Gait variability, stride dynamics and falls risk in community dwelling older women

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A thesis submitted in total fulfillment of the requirement of the degree of Doctor of Philosophy

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February 2010

Statement of sources

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Abstract

This thesis investigated measures of walking instability in older adults with the aim to establish if stride dynamics and gait variability can predict falls in active older women. Falls are a major problem for older adults and the majority occur when walking. The identification of markers of walking instability that predict falls, particularly in active and healthy older adults, would help prevent injury, loss of independence and institutionalisation. Three studies were conducted as part of this thesis. These studies investigated the effect of walking protocol on measures of gait variability in healthy adults and examined the relationship between stride dynamics, gait variability and falls in older adults. Women were used in each study due to their higher incidence of falls and falls-related injury (Stevens & Sogolow, 2005).

Study 1 and 2 recorded spatial, temporal and gait variability data in older (age range 57 to 79 years) and younger women (age range 19 to 21 years) screened for conditions that might impact upon balance or walking. Gait data were collected with an 8.1m GAITRite mat for 10 trials of discrete single walks and 10 laps of a continuous circuit, presented in random order. Study 1 examined the test-retest reliability and systematic bias of data recorded during repeated single and continuous over-ground walking trials over two separate test sessions that were seven days apart (median ± SD, 7±1.58d). Paired *t* tests, intraclass correlation coefficients, standard errors of measurement, and coefficients of variation were calculated. Study 2 investigated if gait variability data captured during repeated single over-ground walking differed from variability data captured during continuous over-ground walking. To quantify variability, standard deviation and coefficients of variation were calculated for each gait parameter, and paired *t* tests were used to compare the measures of variability recorded for each walking protocol.

A major finding from the first two studies was that gait parameters, including gait variability, differed between walking protocols. Study 1 showed that although both continuous and repeated single walking protocols were reliable, the continuous over-ground walking protocol produced

less bias (19% of gait parameters) in test-retest spatiotemporal gait data compared to the single trial condition (65% of gait parameters). Between-protocol differences were more apparent for the older than younger women, with 77% of the gait parameters showing bias in the single trial condition. In contrast, no systematic bias was found in the continuous condition for older adults.

In study 2, walking protocol differences were found between the gait variability data. Compared with a continuous over-ground walking protocol, a repeated single over-ground protocol resulted in increased variability of velocity, step length and stride length data ($p<0.01$) for the older women. In the younger women, increased variability of velocity ($p\leq0.02$), step length ($p=0.04$), stride time ($p\leq0.02$) and step time ($p=0.02$) were found for the single walking trials. The findings from studies 1 and 2 suggest that a continuous protocol may be more stable and may detect gait changes more readily, especially for older women.

Based on the outcomes of study 1 and 2, gait data recorded using a continuous walking protocol were used for the major analyses in study 3 which examined gait variability and stride dynamics. Additionally, data were also recorded from the equivalent number of repeated single walking trials to investigate the influence of walking protocol upon gait variability and falls. Ninety seven active and healthy community dwelling women (mean age=68.73 ± 7.07 years) underwent screening procedures and completed seven minutes of walking around a continuous circuit. Gait data were collected with an 8.1m GAITRite® mat and with two tri-axial Crossbow® accelerometers. A small subset (n=12, mean age=67.17 + 5.27) of participants also attended on a second visit one week following their initial testing session to evaluate the test-retest reliability of the accelerometer data. Participants were then followed prospectively for one year to record fall incidence. Differences in physical (demographic and screening), balance, gait variability and stride dynamic measures between fallers (one or more falls) and non-fallers were examined using Multivariate Analyses of Variance (MANOVAs) and independent samples *t* tests. Between-leg differences in stride dynamics were assessed using a paired samples *t* test. To evaluate the ability

of stride dynamics and gait variability to predict future fallers, direct logistic regression was performed. All analyses were repeated with the sample stratified into non-fallers, single fallers (one fall) and multiple fallers (two or more falls), as well as into multiple fallers and a combined group of single and non-fallers, to investigate the effect of multiple falls as an independent variable. Finally, to evaluate whether walking protocol influenced study outcomes, all between-group and prediction analyses were again repeated using data collected from the repeated single walking protocol.

The major finding of this study was that inter-limb dynamics were altered in fallers. Specifically, inter-limb differences ($p \leq 0.04$) were found in the fractal scaling index of fallers (one or more falls) aged over 70 years, and multiple fallers (two or more falls) aged over 55 years, but not in non-fallers, single fallers or the combined group of single and non-fallers. Interestingly, no differences ($p > 0.05$) were found in any physical, balance, gait variability or other stride dynamic measures between those who fell in the subsequent year and those who did not fall. Additionally, no gait variable predicted future falls in the sample of active older women. Similar outcomes were found when data from a repeated single walking protocol were used, and when the sample was stratified in non-fallers, single fallers, multiple fallers and a combined group of single and non-fallers. Therefore, despite no observable difference in other common measures of intrinsic falls risk, control of inter-limb dynamics was reduced in active and otherwise healthy older fallers and multiple fallers. This outcome suggests that inter-limb dynamics could provide a clinically sensitive and possible early detection marker of gait instability and falls risk in high functioning older adults prior to evidence of change in other measures of physical, balance or gait function, including gait variability.

Acknowledgements

I am deeply indebted to the wonderful support of my co-supervisors. Their rigour, insight and intelligence has lifted my own abilities and made me strive to achieve more in my career. My love of research, and in particular, my fascination with human gait was first instilled by Dr Noel Lythgo who helped me to develop a keen critical eye and a deeper understanding of what I am measuring. Noel, I thank you for giving me these first opportunities in research and for all of the continued support and encouragement you have provided along the way. Professor Keith Hill's role in my career has also been crucial. The decision to undertake doctoral studies with a concurrent clinical career was a difficult one, but meeting Keith and seeing not only his transition between the clinical and research worlds, but also his wonderful ability to make a real difference in the lives of older people motivated me to take this next step. Keith, your impact in this field has been an ongoing inspiration and I feel honored to have had you guiding me in my research studies. Together, Keith and Noel have helped shaped my research and my research career, and I thank them both without qualification.

My principal supervisors at ACU have also provided me with tremendous support. In the latter stages of my studies, my wonderful research mentor Professor Geraldine Naughton commenced supervision. Her thoughtful and tireless input was instrumental in helping guide me in these final months. Jeri, thank you so much for your support and for sharing your passion of research. Your valuable critique brought a comprehensive perspective to my writing that has truly helped shape my thesis. Associate Professor Wayne Maschette was also very helpful during my candidature, particularly during the early stages. He facilitated the initial administration of my ACU employment and my doctoral studies, and was crucial in the planning and recruitment for my first two projects. Thank you Wayne for being an important sounding board as I developed the

foundations of my research.

Myself, and the research environment within the School, have benefitted greatly from the research knowledge of Chris, Morgan, Cam, and in particular, Justin. Thanks to each of you for sharing your experience and your invaluable advice along the way. And to Corey, Trent and Bianca, I wish you all the success in your future research studies.

As with most large projects, there are a number of individuals who have helped behind the scenes on my path to completion. In particular though, I'd like to thank Professor David Morgan from Royal Talbot Rehabilitation Centre, who was instrumental to my final study and in the analysis of my acceleration data. David, thank you for giving me your time and expertise, oftentimes at night, on weekends and sometimes even on your holidays. It has been an absolute pleasure working with you and gaining an insight into your knowledge of signal analysis, programming and genealogy amongst many other things.

To all of the amazing older people who volunteered to assist in these studies, and to the many others who offered but were unable, you deserve a special thank you. You gave up your time for the sole aim of helping to prevent a serious health problem occurring to other people. I admire your selflessness and I thoroughly enjoyed my small window into your rich and diverse lives. And to my two fantastic research assistants, Tania and Melissa, thank you for your help with the testing phase of my final study. Some of those sessions were long and you were both a constant source of energy and professionalism. With the two of you now stepping further into the research arena, I'd be more than happy to return the favour and assist in any way if I can.

My family always worked hard to provide me with more than they had. So to my wonderful mother and father, I thank you both very much for the support and unqualified love you've always given me. I love you both very much and I know I am who I am thanks to you.

Finally, about a year before finishing this thesis I was blessed with the most beautiful little boy in the world. And whilst the timing wasn't great for finishing a thesis, it's thanks to his remarkable mother that I was able to. To my beautiful wife, it's been a busy time and you've had to shoulder most of the load. I can't thank you enough for supporting me through these past years and for being an unending pillar of patience, love, strength and wit. I love you dearly.

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Glossary of abbreviations

AbsRC	Absolute reliability coefficient
AP	Antero-posterior
AUD	Australian dollars
BOOMER	Balance outcome measure for elder rehabilitation
CI	Confidence intervals
cm	Centimetre
cm/s	Centimetres per second
CTSIB	Clinical test of sensory interaction and balance
CV	Coefficient of variation
CPG	Central pattern generator
DFA	Detrended fluctuation analysis
EMG	Electromyography
ES	Effect size
fMRI	Functional magnetic resonance imaging
GPS	Global positioning system
GTO	Golgi tendon organs
HLGD	Higher level gait disturbance
Hz	Hertz
ICC	Intraclass correlation coefficient
kg	Kilogram
LPI	Lateral preference inventory
m	Metre
MANOVA	Multiple analysis of variance
MET	Melbourne edge test
mmHg	Millimetres of mercury

MMSE	Mini mental state examination
MRI	Magnetic resonance imaging
m/s	Metres per second
msec	Millisecond
MLR	Mesencephalic locomotor region
PET	Positron emission tomography
RC	Repeatability coefficients
sec	Second
SEM	Standard error of measurement
SD	Standard deviation
SMA	Supplementary motor area
SPECT	Single photon emission computed tomography
SPSS	Statistical package for the social sciences
USD	United States dollars

List of publications

Refereed journals

Paterson, K., Lythgo, N. & Hill, K. (2009). Gait variability in younger and older women is altered by overground walking protocol. Age and Ageing, 38(6), 745-748.

Paterson, K. L., Hill, K. D., Lythgo, N. D., & Maschette, W. (2008). The reliability of spatiotemporal gait data for young and older women during continuous overground walking. Archives of Physical Medicine & Rehabilitation, 89(12), 2360-2365.

Submitted

Paterson, K., Morgan, D. Lythgo, N. & Hill, K. (submitted – 2nd review). Reliability and validity of an accelerometer to collect temporal and stride dynamic gait data. Archives of Physical Medicine and Rehabilitation.

Paterson, K. L., Hill, K. D., Lythgo, N. D., & Maschette, W. (submitted – 2nd review). Stride dynamics, gait variability and falls risk in healthy older women. Gait and Posture.

1 General introduction

1 General introduction

1.1 Statement of the problem

Falls in older adults are a major global public health burden. In the United States, Australia and many European countries for example, approximately one third of people aged over 65 years are reported to fall yearly (Blake, et al., 1988; Lord, Ward, Williams, & Anstey, 1993; Tinetti, Speechley, & Ginter, 1988). In South America, incidence rates are between one fifth to one third of older adults (Reyes-Ortiz, Al Snih, & Markides, 2005), and rates for Asia are around 20% (Yoshida & Kim, 2006; Yu, et al., 2009). Falls are the leading cause of injury deaths in older adults, predominantly through traumatic brain and lower extremity injuries suffered through the fall (Kochanek, Murphy, Anderson, & Scott, 2004; Stevens, Corso, Finkelstein, & Miller, 2006). Additionally, rates of non-fatal injuries requiring hospitalisation from falls are high, ranging between 1.6 and 7.3 per 10,000 people aged over 60 years in many developed countries (Cripps & Carman, 2001; Orces, 2009; Scott, Pearce, & Pengelly, 2009). The incidence of falls and falls-related injury is higher in women and has been shown to increase with age (Orces, 2009; Stevens & Sogolow, 2005).

The physical consequences of falls are associated with a high financial and social burden to nations and the individual. Researchers in the United States estimated the direct financial costs of falls to be approximately $USD20 billion (Stevens, et al., 2006). In Australia the direct and indirect cost of falls have been estimated to exceed AUD$1,000 million (Moller, 1998). Falls also increase levels of fear and anxiety (Howland, et al., 1993) leading to a restriction in levels of physical activity (Moore & Ellis, 2008), resulting in a heightened risk of further falls (Clough-Gorr, et al., 2008), nursing home admission (Tinetti & Williams, 1997) and death (Leibson, Tosteson, Gabriel, Ransom, & III, 2002).

Many falls associated with ageing are considered preventable. To this end, the World Health Organization developed an international strategic framework aimed at encouraging ‘active

ageing' to prevent disability and subsequent falls in older adults (World Health Organisation, 2007). A salient component of this framework was the identification of biological markers of decline associated with a heightened falls risk. Many of these determinants, such as age and gender, are non-modifiable, while others are modifiable with appropriate implementation of individual and public policies. Consequently, there are pressing needs to establish markers predictive of early falls to attenuate risk.

Much of the work aimed at identifying biological markers of falls risk has examined quantifiable changes in the walking pattern of older adults. Outdoor falls are more prevalent than indoor falls (Bergland, Jarnlo, & Laake, 2003; Li, et al., 2006), and the majority of outdoor falls occur during walking (Li, et al., 2006). Not surprisingly, outdoor falls occur more often in active older adults (Bergland, et al., 2003; Li, et al., 2006). Furthermore, compared with indoor falls, outdoor falls are associated with greater injury incidence (Nachreiner, Findorff, Wyman, & McCarthy, 2007) and severity (Luukinen, et al., 2000). Therefore, the early identification of risk in active older adults is particularly valuable since there is potential to prevent outdoor falls and avoid subsequent falling, morbidity and institutionalisation in an otherwise healthy individual.

It is postulated that markers of walking instability might assist in identifying future fallers. The field of gait analysis has the potential to contribute meaningfully to this goal. The mechanistic approach of gait analysis integrates biomechanical knowledge with the underlying neurological basis of locomotor control, and offers an expansive platform with which to investigate walking stability. Despite extensive research in this field, a need exists for clinically useful markers capable of predicting future falls with precision in otherwise healthy older adults. Two measures of walking instability, gait variability and stride dynamics, have evidence suggesting they could potentially be markers of early falls risk.

Gait variability quantifies the stride to stride fluctuations that occur during walking. Increased variability in many common gait parameters is known to predict future falls in community

dwelling older adults (Hausdorff, Rios, & Edelberg, 2001; Maki, 1997; Verghese, Holtzer, Lipton, & Wang, 2009). However, these prospective studies have examined older adults already showing evidence of mobility problems. As such the ability of gait variability to predict a potential for early falls in a more active older population remains unknown. Additionally, a diverse range of methodologies were employed to collect variability data in retrospective and prospective gait variability studies. Given there have been different fall-related outcomes reported in these studies, it is possible that variability is affected by methodological issues such as the walking protocol employed to capture gait variability data, hence contributing to the different outcomes. As such, the influence of walking protocol upon measures of gait variability and fall identification requires investigation.

Stride dynamics provide a measure of the change in stride fluctuations over time and have been associated with falling in clinical populations (Herman, Giladi, Gurevich, & Hausdorff, 2005). Moreover, the dynamics of walking break down in healthy ageing (Hausdorff, et al., 1997). This type of analysis therefore has the potential to produce new information about the underlying dynamic structure of neuromotor mechanisms controlling walking and may assist in identifying age-related changes in gait stability. To date, the stride dynamics of active and otherwise healthy older fallers and non-fallers have not been investigated. Therefore it is unknown if age-related alterations in dynamic coordination decrease stability and increase falls risk in older adults.

Considerable clinical merit exists in indentifying markers of early falls risk in otherwise healthy older adults prior to the occurrence of secondary changes such as fall-related injuries or fear of falling. To date, gait variability and stride dynamics have only been investigated in older adults already displaying signs of mobility limitations or in adults diagnosed with a mobility disorder. Despite these measures having the potential to identify falls risk in more frail populations, research is yet to establish whether gait variability and stride dynamics are sensitive markers of gait instability and early falls risk in higher functioning active older adults.

1.2 General aim of the study

The aim of this study was to advance knowledge about the role of gait variability and stride dynamics in an active older adult population, and to assess whether these measures predict falls in active and otherwise healthy older women.

1.3 Specific aims of the study

The specific aims of this study were to:

- Examine systematic bias and establish test-retest reliability of gait data collected with a continuous over-ground walking protocol in younger and older women.
- Employ two common walking protocols to record gait data in younger and older women and evaluate whether walking variability is altered by gait methodology.
- Evaluate differences in walking variability between a group of active older female fallers and non-fallers, determined using a 12 month prospective study design, and examine the influence of walking protocol on any observed differences.
- Investigate differences in a measure of stride dynamics, the fractal scaling index, between active older female fallers and non-fallers.
- Investigate the effect of inter-limb coordination on falls by examining the symmetry of the fractal scaling index, and evaluate whether inter-limb differences exist in active older female fallers and non-fallers
- Determine whether gait variability and stride dynamics are sensitive markers of gait instability by evaluating their ability to predict future falls in a group of active and otherwise healthy older women, and examine the influence of walking protocol upon this prediction accuracy.

1.4 Limitations

Various factors have the potential to influence locomotor control, walking stability and falls in active older women, many of which are beyond the scope of this investigation. As a consequence of these limitations, the studies contained herein did not examine:

- Motivational and psychological status of participants.
- Social, economic, or cultural influences.
- Nutritional status of participants.
- Potential influence of genetic factors or ethnicity upon walking.
- Lifestyle and physical environment of participants over a prospective 12 month period.
- Types of activities participants engaged in prior to and during the testing period.

