examining issues related to the degrees of freedom problem, the role of physiological systems of postural control, and the assessment of falls-risk, very few newly published voluntary postural sway studies were identified (Freitas et al., 2010, Nolan et al., 2010, Lázaro et al., 2010).

2.2.3 Biomechanics of standing postural control

Biomechanical analysis is a quantitative approach of examining how posture is adjusted to facilitate orientation and stability (Winter, 1995, Hayes, 1982). The biomechanics of postural control are often described with particular reference to the relationship between the centre of mass (COM) and the base of support area. The COM is the central point of the total body mass, and it represents the weighted average of the COM of each body segment in 3D space (Winter, 1995). The COM is located approximately 55% of standing height above the ground (Winter et al., 1990a), and the vertical projection of the COM onto the ground is called the centre of
gravity (COG). To stand upright, an individual’s COG must be located within the base of support area delineated by the feet (Shumway-Cook and Woollacott, 2001, Winter, 1995). Therefore, the horizontal position of the COM is tightly regulated by the postural control system during standing to ensure that it does not exceed the base of support limits.

When an individual stands still there are small displacements of the COM in the AP and ML directions. These small oscillations of the COM are referred to as postural sway and are believed to reflect the regulation of standing stability by the postural control system (Carr and Sheperd, 1998, Winter, 1995). The control of postural sway is a dynamic process that depends on both passive elastic musculotendinous forces and actively generated muscle forces (Balasubramaniam and Wing, 2002). These forces produce torques about the ankle and hip joints, which result in corrective movements of the body towards the neutral, most stable position. As a result of these motor activities, a change occurs in the properties of the ground reaction force, most notably in the movement of the centre of pressure (COP). The COP is defined as the point location of the vertical ground reaction force. It represents a weighted average of all the pressures over the surface area of the feet in contact with the ground (Winter, 1995). For example, during stance on a single leg the COP lies within the area of the supporting foot, and during bipedal stance the COP is located somewhere between the feet depending on the relative pressure exerted by each foot. The relationships between a person’s COM, COG, and COP during standing are shown in Figure 2.2.

The relationships between the COP and COM during postural sway are often modelled as an inverted pendulum. In this model, the sway of the body in the sagittal plane occurs about the ankle joints similar to an inverted pendulum, with no rotation occurring at the knee or hip joints (Karlsson and Lanshammar, 1997, Peterka, 2002). In contrast, postural sway in the frontal plane is often modelled as a parallelogram inverted pendulum with motion occurring primarily about the hip joints and a smaller amount of motion occurring at the ankle joints (Rietdyk et al., 1999a). Postural sway in the AP direction is corrected by the ankle dorsiflexors and plantar flexors, whereas
postural sway in the ML direction is corrected primarily by the hip abductors-adductors with a smaller contribution from the ankle inverters-everters (Day et al., 1993, Kapteyn, 1973). In experiments of quiet standing, Winter and colleagues (1996, 1993) observed that AP and ML postural sway were corrected separately and independently with very little overlap. This finding was taken as evidence that corrective postural responses for the AP and ML directions of postural sway were separately and independently regulated by the postural control system. A sagittal
plane inverted pendulum model showing postural sway one complete oscillation of the body divided into five stages is presented in Figure 2.3a. At stage one, the individual's COG (W) is ahead of their COP (R), which produces an imbalance of torques about the ankle joint and results in forward postural sway. The postural control system detects the forward sway and corrects it by increasing plantar flexor activity. As shown at stage two, this plantar flexor activity shifts the COP further forward with respect to the COG, thereby producing a resultant anticlockwise torque. This anticlockwise torque slows and eventually reverses the sway direction so that the body begins swaying backward at stage three. To correct the backward sway, the individual reduces plantar flexor and/or increases dorsiflexor activity. As shown at stage four, this shifts the COP posterior with respect to the COG, and the resultant clockwise torque corrects the backward sway at stage five.

Taken together, it is evident that the COP closely tracks the COM during postural sway. When the COM sways away from a position of stability, a corrective torque is implemented such that the position of the COP overshoots with respect to the COM and restores it to a position of stability (Gage et al., 2004, Winter, 1995). Figure 2.3b shows COP and COM data in the AP and ML directions, respectively, for a young adult during 30 s of quiet stance. The sequence of events depicted in Figure 2.3a are repeated multiple times, and as predicted by the inverted pendulum model, the COP and COM are very tightly coupled, the COP has a greater dynamic range compared with the COM, and overshoots of the COP are associated with a reversal of the COM direction. Therefore, movements of the COP with respect to the COM reflect the activity of the postural control system for limiting the amplitude of postural sway. It is important to note, however, that the postural control system does not deliberately shift the COP to regulate the displacement of the COM during quiet stance. The COP represents a convenience measure to assess the overall performance of the postural control system to regulate the trajectory of the COM.
Figure 2.3. (a) Sagittal plane inverted pendulum model of a person swaying in the AP direction about the ankle joint at five different stages of oscillation (Reprinted, with permission, from Winter, 1995). The vertical ground reaction force (R) and weight force (W) vectors represent the positions of the COP and COM that are p and g distances, respectively, from the ankle joint. The associated angular accelerations (α) and angular velocities (ω) of the body are shown. Although the biomechanical model of postural sway is different for postural sway in the ML direction, the same principles apply to movements of the COP and COM. (b) COP and COM amplitudes in the AP and ML directions during 30 s of quiet stance for a young male aged 23 years.
2.2.4 Measurement of postural stability

As the manner in which posture is regulated depends on the interaction between individual, task, and environment factors (Horak, 2006, Shumway-Cook and Woollacott, 2001, Newell, 1986), a major difficulty in the development of a valid measure of postural stability is that the stability requirements for one set of conditions may not generalise to other sets of conditions (Slobounov and Newell, 1994, Huxham et al., 2001). Consequently, there is no standard measurement of postural stability, which has resulted in a variety of different measurement techniques and outcome measures reported in the literature (Johansson and Magnusson, 1991, Browne and O'Hare, 2001). New measures of postural stability are continually emerging in the literature, however it is unlikely that a single measure could ever encompass all facets of postural control and could be used to prevent falls completely (Horak, 2006, Lord et al., 2007).

A common method of estimating an individual’s level of standing stability has been to measure the amplitude of postural sway oscillations during quiet stance (Winter, 1995). Such assessments have been called many different terms including posturometry, posturography, stabilometry, and stabilography (Prieto et al., 1993). In a typical assessment, the individual is required to stand as still as possible for approximately 30 s whilst the postural sway oscillations of the body are recorded. Some studies have directly measured the sway movements of the body itself (Accornero et al., 1997, Moe-Nilssen and Helbostad, 2002, Lord et al., 2003, Day et al., 1993), however most studies have measured the COP using a force plate (Shumway-Cook and Woollacott, 2001, Winter, 1995, Carr and Sheperd, 1998, Patla et al., 1990a, Browne and O'Hare, 2001). Based on inverted pendulum models of quiet stance, a less efficient postural control system would result in greater amplitude of postural sway and thus greater amplitudes of COP displacement (Winter, 1995). Therefore, a predominant assumption of standing stability assessments is that small amounts of COP motion indicate optimal balance and large amounts of COP motion indicate poor balance (Shumway-Cook and Woollacott, 2001). However, there is evidence to suggest that COP displacement is not always a valid indicator of postural stability. For
example, Parkinson’s disease patients may exhibit normal or even reduced levels of standing COP amplitude compared with aged-matched controls (Schieppati et al., 1994, Horak et al., 1992, Beckley et al., 1993). The reduced sway is likely to be due to increased coactivation associated with the disease that results in a rigid posture (Horak et al., 1992, Romero and Stelmach, 2003). Although these effects may facilitate static postural stability, they can be particularly detrimental to dynamic postural stability and the ability to respond to external postural perturbations (Allum et al., 2002). Therefore, a reduced capacity to recover from external perturbations is regarded as a defining feature of an unstable postural system (Hayes, 1982, Patla et al., 1990a, Johansson and Magnusson, 1991). In addition, the COM is not confined to the base of support during more dynamic ambulatory movements such as sit-to-stand transfers, stepping, and locomotion. Dynamic postural stability is achieved under these conditions through the maintenance of an appropriate spatiotemporal relationship between the COP, COM, and the moving base of support (Winter, 1995, Patton et al., 1999, MacKinnon and Winter, 1993, Pai et al., 1994). Therefore, standing COP amplitude may not necessarily be an accurate representation of an individual’s static or dynamic postural stability.

The limitation of an exclusive focus on COP motion to measure postural stability may be overcome by taking into account the relationships between the base of support dimensions and/or the movements of the COM. An approach of quantifying postural stability that was employed in this thesis was calculation of the horizontal separation distance between the COP and COM (COP-COM). This measure was described by Winter (1995) based on an inverted pendulum model of postural sway and has since been used to characterise postural stability during standing, walking, obstacle crossing, and stair negotiation tasks. The distance of COP-COM separation is proportional to the horizontal acceleration of the COM (Gage et al., 2004, Winter, 1995, Masani et al., 2007). Therefore, the measure provides an indication of the postural control system’s capacity to regulate the accelerations of the COM. Consistent with this prediction, deficits in postural control associated with ageing (Yu et al., 2008, Berger et al., 2005a, Masani et al., 2007), falls-history (Berger et al.,
Altered COP-COM separation has also been linked with reduced dynamic postural stability under more challenging task conditions. For level walking and obstacle crossing, stroke patients (Lee and Chou, 2006), traumatic brain injury patients (Chou et al., 2004), and older adults with balance impairment (Said et al., 2008) exhibit greater COP-COM separation in the AP and/or ML directions compared with matched-controls. One study also found that older adults had decreased COP-COM separation during walking and obstacle crossing compared with younger adults (Hahn and Chou, 2004). This result was taken to reflect that the older adults adopted a more conservative movement strategy to facilitate their gait stability compared with the young.

When the requirement of a postural task is to rapidly generate or arrest body momentum (e.g., gait initiation or termination, balance recovery by stepping), increased COP-COM separation is associated with better postural stability. For example, older adults with neurological impairments affecting postural control have a reduced capacity to initiate gait as indicated by their reduced COP-COM separation in the target AP direction of movement (Martin et al., 2002, Chang and Krebs, 1999). Similarly, superior balance control during a sudden translation of the standing surface is associated with increased COP-COM separation during the postural response (Horak et al., 2005, Horak et al., 1996). Increased COP-COM separation is desirable because it reflects a greater capacity to decelerate the movement of the COM towards the base of support limits resulting from the translation. If the COM position closely approaches the maximum COP position (i.e., small COP-COM separation), the individual is likely to stumble or fall (Yang et al., 1990). Taken together, a growing body of literature indicates that COP-COM separation is a valid indicator of postural stability. The association between increased or decreased COP-COM separation and improved postural stability is task dependent.
2.3 Postural Control Physiology

The physiological tasks associated with avoiding a fall can be viewed as a three stage process (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). This three stage process is presented in Figure 2.4. The first stage is called stimulus detection and involves detecting an impending hazard within the environment (Patla, 1997), detecting an external perturbation from its direct biomechanical affect upon the body (Wolfson et al., 1986), or detecting an anticipated perturbation based on a planned voluntary movement (Hall et al., 2010). Stimuli are detected using the visual, vestibular, somatosensory, and interoceptive systems (Mergner et al., 2003). Studies have revealed that ageing and falls are associated with a reduced capability to detect stimuli because of deterioration in the visual, vestibular, and somatosensory systems. For instance, in a large sample of 341 community-dwelling older women aged 65 to 99 years, it was found that multiple falls status was significantly associated with poor visual acuity and visual contrast sensitivity and reduced vibration sense and leg proprioception (Lord et al., 1991a). The second stage in the process of avoiding a fall is response selection, which involves the transmission of detected stimuli along neural pathways to the CNS. Information contained within the stimuli, such as the magnitude and direction of the perturbation, are then processed centrally so that an appropriate motor response is selected (Horak and Nashner, 1986, Runge et al., 1998). These sensory integration and response selection tasks require cognitive resources (Redfern et al., 2001), and therefore reaction time measures of CNS processing are useful indicators of performance of this stage (Corriveau et al., 2004b). In the study of Müller and colleagues (2004), participants were required to react by voluntarily depressing a handheld microswitch when the cue to react was delivered at the same time as a destabilising forwards or backwards translation of the standing surface. It was found that reaction time was significantly increased during platform translation compared with the baseline standing condition, presumably because the tasks competed for central processing resources associated with the response selection stage. The third stage in the process of avoiding a fall is response execution, which involves execution of the selected motor response quickly and effectively so that the individual maintains their balance. The physiological systems involved in the response execution phase
Figure 2.4. Overview of the major events that underlie a fall due to a perturbation such as an environmental hazard. Perturbations give rise to sensory information which may be detected in anticipation of its destabilising effect, or as a result of its direct mechanical effect upon the body. The sensory information is detected by visual, vestibular, somatosensory, auditory, or interoceptive systems, and is processed by the CNS so that an appropriate motor response is selected (e.g., voluntary or automatic). Features of the response execution such as reaction time dictate the response efficacy and whether a fall is avoided. Reprinted, with permission, from Grabiner and Enoka (1995).
include neuromuscular control and muscular strength, which have also been found to deteriorate with ageing and in older adults who fall (Wolfson et al., 1995, Woollacott et al., 1986).

The three stage process of maintaining postural stability and avoiding falls is based on an information processing framework (Redfern et al., 2001, Müller et al., 2004). As such, the sequential nature of the three stage process suggests that deficits in any one component can be detrimental to the ability to avoid falls, and that success or failure of the overall response depends on the functional capabilities of all components (Grabiner and Enoka, 1995). Although a relatively healthy individual may be able to compensate for impairment to a single component, frail older adults with impairments to multiple components may only achieve partial compensation, which would increase their susceptibility to falls (Lord et al., 2007). As the normal ageing process results in degenerative changes in the sensory, cognitive, motor, neuromuscular, and musculoskeletal systems of postural control, the abilities of older adults to detect and anticipate perturbations, select appropriate responses, and to execute responses effectively within a critical time period is often compromised (Romero and Stelmach, 2003, Grabiner and Enoka, 1995, Stelmach and Worringham, 1985, Horak et al., 1989, Woollacott, 1993). The specific physiological systems of postural control which undergo age-related deterioration and are associated with falls include the visual, vestibular, and somatosensory senses, muscle strength, and reaction time (Lord et al., 2007, Lord and Sturnieks, 2005, Lord et al., 2003, Sturnieks et al., 2008b). The age and falls-status associated changes in these physiological systems are reviewed in the following sections.

2.3.1 Visual system

Vision provides continually updated information regarding the physical structure of the environment, movement of the body and external objects, and the spatial positioning of the body and its segments (Shumway-Cook and Woollacott, 2001). This visual information is used by the postural control system to regulate orientation and stability through both feed-forward and feedback mechanisms. Ageing results in
deterioration of the biological composition of the eye from the age of 40-50 years onwards (Birren and Williams, 1982). Consequently, older adults often have visual deficits for directing light onto the retina (Weale, 1963). Ocular diseases such as cataract, macular degeneration, glaucoma, and diabetic retinopathy are also relatively common among older people (Sturnieks et al., 2008b). These age- and disease-related changes have adverse effects on many aspects of visual function such as acuity, contrast sensitivity, depth perception, glare susceptibility, accommodation, and dark adaptation (Pitts, 1982). Older adults often have a reduced ability to judge distances and to perceive the spatial relationships between objects (Lord, 2006, Lord and Dayhew, 2001, Owen, 1985). In addition, older people also have a reduced capacity to detect low contrast hazards such as pavement cracks, raised edges, street gutters, doorway ledges, steps, and floor coverings (Lord, 2006, Lord and Dayhew, 2001, Owen, 1985). Therefore, deterioration of the visual system places older adults at an increased risk of obstacle contact and tripping during navigation through hazardous environments. Research also supports a strong link between visual dysfunction and falls. Large-scale studies have found significant associations between multiple falls by older adults and impaired visual acuity (Lord and Dayhew, 2001, Lord et al., 1994b, Ivers et al., 1998), contrast sensitivity (Lord and Clark, 1996, Lord et al., 1991a, Lord et al., 1994b, Ivers et al., 1998), and depth perception (Lord and Dayhew, 2001).

2.3.2 Vestibular system

The vestibular system provides the CNS with information regarding the position and motion of the head. This information is used to trigger neural reflexes which stabilise the head, eyes, and lower limbs during voluntary movements (Hain et al., 2000). The vestibular system also has important roles in sensing and perceiving self-motion, vertically orienting the body over the base of support, and maintaining standing stability through the recruitment of hip motion (Horak and Shupert, 2000). Age-related changes in vestibular anatomy include a loss of sensory hair cells from the otoliths and semi-circular canals (Rauch et al., 2001, Rosenhall, 1973), and loss of neurons from the vestibular nerve (Park et al., 2001, Bergstrom, 1973) and vestibular nuclei (Alvarez et al., 1998, Lopez et al., 1997). Declines in sensory receptors and
neurons with ageing have been linked to reduced excitability of the vestibular system and abnormal vestibular-ocular reflexes (Balo et al., 2001, Peterka et al., 1990). An epidemiological study involving 5,086 people aged 40 years and older found that 35% of individuals had some form of vestibular dysfunction (Agrawal et al., 2009). Despite these degenerative age-related changes in vestibular function, it is currently unknown whether they predispose older people to falls especially since vestibular dysfunction does not necessarily result in symptoms of dizziness or disequilibrium (Matheson et al., 1999, Baloh et al., 2001). In older adults without clinical symptoms of vestibulopathy, some studies have found a significant relationship between vestibular function and falls (Jacobson et al., 2008, Murray et al., 2005, Kristinsdottir et al., 2001b), whereas other studies have found no significant relationship (Balo et al., 2001, Lord et al., 1994b). Perhaps the discrepancy of findings is due to the poor sensitivity of some tests in detecting subtle yet important changes in vestibular function (Lord et al., 2007). Individuals with chronic vestibular deficits experience vertigo, nystagmus, ataxic gait, lateral instability (Lord and Sturieks, 2005), and are often recurrent fallers (Balo et al., 2001, Gazzola et al., 2006). Therefore, the general consensus is that degenerative age-related changes in the vestibular system contribute to falls, especially under environmental conditions where the visual and somatosensory systems cannot be used compensate for vestibular dysfunction (Horak and Shupert, 2000, Lord and Sturieks, 2005, Sturieks et al., 2008b).

2.3.3 Somatosensory system

The somatosensory system consists of muscle and joint receptors that contribute to sense of joint position and movement (proprioception), and also cutaneous mechanoreceptors which provide feedback to the CNS regarding contact forces that are applied to the body (Goble et al., 2009, Shaffer and Harrison, 2007). Somatosensory information has been consistently identified as the most important source of sensory information for postural control compared with the visual and vestibular systems (Colledge et al., 1994, Lord and Ward, 1994, Fitzpatrick and McCloskey, 1994, Lord et al., 1991b, Baci and Colebatch, 2005, Kristinsdottir et al., 2001a, Peterka, 2002, Benjuya et al., 2004). Cross-sectional studies of somatosensory
anatomy have revealed that ageing results in diminished numbers and altered morphology of muscle proprioceptors (Liu et al., 2005, Swash and Fox, 1972), joint receptors (Aydog et al., 2006, Morisawa, 1998), and cutaneous receptors (Bolton et al., 1966, Cauna and Mannan, 1958, Iwasaki et al., 2003). These structural age-related changes are associated with significant reductions of tactile sensitivity (Lord and Ward, 1994, Bruce, 1980, Perry, 2006), vibration sensitivity (Lord and Ward, 1994, Inglis et al., 2002, Verrillo et al., 2002, Perry, 2006), two-point discrimination (Stevens et al., 2003, Stevens and Choo, 1996), and proprioception (Lord and Ward, 1994, Hurley et al., 1998, Tsang and Hui-Chan, 2004, You, 2005, Verschueren et al., 2002, Madhavan and Shields, 2005). As a result, older adults often have reduced sense of segmental orientation, which may lead to movement errors when using tactile feedback to guide movement (Goble et al., 2009, Shaffer and Harrison, 2007). Deterioration of lower limb sensation in older adults has also been linked to reduced postural stability during standing tasks (Madhavan and Shields, 2005, Tsang and Hui-Chan, 2004, Lord et al., 1991b, Hurley et al., 1998, Duncan et al., 1992a, Brocklehurst et al., 1982, Choy et al., 2008), and has been identified as an important marker of falls-risk (Lord et al., 1991a, Lord et al., 1992, Lord et al., 1994a, Lord et al., 1994b, Lord et al., 1999). More severe declines in somatosensory function are observed in older individuals with peripheral neuropathy disorders (Simoneau et al., 1995). These disorders have been reported to increase falls-risk by as much as 20-fold compared with age- and sex-matched controls (Richardson and Hurvitz, 1995, Richardson et al., 1992).

2.3.4 Muscle strength

The maximum force produced by a muscle, which is known as muscle strength, is an important determinant of balance, gait function, physical performance, and independent mobility in older adults (Rogers et al., 2003a). A well-documented consequence of the normal ageing process is the decline in muscle mass, muscle quality, and muscle strength in older age, a condition which is referred to as sarcopenia (Deschenes, 2004, Rolland et al., 2008, Kamel, 2003). Cross-sectional studies have found that general muscle strength reaches its peak between 20 to 30
years, and is well preserved until approximately 50 years (Larsson et al., 1979). Muscle
strength then declines at approximately 12% to 15% per decade after age 60 (Kamel,
2003, Deschenes, 2004, Vandervoort, 2002). These rates of muscle strength reduction
with ageing are conservative estimates as considerably higher rates of loss have been
found in longitudinal studies (Aniansson et al., 1992, Frontera et al., 2000, Hughes et
cross-sectional area and strength in sedentary older men and found that leg muscle
strength decreased by 2.5% per annum and that sarcopenia accounted for 90% of the
age-related decline in strength after 12 years follow-up.

Population-based studies have found that sarcopenia is significantly associated with
self-reported physical disability (Baumgartner et al., 1998, Melton et al., 2000) and
reduced physical performance among older adults (Visser et al., 2002). Older people
with leg muscle weakness tend to walk slower (Wolfson et al., 1995, Lord et al., 1996,
Callisaya et al., 2009, Bendall et al., 1989, Bassey et al., 1992), exhibit reduced
standing stability (Lord et al., 1991b, Lord and Menz, 2000, Lord et al., 1999), and
experience difficulty in performing simple functional movements such as rising from a
chair (Brown et al., 1995, Lord et al., 2002b, Skelton et al., 1994) and negotiating stairs
(Tiedemann et al., 2007, Bassey et al., 1992). Muscle weakness of the lower limbs is
also considered to be a very strong indicator of falls-risk (Lord et al., 2007, Luukinen
strength of the quadriceps is often used as a simple indicator of overall muscle strength
(Lord et al., 2003). Lord and colleagues (1999, 1992, 1996, 1994a, 1994b) have
consistently found that fallers have reduced quadriceps strength compared with non-
fallers in community, institutional, and hospital settings. It has also been reported that
for older adults living in nursing homes, fallers have ankle, knee (Whipple et al., 1987,
Studenski et al., 1991), and hip weakness (Robbins et al., 1989) compared with non-
fallers. In one nursing home, fallers had approximately 48% to 61% less knee strength
and 90% less ankle dorsiflexion strength compared with non-fallers (Wolfson et al.,
1995). Taken together, these findings are consistent with the view that muscle
weakness is one of the strongest determinants of falls-risk (Rubenstein, 2006).
2.3.5 Reaction time

Motor responses of the postural control system often need to be initiated and executed in a rapid and timely manner to maintain postural stability. Therefore, a slowing of motor responses following a perturbation increases the susceptibility of older persons to falls (Grabiner and Jahnigen, 1992, Stelmach and Worringham, 1985, Grabiner and Enoka, 1995). Three measures which have been used to assess the speed of response in young and older adults include reaction time, movement time, and response time. Reaction time is defined as the time required to initiate a movement as rapidly as possible in response to a suddenly presented and unanticipated visual, auditory, or tactile stimulus (Schmidt and Lee, 1999). Movement time is the interval between the initiation of the response (i.e., the end of reaction time) to the completion of the movement, and is therefore dependent on the nature of the movement task (Schmidt and Lee, 1999). Response time is the overall time required to initiate and complete the movement response and represents the sum of reaction time and movement time.

To improve understanding of the central and peripheral processes involved in reaction time responses, researchers have placed electrodes over the belly of the appropriate muscle to measure when the muscle is activated. The EMG data revealed that the overall reaction time can be divided into two periods called premotor reaction time and motor reaction time (Figure 2.5; Botwinick and Thompson, 1966a, Botwinick and Thompson, 1966b). For a period of time after the stimulus has been presented, the EMG signal remains silent. This period is premotor reaction time, and includes events such as receptor activation by the stimulus, central processing of the stimulus (e.g., stimulus identification and movement programming), and the transmission of neural impulses along sensory and motor nerves (Schmidt and Lee, 1999, Marteniuk, 1976). The motor reaction time is the period where there is increased EMG activity but no observable movement has been produced. Within this period of electromechanical delay, processes within the muscle itself are converting electrical signals into muscle force (Schmidt and Lee, 1999). Movement occurs when the amount of generated muscle force is sufficient to overcome the inertial resistance of the limb or body.
Figure 2.5. Hypothetical sequence of events involved in a typical reaction time (RT) paradigm. Note that the premotor and motor RT periods are not to scale. The EMG signal is recorded from the muscle that is used to perform the movement task. Reprinted, with permission, from Schmidt and Lee (1999).

segment to be moved (Corcos et al., 1992). Given that the time taken for muscle force generation and nerve conduction are relatively small, the overall reaction time period is generally taken as a measure of an individual’s speed of cognitive processing (Birren and Fisher, 1995, Salthouse, 1985a, Welford, 1977).

In simple reaction time designs, participants are presented with only one stimulus to which they must initiate one specific response. As participants know the required response in advance, the simple reaction time is essentially a measure of the time required to detect the stimulus and initiate the response without any requirement to make decisions about the type of response (Amrhein, 1996, Welford, 1977). To increase the level of cognitive demand that is required to perform reaction time tasks,
researchers have increased the number and complexity of potential responses (Welford, 1977). Experiments in which there is more than one possible stimulus have an added decision making component and thus require a greater period of cognitive processing. A popular design is the choice reaction time experiment, in which participants must execute a unique motor response to each of several different stimuli. The longer reaction time under choice conditions is believed to reflect the additional requirements to identify which signal occurred and to select the corresponding motor response (Welford, 1980, Amrhein, 1996, Welford, 1977).

Since the initial study of Galton (1885), over 100 years of subsequent research has confirmed that the normal ageing process results in a slowing of reaction time (Birren and Fisher, 1995). This slowing of reaction time in older age is perhaps one of the most extensively verified findings in the ageing and sensorimotor literature (Salthouse, 1985b, Welford, 1984, Cerella, 1991). Ageing also results in slowing of movement execution, and therefore a general slowing of overall movement responses is typically observed for older adults (Cerella, 1991, Birren and Fisher, 1995). Despite adjusting for potentially confounding factors such as education, intelligence, motivation, disease, and physical activity habits, studies have still found that reaction time slows with ageing (Welford, 1977). The general view is that reaction time lengthens continuously throughout the adult lifespan. An eight year longitudinal study by Fozard et al (1994) of 1,265 community-living people aged 17 to 96 years found that simple reaction time increased at a rate of 0.5 ms per year after age 20. Cerella (1985) performed a meta-analysis of 18 reaction time studies and found that a generalised slowing coefficient explained over 94% of the variance in the average reaction times of older adults compared with young adults across 189 different task conditions. Factors such as sensory deficits, increased joint stiffness, slower nerve conduction speed, slower build up of muscular force, and cautious movement strategies have been suggested to influence age-related slowing of reaction time (Welford, 1977, Welford, 1980, Welford, 1984). However, the major determinant of the age-related slowing in reaction time is believed to result from deterioration of CNS mechanisms such as information processing capacity and central integrative function (Birren and Fisher, 1995, Salthouse, 1985a, Welford, 1977, Welford, 1984).
Slowed reaction time has also been consistently identified as a significant risk factor for falls. For example, a number of studies have shown that slower simple visual reaction time requiring a hand-press response is both significantly associated with falls (Lord et al., 1991a, Lord et al., 1994b, Lord and Clark, 1996) and is a significant predictor of multiple fallers versus non-multiple fallers (Lord et al., 1991a, Lord and Dayhew, 2001, Lord et al., 1994b). Nevitt and colleagues (1991) found that slower simple visual reaction time requiring a hand response was significantly associated with injurious falls in a 1 year prospective study involving 326 community-living older people. Lajoie and Gallagher (2004) examined simple auditory reaction time during upright stance in 125 older individuals that were recruited from the community, senior residences, and nursing homes. They found that reaction time was significantly slower in the individuals with a history of falls. Anstey and colleagues (2009) examined the reaction times of a sample of 658 community-dwelling older adults that included a small number of individuals with chronic diseases such as diabetes, stroke, and Parkinson’s disease. They found that multiple fallers had slower reaction times across a variety of different choice tasks involving responses with both the hands and feet compared with non-fallers. Taken together, these findings suggest that slower reactive motor responses may be implicated in falls. However, more studies are required that examine the reaction times of older adults under conditions where whole body postural stability must be maintained.

2.4 Task Specific Postural Control in Relation to Ageing and Falls

2.4.1 Quiet stance

Quiet stance is a postural task in which an individual stands as still as possible for a given period of time. Although quiet stance is not normally performed in daily life unless an individual is instructed to do so (e.g., soldiers or performing arts), the task has still been extensively researched (Balasubramaniam and Wing, 2002, Winter, 1995). The performance of quiet stance is usually measured from the postural sway oscillations of the body (see also sections 2.2.3 and 2.2.4). As quiet stance is a relatively easy postural task, an individual’s performance is often considered to
represent their baseline level of postural stability with respect to more dynamic tasks (Hsiao-Wecksler et al., 2003). However, this may not always be a valid assumption because deterioration of postural control is best detected under more dynamic and challenging movement conditions compared with quiet stance (Baloh et al., 1995, Baloh et al., 1994, Lajoie et al., 1996, Prioli et al., 2006). For example, Liaw and colleagues (2009) found no significant age-related differences in COP excursions during quiet stance, however significant age-related differences emerged when participants performed voluntary postural sway movements or reacted to translations of the standing surface. Similarly, Delbaere and colleagues (2006a, 2006b) found significant differences between fallers and non-fallers in the performance of AP voluntary postural sway, but did not detect any differences between these groups during quiet stance. Therefore, quiet stance may only provide limited information about a person’s capacity to maintain their balance during more dynamic activities.

Although the amplitude of COP displacement is the most common measure of quiet stance, many other COP indicators of standing steadiness have been calculated from the domains of position, area, velocity, frequency, and complexity/chaos (Raymakers et al., 2005, Demura et al., 2008a). Overall, COP velocity and frequency measures appear to be the most sensitive to ageing (Raymakers et al., 2005, Baloh et al., 1994, Winter, 1995, Demura et al., 2008a, Prieto et al., 1996). With regards to falls, a relatively consistent finding is the decline in ML postural stability for fallers compared with non-fallers (Maki and McIlroy, 1996, Lord et al., 1999, Rogers and Mille, 2003). In a review of nine prospective studies that used force platform measures to predict falls (Piirtola and Era, 2006), five studies found significant associations with falls (Bergland et al., 2003, Bergland and Wyller, 2004, Topper et al., 1993, Maki, 1993, Maki et al., 1994, Thapa et al., 1996, Stel et al., 2003b), whereas the other four studies did not (Kario et al., 2001, Brauer et al., 2000, Baloh et al., 1998, Hill et al., 1999). The force platform measures that were significantly predictive of falls in the five studies were related to the amplitude and speed of the ML COP.

As quiet stance does not challenge postural stability to any great extent, it is not surprising that many studies have also found no significant differences in postural sway with ageing (Collins et al., 1995, King et al., 1994, Prioli et al., 2006) or balance impairment (Schieppati et al., 1994, Horak et al., 1992, Beckley et al., 1993, Gauthier-Gagnon et al., 1986). It is likely that the ability to stand still can be maintained until severe functional decline in postural control (Maki and McIlroy, 1996, Prioli et al., 2006, Panzer et al., 1995, Carr and Sheperd, 1998, Rogers et al., 2003a, Hageman et al., 1995). Older adults are typically well within their balance capacity during quiet stance, and therefore very few falls are thought to occur during quiet stance (Patla et al., 1990b, Wolfson et al., 1992). Furthermore, age-related increases in COP amplitude during postural sway may reflect a postural strategy used by older adults to augment the level of sensory feedback rather than a sign of instability (Patla et al., 1990a, Carpenter et al., 2010).
Given the limitations of quiet stance assessment, a growing body of literature has examined standing steadiness under more challenging postural conditions. One method of challenging quiet stance is to reduce the base of support area in which standing stability must be regulated. Four progressively more difficult stances have been investigated: (1) feet side-by-side touching (Romberg stance), (2) one foot placed forwards with its heel aligned with the toes of the other foot (semi-tandem stance), (3) heel of one foot directly in front of and touching the toes of the other foot (tandem or sharpened Romberg stance), and (4) standing on one foot only (single leg stance) (Rogers et al., 2003a). Progressive increases in the amplitude of postural sway have been noted across these conditions respectively (Day et al., 1993, Kirby et al., 1987, Guralnik et al., 1994). The reduced base of support area also accentuates differences in postural sway between young and older adults (Hurley et al., 1998, Prioli et al., 2006, Ekdahl et al., 1989, Jonsson et al., 2007, Iverson et al., 1990, Briggs et al., 1989, Bohannon et al., 1984, Balogun et al., 1994, Speers et al., 1998, Amiridis et al., 2003) and between fallers and non-fallers (Vellas et al., 1997, Studenski et al., 1991, Heitmann et al., 1989, Maki et al., 1991, Delbaere et al., 2006b, Melzer et al., 2004, Lord et al., 1999).