1.5 Delimiters

Due to the presence of limiting factors, this investigation was delimited to the study of:

- Common spatial and temporal walking parameters including measures of gait variability and walking dynamics.
- Healthy young women aged between 18 and 35 (studies 1 and 2).
- Healthy older women aged between 55 and 90 years (studies 1 to 3).
- Volunteers with mobility levels enabling attendance at the testing laboratory.
- Women able to meet the selection criteria outlined in Chapter 3.2.2 and pass the screening measures outlined in Chapter 3.4.1.

2 Literature review

2 Literature review

This chapter is comprised of three major sections that critically review the literature pertaining to falls in older adults (Chapter 2.1), normal and ageing gait patterns (Chapter 2.2) and gait variability, stride dynamics and falls risk (Chapter 2.3).

2.1 Falls

2.1.1 Preamble

Falls have a major impact upon the physical and social well-being of older adults. Approximately one-third of community-dwelling Australians over the age of 65 fall each year (1993; Morris, et al., 2004), with this increasing to almost half of those aged over 75 (Hill, Schwarz, Flicker, & Carroll, 1999; Tinetti, Speechley, & Ginter, 1988). The high incidence of falling exhibited by older adults is directly associated with a reduction in quality of living through increased morbidity (e.g., hip fractures), loss of living independence and reduced physical activity levels (Hill, et al., 1999; Murphy, Williams, & Gill, 2002; Stevens, 2005; Stevens, Corso, Finkelstein, & Miller, 2006). Moreover, after an initial fall, the risk of further falls and subsequently morbidity and mortality increases (Clough-Gorr, et al., 2008; Kochanek, Murphy, Anderson, & Scott, 2004). Consequently, the prevention of falls through the early identification of at-risk individuals is an important social and medical objective. The following section will examine the costs of falls in more detail, and will finish with a review of falls-related risk factors for older adults.

2.1.2 Definition of a fall

For this study, a fall is defined as “inadvertently coming to rest on the ground, floor or other lower level, excluding intentional change in position to rest in furniture, wall or other objects” (World Health Organisation, 2007, p. 1).

2.1.3 Financial consequences of falls

Falls have a substantial economic impact on the Australian health care system. In the 1993-94 period, fall-related injuries in Australians aged over 65 were estimated to cost AUD$406.4 million (Mathers & Penm, 1999). If indirect and non-medical costs are included in the estimation, the cost is more than AUD$1,000 million (Moller, 1998). More recently, Hendrie and colleagues (2004) reported that the cost of non-fatal fall injuries requiring admission to an Australian emergency department was $AUD86.4 million.

In many developed countries the financial cost of injuries due to fatal and non-fatal falls is much higher. In the United States for instance, the direct medical costs for hospital, emergency department and outpatient care was $USD19 billion dollars for non-fatal falls and USD$0.2 billion dollars for fatal falls in 2000 (Stevens, et al., 2006). In the United Kingdom, fall-related injuries in older adults aged over 60 years cost almost £1 billion in acute emergency department and longer-term hospital care expenses in 1999 (Scuffham, Chaplin, & Legood, 2003). In smaller countries such as Ireland, health service costs for fall-related injuries in a single hospital were reported to be €10.8 million per year (Cotter, Timmons, O'Connor, Twomey, & O'Mahony, 2006).

With the projected increase in both the Australian and global population, particularly in the over 65 age group, coupled with expected improvements in life expectancy over the coming decades, health care costs associated with falling are likely to rise dramatically (Strategic Injury Prevention Partnership, 2000). Whilst exact figures are difficult to predict, a 1996 study estimated that falls-related health care costs in the United States could reach USD$43.8 billion by the year 2020 (Englander, Hodson, & Terregrossa, 1996). Other work has estimated that medical costs from falls-related hip fractures would reach USD$16 billion in the United States by 2040 (Cummings, Rubin, & Black, 1990). In Western Australia, costs of falls presenting to emergency departments were projected to reach AUD$180 million by 2021 (Hendrie, et al., 2004), and Australia wide, the projected cost of health care for falls-related injuries is expected to triple to

AUD$1375 million by 2051 (Moller, 2003).

2.1.4 Behavioural and psychological consequences of falls

Falling often results in negative psychological consequences that reduce quality of life for older adults. For example, individuals who have fallen often report reduced confidence and decreased perceived ability in activities of daily living (Yardley & Smith, 2002), an increased fear of falling (Jørstad, Hauer, Becker, & Lamb, 2005), further activity restrictions (Murphy, et al., 2002) and in some instances, anxiety during standing and walking activities (Bhala, O'Donnell, & Thoppil, 1982; Murphy & Isaacs, 1982). Even in the absence of a physical injury sustained by falling, older adults often lose confidence in their physical abilities and restrict physical activity (Moore & Ellis, 2008; Yardley & Smith, 2002). It has been suggested that the physical outcomes from these psychological consequences may be more disabling than the actual fall (Salkeld, et al., 2000).

Fear has become ubiquitous in the older adult population, with between one quarter to one half of those over 60 years reporting some fear of falling (Arfken, Lach, Birge, & Miller, 1994; Howland, et al., 1998; Howland, et al., 1993; Lach, 2005; Tinetti, De Leon, Doucette, & Baker, 1994). Amongst those who have fallen, 29% of non-injured fallers reported some fear of falling, whereas 41% of injured fallers were afraid of falling (Howland et al. 1993). Fear of falling has also been reported in non-fallers. Approximately 20% of older non-fallers have reported to be very or somewhat afraid of falling (Howland, et al., 1993; Lach, 2005). Interestingly, this fear is greater than other anxieties such as being robbed or experiencing financial problems (Howland, et al., 1993). Prevalence of fear is also greater with advancing age, with between 14-29% of 65-70 year olds reporting some fear of falling (Arfken, et al., 1994; Lach, 2005). This rises to nearly 56% for those over the age of 80 (Lach, 2005).

Fear and its associated loss of confidence can lead to restrictions in the level of physical and functional activities (Cumming, Salkeld, Thomas, & Szonyi, 2000; Lachman, et al., 1998; Tinetti, et

al., 1994; Tinetti, Richman, & Powell, 1990; Vellas, Wayne, Romero, Baumgartner, & Garry, 1997). For instance, fear of falling has been associated with being sedentary and a reduction in recreational or other physical activities (Bruce, Devine, & Prince, 2002; Murphy, et al., 2002). Older adults with an increased fear of falling have also been shown to have reduced performance on activities of daily living such as housework and dressing (Cumming, et al., 2000). Importantly, these behavioural changes are associated with progressive functional declines and poorer health outcomes (Cumming, et al., 2000; Martin, Hart, Spector, Doyle, & Harari, 2005; Mendes de Leon, Seeman, Baker, Richardson, & Tinetti, 1996), increased risk of future falls (Cumming, et al., 2000; Hill, et al., 1999; Pluijm, et al., 2006; Tromp, et al., 2001) and an increased rate of nursing home admission (Cumming, et al., 2000).

2.1.5 Physical consequences of falls

The increased average age in recent decades has been associated with an increase in the incidence of falls and a subsequent increase in fall-related injuries in older adults (Roudsari, Ebel, Corso, Molinari, & Koepsell, 2005). During the 2005-06 period 2,415 per 100,000 older Australians suffered a fall-related injury requiring hospitalisation, an increase of 10% over the 2003-04 period (Bradley & Harrison, 2007; Bradley & Pointer, 2008). The length of hospital stay due to a fall has also risen during this period (Bradley & Pointer, 2008). With the percentage of Australians over 65 increasing by 17% between 1991 and 1999 (Cripps & Carman, 2001), and projected to comprise a quarter of the population by 2056 (Australian Bureau of Statistics, 2008), the incidence of falls and falls-related injuries can also be expected to increase.

Falls are the leading cause of injury deaths in older adults, accounting for approximately 38% of all unintentional injury deaths (Kochanek, et al., 2004). Traumatic brain injury is the most common cause of death from a fall (46%), whereas injuries to the lower extremities account for 32% of fatal fall injuries (Stevens, et al., 2006). Falls are also the leading cause of non-fatal injuries (Stevens, 2005), with between 10 to 30% of falls resulting in moderate to severe injuries

(Alexander, Rivara, & Wolf, 1992; Dellinger & Stevens, 2006; Hill, et al., 1999; Tinetti, et al., 1988). Approximately 10 to 15% of fall-related injuries require hospitalisation (Alexander, et al., 1992; Hendrie, et al., 2004; Sattin, et al., 1990; Stevens, et al., 2006), with the most serious and the most frequent injuries again being traumatic brain injuries and hip fractures (Stevens, et al., 2006; Thomas, Stevens, Sarmiento, & Wald, 2008). In the United States, hip fractures and head trauma such as contusions and intracranial injuries accounted for 50% of all injuries in older adults hospitalised due to a fall (Roudsari, et al., 2005). In Australia the statistics are similar, with injuries to the hip and thigh accounting for the majority of fall-related injuries (31%), and head injuries the second most common outcome (17%) (Bradley & Pointer, 2008).

Injuries from non-fatal falls have serious health and quality of life consequences. Studies have shown increased mortality following a fall-related hip fracture (Rapp, Becker, Lamb, Icks, & Klenk, 2008), with up to 20% of older adult fallers dying within 12 months of the accident (Leibson, Tosteson, Gabriel, Ransom, & III, 2002; Magaziner, et al., 2000). Of the survivors, up to a quarter of formerly independent community-dwelling older adults are institutionalised for at least one year following a fall (Magaziner, et al., 2000; Magaziner, Simonsick, Kashner, Hebel, & Kenzora, 1990) and up to 13% require long term care in a nursing home (Tinetti & Williams, 1997). Older adults who have experienced a fall are also more likely to fall again in the subsequent 12 month period (Clough-Gorr, et al., 2008), further increasing the likelihood of fall-related injuries, institutionalisation and death. Clearly, early identification of falls risk factors that might lead to the prevention of falling in older adults is a major health care goal.

2.1.6 Falls Risk Factors

Studies have identified a multitude of risk factors for falls in older adults (Brown, 1999; Oliver, Daly, Martin, & McMurdo, 2004). These are commonly described as either extrinsic or intrinsic falls risk factors. Extrinsic or environmental risk factors refer to indoor or outdoor fall hazards that increase an individual's susceptibility of experiencing a fall, and include factors such as poor

lighting and uneven surfaces (Stevens, 2005). Intrinsic falls risk factors include individual characteristics that increase risk of falling, such as cardiovascular or neurological pathology and increasing age (Stevens, 2005). Falling is the consequence of a complex interaction of these extrinsic and intrinsic risks (Figure 2.1) (Tromp, et al., 2001).

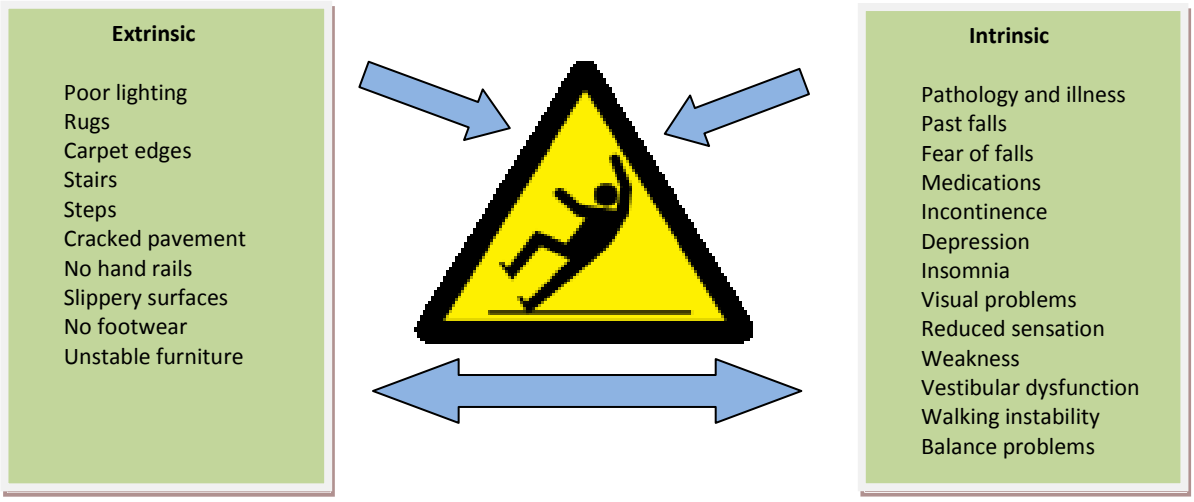


Figure 2.1. Interaction between extrinsic risk factors, intrinsic risk factors and falls.

2.1.6.1 Extrinsic falls risk factors

A number of environmental hazards have been identified to increase a person's risk of falling. For community-dwelling older adults, these hazards are typically located in and around a person's home and community, and include rugs and carpet edges, a lack of hand rails, poor lighting conditions, obstructed walkways, uneven or slippery surfaces and unstable furniture (Carter, Campbell, Sanson-Fisher, Redman, & Gillespie, 1997; Connell, 1996; Gill, Williams, Robison, & Tinetti, 1999; Li, et al., 2006; Masud & Morris, 2001). Additionally, falls in older adults have also been attributed to walking barefoot (Koepsell, et al., 2004; Menz, Morris, & Lord, 2006), in socks (Menz, et al., 2006) or in stockings (Koepsell, et al., 2004).

Indoor falls can occur in the home, in a residential setting such as an aged care facility or in shopping centres, schools or other public administration buildings (Berg, Alessio, Mills, & Tong,

1997; Bradley & Pointer, 2008; Cripps & Carman, 2001; National Ageing Research Institute, 2004). Of these, falls in the home account for nearly half of all falls requiring hospitalisation in Australians aged over 65 years (Bradley & Pointer, 2008). It is important to note however that home falls reported by Bradley and Pointer included falls occurring in the driveway, and therefore would include some outdoor falls. Residential settings are the second most common place for a fall, accounting for 22% of falls in Australia (Bradley & Pointer, 2008). Overseas, the figures for fall location are similar, with between 44% and 65% of falls occurring inside the place of usual residence, including both community and residential dwellings (Masud & Morris, 2001).

The prevalence of household fall hazards such as loose flooring or poor lighting has been shown to be greater in the homes of fallers than non-fallers (Isberner, et al., 1998). However frailty appears to mediate the extent of falls risk from environmental home hazards (Lord, Menz, & Sherrington, 2006). That is, whilst some studies have shown that only a small number of falls were attributed to home falls hazards (Norton, Campbell, Lee-Joe, Robinson, & Butler, 1997; Sattin, Rodriguez, DeVito, Wingo, & Group, 1998), it is vigorous but not frail older adults that have been found to have a greater number of home hazards (Northridge, Nevitt, Kelsey, & Link, 1995). This apparent paradox between increased mobility and falls risk might be explained by the greater exposure to environmental home hazards due to the increased activity in the more vigorous older adults (Lord, et al., 2006), a finding also observed in outdoor falls (Hill, et al., 1999; Li, et al., 2006; Speechley & Tinetti, 1991).

Outdoor falls occur more often than indoor falls (Bath & Morgan, 1999; Bergland, Jarnlo, & Laake, 2003; Li, et al., 2006; Weinberg & Strain, 1995). These falls are also associated with higher injury rates (Nachreiner, Findorff, Wyman, & McCarthy, 2007) and more serious injuries such as hip fractures (Luukinen, et al., 2000) than indoor falls. In a large American study of 2,193 community-dwelling older adults, Li and colleagues (2006) found 58% of falls occurred outdoors. These were most commonly attributed to tripping or slipping on uneven or wet surfaces.

Outdoor falls also occur more frequently in active than frail older adults (Bergland, et al., 2003; Li, et al., 2006; Speechley & Tinetti, 1991). For example, outdoor fallers were more likely to report excellent health status, no physical difficulties and not require assistance for performing activities of daily living compared to indoor fallers (Li, et al., 2006). It is likely that active older adults have greater exposure to environmental falls hazards leading to a higher incidence of outdoor falls (Lord, et al., 2006). The increased prevalence of outdoor falls in active community-dwelling older adults is especially pertinent given that the risk of further falls (Clough-Gorr, et al., 2008), hospitalisation (Alexander, et al., 1992; Hendrie, et al., 2004; Sattin, et al., 1990; Stevens, et al., 2006), and nursing home admission (Tinetti & Williams, 1997) all increase following a fall (Chapter 2.1.5).

Forty-seven percent of outdoor falls occur during walking with 34% occurring on a footpath, curb or street (Li, et al., 2006). It is possible therefore that continuous walking over longer periods, as opposed to shorter walking bouts, heightens the risk of falls in older adults. Interestingly, recent work with young adults found that more than 90% of walking bouts were for short periods of less than 100 continuous steps (Orendurff, Schoen, Bernatz, Segal, & Klute, 2008). If older adults have similar walking behaviour, the limited exposure to longer continuous walking bouts might contribute in part to the high incidence of outdoor falls. To date, no study has explored differences in the walking pattern of older adults during longer continuous compared to shorter discontinuous walking bouts. Given many studies have shown an association between age-related gait changes and falls risk (Chapters 2.1.6.2, 2.2.3 and 2.3.4), the influence of walking protocol upon age-related gait changes and measures of falls risk in older adults is worthy of investigation. This will be explored further in Chapter 2.3.6.