An alternative approach of challenging standing postural control is to manipulate the sensory conditions of the visual, vestibular, and somatosensory systems. When vision is withdrawn during quiet stance, the increased postural sway associated with ageing (Hay et al., 1996, Whipple et al., 1993, Shumway-Cook and Woollacott, 2001, Panzer et al., 1995, Pyykko et al., 1990) and increased falls-risk (Lord et al., 1991a, Melzer et al., 2004, Lord et al., 1994a) becomes more pronounced. Manipulation of static and dynamic visual cues within the environment also results in stronger postural sway responses for older adults compared with young adults (Redfern et al., 1997, Poulain and Giraudet, 2008, Haibach et al., 2008, Haibach et al., 2007, Prioli et al., 2005, Prioli et al., 2006, Wade et al., 1995) and for fallers compared with non-fallers (Haibach et al., 2008, Tobis et al., 1985, Ring et al., 1988). Ring and colleagues (1988) found that recent fallers (within two weeks) and remote fallers (within one year) exhibited significantly more sway compared with non-fallers when they were subjected to
destabilising visual illusions. These results suggest that fallers may have an increased reliance on visual information for postural control compared with non-fallers, which would increase their likelihood of falling when visual information is reduced or conflicts with the other senses (Tobis et al., 1985, Lord et al., 1994a).

Somatosensory perturbations have been employed by requiring participants to stand on thick foam surfaces, which reduce proprioceptive feedback from the feet and ankles (Lord and Menz, 2000). Under these conditions, increased sway has been noted with ageing (Tanaka et al., 1997, Redfern et al., 1997, Raymakers et al., 2005, Gill et al., 2001, Choy et al., 2008) and for fallers compared with non-fallers (Lord et al., 1991a, Lord and Clark, 1996, Anacker and Di Fabio, 1992, Lord and Dayhew, 2001, Lord et al., 1994a, Lord et al., 1994b). Other somatosensory perturbations such as high frequency tendon vibration also tend to exacerbate age-related increases in postural sway (Hay et al., 1996, Teasdale and Simoneau, 2001). The disruption of vestibular information for postural control is another method of challenging quiet stance. In particular, a backward tilt of the head from its normal position is suggested to reduce the gain of the utricular otolith organs (Diener and Dichgans, 1988, Brandt et al., 1981). Jackson and colleagues (1991, 1996) observed that older adults compared with young adults had increased postural sway during head extension, and also increased rates of balance loss when head extension was added to visual and somatosensory perturbations.

Taken together, the literature indicates that postural sway measures obtained during normal quiet stance are not particularly valid indicators of an individual’s level of postural stability. The reduction in postural stability associated with ageing and falls appears to be more readily detected during dynamic, challenging postural tasks in which older adults are more likely to lose their balance and fall. Therefore, caution should be exercised when using quiet stance measures to infer differences in postural stability between different age and falls-risk groups. Quiet stance may be more useful for investigating the sensory and motor mechanisms of postural control rather than as a screening test for falls-risk (Winter, 1995, Baloh et al., 1994).
2.4.2 External perturbations of stance

To improve understanding of the full repertoire of responses that the postural control system uses to maintain stability, mechanical perturbations have been applied to the body of participants to induce an unexpected loss of balance. A number of studies have employed push or pull forces that were applied to various segments of the body such as the legs, waist, or head to evoke compensatory responses (Shupert and Horak, 1996, Horak et al., 2001, Horak et al., 1994, Hsiao-Wecksler et al., 2003, Rietdyk et al., 1999b, Wolfson et al., 1986, Hilliard et al., 2008, Patton et al., 2006, Rogers et al., 2001a). However, a more widely used experimental technique has been to tilt or translate the support surface on which the participant stands to induce relative motion between the base of support and the COM (Horak et al., 1997). In the first studies to examine postural responses following external perturbations (Nashner, 1976, Nashner, 1977), participants were required to stand on a moving platform that could translate in the AP direction or tilt about the ML axis to induce toes-up and toes-down rotations. Three discrete pre-programmed postural responses called the ankle, hip, and stepping strategies were documented (Nashner and McCollum, 1985, Horak and Nashner, 1985). These strategies repositioned the COM following a forward or backward translation of the moving platform, and were characterised by different patterns of segmental coordination and muscle activation.

The ankle strategy was characterised by a distal-to-proximal sequence of ankle, thigh, and trunk muscle activity that repositioned the COM by rotating the body primarily about the ankle joint similar to a single-segment inverted pendulum (Horak and Nashner, 1986, Horak et al., 1989). The ankle strategy was used in response to small and low velocity perturbations during stance on a firm and wide surface capable of resisting ankle torque (Horak et al., 1989). The hip strategy was characterised by a proximal-to-distal sequence of trunk to thigh muscle activity, and a corresponding angular rotation of the trunk about the hip joint (Horak and Nashner, 1986, Nashner and McCollum, 1985). The hip strategy was used in response to larger and faster perturbations or in situations that limited the production of ankle torque, such as stance on a short or compliant surface (Horak and Nashner, 1986, Nashner and
McCollum, 1985). See Figure 2.6 for examples of the ankle and hip strategies. The stepping strategy was characterised by rapid step, hop or stumble movements which created a new base of support to capture the new position of the COM (Horak et al., 1989). The stepping strategy was initially thought to be used when the ankle and hip strategies were unsuccessful in compensating for very large or fast perturbations (Horak and Nashner, 1986, Horak et al., 1989). However, participant’s responses were constrained in the initial studies because they were instructed not to step, except to avoid a fall. More recent studies that did not constrain participant responses found that compensatory stepping and arm grasping were commonly initiated early after the onset of perturbation, even when the perturbations were small and stability could have been maintained using ankle or hip strategies (Maki and McIlroy, 1997, Maki et al., 2003).

Ageing is known to have a detrimental influence on the speed and coordination of compensatory postural responses. In studies where stepping was discouraged, older adults exhibit delayed responses (Okada et al., 2001, Stelmach et al., 1989, Woollacott et al., 1988), increased levels of induced sway (Wu, 1998, Stelmach et al., 1989, Baloh et al., 1994, Gu et al., 1996), reversals in the typical sequence of muscle activation (Woollacott et al., 1986, Woollacott, 1993), breakdown of functional coupling between muscles (Woollacott et al., 1986, Woollacott, 1986, Stelmach et al., 1989, Woollacott et al., 1988), increased muscular co-contraction (Okada et al., 2001, Manchester et al., 1989, Woollacott et al., 1988), greater reliance on hip strategy compared with ankle strategy (Okada et al., 2001, Manchester et al., 1989, Gu et al., 1996), and greater occurrence of balance loss (Wolfson et al., 1992, Camicioli et al., 1997, Woollacott et al., 1986) compared with younger adults. When stepping and grasping reactions were not discouraged, older adults initiated steps at lower levels of perturbation (Mille et al., 2003, Jensen et al., 2001, Luchies et al., 1994, Pai et al., 1998), and were more likely to take multiple steps and initiate arm movements to grasp for safety rails (Pai et al., 1998, Maki et al., 2000, Luchies et al., 1994, Wolfson et al., 1986, McIlroy and Maki, 1996) compared with young adults. Older adults also experienced lateral instability during compensatory stepping as observed from more
Figure 2.6. Muscle activation patterns and segmental motions associated with the ankle and hip strategies for correcting forward (top) and backward (bottom) translations of the standing surface. (a-b) On a normal surface, the ankle strategy corrects sway via a bottom-up coordination pattern of EMG activity. (c-d) On a short surface that limits ankle torque production, the hip strategy corrects sway via a top-down coordination pattern of trunk and thigh EMG activity. The dashed stick figures represent the corrected body position. Para: lumbar paraspinal muscles; Abd: rectus abdominis; Ham: hamstrings; Quad: quadriceps; Gast: gastrocnemius; Tib: tibialis anterior. Reprinted, with permission, from Horak and Nashner (1986).

laterally directed foot placement following AP perturbations (McIlroy and Maki, 1996) and increased incidence of inter-limb collisions when responding to ML perturbations (Maki et al., 2000, Rogers and Mille, 2003). Allum and colleagues (2002) examined age-related differences in compensatory responses during platform tilts about the AP axis. Young adults responded with early and large trunk rotations in the opposite
direction to the platform tilt. In contrast, older adults had trunk rotations in the same direction as platform tilt (i.e., towards the fall side), which was also accompanied by protective arm movements presumably to cushion a potential fall. The authors concluded that increased trunk stiffness was a key biomechanical change with ageing that interferes with compensatory trunk movements and facilitates an impending fall. Collectively, the literature indicates that the compensatory postural responses of older adults are less effective for recovering from a sudden loss of balance compared with young adults.

The stepping and grasping strategies are often the last line of defence to recover from a loss of balance (Maki and McIlroy, 2006). For example, video surveillance of falls and near-fall events in aged-care institutions has shown that reactive limb movements were very common following a loss of balance with compensatory stepping observed in approximately 32% to 45% of incidents and compensatory grasping observed in approximately 65% to 72% of incidents (Holliday et al., 1990, Connell, 1996a, Maki and McIlroy, 1997). This evidence suggests that stepping and grasping responses play a vital role in preventing falls, and also that fallers and non-fallers may differ in the characteristics of compensatory stepping and grasping. Maki and colleagues (2001) monitored 64 older adults for 12 months following an initial assessment of their stepping and grasping responses to AP support-surface translations. The use of arm reactions and a slowing of arm reactions were found to be predictive of falls. In addition, the tendency to take multiple steps to recover balance was predictive of AP falls, and the tendency to use more laterally directed steps was predictive of ML falls. Further evidence for reduced lateral stability in fallers compared with non-fallers has been reported in recent prospective studies of stepping responses to laterally directed perturbations (Hilliard et al., 2008, Patton et al., 2006, Rogers et al., 2001a).

In contrast, a number of studies have also found no significant differences between fallers and non-fallers in the characteristics of compensatory balance reactions (Maki et al., 1990, Maki et al., 1991, Baloh et al., 1998, Baloh et al., 1995). However, falls were measured retrospectively in three out of four of these studies. Therefore, these
disparate findings may be partly related to experimental design limitations. Although compensatory stepping and grasping responses appear to be very important in avoiding falls, more scientific evidence is required to verify the capacity of external perturbation paradigms to predict falls in older adults (Horak et al., 1997, Maki and McIlroy, 1997). In particular, many studies of compensatory postural responses have used relatively small sample sizes, and therefore may have lacked statistical power to detect important differences between groups. Although a number of studies have demonstrated that older adults’ compensatory postural responses can be improved with training interventions (Mansfield et al., 2010, Rogers et al., 2003b, Shimada et al., 2004), more research is required to determine whether these improvements can significantly reduce falls.

2.4.3 Rapid voluntary stepping

Rapid voluntary steps are an important response for protecting postural stability during activities of daily living (Rogers et al., 2003b, St George et al., 2007). In particular, rapid voluntary steps are often used during walking when a hazard in the gait path is suddenly detected (St George et al., 2007). In the laboratory, testing of rapid voluntary step performance involves measurement of step responses under reaction time conditions. For ease and simplicity of testing, most studies have examined rapid step responses initiated from a standing position rather than during walking. Although measurement of rapid stepping from a static position is beneficial for clinical tests of falls-risk, the lack of a preliminary dynamic movement component may underestimate the true effects of ageing and falls-status on voluntary stepping in real world environments. The first study to investigate the effect of ageing on rapid voluntary stepping was conducted by Patla and colleagues (1993). In this study, young and older adults stepped with the right limb in the forward, backward or right directions in response to a three-choice visual cue. The older adults exhibited slower reaction time and took longer to transfer their body weight between the limbs during the postural adjustment compared with the young adults. The authors concluded that slower steps by older people may reduce their ability to avoid falls. Similarly, Melzer and colleagues (2004, 2007) investigated rapid step responses of the dominant limb in
the forward, backward, and lateral directions in response to a cutaneous cue applied to the leg. It was found that ageing resulted in slower reaction, weight transfer, and swing execution times (Melzer and Oddsson, 2004). In two other prospective studies, multiple fallers compared with non-fallers (Melzer et al., 2007) and older individuals who were seriously injured from a fall compared with those not seriously injured (Melzer et al., 2009b) were slower to complete the sub-events of rapid voluntary stepping whilst concurrently performing a cognitive task. A potential limitation of the above studies was that participants always stepped with the same limb. Therefore, participants knew the appropriate weight shift response in advance of receiving the cue, which may have underestimated group differences in stepping performance.

Rapid voluntary stepping with both limbs in multiple directions has also been compared between younger and older adults and between older adults with different classifications of falls-risk. Lord and Fitzpatrick (2001) designed a test of ‘choice stepping reaction time’ in which participants were required to rapidly step onto one of four randomly illuminated panels with either limb (i.e., two panels were positioned anterior and lateral to each foot). Choice stepping reaction time was defined as the period from the illumination of the panel to foot contact with the panel. In a large study of 30 younger adults and 477 retirement village residents, it was found that ageing and a history of falls were significantly associated with slower choice stepping reaction time (Lord and Fitzpatrick, 2001). Interestingly, slower reaction time was the strongest predictor of falls-history above a range sensorimotor, balance, and neuropsychological variables and was significantly correlated with visual contrast sensitivity, leg strength, simple reaction time, and standing and leaning balance. The authors suggested that impaired voluntary stepping may contribute to falls and that their choice stepping reaction task was a composite measure of falls-risk in older people. Subsequent studies from this research group have examined choice stepping responses when participants were required to step over small obstacles and/or perform a concurrent cognitive task (St George et al., 2007, Sturmieks et al., 2008a). Slower choice stepping reaction time (cue onset to EMG onset) and step execution time (EMG onset to foot contact) were found with ageing and in older adults with high
falls-risk compared with low falls-risk. Older adults also experienced difficulty in coordinating their rapid voluntary steps under these challenging task conditions because they frequently stepped in the wrong direction, contacted the obstacles, and made more errors on the cognitive task compared with the young.

Other investigations of age-related differences in rapid voluntary stepping have examined the additional influences of cue type, simple versus choice design, step direction, and moving targets. Rogers et al (2001b) investigated the influence of light, auditory, and electrotactile cues on forward stepping responses in young and older adults and found that age-related differences reaction and weight transfer times were unaffected by cue type. Luchies and colleagues (2002) found that reaction, weight shift, and landing times were significantly slower with ageing under both simple and choice reaction time conditions, and that weight shift and landing times were progressively slower for the lateral, diagonal, and anterior-posterior step directions, respectively. Tseng and colleagues (2009) investigated the effect of ageing and perturbations on voluntary steps by unexpectedly moving an anterior step target by 20 cm to the right midway through the voluntary step. Older adults were particularly susceptible to the target shifts as they had slower reactive responses to alter the propulsive forces of the supporting foot and the swing trajectory of the stepping foot compared with the young. Challenge to ML stability has been increased during rapid voluntary steps by requiring participants to step forwards onto a heightened block. Slowed stepping during these types of stepping tasks was found to be significantly associated with falls among older adults (Brauer et al., 2000, Miyamoto et al., 2008).

Collectively, the evidence indicates that the speed of rapid voluntary stepping actions slows with ageing and increased falls-risk across a variety of different task conditions. This slowing of voluntary stepping with ageing and increased falls-risk is consistent because almost all studies found significant differences between age groups and falls groups. The strong association of voluntary stepping reaction time with falls and postural control physiology suggests that slowed postural reaction time contributes to
falls and would therefore be useful to screen the falls-risk of older adults clinically. More research is required to determine the effects of ageing and falls-status on rapid voluntary stepping whilst walking through real world environments.

2.4.4 Gait tasks

During gait tasks such as walking and obstacle crossing, the COM travels outside the base of support area provided by the supporting limb for significant periods of time (Woollacott and Tang, 1997, Sutherland et al., 1995, Winter, 1991). This represents a challenging situation for the postural control system compared with quiet stance because stability must be regulated with respect to a moving base of support. Moreover, the human body is a top-heavy structure because two-thirds of the body’s mass is located within the head, arms, and trunk segments (Winter, 1991). This top heaviness is believed to pose a significant challenge to the regulation of upper body postural motions during gait (Winter et al., 1990c, Prince et al., 1997). Given the high demands placed upon posture and balance control during gait-related tasks, it is not surprising that a significant proportion of older adults experience difficulty with these types of movements. Approximately 8% of community-dwelling older adults report difficulty walking and require the assistance of a person or gait device for safe ambulation (Rogers et al., 2003a). Consistent with the self-reported difficulty of the elderly in performing gait tasks, episodes of falls and balance loss by older people are most likely to occur during gait (Cali and Kiel, 1995, Berg et al., 1997, Norton et al., 1997).

kinetic differences in walking patterns are also significantly associated with ageing (Kerrigan et al., 1998, Prince et al., 1997) and with falls among older adults (Lord et al., 1996, Maki, 1997, Barak et al., 2006). Wolfson and colleagues (1990) performed video-based analysis of the self-selected walking pattern of 49 nursing home residents with the aim of identifying gait markers that were associated with falling. Individuals with a recent history of falls had a more ‘guarded’ posture as defined by greater hesitancy, slowness, diminished propulsion, and a lack of commitment in the stepping and arm swing actions. Collectively, the slower and altered walking patterns associated with ageing and increased falls-risk reflects the adoption of a conservative and safer gait pattern to facilitate dynamic stability (Lord et al., 2007, Lajoie et al., 1996, Menz et al., 2003c, Winter et al., 1990b, Woollacott and Tang, 1997). A growing body of literature has also identified that fallers have more variable walking patterns compared with non-fallers (Barak et al., 2006, Lord et al., 1996, Maki, 1997, Verghese et al., 2009, Guimaraes and Isaacs, 1980). These more variable patterns reflect a loss of automaticity and stability of walking, and suggest that variability of movement patterns is an important marker of falls-risk (Maki, 1997, Barak et al., 2006, Verghese et al., 2009, Guimaraes and Isaacs, 1980).

The ability to adapt level walking to cross over obstacles or to sidestep around them is a key requirement for successful navigation through complex environments. The large proportion of falls attributed to slips, trips, and stumbles indicates that deficits in obstacle negotiation performance by older adults are implicated in many falls (Bradley and Pointer, 2009). Studies examining age-related differences in the ability to avoid obstacles under time critical conditions have reported that older adults were slower to sidestep to the left or right during straight line walking (Gilchrist, 1998), and had a lower rate of success in completing rapid 90 degree turns within a series of prescribed available response times (Cao et al., 1997). These findings suggest that older adults have a reduced ability to incorporate a rapid sideways shift into their normal walking pattern to avoid contact with an unexpected hazard. When required to step over obstacles, older adults exhibit more cautious obstacle crossing strategies compared with young adults (Galna et al., 2009). These more cautious strategies are typically
defined by slower approach and crossing speeds, earlier adjustment of the approach walking pattern, shorter distances between the obstacle and the crossing step, and increased foot clearance over the obstacle (Wantanabe and Miyakawa, 1994, Chen et al., 1994b, Chen et al., 1994a, Chen et al., 1991, Lu et al., 2006). Older adults also exhibit altered coordination between COP and COM motions during obstacle crossing. In general, ageing is associated with reduced COP-COM separation in the target and non-target directions of travel (Huang et al., 2008, Hahn and Chou, 2004). Reduced COP-COM separation is beneficial for obstacle crossing under self-paced conditions because it reduces the acceleration of the COM, which is presumably easier for older people to stabilise during the crossing phase (Huang et al., 2008, Hahn and Chou, 2004). Collectively, this body of research suggests that trips over obstacles by older people are due to their reduced ability to visually detect the obstacle rather than a risky crossing strategy (Brown et al., 2006, Patla et al., 1996). However, this may not always be the case because older adults are more likely to inadvertently contact obstacles under time critical conditions (Chen et al., 1996, Galna et al., 2009, Chen et al., 1991, Chen et al., 1994a).

Very few studies have contrasted the performance of fallers and non-fallers in obstacle negotiation tasks. Weerdesteyn and colleagues (2005) examined obstacle crossing skills in multiple fallers compared with single fallers and non-fallers during walking on a treadmill. A triggered electromagnet suddenly released an obstacle onto the conveyor belt of the treadmill in line with the left or right leg at various phases of the gait cycle. The multiple fallers had a significantly lower rate of success to step over the obstacles when only 200 to 250 ms of response time was available to execute the response. Chou and colleagues (2003) measured COM motions during level walking and obstacle crossing in 9 healthy older adults and 6 patients with balance complaints (three had vestibular dysfunction) and found that the patients had significantly greater and faster ML COM motion when crossing the obstacles. These results are in agreement with other reports that balance impaired older adults (Lee and Chou, 2006) and traumatic brain injured patients (Chou et al., 2004) have greater amplitude and speed of ML COM movement, and greater ML COP-COM separation during obstacle
crossing compared with age- and sex-matched controls. Taken together, these findings suggest that obstacle crossing substantially challenges postural stability, and that individuals with balance impairment may be at risk of falling laterally during this activity.

2.5 Voluntary Postural Sway and Leaning Movements

Voluntary postural sway is a task in which participants are required to rhythmically oscillate the body in the AP or ML directions whilst maintaining their standing balance. Voluntary leaning tasks are similar to voluntary postural sway as the initial phase of the lean movement involves a voluntary postural movement in a specific direction, which is usually followed by the requirement to maintain the lean statically for a given period of time. Voluntary postural sway and leaning actions examine a person’s capacity to voluntarily move the COM throughout the base of support area without losing their standing balance. These movements challenge the maintenance of standing postural stability for two main reasons. Firstly, large movements of the COM towards the limits of stability must be accompanied by postural responses that arrest the angular momentum of the body and ensure that the COM remains positioned within the base of support area (Krishnamoorthy et al., 2003, Krishnamoorthy and Latash, 2005). Secondly, the execution of these corrective postural responses, and the maintenance of any subsequent body lean, are likely to involve significant contributions from the sensory (Kavounoudias et al., 1998, Kavounoudias et al., 1999, Gurfinckel et al., 1995, Horak and Shupert, 2000, Maurer et al., 2001), cognitive (Lajoie et al., 1996, Woollacott and Shumway-Cook, 2002), and muscular systems (Fujiwara et al., 1982, Melzer et al., 2009a, Duarte and Zatsiorsky, 2002). Therefore, voluntary postural sway and leaning tasks represent useful models for examining standing postural control under challenging postural conditions. Importantly, these tasks overcome the limitations of quiet stance assessment because of the requirement for dynamic voluntary postural sway. Given the simplicity of these tasks and the ease of assessment, measures of voluntary postural sway and lean performance have been used to examine differences in postural control with ageing and increased falls-risk. A
The commercially available Limits of Stability and Rhythmic Weight Shifts tests offered by Balance Master® have been popular protocols for examining leaning and voluntary postural sway movements. Both tests examine the ability of participants to quickly and accurately move their COG within their base of support area via voluntary shifts of the COP. In the Limits of Stability test, participants shift their COP as quickly and accurately as possible using visual feedback towards one of eight targets (forward, backward, left, right, and four diagonal). Whilst leaning, participants then hold their COP as close as possible to the highlighted target for the remainder of an 8 s trial. Target amplitudes are usually set at the participant’s theoretical 100% limits of stability, which is the approximate maximum distance a person of a given height can move their COG within the base of support and still maintain their standing balance (Rose and Clark, 2000). In the Rhythmic Weight Shifts test, participants reciprocally move their COP between two square AP or ML targets to match the speed and trajectory of a visual target as accurately as possible (Boulgarides et al., 2003). As performance is guided by visual feedback in both the Limits of Stability and Rhythmic Weight Shifts tests, it may be argued that the primary task goal is to voluntarily shift the COP rather than to lean or produce voluntary postural sway. As the movements of the COP and COM are not causally related, it is possible to voluntarily shift the COP during upright stance without producing voluntary postural sway movements (i.e., oscillations of the COM) (Duarte and Freitas, 2005).

Ageing has been associated with slower and less accurate leaning and voluntary postural sway movements. In general, older adults exhibit slower reaction time to rapidly initiate leans, slower leaning velocity during the approach to the target (Nolan et al., 2010, Borah et al., 2007, Liaw et al., 2009, Nitz et al., 2003), reduced COP path accuracy during voluntary postural sway and target-directed leaning (Blaszczyk et al., 1994, Borah et al., 2007, Hageman et al., 1995, Liaw et al., 2009, Nitz et al., 2003),
<table>
<thead>
<tr>
<th>Study</th>
<th>Participants</th>
<th>Tasks</th>
<th>Summary of Findings</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Ageing</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Nolan et al. (2010)</td>
<td>Men (106) aged 30 to 80 years</td>
<td>Limits of Stability test.</td>
<td>Ageing was associated with slower RT to initiate leans by the 60th decade.</td>
</tr>
<tr>
<td>Liaw et al. (2009)</td>
<td>Young adults (45) Mid-aged adults (27) Older adults (35)</td>
<td>Limits of Stability 75% and AP Rhythmic Weight Shifts tests.</td>
<td>Ageing was associated with slower RT to initiate leans, slower COP velocity during lean approach, and reduced COP path accuracy during lean approach and Rhythmic Weight Shifts.</td>
</tr>
<tr>
<td>Borah et al. (2007)</td>
<td>Older adults (64)</td>
<td>Limits of Stability and Rhythmic Weight Shifts tests averaged over the AP and ML directions.</td>
<td>Ageing was associated with slower RT to initiate leans, slower COP velocity during the lean approach, and greater non-target COP velocity during Rhythmic Weight Shifts.</td>
</tr>
<tr>
<td>Holbein-Jenny et al.</td>
<td>Adults (52) aged 23 to 73 years</td>
<td>Forward, backward left, and right maximum leans for 2-3 s.</td>
<td>Ageing was associated with reduced COP lean amplitude as a percentage of the base of support area.</td>
</tr>
<tr>
<td>Nitz et al. (2003)</td>
<td>Older adults (366)</td>
<td>Limits of Stability test ML only.</td>
<td>Ageing associated with slower RT to initiate leans, slower COP velocity during lean approach, and reduced COP lean amplitude.</td>
</tr>
<tr>
<td>Okada et al. (2001)</td>
<td>Young men (8) Older men (8)</td>
<td>Forward, backward left, and right maximum leans for 20 s.</td>
<td>Older men had reduced backward COP lean amplitude and AP COP lean range normalised to foot length compared with young.</td>
</tr>
<tr>
<td>van Wegen et al. (2002)</td>
<td>Young adults (7) Older adults (7)</td>
<td>Forward and backward leans performed at 50% and 100% of maximum amplitude for 10 s.</td>
<td>Older adults had greater COP variability and reduced ‘time-to-contact’ of the COP with the base of support boundary whilst holding the leans compared with young adults.</td>
</tr>
<tr>
<td>Study</td>
<td>Participants</td>
<td>Tasks</td>
<td>Summary of Findings</td>
</tr>
<tr>
<td>-------------------------------</td>
<td>-------------------------</td>
<td>--------------------------------------------</td>
<td>-------------------------------------------------------------------------------------</td>
</tr>
<tr>
<td>Tanaka et al. (1997)</td>
<td>Young adults (13)</td>
<td>Forward, backward left, and right maximum leans for 10 s.</td>
<td>Older adults had reduced COP lean amplitude in all directions compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (7)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hageman et al. (1995)</td>
<td>Young adults (24)</td>
<td>Limits of Stability test.</td>
<td>Older adults had slower MT and reduced COP path accuracy during lean approach compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (24)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Blaszczyk et al. (1994)</td>
<td>Young adults (9)</td>
<td>Forward, backward, left, and right maximum leans for 10 s.</td>
<td>Older adults had reduced COP lean amplitude in the forward and backward directions, and slower MT and reduced COP path accuracy during lean approach compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (9)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>King et al. (1994)</td>
<td>Young adults (33)</td>
<td>Forward and backward maximum leans for 7 to 9 s.</td>
<td>Older adults had reduced forward and backward COP lean amplitudes normalised to foot length and greater COP amplitude whilst holding the leans compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (90)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Schieppati et al. (1994)</td>
<td>Young adults (18)</td>
<td>Forward and backward maximum leans for 51 s.</td>
<td>Group means were not compared for significance. AP COP lean amplitude was inversely related to PD severity.</td>
</tr>
<tr>
<td></td>
<td>Older adults (18)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>PD patients (18)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Blaszczyk et al. (1993a)</td>
<td>Young adults (13)</td>
<td>AP and circular voluntary postural sway.</td>
<td>Older adults had reduced COP amplitude for the voluntary postural sway movements compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (13)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Blaszczyk et al. (1993b)</td>
<td>Young adults (11)</td>
<td>Forward, backward left, and right maximum leans for 20 s.</td>
<td>Older adults had reduced COP amplitude for backward and left leans, and reduced COP amplitude whilst holding the leans compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (11)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
### Table 2.4. Continued.

<table>
<thead>
<tr>
<th>Study</th>
<th>Participants</th>
<th>Tasks</th>
<th>Summary of Findings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stelmach et al. (1989)</td>
<td>Young adults (6)</td>
<td>AP platform translations during forward or backward voluntary sways.</td>
<td>Older adults had slower and less stereotypically organised muscle activation patterns to coordinate automatic reflexes with the voluntary sways compared with young adults.</td>
</tr>
<tr>
<td></td>
<td>Older adults (6)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Fujiwara et al. (1982)</td>
<td>Adults (239) aged 20-79 years</td>
<td>Forward and backward maximum leans for 10 s.</td>
<td>Ageing was associated with reduced forward and backward COP lean amplitude and greater COP variability whilst holding the leans. COP lean amplitude was positively correlated with muscle strength.</td>
</tr>
<tr>
<td>Murray et al. (1975)</td>
<td>Young men (8)</td>
<td>Forward, backward left, and right maximum leans for 10 s.</td>
<td>Ageing was associated with reduced COP amplitude in all lean directions.</td>
</tr>
<tr>
<td></td>
<td>Mid-aged men (8)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Older men (8)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Falls-status</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lázaro et al. (2010)</td>
<td>Multiple fallers (113)*</td>
<td>Rhythmic Weight Shifts in the AP and ML directions.</td>
<td>Multiple fallers had reduced path accuracy during AP Rhythmic Weight Shifts compared with non-fallers.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (113)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Non-fallers (30)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Delbaere et al. (2006a)</td>
<td>Fallers (128)†</td>
<td>Rhythmic Weight Shifts in the AP and ML directions.</td>
<td>Fallers had reduced COP velocity and path accuracy during AP Rhythmic Weight Shifts compared with non-fallers.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (29)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Delbaere et al. (2006b)</td>
<td>Multiple fallers (52)†</td>
<td>Rhythmic Weight Shifts test in the AP direction.</td>
<td>Slower AP COP velocity was predictive of multiple falls.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (129)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
### Table 2.4. Continued.

<table>
<thead>
<tr>
<th>Study</th>
<th>Participants</th>
<th>Tasks</th>
<th>Summary of Findings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Melzer et al. (2004)</td>
<td>Multiple fallers (19)†</td>
<td>Limits of Stability test in the AP and ML directions.</td>
<td>No significant differences between groups were found for AP and ML COP lean amplitudes.</td>
</tr>
<tr>
<td></td>
<td>Non-multiple fallers (124)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Boulgarides et al. (2003)</td>
<td>Multiple fallers (20)†</td>
<td>Limits of Stability test.</td>
<td>COP measures were not significantly associated with falls-status.</td>
</tr>
<tr>
<td></td>
<td>Non-multiple fallers (79)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Girardi et al. (2001)</td>
<td>Multiple fallers (20)*</td>
<td>Limits of Stability test.</td>
<td>A higher proportion of the multiple fallers had reduced COP lean amplitudes (i.e., abnormal test results) based on age-matched normative data.</td>
</tr>
<tr>
<td></td>
<td>Single fallers (13)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Lord et al. (2001)</td>
<td>Fallers (174)*</td>
<td>Maximum voluntary AP postural sway measured using a swaymeter.</td>
<td>Fallers had reduced AP sway range compared with non-fallers.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (303)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Wallmann (2001)</td>
<td>Fallers (10)*</td>
<td>Limits of Stability test in forward direction only.</td>
<td>No significant differences between groups were found for forward COP lean amplitude.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (15)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Brauer et al. (2000)</td>
<td>Fallers (35)†</td>
<td>Forward, backward left, and right maximum leans for 5 s.</td>
<td>COP lean amplitudes were not significantly associated with falls-status.</td>
</tr>
<tr>
<td></td>
<td>Non-fallers (65)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Number of participants in each group displayed in parentheses.
*Falls retrospectively measured.
†Falls prospectively measured.