2.1.6.2 Intrinsic falls risk factors

A number of important intrinsic factors have been identified that increase the risk of falling in the older adult population. Of these, past falls and acute and chronic illness and disability are

often reported as leading causes (Clough-Gorr, et al., 2008; Lord, et al., 1993; Pluijm, et al., 2006; Tinetti, et al., 1988; Tinetti, Williams, & Mayewski, 1986; Tromp, et al., 2001). For example, having experienced a previous fall increases the risk of both future single and multiple falls, with reported odds ratios of between 2.6 and 5.5 (Clough-Gorr, et al., 2008; Tromp, et al., 2001). Similarly, a diagnosis of stroke, arthritis and Parkinson's disease have been shown to increase a person's risk of falling (Dolinis, Harrison, & Andrews, 1997). Other work has shown that 14% of unexplained falls were attributable to cardiovascular disorders (Montero-Odasso, Schapira, Duque, et al., 2005), whilst urinary incontinence (Brown, et al., 2000; Tromp, et al., 2001), insomnia (Avidan, et al., 2005) and depression (Whooley, et al., 1999) have also been identified as important intrinsic falls risk factors.

Treatment of chronic disease and other illness through prescription medication has also been associated with falling in older adults. Use of more than four medications, or use of benzodiazepines or antiepileptic drugs, has been consistently reported to increase falls risk in older adults (Ensrud, et al., 2002; Leipzig, Cumming, & Tinetti, 1999; Mustard & Mayer, 1997; Neutel, Perry, & Maxwell, 2002; Petty, et al., 2010; Stenbacka, Jansson, Leifman, & Romelsjo, 2002). Further, reducing the number of prescribed medications in older adults has been shown to lower falls risk (Pit, et al., 2007) and a reduction in psychotropic medication use has also been associated with a lower risk of falls (Campbell, Robertson, Gardner, Norton, & Buchner, 1999).

Whilst underlying pathology and pharmacological treatment may predispose an individual to falling, variations in the severity of the condition, and in many cases the absence of a recognised medical diagnosis, may render this method of falls risk classification problematic. For example, Clough-Gorr and colleagues (2008) found that preclinical disability can prospectively identify individuals at an increased risk of falling in the absence of a recognised medical condition. Other researchers (Lord, Menz, & Tiedemann, 2003) therefore have argued against this "disease-oriented approach" to falls risk identification, where falls risk is attributed to a specific medical

diagnosis. Rather, they propose a "physiologic approach" to falls risk assessments whereby the underlying pathophysiology is evaluated, and deficits in specific physiological systems are associated with an increased falls risk. As an example of this approach, the presence of peripheral neuropathy in an individual with diabetes would be more likely to provide useful information on falls risk than the diagnosis of diabetes.

Based on such an assessment of physiological systems, a number of intrinsic risk factors have been identified. Deficits in visual acuity (Dargent-Molina, et al., 1996; Felson, Anderson, & Annan, 1989; Ivers, Norton, Cumming, Butler, & Campbell, 2000; Tromp, et al., 2001), visual contrast sensitivity (Cummings, et al., 1995; de Boer, et al., 2004; Lord & Dayhew, 2001) and depth perception (Cummings, et al., 1995; Ivers, et al., 2000; Lord & Dayhew, 2001) have all been associated with an increased risk of falling, with the latter two being the most important visual risk factors (Lord, 2006; Lord & Dayhew, 2001). Other work has shown that reduced strength (Graafmans, et al., 1996; Hill, et al., 1999; Lord, et al., 1993; Pluijm, et al., 2006), balance deficits (Di Fabio & Anacker, 1996; Djaldetti, Lorberboym, & Melamed, 2006; Hill, et al., 1999; Tinetti, et al., 1986), reduced peripheral sensation (Richardson & Hurvitz, 1995), vestibular dysfunction and dizziness (Brandt & Dieterich, 1993; Pluijm, et al., 2006) and foot problems (Menz & Lord, 2001; Menz, Morris, & Lord, 2006; Pluijm, et al., 2006) are risk factors for falls in older adults. Although each of these risk factors in isolation increases the chance of a fall, risk of single and multiple falls increases with greater numbers of falls risk factors (Figure 2.2) (Graafmans, et al., 1996; Lord, et al., 2003; Tinetti, et al., 1986). Further, the prevalence of many of these falls risk factors has been reported to increase with age (Stevens, 2005).

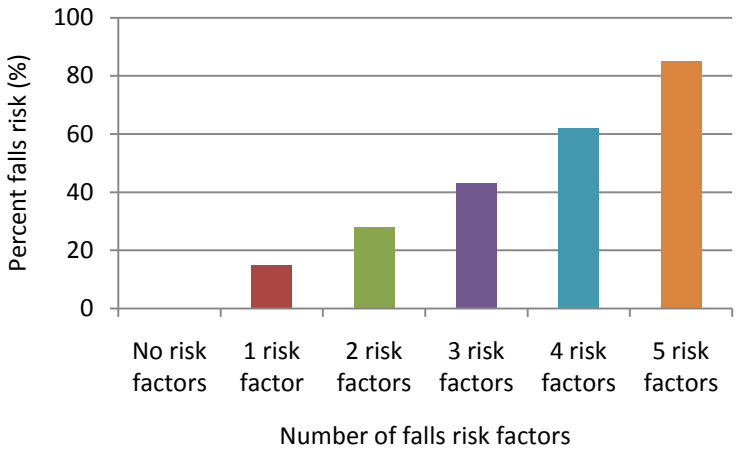


Figure 2.2. Probability of two or more falls in a 28 week follow up period. Risk factors include immobility, poor mental state, orthostatic hypotension, dizziness and stroke. Based on data from Graafmans et al. (1996).

An association between reduced mobility, functional limitations and increased falls risk has also been shown. For example, Shumway-Cook and colleagues (2000) found that reduced mobility, as measured using the timed up and go, was successful at identifying community-dwelling older fallers. Other work has reported that three or more functional limitations (Pluijm, et al., 2006), abnormal or poor mobility (Beauchet, Dubost, Herrmann, Rabilloud, et al., 2005; Tinetti, et al., 1986), reductions in activity levels and reduced measures of physical performance (Tromp, et al., 2001) increased the risk of falling in older adults. However, as outlined in Chapter 2.1.4, it is possible that fear of falling is a confounding factor in the association between mobility, function and falls (Cumming, et al., 2000; Lachman, et al., 1998; Tinetti, et al., 1994; Tinetti, et al., 1990; Vellas, et al., 1997).

The relationship between decreased mobility and falls risk may also be associated with age-related gait changes. For example, self reported and clinically diagnosed gait abnormalities have been reported as risk factors for falling in older adults (Stevens, 2005; Tinetti, et al., 1988; Verghese, Holtzer, Lipton, & Wang, 2009). Similarly, many studies have shown that alterations in the mean value of a number of gait parameters can retrospectively (Besser, et al., 2000; Guimaraes & Isaacs, 1980; Montero-Odasso, Schapira, Duque, et al., 2005; VanSwearing, Paschal,

Bonino, & Chen, 1998; Woo, Ho, Lau, Chan, & Yuen, 1995) and prospectively (Hill, et al., 1999; Lord, Lloyd, & Keung Li, 1996; Verghese, et al., 2009) identify fallers. Interestingly, other work has shown that stride-to-stride variations in many gait parameters, or gait variability, may provide greater accuracy in prospectively identifying older adult fallers than changes in the mean values of gait parameters (Besser, Selby-Silverstein, & Prickett, 2001; Hausdorff, 2005; Hausdorff, Rios, & Edelberg, 2001; Lord, et al., 1996; Maki, 1997; Verghese, et al., 2009).

Consequently, it would appear that averaged gait values, and the fluctuations that occur around these values, are clinically useful measures for identifying mobility problems in older adult fallers. Further, given that the majority of falls occur outdoors, particularly in less frail older people (Bath & Morgan, 1999; Li, et al., 2006; Weinberg & Strain, 1995) whilst walking (Li, et al., 2006), the assessment of older adult's walking pattern, and the documentation of age and fall-related changes, is an important research and clinical objective. The following section therefore will review work examining the gait of older adults, the walking changes that occur with normal ageing and the role of these gait alterations in increasing falls risk. Prior to discussing these age and fall-related changes in gait however, a description of normal walking will be provided. The ability of gait variability to differentiate between fallers and non-fallers, and prospectively identify future fallers will be explored in Chapter 2.3.4.

2.2 The Gait Cycle

2.2.1 Preamble

Mobility is an important human function that provides independence and augments health. Mobility limitations such as falls impact upon a person's freedom and autonomy, and upon physical health, psychological health and well-being (Shumway-Cook, Ciol, Yorkston, Hoffman, & Chan, 2005). Much work has been conducted to identify mobility restrictions associated with ageing and pathology, with the goal being to prevent these restrictions and restore normal function. Prior to identifying such limitations however, a thorough understanding of normal gait

is necessary. Therefore, the following section reviews the normal adult gait pattern, the common parameters used to describe gait and the neural control of walking. The consequences of ageing upon these factors will then be presented.

2.2.2 Normal Gait Patterns

Normal human gait is described as a succession of repetitive events termed the gait cycle. One complete gait cycle is the period between two consecutive gait events and contains both a stance phase and a swing phase (Whittle, 2002). The stance phase is the period of time when the foot is in contact with the ground and comprises approximately 60% of the gait cycle, whereas the swing phase is when the foot is non-weight bearing and comprises approximately 40% of the gait cycle. The stance phase is further divided into heel contact, foot flat, heel off and toe off events, and the swing phase is divided into early and late swing events (Ashton-Miller, 2005). These gait cycle events are illustrated for the right limb in Figure 2.3.

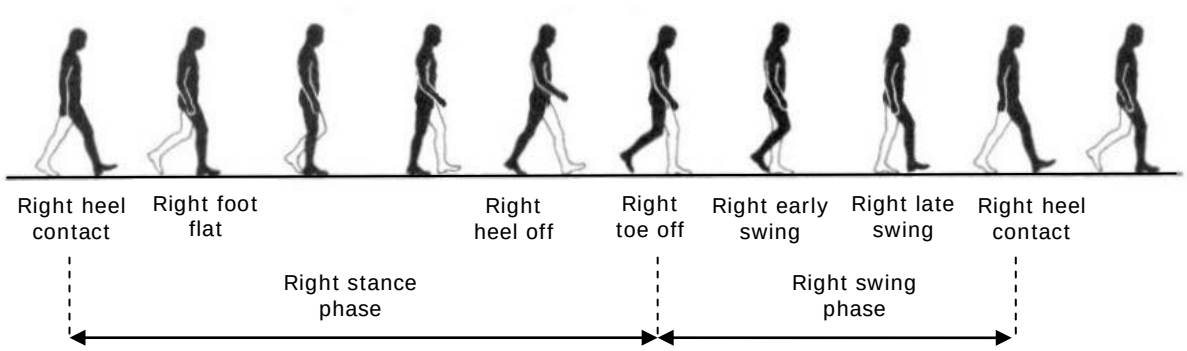


Figure 2.3. Right limb gait cycle and sub phases. Adapted from Inman, Ralston and Todd (1981, p. 26).

Gait may be described quantitatively using kinematic and kinetic variables or electromyography (EMG). Kinematic variables are those that describe motion regardless of the forces producing that motion, and may be linear or angular. Linear kinematic variables include distance and time descriptors of walking, whereas angular kinematic variables include angular motions such as knee and hip joint angles. Gait kinetics and EMG describe the forces and muscle activity that produce walking, such as hip and knee joint powers and hip flexor myoelectric activity. The following sections describe normal lower extremity function using these variables.

2.2.2.1 Linear kinematics

Linear kinematics describe straight-line motion, and include measures of position, displacement, velocity, speed and acceleration (Hamill & Knutzen, 2003). When walking along a straight path therefore, typical linear kinematics include spatial measures such as stride length and temporal measures such as stride time. Such gait variables are commonly termed the spatiotemporal stride parameters.

As outlined above, human walking is cyclical, with repetitive phases where the body is supported initially by one leg followed by the other leg. A complete walking cycle, termed a stride or gait cycle, is the period between gait events of one leg to the same event for the subsequent contact of the same leg (Hamill & Knutzen, 2003). For example, in healthy adults the heel contact event is commonly used to indicate the limits of a gait cycle. Thus a stride may be defined as heel contact of one leg to the next heel contact of the same leg. The horizontal distance between these two events, and the time interval between them, are termed the stride length (line AG, Figure 2.4) and stride time variables, measured in centimetres and seconds, respectively.

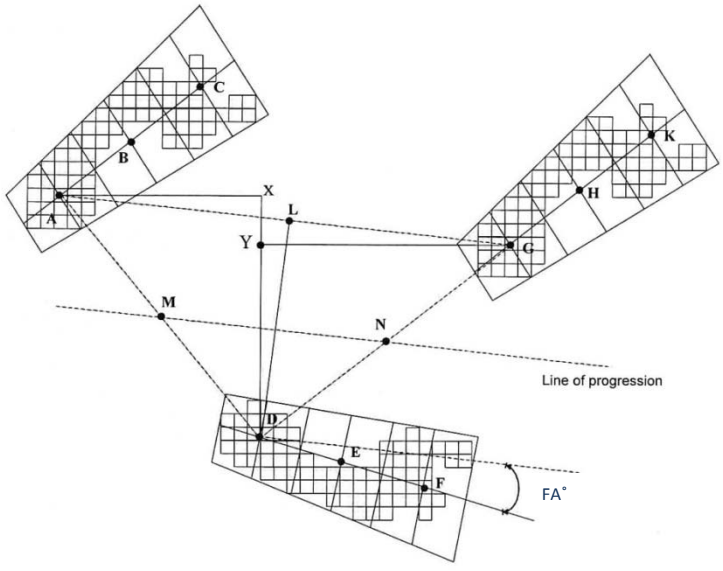


Figure 2.4. Schematic diagram showing the determination of spatial gait measures. Adapted from the GAITRite® Operating Manual ("GAITRite Operating Manual," CIR Systems Inc.). FA, foot angle. See text for explanation of gait measurement determination.

A stride may be further divided into two steps, with a step being the period between gait events of one leg to the same event occurring on the following contact of the opposite leg. Consequently, step length (line AX, Figure 2.4) and time are defined as the horizontal distance and time between the heel contact event of one leg to the heel contact of the opposite foot, measured in centimetres and seconds respectively (Whittle, 2002).

Another commonly studied gait variable is cadence. Cadence is the number of steps within a given time period, typically a second or minute, and the product of this variable and stride length is walking speed (Rose & Gamble, 2006). Walking speed is defined as the average horizontal speed of movement in the direction of walking, and is measured over one or more strides and reported in meters or centimetres per second (Winter, 1991). Walking speed may be calculated using stride length and cadence as follows (Winter, 1991):

$$velocity = \frac{stride\ \ length}{120}\ cm/s, \quad \text{Equation 2.1}$$

Alternatively, gait measurement software applications, such as those from the GAITRite® (CIR Systems, Inc) walkway, often calculate walking speed by dividing a given walking distance by the time taken to walk the distance. The distance commonly used is the horizontal distance between heel contact events of the first and last footfalls on the walkway.

Another useful gait variable is the base of support, sometimes termed the step or stride width. Base of support is defined as the side to side distance between the heel centres of two successive steps and is expressed in centimetres (line LD, Figure 2.4) (Whittle, 2002). Finally, foot angle is also used to describe gait. Foot angle is defined as the angle formed between a line bisecting the foot and the line of progression (angle FA°, Figure 2.4) (Whittle, 2002). The reference line for the foot bisection varies slightly between studies but is commonly reported as a line bisecting the heel and the second metatarsal.

Spatial and temporal stride parameters are relatively simple kinematic measures that enable comparison between and within individuals. For example, a frail older adult's gait pattern may be measured to compare it to that exhibited by a healthy older adult for the purpose of evaluating falls risk. Alternatively, the walking pattern of an older frail adult may be assessed over repeated testing sessions to determine the effectiveness of a falls intervention program. For this reason, the stride parameters of healthy adults are useful measures for both clinicians and researchers to compare deviations and monitor gait changes.

Minimum and maximum averaged values of common spatiotemporal stride parameters recorded from healthy younger adults reported by a selection of studies published over the previous 45 years are presented in Table 2.1. Values for healthy older adults will be presented in Table 2.2 in Chapter 2.2.3.1.

Table 2.1. Minimum and maximum value ranges of common spatial and temporal gait parameters from healthy adults.

	Participants	N	Stride length (cm)	Base of support (cm)	Foot angle (˚)	Stride time (sec)	Speed (cm/s)	Cadence (steps/s)
Murray et al. (1964)	Men aged 20-45 years	36	156-159	7.2-9.6	4.6-6.8	0.98-1.04	NA	1.85-2.03
Blanke & Hageman (1989) [†]	Men aged 20-32 years	12	175 *	10.8	NA	NA	131	NA
Oberg et al. (1993)	Men and women aged 20-49 years	90	114-130*	NA	NA	NA	123-133	1.98-2.16
Stolze et al. (2000)[a]	Women aged 21-37 years	22	148-157	6.6-8.1	6.4-9.5	NA	138-160	2.22-2.33
Whittle (2002)[a]	Women aged 18-49 years	NA	106-158	NA	NA	0.87-1.22	94-166	1.63-2.30
Menz et al. (2004) [†b]	Men and women aged 22-40 years	30	155*	8.3	5.9	NA	143	1.85
Van Uden & Besser (2004) [†b]	Men and women aged 19-59 years	21	156	8.3	5.1	1.10	142	NA
Rose & Gamble (2006)	Men and women aged 20-45 years	NA	156-159	NA	NA	NA	145-159	1.85-2.03

Note: [†] indicates mean is listed as range was not reported; NA, not available; * indicates data was reported as step length and has been doubled to derive stride length; [a] indicates 95% confidence intervals; [b] indicates data from first day of reliability study used.