Limits of Stability and Rhythmic Weight Shifts tests performed at 100% of theoretical maximum lean amplitude unless otherwise stated.
All lean and voluntary sway movements were measured using a force plate with the exception of Lord, et al. (Lord and Fitzpatrick, 2001).
reduced COP amplitudes during maximum leaning and voluntary postural sway (Blaszczyk et al., 1993b, Blaszczyk et al., 1993a, Blaszczyk et al., 1994, Fujiwara et al., 1982, King et al., 1994, Murray et al., 1975, Nitz et al., 2003, Okada et al., 2001, Tanaka et al., 1997, Holbein-Jenny et al., 2007), and slower movement time to attain final leaning positions (Blaszczyk et al., 1994, Hageman et al., 1995) compared with young adults. Stelmac and colleagues (1989) examined a novel task in which AP platform translations were delivered to participants whilst they were in the process of swaying forward or backward. EMG patterns of the lower limbs demonstrated that the older adults had slower and less stereotypically organised muscle responses to the perturbation, and a reduced capacity to coordinate reflex and voluntary postural responses to facilitate their stability. With the exception of a study by Blaszczyk et al (1993b), four studies have found that older adults have greater COP motion whilst sustaining critical leaning positions compared with young adults (Hasselkus and Shambes, 1975, van Wegen et al., 2002, King et al., 1994, Fujiwara et al., 1982). This has been suggested to reflect reduced leaning stability of the older adults, however it may also reflect a deliberate action of the older adults to attain a greater lean amplitude (Schieppati et al., 1994). Taken together, the results suggest that older adults have a reduced capacity to stabilise leaning postures and to shift body position within the base of support area compared with young adults.

In contrast to investigations of ageing, fewer studies have compared healthy older adult fallers and non-fallers in the performance of voluntary postural sway movements. In a recent study, Lázaro et al (2010) investigated AP and ML Rhythmic Weight Shifts test performance in 226 community-dwelling older adults divided into multiple faller and non-faller groups based on self-reported 6 months history of falling. No significant group differences were found in COP velocity during test performance, however the multiple fallers exhibited reduced COP path accuracy with respect to the visual tracking target compared with non-fallers. Similarly, Delbaeare and colleagues (2006a, 2006b) examined associations between Rhythmic Weight Shifts task performance and 12 months prospective falls-status in a cohort of 263 community-dwelling older adults. It was found that fallers had slower COP velocity and reduced
COP path accuracy compared with non-faller, and that slower COP velocity was a significant predictor of multiple falls. In agreement with these findings, Ben Achour Lebib and colleagues (2006) reported that fallers exhibit slower movement time to attain final lean amplitude during the Limits of Stability task compared with non-fallers. Although falls have been associated with slower and more variable COP motion during leaning and voluntary postural sway, all studies (Melzer et al., 2004, Ben Achour Lebib et al., 2006, Boulgarides et al., 2003, Brauer et al., 2000, Wallmann, 2001) except one by Girardi and colleagues (2001) found that maximum COP lean amplitude during the Limits of Stability task was not reduced for fallers compared with non-fallers. The study of Girardi and colleagues (2001) found that a higher proportion of retrospectively determined multiple fallers compared with single fallers had reduced maximum COP lean amplitudes based on age-matched normative data reported by Amin et al (2000). As the average lean amplitudes were not reported or contrasted for significance between the multiple fallers and single fallers, no inferences can be made from this study regarding associations of falls-status with maximum lean amplitude.

The PPA involves a test of maximum voluntary AP sway among a number of other physiological and balance tests in the prediction of a person’s falls-risk (Lord et al., 2003). In this maximum voluntary sway test, a ‘swaymeter’ fastened to the waist at the approximate location of the COM directly records the sway displacement of participants as they sway forwards and backwards using their ankles. Significant reductions in maximum AP sway range have been reported for retrospectively determined fallers compared with non-fallers (Lord and Fitzpatrick, 2001) and multiple fallers compared with non-multiple fallers (Sturnieks et al., 2004). However, a high number of older adults in these studies had severe functional impairments including stroke, diabetes, and lower limbs arthritis. Therefore, it appears as though falls are associated with a reduced ability to quickly and effectively execute postural movements in specific directions, but the ability sway maximally or lean maximally within the base of support area is relatively well preserved in older adults without observable neurological or musculoskeletal disease.
2.6 Summary of Major Findings

2.1 Falls in older adults: Falls are a major international health problem for older adults aged 65 years and above. This problem is predicted to grow because current demographic change will result in worldwide increases in the proportion of adults who are most at risk of falling. Effective falls prevention initiatives need to be developed and implemented, however this is a difficult undertaking because the rates of falls, associated risk factors, accuracy of falls-risk screening tools, and methods of rehabilitation are diverse and depend on whether the older adults are dwelling in community, aged-care or hospital settings. The accuracy of common falls-risk assessment tools to predict falls based on qualitative measures of task performance has not been consistent between studies and is controversial.

2.2 Postural control: The postural control system represents a functional integration of the sensory, cognitive, motor, neuromuscular, and musculoskeletal physiological systems, which act cooperatively to regulate the orientation and stability of the body. Postural control is an extremely complex task as there are many different individual, task, and environment factors that have interactive effects on postural orientation and stability. To continually meet the demands of maintaining stability, the postural control system exhibits considerable robustness, adaptability, and redundancy of control. Postural control and the measurement of an individual’s level of postural stability can be described with reference to the spatiotemporal relationships between the COP, COM, and the base of support.

2.3 Postural control physiology: The normal ageing process results in structural and functional deterioration of the visual, vestibular, somatosensory, muscular, and cognitive systems. These age-related physiological deficits accumulate, and may also be accompanied by disease conditions, which ultimately results in balance impairment and increased falls-risk for many older adults. The deterioration in postural control physiology with ageing reduces the abilities of older adults to detect perturbations, and to execute proactive and reactive postural responses quickly and effectively to avoid falling. Fast reaction time may be important for avoiding falls, however most
investigations of reaction time have not been conducted under task conditions that are relevant to the maintenance of standing postural stability.

2.4 Task specific postural control in relation to ageing and falls: Accumulated deficits in postural control associated with ageing often translate into greater difficulty for older adults to perform basic movement tasks such as standing, balance recovery, rapid voluntary stepping, and gait. Poor performance of these activities by older adults has been associated with falls to varying degrees. Older adults tend to adopt compensatory movement patterns in an attempt to facilitate their postural stability. Such compensatory responses are typically defined by slower and more cautious movement strategies.

2.5 Voluntary postural sway and leaning movements: These tasks represent a simple approach to examining the speed, accuracy, and coordination of whole body postural movements under self-paced and reaction time conditions. The Limits of Stability and Rhythmic Weight Shifts tests have been the most commonly investigated tasks. Cross-sectional studies of healthy young and older adults have consistently revealed that ageing is significantly associated with slower and less accurate voluntary postural sway/leaning movements and also a reduced capacity to maximally shift the COP within the base of support area. In contrast, a survey of the literature revealed only ten studies which have examined associations between falls-status and voluntary postural sway/leaning performance in healthy older adults. Six of these studies examined maximum leaning amplitudes, the findings of which collectively suggest that maximum lean amplitudes are poor predictors of falls. The findings of the other four studies suggest that fallers exhibit slower and less accurate COP motions during voluntary postural sway compared with non-fallers. Given that only a small number of studies have examined differences in voluntary postural sway performance between young and older adults, significant gaps in knowledge were evident. In particular, it is unknown whether reaction time and coordination measures of voluntary postural sway performance decline with ageing and increased falls-risk, and whether these reaction time and coordination measures could accurately predict falls. Despite the
potential of voluntary postural sway as a convenient model for scientific research, there has been no comprehensive investigation as to whether voluntary postural sway tasks can improve understanding of postural instability among older people.
CHAPTER 3

EXPERIMENT ONE

Age-related differences in postural reaction time and coordination during voluntary sway movements

3.1 Introduction

Investigations of age-related differences in posture and movement control show that the older people tend to adopt cautious movement strategies that are typically defined by slower motor responses. For example, compared to the young, older individuals exhibit slower walking velocity (Maki, 1997, Prince et al., 1997), increased obstacle crossing time (Chen et al., 1991), and increased movement time during voluntary postural leaning (Blaszczyk et al., 1994, Hageman et al., 1995). These slower responses may be used in an effort to facilitate dynamic postural stability (Murray et al., 1969, Rosengren et al., 1998), but may also simply reflect age-related degenerative changes in postural control.

The normal ageing process is consistently associated with slower movement responses and hence increased reaction time (Fozard et al., 1994, Welford, 1977). However, the majority of age-related reaction time increases have been observed during tasks that involve initiation of upper-limb movement from a resting position while the subject adopts a seated posture (see for example, Era et al., 1986, Yan et al., 2000). Fewer studies have examined age-related differences in reaction time under conditions where participants are concurrently required to maintain standing postural stability. As the requirements of achieving stability are highly influenced by task constraints (Newell, 1986), it is important to examine age-related differences in reaction time...
under challenging postural conditions (Lajoie et al., 1993, Lajoie et al., 1996, Teasdale et al., 1993). The reaction time associated with initiating whole body motion would seem more relevant to maintaining dynamic postural stability compared with reaction time obtained under task conditions that do not substantially perturb balance (Lord and Fitzpatrick, 2001).

The maintenance of postural stability depends also on the response strategy implemented by the postural control system. Assessment of whole body postural coordination responses may therefore provide improved understanding of age-related declines in stability (Nardone et al., 2000, Wu, 1998). However, many investigations of standing postural control use measures derived from the centre of pressure (COP) alone. An approach based solely on COP excursions may be limited as they can only provide a global indication of body motion. To obtain information about the inter-segmental dynamics of maintaining stability, it may be necessary to measure postural motion at additional locations such as the trunk and head (Accornero et al., 1997, Cromwell et al., 2001, Menz et al., 2003c, Menz et al., 2003b).

Increasingly rigid movement patterns for older adults compared with young adults have been reported in studies examining the strength of coupling between motions of the body’s segments. This increased rigidity of older individuals may be attributed to passive osseoligamentous changes in the joints (Berne et al., 1999, Einkauf et al., 1987, Svanborg, 1988), or due to an active postural response strategy (Hortobagyi and Devita, 2006). Evidence of strategies of increased rigidity by older adults compared with young adults have been noted during Romberg stance (Accornero et al., 1997, Benjuya et al., 2004), voluntary downward stepping movements (Hortobagyi and DeVita, 2000), horizontal support surface translations (Manchester et al., 1989, Okada et al., 2001, Wu, 1998), and during gait (Kavanagh et al., 2004, Murray et al., 1969). Although a rigid movement strategy can reduce the amplitude of COP excursion during quiet stance (Horak et al., 1992, Melzer et al., 2001), this strategy may not always represent the most optimal method for achieving dynamic stability. For example, a rigid movement strategy reduces the degrees of freedom available for the

The purpose of this study was to examine age-related differences in reaction time and coordination between COP, trunk, and head motion during standing voluntary postural sway movements. It was predicted that the older adults would exhibit slower reaction time and altered coordination responses characterised by increased strength of coupling between COP, trunk and head motions compared with the young adults.

3.2 Methods

3.2.1 Participants

Ten healthy young men (24 ± 5 years) and eight older men (75 ± 2 years) participated in this study. Young participants were tertiary students recruited from the University population, and older participants were recruited from the surrounding community. Prior to testing, participants provided written informed consent, completed a medical history questionnaire, and had their height and weight measured. All participants had normal or corrected-to-normal vision, and reported no known neurological or cognitive disorders, proprioceptive deficits, or history of neuromuscular injury that could influence performance. All experimental procedures complied with the guidelines of the Griffith University Human Research Ethics Committee.

3.2.2 Experimental design

Participants were required to react to a suddenly presented stimulus during two different conditions: (1) sway initiation, where participants rapidly initiated anterior-posterior (AP) or medial-lateral (ML) voluntary postural sway from quiet stance, and (2) orthogonal switches of sway, where participants performed rapid orthogonal switches of voluntary postural sway between the AP and ML directions (Figure 3.1). Prior to the presentation of a two-choice auditory cue, participants stood quietly (sway initiation task) or performed AP or ML voluntary postural sway at their preferred frequency (orthogonal switch task). Two-choice auditory cues were “forward” or
Figure 3.1. Overview of the sway initiation and orthogonal switch voluntary postural sway tasks. (a) Accelerometer placement and standing position on the force plate. Arrows delineate the AP and ML directions for the voluntary postural sway movements. (b) The experimental design showing all conditions for the sway initiation and orthogonal switch tasks. (c) AP and ML COP excursions for a sway initiation trial where the participant initially stands quietly, reacts to a “left” auditory cue, and then performs voluntary postural sway in the ML direction. (d) AP and ML COP excursions for an orthogonal switch trial where the participant initially performs AP voluntary postural sway, reacts to a “left” auditory cue, and then performs ML voluntary postural sway.

<table>
<thead>
<tr>
<th>Initial Condition</th>
<th>Two-Choice Audio Cue</th>
<th>Sway Response</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quiet stance</td>
<td>forward or backward</td>
<td>AP sway</td>
</tr>
<tr>
<td>Quiet stance</td>
<td>left or right</td>
<td>ML sway</td>
</tr>
</tbody>
</table>

Switch of Sway

<table>
<thead>
<tr>
<th>Initial Condition</th>
<th>Audio Cue</th>
<th>Sway Response</th>
</tr>
</thead>
<tbody>
<tr>
<td>AP sway</td>
<td>left or right</td>
<td>ML sway</td>
</tr>
<tr>
<td>ML sway</td>
<td>forward or backward</td>
<td>AP sway</td>
</tr>
</tbody>
</table>
“backward” and “left” or “right,” and were used to specify the direction in which participants had to initiate/switch their voluntary postural sway. After reacting in the specified direction, participants performed voluntary postural sway at their preferred frequency in the AP direction if the cue was forward or backward, or in the ML direction if the cue was left or right. Auditory cues were pre-recorded using Waveform Hold and Modify software (WHAM, version 1.33), and were delivered to participants through PC speakers. For the voluntary postural sway movements, participants were instructed to restrict their actions to the ankle joint for AP sway and to sequentially load and unload each leg for ML sway (Winter et al., 1996). All movements were performed with arms held parallel to the body.

Five 25 s trials were collected for each direction of response (i.e., forward, backward, left, right). Anticipatory responses were discouraged by randomising the auditory cue within a 15 to 19 s period of each trial, and implementing 10 ‘catch’ trials per task in which no auditory cue was presented. Sequence of cue presentation was counterbalanced within the sway initiation and orthogonal switch tasks for all participants. Prior to data collection, participants self-selected a comfortable barefoot stance position on the force plate, which was traced onto sheets of paper and standardised between trials. Foot positioning variables of inter-malleolar distance, big-toe distance, effective foot length, and base of support area were computed as described by Chiari and colleagues (2002).

3.2.3 Instrumentation and data acquisition

Participants stood on a three-component strain-gauge force plate (Bertec BP5050) from which the co-ordinate location of the participant’s COP was computed. Three-dimensional accelerations of the head and lower trunk were measured with two lightweight tri-axial accelerometers (Crossbow CXL02LF3, range ± 2g). The head accelerometer was positioned over the occipital pole using an elastic headband, and the lower trunk accelerometer was secured over the L3 spinous process with rigid sports tape (see Figure 3.1). Prior to data collection, both accelerometers were statically calibrated using gravity to obtain voltage outputs equivalent to -1g of
acceleration for each axis. Force plate and accelerometer data were synchronised and sampled at 100 Hz using custom LabVIEW software (National Instruments, version 7.1).

3.2.4 Data analysis

COP and accelerometer data were band-pass filtered using a fourth-order Butterworth filter with cut-off frequencies of 0.1 and 5.0 Hz. A trigonometric correction similar to that described by Moe-Nilssen and Helbostad (2002) was applied to the AP and ML accelerations to account for gravitational artefact introduced into acceleration data during accelerometer tilt.

The dependent measures examined in this study included reaction time, COP-trunk-head reaction time difference, COP-trunk-head coupling, sway frequency, and sway amplitude. Reaction time was the period between cue onset and the first observable change in COP or acceleration. The first observable change was determined using a custom designed algorithm. When the subject reacted to the auditory cue, an initial peak in COP/acceleration was clearly discernable (see Figures 3.1 and 3.2). The algorithm searched backwards from this peak to the point at which COP/acceleration amplitude first exceeded 2 SDs of baseline data (i.e., the 3 s period immediately prior to auditory cue onset). For the reaction time difference measures, the difference in timing between COP-trunk, COP-head, and trunk-head reaction times was computed. The reaction time difference magnitude was used to determine the degree to which the COP, trunk, and head reacted simultaneously. The polarity of the reaction time difference was used to examine the sequence of movement initiation between the COP, trunk, and head.

In addition to the reaction time based measures, data analysis was also performed on periods of voluntary postural sway. Pearson product-moment cross-correlation analysis with time lags of ± 1500 ms was used to examine coupling between COP-trunk, COP-head, and trunk-head motion. Peak correlation coefficients and phase differences were extracted from this analysis. Strength of coupling was assessed with the absolute value of the peak correlation coefficient. The phase difference between
the two signals was defined as the time lag corresponding with the peak correlation coefficient. Similar to the reaction time difference measures, the polarity of the phase difference determined the signal that was leading or lagging during the voluntary postural sway. Sway frequency was determined by manual selection of successive COP peaks and troughs, and was computed as the inverse of the average period. Sway amplitude was measured with COP range, which was computed as the difference between minimum and maximum excursion for the AP and ML directions. Sway frequency and amplitude were normalised to the participant’s height and weight prior to statistical analysis. Cross-correlation, frequency, and amplitude analyses were performed on periods of voluntary postural sway (3 s pre- and post-cue) for the orthogonal switch task only. All data analysis was performed using custom designed software in Matlab version 7.1 (MathWorks, R14).

3.2.5 Statistical analysis

Analysis of Variance (ANOVA) was used to determine the effect of: (1) age (2 levels: young adults, older adults) on anthropometric and foot positioning measures; (2) age (2 levels: young adults, older adults), task (2 levels: sway initiation, orthogonal switch), and body region (3 levels: COP, trunk, head) on reaction time; (3) age (2 levels: young adults, older adults), task (2 levels: sway initiation, orthogonal switch), and body region difference (3 levels: COP-trunk, COP-head, trunk-head) on reaction time difference; (4) age (2 levels: young adults, older adults), and time (2 levels: pre-cue, post-cue) on COP sway frequency and amplitude for the orthogonal switch task; and (5) age (2 levels: young adults, older adults), time (2 levels: pre-cue, post-cue), and body region difference (3 levels: COP-trunk, COP-head, trunk-head) on coupling for the orthogonal switch task. Repeated measures were specified for ANOVAs involving the time and task factors. Planned contrasts were used to compare individual means. All statistical tests were performed using custom software developed in SAS for Windows (SAS Institute, version 9.1), with significance accepted at $P < .05$. 
3.3 Results

3.3.1 General participant characteristics

General participant characteristics of height, mass and foot positioning are presented in Table 3.1. Young adults were taller than the older adults ($F_{1,16} = 9.55$, $P = .007$).

3.3.2 Representative data

Representative examples of COP excursion and accelerations of the trunk and head during the sway initiation and orthogonal switch tasks are displayed in Figure 3.2. The data revealed a similar pattern between COP and acceleration of the trunk and head, although the pattern of COP excursion was approximately 180° out-of-phase with respect to the accelerations. Following the cue onset, there was a delay in movement initiation for both sway initiation and orthogonal switch tasks.

Table 3.1
Anthropometric and foot positioning characteristics of the young and older adults.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Young adults (n = 10)</th>
<th>Older adults (n = 8)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height (cm)</td>
<td>179.3 ± 7.5</td>
<td>169.7 ± 5.1*</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>74.8 ± 10.6</td>
<td>82.3 ± 8.7</td>
</tr>
<tr>
<td>Inter-malleolar distance (cm)</td>
<td>16.4 ± 1.6</td>
<td>14.6 ± 3.3</td>
</tr>
<tr>
<td>Big toe distance (cm)</td>
<td>24.1 ± 2.8</td>
<td>27.4 ± 4.0</td>
</tr>
<tr>
<td>Effective foot length (cm)</td>
<td>26.3 ± 1.6</td>
<td>25.2 ± 1.0</td>
</tr>
<tr>
<td>Base-of-support area (cm$^2$)</td>
<td>532.7 ± 79.9</td>
<td>529.0 ± 80.0</td>
</tr>
</tbody>
</table>

Values are mean ± standard deviation.
*Significantly different from young adults, $P < .05$. 
Figure 3.2. Representative COP excursion and trunk and head accelerations of a young adult participant during the initiation of ML voluntary postural sway and during an orthogonal switch of AP to ML voluntary postural sway. Data is shown for 8 s before and after the onset of the auditory cue ($t = 0$ s).
3.3.3 Effect of age, task and body region on reaction time

A main effect of age ($F_{1,16} = 8.57$, $P = .009$) indicated that older adults were slower to react compared with the young adults (mean ± SD; young adults: 452 ± 5 ms; older adults: 597 ± 13 ms). A main effect of task ($F_{1,16} = 80.48$, $P < .001$) also indicated that participants exhibited slower reaction time during the orthogonal switch task compared with the sway initiation task (mean ± SD; sway initiation task: 483 ± 9 ms; orthogonal switch task: 548 ± 10 ms). A significant age-by-task interaction ($F_{1,16} = 17.54$, $P < .001$) was also detected (Figure 3.3). Planned contrasts revealed that the older adults exhibited slower reaction time compared with the young adults during the sway initiation task ($F_{1,15} = 5.09$, $P = .039$) and the orthogonal switch task ($F_{1,15} = 12.46$, $P = .003$). Orthogonal switch reaction time was also slower compared with sway initiation reaction time for both the young adults ($F_{1,15} = 12.86$, $P = .003$) and older adults ($F_{1,15} = 77.84$, $P < .001$).

3.3.4 Effect of age, task and body region difference on reaction time difference

A main effect of age ($F_{1,16} = 8.27$, $P = .011$) indicated that older adults reacted with less reaction time difference compared with the young adults (mean ± SD; young adults: -22 ± 4 ms; older adults: mean = -2 ± 5 ms). That is, the older adults exhibited reduced differences in reaction time between the COP, trunk, and head body regions compared with the young adults.

3.3.5 Effect of age and time on the frequency and amplitude of voluntary postural sway

No significant age differences were observed for the frequency or amplitude of voluntary postural sway during the orthogonal switch task ($P > .05$).
**Figure 3.3.** Reaction time of the young and older adults for the sway initiation and orthogonal switch voluntary postural sway tasks. Error bars represent one standard error of the mean. *Significantly different between age-groups or sway tasks, $P < .05$.

### 3.3.6 Effect of age, time and body region difference on coupling during voluntary sway for orthogonal switches of voluntary postural sway

A main effect of age ($F_{1,16} = 4.92$, $P = .041$) indicated that older adults exhibited reduced phase differences compared with the young adults (mean ± SD; young adults: $-19 ± 3$ ms; older adults: $2 ± 3$ ms). A significant age-by-body region difference interaction ($F_{2,32} = 4.54$, $P = .018$) was also observed (Figure 3.4a). Planned contrasts revealed that the older adults exhibited a reduced phase difference between COP and head motion compared with the young adults ($F_{1,31} = 10.65$, $P = .003$). A significant age-by-time interaction ($F_{1,16} = 6.36$, $P = .023$) was also detected (Figure 3.4b). Planned contrasts revealed that the older adults exhibited reduced phase differences compared with the young adults during post-cue sway ($F_{1,15} = 9.19$, $P = .008$). No significant age-related differences in strength of coupling were detected ($P > .05$).
Figure 3.4. (a) Phase difference between COP-trunk, trunk-head, and COP-head motion during voluntary postural sway of the orthogonal switch task for the young and older adults. (b) Phase difference during pre- and post-cue voluntary postural sway of the orthogonal switch task for the young and older adults. Error bars represent one standard error of the mean. *Significant difference between age-groups, $P < .05$. 

- Table 3.1: Time
  - Pre-Cue
    - COP-trunk: 0
    - trunk-head: 0
    - COP-head: 0
  - Post-Cue
    - COP-trunk: 0
    - trunk-head: 0
    - COP-head: 0

- Table 3.2: Phase Difference (ms)
  - Body Region Difference: COP-trunk, trunk-head, COP-head
  - Young adults: 0
  - Older adults: 0

- Figure 3.4a: Graph showing phase difference between COP-trunk, trunk-head, and COP-head motion during voluntary postural sway of the orthogonal switch task for the young and older adults.

- Figure 3.4b: Graph showing phase difference during pre- and post-cue voluntary postural sway of the orthogonal switch task for the young and older adults. Error bars represent one standard error of the mean. *Significant difference between age-groups, $P < .05$. 

- Table 3.1: Time
  - Pre-Cue
    - COP-trunk: 0
    - trunk-head: 0
    - COP-head: 0
  - Post-Cue
    - COP-trunk: 0
    - trunk-head: 0
    - COP-head: 0

- Table 3.2: Phase Difference (ms)
  - Body Region Difference: COP-trunk, trunk-head, COP-head
  - Young adults: 0
  - Older adults: 0
3.4 Discussion

The purpose of this study was to examine age-related differences in reaction time and coordination between COP, trunk, and head motion during voluntary postural sway movements. It was hypothesised that healthy older adults would exhibit slower reaction time and altered postural coordination characterised by increased strength of coupling between COP, trunk and head motions compared with the young adults. The results confirmed these hypotheses as the older adults exhibited slower reaction time during both sway initiation and orthogonal switch tasks, and reduced differences in reaction time and phasing between COP, trunk, and head motion compared with the young adults. These findings suggest that the normal ageing process is associated with altered postural responses during the execution of a rapid change in the direction of whole body motion.

The older adults exhibited significantly slower reaction time compared with the young adults during the sway initiation and orthogonal switch tasks. Age-related slowing of reaction time is a finding that is consistent with many studies investigating the effect of age on the speed of motor performance (Welford, 1977). However, the majority of studies which have reported age-related slowing of reaction time were conducted under task conditions that were not associated with the maintenance of whole body balance and stability. The present study has built on these findings by showing that older adults were slower than young adults to initiate a change in the direction of voluntary postural sway when postural stability had to be concurrently maintained. Age-related declines in the speed of postural responses under conditions of whole body stability have also been observed during rapid voluntary stepping (Luchies et al., 2002, Melzer and Oddsson, 2004, Patla et al., 1993), sudden turns during gait, and termination of gait (Cao et al., 1997, Cao et al., 1998a). The slower reaction time of the older adults in this study may be attributed to common characteristics of the ageing process such as decreased central processing speed, increased passive joint stiffness, a more cautious strategy of response, increased preference for accuracy (Welford, 1977, Welford, 1984), and/or altered neuromuscular properties (Narici and Maganaris, 2006).
Reaction time was found to be significantly slower during the orthogonal switch task compared with the sway initiation task for both the young adults and older adults. The main difference between these tasks was the initial condition, which was quiet stance for the sway initiation task and was voluntary postural sway for the orthogonal switch task. These results suggest that an orthogonal switch of voluntary postural sway was more demanding compared with initiation of voluntary postural sway, presumably because of the requirement to switch angular momentum between orthogonal axes. Furthermore, given that postural control of quiet stance requires a certain degree of attention (Melzer et al., 2001), it is also likely that greater attention was necessary when the regulation of postural stability was challenged by voluntary postural sway (Cordo and Nashner, 1982, Woollacott and Shumway-Cook, 2002). It has been noted that greater attention is required to maintain postural stability in other tasks which elicit increased sway above that observed during quiet stance such as Romberg (Teasdale et al., 1993) or tandem stance (Barra et al., 2006). Therefore, central processing time of the auditory cue may have been lengthened in the orthogonal switch task due to the attentional demand of stabilising the voluntary postural sway.

Older adults exhibited altered voluntary postural sway responses as assessed by reaction time differences with respect to the auditory cue and phase differences between COP, trunk, and head motion compared with the young adults. The reduced reaction time difference for the older adults indicates that motion at each body region was initiated more simultaneously compared with the young adults. Reduced phase-differences were also observed between body regions for older adults compared with young adults during voluntary postural sway. In addition, a post hoc analysis was conducted which indicated that the phase differences for the COP-trunk, COP-head, and trunk-head body regions were significantly different from zero for the young adults, but not for the older adults. The phase differences for the young adults were negative (see Figure 3.4a), suggesting a bottom-up coordination sequence with the inferior body region leading the superior body region (i.e., COP leads trunk, trunk leads head). In contrast, the older adults exhibited a more synchronous coordination pattern between COP, trunk, and head motion, suggesting that they swayed more like a single
rigid unit. Taken together, these results indicate that the older adults had stronger
temporal coupling between locations compared with the young adults during both
their voluntary postural sway and reaction responses.

A more rigid posture effectively reduces the degrees of freedom to be organized by
the postural control system, and thus decreases the demand to control the upper-body
separately from the lower-body (Vereijken et al., 1992). Consistent with the current
findings, older adults tend to exhibit increased postural rigidity during other
challenging postural tasks. For example, older individuals exhibit increased whole-body
rigidity (Wu, 1998) and higher probability of lower limb muscle co-activation
is perturbed by a moving platform. Despite the increased stability provided by a more
rigid posture, it has the mechanical effect of reducing joint mobility and flexibility
(Berne et al., 1999, Einkauf et al., 1987, Kornecki, 1992, Svanborg, 1988). Reduced
mobility is unlikely to be beneficial to dynamic postural stability and may actually
compromise the execution of voluntary or compensatory responses (Allum et al., 2002,
Wu, 1998). In fact, the results of this study suggest that the increased whole body
rigidity of older adults could be a contributing factor to their slower reaction time for
initiating a change in the direction of voluntary postural sway.

3.5 Conclusion

A novel approach for examining age-related differences in reaction time and dynamic
postural stability was presented. Older adults exhibited significantly slower reaction
time compared with young adults when required to initiate a rapid change in the
direction of voluntary postural sway. The older adults also adopted more rigid
coordination strategies between COP, trunk, and head motion during both their
reaction and voluntary postural sway responses compared with the young adults. The
rigid movement strategy of the older adults was presumably generated in an effort to
compensate for increased challenge to the maintenance of stability.
CHAPTER 4

EXPERIMENT TWO

Voluntary sway and rapid orthogonal transitions of voluntary sway in young adults, and low and high falls-risk older adults

4.1 Introduction

The risk of falling increases with older age (Lord et al., 2003). Although the causes of falling among older people are multi-factorial, deterioration of balance control is a key factor (Lord et al., 2003, Horak, 2006). In the simplest context, the ability to avoid a fall involves three processes: (1) to detect a stimulus from the environment, (2) to process the information contained in the stimulus, and (3) to correctly execute the appropriate response within a critical time frame (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Age-related degeneration of the sensory, cognitive, and musculoskeletal systems has a negative influence on the execution of these processes, which reduces the ability of older people to regulate the orientation and stability of the body during everyday tasks (Horak, 2006). In particular, the slowing of postural movements (Lord and Fitzpatrick, 2001, St George et al., 2007), decreased leg muscle strength (Pijnappels et al., 2008), and deterioration in the coordination of reactive responses to postural perturbations (Allum et al., 2002) with ageing are believed to underlie the increased susceptibility of older people to falls.

Investigations of age-related differences in postural control have typically focused on movements performed in a single plane (Winter, 1995, Maki and McIlroy, 1996). In tasks that primarily involve anterior-posterior (AP) motion such as walking (Prince et al., 1997), obstacle crossing (Hahn and Chou, 2004), and recovering balance from a
forward lean by stepping (Wojcik et al., 1999), age-related deterioration of postural control is reflected in the slower and generally less effective postural responses of older adults compared with young adults. In tasks that involve medial-lateral (ML) motion such as laterally-directed waist pulls (Mille et al., 2005) or sideways translations of the standing surface (Maki et al., 2000), older adults are more likely to experience inter-limb collisions and also require a greater number of steps to recover balance compared with the young. In addition, older adult fallers have pronounced ML sway during quiet stance compared with non-fallers (Maki et al., 1994, Lord et al., 1999, Delbaere et al., 2006a), which is a finding that supports strong associations between deterioration of ML postural stability and increased falls-risk amongst older people (Rogers and Mille, 2003).

Given these age-related declines in whole-body movement for a single plane, a postural task that requires coordination between the AP and ML directions may be especially difficult for older people to stabilise. One simple and easy technique to investigate the effect of ageing on combined AP and ML movement is to examine postural sway in multiple directions (Hageman et al., 1995). A previous study (Tucker et al., 2008) assessed age-related differences in postural responses during rapid orthogonal switches of voluntary postural sway between the AP and ML directions under choice reaction time conditions. The results demonstrated that older individuals exhibited slower reaction time and tighter temporal coupling between centre of pressure (COP), trunk, and head motion compared with the young. Although age-related differences were detected in the speed and coordination of the postural response during orthogonal sway transitions, it remains unclear how these differences influenced postural stability. In addition, because there is substantial heterogeneity of postural control amongst older people (Horak et al., 1989), it is unclear whether these results generalise to subpopulations of older adults such as those with different levels of falls-risk.