2.2.2.2 Angular kinematics

Angular kinematics describe joint motions occurring during walking. The following discussion will be limited to hip, knee and ankle joint motion in the sagittal plane (Figure 2.5). Consideration of motion in other planes and occurring in the head, arms and trunk segment is beyond the scope of this review.

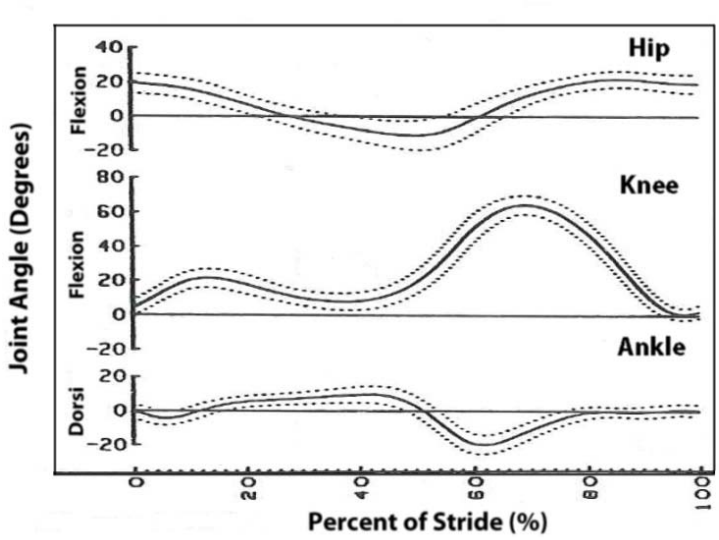


Figure 2.5. Normal hip, knee and ankle join kinematics in the sagittal plane. Adapted from Winter (1991).

Normal hip joint motion (Figure 2.5) is relatively straightforward, with the stance limb extending as the centre of mass passes over the fixed lower extremity and the swing limb flexing at the hip to propel the non weight-bearing limb forward (Riley & Kerrigan, 2005). This motion commences in a flexed position of approximately 20° to 30° of flexion at heel contact and extends to about 20° extension just prior to toe off. During late stance and early swing, the hip joint undergoes flexion to a maximum of 20° to 30° of flexion in preparation for heel contact on the subsequent gait cycle (Rose & Gamble, 2006; Winter, 1991).

Normal knee joint motion (Figure 2.5) is comprised of two flexion waves. The first flexion wave occurs early in stance and acts to absorb the impact of initial contact, whilst the second occurs in early swing to assist with toe clearance (Rose & Gamble, 2006). Specifically, the knee

joint is in about 5° of flexion at heel contact and rapidly flexes to 20° by 15% of the gait cycle. The knee then extends slightly, returning to approximately 5° of flexion, until the heel comes off the ground as the knee rapidly flexes again. This knee joint flexion continues during late stance into early swing, achieving its maximum of approximately 60° of flexion by mid swing. The joint then undergoes extension in preparation for the following heel contact (Rose & Gamble, 2006; Winter, 1991).

Normal ankle angular kinematic plots (Figure 2.5) are more complex than those of the hip and knee joints, and include three foot rockers originally described by Perry (1992), in addition to a fourth movement occurring during swing. The first rocker is the initial plantar flexion motion of approximately 10° following heel contact, and is attributed to the geometry of the calcaneus (Perry, 1992). The second rocker occurs as the tibia passes over the planted foot producing a gradual passive dorsiflexion of approximately 10° by late stance (Rose & Gamble, 2006). As the heel is then pulled off the ground with the advancing centre of mass, the third rocker occurs with first metatarsophalangeal joint dorsiflexion as the ankle joint rapidly plantarflexes to a maximum of 20° just after toe off. During swing, the ankle joint displays its fourth major motion with a dorsiflexion movement just after toe off to assist with toe clearance and is followed by preparation for heel contact (Rose & Gamble, 2006; Winter, 1991).

2.2.2.3 Kinetics

Movement occurs through the application of force. The repetitive segmental kinematic motion that occurs during human walking is the product of internal and external forces acting on the body and occurring at each joint (Hamill & Knutzen, 2003). The study of these forces is known as kinetics. Locomotor kinetics include ground reaction and muscular forces, the moments and powers produced by those forces across specific joints, and the energy produced or absorbed at these joints (Winter, 1991). An examination of these forces during gait is beyond the scope of this review, hence the following discussion will be limited to the internal hip, knee and ankle joint

powers occurring in the sagittal plane.

Muscle mechanical power is the net product of joint angular velocity and the moment of force, and may be either positive or negative (Winter, 1991). Positive mechanical power reflects power generation and is associated with concentric muscular activity, whereas negative mechanical power reflects power absorption and is associated with eccentric muscular activity (Rose & Gamble, 2006; Winter, 1991). Whilst variation exists in the joint power profiles within and between individuals, common important power generation or absorption bursts in the gait cycle have been identified and labeled (Figure 2.6) (Eng & Winter, 1995; Winter, 1991). For example, the second ankle power burst is labelled A2, the first hip power burst is labelled H1 and so on. The following discussion presents the common powers that have been reported on healthy adults.

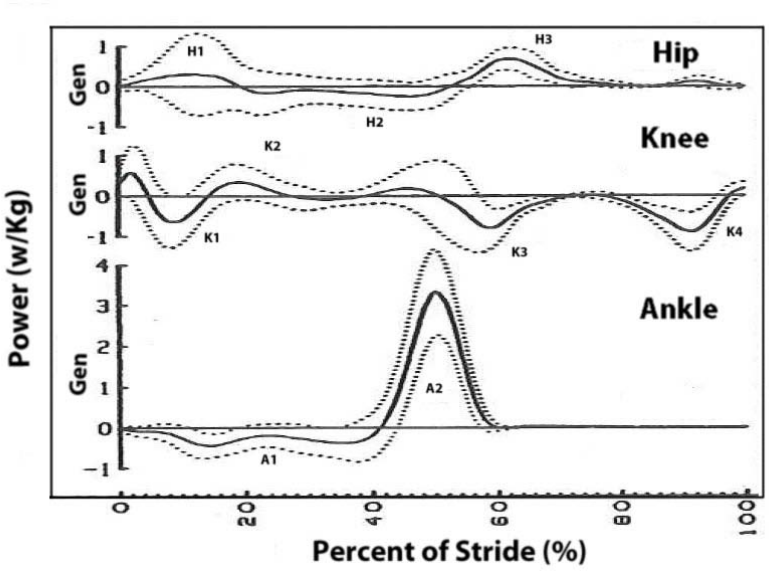


Figure 2.6. Normal hip, knee and ankle kinetics in the sagittal plane. Adapted from Winter (1991).

At the hip joint, there is an early generation of power (H1) during initial loading of the limb resulting in stance phase hip extension. This is followed by the H2 power absorption as the extension is decelerated by an eccentric contraction of the hip flexors. During late stance and into early swing, the swing limb is pulled through by a hip flexor concentric contraction to propel

the swinging limb forward, seen as the H3 power generation burst (Prince, Sadeghi, Zabjek, & Allard, 2000; Sadeghi, Prince, Zabjek, & Allard, 2001; Sadeghi, Sadeghi, et al., 2001; Winter, 1991).

At the knee joint, initial knee flexion upon loading is controlled eccentrically by the knee extensors, evidenced by the K1 power absorption. This is quickly followed by a power generation (K2) as the knee extends by a concentric quadriceps contraction, and then a second power absorption burst (K3) as the late stance knee flexion is again controlled eccentrically by the knee extensors. Finally, in late swing as knee extension slows in preparation for the subsequent heel contact, the knee extensors again eccentrically control extension by a final power absorption burst (K4) (Hamill & Knutzen, 2003; Winter, 1991).

There are two power bursts at the ankle joint. The first of these is the A1 power absorption burst throughout the foot flat phase. This occurs with an eccentric contraction of the plantarflexors to control the passive dorsiflexion of the tibia passing over the planted foot. The second is a large power generation burst (A2) by the plantarflexors in late stance as the heel is pulled off the ground, accelerating the limb forward into swing (Rose & Gamble, 2006; Vardaxis, Allard, Lachance, & Duhaime, 1998; Winter, 1991).

2.2.2.4 Electromyography

The activation of skeletal muscles during human walking provides the forces necessary to propel the body forward. The activation of these muscles is the net product of the recruitment of each motor unit comprising that muscle, which generates an electrical signal termed the motor unit action potential (Winter, 2005). Electromyography (EMG) is the process of recording the sum of these motor potentials from a given muscle, and provides valuable information regarding the activation of that muscle during walking.

Figure 2.7 illustrates the activity of the major muscle groups of the lower limb during the stance and swing phases of walking. During the heel contact to foot flat phase, tibialis anterior is active to eccentrically control plantarflexion whilst gluteus maximus and the medial and lateral hamstrings extend the hip and stabilize the pelvis. Also during this phase, rectus femoris, vastus medialis and vastus lateralis eccentrically control knee flexion as the stance limb is loaded (Kwon, Minor, Maluf, & Mueller, 2003; Uustal & Baerga, 2004; Vaughan, Davis, & O'Connor, 1999).

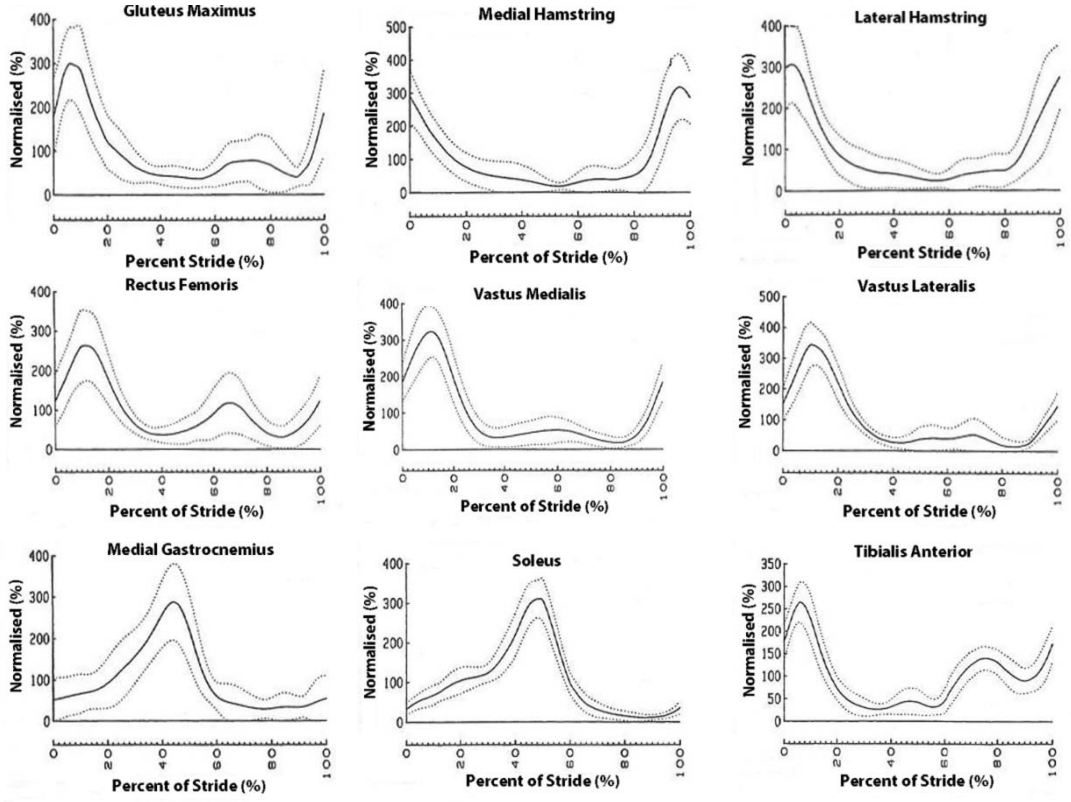


Figure 2.7. Normal EMG activity during walking. Adapted from Winter (1991).

Activity from the quadriceps muscle group continues into midstance to extend the knee as the gastrocnemius and soleus muscles control the passive dorsiflexion of the tibia over the planted foot. Concentric activity from the plantarflexors then pulls the heel off the ground following which rectus femoris commences hip flexion to pull the limb into swing. This activity continues into early swing, pulling the limb forward and concentric contraction of tibialis anterior dorsiflexes the foot for toe clearance. In late swing, gluteus maximus and the medial and lateral hamstrings eccentrically decelerate the swinging limb in preparation for subsequent heel contact

(Perry, 1992; Rose & Gamble, 2006; Vaughan, et al., 1999; Winter, 1991).

2.2.2.5 Neural control of gait

The generation and control of human walking is achieved through the normal functioning and interaction of three components of the nervous system (Barrière, Leblond, Provencher, & Rossignol, 2008). Firstly, a spinal central pattern generator (CPG) produces the coordinated rhythmic muscle activity required for walking. Secondly, the CPG is under the instruction of higher centres that initiate and terminate locomotion, and control variables such as the rate and direction of walking. Finally, the generator and higher centres receive phasic sensory feedback from peripheral receptors regarding factors such as the timing, orientation and loading that occur during walking. Whilst the role of neuromodulators in modifying spinal locomotor outputs is receiving increasing attention of late [for a brief discussion, see (MacKay-Lyons, 2002)], the following section will be limited to a discussion of each of the former three components on human walking.

2.2.2.5.1 Central pattern generators

Many studies have investigated the role of CPGs in locomotion. Early work by Sherrington (1910) proposed that flexion and crossed extensor reflexes provide the basis for generating the alternating patterns of flexion and extension of the lower limbs in human walking. However, in 1914, Graham Brown suggested that locomotor activity is independent of reflexes and is instead generated by a network of specialized neurons called central pattern generators (CPGs). Theoretically, CPGs were thought to have two half centres within the spinal cord, connected by reciprocal inhibition and controlled by a common excitatory input (Graham Brown, 1914). It was proposed that activation of one half-centre generated flexor and extensor muscle activity of the ipsilateral limb in opposite phase to the contralateral limb (Jankowska, Jukes, Lund, & Lundberg, 1967b). Fatigue then released the reciprocal inhibition allowing activation of the alternate half centre, resulting in alternate stepping-like activity in the lower limbs (Figure 2.8) (Graham Brown,

1914; Hultborn & Nielsen, 2007) .

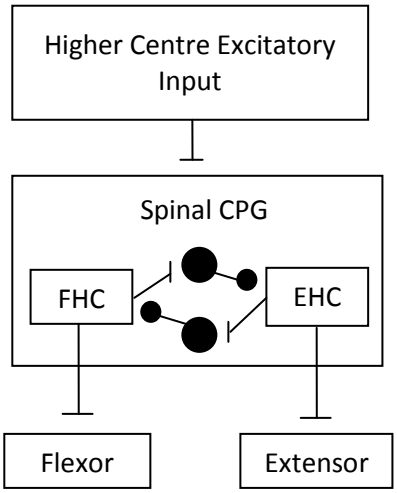


Figure 2.8. Central pattern generator (CPG) with mutually inhibiting (filled circles) flexor and extensor half centres (FHC and EHC respectively). Adapted from Van de Crommert et al. (1998).

Graham Brown's hypothesis was supported by earlier work showing locomotor behaviour in deafferented cats (Graham Brown, 1911, 1912), and was subsequently confirmed when the network was found in the transected spinal cords of cats in 1967 (Jankowska, et al., 1967b; Jankowska, Jukes, Lund, & Lundberg, 1967a). Evidence for the existence of CPGs in a variety of vertebrates other than humans is now unequivocal, and its role in producing coordinated rhythmical stepping is widely accepted (MacKay-Lyons, 2002). Whilst the network is yet to be directly observed in humans, a number of reviews have presented support of its existence (Dietz, 2003; Duysens & Van de Crommert, 1998; Hultborn & Nielsen, 2007; MacKay-Lyons, 2002).

2.2.2.5.2 Higher centres

To evoke locomotor activity from CPG neural networks in deafferented spinal animals, it is necessary to apply external pharmacological stimuli such as L-DOPA and Nialamide (Jankowska, et al., 1967a; Lundberg, 1979). In the intact animal however, this behaviour is thought to be generated from higher neural centres within the brainstem and midbrain (Grillner, Wallén, Saitoh, Kozlov, & Robertson, 2008; MacKay-Lyons, 2002). Work in decerebrate cats has shown that stimulation of a brainstem centre termed the mescencephalic locomotor region (MLR)

produces locomotion in the absence of afferent limb feedback (Jordan, Pratt, & Menzies, 1979; Shik & Orlovsky, 1976; Whelan, 1996). Given the MLR projects bilaterally onto spinal CPGs via reticulospinal neurons (Brocard & Dubuc, 2003; Garcia-Rill & Skinner, 1987; Jahn, et al., 2008), it is believed that the MLR initiates and regulates locomotion through direct activation of the pattern generators (Armstrong, 1988; Grillner, Hellgren, Ménard, Saitoh, & Wikström, 2005; Noga, Kriellaars, Brownstone, & Jordan, 2003; Shefchyk & Jordan, 1985).