A simple method to quantify postural stability is to examine the difference between the horizontal locations of the COP and centre of mass (COM) (Masani et al., 2007,
Winter, 1995). Greater difference, or separation, between the COP and COM (COP-COM) increases the moment-arm between the body weight vector and the vertical ground reaction force. This in turn produces a net joint torque about the ankles or hips that accelerates the COM in the opposite direction to the COP (Winter, 1995). During quiet stance, greater COP-COM separation in the AP and ML directions has been observed with ageing (Masani et al., 2007, Berger et al., 2005a), stroke (Corriveau et al., 2004b), and peripheral neuropathy (Corriveau et al., 2000). Ageing and neurological impairment therefore reduce the ability to minimise horizontal accelerations of the COM in a task where less postural motion is typically associated with better performance. During more dynamic tasks such as walking and obstacle crossing, the amplitude of COP-COM separation is significantly altered with ageing (Hahn and Chou, 2004), increased falls-risk among older people (Lee and Chou, 2006), traumatic brain injury (Chou et al., 2004), and stroke (Said et al., 2008). Collectively, these studies suggest that COP-COM separation is sensitive to the decline in postural stability that occurs with ageing and pathology to the balance control system across a range of different postural tasks.

The purpose of this study was to examine differences between young adults, low falls-risk older adults, and high falls-risk older adults in the speed of response and postural stability during voluntary postural sway and rapid orthogonal transitions of voluntary postural sway between the AP and ML directions. It was hypothesised that ageing would result in slower reaction time and movement time, and altered COP-COM separation during voluntary postural sway and orthogonal transitions, as observed for the low and high falls-risk older adults compared with the young adults. It was also hypothesised that similar differences in the outcome measures during voluntary postural sway and orthogonal transitions would be observed for the high falls-risk older adults compared with the low falls-risk older adults.
4.2 Methods

4.2.1 Participants

Twenty-five younger (age range: 19-35 years) and 48 older (age range: 65-93 years) men and women participated in this study. Young adults were recruited from the University population, and older adults were recruited from the local community by written invitation, newspaper advertisements, and fliers placed within retirement villages. Older adults were required to be older than 65 years of age, and volunteers were excluded if they reported neurological, cognitive or proprioceptive disorders and recent or recurrent history of musculoskeletal injury and/or surgery. All participants provided written informed consent. The guidelines of the Institutional Human Research Ethics Committee were followed during all experimental procedures.

4.2.2 Falls-risk assessment

Falls-risk of the older adults was calculated using the long-form Physiological Profile Assessment (PPA), which has been validated in prospective studies of falls in community and institutional settings, and predicts those at increased risk of falling with 75% accuracy (Lord et al., 2003). The long-form PPA includes tests of vision, sensation, leg muscle strength, reaction time, postural sway, and postural coordination. Scores from each of these tests were combined to provide an overall falls-risk score that ranged from -2 (very low falls-risk) to 4 (very marked falls-risk). The older adults were divided into two groups that represented high falls-risk (≥ 1) and low falls-risk (< 1) (St George et al., 2007). The modified Baecke questionnaire and the Falls Efficacy Scale International (FES-I) were used to determine physical activity levels and the fear of falling, respectively, of the low and high falls-risk older adults. The number of falls in the past year was also obtained from self-reports, where a fall was defined as “an event which resulted in a person coming to rest unintentionally on the ground or other lower level, not as the result of a major intrinsic event or an overwhelming hazard” (Lord et al., 1999).
4.2.3 **Instrumentation**

Reflective markers (14 mm) were placed on participants according to the full-body VICON Plug-In Gait model (Oxford Metrics Group Plc., West Way, Oxford, UK). During testing, participants stood on a multicomponent force plate (Type 9287A, Kistler Instrument Corporation, Amherst, NY, USA), which was surrounded by eight VICON MX-13 infrared cameras for 3D motion capture (Oxford Metrics Group). Force plate and 3D kinematic data were synchronised and collected using Nexus software v1.3 (Oxford Metrics Group) with a sampling frequency of 100 Hz for marker trajectories and 1000 Hz for ground reaction force data. To prevent injuries resulting from falls, participants wore a light-weight safety harness during testing which was secured to the roof of the laboratory using a pulley system. The harness and pulley system were adjusted for each participant prior to testing to ensure that their voluntary postural sway movements were not restricted.

4.2.4 **Task and experimental design**

Foot position was traced onto sheets of paper that were affixed to the force plate. Stance width was standardised to 10% of the participant’s height with an outward foot angle of 15° (McIlroy and Maki, 1997). Participants matched their measured foot position to their footprints prior to the commencement of each trial. During testing, participants executed rapid, orthogonal transitions of voluntary postural sway between the AP and ML directions in response to a two-choice auditory cue (Tucker et al., 2008). For AP-ML transitions, participants initially swayed in the AP direction and then reacted to a “left” or “right” auditory cue. After reacting left or right, participants commenced and continued ML sway. For ML-AP transitions, participants initially swayed in the ML direction and then reacted to a “forward” or “backward” auditory cue. After reacting forward or backward, participants commenced and continued AP sway.

Auditory cues were presented to participants at approximately their neutral stance position after they had completed 2.0, 2.5, 3.0, 3.5, 4.0, or 4.5 oscillations of voluntary
postural sway. The COP amplitude in the direction of sway at cue onset was not significantly different between groups \( (P > .05) \). The number of oscillations prior to cue onset was randomised, and the sequence of auditory cue presentation was counterbalanced for each participant. All participants were instructed to react and to move as quickly as possible in the direction indicated by the auditory cue. Participants were also instructed to restrict motion to the ankle joint for AP sway, to sequentially load and unload each leg for ML sway, and for all sway movements, to keep the arms relaxed alongside the trunk and refrain from lifting the feet off the ground. Following one practice trial, five experimental trials were collected for each direction of response (i.e., forward, backward, left, right), with a minimum of three trials per direction suitable for subsequent analysis. All participants were provided with 30 s rest intervals between trials, and were given seated breaks as required. Trials in which the participant responded in the wrong direction, lost their balance and/or stepped were recorded.

The amplitude of voluntary postural sway was standardised during the AP-ML and ML-AP sway tasks by on-line monitoring of a marker placed over the tenth thoracic spinous process (T10). Prior to testing, T10 position was recorded during maximum static leans performed forward and backward (AP range), and left and right (ML range). Nexus software (Oxford Metrics Group) was then used to implement T10 biofeedback reference points that permitted sway within the middle 60% of the AP and ML ranges. When the real-time T10 position exceeded the reference point during voluntary postural sway, an auditory beep indicated the need to reverse the current direction of oscillation. During testing, all participants swayed within their designated AP and ML ranges at their preferred frequency. These frequencies were not significantly different between groups during pre- or post-transition voluntary sway for the AP-ML and ML-AP tasks \( (P > .05) \).

4.2.5 Dependent measures

Reaction time to the auditory cue was determined from the COP of the post-transition sway direction (e.g., ML COP for AP-ML transitions) using the algorithm of Mills et al
Experiment Two

A 20 ms sliding window moved forwards from the auditory cue at 1 ms intervals until all data points within the sliding window exceeded a threshold. The threshold was the average amplitude of the COP 100 ms immediately prior to the auditory cue, plus or minus 2 SDs of non-target COP data computed during two voluntary postural sway oscillations prior to the cue. Movement time was calculated from the end of reaction time to the first COP peak of voluntary postural sway in the direction indicated by the auditory cue. The raw amplitude of this first COP peak was recorded. The transition phase was defined as the movement time period. The pre- and post-transition phases were two complete oscillations of voluntary postural sway, which were immediately prior to the cue for the pre-transition phase, or immediately following the transition for the post-transition phase (see Figure 4.1). The COP, COM, and COP-COM dependent measures were calculated from the pre-transition, transition, and post-transition phases.

The COM was computed from the weighted sum of the centres of mass of the modelled body segments, which were obtained using the full-body VICON Plug-In Gait model. Subsequent data analysis was performed using custom designed software in Matlab v7.1 (Release 14, The Mathworks Inc., Natick, MA, USA). COM data was up-sampled to 1000 Hz, and then all data were filtered with a fourth-order, zero phase shift, band-pass Butterworth filter with cut-off frequencies at 0.1 and 10 Hz. This filter removed high frequency noise and also de-trended the COP and COM signals to ensure that their trajectories accurately reflected the AP and ML directions of body sway. COP and COM data were then normalised to the maximum (forward and right), and minimum (backward and left) COP amplitudes that were obtained from two trials of maximum voluntary AP and ML sway (see Table 4.1 for maximum COP amplitudes). For each phase of the task, root mean square (RMS) amplitude was calculated for the COP and COM in the target sway direction, and also in the non-target (orthogonal) direction. The separation difference between the COP and COM trajectories was also obtained by subtracting the COM from the COP, and then calculating the RMS of the data (COP-COM) in the target and non-target directions.
4.2.6 Statistical analysis

One-factor repeated measures Analysis of Variance (ANOVA), and Analysis of Covariance (ANCOVA) were used to detect main group effects (3 levels: young adults, low falls-risk older adults, high falls-risk older adults) for the AP-ML and ML-AP sway tasks separately. ANOVA was used to test for group differences in general characteristics and the pre- and post-transition variables. ANCOVA was used to test for group differences in the transition variables. The covariates for this analysis were the pre-transition variables that were significantly different between groups and COP amplitude in the direction of sway at cue onset. In the event of a significant main group effect, planned contrasts were used to compare individual means. All statistical analyses were performed using custom software developed in SAS for Windows v9.1 (SAS Institute Inc., Cary, NC, USA), with significance accepted at $P < .05$.

4.3 Results

4.3.1 Group characteristics

A significant main effect was detected between groups for age, PPA score, and maximum COP amplitude in the forward, left, and right directions (Table 4.1). The high falls-risk older adults were older and had a higher PPA score compared with the low falls-risk older adults ($P < .01$). The low and high falls-risk older adults were older and had reduced forward, left, and right COP amplitudes during maximum voluntary sway compared with the young adults ($P < .01$).

4.3.2 Representative data

Representative time-series plots of COP, COM and COP-COM data, and AP versus ML COP plots for an ML-AP trial for young, low falls-risk, and high falls-risk participants are displayed in Figure 4.1. Following cue onset, there was a delay in movement initiation prior to the reaction response. The reaction response was characterised by an initial peak of the COP with respect to the COM, which was followed by oscillations in the post-transition direction and diminished oscillation in the pre-transition direction. COP
Table 4.1
Group characteristics and maximum voluntary sway COP amplitudes.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Young adults</th>
<th>Low falls-risk older adults</th>
<th>High falls-risk older adults</th>
<th>$F$, $P$ value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Anthropometry</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>N (males, females)</td>
<td>25 (13,12)</td>
<td>32 (17,15)</td>
<td>16 (10,6)</td>
<td>NA</td>
</tr>
<tr>
<td>Age (years)</td>
<td>25 ± 4</td>
<td>74 ± 5*</td>
<td>79 ± 7†</td>
<td>$F_{2,70} = 719.19$, $P &lt; .001$</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>173.2 ± 10.7</td>
<td>166.6 ± 10.5*</td>
<td>166.4 ± 12.0</td>
<td>$F_{2,70} = 3.05$, $P = .054$</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>70.8 ± 14.5</td>
<td>79.8 ± 14.0</td>
<td>75.7 ± 18.3</td>
<td>$F_{2,70} = 2.44$, $P = .094$</td>
</tr>
<tr>
<td><strong>Falls-Risk Assessment</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PPA score</td>
<td>NA</td>
<td>0.4 ± 0.5</td>
<td>1.8 ± 0.6†</td>
<td>$F_{1,46} = 77.03$, $P &lt; .001$</td>
</tr>
<tr>
<td>Fallen in previous year (%)</td>
<td>NA</td>
<td>22</td>
<td>50</td>
<td>NA</td>
</tr>
<tr>
<td>Modified Baecke score</td>
<td>NA</td>
<td>16.3 ± 8.4</td>
<td>14.3 ± 9.3</td>
<td>$F_{1,46} = 0.50$, $P = .483$</td>
</tr>
<tr>
<td>FES-I</td>
<td>NA</td>
<td>21.7 ± 6.3</td>
<td>26.3 ± 10.6</td>
<td>$F_{1,46} = 3.30$, $P = .076$</td>
</tr>
<tr>
<td><strong>COP Amplitude</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Forward (mm)</td>
<td>146 ± 66</td>
<td>113 ± 48*</td>
<td>103 ± 35*</td>
<td>$F_{2,70} = 4.13$, $P = .020$</td>
</tr>
<tr>
<td>Backward (mm)</td>
<td>87 ± 34</td>
<td>83 ± 38</td>
<td>61 ± 23</td>
<td>$F_{2,70} = 3.00$, $P = .057$</td>
</tr>
<tr>
<td>Left (mm)</td>
<td>193 ± 86</td>
<td>156 ± 60*</td>
<td>117 ± 39*</td>
<td>$F_{2,70} = 6.36$, $P = .003$</td>
</tr>
<tr>
<td>Right (mm)</td>
<td>202 ± 86</td>
<td>150 ± 62*</td>
<td>119 ± 26*</td>
<td>$F_{2,70} = 8.32$, $P &lt; .001$</td>
</tr>
</tbody>
</table>

Values are means ± standard deviation unless otherwise indicated.

*Significantly different to young adults, $P < .05$.
†Significantly different to low falls-risk older adults, $P < .05$.
PPA = Physiological Profile Assessment; FES-I = Falls Efficacy Scale International; COP = centre of pressure; NA = not available.
Figure 4.1. Representative plots of normalised COP, COM and COP-COM amplitudes in the AP and ML directions during an ML-AP trial for young, low falls-risk, and high falls-risk participants in which a forward cue was presented. The time series plots illustrate 10 s of data prior to, and following the onset of the cue (dot vertical line at \( t = 0 \) s). The reaction time period starts at the cue and ends at the vertical dash line (– –) and the movement time period starts at the dash line and ends at the vertical dash-dot line (— · —). Adjacent to the time plots are the
corresponding normalised AP versus ML COP plots for each participant. The symbols represent the coordinate location of the COP when a specific event occurred. ◆ = cue onset; ⊕ = reaction time; ● = movement time. The bold path highlights the COP trajectory from cue onset to movement time. With respect to neutral stance (zero amplitude), positive and negative values denote that for the AP direction, the COP and COM were located forward and backward, respectively, and that for the ML direction, the COP and COM were located right and left, respectively.

and COM excursions appeared smaller and more visually irregular in the non-target direction compared with the target direction of sway.

**4.3.3 Pre-transition variables for the AP-ML task**

A significant main effect was detected between groups for AP COP ($F_{2,70} = 5.28, P = .007$) and ML COP amplitudes ($F_{2,70} = 6.03, P = .004$). The low and high falls-risk older adults had reduced AP COP and increased ML COP amplitudes compared with the young adults ($P < .01$) (Figure 4.2a).

**4.3.4 Transition variables for the AP-ML task**

A significant main effect was detected between groups for reaction time ($F_{2,70} = 8.90, P < .001$), movement time ($F_{2,70} = 3.69, P = .031$), raw COP amplitude at the end of movement time ($F_{2,70} = 12.66, P < .001$), and the AP COP ($F_{2,70} = 7.29, P < .001$), AP COP-COM ($F_{2,70} = 6.71, P = .002$), and ML COP-COM amplitudes ($F_{2,70} = 3.58, P = .033$). Reaction time and movement time were slower for the low and high falls-risk older adults compared with the young adults, and the high falls-risk older adults had slower reaction time compared with the low falls-risk older adults ($P < .05$) (Figures 4.3a, and 4.3c). Raw COP amplitude was reduced for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults ($P < .05$) (mean ± SD; young adults: 139 ± 31 mm; low falls-risk older adults: 126 ± 29 mm; high falls-risk older adults: 106 ± 21 mm). The low and high falls-risk older adults also had reduced AP COP and AP COP-COM amplitudes compared with the young adults, and the high falls-risk older adults had reduced ML
COP-COM amplitude compared with the young adults and low falls-risk older adults ($P < .05$) (Figures 4.2b, and 4.2h).

**Figure 4.2.** Pre-transition, transition and post-transition amplitudes of the COP, COM, and COP-COM in the AP and ML directions for the young adults (YA), low falls-risk older adults (LFROA) and high falls-risk older adults (HFROA) for the AP-ML sway task. Values are means ± standard deviation in normalised units. *Significantly different to young adults, $P < .05$. †High falls-risk older adults significantly different to low falls-risk older adults, $P < .05$. 

```
Figure 4.3. Reaction and movement times of the young adults (YA), low falls-risk older adults (LFROA) and high falls-risk older adults (HFROA) during the AP-ML and ML-AP sway tasks. Values are means ± standard deviation. *Significantly different to young adults, \(P < .05\). †High falls-risk older adults significantly different to low falls-risk older adults, \(P < .05\).

4.3.5 Post-transition variables for the AP-ML task

A significant main effect was detected between groups for AP COM (\(F_{2,70} = 7.11, P = .002\)) and ML COP-COM amplitudes (\(F_{2,70} = 3.80, P = .027\)). The high falls-risk older adults had increased AP COM amplitude and reduced ML COP-COM separation compared with the young adults and low falls-risk older adults (\(P < .01\)) (Figures 4.2f, and 4.2i).
4.3.6 Pre-transition variables for the ML-AP task

A significant main effect was detected between groups for AP COM ($F_{2,70} = 10.01, \, P < .001$), ML COM ($F_{2,70} = 4.28, \, P = .018$), and ML COP-COM amplitudes ($F_{2,70} = 3.42, \, P = .038$). The low falls-risk older adults had greater AP COM amplitude compared with the young adults ($P = .020$) (Figure 4.4d). The high falls-risk older adults had increased AP COM and ML COM amplitudes compared with the young adults and low falls-risk older adults, and reduced ML COP-COM separation compared with the low falls-risk older adults ($P < .05$) (Figures 4.4d, and 4.4g).

4.3.7 Transition variables for the ML-AP task

A significant main effect was detected between groups for reaction time ($F_{2,70} = 24.13, \, P < .001$), movement time ($F_{2,70} = 4.82, \, P = .018$), raw COP amplitude at the end of movement time ($F_{2,70} = 6.92, \, P = .002$), and the AP COP ($F_{2,70} = 6.98, \, P = .002$), AP COM ($F_{2,70} = 3.96, \, P = .024$), and ML COP-COM amplitudes ($F_{2,70} = 11.83, \, P < .001$). Reaction time was slower for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults ($P < .05$) (Figure 4.3b). The high falls-risk older adults also had slower movement time compared with the young adults and low falls-risk older adults ($P < .05$) (Figure 4.3d). Raw COP amplitude was reduced for the low and high falls-risk older adults compared with the young adults ($P < .05$) (mean ± SD; young adults: $85 ± 18$ mm; low falls-risk older adults: $76 ± 20$ mm; high falls-risk older adults: $68 ± 18$ mm). Compared with the young adults, the low falls-risk older adults also had reduced AP COM amplitude, and the low and high falls-risk older adults had reduced AP COP and ML COP-COM amplitudes ($P < .05$) (Figures 4.4e, 4.4b, and 4.4h).

4.3.8 Post-transition variables for the ML-AP task

A significant main effect was detected between groups for AP COP ($F_{2,70} = 5.06, \, P = .009$), ML COP ($F_{2,70} = 3.50, \, P = .035$), ML COM ($F_{2,70} = 14.37, \, P < .001$), and ML COP-COM amplitudes ($F_{2,70} = 3.93, \, P = .024$). The low and high falls-risk older adults had reduced AP COP amplitude compared with the young adults ($P < .05$) (Figure 4.4c). The
low falls-risk older adults also had increased ML COP, ML COM, and ML COP-COM amplitudes compared with the young adults \((P < .05)\) (Figures 4.4c, 4.4f, and 4.4i). The high falls-risk older adults had increased ML COM amplitude compared with the young adults and low-falls-risk older adults \((P < .05)\) (Figure 4.4f).

**Figure 4.4.** Pre-transition, transition, and post-transition amplitudes of the COP, COM, and COP-COM in the AP and ML directions for the young adults (YA), low falls-risk older adults (LFROA) and high falls-risk older adults (HFROA) during the ML-AP sway task. Values are means ± standard deviation in normalised units. *Significantly different to young adults, \(P < .05\). †High falls-risk significantly different to low falls-risk older adults, \(P < .05\).
4.4 Discussion

4.4.1 Voluntary postural sway in the young adults, and low and high falls-risk older adults

The low and high falls-risk older adults compared with the young adults exhibited reduced COP amplitude in the forward and sideways directions during maximum voluntary postural sway and reduced target AP COP amplitude during voluntary postural sway and orthogonal transitions. These findings suggest that older people have a reduced ability to shift the body maximally within the base of support, and also to control voluntary postural sway via AP shifts of the COP. Age-related reduction in COP amplitude during voluntary postural sway movements may be due to reduced leg muscle strength, impaired perception of stability boundaries, or an attempt to increase the margin of postural stability (Blaszczyk et al., 1993b, Blaszczyk et al., 1994). As hypothesised, the high falls-risk group generally exhibited reduced target ML COP-COM separation compared with the young adults and low falls-risk older adults during ML voluntary postural sway and orthogonal transitions. The high falls-risk group therefore had a closer ML alignment between the COP and COM, which although stable under static conditions, would be unlikely facilitate postural stability during voluntary postural sway as the ability to generate a stabilising torque and accelerate the COM in the desired direction is reduced (Winter, 1995).

Increased postural sway in the non-target direction of movement was exhibited by the low and high falls-risk older adults compared with the young adults, and the high falls-risk older adults compared with the low falls-risk older adults. The increased non-target sway for the older adults compared with young adults was associated with increased COP and/or COM amplitude, whereas the increased non-target sway for the older adults with a higher risk of falling was more exclusively associated with increased COM amplitude. These results suggest that ageing and increased falls-risk reduce the ability to concurrently regulate AP and ML sway and to control multi-directional postural movements. Similar to the current results, increased non-target sway has been observed for older compared with young adults (Blaszczyk et al., 1994, Hageman
et al., 1995), and fallers compared with non-fallers (Delbaere et al., 2006a). As ageing and increased falls-risk are associated with a reduction in the capability to detect sensory stimuli (Grabiner and Enoka, 1995, Lord et al., 2003, Stelmach and Worringham, 1985), the increased non-target sway may be a mechanism to enhance sensory feedback and improve postural stability (Patla et al., 1990b). Alternatively, individuals with reduced postural control may have used more attention to stabilise the target direction of sway, which was presumably at the expense of the non-target direction (Woollacott and Shumway-Cook, 2002). The amplitude of sway in the non-target direction during voluntary postural sway may be a simple and easy test to screen older individuals for risk of falls.

4.4.2 Reaction and movement times of the young adults and low and high falls-risk older adults during orthogonal sway transitions

The results confirmed the hypotheses that ageing and increased falls-risk would be significantly associated with slower reaction and movement responses during rapid orthogonal transitions of voluntary postural sway. The slower responses of the older adults compared with the young adults also supports the results of a previous study involving young and older men (Tucker et al., 2008). In addition, the groups with slower movement time also had reduced raw amplitude of their COP responses to the auditory cue. Therefore, the associations between slower movement time with ageing and increased falls-risk occurred over shorter distances of COP displacement. Given the importance of being able to respond quickly to unexpected stimuli to avoid falls, a negative influence of aging and increased falls-risk to slow reaction and movement responses is important clinically (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Slower reaction time and movement time for high compared with low falls-risk individuals has been observed during voluntary stepping tasks that challenge whole-body stability (Lord and Fitzpatrick, 2001, St George et al., 2007), which together with the findings of the present study, suggest that the speed of postural responses are an important determinant of falls-risk in older people.
In agreement with the study hypotheses, the low and high falls-risk older adults generally exhibited reduced AP and ML COP-COM separation compared with the young adults during orthogonal transitions. Based on previous reports of high positive correlations between COP-COM separation and COM acceleration during postural sway (Masani et al., 2007, Yu et al., 2008), the current findings suggest that the older adults had a reduced ability to accelerate the COM in the desired direction during the transition phase compared with the young. The results suggest that the primary mechanism underlying the greater AP COP-COM separation for the younger adults compared with the older adults was increased AP COP amplitude. Another finding in agreement with the study hypotheses was that the high falls-risk older adults had decreased ML COP-COM separation during AP-ML transitions compared with the low falls-risk older adults. Considering that previous studies have also shown that individuals with reduced postural stability exhibit altered COP-COM separation in the ML direction (Chou et al., 2004, Lee and Chou, 2006), there is growing evidence that measures of ML COP-COM separation may be useful in differentiating between older individuals with low and high falls-risk.

One prominent result of this study concerned the generalised effect of ageing and increased falls-risk on the control of ML postural stability. Following ML-AP transitions, the low falls-risk older adults had increased non-target ML COP, COM, and COP-COM amplitudes compared with the young adults. These results suggest that the low falls-risk older adults experienced difficulty in suppressing their pre-transition ML voluntary postural sway during the post-transition phase compared with the young adults. In addition, five out of the seven significant differences between the low and high falls-risk older adults for the COP, COM, and COP-COM variables were related to the ML direction. The greater target and non-target ML COM amplitude of the high falls-risk
older adults compared with the low falls-risk older adults suggests they had poor control of their ML COM trajectory which reduced their ML COP-COM separation during voluntary postural sway and orthogonal transitions. Collectively, these results suggest that deterioration of postural control with ageing and increased falls-risk may be pronounced in the ML direction during voluntary postural sway movements. In support of our findings, particular declines in the control of ML postural stability with increased falls-risk have been reported during standing postural sway (Delbaere et al., 2006a, Lord et al., 1999, Maki et al., 1994).

4.4.5 Limitations

The findings of this study must be interpreted with caution because of limitations related to the statistical analysis. It may be argued that multivariate analysis would have been a more suitable approach for examining closely related variables such as the COP, COM, and COP-COM amplitudes during voluntary postural sway. In contrast, a more exploratory univariate approach was used in the current study so that it could be determined whether changes in COP-COM separation were associated with changes in COP or COM amplitudes, and whether group differences were present for the different tasks, phases of the task, and sway directions. The findings of the study support the use of univariate analyses because the significant group differences were not identical for all variables, tasks, and sway directions. However, the greater number of statistical tests without correction for multiple comparisons implies that some of the significant results may have occurred by chance. The high falls-risk older adults were also significantly older compared with the low fall-risk older adults. Therefore, it is possible that increased age and not necessarily increased falls-risk contributed to the significant differences between the low and high falls-risk older adults. To verify the effect of age on the results, the comparisons between the low and high falls-risk older adults were re-examined using a statistical adjustment for age. The results were virtually identical, and therefore the age-difference between the falls-risk groups does not affect the results of this study in any meaningful way.
4.5 Conclusions

The overall findings of the present study suggest that during voluntary postural sway tasks, older adults with increased susceptibility to falling have slowed postural reaction and movement times, and also exhibit differences in coordination between their COP and COM motions that are suggestive of reduced postural stability. The association between the PPA classifications of falls-risk with speed of response, non-target postural sway, and the amplitude of COP-COM separation suggests that voluntary postural sway task measures may be useful for quick screening tests of falls-risk prior to more detailed assessment of falls-risk by the PPA.
CHAPTER 5

EXPERIMENT THREE

Differences in rapid initiation and termination of voluntary postural sway associated with ageing and falls-risk

5.1 Introduction

Falls are frequent events among older people and can result in disability, fear of falling, admission into aged-care, and reduced quality of life (Hill et al., 2004). Accumulated deficits in the sensorimotor systems underlying balance are strongly predictive of falls among older adults (Lord et al., 2003). In particular, the normal ageing process results in progressive deterioration of the visual, vestibular and somatosensory systems, reduced speed of central processing and nerve conduction, sarcopenic reductions in muscle strength, and spatiotemporally disorganised muscular activation patterns (Horak et al., 1989, Woollacott, 1993). These physiologic changes are often accompanied by a generalised slowing of postural responses and altered movement patterns for older adults compared with young adults (Stelmach and Worringham, 1985). The movement patterns adopted by older adults often reflect a more conservative or cautious approach to balance maintenance (Hahn and Chou, 2004, Menz et al., 2003c). Heterogeneity in the normal ageing process also results in different rates of sensorimotor and balance decline in the older population such that some older adults experience high risk of falls whereas others experience low risk of falls (Horak et al., 1989).

Voluntary postural sway movements represent a convenient model for examining postural control. Not only does voluntary postural sway allow the potential to
systematically manipulate the aspects of the task and to objectively quantify the motor response (Borah et al., 2007, Rose and Clark, 2000), it is also a task that is sensitive to declines in movement performance with ageing and balance impairment among older people. Age-related differences in voluntary postural sway include slower reaction time to initiate sway during the Limits of Stability task (Borah et al., 2007, Liaw et al., 2009, Nitz et al., 2003), increased postural sway in the non-target direction of movement (Blaszczyk et al., 1994, Hageman et al., 1995), and reduced maximum lean amplitude (Blaszczyk et al., 1994, Murray et al., 1975). In a 12 month prospective study of 263 community-dwelling older adults, fallers were found to exhibit slower and less accurate voluntary postural sway movements compared with non-fallers (Delbaere et al., 2006a). Furthermore, tests of voluntary postural sway and leaning actions are significantly predictive of falls in older adults with foot problems, lower limbs arthritis, stroke, and diabetes (Lord and Fitzpatrick, 2001, Menz and Lord, 2001, Sturnieks et al., 2004). Taken together, previous research suggests that measures associated with voluntary postural sway tasks could be useful and accurate predictors of ageing and falls-risk related declines in postural stability.

During normal upright stance, postural sway in the anterior-posterior (AP) direction is regulated by the ankle dorsiflexors and plantar flexors, whereas postural sway in the medial-lateral (ML) direction is regulated primarily by the hip abductors and adductors with smaller contribution from the ankle inverters and everters (Day et al., 1993, Gatev et al., 1999). The differing neuromuscular control processes that regulate AP and ML postural sway have been called the ankle and hip mechanisms, respectively (Winter et al., 1993). It was suggested that the ankle and hip mechanisms were separately and independently regulated by the postural control system because they were executed orthogonally and with very little overlap during normal bipedal stance (Winter, 1995, Winter et al., 1996). A consistent finding in the literature is that older adults high falls-risk have pronounced declines in their ML postural stability, which is presumably due to reduced control of the abductor/adductor muscles, reduced torque generating capacity, or different response strategies (Maki et al., 1994, Nitz et al., 2003, Rogers and Mille, 2003). Examination of the abilities of high falls-risk older adults to perform
ML voluntary postural sway may therefore provide novel insights regarding ML instability. Given that most voluntary sway studies have averaged over sway directions (Borah et al., 2007), or examined only the AP direction (Blaszczyk et al., 1993b, Liaw et al., 2009, Stelmach et al., 1989), little is currently known about whether the ability of balance impaired older adults to perform voluntary postural sway depends on the direction of movement.

The regulation of standing stability by the postural control system is characterised by postural shifts of the centre of pressure (COP) with respect to the horizontal position of the centre of mass (COM) (Winter, 1995). However, most studies examining the stability of voluntary postural sway movements have reported the excursions of the COP alone. Investigation of the underlying control processes that regulate postural stability during voluntary postural sway movements may therefore benefit from a combined interpretation of COP and COM motions (Tucker et al., 2009). One such measure is the separation distance between the COP and COM (COP-COM), which is directly proportional to horizontal acceleration of the COM (Corriveau et al., 2004b, Masani et al., 2007, Winter, 1995, Yu et al., 2008). During reactive transitions of voluntary postural sway between the AP and ML directions, ageing and increased falls-risk are associated with reduced COP-COM separation and a reduced capacity to accelerate the COM (Tucker et al., 2009). Walking and obstacle crossing studies have also found that older adults with balance disorders exhibit greater ML instability as identified by their greater ML COP-COM separation in the non-target direction of movement compared with matched controls (Chou et al., 2004, Lee and Chou, 2006). Therefore, the measurement of COP-COM separation in the target and non-target directions during voluntary postural sway may provide insights into the decline in postural stability that occurs with the normal ageing process.

An individual’s reaction time to a balance perturbation is also an important determinant of their capacity to maintain postural stability during everyday activities (Stelmach and Worringham, 1985). For example, if an effective postural response is not initiated and executed within the available response time following contact with a
balance hazard, a fall may be more likely to occur. The majority of studies investigating reaction time during voluntary postural sway movements have examined age-related differences in the ability to initiate movement in a specific direction as rapidly as possible (Liaw et al., 2009, Nitz et al., 2003). A complementary ability to rapidly terminate body movement is an essential movement skill that has received much less research attention despite the potential importance of such responses in avoiding falls (Menant et al., 2009, Cao et al., 1998a, Tirosh and Sparrow, 2004).

The aim of this study was to determine whether the reaction time to rapidly initiate and terminate voluntary postural sway and COP-COM separation during the initiation and performance of voluntary postural sway were significantly different between young adults, low falls-risk older adults, and high falls-risk older adults. It was also of interest to determine which combination of balance measures, sway tasks, and movement directions would best differentiate between the young adults, low falls-risk older adults and high falls-risk older adults. It was hypothesised that: (1) the high falls-risk older adults and low falls-risk older adults would exhibit slower reaction time, reduced amplitudes of AP and ML COP-COM separation and increased non-target postural sway compared with the young adults, (2) the high falls-risk older adults would exhibit slowed reaction time, reduced ML COP-COM separation and increased non-target postural sway compared with the low falls-risk older adults, and (3) the voluntary postural sway measures assessed would accurately predict the young adults, low falls-risk older adults, and high falls-risk older adults.

5.2 Methods

5.2.1 Participants

Twenty-five young adults (age range: 19-35 years, 52% male), 33 low falls-risk older adults (age range: 67-84 years, 52% male) and 15 high falls-risk older adults (age range: 65-84 years, 52% male) were recruited from Griffith University Gold Coast campus and the surrounding community. Volunteers were excluded if they reported neurological, cognitive or proprioceptive disorders and recent or recurrent history of
musculoskeletal injury and/or surgery. All participants provided written informed consent prior to testing. The study was approved by the Institutional Human Research Ethics Committee and all guidelines of the Committee were followed during the experimental procedures.

5.2.2 Falls-risk assessment

The falls-risk of the older adult participants was calculated using the long-form Physiological Profile Assessment (PPA) (Lord et al., 2003). In addition, PPA data was also collected from 12 of the 25 young adult participants. Tests of vision, sensation, leg strength, reaction time, postural sway, and dynamic balance provided an overall falls-risk score that ranged from -2 (very low falls-risk) to 4 (very marked falls-risk). A falls-risk score of 1 was used to categorise older individuals as either low falls-risk (< 1) or high falls-risk (≥ 1) (St George et al., 2007). The PPA has been validated on over 2,000 adults and predicts multiple faller and non-multiple faller individuals with 75% accuracy in community and institutional settings (Lord et al., 2003).

5.2.3 Voluntary postural sway tasks

The experimental procedures used in this study were similar to our previous studies of voluntary postural sway (Tucker et al., 2008, Tucker et al., 2009). Participants were instructed to sway primarily using their ankles with as little hip motion as possible for AP voluntary postural sway, and to sequentially load and unload the limbs for ML voluntary postural sway (Winter, 1995, Winter et al., 1996). These AP and ML voluntary postural sway movements were performed in separate trials. For the sway initiation task, participants were required to rapidly initiate voluntary postural sway in response to a two-choice auditory cue following a random period of quiet stance of between 5 and 10 s. ‘Forward’ or ‘backward’ cues were presented for AP sway trials, and ‘left’ or ‘right’ cues were presented for ML sway trials. Participants were instructed to react and to move as quickly as possible in the direction indicated by the auditory cue.
Following sway initiation, participants immediately commenced continuous voluntary postural sway at their preferred frequency. In 50% of trials, a “stop” auditory cue was presented after the participant had completed a randomly selected number of continuous voluntary sway oscillations (i.e., 2.0, 2.5, 3.0, 3.5, 4.0, or 4.5 cycles). The stop cue was manually presented by the principal investigator (M.G.T.) when the participant’s body lean was visually estimated as close to neutral stance during the subsequent sway oscillation. The other 50% of trials in which a stop cue was not presented were ‘catch’ trials in which the participant swayed until they had completed 5.0 continuous voluntary postural sway oscillations. When presented with a stop cue, participants were required to terminate voluntary sway as rapidly as possible and then return to quiet stance. The instantaneous amplitude of the COP in the direction of sway when the stop cue was presented was recorded for all termination trials.

5.2.4 Instrumentation

COP data were collected with a multicomponent force plate (Type 9287A, Kistler Instrument Corporation). COM displacement was calculated with 3D motion analysis, using 14 mm diameter retro-reflective markers that were attached according to the full body VICON Plug-in-Gait model (Oxford Metrics Group Plc.) (Figure 5.1a-c). Marker trajectories and force plate data were synchronised and collected using Nexus software version 1.3 (Oxford Metrics Group) at sampling frequencies of 100 and 1000 Hz, respectively. The force data was collected at 1000 Hz so that reaction times could be measured with precision of milliseconds. Participants were fitted with a light-weight harness that was secured to the roof of the laboratory via a safety line. The harness and safety line were adjusting prior to testing to ensure that voluntary postural sway movements were not restricted in any way but impact with the ground would be prevented if a fall occurred (Figure 5.1a-b).
Figure 5.1. Marker placement, biomechanical modelling, and representative data. (a-b) Anterior and posterior views of the retro-reflective marker locations, the harness and safety line, and the measured stance position on the force plate for an older participant. (c) VICON Plug-in-Gait model of a participant during AP voluntary sway that shows the 3D orientation of each segment, the COM (shaded circle), and the vertical ground reaction force (white arrow). (d) Normalised amplitudes of the COP, COM, and COP-COM separation in the target AP and non-target ML directions for a high falls-risk older adult during quiet stance, sway initiation, continuous voluntary sway, and sway termination. The participant was presented with a “backward” cue to initiate AP sway, and after a period of continuous voluntary sway, was presented a “stop” cue to terminate AP sway. The instants of cue presentation are indicated by the dashed vertical lines and the ends of the initiation and termination reaction times (RT) are indicated by the solid vertical lines.
5.2.5 Experimental design

Foot position on the force plate was standardised so that stance width was equal to 10% of the participant’s height and the outward foot angle was 15° (McIlroy and Maki, 1997). Prior to testing, the position of the T10 marker was recorded during maximum static leans in the forward and backward (AP range), and left and right (ML range) directions. Biofeedback reference points were then implemented using Nexus software (Oxford Metrics Group) so that movement of T10 was standardised to the middle 60% of the AP and ML ranges during testing (Tucker et al., 2009). The initiation and stop auditory cues were counterbalanced for each participant. Following two practice trials, five experimental trials were collected for each sway initiation direction (i.e., forward, backward, left, and right) for a total of 10 AP trials and 10 ML trials. Trials in which participants responded in the incorrect direction during the initiation of sway, or lost their balance and/or stepped during all sway tasks were repeated.

5.2.6 Data analysis

COM data were up-sampled to 1000 Hz using cubic spline interpolation, and then the COP and COM data were filtered with a 4th order, zero phase shift, band-pass Butterworth filter. For both the COP and COM signals, the cut-off frequencies of the filter were 0.1 and 10 Hz. This filter removed high frequency noise and also de-trended the COP and COM signals to ensure that their trajectories accurately reflected the AP and ML directions of postural sway. Prior to the calculation of the dependent measures, COP and COM data were normalised for each participant using maximum COP amplitudes that were obtained from two trials of maximum dynamic voluntary AP and ML sway. The COP, COM, and COP-COM values reported in the text are in normalised units. Data analysis was performed using Matlab version 7.6 (Release 2008a, The Mathworks Inc.).

Initiation reaction time was calculated as the period from the auditory cue until the first observable change in COP amplitude associated with sway initiation. The first observable change of COP was taken as the point where the COP amplitude exceeded
2 SDs of baseline, which was calculated from the 3 s period immediately prior to the auditory cue (Tucker et al., 2008). The sway initiation phase was the period from the end of the initiation reaction time to the response peak of COP displacement in the direction indicated by the auditory cue. The continuous voluntary sway phase was the period from the zero-crossing of COP displacement immediately following the sway initiation phase to the zero-crossing of COP displacement immediately prior to a stop cue or the end of the trial when a stop cue was not presented. Termination reaction time was the period from the stop cue to the zero-crossing of COP velocity (i.e., change in COP direction) immediately following the cue (Buchanan and Horak, 2003).

For the initiation and continuous voluntary sway phases, root mean square (RMS) amplitude of the COP and COM was calculated using the method of Prieto et al (1996) in the target direction of voluntary sway and also in the non-target (orthogonal) sway direction. For example, during an AP sway trial, movement in the AP direction is target and movement in the ML direction is non-target. RMS amplitudes were used in this study because the method provides a global measure of signal amplitude that is suitable for periods of voluntary postural sway. The separation distance between COP and COM trajectories (COP-COM) was obtained by subtracting the COM from the COP, and then calculating the RMS of the data. It was also of interest to determine whether the groups had the same frequency and velocity of continuous voluntary sway as they may affect COP-COM separation and the ability to rapidly terminate sway. The frequency of the COP was computed during the continuous voluntary sway phase as the inverse of the average sway period. The average sway period was determined using auto-correlation analysis. COP velocity was obtained by differentiation of displacement data using three-point central difference equations, and then calculating the RMS of the data. A representative example of baseline quiet stance, initiation and termination reaction times, and the sway initiation and continuous voluntary sway phases in presented in Figure 5.1d.
5.2.7 Statistical analysis

One-factor Analysis of Variance (ANOVA) was used to test for main group effects (3 levels: young adults, low falls-risk older adults, high falls-risk older adults) in general characteristics and COP measures (see Table 5.1) and the initiation and termination reaction times. Two-factor ANOVA including group (3 levels: young adults, low falls-risk older adults, high falls-risk older adults) and sway direction (2 levels: target, non-target) factors was used to test for differences in COP-COM separation. Planned contrasts were used to identify specific differences between groups. Type 1 error rate was reduced by setting the significance level at 0.01. Effect sizes (ES) were reported as Cohen’s d (Cohen, 1988).

To develop predictive models of group status, three forward stepwise discriminant analyses were performed: (1) young adults versus low falls-risk older adults, (2) young adults versus high falls-risk older adults, and (3) low versus high falls-risk older adults. Reaction time and COP-COM variables that were identified as significantly different between the groups were used as predictor variables in the discriminant analyses. A probability to enter criterion of $P < .15$ was used to ensure that potentially important discriminators entered each model (Constanza and Afifi, 1979). In the stepwise procedure, the variable that was the strongest discriminator between groups, as measured by the Wilks’ Lambda statistic, was selected in the first step. In subsequent steps, the variable that explained most of the remaining between-groups variance and significantly improved the model was selected. This method ensured that the variables that provided the maximum combined between-groups discrimination were selected (Klecka, 1980).

The predictive capabilities of the selected sway variables were evaluated using a generalised squared-distance classification procedure, which was adjusted for unequal group sizes (Klecka, 1980). The percentage of individuals that were correctly classified from each group (i.e., sensitivity and specificity scores) were compared to the percentage of individuals that could be correctly classified due to group size proportions (prior probability) using Z tests for binomial percentages. Positive and
negative likelihood ratios were also computed which indicate the odds that an individual belongs to a particular group following a particular classification result (Deeks and Altman, 2004). Variables that were not normally distributed were log-transformed prior to discriminant analysis. Statistical analyses were performed using SAS for Windows version 9.1 (SAS Institute Inc.).

5.3 Results

5.3.1 Group differences in general characteristics and COP measures

Significant main group effects were detected for age, PPA score, and COP amplitudes during maximum voluntary sway and continuous voluntary sway (Table 5.1). The young adults were significantly younger compared with the low falls-risk older adults ($F_{1,70} = 1518.49, P < .001, ES = 10.86$) and the high falls-risk older adults ($F_{1,70} = 1138.94, P < .001, ES = 11.25$) and also had a lower PPA score (decreased falls-risk) compared with the low falls-risk older adults ($F_{1,57} = 35.69, P < .001, ES = 2.38$) and high falls-risk older adults ($F_{1,57} = 148.74, P < .001, ES = 6.05$). Low falls-risk older adults also had a lower PPA score compared with the high falls-risk older adults ($F_{1,57} = 75.70, P < .001, ES = 2.71$). For the COP measures, the young adults had increased COP amplitude during ML maximum voluntary sway ($F_{1,70} = 7.37, P < .01, ES = 0.66$) and AP continuous voluntary sway ($F_{1,70} = 8.91, P < .01, ES = 0.76$) compared with low falls-risk older adults. The young adults also had increased COP amplitude during AP maximum voluntary sway ($F_{1,70} = 9.44, P < .01, ES = 1.05$), ML maximum voluntary sway ($F_{1,70} = 15.28, P < .001, ES = 1.22$), and AP continuous voluntary sway ($F_{1,70} = 8.63, P < .01, ES = 0.93$) compared with the high falls-risk older adults.

5.3.2 Group differences in initiation and termination reaction times

Significant main group effects were detected in reaction times for the initiation of AP sway ($F_{2,70} = 16.58, P < .001$, Figure 5.2a), initiation of ML sway ($F_{2,70} = 10.76, P < .001$, Figure 5.2b), termination of AP sway ($F_{2,70} = 11.79, P < .001$, Figure 5.2c), and termination of ML sway ($F_{2,70} = 12.20, P < .001$, Figure 5.2d). High falls-risk older adults had significantly slower reaction times compared with the young adults for initiation
Table 5.1
General characteristics and COP measures for the young adults (YA), low falls-risk older adults (LFROA), and high falls-risk older adults (HFROA).

<table>
<thead>
<tr>
<th>Variable</th>
<th>YA (n = 25)</th>
<th>LFROA (n = 33)</th>
<th>HFROA (n = 15)</th>
<th>F, P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>25 ± 4</td>
<td>74 ± 5*</td>
<td>77 ± 5*</td>
<td>$F_{2,70} = 916.89, P &lt; .001$</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>173 ± 11</td>
<td>167 ± 10</td>
<td>165 ± 13</td>
<td>$F_{2,70} = 3.21, P = .046$</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>71 ± 15</td>
<td>82 ± 20</td>
<td>77 ± 19</td>
<td>$F_{2,70} = 2.69, P = .075$</td>
</tr>
<tr>
<td>PPA score</td>
<td>-0.6 ± 0.4†</td>
<td>0.3 ± 0.5*</td>
<td>1.6 ± 0.5*†</td>
<td>$F_{2,57} = 77.13, P &lt; .001$</td>
</tr>
<tr>
<td>Incorrect responses (% trials)</td>
<td>1.0 ± 2.0</td>
<td>2.0 ± 3.9</td>
<td>2.1 ± 4.0</td>
<td>$F_{2,70} = 0.87, P = .422$</td>
</tr>
<tr>
<td>Balance loss (% trials)</td>
<td>1.0 ± 2.5</td>
<td>0.3 ± 1.2</td>
<td>1.2 ± 2.6</td>
<td>$F_{2,70} = 1.37, P = .262$</td>
</tr>
<tr>
<td>AP max sway range (mm)</td>
<td>233 ± 89</td>
<td>194 ± 77</td>
<td>158 ± 30*</td>
<td>$F_{2,70} = 4.92, P = .010$</td>
</tr>
<tr>
<td>ML max sway range (mm)</td>
<td>394 ± 170</td>
<td>302 ± 116*</td>
<td>230 ± 46*</td>
<td>$F_{2,70} = 8.16, P &lt; .001$</td>
</tr>
<tr>
<td>AP CVS amplitude (% max)</td>
<td>82.9 ± 20.9</td>
<td>69.4 ± 15.7*</td>
<td>66.5 ± 11.9*</td>
<td>$F_{2,70} = 6.00, P = .004$</td>
</tr>
<tr>
<td>ML CVS amplitude (% max)</td>
<td>72.4 ± 12.1</td>
<td>75.3 ± 12.4</td>
<td>63.5 ± 11.4</td>
<td>$F_{2,70} = 0.41, P = .669$</td>
</tr>
<tr>
<td>AP CVS velocity (units/s)</td>
<td>2.08 ± 0.68</td>
<td>1.94 ± 0.64</td>
<td>1.63 ± 0.40</td>
<td>$F_{2,70} = 2.48, P = .090$</td>
</tr>
<tr>
<td>ML CVS velocity (units/s)</td>
<td>1.74 ± 0.55</td>
<td>1.88 ± 0.59</td>
<td>1.59 ± 0.36</td>
<td>$F_{2,70} = 2.71, P = .071$</td>
</tr>
<tr>
<td>AP sway frequency (Hz)</td>
<td>0.31 ± 0.05</td>
<td>0.33 ± 0.05</td>
<td>0.30 ± 0.04</td>
<td>$F_{2,70} = 2.61, P = .085$</td>
</tr>
<tr>
<td>ML sway frequency (Hz)</td>
<td>0.33 ± 0.06</td>
<td>0.34 ± 0.05</td>
<td>0.30 ± 0.04</td>
<td>$F_{2,70} = 1.99, P = .144$</td>
</tr>
</tbody>
</table>

Values are means ± standard deviation. CVS = continuous voluntary sway.
+Low falls-risk older adults significantly different compared with young adults, $P < .01$.
*High falls-risk older adults significantly different compared with young adults, $P < .01$.
†High falls-risk older adults significantly different compared with low falls-risk older adults, $P < .01$.
‡PPA score for young adults was measured from 12 of the 25 participants.

of AP sway ($F_{1,70} = 33.15, P < .001, ES = 1.73$), initiation of ML sway ($F_{1,70} = 16.77, P < .001, ES = 1.55$), termination of AP sway ($F_{1,70} = 9.06, P < .01, ES = 1.35$), and termination of ML sway ($F_{1,70} = 14.75, P < .001, ES = 1.56$). High falls-risk older adults also had significantly slower reaction times compared with the low falls-risk older adults for the initiation of AP sway ($F_{1,70} = 14.13, P < .001, ES = 1.00$), initiation of ML sway ($F_{1,70} = 9.88, P < .01, ES = 0.86$), termination of AP sway ($F_{1,70} = 21.62, P < .001, ES = 1.37$), and termination of ML sway ($F_{1,70} = 16.89, P < .001, ES = 1.32$). Low falls-risk older adults had significantly slower reaction time compared with the young adults for the initiation of AP sway ($F_{1,70} = 7.17, P < .01, ES = 1.05$).
Figure 5.2. Reaction times of the young adults (YA), low falls-risk older adults (LFROA), and high falls-risk older adults (HFROA). (a) Initiation of AP sway, (b) Initiation of ML sway, (c) Termination of AP sway, and (d) Termination of ML sway. Values are mean ± standard error. +LFROA significantly different to YA, $P < .01$, *HFROA significantly different to YA, $P < .01$. †HFROA significantly different to LFROA, $P < .01$.

5.3.3 Group differences in COP-COM separation for sway initiation and continuous voluntary sway tasks

Significant group-by-sway direction interactions were found for COP-COM separation during the initiation of AP sway ($F_{2,70} = 13.21$, $P < .001$, Figure 5.3a), AP continuous voluntary sway ($F_{2,70} = 10.74$, $P < .001$, Figure 5.3b), and ML continuous voluntary sway
(F_{2,70} = 4.97, P < .01, Figure 5.3d). For the initiation of AP sway, the young adults had greater target AP COP-COM separation compared with the low falls-risk older adults (F_{1,68} = 29.36, P < .001, ES = 1.03) and the high falls-risk older adults (F_{1,68} = 22.33, P < .001, ES = 1.30). For AP continuous voluntary sway, the young adults had greater target AP COP-COM separation compared with the low falls-risk older adults (F_{1,68} = 15.42, P < .001, ES = 0.71) and high falls-risk older adults (F_{1,68} = 30.65, P < .001, ES = 1.22). For ML continuous voluntary sway, the high falls-risk older adults had reduced target ML COP-COM separation compared with the young adults (F_{1,68} = 13.88, P < .001, ES = 0.67) and the low falls-risk older adults (F_{1,68} = 14.04, P < .001, ES = 0.86). For the initiation of ML sway (Figure 5.3c), the group-by sway direction interaction was not significant, however the planned contrasts revealed significantly reduced target ML COP-COM separation for the low falls-risk older adults (F_{1,68} = 7.21, P < .01, ES = 0.57) and high falls-risk older adults (F_{1,68} = 15.31, P < .001, ES = 1.05) compared with the young adults.

5.3.4 Predictive models of group status

A summary of the selected variables, predictive capabilities, and prior probabilities for the three discriminant analyses are displayed in Table 5.2. The young adults and low falls-risk older adults were significantly discriminated by two variables that were both related to the rapid initiation of AP sway (Wilks’ Lambda = 0.72, P < .001). Two reaction time measures and one COP-COM measure were selected as significant discriminators between the young adults and the high falls-risk older adults (Wilks’ Lambda = 0.31, P < .001). This discriminant model produced the best overall classification accuracies and likelihood ratios. Two termination reaction time measures and target ML COP-COM separation during continuous voluntary sway significantly discriminated between the low and high falls-risk older adults (Wilks’ Lambda = 0.59, P < .001). Sensitivities and specificities ranged from 72% to 100% and were significantly greater than their corresponding prior probabilities (P < .01).
Figure 5.3. COP-COM separation amplitudes in the target and non-target sway directions for the young adults (YA), low falls-risk older adults (LFROA) and high falls-risk older adults (HFROA) during sway initiation and continuous voluntary sway (CVS) in the AP and ML directions. (a) Initiation of AP sway, (b) AP CVS, (c) Initiation of ML sway, and (d) ML CVS. Values are mean ± standard error in normalised units. +LFROA significantly different to YA, \( P < .01 \). *HFROA significantly different to YA, \( P < .01 \). †HFROA significantly different to LFROA, \( P < .01 \).
Table 5.2
Results of the discriminant analyses between the young adults (YA), low falls-risk older adults (LFROA) and high falls-risk older adults (HFROA).

<table>
<thead>
<tr>
<th>Selected Variables</th>
<th>YA vs. LFROA</th>
<th>YA vs. HFROA</th>
<th>LFROA vs. HFROA</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>Selected Variables</strong></td>
<td>1. Target AP COP-COM to initiate AP sway</td>
<td>1. Reaction time to initiate AP sway</td>
<td>1. Reaction time to terminate AP sway</td>
</tr>
<tr>
<td></td>
<td>2. Reaction time to initiate AP sway</td>
<td>2. Reaction time to terminate ML sway</td>
<td>2. Reaction time to terminate ML sway</td>
</tr>
<tr>
<td></td>
<td>3. Target AP COP-COM to initiate AP sway</td>
<td></td>
<td>3. Target ML COP-COM during ML continuous voluntary sway</td>
</tr>
<tr>
<td><strong>Sensitivity</strong></td>
<td>LFROA: 76%</td>
<td>HFROA: 100%</td>
<td>HFROA: 73%</td>
</tr>
<tr>
<td><strong>Specificity</strong></td>
<td>YA: 72%</td>
<td>YA: 88%</td>
<td>LFROA: 88%</td>
</tr>
<tr>
<td><strong>Likelihood Ratio (+)</strong></td>
<td>2.71</td>
<td>8.83</td>
<td>6.05</td>
</tr>
<tr>
<td><strong>Likelihood Ratio (-)</strong></td>
<td>0.34</td>
<td>0.00*</td>
<td>0.30</td>
</tr>
<tr>
<td><strong>Prior Probability</strong></td>
<td>YA: 43%</td>
<td>YA: 62%</td>
<td>LFROA: 69%</td>
</tr>
<tr>
<td></td>
<td>LFROA: 57%</td>
<td>HFROA: 38%</td>
<td>HFROA: 31%</td>
</tr>
</tbody>
</table>

*High sensitivity but low negative likelihood ratio paradox.
5.4 Discussion