Although the majority of work on the MLR has involved non-human vertebrates, clinical case reports on humans with mesencephalic lesions (Hanna & Frank, 1995; Hathout & Bhidayasiri, 2005; Masdeu, Alampur, Cavaliere, & Tavoulareas, 1994) and recent imaging studies on humans imagining walking (Jahn, et al., 2008) suggest the MLR is also likely to play a role in the initiation of CPGs in human walking. The role of supraspinal structures other than the brainstem in locomotion however remains less clear. These roles can be arbitrarily divided into locomotor program selection, attention and adaptation demands, and steering and postural requirements (Grillner, et al., 2008; Orlovsky, 1991). Research has identified that the basal ganglia, the sensorimotor cortex, visual and vestibular inputs and the cerebellum contribute to these functions during locomotion.

The important role of the basal ganglia in walking is demonstrated by the gait changes observed in patients with basal ganglia disease (Baltadjieva, Giladi, Gruendlinger, Peretz, & Hausdorff, 2006; Peppe, Chiavalon, Pasqualetti, Crovato, & Caltagirone, 2007). Work has shown that the inhibitory system of the basal ganglia aids in the modulation of rhythmic limb movements and the maintenance of muscle tone and posture during walking (Garcia-Rill, 1986; Takakusaki, Kohyama, Matsuyama, & Mori, 2001; Takakusaki, Oohinata-Sugimoto, Saitoh, & Habaguchi, 2004). Consequently, basal ganglia pathology such as Parkinson's disease produces changes in locomotor rhythm (Baltadjieva, et al., 2006; Frenkel-Toledo, et al., 2005), tone (Prochazka, et al., 1997; Rao, Hofmann, & Shakil, 2006) and posture (Jacobs, Dimitrova, Nutt, &

Horak, 2005; Nallegowda, et al., 2004). Other authors have also demonstrated that stimulation of the substantia nigra within the basal ganglia increases stance phase duration, and alters rhythmic limb movements and 'MLR activated step cycles' (Takakusaki, Habaguchi, Ohtinata-Sugimoto, Saitoh, & Sakamoto, 2003), suggesting a role in modulating the cyclical rhythmic properties of walking. Again, these gait changes are also consistent with other Parkinson's-induced gait disturbances, including alterations in the temporal fluctuations during walking (Morris, Iansek, Matyas, & Summers, 1994). The potentially important role of the basal ganglia in modulating rhythmic components of locomotion will be explored further in Chapter 2.3.5.

The basal ganglia also has an important role in the selection and initiation of the locomotor program (see Figure 2.9) (Grillner, et al., 2005; Grillner, et al., 2008). Previous work has demonstrated that basal ganglia output nuclei such as the substantia nigra and the globus pallidus project directly to the MLR (Garcia-Rill, 1986; Takakusaki, et al., 2003; Takakusaki, Oohinata-Sugimoto, et al., 2004; Takakusaki, Saitoh, Harada, & Kashiwayanagi, 2004). These projections are inhibitory, thus at rest the target locomotor centre is under inhibitory control, and locomotion is only initiated when the inhibition is removed (Grillner, et al., 2005; Hikosaka, Takikawa, & Kawagoe, 2000; Ménard, Auclair, Bourcier-Lucas, Grillner, & Dubuc, 2007). The disinhibition occurs through activation of the input region of the basal ganglia, termed the striatum (Brudzynski, Wu, & Mogenson, 1993; Grillner, et al., 2005). In turn, the striatum receives input from the cerebral cortex (Grillner, et al., 2005) (Figure 2.9). Consequently, Grillner and colleagues (2005) proposed that the various basal ganglia nuclei function as a filter for motor behaviour, selecting and releasing activities such as locomotion. The occurrence of Parkinson's induced hypokinesias such as the freezing phenomenon, and L-DOPA-induced hyperkinesias would appear to support this proposition (Takakusaki, Oohinata-Sugimoto, et al., 2004).

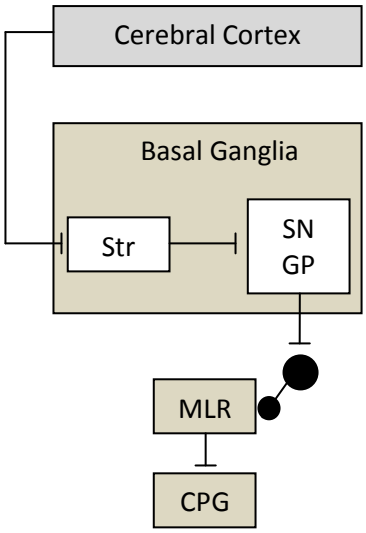


Figure 2.9. Pathway depicting basal ganglia control of locomotion. Filled circles indicates inhibitory control. Str, striatum; SN, substantia nigra; GP, globus pallidus; MLR, mescencephalic locomotor region; CPG, central pattern generator.

The basal ganglia is also part of a larger neural circuit that receives input from the cerebral cortex and thalamus (Grillner, et al., 2005). Researchers have demonstrated that this input from the cerebral cortex is crucial in the planning, initiation and termination of walking (Nutt, Marsden, & Thompson, 1993; Wichmann & DeLong, 1996). For instance, Yazawa et al. (1997) showed that bilateral activation in the supplementary motor area (SMA) of the frontal lobe was involved in gait initiation, helping to explain why underactivity in this area is seen in the cortices of Parkinson's patients (Sato, Hashimoto, Nakamura, & Ikeda, 2001; Shibasaki, Fukuyama, & Hanakawa, 2004). Miyai et al. (2001) also showed activity in the SMA and the medial portion of the primary sensorimotor cortex of healthy adults during treadmill walking. These activation patterns were not present during arm swing or foot movement tasks suggesting locomotor specific roles for these regions. Using single photon emission computed tomography (SPECT) whilst treadmill walking, Hanakawa and colleagues (1999) also showed cortical activity in foot, leg and trunk regions of the primary sensorimotor cortex and the premotor region, which has been attributed to higher control of human walking (Shibasaki, et al., 2004).

Other work has shown that although there is minor activation of these cortical areas during

unconstrained walking, the supplementary motor area and sensorimotor cortex are increasingly involved as the locomotor task difficulty increases (Malouin, Richards, Jackson, Dumas, & Doyon, 2003). For instance, functional magnetic resonance imaging (fMRI) studies have shown increased activity in the neurons in the motor cortex when stepping over obstacles (Drew, Prentice, & Schepens, 2004; Matsuyama, et al., 2004) and when adjusting posture (Beloozerova, et al., 2003). Other authors have suggested that the cerebral cortex becomes more involved in locomotor control with increasing effort and attentional requirements (Shibasaki, et al., 2004). Consequently, it would appear from these studies that following gait initiation, the basal ganglia and brainstem may be able to control unconstrained steady state locomotion with minimal cortical activity, although some background activation is still required to adapt to any unexpected environmental demands (Jahn, et al., 2004; Shibasaki, et al., 2004). However, for increasingly complex walking, where exact foot placement for obstacle avoidance or external perturbations is required, greater cortical control appears necessary (Drew, Jiang, Kably, & Lavoie, 1996; Jahn, et al., 2004; Orlovsky, 1991).

Visual and vestibular inputs also play an important role during walking. Although coordination between the visual and locomotor systems is required for precise foot placement during obstacle negotiation (Takakusaki, Oohinata-Sugimoto, et al., 2004), Fukuyama et al. (1997) has shown that the visual cortex is also active during walking across flat or even terrain. Other components of the visual system that are active during normal walking include the fusiform and parahippocampal gyri for visuospatial navigation and imagination of the visual environment (Ekstrom, et al., 2003; Jahn, et al., 2008; Jahn, et al., 2004), and projections from the superior colliculus to reticulospinal neurons in the MLR for locomotor steering (Fagerstedt, Orlovsky, Deliagina, Grillner, & Ullén, 2001; Saitoh, Ménard, & Grillner, 2007). Vestibular inputs have also been shown to project to reticulospinal neurons in the brainstem, providing information about orientation and movements of the head during locomotion, and integrating propulsion and body orientation (Grillner, et al., 2008; Zelenin, Orlovsky, & Deliagina, 2007).

The final supraspinal structure known to have an important role in locomotion is the cerebellum. Positron emission tomography (PET) and SPECT studies have shown that the cerebellum is widely active during walking and running (Fukuyama, et al., 1997; Mishina, et al., 1999; Shibasaki, et al., 2004; Tashiro, et al., 2001). Various locomotor functions have been proposed for cerebellar activity, including the maintenance of balance and postural control, fine control and coordination of multijoint dynamics, and speed modulation of spinal CPGs.

The contribution of the cerebellum to postural control and multijoint coordination and dynamics is evident from locomotor disorders such as gait ataxia observed in patients with cerebellar pathology (Ebersbach, et al., 1999). Studies have shown that the medial aspect of the cerebellum is important for the former postural control function (Thach & Bastian, 2004), whereas the intermediate (cerebellar vermis) and lateral aspects appear to control the coordination (Marple-Horvat & Criado, 1999; Pardoe, Edgley, Drew, & Apps, 2004). However, despite deficits in balance and coordination in patients with cerebellar pathology, these two functions may not necessarily be interdependent. For instance, recent work by Ilg and colleagues (2007) has shown that locomotor coordination deficits in patients with cerebellar disorders are strongly correlated with impaired multijoint temporal variability rather than balance impairments. This finding suggests the cerebellum plays an important temporal role in locomotor coordination independent of postural control.

Other work supports the idea of the cerebellum contributing to locomotor timing mechanisms. Mori et al. (2001) reported that increasing the strength of stimuli to the cerebellum increased the speed of locomotion in cats. Using fMRI on healthy adults imagining walking and running, Jahn and colleagues (2004) found increased cerebellar activation with increased locomotor speed. This led the authors to propose that the cerebellar vermis might function as a speed generator, further highlighting the role of the cerebellum in temporal locomotor coordination. Consequently, whilst it is evident that the cerebellum contributes to postural

control during walking, there is increasing evidence that cerebellar regions may also play a fundamental role in the temporal regulation of the stepping pattern. The neural control of stepping rhythm will be explored further in Chapter 2.3.5. A schematic diagram illustrating the various higher inputs, including those to and from the cerebellum, to the spinal CPG is displayed below in Figure 2.10.

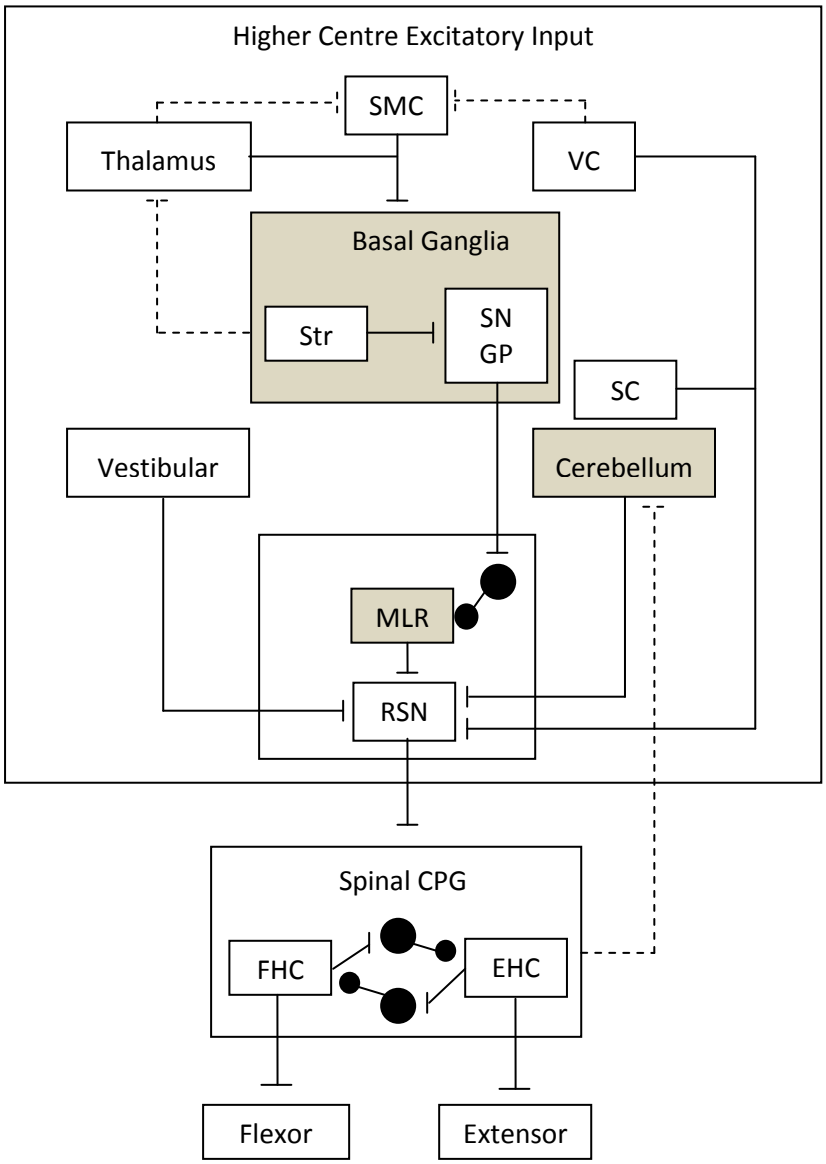


Figure 2.10. Diagram of Higher centres directing locomotion and providing input to the spinal CPG. Solid lines indicate feed forward and dashed lines indicate feedback mechanisms. Shaded boxes indicate purported regions with rhythmic locomotor roles. Filled circles indicate inhibitory control. SMC, supplementary motor cortex; VC, visual cortex; Str, striatum; SN, substantia nigra; GP, globus pallidus; SC, superior colliculus; MLR, mescencephalic locomotor region; RSN, reticulospinal neurons; CPG, central pattern generator; FHC, flexor half centre; EHC, extensor half centre. Spinal CPG component adapted from Van de Crommert et al. (1998).

2.2.2.5.3 Afferent feedback

The studies reviewed in the previous section show that CPG activity may be initiated pharmacologically or by stimulation of the brainstem. However, early experiments using spinal cats reported that electrical stimulation of the dorsal roots of the spinal cord (simulating afferent sensory feedback) initiated locomotor behaviour (Grillner & Zangger, 1979). While the ability of sensory feedback to evoke spontaneous locomotion in animal preparations such as the spinal cat is of some interest, of much greater relevance to human locomotion is the role of afferent feedback in the entrainment of the locomotor rhythm through stance and swing phase transitions (Hultborn & Nielsen, 2007; Pearson, 2004).

Van de Crommert and colleagues (1998) have identified three sensory sources for locomotor entrainment: information on loading from force-sensitive golgi tendon organs (GTO) in extensor muscles, loading feedback from mechanoreceptors in the plantar foot, and positional information from stretch-sensitive muscle spindles in hip musculature. The first two function to increase stance muscle bursts and duration during the stance phase of gait, whereas the third facilitates the onset of the swing phase (Duysens, Clarac, & Cruse, 2000). Other roles of sensory feedback during locomotion have also been reported, including adaptive mechanisms and phase and task-dependent modulation of muscle activity [see (Pearson, 2004; Zehr & Stein, 1999) for reviews], however the following section will be restricted to the ability of afferent feedback to modulate the timing of the phase transitions during gait.

Initial studies proposed that the role of GTOs was to prevent excessive loading in skeletal muscles by autogenic inhibition (Hultborn & Nielsen, 2007). Whilst this may be the case under static conditions, other experiments have demonstrated that during locomotion, the receptors enhance extensor activity when the extensors are loaded during stance (Pearson, 1995). For example, selective stimulation of Ib afferents from GTOs (simulating loading) from the plantaris muscle in spinal cats increased extensor muscle bursts and delayed the onset of flexor activity

(Conway, Hultborn, & Kiehn, 1987). This supports earlier work by Duysens and Pearson (1980) who reported increased ankle extensor EMG amplitude and duration coupled with reduced flexor bursts following an increased stretch of the cat Achilles tendon. The extensor activity was attributed to the increase in muscle loading from the stretch (due to force-length muscle characteristics) stimulating Ib afferents, although stimulation of Ia afferents from muscle spindles cannot be ruled out (Van de Crommert, et al., 1998). Indeed, Guertin et al. (1995) reported similar findings and suggested that convergence of Ia and Ib afferents contributed to the modulation of extensor burst activity. More recent work on humans however has shown that a sudden unloading of the ankle plantarflexors during stance reduces soleal EMG activity, even following blocking of Ia afferents, suggesting a likely involvement of the Ib pathway (Sinkjaer, Andersen, Ladouceur, Christensen, & Nielsen, 2000). Therefore, it is probable that the positive feedback from GTOs functions to reinforce extensor activity in response to limb loading during the stance phase of gait, and prevent the onset of swing until the limb is sufficiently unloaded (Duysens, et al., 2000; Pearson, 2004).

Loading feedback from cutaneous mechanoceptors in the foot has also been shown to prolong extensor bursts and delay flexor activity during the stance phase of locomotion (Duysens, 1977a, 1977b; Duysens & Pearson, 1976). Conversely, withdrawal of cutaneous load feedback from the foot has been shown to initiate flexion (Jankowska, et al., 1967a). Conway et al. (1994) has also demonstrated that stimulation of foot cutaneous nerves during late flexor activity terminated the flexor burst in spinal cats. Van de Crommert and colleagues (1998) argue that these cutaneous afferents are able to enhance extensor activity when loaded, such as during stance, and initiate flexor activity when unloaded, such as in the phase transition to swing. It was suggested that this could be achieved via an excitatory input to CPG extensor motoneurons and an inhibitory input to CPG flexor motoneurons, with the purpose of regulating extensor activity during stance in concert with Ib GTO feedback. The finding of extensive convergence of Ib and cutaneous afferents would appear to support this proposal (Lundberg, Malmgren, & Schomburg, 1977).