The objectives of this study were to determine whether reaction time and COP-COM measures associated with voluntary postural sway tasks could differentiate between groups of adults with differing age and falls-risk status, and also to determine which combination of balance measures, sway tasks, and movement directions best discriminated between the groups. The main findings were that older adults with high falls-risk compared with low falls-risk had particularly slowed reaction time to terminate voluntary sway, and decreased COP-COM separation when performing voluntary sway in the ML direction. The older groups compared with the young also had higher PPA scores, slower reaction times, and reduced COP-COM separation, however the specific differences in these measures for the voluntary postural sway tasks and directions was dependent on whether the older group was high or low falls-risk. The voluntary sway measures were also effective in predicting group status, with classification accuracies ranging from 72% to 100%.

5.4.1 Reaction time

The high falls-risk older adults had significantly slower reaction time to initiate and terminate voluntary postural sway compared with the young adults and low falls-risk older adults for all task conditions. These findings indicate that high falls-risk older adults with impaired sensorimotor function as measured by the PPA (Lord et al., 2003) had a corresponding reduction in the ability to respond to the reaction time stimulus during initiation and termination of voluntary postural sway in the AP and ML directions. While it is not possible from the findings of the present study to identify how specific sub-components of the task (i.e., stimulus detection, processing of stimulus, postural response) contributed to the observed group differences in reaction time, it is clear that the high falls-risk older adults had deficits in any one or combination of these subtasks when reacting to the auditory stimuli. Reaction time tasks requiring the initiation and termination of voluntary postural sway may therefore be a useful marker of increased falls-risk among community-dwelling older people.
The hypothesis that low falls-risk older adults would have slower voluntary postural sway reaction times compared with young adults was only partially supported by the findings of the study. The low falls-risk older adults had reaction times that were significantly slower compared with the young adults during the initiation of AP sway and reaction times that approached significance for the initiation of ML sway ($P = 0.047$). These findings are in general agreement with previous reports that ageing slows the reaction time of multidirectional leaning movements during the Limits of Stability task (Liaw et al., 2009, Nitz et al., 2003) and during rapid transitions of voluntary postural sway (Tucker et al., 2008, Tucker et al., 2009). A result contrary to one of the study hypotheses was that no significant differences in termination reaction time were observed between the young adults and low falls-risk older adults. Perhaps the main reason for this absence of significance was related to the manner by which termination reaction time was calculated. Termination reaction time was the period from the stop cue to the zero-crossing of COP velocity (i.e., change in COP direction) immediately following the cue (Buchanan and Horak, 2003). Factors with the potential to influence termination reaction time included the COP position and velocity at the time of the cue, and the amplitude of COP displacement during the reaction time period (i.e., from the stop cue to the zero-crossing of COP velocity). An additional analysis was performed to determine whether these factors were significantly different between groups. This analysis revealed no significant differences in COP position and velocity at the time of the cue between the young adults and low falls-risk older adults ($P > .01$). However, COP displacement during the reaction time period was significantly greater for the young adults compared with the low falls-risk older adults for termination of sway in the AP direction (Mean ± SD; young adults: 123 ± 32 mm; low falls-risk older adults: 89 ± 33 mm; $F_{1,70} = 7.12$, $P < .01$, $ES = 1.06$) and ML direction (Mean ± SD; young adults: 215 ± 53 mm; low falls-risk older adults: 172 ± 46 mm; $F_{1,70} = 8.39$, $P < .01$, $ES = 0.91$). Therefore, the additional analysis suggests that the main reason for the lack of difference in termination reaction time between the groups was that the young adults, who were expected to have a reduced termination reaction time, displaced their COP further during the reaction time period compared with the low falls-risk older adults.
5.4.2 COP-COM separation

As hypothesised, the low and high falls-risk older adults had reduced COP-COM separation compared with the young adults. For the low falls-risk older adults, the reduced COP-COM separation was observed during AP and ML sway initiation and AP continuous voluntary sway. The high falls-risk older adults had reduced COP-COM separation compared with the young for the sway initiation and continuous voluntary sway tasks in all directions. Collectively, these findings suggest that the older groups had a reduced capacity to shift the COP with respect to the COM during voluntary postural sway movements which decreased their COP-COM separation. As participants were instructed to react as quickly as possible, the smaller COP-COM separation during sway initiation was undesirable because it limited the acceleration of the COM in the initiation direction (Masani et al., 2007, Yu et al., 2008). Decreased COP-COM separation during continuous voluntary sway also limited the capacity to halt the trajectory of the COM at the end of a sway cycle (e.g., finish swaying forward) and reverse its direction of oscillation (e.g., begin swaying backward). These findings might indicate that the older adults had reduced postural stability during voluntary postural sway tasks due to underlying sensorimotor deficits in controlling COM motions (Corriveau et al., 2000). However, as participants were only required to sway to 60% of maximum lean amplitude, it may also be argued that the reduced COP-COM separation could represent a conservative sway strategy by the older adults aimed at reducing COM acceleration near the limits of stability (Hahn and Chou, 2004).

The findings of the study also supported the hypothesis that high falls-risk older adults would have reduced ML COP-COM separation compared with the low falls-risk older adults during ML voluntary postural sway. The reduced ML COP-COM separation for the high falls-risk older adults suggests a reduced capacity to control accelerations of the ML COM (Masani et al., 2007, Yu et al., 2008). The current findings also add to the growing literature that older adults with an increased susceptibility to falls have deficits in ML postural control as identified by altered ML COP-COM separation (Chou et al., 2003, Chou et al., 2004). As participants performed the ML sway movements via sequential loading and unloading of each limb, the underlying cause of these findings
may be deterioration in neuromuscular control of hip abductor-adductors and/or ankle inverter-everters, or an altered lateral sway strategy (Rogers and Mille, 2003, Winter et al., 1996).

No significant differences in non-target COP-COM separation were observed between the young adults, low falls-risk older adults and high falls-risk older adults during the voluntary postural sway tasks. Contrary to the current findings, greater non-target ML COP-COM separation has been previously reported for traumatic brain injury patients and older adults with balance impairments and a history of falls during level walking and obstacle crossing compared with matched controls (Chou et al., 2004, Lee and Chou, 2006). Therefore, increased non-target COP-COM separation may be more readily observed during more dynamically challenging balance tasks or in individuals with neurological disorders. The lower COP-COM separation for the non-target direction compared with the target direction indicates that all three groups successfully restricted the accelerations of the COM to the target direction of motion. The current results also demonstrate that group differences in COP-COM separation were most prominent in the target direction of sway, which is also more important from a stability perspective as the greatest threat to loss of standing balance was in the direction of voluntary sway.

5.4.3 Prediction of group status

The classification accuracies and likelihood ratios of the discriminant models indicated that the reaction time and COP-COM separation measures had useful predictive capacities for group status with sensitivities and specificities ranging from 72% to 100%. The models with the strongest associations with group status were high falls-risk older adults versus young adults (sensitivity: 100%, specificity: 88%) and high versus low falls-risk older adults (sensitivity: 73%, specificity: 88%). These results demonstrate that the high falls-risk older adults were accurately identified from the voluntary postural sway measures assessed in this study. The elevated PPA scores of the high falls-risk older adults also suggests that their reduced capacity to perform voluntary postural sway tasks was due to sensorimotor impairment (Lord et al., 2003). The
discriminant model for young adults versus low falls-risk older adults was not as strongly predictive of group status (sensitivity: 76%, specificity: 72%), presumably because of less severe declines in sensorimotor function associated with a lower risk of falling.

The strongest predictors of high versus low falls-risk status out of all the task measures assessed were the reaction times to terminate AP and ML voluntary sway. Therefore, a prominent finding of the study is that a reduced capacity to rapidly terminate body movement is strongly associated with PPA defined classifications of increased falls-risk amongst community-dwelling older adults. Moreover, slower termination reaction time was found to be a stronger and more important predictor of high falls-risk compared with slower initiation reaction time. These novel findings suggest that future research should examine the abilities of high and low falls older adults to rapidly terminate voluntary body movement, especially as reaction time research on fallers and non-fallers has almost exclusively focussed on movement initiation tasks. Indications of slower termination responses by older adults compared with the young have been reported in the literature (Cao et al., 1998a, Menant et al., 2009, Tirosh and Sparrow, 2004), however it is currently unclear whether a reduced capacity to terminate voluntary body movement could be a causative factor in falls.

With regards to the influence of movement direction on group differences in voluntary postural sway performance, it was found that reduced COP-COM separation during continuous voluntary sway in the ML direction was an important predictor of high versus low falls-risk. In contrast, the best predictors of the low falls-risk older adults versus the young adults were target COP-COM separation and reaction time during rapid initiation of voluntary sway in the AP direction. Although the low falls-risk older adults also exhibited reduced target COP-COM separation during initiation of ML sway compared with the young adults, this measure was not a significant and independent predictor of group status. High falls-risk older adults were most accurately differentiated from the young adults by reaction time and COP-COM separation task measures in both the AP and ML sway directions. Taken together, these findings
suggest that deterioration in voluntary postural sway movements was most prominent in the AP direction for the low falls-risk older adults compared with the young adults. An extra level of reduced control was apparent in the ML direction for older adults with high risk of falling compared with low risk of falling. Given that there are separate mechanisms underlying voluntary postural sway in the AP and ML directions (Winter et al., 1996, Winter et al., 1993), the overall group differences in voluntary postural sway performance are likely to have emerged from reduced neuromuscular control at the ankles and hips.

5.5 Conclusion

The current findings demonstrate that increased age and falls-risk result in deterioration in the speed and stability of postural responses associated with voluntary postural sway movements. Most notably, the time taken to terminate voluntary postural sway was a highly specific indicator of high versus low falls-risk older adults, and therefore may be an important marker of falls-risk. The high falls-risk older adults also had reduced separation between motions of the COP and COM in the target direction during all sway tasks compared with young adults, and during continuous ML sway compared with low falls-risk older adults. The low falls-risk older adults also had reduced target COP-COM separation compared with young adults primarily when swaying in the AP direction. These findings suggest that increasing age and falls-risk results in reduced postural stability during voluntary postural sway actions because of a reduced capacity to regulate COM motions. The overall results indicate that voluntary postural sway is a useful model for examining the biomechanical mechanisms underlying declines in postural stability with ageing and increased falls-risk.
What are the relationships between voluntary postural sway measures and falls-history status in community-dwelling older adults?

6.1 Introduction

The normal ageing process increases the susceptibility of older persons to falls. For community-living older adults 65 years and older, approximately one in three individuals fall at least once a year and one in ten individuals fall on multiple occasions (Lord et al., 1993, Hill et al., 2004). Given this high rate of falls, a major focus of falls-related research has been on the development of approaches to predict an individual’s susceptibility to falls (Lord et al., 2003, Perell et al., 2001, Scott et al., 2007). However, this is not a trivial undertaking because over 400 potential risk factors have been identified with falls (Oliver et al., 1997). For many older individuals, impaired balance emerges as the biggest falls risk factor (Close, 2005b). Epidemiological studies have found that most falls result from an inability to rapidly recover from a loss of balance during daily activities (Lord et al., 1993, Bradley and Pointer, 2009). Therefore, a key element in the ability to avoid a fall is the capacity to react quickly and effectively with the body (Lord and Fitzpatrick, 2001, Maki and McIlroy, 2006), a task crucially dependent on sensory feedback, muscular strength, motor coordination, and reaction time (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Simple quantitative assessments that emphasised speed of response during voluntary multidirectional body movements have accurately identified older individuals with increased falls-risk (Lord and Fitzpatrick, 2001, Dite and Temple, 2002, Melzer et al., 2007).
Voluntary postural sway movements represent a simple approach to examining deficits in postural control which may contribute to falls. The three main categories of voluntary postural sway tasks include maximum voluntary leans held statically, continuous steady-state voluntary postural sway, and rapidly initiated voluntary postural sway movements performed under reaction time conditions. Maximum static lean amplitudes (cf. Limits of Stability and Functional Reach tests) have been found to be weakly associated or non-predictive of falls in healthy older adults (Boulgarides et al., 2003, Brauer et al., 2000, Melzer et al., 2004, Thomas and Lane, 2005, Wallmann, 2001). In contrast, reactive voluntary postural sway movements significantly differentiate between younger and older men (Tucker et al., 2008), and between young adults, older adults at low risk of falls, and older adults at high risk of falls (Tucker et al., 2009). For rapid orthogonal switches of voluntary postural sway between the anterior-posterior (AP) and medial-lateral (ML) directions, ageing and elevated falls-risk among the elderly results in a slower reaction time, reduced centre of pressure (COP) response amplitudes, and altered postural coordination (Tucker et al., 2008, Tucker et al., 2009). Despite deterioration in the reaction time and coordination of voluntary postural sway movements with ageing and increased falls-risk, there has been no comprehensive investigation of associations between different types of voluntary postural sway tasks and falls-history status. Given that falls are strongly related to declines in postural control physiology (Lord and Sturmsieks, 2005), examination of sensorimotor and balance function and falls-risk using the Physiological Profile Assessment (PPA) may improve understanding of differences in voluntary postural sway performance among older adults. The PPA is a valid and reliable predictor of falls among older adults as it differentiates between multiple fallers and non-multiple fallers with 75% accuracy (Lord et al., 2003). As voluntary postural sway actions are likely to require significant contributions from the cognitive (Lajoie et al., 1993), sensory (Kavounoudias et al., 1998), and muscular systems (Melzer et al., 2009a), reaction time and COP measures of voluntary postural sway performance may provide similar information on falls-risk compared with the PPA.
The general objective of this study was not to develop a new clinical test of falls-risk but rather to examine the factors related to the performance of voluntary postural sway that may contribute to falls-history status. The specific aims of this study were to examine the effect of falls-history status on the performance of voluntary postural sway tasks as assessed by reaction time and COP amplitude measures, and also to examine the strength of relationship between these measures and PPA score. It was hypothesised that: (1) multiple fallers would have significantly slower reaction time and reduced COP amplitude compared with non-fallers, (2) voluntary postural sway measures would significantly identify the falls-history status of older adults with comparable accuracy to PPA score, and (3) the tasks with the best capacity for discriminating falls-history status would involve a reaction response rather than a static lean movement.

6.2 Methods

6.2.1 Participants

Fifty-one older adults aged 65 to 94 years were recruited from retirement villages and independent housing within the local community. Participants were excluded if they reported neurological, cognitive or proprioceptive disorders and recent or recurrent history of musculoskeletal injury, surgery, or both. All participants provided written informed consent prior to testing. The guidelines of the Institutional Human Research Ethics Committee were followed during all experimental procedures.

6.2.2 Experimental design and protocol

Participants attended the laboratory on two occasions. All data were collected by the primary author (M.G.T.). On the first visit, participants completed a questionnaire regarding their medical history and falls history. A fall was defined as an event within the previous 12 months during activities of daily living which resulted in a person coming to rest unintentionally on the ground or other lower level, and not as the result of a major intrinsic event or an overwhelming hazard. Individuals reporting no falls were classified as non-fallers, individuals reporting one fall were classified as single
fallers, and individuals reporting two or more falls were classified as multiple fallers. After measurement of participant characteristics (e.g., age, sex, height, mass), a fall-risk assessment was performed using the short-form PPA. Briefly, this assessment included tests of visual contrast sensitivity, knee joint proprioception, leg extension strength of the dominant limb, visual reaction time using a hand-press response, and amplitude of postural sway whilst standing for 30 s on a rubber foam mat (Lord et al., 2003). Scores from each test were combined to yield a falls-risk score that ranged from -2 (very low falls-risk) to 4 (very marked falls-risk) (Lord et al., 2003).

On the second visit, participants underwent voluntary postural sway testing. The voluntary postural sway assessments have been described previously (Tucker et al., 2008, Tucker et al., 2009). Initially, participants were fitted with a lightweight, non-restrictive safety harness that was secured to the roof of the laboratory using a safety line. Participants then stood on a Kistler Type 9287A force plate, which was used to collect the COP in the AP and ML directions at 1000 Hz. The following six postural reaction and voluntary postural sway tasks were assessed: (1) maximum static leans — without losing balance, participants leaned as far as they could in the forward, backward, left and right directions and held their maximum lean position for 4 s (Figure 6.1a); (2) maximum voluntary sway — participants swayed maximally for three continuous cycles in the AP direction and then the ML direction (Figure 6.1b); (3) initiation of voluntary sway — participants initiated AP or ML voluntary postural sway as rapidly as possible in response to an auditory cue following a random 5 to 10 s period of quiet stance (Figure 6.1c); (4) orthogonal switches of voluntary sway — following 2 to 5 oscillations of voluntary postural sway, participants switched their sway to the orthogonal direction as rapidly as possible in response to an auditory cue (e.g., sway in the AP direction before rapidly switching sway to the ML direction, Figure 6.1d); (5) termination of voluntary sway — participants terminated AP or ML voluntary postural sway as rapidly as possible in response to a “stop” auditory cue following 2 to 5 oscillations of sway (Figure 6.1e); and (6) continuous voluntary sway — participants performed 4 to 5 oscillations of AP or ML voluntary postural sway at their preferred frequency between the initiation and termination tasks (Figure 6.1f).
Figure 6.1. Sample COP data from the six voluntary postural sway tasks assessed in this study. (a) Maximum static leans, (b) Maximum voluntary sway, (c) Initiation of voluntary sway, (d) Orthogonal switches of voluntary sway, (e) Termination of voluntary sway, and (f) Continuous voluntary sway. For the initiation, termination, and orthogonal switch tasks, the dotted vertical line represents the onset of the auditory cue, and the dashed vertical line represents the participant’s reaction time to the stimulus.
Participants were instructed to restrict motion to the ankle joint during AP sway movements, and to sequentially load and unload each leg during ML sway movements (Tucker et al., 2008, Tucker et al., 2009). For the initiation and orthogonal switch tasks, participants were presented with a “forward” or “backward,” or a “left” or “right” auditory cue (i.e., a two-choice reaction time design). After a forward or backward reaction, participants immediately commenced and continued AP voluntary postural sway. After a left or right reaction, participants immediately commenced and continued ML voluntary postural sway. Two trials were collected for maximum voluntary sway and maximum static lean tasks, and 20 trials (10 AP and 10 ML) were collected for the initiation, orthogonal switch, and termination tasks. Trials in which participants lost their balance and stepped or responded in the incorrect direction were repeated.

6.2.3 Voluntary postural sway measures

Reaction times and COP response amplitudes were measured for the initiation, orthogonal switch, and termination tasks. Start reaction time (initiation task) and switch reaction time (orthogonal switch task) were the periods from the auditory cue until the COP exceeded a threshold of 2 SDs of baseline data, which were calculated 3 s before the cue (Tucker et al., 2009). Stop reaction time (termination task) was the period from the stop cue until the direction of COP oscillation reversed. Response amplitudes for sway initiation and orthogonal switches of sway were measured as the peak of AP or ML COP displacement in the direction indicated by the auditory cue. Termination response amplitude was the peak of AP or ML COP amplitude at the end of the stop reaction time period.

Ranges of COP displacement in the AP and ML directions were measured for the maximum static lean, maximum voluntary sway, and continuous voluntary sway tasks. For each task, a peak detection algorithm identified the peak forward, backward, left, and right COP amplitudes. The peak amplitudes were averaged for each direction, and then the average peak amplitudes were used to calculate the AP and ML COP ranges. The AP and ML ranges were normalised to the participant’s height and mass for the
maximum static lean and maximum voluntary sway tasks. All data analysis was performed using Matlab software version 7.6.0 (Release 2008a).

6.2.4 Statistical analysis

Analysis of Covariance (ANCOVA) was used to test for main group effects of falls-history status (3 levels: non-fallers, single fallers, multiple fallers) in participant characteristics and voluntary postural sway measures. Planned contrasts were used to identify specific between-group differences. Age was used as the covariate for all statistical tests. The relationships between the voluntary postural sway measures identified as significantly different between groups and the relationships between these voluntary sway measures and PPA score were examined using Pearson product-moment partial correlation coefficients adjusted for age. Compound measures for this analysis represented the average of that measure over the AP and ML directions.

Forwards stepwise discriminant analyses were performed with sensitivities, specificities, positive and negative likelihood ratios and classification coefficients provided for (1) age, (2) PPA score, (3) age and PPA score, (4) the first selected voluntary postural sway measure, and (5) all selected voluntary postural sway and age measures. Participants were divided into non-multiple fallers and multiple fallers for discriminant analysis because few significant differences were identified between the non-fallers and single fallers in the ANCOVA analysis. The significance of each analysis was tested using the Wilks’ lambda statistic where values closer to zero and one denote high and low between-group discrimination, respectively (Klecka, 1980). Statistical analyses were performed using SAS version 9.1 with significance accepted at $P < .05$.

6.3 Results

6.3.1 Participant characteristics of the falls groups

Out of the 51 older adults that participated in the study, 36 individuals (70.6%) reported no falls, 10 individuals (19.6%) reported falling once, and 5 individuals (9.8%)
reported falling on two or more occasions. The multiple fallers were significantly older and had a higher PPA score compared with the non-fallers and single fallers \( (P < .05) \) (Table 6.1). For the PPA subtests, the multiple fallers had significantly slower hand-press reaction time compared with the single fallers and non-fallers, and increased postural sway whilst standing on the rubber foam matt compared with the non-fallers \( (P < .05) \). All participants were able to complete the required number of trials following episodes of balance loss or incorrect responses.

### 6.3.2 Effect of falls-history status on voluntary postural sway measures

Multiple fallers had significantly slower switch reaction time during AP-ML and ML-AP orthogonal switches of voluntary sway, slower stop reaction time during termination of ML voluntary sway, and reduced COP amplitude during initiation of AP and ML voluntary sway and continuous voluntary AP and ML sway compared with non-fallers \( (P < .05) \) (Table 6.2). Multiple fallers also had slower start reaction time during initiation of AP voluntary sway and a higher percentage of incorrect responses and balance loss trials and compared with the single fallers and non-fallers \( (P < .05) \).

### Table 6.1

Participant characteristics of the non-fallers (NF), single fallers (SF), and multiple fallers (MF).

<table>
<thead>
<tr>
<th>Descriptive Measures</th>
<th>NF</th>
<th>SF</th>
<th>MF</th>
<th>( F, P ) value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (yr)</td>
<td>74 ± 5</td>
<td>75 ± 5</td>
<td>86 ± 7*†</td>
<td>( F_{2,47} = 10.48, P &lt; .001 )</td>
</tr>
<tr>
<td>Sex (% men)</td>
<td>56</td>
<td>50</td>
<td>40</td>
<td>NA</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>168 ± 11</td>
<td>164 ± 9</td>
<td>157 ± 13</td>
<td>( F_{2,47} = 2.84, P = .068 )</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>81 ± 20</td>
<td>79 ± 15</td>
<td>62 ± 16</td>
<td>( F_{2,47} = 2.33, P = .108 )</td>
</tr>
<tr>
<td>PPA score</td>
<td>0.61 ± 0.69</td>
<td>0.82 ± 0.60</td>
<td>2.31 ± 0.76*†</td>
<td>( F_{2,47} = 9.24, P &lt; .001 )</td>
</tr>
</tbody>
</table>

**PPA Subtests**

<table>
<thead>
<tr>
<th></th>
<th>NF</th>
<th>SF</th>
<th>MF</th>
<th>( F_{2,47} ) value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Visual contrast sensitivity (dB)</td>
<td>20 ± 1.4</td>
<td>20 ± 0.9</td>
<td>18 ± 2.2</td>
<td>( F_{2,47} = 1.66, P = .201 )</td>
</tr>
<tr>
<td>Knee proprioception (*)</td>
<td>2.0 ± 1.2</td>
<td>1.9 ± 1.0</td>
<td>2.8 ± 1.0</td>
<td>( F_{2,47} = 1.05, P = .376 )</td>
</tr>
<tr>
<td>Leg extension strength (kg)</td>
<td>40.6 ± 17.4</td>
<td>34.5 ± 9.0</td>
<td>20.3 ± 3.4</td>
<td>( F_{2,47} = 2.84, P = .069 )</td>
</tr>
<tr>
<td>Hand-press reaction time (ms)</td>
<td>227 ± 30</td>
<td>233 ± 38</td>
<td>305 ± 46*†</td>
<td>( F_{2,47} = 8.51, P &lt; .001 )</td>
</tr>
<tr>
<td>Sway on foam (mm)</td>
<td>154 ± 55</td>
<td>182 ± 79</td>
<td>255 ± 59*†</td>
<td>( F_{2,47} = 4.47, P = .017 )</td>
</tr>
</tbody>
</table>

Values are mean ± standard deviation unless otherwise indicated.

*Significantly different to non-fallers, \( P < .05 \).
†Multiple fallers significantly different to single fallers, \( P < .05 \).
NA = not available.
Table 6.2
Differences between non-fallers (NF), single fallers (SF), and multiple fallers (MF) in reaction time and COP amplitude measures for the voluntary postural sway tasks.

<table>
<thead>
<tr>
<th>Voluntary Sway Task</th>
<th>NF</th>
<th>SF</th>
<th>MF</th>
<th>F, P value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>General Performance</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Incorrect responses (% trials)</td>
<td>2.7 ± 3.2</td>
<td>2.1 ± 2.9</td>
<td>9.6 ± 12.6*†</td>
<td>F_{2,47} = 3.70, P = .032</td>
</tr>
<tr>
<td>Loss of balance (% trials)</td>
<td>0.5 ± 1.5</td>
<td>0.2 ± 0.7</td>
<td>5.1 ± 7.2*†</td>
<td>F_{2,47} = 5.63, P = .006</td>
</tr>
<tr>
<td><strong>Maximum Static Leans</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP COP range (units)</td>
<td>0.94 ± 0.35</td>
<td>0.90 ± 0.33</td>
<td>1.04 ± 0.64</td>
<td>F_{2,47} = 0.17, P = .841</td>
</tr>
<tr>
<td>ML COP range (units)</td>
<td>1.69 ± 0.62</td>
<td>1.56 ± 0.54</td>
<td>1.73 ± 0.92</td>
<td>F_{2,47} = 0.18, P = .839</td>
</tr>
<tr>
<td><strong>Maximum Voluntary Sway</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP COP range (units)</td>
<td>1.02 ± 0.39</td>
<td>0.96 ± 0.44</td>
<td>1.37 ± 0.46</td>
<td>F_{2,47} = 1.42, P = .251</td>
</tr>
<tr>
<td>ML COP range (units)</td>
<td>1.81 ± 0.72</td>
<td>1.60 ± 0.71</td>
<td>1.88 ± 0.40</td>
<td>F_{2,47} = 0.40, P = .675</td>
</tr>
<tr>
<td><strong>Continuous Voluntary Sway</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP COP range (mm)</td>
<td>261 ± 49</td>
<td>234 ± 47</td>
<td>183 ± 51*</td>
<td>F_{2,47} = 4.34, P = .019</td>
</tr>
<tr>
<td>ML COP range (mm)</td>
<td>154 ± 31</td>
<td>136 ± 36</td>
<td>109 ± 24*</td>
<td>F_{2,47} = 3.67, P = .033</td>
</tr>
<tr>
<td><strong>Initiation of Voluntary Sway</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Start AP reaction time (ms)</td>
<td>492 ± 146</td>
<td>543 ± 195</td>
<td>792 ± 197*†</td>
<td>F_{2,47} = 5.28, P = .009</td>
</tr>
<tr>
<td>Start ML reaction time (ms)</td>
<td>434 ± 120</td>
<td>397 ± 81</td>
<td>548 ± 117</td>
<td>F_{2,47} = 2.26, P = .116</td>
</tr>
<tr>
<td>AP COP response (mm)</td>
<td>120 ± 21</td>
<td>107 ± 24</td>
<td>91 ± 14*</td>
<td>F_{2,47} = 3.81, P = .029</td>
</tr>
<tr>
<td>ML COP response (mm)</td>
<td>73 ± 14</td>
<td>65 ± 16</td>
<td>51 ± 6*</td>
<td>F_{2,47} = 4.50, P = .016</td>
</tr>
<tr>
<td><strong>Switches of Voluntary Sway</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Switch ML-AP reaction time (ms)</td>
<td>664 ± 140</td>
<td>743 ± 141</td>
<td>888 ± 118*</td>
<td>F_{2,47} = 4.57, P = .015</td>
</tr>
<tr>
<td>Switch AP-ML reaction time (ms)</td>
<td>613 ± 114</td>
<td>695 ± 113</td>
<td>838 ± 205*</td>
<td>F_{2,47} = 5.88, P = .005</td>
</tr>
<tr>
<td>ML-AP COP response (mm)</td>
<td>75 ± 16</td>
<td>70 ± 15</td>
<td>52 ± 9</td>
<td>F_{2,47} = 3.14, P = .052</td>
</tr>
<tr>
<td>AP-ML COP response (mm)</td>
<td>119 ± 24</td>
<td>111 ± 23</td>
<td>88 ± 17</td>
<td>F_{2,47} = 2.83, P = .082</td>
</tr>
<tr>
<td><strong>Termination of Voluntary Sway</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stop AP reaction time (ms)</td>
<td>490 ± 70</td>
<td>548 ± 81</td>
<td>536 ± 118</td>
<td>F_{2,47} = 2.63, P = .084</td>
</tr>
<tr>
<td>Stop ML reaction time (ms)</td>
<td>549 ± 64</td>
<td>588 ± 80</td>
<td>660 ± 105*</td>
<td>F_{2,47} = 4.97, P = .011</td>
</tr>
<tr>
<td>AP COP response (mm)</td>
<td>98 ± 23</td>
<td>88 ± 21</td>
<td>71 ± 29</td>
<td>F_{2,47} = 2.54, P = .090</td>
</tr>
<tr>
<td>ML COP response (mm)</td>
<td>56 ± 15</td>
<td>50 ± 18</td>
<td>44 ± 7</td>
<td>F_{2,47} = 1.27, P = .291</td>
</tr>
</tbody>
</table>

Values are mean ± standard deviation.
*Significantly different to non-fallers, P < .05.
†Multiple fallers significantly different to single fallers, P < .05.
For the orthogonal switch task, ML-AP represents a switch of ML to AP voluntary postural sway, and AP-ML represents a switch of AP to ML voluntary postural sway.
6.3.3 Relationships between voluntary sway measures

Scatter plots for the relationships between variables identified as significantly different between groups are displayed in Figure 6.2. Correlations between the AP and ML directions for measures of switch reaction time, sway initiation amplitude and continuous voluntary sway amplitude ranged from .72 to .85 (Figure 6.2a-c). Start reaction time to initiate AP sway was significantly correlated with sway initiation AP COP amplitude \( (r = -.51) \) (Figure 6.2d), and also with the stop reaction time to terminate ML sway \( (r = .44) \) (Figure 6.2e). Compound orthogonal switch reaction time was significantly correlated with the compound measures of sway initiation amplitude \( (r = -.38) \) (Figure 6.2f), and continuous voluntary sway range \( (r = -.44) \) (Figure 6.2g). Compound continuous voluntary sway range was also significantly correlated with compound sway initiation amplitude \( (r = .93) \) (Figure 6.2h).

6.3.4 Relationships between voluntary postural sway measures and PPA score

Scatter plots for the relationships between PPA score and voluntary sway measures are displayed in Figure 6.3. Correlations of PPA score with age, and the initiation, orthogonal switch and termination reaction times were significant and ranged from .33 to .45 (Figure 6.3a-e). PPA score was also significantly correlated with sway initiation and continuous voluntary sway COP amplitudes for both AP and ML directions, with \( r \) values between -.42 and -.50 (Figure 6.3f-i).

6.3.5 Classification of non-multiple fallers and multiple fallers using age, PPA score and voluntary postural sway measures

The results of the discriminant analyses are presented in Table 6.3. Age alone, PPA score alone, and age combined with PPA score significantly discriminated between the multiple fallers and non-multiple fallers \( (P < .001) \). The first voluntary sway measure selected as the strongest discriminator between groups was stop reaction time to terminate ML sway \( (P < .001) \). With all of the significantly different voluntary sway measures available for selection, the selected measures were stop reaction time to terminate ML sway, age, and start reaction time to initiate AP sway \( (P < .001) \).
Fig 6.2. Scatter plots of non-faller (white circles), single faller (gray circles), and multiple faller (black circles) participant data with fitted linear regression lines for (a) Switch reaction time (RT) AP and switch RT ML, (b) Sway initiation AP and ML amplitude, (c) Continuous voluntary sway (CVS) AP and ML amplitude, (d) Start RT AP and sway initiation AP amplitude, (e) Start RT AP and stop RT ML, (f) Sway initiation compound (cmp) amplitude and switch RT cmp, (g) CVS cmp amplitude and switch RT cmp, and (h) CVS cmp amplitude and sway initiation cmp amplitude. A cmp measure is the average of that measure over the AP and ML directions. The $r$ values represent the age-adjusted partial correlation coefficient.
Figure 6.3. Scatter plots of non-faller (white circles), single faller (gray circles), and multiple faller (black circles) participant data with fitted linear regression lines for PPA score and (a) Age, (b) Start reaction time (RT) AP, (c) Stop RT ML, (d) Switch RT AP, (e) Switch RT ML, (f) Sway initiation AP amplitude, (g) Sway initiation ML amplitude, (h) Continuous voluntary sway (CVS) AP amplitude, and (i) CVS ML amplitude. The $r$ values for plots 6.3b-i represent the age-adjusted partial correlation coefficient.
6.4 Discussion

The results confirmed the hypothesis that multiple fallers would have slower reaction time during voluntary postural sway tasks compared with non-fallers. Not only was stop reaction time to terminate ML sway identified as the strongest overall discriminator of falls-history status from the voluntary postural sway measures, the multiple fallers also had reaction times which were 20% to 60% slower compared with the non-fallers for the initiation, termination, and orthogonal switch tasks. These slowed postural reactions would seem detrimental to the ability to avoid falls because there is often a limited amount of time in which to initiate an effective response to a perturbation of balance (Grabiner and Enoka, 1995, Maki and McIlroy, 2006, Pavol et al., 2001, Stelmach and Worringham, 1985). In agreement with the current findings, the best determinants of falls-history status from the PPA subtests were hand-press reaction time and the ability to regulate postural sway oscillations whilst standing on a challenging foam surface. Therefore, the findings of the study suggest that slower voluntary postural reactions may be an important determinant of falls among older people (Lord and Fitzpatrick, 2001, Melzer et al., 2007, Brauer et al., 2000).

Multiple fallers had reduced COP amplitudes compared to non-fallers when rapidly initiating sway and when performing continuous voluntary sway in the AP and ML directions. These results suggest that the multiple fallers had weaker postural responses to shift the COP, which is a reflection of their reduced standing stability. As the multiple fallers also lost their balance more frequently during the six sway tasks, the results suggest they had difficulty in rapidly correcting whole body momentum (Cao et al., 1998b, Tucker et al., 2009). The increased PPA scores of the multiple fallers compared to the non-fallers and single fallers is consistent with the view that the multiple fallers had increased falls-risk and poor balance control because of impairments to their postural control physiology (Lord et al., 2003). The significant correlations between PPA score and the reaction time and COP measures suggests that global declines in cognitive, sensory, strength, and coordination factors might have contributed to the group differences in voluntary postural sway performance.
Table 6.3
Discriminant variables, classification accuracies, likelihood ratios, and classification equations for the non-multiple Fallers (NMF; prior probability = 90%) and multiple fallers (MF; prior probability = 10%).

<table>
<thead>
<tr>
<th>Discriminant Variables</th>
<th>Sensitivity (MF)</th>
<th>Specificity (NMF)</th>
<th>Likelihood Ratio (+)</th>
<th>Likelihood Ratio (-)</th>
<th>Wilks’ lambda</th>
<th>Classification Equations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>60*</td>
<td>98*</td>
<td>27.60</td>
<td>0.41</td>
<td>0.70</td>
<td></td>
</tr>
<tr>
<td>PPA score</td>
<td>60*</td>
<td>100*</td>
<td>0†</td>
<td>0.40</td>
<td>0.65</td>
<td></td>
</tr>
<tr>
<td>PPA score + age</td>
<td>80*</td>
<td>100*</td>
<td>0†</td>
<td>0.20</td>
<td>0.51</td>
<td></td>
</tr>
<tr>
<td>StopRT&lt;sub&gt;ML&lt;/sub&gt;</td>
<td>60*</td>
<td>96</td>
<td>13.80</td>
<td>0.42</td>
<td>0.69</td>
<td></td>
</tr>
<tr>
<td>StopRT&lt;sub&gt;ML&lt;/sub&gt; + age + StartRT&lt;sub&gt;AP&lt;/sub&gt;</td>
<td>80*</td>
<td>98*</td>
<td>36.80</td>
<td>0.20</td>
<td>0.55</td>
<td></td>
</tr>
</tbody>
</table>

*Significantly higher compared with prior probability using one-tailed Z test, \( P < .05 \).
†Low score but high positive likelihood ratio paradox.
StartRT<sub>AP</sub> is the reaction time to initiate AP sway.
StopRT<sub>ML</sub> is the reaction time to terminate ML sway.
To classify cases, age (years), PPA score (unitless), and reaction time (ms) were entered into the appropriate MF and NMF equations. Each case was assigned to the group for which it had the highest classification score.
The second hypothesis was also confirmed, because the voluntary postural sway measures significantly identified falls-history status with comparable accuracy to PPA score. Age alone was sufficient to classify a high percentage of multiple faller and non-multiple faller individuals, which confirms that advanced age is a risk factor for multiple falls (Lord et al., 1993). Two reaction time measures (stop reaction time ML and start reaction time AP) and age improved sensitivity and specificity, and the likelihood ratios for this discriminant model further demonstrated that these measures formed a useful classification model for multiple falls status. Not only were sensitivity and specificity similar between voluntary postural sway measures and PPA score, but voluntary postural sway measures were also found to be significantly correlated with PPA score. Therefore, voluntary postural sway measures provide important information regarding falls-risk and may have some clinical value (Rose and Clark, 2000, Tucker et al., 2009). The significant relationships among the voluntary postural sway measures also indicate that they provided some overlapping information across tasks and response directions. Reaction time and COP measures other than those identified as the best discriminators between groups may thus provide useful information regarding voluntary postural sway performance.

The study findings also confirmed the hypothesis that reactive postural tasks would be better predictors of falls-history status compared with static leaning actions. Significant differences were detected between multiple fallers and non-fallers for sway tasks involving a reaction response. In contrast, no significant differences were observed for the maximum lean or maximum voluntary sway tasks, a finding that is consistent with previous reports (Tucker et al., 2009, Boulgarides et al., 2003, Brauer et al., 2000, Melzer et al., 2004, Thomas and Lane, 2005, Wallmann, 2001). The maintenance of a static postural position may be less challenging to the sensorimotor and muscular systems compared to a dynamic reaction movement (Prioli et al., 2006). Alternatively, older adults regardless of their falls-risk may adopt a cautious approach to balance maintenance when approaching the limits of stability (Hahn and Chou, 2004, Menz et al., 2003c). Furthermore, most differences in the sway measures were observed between the multiple fallers and non-fallers. This indicates that voluntary
postural sway measures have the best capacity to identify older adults with balance impairments that are systematically related to falls rather than to identify older adults with more subtle balance limitations who may fall only once within 12 months (Nevitt et al., 1991, Ivers et al., 1998).

The findings of this study must be interpreted with caution as the retrospective questionnaires may have underestimated the number of falls recalled by participants (Ganz et al., 2005, Hill et al., 2004). A retrospective study also cannot determine whether the differences observed between groups contributed to the falls or were simply a consequence of the falls. What this study established, however, is that the variables investigated were significantly different between groups, and more importantly, that they were able to discriminate retrospective falls with a high degree of sensitivity and specificity. In addition, the primary author of the study (M.G.T) was not blinded to falls group membership of the individual participants which might have influenced the results. The PPA may also be more suitable as a prospective rather than retrospective measure of falls-risk (Lord et al., 2003). Although the sample size for this study was relatively small (N = 51), the group size proportions of the fallers (29%) and multiple fallers (10%) are consistent with the accepted rates of falls among community-dwelling older people (Hill et al., 2004, Lord et al., 1993).

6.5 Conclusion

The ability to react quickly with the whole body and to regulate voluntary postural sway under dynamic and challenging conditions is a strong indicator of a recent history of multiple falls in healthy, community-dwelling older people. The slower postural reactions and less effective COP stabilizing responses of the multiple fallers compared with non-fallers, and also the comparable classification accuracies between PPA score and reactive voluntary sway measures suggest that reactive postural sway tasks provide important information regarding falls-risk.
7.1 Summary of Experimental Findings

The general aim of this thesis was to examine the postural responses of young and older adults during reactive and self-paced voluntary postural sway tasks. It was anticipated that the findings would improve understanding of how posture and balance is controlled during voluntary postural sway movements. It was also expected that examination of reaction time and coordination measures of voluntary postural sway performance would provide insights into the mechanisms underlying postural instability among older people. The following section provides a summary of the four experiments that comprise this thesis.

Chapter 3. Age-related differences in postural reaction time and coordination during voluntary sway movements.

The purpose of this study was to examine age-related differences in reaction time and the pattern of temporal coordination between centre of pressure (COP), trunk, and head motion during rapid initiation of voluntary postural sway and rapid orthogonal switches of voluntary postural sway between the anterior-posterior (AP) and medial-lateral (ML) directions. The main findings of the study were that older compared with young adults had slower reaction time during both voluntary postural sway tasks, and reduced time-lag differences between COP, trunk, and head motion during the orthogonal switch reaction response and during the voluntary postural sway immediately following the orthogonal switch reaction response. Taken together, these results suggest that the older adults adopted more rigid coordination strategies when executing a rapid change in the direction of voluntary postural sway compared with
the young. The rigid movement strategy of the older adults was presumably generated in an effort to compensate for challenge to the maintenance of postural stability.

Chapter 4. Voluntary sway and rapid orthogonal transitions of voluntary sway in young adults, and low and high fall-risk older adults.

The objective of this study was to examine differences in postural stability and the speed of response between young adults, and low and high falls-risk older adults during voluntary postural sway and orthogonal switches of voluntary postural sway between the AP and ML directions. The main findings were that both falls-risk groups had slower reaction and movement time, greater COP and/or centre of mass (COM) amplitude in the non-target direction during voluntary postural sway, and reduced AP and ML COP-COM separation during voluntary postural sway and orthogonal transitions compared with the young. High falls-risk older adults had slower reaction and movement time, increased non-target COM amplitude during voluntary postural sway, and reduced ML COP-COM separation during voluntary postural sway and orthogonal transitions compared with the low falls-risk older adults. Overall, the findings suggest that age-related deterioration of postural control resulted in slower reactive responses and reduced control of the direction of body movement during voluntary postural sway and orthogonal transitions. High falls-risk older adults had a particularly reduced capacity to respond quickly during orthogonal transitions, and also had decreased control of their ML COM motions which reduced their ML stability.

Chapter 5. Differences in rapid initiation and termination of voluntary postural sway associated with ageing and falls-risk.

This study examined differences between young adults, and low and high falls-risk older adults in reaction time and the relationships between COP and COM motions during rapid initiation and termination of voluntary postural sway and continuous voluntary postural sway. Low falls-risk older adults had slower reaction time during initiation of AP sway and decreased COP-COM separation during initiation of AP and ML sway and continuous AP voluntary sway compared with the young adults. High
falls-risk older adults had slower initiation and termination reaction times in all response directions and decreased COP-COM separation during sway initiation and continuous voluntary sway in the AP and ML directions compared with the young adults. High falls-risk older adults also had slower initiation and termination reaction times in all response directions, and decreased ML COP-COM separation during continuous ML sway compared with the low falls-risk older adults. All significant differences between groups were observed in the target direction of voluntary postural sway. Reaction time and COP-COM measures significantly predicted group status in discriminant models with sensitivities and specificities ranging from 72% to 100%. Overall, these findings highlight important associations of age-related declines in sensorimotor function related to an increased risk of falling with slower postural reaction time and reduced postural stability.

**Chapter Six. What are the relationships between voluntary postural sway measures and falls-history status in community-dwelling older adults?**

The objectives of this study were to determine whether a series of voluntary postural sway tasks could differentiate and accurately identify the falls-history status of older adults, and to examine the relationships between voluntary postural sway measures and falls-risk. The six voluntary postural sway tasks were maximum static leans, maximum voluntary postural sway, continuous voluntary postural sway, rapid initiation of voluntary postural sway, rapid termination of voluntary postural sway, and rapid orthogonal switches of voluntary postural sway between the AP and ML directions. The results revealed few differences in the task measures between single fallers and non-fallers. Multiple fallers had increased age, increased Physiological Profile Assessment (PPA) falls-risk score, slower initiation, termination, and orthogonal switch reaction times, and reduced COP amplitude during sway initiation and continuous voluntary sway compared with the non-fallers. Voluntary postural sway measures were significantly correlated with each other and with PPA score. Two postural reaction time measures with age identified 80% of multiple fallers and 98% of non-multiple fallers. Similarly, PPA score with age identified 80% of multiple fallers and 100% of non-multiple fallers. The slower and less effective balance responses of the
multiple fallers compared with non-fallers and the comparable sensitivity and specificity of PPA score and reactive voluntary postural sway measures suggests that postural reaction time is a strong determinant of falls-risk.

7.2 Synthesis of Experimental Findings

In the General Introduction, three specific aims and hypotheses of this thesis were stated. In the following sections, these three aims and hypotheses are addressed based on a synthesis of the experimental findings of this thesis. The overall findings are discussed with respect to previous literature, and with respect to the unique contribution that this work provides to a greater understanding of standing postural control and the differences in standing postural control associated with ageing and increased falls-risk.

7.2.1 Reaction time during voluntary postural sway in young and older adults

The first aim of this thesis was to determine if differences exist in the reaction time of voluntary postural sway movements between young adults, low falls-risk older adults, and high falls-risk older adults. It was hypothesised that voluntary postural sway reaction times would be slower for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults. Overall, the findings of the four experiments confirmed this hypothesis as voluntary postural sway reaction times were significantly slowed with ageing and an increased risk of falling among older adults. A summary of the group differences in voluntary postural sway reaction times for each experiment is presented in Table 7.1. As detailed in the synthesis below, these findings provide new insights into the slowing of postural reaction responses with ageing and falls-risk and also demonstrate that older adults with a heightened susceptibility to falls experience a particular slowing in their speed of response.
Table 7.1

Summary of group differences in reaction times (RT) during the initiation, termination, and orthogonal switch voluntary postural sway tasks.

<table>
<thead>
<tr>
<th>Experiment 1</th>
<th>YA vs. OA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Initiation RT</td>
<td>OA &gt; YA</td>
</tr>
<tr>
<td>Orthogonal switch RT</td>
<td>OA &gt; YA</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 2</th>
<th>YA vs. LFROA</th>
<th>YA vs. HFROA</th>
<th>LFROA vs. HFROA</th>
</tr>
</thead>
<tbody>
<tr>
<td>AP-ML orthogonal switch RT</td>
<td>LFROA &gt; YA</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
<tr>
<td>ML-AP orthogonal switch RT</td>
<td>LFROA &gt; YA</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 3</th>
<th>YA vs. LFROA</th>
<th>YA vs. HFROA</th>
<th>LFROA vs. HFROA</th>
</tr>
</thead>
<tbody>
<tr>
<td>AP Initiation RT</td>
<td>LFROA &gt; YA</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
<tr>
<td>ML Initiation RT</td>
<td>LFROA &gt; YA</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
<tr>
<td>AP Termination RT</td>
<td>NSD</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
<tr>
<td>ML Termination RT</td>
<td>NSD</td>
<td>HFROA &gt; YA</td>
<td>HFROA &gt; LFROA</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 4</th>
<th>NF vs. SF</th>
<th>SF vs. MF</th>
<th>MF vs. NF</th>
</tr>
</thead>
<tbody>
<tr>
<td>AP Initiation RT</td>
<td>NSD</td>
<td>MF &gt; SF</td>
<td>MF &gt; NF</td>
</tr>
<tr>
<td>ML Termination RT</td>
<td>NSD</td>
<td>NSD</td>
<td>MF &gt; NF</td>
</tr>
<tr>
<td>AP-ML orthogonal switch RT</td>
<td>NSD</td>
<td>NSD</td>
<td>MF &gt; NF</td>
</tr>
<tr>
<td>ML-AP orthogonal switch RT</td>
<td>NSD</td>
<td>NSD</td>
<td>MF &gt; NF</td>
</tr>
</tbody>
</table>

Where a ‘greater than’ symbol is indicated, reaction time was significantly different between groups, \(P < .05\). NSD = not significantly different.

Group name abbreviations: YA = young adults; OA = older adults; LFROA = low falls-risk older adults; HFROA = high falls-risk older adults; NF = non-fallers; SF = single fallers; MF = multiple fallers.

In general, it was found that ageing was significantly associated with a slowing of voluntary postural sway reaction times. These results were evident from the slower initiation and orthogonal switch reaction times for older men compared with young men (experiment 1), the slower initiation and orthogonal switch reaction times for low falls-risk older adults compared with young adults, and the slower initiation, orthogonal switch, and termination reaction times for high falls-risk older adults compared with young adults (experiments 2 and 3). These findings have built upon four previous studies using Limits of Stability, which demonstrated that ageing results in slower reaction time to initiate leaning actions in multiple directions (Nolan et al., 2010, Liaw et al., 2009, Borah et al., 2007, Nitz et al., 2003). The current findings indicate that ageing additionally results in slower reaction time during dynamically
challenging postural tasks requiring orthogonal transitions of voluntary postural sway between the AP and ML directions. The degree of slowing of voluntary postural sway reaction times was related to the degree of sensorimotor deterioration as measured by the PPA. This finding is consistent with the idea that individuals who undergo the most severe age-related declines in sensorimotor function are less able to compensate for their deficits in postural control, and therefore experience particular difficulty when responding rapidly to stimuli whilst concurrently maintaining their standing balance (St George et al., 2007, Lord and Fitzpatrick, 2001).

To perform the voluntary postural sway reaction time tasks, participants had to rapidly detect an auditory stimulus, process the information contained in the stimulus, select the appropriate response, and then execute that response in the target direction of sway as rapidly as possible (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Given that a reaction time response is composed of a series of sensory, cognitive, and motor subtasks, it is likely that a combination of factors which have been previously verified as contributing to age-related changes in reaction time contributed to the slower responses observed between the participant groups (Stelmach and Worringham, 1985, Grabiner and Enoka, 1995). Some of these factors include a reduced capacity to detect stimuli, decreased central processing speed, increased passive joint stiffness, a more cautious strategy of response, increased preference for accuracy, reduced muscular strength (Welford, 1977, Welford, 1984), and altered neuromuscular properties (Narici and Maganaris, 2006). As suggested in experiment 1, the adoption of a more rigid postural coordination strategy by older adults during challenging postural tasks may also contribute to age-related slowing of reaction time during voluntary postural sway movements.

The combined results of experiments 2 and 3 revealed that older adults with more subtle age-related sensorimotor deficits do not have slower reaction times for all types of voluntary postural sway tasks compared with the young. In particular, there were no significant differences in the reaction times to terminate AP and ML voluntary postural sway between the low falls-risk older adults and young adults. This finding is
counterintuitive to previous reaction time research which has consistently found that ageing causes a slowing of reaction time (Welford, 1984, Cerella, 1985, Cerella, 1991, Salthouse, 1985b, Era et al., 1986). However, in most previous reaction time studies, participants were usually required to react with their upper limbs from a static position whilst seated in a chair (see for example, Yan et al., 2000, Era et al., 1986). As postural control requires a significant degree of attention (Woollacott and Shumway-Cook, 2002), the level of demand placed upon the postural control system to stabilise the body influences reaction time (Lajoie et al., 1993, Lajoie et al., 1996, Teasdale et al., 1993). Therefore, the current results highlight the importance of examining age-related differences in reaction time under conditions when standing postural stability must be maintained. Caution should be exercised when generalising age-related differences in reaction time from tasks involving movements of the upper limbs to standing postural tasks.

The current results suggest that the level of challenge to postural stability during voluntary postural sway movements influences age-related differences in reaction time. In particular, the results of experiments 1 and 2 revealed that older adults had slower reaction responses compared with the young during challenging voluntary postural sway tasks that required a rapid acceleration of the COM towards the limits of stability (e.g., the sway initiation and orthogonal switch tasks). To maintain balance and avoid stepping during these reaction responses, a complementary ability to rapidly halt the COM and restrict its trajectory before it reached the limits of stability was also required (Winter, 1995). Therefore, fast reaction movements of the body towards the limits of stability required an increased capacity to stabilise self-induced perturbations to standing balance. Presumably, the older adults were not able to withstand the same level of balance perturbation compared with the young due to age-related declines in sensorimotor and balance function (Okada et al., 2001, Stelmach et al., 1989, Wu, 1998). Therefore, age-related declines in postural control are likely to have contributed to the slower reaction times of the older adults during the sway initiation and orthogonal switch tasks compared with the young adults.
In contrast, the sway termination reaction times were not significantly different between the low falls-risk older adults and the young adults. This result may be due to differences in task requirements for the sway termination task compared with the sway initiation and orthogonal switch tasks. In particular, the sway termination task required the capacity to halt the movement of COM and return it to a position of stability rather than accelerating it towards the limits of stability. Therefore, a fast reaction response during sway termination represents less of a balance perturbation, and is less challenging to stability, compared with a fast reaction response during sway initiation or orthogonal transitions of sway. Further analysis of COP data in experiment 3 also revealed that the young adults shifted their COP a greater distance compared with the low falls-risk older adults during both AP and ML termination responses. Therefore, another potential reason for the similar termination reaction times between the young and low falls-risk older adults was an age-related reduction in the magnitude of the termination response within the reaction time period. In addition, the sway termination task required only one reaction response (i.e., “stop”), whereas the initiation and orthogonal switch tasks required one of two potential responses (i.e., “forward” or “backward” and “left” or “right”). Therefore, the sway initiation and orthogonal switch tasks had an additional cognitive requirement to identify which stimulus was presented and to select the corresponding motor response (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). As older adults have reduced information processing capacity compared with the young, age-related differences in reaction times are exacerbated under choice reaction time conditions compared with simple time reaction conditions (Ketcham and Stelmach, 2001, Welford, 1977). Taken together, the current results suggest that age-related slowing of reaction time in older adults with subtle sensorimotor deficits is most likely to be observed during postural tasks that place greater demands on cognitive processing and stability control. In agreement with these findings, other studies have reported that age-related slowing of reaction time is accentuated during task conditions that increasingly challenge postural stability such as standing compared with sitting, narrow stance compared with broad stance, and leaning compared with normal erect stance (Lajoie et al., 1996, Teasdale et al., 1993).
The experiments comprising this thesis were the first to examine voluntary postural sway reaction times in healthy older adults with different classifications of falls-risk status and falls-history status. The high falls-risk older adults had slower reaction times for all voluntary postural sway tasks and response directions compared with the low falls-risk older adults. Given the significantly slower reaction time of the high falls-risk older adults and also the significant correlations between reaction times and PPA score, the collective experimental findings suggest voluntary postural sway reaction times represent a composite measure of sensorimotor and balance function (Lord and Fitzpatrick, 2001). Therefore, a key finding to emerge from the current studies is that voluntary postural reaction times represent a simple measure of the integrated functions of the physiological systems involved in the maintenance of posture and balance and provide an indication of the falls-risk of healthy older adults. Older adults with a recent history of multiple falls also had voluntary postural sway reaction times which were 20% to 60% slower compared with non-fallers. However, the reaction responses identified as significantly slower for the multiple fallers were more task and sway direction specific in contrast to the universal slowing of voluntary postural sway reaction time for the high falls-risk older adults compared with the low falls-risk older adults. Generally, there were no significant differences in voluntary postural sway reaction times between multiple fallers compared with single fallers, and single fallers compared with non-fallers, however tendencies towards significance were observed. Taken together, these results highlight the multifactorial nature of falls, which are caused by many interacting individual, environment, and task factors (Horak et al., 1997, Horak et al., 1989, Lord and Sturnieks, 2005, Newell, 1986, Shumway-Cook and Woollacott, 2001, Woollacott, 1993). Voluntary postural sway reaction times were not significantly different between all falls-history groups because a slower voluntary postural reaction time may not be a contributing factor to all types of falls (Horak, 2006). What the current findings demonstrate is that voluntary postural sway reaction times are slowed in older adults with sensorimotor impairments that are systematically related to falls (i.e., multiple fallers), and are not particularly slowed in older adults with more subtle physiological deficits who may only fall once in a 12 month period.
7.2.2 Coordination during voluntary postural sway in young and older adults

The second aim of this thesis was to determine if differences existed in coordination during performance of voluntary postural sway movements between young adults, low falls-risk older adults, and high falls-risk older adults. It was hypothesised that coordination during voluntary postural sway movements would be altered for the low and high falls-risk older adults compared with the young adults, and for the high falls-risk older adults compared with the low falls-risk older adults. Overall, the findings of the experiments confirmed this hypothesis as ageing and increased falls-risk resulted in altered patterns of body segmental motion and reduced amplitudes of COP-COM separation during the voluntary postural sway tasks.

The findings of experiment 1 revealed age-related differences in inter-segmental coordination responses during voluntary postural sway. Reaction time differences and phase lag differences between COP, trunk, and head motion demonstrated that young men exhibited a ‘bottom-up’ sequence of COP-trunk-head movement during voluntary postural sway and orthogonal transitions. In contrast, the older men exhibited tighter temporal coupling between segmental locations, suggesting that they swayed and reacted a single rigid unit. This rigid response strategy of the older men in contrast to the more flexible strategy of the young was presumably generated in an effort to compensate for challenge to postural stability associated with the voluntary postural sway movements. In support of these findings, it has been reported that older adults adopt a stiffening strategy when stability is challenged by narrowing the base of support area during quiet stance (Benjuya et al., 2004) or by translating the standing surface in the AP direction (Wu, 1998). As the lower trunk accelerometer was placed over the approximate location of the COM (Moe-Nilssen, 1998, Moe-Nilssen and Helbostad, 2002), the tighter COP-trunk coupling for the older men is consistent with age-related reduction in COP-COM separation observed in experiments 2 and 3 when motions of the COM were more accurately measured. Similarly, a rigid voluntary postural sway strategy would result in a reduced capacity to shift the COP with respect to the COM as was also observed in experiments 2 and 3.
Similar to postural sway oscillations during quiet standing (Winter, 1995), a high level of coordination was observed between the horizontal motions of the COP and COM during voluntary postural sway. COP and COM data indicated that the control of voluntary postural sway was characterised by movements of the COP with respect to the COM in the AP and ML directions (Lafond et al., 2004). On the basis of previous research, it can be inferred that when the COM approached the limits of stability in the target direction of sway, ankle and/or hip postural muscle activity produced a stabilising net joint torque, which resulted in an overshoot of the COP trajectory with respect to the maximum amplitude of the COM (Day et al., 1993, Winter et al., 1996, Gatev et al., 1999). The overshoot of COP trajectory increased the separation distance between the COP and COM in the target direction of sway, which had the biomechanical effect of halting the movement of the COM and accelerating it in the opposite direction to the COP (Gage et al., 2004, Masani et al., 2007, Winter, 1995, Yu et al., 2008). Therefore, greater COP-COM separation during voluntary postural sway is consistent with a greater level of control over COM movements within the base of support area and increased postural stability.

The results of experiments 2 and 3 indicate that ageing and increased falls-risk were associated with reduced COP-COM separation during voluntary postural sway. A summary of the significant group differences in COP-COM separation during the voluntary postural sway tasks is presented in Table 7.2. These results are in agreement with previous reports that the level of COP-COM separation is significantly altered for older compared with young adults during quiet stance (Yu et al., 2008, Berger et al., 2005a, Masani et al., 2007), walking, and obstacle crossing (Hahn and Chou, 2004). In addition, COP-COM separation is significantly altered for balance impaired older adults compared with matched controls during quiet stance (Berger et al., 2005b, Yu et al., 2008, Corriveau et al., 2004b, Corrieveau et al., 2000), walking, and obstacle crossing (Lee and Chou, 2006, Said et al., 2008, Chou et al., 2004). In the current experiments, the low and high falls-risk older adults had reduced target COP-COM separation for all voluntary postural sway tasks and response directions compared with the young, with the exception of ML continuous voluntary sway for the low falls-risk older adults. In
addition, the high falls-risk older adults had reduced ML COP-COM separation during ML continuous voluntary sway and AP-ML orthogonal transitions compared with the low falls-risk older adults. The use of biofeedback cues to limit sway to 60% of maximum lean amplitude, and also the non-significant group differences in COP frequency and velocity further suggests that the reductions in COP-COM separation associated with ageing and increased falls-risk were due to differences in postural control rather than the amplitude, frequency, or velocity characteristics of the voluntary postural sway. Overall, these findings suggest that the older groups had closer alignment between the COP and COM compared with the young because of age-related deterioration in postural responses. Similar results were observed in experiment 4 because the multiple fallers had weaker postural responses to shift the COP during rapid sway initiation and continuous voluntary sway compared with the non-fallers. These weaker postural responses were associated with an increased occurrence of balance loss during voluntary postural sway tasks for the multiple fallers, indicating that they had a reduced ability to rapidly correct body momentum.

**Table 7.2**

Summary of group differences in COP-COM separation during voluntary postural sway tasks.

<table>
<thead>
<tr>
<th>Experiment 2</th>
<th>YA vs. LFROA</th>
<th>YA vs. HFROA</th>
<th>LFROA vs. HFROA</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>AP-ML task</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Transition AP COP-COM</td>
<td>YA &gt; LFROA</td>
<td>YA &gt; HFROA</td>
<td>NSD</td>
</tr>
<tr>
<td>Transition ML COP-COM</td>
<td>NSD</td>
<td>YA &gt; HFROA</td>
<td>LFROA &gt; HFROA</td>
</tr>
<tr>
<td>Post sway COP-COM target</td>
<td>NSD</td>
<td>YA &gt; HFROA</td>
<td>LFROA &gt; HFROA</td>
</tr>
<tr>
<td><strong>ML-AP task</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pre sway COP-COM target</td>
<td>NSD</td>
<td>NSD</td>
<td>LFROA &gt; HFROA</td>
</tr>
<tr>
<td>Transition ML COP-COM</td>
<td>YA &gt; LFROA</td>
<td>YA &gt; HFROA</td>
<td>NSD</td>
</tr>
<tr>
<td>Post sway COP-COM non-target</td>
<td>LFROA &gt; YA</td>
<td>NSD</td>
<td>NSD</td>
</tr>
<tr>
<td><strong>Experiment 3</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP Initiation COP-COM target</td>
<td>YA &gt; LFROA</td>
<td>YA &gt; HFROA</td>
<td>NSD</td>
</tr>
<tr>
<td>ML Initiation COP-COM target</td>
<td>YA &gt; LFROA</td>
<td>YA &gt; HFROA</td>
<td>NSD</td>
</tr>
<tr>
<td>AP CVS COP-COM target</td>
<td>YA &gt; LFROA</td>
<td>YA &gt; HFROA</td>
<td>NSD</td>
</tr>
<tr>
<td>ML CVS COP-COM target</td>
<td>NSD</td>
<td>YA &gt; HFROA</td>
<td>LFROA &gt; HFROA</td>
</tr>
</tbody>
</table>

Where a ‘greater than’ symbol is indicated, COP-COM separation was significantly different between groups, \( P < .05 \). NSD = not significantly different.

Group name abbreviations: YA = young adults; LFROA = low falls-risk older adults; HFROA = high falls-risk older adults.
As the magnitude of COP-COM separation is directly proportional to the horizontal acceleration of the COM (Gage et al., 2004, Masani et al., 2007, Winter, 1995, Yu et al., 2008), any reduction in COP-COM separation during voluntary postural sway reaction movements is undesirable. This is consistent with the view that an important determinant of standing postural stability is the capacity to voluntarily move the COM quickly in a specific direction within the base of support area without incurring a step (Liaw et al., 2009, Borah et al., 2007, Nitz et al., 2003). Therefore, the reduced target COP-COM separation observed with ageing and high compared with low falls-risk during the rapid sway initiation and orthogonal switch tasks limited the acceleration of the COM in the direction of response. Furthermore, the reduced COP-COM separation observed with ageing and increased falls-risk during continuous voluntary sway is also reflective of decreased postural stability because it limits the capacity to generate a stabilising torque at the end of a sway cycle (e.g., swaying forward) and to reverse the direction of sway (e.g., begin swaying backward). Taken together, the current findings build upon previous research to demonstrate that ageing and increased falls-risk are associated with altered coordination between COP and COM motions and reduced postural stability during voluntary postural sway movements.

Fear of falling among older adults is a psychological factor that may influence the control of standing stability (Maki et al., 1991), and might be regarded as a contributing factor to differences in COP-COM separation. Carpenter and colleagues (2001) induced increased fear of falling in young adults by raising the platform on which participants stood to a height of 81 cm above the ground. It was found that ankle stiffness computed from the COP-COM difference was increased and also that COP and COM amplitudes were significantly decreased with the increased postural threat. Given the potential influence of psychological factors on COP-COM separation, fear of falling was measured in the low and high falls-risk older adults in experiment 2 using the Falls Efficacy Scale-International (FES-I). Despite a greater proportion of individuals with a history of falling in the high falls-risk group compared with the low falls-risk group (50% versus 22%), no significant differences were detected between the groups in FES-I scores (mean scores: 21.7 and 26.3 for the low and high falls-risk
groups, respectively). This finding may reflect that the current sample of older adults were living independently in the community and were generally healthy and physically active irrespective of previous falls. Considerably higher FES-I scores (i.e., increased fear of falling) have been reported in previous studies of 161 frail geriatric inpatients (median scores: 32 to 35; Denkinger et al., 2010) and 704 older adults from different socioeconomic backgrounds and with a wide range of falls and medical histories and levels of physical function (mean scores: 31 to 35; Yardley et al., 2005). Taken together, the findings suggest that fear of falling was unlikely to have had any strong influence on the group differences in COP-COM separation in the current sample of healthy community-dwelling older adults.

An interesting finding was that the low and high falls-risk older adults had increased COP and/or COM motion in the non-target direction during voluntary postural sway compared with the young adults. Studies involving voluntary leaning, Limits of Stability, and Rhythmic Weight Shifts have also found reduced path accuracy of the COP with ageing (Liaw et al., 2009, Borah et al., 2007, Hageman et al., 1995, Blaszczyk et al., 1994). These results indicate that ageing results in a reduced capacity to regulate the target and non-target sway directions concurrently, which is in general agreement with the other findings of age-related declines in postural stability during voluntary postural sway movements. However, these increased non-target COP and/or COM amplitudes did not necessarily result in significantly altered non-target COP-COM separation. In fact, the collective findings indicate that non-target COP-COM separation during voluntary postural sway tasks was both small in magnitude and was not altered by ageing or increased falls-risk. In experiment 2, the low falls-risk older adults had significantly increased non-target COP-COM immediately following an ML-AP switch of voluntary postural sway compared with the young adults. However, this was an isolated finding that was task and direction specific. Taken together, the results suggest that healthy older adults are at minimal risk of losing their balance in the non-target direction during planar voluntary postural sway movements.
A prominent finding of the current studies was the association of increased falls-risk with ML instability during voluntary postural sway. In particular, the COP-COM separation measures that were significantly reduced for the high falls-risk older adults compared with the low falls-risk older adults were predominantly related to the ML direction. In addition, increased falls-risk was associated with increased ML COM amplitude during ML continuous voluntary sway. Collectively, these findings suggest that high falls-risk older adults had poor control over the trajectory of their ML COM which reduced their ML COP-COM separation. Therefore, high falls-risk older adults might be more susceptible to lateral falls during movements that challenge ML stability. These findings add to a growing body of literature that high falls-risk older adults have particular declines in their ML stability, which places them at increased risk of lateral falls and hip fractures (Rogers and Mille, 2003). In particular, older adults with a recent history of falls (Lee and Chou, 2006) and traumatic brain injury (Chou et al., 2004) exhibit increased ML COM amplitude and reduced COP-COM separation during obstacle crossing compared with matched controls. Furthermore, the strongest quiet stance markers of falls-status are related to the amplitude and speed of postural sway in the ML direction (Piirtola and Era, 2006, Maki and McIlroy, 1996, Lord et al., 1999). As participants were required to sway in the ML direction using a sequential load-unload action of the limbs, the ML instability of the high falls-risk older adults was possibly related to altered hip abductor-adductor and/or ankle inverter-everter neuromuscular control (Hilliard et al., 2008, Winter et al., 1996, Rogers and Mille, 2003).

7.2.3 Prediction of falls-risk and falls-history using voluntary postural sway measures

The third aim of this thesis was to determine which combination of voluntary postural sway tasks, sway directions, and balance measures best predict the falls-risk status and falls-history status of older adults. It was hypothesised that reaction time and coordination measures of voluntary postural sway tasks would accurately predict the group status of the high and low falls-risk older adults and the multiple fallers and non-multiple fallers. Overall, the findings of experiments 3 and 4 confirmed this hypothesis. It was found that voluntary postural sway measures obtained under reaction time
conditions significantly predicted the falls-risk and falls-history groups with high degree of sensitivity and specificity and favourable likelihood ratios. These findings indicate that the current voluntary postural sway measures provide important information regarding the deterioration of postural control that occurs with an increased susceptibility to falling among older people.

An objective of this thesis was to determine whether maximum static lean and maximum voluntary postural sway tests had any predictive value for classifications of falls-risk and falls-history status. In experiments 2 and 3, it was found that the low and high falls-risk older adults had reduced COP amplitude in the forwards and sideways directions during maximum voluntary postural sway compared with the young adults. However, no significant differences were observed between the low and high falls-risk older adults in these maximum COP amplitudes. Similarly, it was observed in experiment 4 that the range of COP excursion during maximum voluntary sway and maximum static leans in the AP and ML directions were not significantly different between non-fallers, single fallers, and multiple fallers. Therefore, the current results have confirmed the results of previous investigations that maximum voluntary sway and maximum lean COP amplitudes are reduced with ageing (Holbein-Jenny et al., 2007, Nitz et al., 2003, Okada et al., 2001, Blaszczyk et al., 1993b, Blaszczyk et al., 1994, Fujiwara et al., 1982, Murray et al., 1975) but are not predictive of falls in healthy older adults (Melzer et al., 2004, Boulgarides et al., 2003, Brauer et al., 2000, Wallmann, 2001). Healthy and active community-dwelling older adults retain their capacity to shift their body maximally within the base of support area despite any differences in their falls-risk.

In contrast to measures of maximum lean and sway amplitudes, reaction time and coordination measures of voluntary postural sway performance were significantly predictive of group status. With regards to the predictive model for the low falls-risk older adults and young adults, reaction time and target COP-COM separation during the rapid initiation of AP sway predicted group status with 76% sensitivity and 72% specificity. In other predictive models involving the high falls-risk older adults or
multiple fallers, the sensitivities and specificities were generally much higher, presumably because greater age-related declines in sensorimotor function produce greater declines in voluntary postural sway performance. High falls-risk older adults and young adults were predicted with 100% sensitivity and 88% specificity by AP initiation reaction time, ML termination reaction time, and target COP-COM separation during AP sway initiation. The high and low falls-risk older adults were predicted with 73% sensitivity and 88% specificity by slower AP and ML termination reaction times and reduced target COP-COM separation during ML continuous voluntary sway. These relatively high prediction accuracies for the older adult groups suggests that the voluntary postural sway measures provided similar information on the falls-risk of older adults compared with the PPA. In agreement with the aforementioned falls-risk model, multiple fallers and non-multiple fallers were predicted with 80% sensitivity and 98% specificity by ML termination reaction time, age, and AP initiation reaction time. PPA score and age also predicted the multiple fallers and non-multiple fallers with 80% sensitivity and 100% specificity, which are considerably higher accuracies than was reported during the validation of the PPA as a falls-risk assessment instrument. In a 12 months prospective study of 341 older women randomly selected from the community, it was found that the PPA classified the multiple fallers and non-multiple fallers with 75% accuracy (Lord et al., 1994b). Therefore, it is likely that the predictive accuracies for multiple falls-history status reported in experiment 4 using voluntary postural sway measures, age, PPA score represent an overestimation of the predictive accuracies that would be obtained in a larger prospective study.

An important outcome of the current experiments is that a reduced capacity to terminate voluntary postural sway movements is strongly associated with an increased susceptibility to falls. These findings are important because the majority of previous reaction time research in relation to falls have examined movement initiation tasks rather than movement termination tasks (Cerella, 1985, Welford, 1977, Welford, 1984). The current findings indicate that termination reaction times were not only better predictors of group status compared with initiation reaction times, they were also the strongest overall predictors of both high falls-risk and multiple falls-history
status out of all the voluntary postural sway measures assessed in experiments 3 and 4. Therefore, these novel findings highlight the need for investigation into the predictive capacity of termination reaction times for falls and whether a slower reaction time to terminate voluntary postural movements is a contributing factor to falls and balance loss in older people.

Irrespective of whether participants were divided into groups based on their age, falls-risk, or falls-history, the measures that best predicted group status were overwhelmingly related to the reactive voluntary postural sway tasks. Overall, the reaction times to initiate and/or terminate voluntary postural sway were selected as significant and independent predictors of group status in all discriminant models. In addition, an initiation or termination reaction time was the strongest overall predictor of group status in five out of the six discriminant models. Therefore, voluntary postural sway reaction times are strong and important indicators of the functional capacity of the postural control system. These results are consistent with those of Lord and Fitzpatrick (2001) who found that choice stepping reaction time was a significant and independent predictor of falls over and above a comprehensive range of physiological, neuropsychological and balance measures. The high sensitivities and specificities for prediction of falls-risk and falls-history, and also the similar classification accuracies achieved using PPA score and voluntary postural sway measures suggests that the current voluntary postural sway tasks may have some clinical value for screening the falls-risk of older adults. An advantage of the PPA is that it involves a detailed assessment of sensorimotor systems and provides information about the underlying physiological impairments (i.e., the likely causes of increased falls-risk) so that a targeted intervention program can be developed for the individual (Lord et al., 2003, Lord et al., 2007). However, there is also a need for quick screening tests of falls-risk prior to more detailed assessment (Whitney et al., 2005), which the current voluntary postural sway task measures may be suitable if adapted to the clinical environment.
7.3 Physiological Mechanisms of Voluntary Postural Sway