The third sensory feedback source regulating locomotor activity originates from muscle spindle afferents from the hip. Early work by Grillner and Rossignol (1978) showed that hip flexion prevented transition from stance to swing, whereas hip extension initiated swing, suggesting that hip afferents may have an important role in swing phase initiation. Further work by Anderson and Grillner (1983) demonstrated entrainment of the locomotor rhythm was able to be achieved by flexion and extension movements of the hip. Investigating whether this entrainment arises from hip articular or Ia muscle spindle afferents, Kriellaars et al. (1994) reported that the activity persisted following both joint denervation and anaesthesia in decerebrate cats, implicating Ia afferent feedback. Other research has shown that hip denervation in humans did not significantly alter either the locomotor pattern or position sense, further supporting the importance of Ia muscle spindle afferents (Grillner & Wallén, 1985). Consequently, it appears that proprioceptive feedback from hip stretch receptors play an important role generating swing phase flexion at the end of stance during gait (Hultborn & Nielsen, 2007), possibly via an excitatory input from Ia afferents to CPG flexor motoneurons (Van de Crommert, et al., 1998). This input, along with feedback from GTO Ib and cutaneous afferents, and central feed forward and feedback mechanisms into the spinal CPG are illustrated below in Figure 2.11.

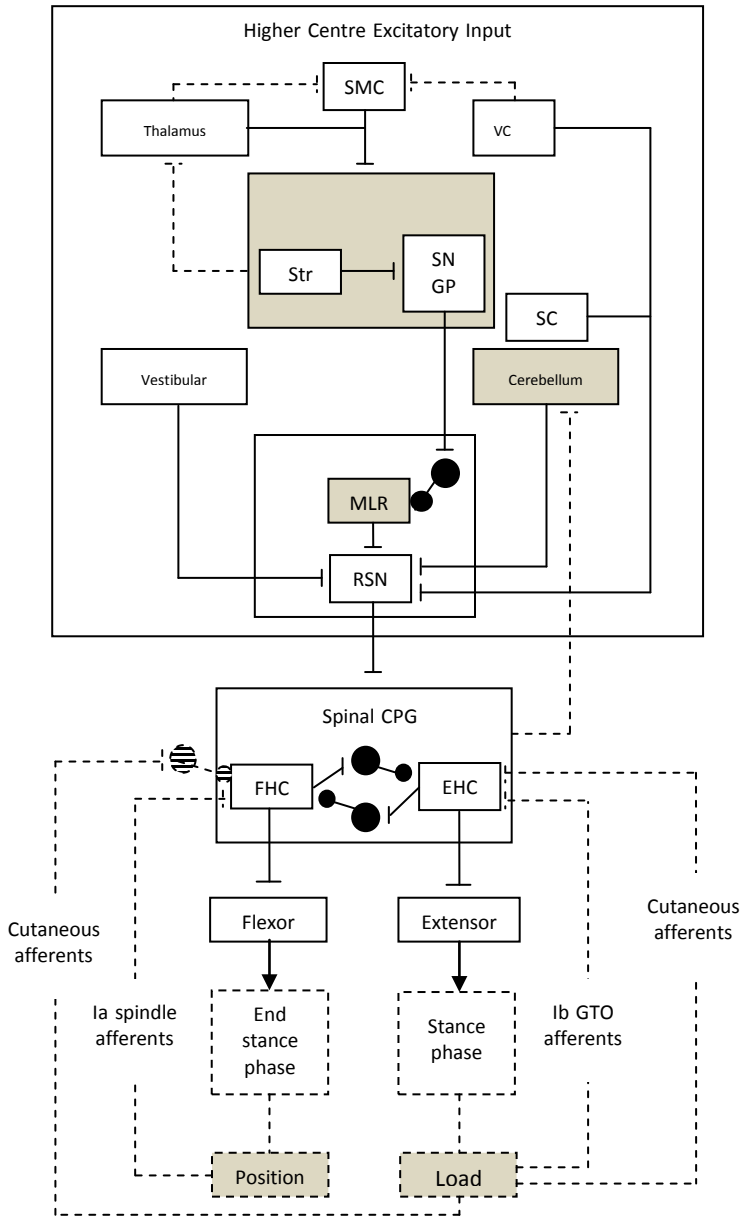


Figure 2.11. Diagram of Higher centres directing locomotion and providing input to the spinal CPG and feedback from peripheral receptors. Solid lines indicate feed forward and dashed lines indicate feedback mechanisms. Shaded boxes indicate purported regions with rhythmic locomotor roles. SMC, supplementary motor cortex; VC, visual cortex; Str, striatum; SN, substantia nigra; GP, globus pallidus; SC, superior colliculus; MLR, mescencephalic locomotor region; RSN, reticulospinal neurons; CPG, central pattern generator; FHC, flexor half centre; EHC, extensor half centre. Spinal CPG component adapted from Van de Crommert et al. (1998).

2.2.3 Older Gait Patterns

In Australia, 43% of community dwelling older adults over 65 years have difficulty in walking one kilometre (Barr, Browning, Lord, Menz, & Kendig, 2005) and approximately one third require assistance with walking (Lord & Menz, 2002). The incidence of walking difficulties rises rapidly with increasing age, with 55% of 80 to 84 year olds having some level of difficulty when walking for less than 100 metres (Mottram, Peat, Thomas, Wilkie, & Croft, 2008). These age-related mobility problems commonly lead to an associated reduction in physical activity and an increase in mobility-related accidents such as a fall (see Chapter 2.1).

Mobility difficulties have been associated with alterations in the gait pattern of older adults (Brach, Studenski, Perera, VanSwearingen, & Newman, 2007; Chu, Chiu, & Chi, 2006; Lichtenstein, Burger, Shields, & Shiavi, 1990; Talkowski, Brach, Studenski, & Newman, 2008). It follows therefore that the prospective identification of gait patterns that lead to mobility difficulties may help prevent these age-related declines in the health of older adults. The following sections will review the kinematic, kinetic, EMG and neural changes that occur with age in an attempt to provide insight into the abnormal gait patterns that lead to mobility restrictions.

2.2.3.1 Linear Kinematics

It is well known that gait changes with ageing. Table 2.2 below presents findings from a number of studies published over the previous 45 years that have compared averaged values of spatial and temporal gait variables in younger and older adults. Many of these studies have shown significant changes in locomotor linear kinematics with increasing age [e.g. (Blanke & Hageman, 1989; Hageman & Blanke, 1986; Menz, et al., 2004; Stolze, et al., 2000; Winter, 1991; Winter, Patla, Frank, & Walt, 1990)].

The most common spatial or temporal alteration in the walking pattern of older adults is reduced walking velocity (Prince, Corriveau, Hébert, & Winter, 1997). For example, Finley et al. (1969) found a 12 cm/s reduction in walking velocity between a group of healthy younger (mean

age 29.9 years) and older women (mean age 74.4 years). Other studies have reported reductions ranging between 3 cm/s and 27.59 cm/s for older compared to younger adults (Hageman & Blanke, 1986; Whittle, 2002). Similarly, Woo and colleagues (1995) found a yearly reduction of 0.1 to 0.7% in walking velocity for adults aged over 70, consistent with other studies (Craik & Dutterer, 1995; Oberg, et al., 1993). Age was also reported as the best predictor of walking speed for their male participants, whereas in women, in addition to age, height and physical activity levels were also associated with velocity (Woo, et al., 1995). Age-related reductions in walking velocity are consistently greater in females than males (see Table 2.2) and these reductions alone have been shown to predict future hospitalisation, falls and needing a caregiver in what were previously community dwelling older adults aged 75 years and over (Montero-Odasso, Schapira, Soriano, et al., 2005).

Table 2.2. Walking velocity, stride length and cadence in younger and older male, female and mixed samples.

	Young				Older			
	Sample	Velocity (cm/s)	Stride length (cm)	Cadence (steps/min)	Sample	Velocity (cm/s)	Stride length (cm)	Cadence (steps/min)
Men								
Murray et al. (1964)	20-25 years (N = 12)	152.2	158.8	115	60-65 years (N = 12)	146.6	153	115
Blanke & Hageman (1989)	20-32 years (N = 12)	131.3	192.6	NA	60-74 years (N = 12)	138.9	189.6	NA
Oberg et al. (1993)[a]	20-29 years (N = 15)	122.7	122.4	118.8	60-69 years (N = 15)	127.7	130	117
Watelin et al. (2000)	Mean 28.04 years (N = 16)	134.8	146	110.7	Mean 61.7 years (N = 16)	122.2*	133*	112.9*
Whittle (2002)[b]	18-49 years (N = NA)	110-182	125-185	91-135	65-80 years (N = NA)	96-168	111-171	81-125
Women								
Finley et al. (1969)	Mean 29.9 years (N = 23)	84	NA	105	Mean 74.4 (N = 23)	70	NA	109
Hageman & Blanke (1986)	20-35 years (N = 13)	159.5	162.7	NA	Aged over 60 years (N = 13)	131.9**	134.9	NA
Oberg et al. (1993)[a]	20-29 years (N = 15)	124.2	118.2	124.8	60-69 years (N = 15)	115.7	110.6	123.6
Lord et al. (1996)	NA	NA	NA	NA	Mean 71.2 years (N = 80)	114	115	118.9

Table 2.2. *Continued*.

	Young				Older			
	Sample	Velocity (cm/s)	Stride length (cm)	Cadence (steps/min)	Sample	Velocity (cm/s)	Stride length (cm)	Cadence (steps/min)
Women (cont'd)								
Stolze et al. (2000)[a]	21-37 years (N = 22)	149	152	136	64-92 years (N = 22)	117***	118**	129
Whittle (2002)[b]	18-49 years (N = NA)	94-166	104-156	97-137	65-80 (N = NA)	91-163	94-146	96-136
Mixed								
Winter et al. (1990)	21-28 years (N = 12)	143.4	155	111.0	62-78 (N = 15)	128.0	139**	110.5
Elble et al. (1991)	Mean 30 years (N = 20)	118	NA	NA	Mean 74.7 years (N = 20)	94	NA	NA
Menz et al. (2004)[a, c]	22-40 years (N = 30)	143	154.9	110.8	76-87 years (N = 31)	116*	127.8	107.9

Note: Where possible ranges have been used. NA, not available; [a] indicates stride length derived from doubling the reported step length; [b] Indicates 95% confidence intervals; [c] indicates data from first day of reliability study used; *p<0.05; ** p<0.01; *** p<0.001.

As noted by Findlay (1969), the reduced walking velocity with age can be mainly explained by a reduction in stride length. Equation 2.1 demonstrates that walking velocity is dependent upon both stride length and cadence. As shown in Table 2.2 however, the majority of studies reporting age-related reductions in walking velocity also report a reduction in stride length, whereas changes in cadence are generally minor. For males, the reductions in stride length range from 3 cm to 14 cm (Blanke & Hageman, 1989; Watelain, Barbier, Allard, Thevenon, & Angue, 2000), whereas for females they range from 7.6 cm to 34 cm (Oberg, et al., 1993; Stolze, et al., 2000). Winter (1991) states that the dependence of walking velocity upon stride length in older adults is strong when the sample under investigation is healthy and screened for gait and other disorders, as is the case with the majority of studies presented in Table 2.2.

Whilst the majority of studies report decreases in velocity and stride length with age, some studies have reported small increases in these gait parameters. For example, Blanke and Hageman (1989) reported a 7.61 cm/s increase in velocity, however this difference was not statistically significant. Similarly, Oberg et al. (1993) reported a 5 cm/s increase in velocity and a 7.6 cm increase in stride length in adults aged 60 to 69 compared to young adults aged 20 to 20. Whilst this difference is not easily explained, when all adults aged over 60 (i.e. between 60 and 79) are included in the comparison, the age related differences are only minor (0.25 cm/s and 4.1 cm increases for walking speed and stride length respectively).

Age-related changes in spatial and temporal parameters other than velocity and stride length are inconsistent. For example, some studies have reported a slight increase in base of support with age (Hageman & Blanke, 1986; Menz, et al., 2004), whereas others report a slight decrease (Blanke & Hageman, 1989; Stolze, et al., 2000). Similarly, although foot angle is typically reported as being larger with age, these changes are relatively minor and unlikely to be meaningful (Menz, et al., 2004; Murray, et al., 1964; Stolze, et al., 2000). Consequently, it would appear that the most consistent age-related changes in spatiotemporal gait parameters are reductions in velocity

and stride length. The kinematic and kinetic mechanisms responsible for these alterations will be discussed in the following sections.

2.2.3.2 Angular kinematics

Lower extremity joint angular kinematic profiles have generally been found to be similar between older and younger adults (Finley, et al., 1969; Murray, et al., 1964; Winter, 1991). However, important differences do exist in the motion of the ankle, knee and hip joints during normal walking, which may help explain the age-related alterations in velocity and stride length discussed previously. For example, Oberg and colleagues (1994) found that the total ankle joint range of motion for a group of healthy older adults was 24.9° whereas younger adult ankle motion was 29.3°. Similarly, Judge et al. (1996) reported lower plantar flexion angles in their older (13°) compared to their younger sample (17°). With perhaps greater relevance for stride length, Winter (1991) found that ankle angle was reduced during the push off phase, with older adults plantarflexing to 24.9° compared to 29.3° for the younger adults. It has been suggested that these age-related reductions in ankle range of motion might provide a possible explanation for the shorter stride and less vigorous push off of older adults (DeVita & Hortobágyi, 2000; Judge, et al., 1996).

Other work has reported minor reductions in knee joint range of motion with age, which may also contribute to the smaller stride of older adults (DeVita & Hortobágyi, 2000; Judge, et al., 1996). For instance, compared to younger adults, knee extension increased during stance and decreased during swing in older adults (Judge, et al., 1996; Oberg, et al., 1994). Similarly, Winter (1991) found reduced knee extension with age during late swing, with older adults exhibiting 5.3° of knee flexion compared to 0.5° for the younger adults at terminal swing. This continued flexion in late swing was correlated with shorted steps in the older adults.

Finally, many authors have reported reduced hip extension in older adults, potentially reducing stride length and walking velocity. Both Kerrigan et al. (1998) and Riley et al. (2001) found that older adults walking at self-selected speed exhibited less hip extension than younger adults. This extension did not increase with increased walking speed. Significantly, Riley et al. (2001) argued that this action reduces walking speed in older adults. However, other authors have reported either an increase (DeVita & Hortobágyi, 2000; Winter, et al., 1990) or no change (Oberg, et al., 1994) in hip joint range of motion with age. Whilst the cause of the different findings in hip kinematics is not immediately clear, a possible explanation lies in the different self-selected speeds of the two age groups in these studies. Similarly, whilst reduced kinematic profiles of the ankle, knee and hip joints with age might provide a partial explanation for the reduced stride length and speed of older adults, it is likely self selected walking speed is a confounding factor in these findings. Age-related alterations in lower extremity joint kinetics may provide further information regarding the mobility limitations. The influence of self-selected walking speed on joint kinetics will therefore be discussed further in the next section.

2.2.3.3 Kinetics

As explained previously, older adults have reduced joint excursions resulting in shorter strides and slower walking speeds. To determine whether these alterations are a conscious choice to improve stability or if they reflect an age-related change in the neuromuscular control of the lower limbs during walking, Winter and colleagues (1990) examined the kinematic and kinetic profiles of 15 healthy older adults compared to an existing database of healthy young adults. The authors reported reduced ankle plantar-flexor power during the propulsive phase of gait for the older adults, resulting in a shortened stride length and reduced velocity. Similarly, a more recent study demonstrated that augmenting the push off power in a sample of older adults using powered ankle-foot orthoses increased their walking velocity, albeit the change was not significant (Norris, Granata, Mitros, Byrne, & Marsh, 2007).

Reduced push off ankle power in older adults has also been observed by Judge et al. (1996). Additionally, after adjusting for differences in step length, these authors found greater hip flexor power during late stance for older adults when compared to young adults. Interestingly however, they showed that when the older adults walked at maximal pace, hip flexor power significantly increased (p<0.05) by 72% but there was no change in ankle power, further supporting reduced ankle power as a limiting factor in the altered angular kinematics and slower walking speed of older adults. Conversely, Kerrigan et al. (1998) found that differences in ankle plantar-flexor power between young and older adults remained when older adults walked at faster velocities whereas there was no difference in hip joint power between the two age groups at this speed. Riley et al. (2001) also found that older adults were unable to increase their plantar-flexor power at faster walking velocities. In a review of kinetic and kinematic changes with age, McGibbon (2003) attributed the conflicting findings between these studies to the different self selected walking velocities of the young and older sample. This idea may also explain the different findings on altered hip kinematics from studies reported previously in Chapter 2.2.3.2.

To address the issue of age-related changes in neuromuscular control at different self selected walking velocities, DeVita and Hortobágyi (2000) recruited young and older participants with the same natural walking speed. They found that older adults produced more power at the hip extensors, and less at the knee extensors and ankle plantar-flexors, than the younger counterparts. The authors suggested that this was a change in the locus of control for the older adults during walking, where neuromuscular function shifted from distal to more proximal muscles with age (DeVita & Hortobágyi, 2000). Again, the speed-dependent nature of these findings could further explain the conflicting findings on hip kinematics reported in Chapter 2.2.3.2.

Other results from DeVita and Hortobágyi (2000) provide further support for an increased reliance on hip function in older adults. Using data from a single limb, the older cohort were

found to use their hip extensors much more for support, with similar work at the hip and ankle for propulsion, whereas younger adults had similar contributions from the hip and knee for support but used the plantar-flexors predominantly for propulsion. Once again, this was attributed to a more proximal neuromuscular control strategy in the older adults. To determine whether the altered kinetics do indeed reflect age-related changes in neuromuscular lower extremity control, an examination of differences in the muscle activation patterns of older and younger adults is necessary.

2.2.3.4 Electromyography

Few studies have investigated age-related changes in muscle activation patterns during walking (Chung & Giuliani, 1997). However, a consistent finding in research employing locomotor-like tasks such as downward stepping and stair walking is that older adults have increased lower limb coactivation patterns (Hortobágyi & DeVita, 2000; Hsu, Wei, Yu, & Chang, 2007; Larsen, Puggaard, Hämäläinen, & Aagaard, 2008). Recently, this finding has also been extended to over-ground (Hortobágyi, et al., 2009) and treadmill (Mian, Thom, Ardigò, Narici, & Minetti, 2006) walking in older adults. For example, Hortobágyi et al. (2009) found increases of between 53% and 62% in the coactivation of lower extremity muscles in the older compared to their younger participants. The authors also reported a reduced temporal separation between antagonist activation in the ankle muscles of older adults. Mian et al. (2006) similarly reported an increase in antagonist coactivation in the lower limb muscles of older adults and associated this with an increase in the metabolic cost of walking. It has been suggested that the greater coactivation in older adults is an adaptive strategy to improve stability during walking via an increase in lower extremity joint stiffness (Hortobágyi & DeVita, 2000; Hortobágyi, et al., 2009; Kästenbauer, Sauseng, Brath, Abrahamian, & Irsigler, 2004).

Earlier work by Winter (1991) compared EMG variability measures between older and younger adults to investigate age-related changes in the consistency of neuromuscular patterns

during walking. Of interest, they reported that lower extremity variation was significantly reduced in older compared to their younger adult participants. As originally proposed by the Russian scientist Nikolai Bernstein, movement is controlled by coordinating a large number of individual joints and muscles (degrees of freedom), enabling a vast number of possible combinations of motor patterns to execute a single consistent task such as walking (Gielen, van Bolhuis, & Vrijenhoek, 1998). The reduced variability in muscle activation patterns of older adults in Winter's (1991) study therefore might reflect age-related declines in the flexibility of the neuromuscular system to control these large numbers of degrees of freedom. The author argues that the conservative walking pattern displayed by older adults, including shorter strides and slower walking speeds, reduces the variability of muscle activation. Interestingly, in contrast to these age-related declines in muscle activation variability, a substantial body of work has shown that the variability of gait output measures such as stride time and length increases with age. This concept of altered variability with age will be explored further in Chapter 2.3.2.2.

2.2.3.5 Neural control of gait

Age-related declines in the central and peripheral nervous systems have been established [e.g. (Lord, et al., 1996; Mold, Vesely, Keyl, Schenk, & Roberts, 2004; Raz, et al., 2005; Salat, et al., 2004)]. These decrements parallel those occurring in locomotor function, such as the altered gait parameters discussed in Chapters 2.2.3.1 to 2.2.3.4, suggesting an association between age-related neural and gait declines (Wolfson, 2001). The significance of age-related nervous system changes on walking is also suggested by associated declines in motor (Anstey & Low, 2004), gait and cognitive functions with age (Scherder, et al., 2007; Verghese, Wang, Lipton, Holtzer, & Xue, 2007; Zimmerman, Lipton, Pan, Hetherington, & Verghese, 2009), and the prevalence of gait disorders amongst individuals with neurological pathology [e.g. (Alexander, et al., 2009; Baltadjieva, Giladi, Balash, Herman, & Hausdorff, 2004; Baltadjieva, et al., 2006; Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998; Hausdorff, et al., 2000; Hausdorff, Mitchell, et al., 1997; Rosano, Brach, Longstreth Jr, & Newman, 2006; Yogev, Plotnik, Peretz, Giladi, & Hausdorff,

2007)]. The contribution of subclinical neural declines to deficits in locomotor function in older adults however is less understood (Zimmerman, et al., 2009). The following section will review evidence of declines in both higher locomotor pathways and afferent CPG feedback systems that occur with age. To date, age-related declines in spinal CPGs are yet to be explored.

2.2.3.5.1 Higher centres

Postmortem studies have shown neuronal losses of up to 50% in the brains of older adults (Shefer, 1973). Whilst general shrinkage in both grey and white matter volume with advancing age has now been established using MRI (Coffey, et al., 1992; Guttmann, et al., 2000; Resnick, Pham, Kraut, Zonderman, & Davatzikos, 2003; Salat, et al., 2004), Raz and colleagues (2005) demonstrated that these reductions are not uniform. In a five year longitudinal study of 72 healthy adults (mean age 52.49, range 22 to 77 years), the authors found that important locomotor regions such as the caudate, cerebellum and hippocampus shrunk considerably, with shrinkage in the latter two regions accelerated with advancing age. Age-related losses in cerebellar (Luft, et al., 1999) and hippocampal (Pruessner, Collins, Pruessner, & Evans, 2001) volume have also been reported by other researchers. Similar regional losses in locomotor-related regions such as the inferior frontal cortex (Resnick, et al., 2003), precentral gyrus and prefrontal cortex (Raz, et al., 1997; Salat, et al., 2004; Scahill, et al., 2003) have also been found. Despite some individual variation, neuronal declines have been shown to begin in the mid fifties (Raz, et al., 2005).

Earlier studies have reported an association between overall brain volume and basic gait measures such as walking time in healthy older adults (Camicioli, Moore, Sexton, Howieson, & Kaye, 1999), and between white matter infarctions and functional walking measures (Whitman, Tang, Lin, & Baloh, 2001). However, the relationship between regional brain neuronal loss and locomotor function has only recently been explored. Rosano and colleagues (2006) revealed that age-related gait changes such as reduced walking speed and step length were associated with

subclinical grey and white matter brain infarctions in older adults free from stroke and dementia. Later work by this team (Rosano, et al., 2008; Rosano, Aizenstein, Studenski, & Newman, 2007) was more specific, identifying that losses within the dorsolateral prefrontal regions were associated with gait speed. These authors proposed that networks within this region may maintain "pace regulation". Conversely, other work has shown reduced walking speed was associated with subclinical brainstem white matter lesion severity in older adults free from neurological disease (Starr, et al., 2003), despite previous findings of a sparing of total brainstem volume in healthy older adults (Luft, et al., 1999). Given important ascending and descending neural pathways pass through the brainstem, lesions within this structure could interfere with efferent and afferent locomotor control signals. Interruption of this communication to and from cortical regions such as the basal ganglia, cerebellum and spinal cord could therefore result in the altered gait observed in older adults.

More recently, reduced volume within the pallidum was associated with wider steps, whereas losses within the motor, supplementary motor and sensorimotor corticies of older adults were associated with shorter steps and longer step times (Rosano, et al., 2008). Shorter strides in healthy older adults have also been associated with reduced hippocampal volumes (Zimmerman, et al., 2009). Whilst the association between age-related motor cortex losses and altered stepping in older adults may be evident, the role of the hippocampus is not readily apparent. Previous work has linked hippocampal firing rates and walking speed in guinea pigs (Rivas, Gaztelu, & García-Austt, 1996), and others have associated theta activity in the mouse hippocampus with motor behaviour including locomotor movements (Bland, 2004). Although some work (Pruessner, et al., 2001) has shown small volume decrements of approximately 1.5% per year, the exact role of the hippocampus during walking and the contribution of age-related changes in the hippocampus to locomotor function in humans is yet to be elucidated. Interestingly, metabolism within the hippocampus has also been associated with increased variability of older adult stride length. Associations between age-related changes in cerebral

structures and walking variability will be explored further in Chapter 2.3.5.

2.2.3.5.2 Afferent feedback

Almost all aspects of sensory feedback are affected by normal ageing. For instance, work has shown that pain, temperature and tactile sensitivity, proprioceptive information from muscle spindles and golgi tendon organs, and visual and vestibular function all reduce with advancing age [e.g. (Gibson & Farrell, 2004; Inoue, Kuwahara, & Araki, 2004; Klein, Klein, Lee, Cruickshanks, & Gangnon, 2006; Madhavan & Shields, 2005; Ochi & Ohashi, 2003; Thornbury & Mistretta, 1981)]. Of these, tactile, proprioceptive, visual and vestibular losses have the most relevance for the changes in locomotor function observed in older adults and thus these inputs will be further explored in the following section.

Age-related alterations in the structure and function of peripheral nerves contributing to tactile sensitivity have been attributed to the ageing process itself rather than underlying clinical pathology (Wolfson, 2001). For example, prevalence rates for peripheral neuropathy and somatosensory loss in older adults free from identifiable disease have been shown to increase with advancing age (Mold, et al., 2004; Sands, et al., 1998). Interestingly, these deficits have also been associated with subjective reports of walking difficulties (Mold, et al., 2004). Similarly, reduced vibratory thresholds have been associated with lower limb performance and mobility measures (Baloh, Ying, & Jacobson, 2003; Buchman, Wilson, Leurgans, & Bennett, 2009), and more objective measures of gait function have also been related to peripheral sensation. Deficits in both vibration detection and tactile threshold were found to be significantly (p<0.01) related to reductions in walking speed and stride length in community dwelling older women (Lord, et al., 1996). However, a more recent study of 1721 adults aged between 70 and 79 reported that only vibration detection was significantly (p<0.001) associated with a slower walking speed after adjusting for covariates such as balance and visual measures (Deshpande, et al., 2008). Other work has found an independent association between reduced walking speed and moderate and

severe peripheral nerve dysfunction as measured using vibration threshold (Resnick, et al., 2000).

Associations between gait measures and proprioception are less clear. For example, a number of studies have shown age-related declines in proprioception, measured using position sense matching tasks (Adamo, Martin, & Brown, 2007; Tsang & Hui-Chan, 2004; Westlake, Wu, & Culham, 2007; You, 2005). These reductions in joint position sense, measured using a knee angle matching task, have also been associated with reduced performance on a number of functional tasks including a 15.5 m walk, in older (mean age 72 years), but not middle aged (mean age 56) or younger (mean age 23 years) adults (Hurley, Rees, & Newham, 1998). Despite these findings however, joint position sense has not been found to be correlated with specific parameters of gait. For instance, Lord and colleagues (1996) tested 183 community-dwelling women between the ages of 22 and 99 years and reported no association between first metatarsophalangeal joint position sense and a number of spatial and temporal gait measures. Similarly, Callisaya et al. (2009) found no independent association between a knee joint matching task and gait variables in 278 adults aged over 60.

Studies have shown reductions in balance under reduced visual conditions [e.g. (Judge, King, Whipple, Clive, & Wolfson, 1995; Lord & Menz, 2000)], and other work has reported an association between visual deficits and an increased falls risk (Chapter 2.1.6.2). Consequently, it is reasonable to expect that gait changes should be evident in those older adults with age-related visual declines. Indeed, artificially-induced visual alterations such as dim lighting combined with double, blurred and tunnel vision simulations have been found to reduce walking speed and increase walking variability in older men and women (Helbostad, Vereijken, Hesseberg, & Sletvold, 2009). Other work employing blurred visual conditions have also shown slower walking speeds (Deshpande & Patla, 2007) and increased toe clearances during a step-up task (Heasley, Buckley, Scally, Twigg, & Elliott, 2005) in older and younger adults. Minor deviations in lateral stepping have been reported in participants wearing a head-mounted roll vection stimuli

designed to alter visual input (Schneider, Jahn, Dieterich, Brandt, & Strupp, 2008).

Despite these apparent associations between vision and measures of gait however, few studies have directly explored the effects of visual deficits on gait variables. In a study of 18 Parkinson's disease patients with normal visual acuity, contrast sensitivity was associated with walking speed and step length (Moes & Lombardi, 2009). Of more relevance to ageing, measures of contrast sensitivity have been associated with a number of gait variables such as walking speed and stride length in community-dwelling older adults (Lord, et al., 1996) and older adults with age-related maculopathy (Wood, et al., 2009). Contrast sensitivity was also associated with performance on a six minute walk test in older adults between 62 and 95 years residing in retirement villages (Lord & Menz, 2002). Similarly, contrast sensitivity and visual field measures were associated with preferred walking speed in a population-based study of 1504 older adults aged between 72 and 92 years (Patel, et al., 2006). Interestingly, visual acuity was not associated with gait measures in any of these studies.

Declines in the vestibular system have also been reported with increasing age (Baloh, et al., 2003; Ochi & Ohashi, 2003), and these declines have been associated with increased measures of sway in older adults (Serrador, Lipsitz, Gopalakrishnan, Black, & Wood, 2009). The influence of the age-related vestibular declines on gait measures however has received limited attention. Lord and colleagues (1996) reported that performance in a test of vestibular function termed the vertical X writing test was significantly associated with spatial and temporal measures of gait in community dwelling older women. Using galvanic stimulation to alter vestibular input, Deshpande and Patla (2007) reported an increase in walking path deviation in older but not younger participants. Conversely, Baloh et al. (2003) did not find age-related declines in the vestibulo-ocular reflex gain to be related to gait function measured using the Tinetti gait and balance score. In patients with vestibular pathology, alterations in some measures of gait have been reported (Ishikawa, Edo, Terada, Okamoto, & Togawa, 1993; Ishikawa, Edo, Yokomizo, &

Togawa, 1995; Marchetti, Whitney, Blatt, Morris, & Vance, 2008), and these are further altered under reduced visual conditions (Cohen, 2000; Ishikawa, et al., 1994). Other work has shown that gait function, measured using the dynamic gait index, is associated with severity of vestibular dysfunction (Whitney, Marchetti, Pritcher, & Furman, 2009) and can prospectively identify fallers in patients with a vestibular disorder (Whitney, Hudak, & Marchetti, 2000). Consequently, despite limited research in older populations, it appears likely that there is some degree of association between age-related declines in the vestibular system and walking function.

2.3 Gait variability and stride dynamics

2.3.1 Preamble

Average or mean values of gait parameters recorded from multiple strides are commonly used to describe walking. Deviations from these values are often used to indicate changes in a locomotor system (Chapter 2.2.3). Although highly regular, these gait parameters naturally fluctuate around an average value on a stride to stride basis (Gabell & Nayak, 1984; Hausdorff, et al., 1996). Under normal conditions, these fluctuations, termed gait variability, are relatively minor, reflecting remarkable consistency and stability within the locomotor system (Hausdorff, 2005). However, these fluctuations have been shown to be altered in certain diseases (Baltadjieva, et al., 2006; Hausdorff, Mitchell, et al., 1997; Heiderscheit, Hamill, & van Emmerik, 2002; Ilg, et al., 2007), in subclinical pathology (Srikanth, et al., 2009; Zimmerman, et al., 2009) as well as in normal ageing (Herman, Giladi, Gurevich, & Hausdorff, 2005; Hollman, Kovash, Kubik, & Linbo, 2007; Maki, 1997; Owings & Grabiner, 2004b; Stolze, et al., 2000). Further, alterations in both the magnitude of these fluctuations, as well as their changes over time – termed stride dynamics – may reflect locomotor disturbances that assist in identifying both past (Barak, Wagenaar, & Holt, 2006; Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005) and future fallers (Hausdorff, Rios, et al., 2001; Maki, 1997). Consequently gait variability and dynamics might provide useful clinical information about mobility dysfunction and pathology in the

locomotor control system (Hausdorff, 2007).

This section reviews the literature on spatial and temporal gait variability, stride dynamics and falls risk, and the effect of data collection methodologies upon measures of gait variability. Other sources of variability, including environmental, instrumental and experimental factors, as well as other gait variability parameters, such the variability of intra joint coordination, kinetic, electromyographic data are beyond the scope of this thesis.

2.3.2 Gait variability

Age-related changes in mean spatial and temporal gait data provide important information on the health of the locomotor system which can be useful in identifying underlying biological disease or pathology (Chapter 2.1.6.2 and 2.2.3). Information regarding alterations in the variation of each individual step around mean values can also provide valuable insight into locomotor deficits. In fact, some studies have shown gait variability data to be a more sensitive measure of age and pathology than some averaged measures (Baltadjieva, et al., 2006; Brach, et al., 2005; Maki, 1997; Menz, Lord, & Fitzpatrick, 2003; Thies, Richardson, & Ashton-Miller, 2005; Thies, Richardson, DeMott, & Ashton-Miller, 2005). Consequently, measures of spatial and temporal gait variability are becoming important clinical tools. Prior to exploring the clinical utility of variability measures, it is necessary to briefly discuss the statistical tools that are used to record variability.