The specific aims of this thesis were to examine group differences in reaction time and coordination measures during voluntary postural sway tasks, and to evaluate the predictive capabilities of these task measures for group status in healthy young and older adults. To achieve these aims, the experimental chapters primarily focused on the analysis and interpretation of biomechanical parameters collected during the voluntary postural sway tasks rather than physiological parameters that were collected separately using the PPA. Although the biomechanical parameters proved successful for accomplishing the aims of this thesis, the PPA data provides the unique opportunity to investigate the role of sensorimotor physiology during the performance of voluntary postural sway tasks. As such, additional correlation and multiple regression analyses were performed to evaluate the relationships between the 17 sensorimotor/balance measures of the long-form PPA (Lord et al., 2003) and the voluntary postural sway measures that formed the best predictive models of group status from experiments 3 and 4. The details and results of these analyses are presented in Appendix 1, and the potential physiological mechanisms underlying the group differences in voluntary postural sway performance are discussed in this section.

7.3.1 Relationships between PPA subtests and voluntary postural sway reaction times

It was found that reaction time to initiate AP voluntary postural sway was significantly associated with visual, tactile, strength, reaction time, and balance measures, which formed the most strongly predictive multiple regression model compared with the other voluntary postural sway variables assessed ($R^2 = 0.61$). These significant associations are consistent with the task requirements of rapidly initiating voluntary postural sway. For instance, the ability to rapidly propel the COM towards the limits of stability about the ankle joint and then to recover from this perturbation is likely to depend on sensory feedback, cognitive processing, ankle strength, and dynamic balance control (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Therefore, the results suggest that deterioration in multiple physiological systems with ageing and balance impairment contributed to the group differences in sway initiation.
reaction time in the experimental chapters. The stronger association of sway initiation reaction time with tactile sensitivity compared with the visual function parameters and leg proprioception suggests that participants may have relied preferentially on tactile feedback from the feet during their reaction responses (Maurer et al., 2001). The significant correlation between faster AP sway initiation reaction time and increased ankle strength confirms the importance of ankle strength in facilitating rapid postural responses in the AP direction. The current findings closely correspond with those of Lord and Fitzpatrick (2001) who reported that the choice stepping reaction time of 30 young adults and 477 retirement-village residents was significant associated with a variety of sensorimotor, balance, and neuropsychological tests. The current results have built upon these findings to demonstrate that voluntary postural sway reaction times also represent a composite measure of sensorimotor and balance function.

While it is likely that multiple sensorimotor systems influenced the sway initiation response, the multiple regression analyses also indicated that hand press reaction time was the strongest independent predictor of AP sway initiation reaction time. As hand press reaction time emphasises CNS processing events rather than sensory detection, nerve conduction, or muscle strength (Lord et al., 1991a, Lord et al., 1994b, Corriveau et al., 2004a), the results provide some evidence to suggest that the speed of cognitive processing might have been the primary physiological mechanism contributing to the differences in AP sway initiation reaction time associated with ageing, falls-risk, and falls-history status.

The reaction time to terminate ML voluntary postural sway was also significantly associated with the PPA subtests, particularly the tests of postural sway and maximum balance range. These results suggest that the ability to rapidly terminate ML voluntary postural sway was most strongly related to dynamic balance control compared with the sensory or strength parameters. An unexpected finding was that the correlations between ML sway termination reaction time and the hand and foot press reaction times did not reach significance ($P = 0.067$ and $P = 0.023$, respectively). Similarly, foot press reaction time was the weakest predictor of ML sway termination reaction time in
the multiple regression model. These findings may reflect the additional requirement to maintain standing dynamic postural stability during the sway termination tasks compared with the maintenance of a static seated posture during the PPA hand and foot press reaction time tasks (Lajoie et al., 1993, Lajoie et al., 1996). Consistent with this view, it was found that ML sway termination reaction time was significantly associated with visual depth perception and tactile sensitivity, which are likely to enhance postural control during voluntary postural sway movements and facilitate termination responses.

Consistent with the detrimental influence of slowed reaction time on postural stability, it was found that slower voluntary postural sway initiation and termination reaction times were significantly associated with increased postural sway. These findings are in agreement with a study by Lord, Clark, and Webster (1991b) who found that slower reaction time was significantly associated with increased postural sway with eyes closed on a foam surface in 95 elderly persons recruited from an aged-care institution. Similar to the physiological processes associated with reacting during voluntary postural sway movements, the correction of standing postural sway also depends on the ability to detect a change in sensory input, central processing of the stimulus, selection of the appropriate sway response, and execution of the sway response as rapidly as possible (Grabiner and Enoka, 1995, Stelmach and Worringham, 1985). Faster ML sway termination reaction time was also significantly associated with increased maximum balance range, presumably because an increased capacity to sway towards the limits of stability would require the concurrent ability to rapidly terminate the oscillation to preserve standing balance.

7.3.2 Relationships between PPA subtests and COP-COM separation during voluntary postural sway

It was generally found that target AP and ML COP-COM separation measures were poorly associated with the PPA subtests. Although AP COP-COM separation during the initiation of sway was not significantly correlated with any of the PPA subtests, the multiple regression analysis identified quadriceps muscle strength as a significant and
independent predictor. Quadriceps strength was also the most highly correlated PPA test with AP sway initiation COP-COM separation \((R = 0.25, P = 0.054)\). These results provide some small suggestion for a role of quadriceps strength in the regulation of COP-COM separation during the rapid initiation of AP sway. Older adults with weaker quadriceps might have exhibited reduced postural stability during this voluntary postural sway task. Given the intuitive role of the ankle joint in performing rapid initiations of AP voluntary sway it was expected that ankle strength would have been more strongly related to AP COP-COM separation. Perhaps the weaker than expected association was because the PPA measured ankle dorsiflexion strength when the ankle plantar flexors might have been the dominant control muscles. Similarly, ML COP-COM separation during continuous voluntary sway was not significantly associated with any of the PPA strength test measures, which may be due to the fact that these voluntary sway movements required contributions from the hip abductors-adductors and ankle inverters-everters rather than the quadriceps, hamstrings or dorsiflexors (Day et al., 1993, Winter et al., 1996). The significant association between better coordinated stability score and increased ML COP-COM separation is consistent with the fact that the coordinated stability test required large and tightly regulated voluntary movements of the COM in the ML direction. It is likely that an increased capacity to regulate ML accelerations of the COM would therefore contribute to better coordinated stability tracking.

In a previous study, Corriveau and colleagues (2004a) used structural equation modelling to investigate the relative contributions of visual, somatosensory, strength, and cognitive processing factors to AP and ML COP-COM separation during quiet stance from 75 older adults. Twenty-nine of the participants had sensorimotor and balance disabilities due to stroke or diabetic peripheral neuropathy. It was found that the strongest determinant of AP and ML COP-COM separation was lower extremity muscle strength measured from all of the major muscle groups of the lower limbs \((R^2 = 0.24\) to 0.41\) and that tests of somatosensation and vision similar to that used in the PPA only explained small proportions of variance in COP-COM separation \((R^2 = 0.02\) and 0.001, respectively). Therefore, it is likely that the lack of association between
COP-COM separation and muscle strength during voluntary postural sway movements was because of the muscle groups tested by the PPA. Corriveau et al (2004a) also found that central processing (i.e., reaction time) explained significant proportions of variance in AP and ML COP-COM separation ($R^2 = 0.06$ to 0.27), which is consistent with the current associations between foot press reaction time and target ML COP-COM separation. Given the important role of the ankle joints in stabilising ML postural sway (Day et al., 1993, Winter et al., 1996), the results suggest that the capacity for rapid adjustment of the ankle joint motions might have contributed to wider ML COP-COM separation. It was also suggested in experiments 2 and 3 that the significantly reduced target COP-COM during ML voluntary postural sway for the high falls-risk older adults compared with the young adults and low falls-risk older adults could have been due to reduced ankle neuromuscular control.

There are a number of potential reasons for the poor associations between the PPA sensory function tests and the amplitude of COP-COM separation during the voluntary postural sway tasks. With regards to the poor associations with vision, previous studies have reported that the regulation of postural sway whilst standing on a firm surface is minimally dependent on visual acuity and contrast sensitivity parameters, particularly in older people (Lord et al., 1991b, Corriveau et al., 2004a). During such tasks, somatosensory information is believed to be the primary source of sensory feedback (Anacker and Di Fabio, 1992, Fitzpatrick and McCloskey, 1994, Lord et al., 1991a, Brocklehurst et al., 1982), which is in agreement with the weak but evident association between increased ML COP-COM separation and improved leg proprioception. The current version of the long-form PPA also includes no vestibular function test, and therefore it is unclear whether vestibular feedback played a significant role in the regulation of COP-COM separation. However, previous research suggests that vestibular feedback was probably not essential during voluntary postural sway because individuals with complete loss of vestibular function can produce relatively normal ankle strategy sway responses (Horak et al., 1994, Horak et al., 1990).
It is also possible that participants reduced their reliance on visual, vestibular, and somatosensory feedback after becoming familiar with the auditory biofeedback which was used to regulate each participant’s amplitude of voluntary postural sway. An example of such an effect has been demonstrated in the studies of Dozza and colleagues (2005, 2007). These studies examined whether an audio biofeedback system with pitch and volume encoding of postural sway could improve the standing stability of bilateral vestibular loss patients and age- and sex-matched controls. It was found that the biofeedback system significantly improved standing stability in both the vestibular patients and controls, particularly under conditions of limited visual and somatosensory information. In addition, it may also be argued that the PPA sensory function tests did not accurately represent the type of sensory feedback used during voluntary postural sway movements. For example, the proprioception test was a leg matching task requiring primarily knee rotations, whereas participants generated voluntary postural sway with their knees held in extension. Tactile sensitivity was also measured using microfilaments applied to the lateral malleolus of the ankle joint. However, a growing body of research suggests that tactile feedback from pressures applied to the plantar soles of the feet would have been more relevant to the regulation of COP-COM separation during voluntary postural sway oscillations (Inglis et al., 2002, Kavounoudias et al., 1998, Kavounoudias et al., 1999, Maurer et al., 2001).

7.3.3 Summary of physiological mechanisms of voluntary postural sway

In summary, it was found that the reaction time to initiate AP sway was significantly correlated with the highest numbers of PPA subtests and also formed the most strongly predictive multiple regression model. There is some evidence to suggest that the primary physiological mechanism underlying the group differences in AP sway initiation reaction time due to ageing and increased falls-risk could have been slower cognitive processing of the auditory cues. The sway initiation and termination reaction times provided important information on sensory and cognitive function, ankle muscle strength, postural sway, and dynamic balance in young and older adults. The results provide further confirmatory evidence that voluntary postural sway reaction times represent a composite measure of sensorimotor and balance function and may be
useful in conjunction with the PPA as a quick screening test of falls-risk prior to more detailed physiological examination (Whitney et al., 2005). The amplitude of COP-COM separation during AP and ML voluntary postural sway tasks had fewer correlations with the PPA sensorimotor and balance tests. The COP-COM separation measures were only weakly related to muscle strength, which may be a reflection of the specific muscle groups tested by the PPA. Overall, the weaker association of COP-COM separation with sensorimotor function suggests that COP-COM separation during voluntary postural sway encapsulates different aspects of postural stability compared with the PPA tests.

7.4 Limitations

The findings of any study must be interpreted with respect to the limitations in participant sampling, experimental design, and the techniques of measurement. Some limitations relevant to the current experiments are discussed in the following sections. Other limitations which are not discussed here are acknowledged in the experimental chapters.

7.4.1 Sample bias and safety considerations

The older adult participants of the current studies were volunteers living independently within the surrounding community. It has been well documented that older adults who are frail, unwell, or living in aged-care are more likely to experience mobility limitations and falls (Hill et al., 2004, Lord et al., 2007). Therefore, it is unclear whether the experimental findings can be generalised to these subpopulations of older adults. In addition, some types of community-dwelling older adults may have been deterred from participating in the current studies. For example, the testing protocols may have been considered lengthy and difficult because of the requirement to stand for extended periods. The young adult participants were also a convenience sample recruited from the University campus. Therefore, the participant groups were not truly random samples of community-dwelling younger and older people. In general,
limitations due to sampling bias and limited external validity are common in studies of postural control similar to those that comprise this thesis.

As falls may occur when testing the balance of unstable individuals, it is important to reduce the likelihood of injury due to falls during testing. Experiments 2, 3, and 4 involved groups of older adults who were susceptible to falling, and therefore all young and older participants were fitted with a safety harness to prevent any fall-related impact with the ground. The harness was adjusted prior to testing for each participant to ensure that their voluntary postural sway movements were not restricted in any way. Nevertheless, the harness may have had some small influence on the voluntary postural sway movements, for example by reducing fear of falling or providing additional somatosensory feedback. The results suggest that the harness was unlikely to have had a major influence on the results. For example, the age-differences in voluntary postural sway performance in experiments 1, 2, and 3 were similar despite the fact that no harness was used in experiment 1.

7.4.2 Experimental design limitations

The univariate statistical tests used in the experiments reflected a less conservative approach to the investigation of ageing, falls-risk and falls-history effects on the performance of voluntary postural sway tasks. Given the close relationships between the voluntary postural sway task measures, it may be argued that multivariate analysis would have been a more efficient statistical approach that reduced the likelihood of Type 1 error. However, a multivariate approach can only provide a global indication of movement when it was an objective of this thesis to undertake a highly focussed analysis of group differences in reaction time, COP, COM, and COP-COM amplitudes with respect to the different voluntary postural sway tasks and movement directions. As no corrections were made for multiple comparisons expect for in experiment 3, some of the significant results may have occurred due to chance.

In experiment 4, the number of falls by the older adults was measured retrospectively using self-report questionnaires. Falls are underestimated by approximately 15% to
20% with retrospective measurement compared with prospective measurement (Ganz et al., 2005, Hill et al., 2004, Cummings et al., 1995). Therefore, there may have been some error in dividing the older adults into falls groups in experiment 4. The older adults in experiments 2 and 3 were divided into high and low falls-risk groups based on their PPA score (St George et al., 2007, Menz et al., 2003a). The PPA is currently one of the most accurate and reliable falls-risk instruments and predicts multiple fallers with 75% accuracy in community settings (Lord et al., 2003). Although the PPA is not as accurate in measuring falls-risk compared with prospective designs, the PPA still provided important information regarding associations of physiological function with ageing, falls-risk, and voluntary postural sway measures.

In experiment 2, the high falls-risk older adults were significantly older compared with the low fall-risk older adults. Therefore, it is possible that increased age might have contributed to the significant differences in voluntary postural sway measures between these groups. However, the same pattern of significant differences in reaction time and stability measures was observed between the falls-risk groups in experiment 3 when the low and high falls-risk older adults had similar ages. Therefore, the age-difference between the low and high falls-risk older adults in experiment 2 was unlikely to have had any major impact on the results of the study.

7.4.3 Measurement limitations

Motion analysis and biomechanical modelling were used to calculate the movements of the COM in experiments 2 and 3. To estimate the location of the COM, an accurate anthropometric model and complete kinematic description of markers attached to specific anatomical locations was required (Lafond et al., 2004). Prior to testing, a series of non-invasive anthropometric measurements including mass, height, and segment lengths and widths were taken from each participant. The COM was derived from anthropometric data, and therefore the COM calculation accuracy depends on how the ageing process affects the quality of anthropometric data. For example, older adults are known have differences in the distribution of bone, muscle, fat, and other tissues within a given segment compared with young adults (Kuczmarski, 1989). These
differences were not taken into account during the modelling. Therefore, these factors may have contributed to error in the measurement of COM motions. Despite these potential errors in COM calculation, the relationships observed between the COP and COM for both the young and older adults during voluntary postural sway were consistent with the accepted relationships previously reported in the literature (Lafond et al., 2004, Winter, 1995).

Participants were required generate voluntary postural sway movements primarily using the ankle joint for AP sway and to sequentially load and unload each leg for ML sway. For sway movements in the AP direction, it is known that older adults prefer to use a hip strategy compared with an ankle strategy (Okada et al., 2001, Manchester et al., 1989, Gu et al., 1996). To deter excessive use of a hip strategy, all participants were given instruction and practice prior to testing. Whenever excessive hip motion was noticed during practice or testing, efforts were made to correct it, however, some of the older adults may have still used more hip strategy compared with the young during AP voluntary postural sway. Arguably, participants should be permitted some degree flexibility in their postural responses rather than severely restricting movement by artificial means such as via splints.

7.5 Future Directions

A number of new research questions regarding voluntary postural sway movements and the effects of ageing and falls-risk on voluntary postural sway movements have arisen due to the work undertaken in this thesis. In addition, given that voluntary postural sway is a convenient model for examining postural control, the current protocol has widespread applications for examining the influence individual, task, and environment factors on the dynamics of postural control. Some of the more prominent indications for future research are outlined in the following sections.
7.5.1 Prospective falls study

Experiment 4 established that voluntary postural sway measures were significantly associated with falls-history status and were able to predict multiple falls-status with a high degree of sensitivity and specificity. However, given the limitations of retrospective designs, these findings need to be validated in a prospective falls study to provide a more accurate indication of the voluntary postural sway measures to predict falls. Power calculation suggests that this study would ideally include 200 to 300 older adults, who would be recruited from the surrounding community using the electoral roll to reduce sample bias. The frequency, type, and locations of falls would be collected using monthly falls-calendars and follow-up telephone calls over a study period of 12 months.

7.5.2 Reaction time during voluntary postural sway

As global measures of reaction time were used in the current studies, the division of reaction times into premotor and motor components may provide a greater level of precision regarding the slower responses observed with ageing and increased falls-risk. Another interesting research question would be to determine whether group differences exist in voluntary postural sway reaction times for the forward, backward, left, and right directions separately. Older adults may experience particular difficulty when responding backward because of the lack of visual feedback and the short distance of the foot posterior to the ankle joint in this direction. Greater control over the instant that the reaction time cue is delivered with respect to the position of the swaying COP would be beneficial. This would improve the precision of the voluntary postural sway reaction times, and would allow investigation of a probe reaction time task (Teasdale et al., 1993). In the probe task, the reaction time cue could be presented at different COP amplitudes during voluntary postural sway, thereby providing insights into the attentional demands of stabilising voluntary postural sway movements (Teasdale et al., 1993). Two recent studies have demonstrated that combined resistance and balance training in older adults for 6 to 12 weeks were effective in improving reaction time and PPA falls-risk (Morrison et al., 2010, Vogler et
al., 2009). Given the strong associations of falls-risk and falls-history with termination reaction time, an intervention study to train termination reaction time with the aim of decreasing falls and PPA falls-risk would also be an interesting direction for future research.

7.5.3 Coordination during voluntary postural sway

Examination of other stability and coordination measures of voluntary postural sway performance may be useful to provide supporting evidence for the conclusions of the thesis. The extrapolated COM (Hof et al., 2005) and the time-to-contact of the COM with the limits of stability (Haddad et al., 2006) are common postural stability measures that could be employed. Measures derived from nonlinear dynamical systems theory are also likely to provide useful information regarding the coordination and stability properties of voluntary postural sway movements (Kurz and Stergiou, 2004). Some of these measures include the Lyapunov exponent (stability), approximate entropy (regularity), relative phase (coordination patterns), and power spectral analysis (variability). Examination of joint level kinematics and kinetics and the electromyographic activity of selected postural muscles would improve understanding of coordination during the performance of voluntary postural sway movements. Coactivation measures may provide supporting evidence for rigid coordination strategies adopted by older adults during voluntary postural sway. Examination of ankle and hip joint kinematics, kinetics, and muscle activation patterns could be used to confirm the use of ankle and hip strategies during voluntary postural sway, and could also reveal insights into CNS control of ML stability in older adults with increased falls-risk.

7.5.4 Postural control physiology

The assessment of sensorimotor and balance function by the PPA provided useful insights into the association between global physiological decline and reduced voluntary postural sway performance. Evaluation of relationships between the PPA subtests and voluntary postural sway measures in Appendix 1 also gave an indication
of the contribution of sensorimotor and balance parameters to the performance of voluntary postural sway movements. However, the PPA subtests only provided a general estimate of associations between physiological systems and voluntary postural sway. Physiological tests that are more targeted to the performance of voluntary postural sway movements would provide a better indication of the underlying physiological mechanisms. Some physiological tests of primary interest for future investigation may include static and dynamic tactile sensitivity of the plantar soles of the feet, ankle and hip joint proprioception, and isokinetic dynamometer strength testing of ankle dorsiflexors-plantar flexors, hip abductors-adductors, and the ankle inverters-everters. Stronger empirical evidence for the unique role of each system in voluntary postural sway performance could also be established by employing sensory perturbations and muscle fatigue (Menz, 2002). As high falls-risk older adults are thought to rely heavily on visual feedback for postural control (Tobis et al., 1985, Lord et al., 1994a), an interesting research question would be to determine whether the removal of vision influences voluntary postural sway responses and improves the accuracy of falls prediction. Another question of theoretical interest is whether young adults would adopt voluntary postural sway responses similar to older adults if their sensory feedback was altered (e.g. light scattering lenses, head tilt, stance on foam surface) or their ability to produce muscle force was inhibited by fatigue.

7.5.5 Development of a clinical test

The high sensitivities and specificities of the voluntary postural sway measures for predicting falls-risk and falls-history statuses suggests they might be clinically useful to screen older adults for susceptibility to falls. However, the voluntary postural sway protocols would need to undergo further research and development before clinical suitability. As outlined above, the findings of experiment 4 need to be confirmed in a well-conducted prospective study. It is also important to increase the external validity of the current findings by testing older adults dwelling in aged-care institutions and hospital settings and those with diseases affecting postural control. The voluntary postural sway measures need to be evaluated for test-retest reliability and concurrent validity with respect to other common clinical balance tests. Based on the preliminary
findings presented in this thesis, two reaction time measures and age formed an excellent predictive model for multiple falls status. The initiation, orthogonal switch, and termination reaction times can be measured in a single voluntary postural sway trial, which allows for quick and easy assessment. A forceplate with automated data analysis software would provide the most accurate measurement of voluntary postural sway reaction times and would be most desirable from a cost-benefit perspective. A cheaper but less accurate alternative may involve an instrument to directly record sway movements of the body (similar to the swaymeter of the PPA) used in conjunction with a stopwatch to measure response times.

7.6 Conclusions

Reaction time

The slowing of voluntary postural sway reaction times for older compared with young adults depends on variability in the normal ageing process. Reaction times are universally slowed for healthy older adults who have greater age-related sensorimotor impairments related to an increased risk of falling. However, older adults with more subtle age-related physiological declines exhibit a more task specific slowing of reaction time, which is particularly evident during more challenging voluntary postural sway tasks. The degree of slowing of voluntary postural sway reaction times is consistent with the degree of deterioration in postural control physiology. Collectively, these findings indicate that voluntary postural sway reaction times are a composite measure of sensorimotor and balance function and are strongly related to increased falls-risk and postural instability.

Coordination

Ageing and increased falls-risk result in altered coordination between body segmental regions, and between COP and COM motions during voluntary postural sway. These altered coordination patterns were due to deterioration in postural responses that shift the COP with respect to the COM to stabilise voluntary postural sway. The altered coordination patterns with ageing reduced the capacity of the older adults to rapidly
generate or arrest body momentum during voluntary postural sway. For multiple fallers, this was likely to have underpinned their greater frequency of balance loss during testing compared with non-fallers and single fallers. Ageing also results in a reduced capacity to concurrently regulate postural sway in the target and non-target directions of movement. However, this does not place older adults at increased risk of balance loss in the non-target direction during planar voluntary postural sway. High falls-risk older adults have particular declines in their ML stability during voluntary postural sway. Therefore, healthy older adults with greater levels of sensorimotor impairment may be at increased risk of falling laterally during tasks challenge ML postural stability.

**Prediction**

The capacity to maximally shift the body within the base of support area is retained among older adults with different falls-histories and levels of sensorimotor impairment. Therefore, maximum leaning and sway tests are not good predictors of falls in healthy and active community-dwelling older adults. Reaction time and coordination measures of voluntary postural sway performance are consistent and accurate predictors of age, falls-risk, and falls-history statuses. Therefore, reaction time and coordination measures of voluntary postural sway performance provide important information about the functional status of the postural control system. As voluntary postural sway reaction times, particularly the reaction time to rapidly terminate sway, were the strongest overall predictors of an increased susceptibility to falls, the current findings suggest that a reduced capacity to rapidly terminate postural motion might contribute to episodes of balance loss among older people. The high sensitivities, specificities and favourable likelihood ratios for different classifications of falls-risk suggest that the current protocol of voluntary postural sway reaction times may be useful to screen the falls-risk of older adults.
APPENDIX 1

Relationships between sensorimotor and voluntary postural sway measures

A.1 Background

Voluntary postural sway is a standing balance task that involves rhythmic oscillations of the body about the ankle and hip joints, which can be performed in the anterior-posterior (AP) and medial-lateral (ML) directions. Voluntary postural sway represents a convenient movement task for examining issues related to balance control because of the simple yet challenging movements, its suitability for biomechanical measurement, and the ability to experimentally manipulate the aspects of the task (Rose and Clark, 2000). Previous studies have demonstrated that voluntary postural sway is also useful for examining differences in reaction time and postural stability associated with ageing and an increased susceptibility to falls. Two recent studies found that ageing was significantly associated with slower rapidly initiated voluntary postural sway actions in multiple directions and reduced path accuracy during rhythmic voluntary sway in the AP and ML directions (Borah et al., 2007, Liaw et al., 2009). Investigations involving fallers and non-fallers have also reported that falls were associated with slower velocity and reduced path accuracy of voluntary postural sway (Ben Achour Lebib et al., 2006, Lázaro et al., 2010, Delbaere et al., 2006a).

Voluntary postural sway tasks are likely to place demands on the sensory (Maurer et al., 2001, Radhakrishnan et al., 2010), cognitive (Teasdale et al., 1993), and muscular (Duarte and Zatsiorsky, 2002) systems because of the challenge to stability associated with large movements of the centre of mass (COM) towards the limits of balance. However, there has been little to no investigation of the sensorimotor correlates of voluntary postural sway task measures. The two studies which have examined the sensorimotor correlates of task performance involved the maintenance of static
leaning postures rather than dynamic voluntary postural sway oscillations. Fujiwara and colleagues (1982) measured the forward and backward maximum lean amplitudes of 239 adults aged 20-79 years and found that centre of pressure (COP) path length whilst sustaining the leans was negatively correlated with ankle strength. Similarly, Melzer and colleagues (2009a) found that maximum AP COP lean range was positively correlated with ankle strength but not tactile two-point discrimination of the big toe. While these studies provide some evidence to suggest that ankle strength contributes importantly to the maintenance of maximum leans, the physiological mechanisms underlying the performance of voluntary postural sway and the specific sensorimotor factors which may contribute to ageing and falls-risk differences in voluntary postural sway performance remain unknown.

The purpose of this additional data analysis was to evaluate the relationships between the 17 subtests of the long-form Physiological Profile Assessment (PPA) with the voluntary postural measures that were identified as the strongest predictors of group status from experiments 3 and 4.

A.2 Analysis

Complete sets of PPA and voluntary postural sway data were available for 12 young adults and 51 older adults (N = 63). For details of the specific PPA subtests see Lord et al (2003). The voluntary postural sway measures that were evaluated for their relationships with the PPA subtests included reaction time to initiate AP voluntary sway, reaction time to terminate ML voluntary sway, target AP COP-COM separation during the initiation of AP voluntary sway, and target ML COP-COM separation during continuous voluntary ML sway. Strength of association between PPA subtests and voluntary postural sway measures was determined using correlation coefficients provided by Pearson’s product-moment correlation analysis. Linear relationships between variables were confirmed by visual inspection of scatter plots. Forward stepwise multiple regression analysis with standardised beta weights was also used to determine the relative importance of the PPA subtests in explaining variance in the voluntary postural sway measures. Scatter plots of the residuals revealed no violations.
of linearity, normality, or homoscedasticity. Collinearity diagnostics also revealed no significant indications of multicollinearity (Belsley et al., 1980). Age was used as a covariate for all correlation and regression analyses. Significance was accepted at \( P < .01 \). Statistical analyses were performed using SAS version 9.1.

A.3 Results

Correlation analysis revealed that reaction time to initiate AP sway was significant correlated with three postural sway tests, tactile sensitivity, dorsiflexion strength, hand press reaction time, foot press reaction time, and coordinated stability \((P < 0.01)\). Three postural sway tests, depth perception, tactile sensitivity, and maximum balance range were also significantly correlated with reaction time to terminate ML postural sway \((P < 0.01)\). There were no significant correlations between target AP COP-COM separation during the initiation of AP voluntary sway and the PPA subtests, however the correlation with quadriceps strength approached significance \((P = 0.055)\). Target ML COP-COM separation during continuous voluntary ML sway was significantly correlated with coordinated stability \((P < 0.01)\), and approached significance for foot press reaction time \((P = 0.018)\), and sway with eyes open on foam \((P = 0.024)\). The correlation coefficients for these analyses are presented in Table A1.

Stepwise multiple regression analysis revealed that hand press reaction time, sway with eyes open on floor, sway with eyes closed on floor, sway with eyes open on foam, maximum balance range, and low contrast visual acuity were significant and independent predictors of AP sway initiation reaction time \((F_{7,55} = 12.24, P < 0.001;\) Model \(R^2 = 0.61)\). Standardised beta weights indicated that hand press reaction time accounted for the highest proportion of variance in AP sway initiation reaction time, and that low contrast visual acuity and sway with eyes open on floor explained the least amount of variance. For reaction time to terminate ML voluntary sway, depth perception, foot press reaction time, sway with eyes closed on foam, and maximum balance range were significant and independent predictors \((F_{5,57} = 9.27, P < 0.001;\) Model \(R^2 = 0.45)\). The standardised beta weights indicated that postural sway with
### Table A.1

Age-adjusted sensorimotor and balance correlates of voluntary postural sway measures for the young and older adults (N = 63).

<table>
<thead>
<tr>
<th>PPA Subtests</th>
<th>Initiation RT AP</th>
<th>Initiation AP COP-COM</th>
<th>Termination RT ML</th>
<th>CVS ML COP-COM</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>R</td>
<td>Beta</td>
<td>R</td>
<td>Beta</td>
</tr>
<tr>
<td>Visual acuity high contrast</td>
<td>-0.16</td>
<td>-</td>
<td>0.16</td>
<td>-</td>
</tr>
<tr>
<td>Visual acuity low contrast</td>
<td>0.05</td>
<td>-0.16</td>
<td>0.04</td>
<td>-</td>
</tr>
<tr>
<td>Visual contrast sensitivity</td>
<td>-0.22</td>
<td>-</td>
<td>-0.05</td>
<td>-</td>
</tr>
<tr>
<td>Depth perception</td>
<td>0.15</td>
<td>-</td>
<td>0.05</td>
<td>-</td>
</tr>
<tr>
<td>Tactile sensitivity</td>
<td>0.36*</td>
<td>-</td>
<td>0.16</td>
<td>-</td>
</tr>
<tr>
<td>Proprioception</td>
<td>-0.01</td>
<td>-</td>
<td>-0.10</td>
<td>-</td>
</tr>
<tr>
<td>Quadriceps strength</td>
<td>-0.24</td>
<td>-</td>
<td>0.25</td>
<td>0.26</td>
</tr>
<tr>
<td>Hamstrings strength</td>
<td>-0.25</td>
<td>-</td>
<td>0.14</td>
<td>-</td>
</tr>
<tr>
<td>Ankle strength</td>
<td>-0.38*</td>
<td>-</td>
<td>0.12</td>
<td>-</td>
</tr>
<tr>
<td>Hand press RT</td>
<td>0.52*</td>
<td>0.44</td>
<td>-0.11</td>
<td>-</td>
</tr>
<tr>
<td>Foot press RT</td>
<td>0.34*</td>
<td>-</td>
<td>0.01</td>
<td>-</td>
</tr>
<tr>
<td>Sway EO floor</td>
<td>0.07</td>
<td>-0.17</td>
<td>0.07</td>
<td>-</td>
</tr>
<tr>
<td>Sway EC floor</td>
<td>0.49*</td>
<td>0.33</td>
<td>-0.02</td>
<td>-</td>
</tr>
<tr>
<td>Sway EO foam</td>
<td>0.41*</td>
<td>0.28</td>
<td>-0.15</td>
<td>-</td>
</tr>
<tr>
<td>Sway EC foam</td>
<td>0.46*</td>
<td>-</td>
<td>-0.07</td>
<td>-</td>
</tr>
<tr>
<td>Max balance range</td>
<td>-0.25</td>
<td>-0.21</td>
<td>0.08</td>
<td>-</td>
</tr>
<tr>
<td>Coordinated stability</td>
<td>0.38*</td>
<td>-</td>
<td>-0.04</td>
<td>-</td>
</tr>
</tbody>
</table>

* P < 0.01

Abbreviations: R = correlation coefficient; Beta = standardised beta weight; RT = reaction time; EO = eyes open; EC = eyes closed.
eyes closed on foam and maximum balance range were of greatest importance in explaining variance in ML sway termination reaction time. Quadriceps strength was the only significant and independent predictor of target AP COP-COM separation during the initiation of AP sway ($F_{2,60} = 7.00, P = 0.002$; Model $R^2 = 0.19$). Target ML COP-COM separation during continuous voluntary ML sway was predicted by proprioception, foot press reaction time, and coordinated stability ($F_{4,58} = 4.03, P < 0.01$; Model $R^2 = 0.22$), with coordinated stability explaining the highest proportion of variance. The standardised beta weights for the significant predictor variables are displayed in Table A1.

A.4 Interpretation

The findings of the analysis revealed that the initiation and termination voluntary postural sway reaction times were together correlated with 10 of the 17 PPA subtests, particularly those relating to reaction time, tactile sensitivity, postural sway, and dynamic balance. These results provide further evidence to suggest that voluntary postural sway reaction times represent a composite measure of sensorimotor and balance function. As the hand press reaction time test was most strongly related to AP sway initiation reaction time, the primary physiological mechanism underlying changes in the ability to rapidly initiate voluntary postural sway may be slower cognitive processing. In contrast, the reaction time to terminate voluntary postural sway was best predicted by postural sway with eyes closed on foam and maximum leaning range, suggesting that the ability to terminate voluntary postural sway depends more on global dynamic balance control rather than the individual physiological systems measured by the PPA. In general, the COP-COM separation measures were generally weakly associated with the PPA subtests. Therefore, COP-COM separation during voluntary postural sway tasks may capture different information about a person’s postural stability compared with the PPA tests. Taken together, the results of this analysis provide some useful insights into the role of physiological systems during voluntary postural sway tasks. For a more detailed discussion of the findings of this analysis refer to section 7.3 of the thesis.
REFERENCES


CHEN, H.-C., ASHTON-MILLER, J., ALEXANDER, N. & SCHULTZ, A. (1994a) Age effects on strategies used to avoid obstacles. *Gait Posture*, 2, 139-146.


References


References


References


References


TODD, C. & SKELTON, D. (2004) What are the main risk factors for falls among older people and what are the most effective interventions to prevent these falls? Copenhagen, WHO Regional Office for Europe.


TOUPET, M., GAGEY, P. M. & HEUSCHEN, S. (1992) Vestibular patients and aging subjects lose use of visual input and expend more energy in static postural control. IN VELLAS, B., TOUPET, M., RUBENSTEIN, L., ALBAREDE, J. L. &