2.3.2.1 Statistical analysis of gait variability

Several statistical measures are used to record gait variability. The most common measures are the within subject standard deviation [e.g. (Brach, Studenski, Perera, VanSwearingen, & Newman, 2008; Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Maki, 1997; Mbourou, Lajoie, & Teasdale, 2003; Moe-Nilssen & Helbostad, 2005; Thies, Richardson, & Ashton-Miller, 2005; Thies, Richardson, DeMott, et al., 2005)] and the coefficient of variation [e.g. (Brach, et al., 2005; Brach, Berthold, Craik, VanSwearingen, & Newman, 2001; Dubost, et al., 2006; Gabell &

Nayak, 1984; Hausdorff, Edelberg, et al., 1997; Mbourou, et al., 2003; Öken, Yavuzer, Ergöçen, YorgancIoglu, & Stam, 2008)]. The standard deviation (SD; Equation 2.2) calculates the degree by which individual scores differ from the mean, with higher values indicating a greater spread of scores and hence greater variability (Baumgartner, Jackson, Mahar, & Rowe, 2006). The statistic is expressed in the same units as the measured data therefore providing an estimate of its relative variability. The SD is calculated using the following formula:

$$SD = \sqrt{\frac{\sum \left(X - \overline{X}^2\right)}{N - 1}} \qquad \text{Equation 2.2}$$

In addition to evaluating a gait parameter's relative variability, it is often useful to compare the variability of several gait parameters that are recorded in different units of measurement. Consequently, measures of absolute variability are also important. The coefficient of variation (CV; Equation 2.3) expresses the variability as a percentage of the mean therefore providing a measure of its absolute variability (Hopkins, 2000). The CV is calculated using the following formula:

$$CV = \left(\frac{SD}{\overline{X}}\right)100\% \qquad \text{Equation 2.3}$$

2.3.2.2 Healthy and age-related changes in gait variability

Stride to stride variation in most parameters of walking has generally been found to be small. For instance, CVs of less than 3% have been reported in healthy young adults for gait velocity (Terrier & Schutz, 2003; Terrier, Turner, & Schutz, 2005), step length (Jordan, Challis, & Newell, 2007; Terrier & Schutz, 2003; Terrier, et al., 2005), stride time (Dubost, et al., 2008; Gabell & Nayak, 1984; Hausdorff, Ashkenazy, et al., 2001; Hausdorff, Edelberg, et al., 1997; Hausdorff, Mitchell, et al., 1997; Hausdorff, et al., 1996; Hausdorff, Zemany, Peng, & Goldberger, 1999; Jordan, et al., 2007), and cadence (Terrier & Schutz, 2003; Terrier, et al., 2005). Similarly, SD values of young healthy adults walking under normal conditions have been found to be less than 1.6 cm for step length (Kang & Dingwell, 2008; Owings & Grabiner, 2004a; Owings & Grabiner,

2004b), less than 32 msec for stride time (Dingwell & Cavanagh, 2001; Dingwell, Cusumano, Cavanagh, & Sternad, 2001; Dubost, et al., 2008; Gates & Dingwell, 2007; Hausdorff, Edelberg, et al., 1997; Hausdorff, et al., 1996; Kang & Dingwell, 2008), and less than 2.5 cm for base of support (Kang & Dingwell, 2008; Owings & Grabiner, 2004a; Owings & Grabiner, 2004b; Thies, Richardson, & Ashton-Miller, 2005). These small stride to stride variations in spatial and temporal gait measures reflect the inherent stability and consistency of the neuromotor mechanisms that control normal walking (Hausdorff, 2005).

Interestingly, although ageing has been shown to increase some measures of gait variability, the majority of studies have not found these changes to be statistically significant. In one of the earliest studies investigating the effects of age on walking variability, Gabel and Nayak (1984) reported that CVs for step length, stride width, stride time and double support time were not significantly different between younger and older adults, although specific significance values were not reported. Other researchers have also reported similar magnitudes of variability between healthy younger and older adults for walking velocity (Grabiner, Biswas, & Grabiner, 2001), step width (Thies, Richardson, & Ashton-Miller, 2005), step time (Owings & Grabiner, 2004b; Thies, Richardson, & Ashton-Miller, 2005), stride time (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Hausdorff, Mitchell, et al., 1997), swing time (Hausdorff, Edelberg, et al., 1997; Springer, et al., 2006), stance time (Hausdorff, Edelberg, et al., 1997), step length (Owings & Grabiner, 2004b), stride length (Stolze, et al., 2000) and foot angle (Stolze, et al., 2000). The small magnitude of variability reported for the older adults in these studies implies that healthy ageing might not alter the locomotor mechanisms responsible for fine tuning the spatial and temporal aspects of gait on a stride to stride basis.

Despite similar gait variability measures between younger and older adults, there have been a small number of studies reporting minor age-related alterations in the stride-to-stride variation of some gait parameters. For instance, Grabiner and colleagues (2001) found significant

differences (p<0.05) in the variability of stride width, stride time and step length data. Owings and Grabiner (2004b) also reported significant increases (p=0.037) in the step width of older compared to younger adults. In another study, Owings and Grabiner (2004a) demonstrated that step width variability was able to correctly classify 77% of the sample as either younger or older adults.

Of relevance is the use of a treadmill to collect gait variability data in the latter two studies. Research has shown that treadmill walking alters averaged spatial, temporal, kinetic and angular kinematic gait data (Marsh, et al., 2006; Riley, Paolini, Croce, Paylo, & Kerrigan, 2007; Stolze, et al., 1997). Additionally, there have been suggestions that the imposed constant speed of a treadmill may artificially reduce the natural variation that occurs during over-ground walking (Dingwell, et al., 1999). Indeed, in addition to evidence of altered variability in gait parameters recorded whilst running on a treadmill (Nelson, Dillman, Lagasse, & Bickett, 1972), other work has shown significant differences in measures of gait variability recorded whilst treadmill walking compared to over-ground walking (Dingwell, et al., 2001). Further, whilst it has been reported that more than 400 steps are required to accurately estimate gait variability in younger adults walking on a treadmill (Owings & Grabiner, 2003), it is unclear how many steps are required for older adults. In fact, recent work suggests that even after 15 minutes older adults are not familiarised to treadmill walking based on lower limb kinematic data (Wass, Taylor, & Matsas, 2005). Consequently, it is possible that the conflicting findings of age-related changes in gait variability may be due in part to the use of a treadmill in the studies by Owings and Grabiner (Owings & Grabiner, 2004a; Owings & Grabiner, 2004b).

The finding of altered stride width, stride time and step length variability for the older participants in the Grabiner et al. (2001) study is less easily explained. Interestingly, a major difference between this research and those reporting no age-related differences in the variability of gait parameters is the use of a continuous walking protocol (Hausdorff, Edelberg, et al., 1997;

Hausdorff, Mitchell, et al., 1997; Owings & Grabiner, 2004b; Springer, et al., 2006), or a walking protocol that is not clearly explained (Gabell & Nayak, 1984; Thies, Richardson, & Ashton-Miller, 2005) in many of the latter studies. For instance, both studies by Hausdorff and colleagues (Hausdorff, Edelberg, et al., 1997; Hausdorff, Mitchell, et al., 1997) used footswitches to collect 6 minutes of continuous gait data, whilst Owings and Grabiner employed an instrumented treadmill for 10 to 15 minutes of continuous walking. Similarly, whilst Thies et al. (2005) and Gabel and Nayak (1984) recorded gait data on a walkway, it does not specify whether discrete or continuous trials were used. Unlike these studies, the Grabiner et al. (2001) study employed a protocol of repeated single walking trials. As demonstrated in work comparing treadmill and over-ground walking (Dingwell, et al., 2001; Nelson, et al., 1972), it is possible that the walking protocol employed to collect gait variability data affects the data. To date, the influence of walking protocol upon measures of gait variability is unclear. This topic will be explored further in Chapter 2.3.6.

2.3.3 Stride dynamics

Measures of gait variability provide an indication of the amplitude of fluctuations in a gait parameter compared to its mean over a walking trial. The variability for many of these parameters has been shown to be similar in younger and older adults reflecting a consistency of the locomotor control mechanisms across multiple decades (Chapter 2.3.2.2) (Gabell & Nayak, 1984; Springer, et al., 2006). Despite this long-term stability, when the fluctuations are observed on a stride-to-stride basis, there is random fluctuation around the mean, evident as noise in the plotted signal (Figure 2.12) (Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995). A closer analysis of the fluctuations over extended time periods however reveals a hidden temporal structure (Hausdorff, et al., 1995). The long-term fluctuations in stride variables are termed gait or stride dynamics (Hausdorff, 2007), and analysis has revealed a surprising fractal-like structure. Prior to examining the work investigating stride dynamics, a brief review of the nomenclature and techniques used to analyse the dynamics of biological signals is required.

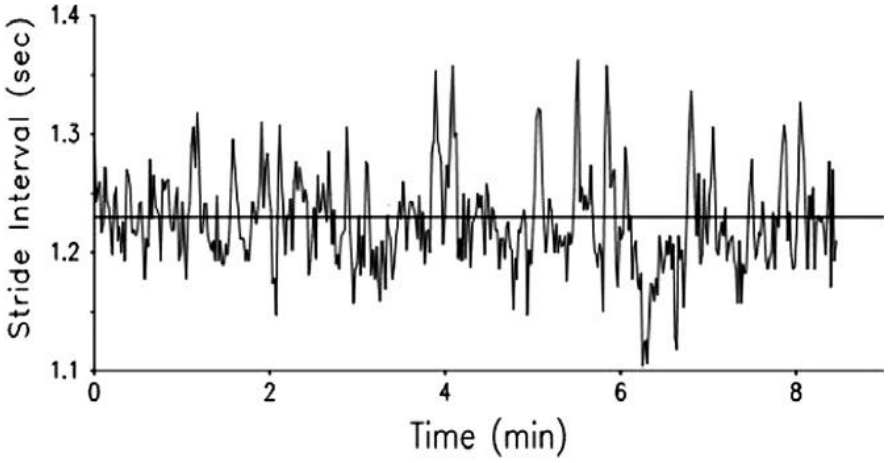


Figure 2.12. Stride to stride fluctuations in stride time. Adapted from Hausdorff et al. (1995).

2.3.3.1 Fractal analysis of stride dynamics

For an object to exhibit fractal behaviour it must possess two important properties. Firstly, it must display self-similarity so that divisions and sub divisions of the entire object statistically resemble the whole over multiple scales (Goldberger, et al., 2000). In biological signals such as stride time, this is reflected in fluctuations on one time scale being self-similar to variations occurring on other time scales. Thus variations in the stride time at any given instant are "related" to variations occurring hundreds or even thousands of strides earlier (Hausdorff, et al., 1995). Secondly, unlike Euclidean objects (such as cubes) that have integer dimensions, the self-similar structure of fractals have fractional scale-free dimensions (Goldberger, et al., 2000). In a temporal sequence with fractal properties, this is reflected in the power-law relationship of an increase in the amplitude of fluctuations with an increased observation window size (Herman, et al., 2005).

A number of techniques exist to examine the temporal structure of biological signals, including spectral analysis (Hausdorff, et al., 1995; Hausdorff, et al., 1999), autocorrelation (Hausdorff, et al., 1999) and detrended fluctuation analysis (DFA) (Goldberger, et al., 2000; Hausdorff, et al., 1995; Hausdorff, et al., 1996; Herman, et al., 2005; Peng, Havlin, Stanley, &

Goldberger, 1995). Of these, DFA has the advantage of detecting self-similarity in nonstationary time series (i.e. series where the mean, standard deviation and higher moments are unaffected by changing the time window), it can be used over longer time scales, and it is able to avoid the false detection of extrinsic as opposed to biologically-derived self-similarity (Goldberger, et al., 2000). Consequently, DFA has become widely used in the analysis of long-range fractal correlations in a number of biological signals (Jordan, Challis, Cusumano, & Newell, 2009).

DFA is a modified root mean square analysis of an integrated and detrended time series performed repeatedly over multiple time scales. For each observation window (i.e. time scale), DFA firstly integrates the time series and divides it into boxes of equal length, *n*. The integrated series in each box, *y(k)*, is then detrended by fitting it with a least squares line and subtracting the local trend, $y_n(k)$. The root mean square of the fluctuation, *F(n)*, at each observation is calculated by:

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}\left[y(k) - y_n(k)\right]^2} \qquad \text{Equation 2.4}$$

The relationship between the fluctuation magnitude, *F(n)*, and the window size, n, is determined by repeating the calculation over the total time scale (Goldberger, et al., 2000). Power-law scaling (an increase in *F(n)* with an increase in window size) indicates fractal-like self-similarity, and this linear relationship, plotted on a log-log graph, is represented by the scaling exponent α (Goldberger, et al., 2002; Herman, et al., 2005). For uncorrelated random fluctuations, α=0.5, whereas for long-range power law correlations 0.5>α≤1.0 and Brownian noise, α=1.5 (Goldberger, et al., 2000; Goldberger, et al., 2002; Hausdorff, 2007).

Fluctuations in the time series of some biological systems have revealed a hidden fractal structure. For instance, using DFA to evaluate the dynamics of the healthy human heartbeat, many studies have shown complex and irregular fluctuations over long time scales with a scaling

exponent close to α=1.0 [e.g. (Havlin, et al., 1999; Peng, et al., 1995; Peng, et al., 1993)]. These long-range fractal correlations have been shown to be altered in patients with ventricular tachycardia (Baumert, Wessel, Schirdewan, Voss, & Abbott, 2007), fatal or near-fatal sudden cardiac death syndrome (Amaral, Goldberger, Ivanov, & Stanley, 1998; Havlin, et al., 1999; Peng, et al., 1995; Rodriguez, Lerma, Echeverria, & Alvarez-Ramirez, 2008), adults with severe heart failure (Amaral, et al., 1998; Havlin, et al., 1999; Peng, et al., 1995; Voss, et al., 2008), as well in healthy older adults (Beckers, Verheyden, & Aubert, 2006; Iyengar, Peng, Morin, Goldberger, & Lipsitz, 1996; Stein, Barzilay, Chaves, Domitrovich, & Gottdiener, 2009). Significantly, the technique has also proved useful in predicting mortality in patients with various cardiac diseases (Ho, et al., 1997; Huikuri, et al., 2000; Mäkikallio, et al., 2001; Stein, et al., 2008).

In conjunction with this work in heart rate dynamics, some researchers also explored the influence of physical activity on heart rate dynamics by concurrently recording stride-to-stride and beat-to-beat signals (Hausdorff, Forman, Pilgrim, Rigney, & Wei, 1992). It was revealed that gait instability was related to poor cardiovascular health, suggesting that the study of stride dynamics might also provide further clinically useful information. This outcome subsequently led to further investigation of long-range fractal dynamics in walking (Hausdorff, et al., 1995).

2.3.3.2 Healthy and age-related changes in stride dynamics

The majority of work investigating stride dynamics has used stride time, also known as the stride interval (Gates & Dingwell, 2007; Hausdorff, Mitchell, et al., 1997; Hausdorff, et al., 1995; Hausdorff, et al., 1996; Hausdorff, Rios, et al., 2001; Hausdorff, et al., 1999; Herman, et al., 2005). The stride interval can be considered the “final output” of the neuromotor control system and as such, variability in this signal might provide insight into higher rhythmic generating centres (Hausdorff, 2007). If the variations were random, then the timing of fluctuations of one stride could be expected to be completely independent and uncorrelated with the value of each subsequent stride (Gates & Dingwell, 2007). Such random noise might be due to experimental

errors or deficits within the neuromotor control of walking (Buzzi, Stergiou, Kurz, Hageman, & Heidel, 2003). Alternatively, the stride-to-stride variations might be attributable to short-range correlations whereby a given stride is only related to the immediate strides around it. If this were the case, a short stride might be followed by a longer stride or vice versa, but the variations would be random and unrelated over the longer-term (Hausdorff, et al., 1995). Over a decade ago however, it was revealed that the fluctuation dynamics of gait are not random, nor are they simply due to short-term correlations. In a sample of ten healthy young males walking continuously for 9 minutes, the duration of each stride was shown to be statistically correlated to each other stride over multiple time periods (α=0.83) (Hausdorff, et al., 1995). The long-range correlations in the stride time fluctuations were found to decay in a power-law manner suggesting a fractal structure, and were also shown to break down following random shuffling of the data ($\alpha \approx 0.50$) (Hausdorff, et al., 1995; Hausdorff, et al., 1996).

Further investigation into the stride dynamics of young healthy adults has revealed that this fractal pattern is stable over much longer time periods, with long-range correlations (α=0.84) found over thousands of strides recorded during approximately one hour of continuous over-ground walking (Hausdorff, et al., 1996). Using DFA, these fractal patterns in stride dynamics have also been found when walking at slower and faster than normal speeds (Hausdorff, et al., 1996; Jordan, et al., 2007), whilst walking over-ground using global positioning system (GPS) devices (Terrier, et al., 2005), and during treadmill walking using both an imbedded force platform (Jordan, et al., 2007) and a 3D optical motion analysis system (Pierrynowski, et al., 2005). Long term fractal correlations have also been shown to be present in a number of other gait parameters in addition to stride time (Jordan, et al., 2007). Combined, these studies show that the fractal nature of the gait rhythm is stable for a variety of locomotor parameters and behaviours thus providing new insight in the neuromotor control mechanisms governing human walking.

Interestingly, age-related changes in stride dynamics have also been reported in healthy older adults free from locomotor-related pathology. In contrast to the findings of similarities in the magnitude of variability between healthy younger and older adults (Gabell & Nayak, 1984; Grabiner, et al., 2001; Hausdorff, Mitchell, et al., 1997; Owings & Grabiner, 2004b; Springer, et al., 2006; Stolze, et al., 2000; Thies, Richardson, & Ashton-Miller, 2005), fractal analysis of stride dynamics has revealed significantly reduced (p<0.003) long-range correlations in the stride interval of healthy older adults (α=0.68) compared to younger controls (α=0.87) (Hausdorff, Mitchell, et al., 1997). Age-related changes in stride dynamics have also been reported for additional locomotor parameters recorded using other gait and analytical methodologies. For instance, Buzzi and colleagues (2003) found that the Lyapunov exponent – a nonlinear measure of local stability – was significantly higher and thus more unstable for all lower extremity kinematic parameters of older adults recorded during treadmill locomotion. Similarly, alterations in the randomness of toe clearance data in healthy older adults and older adult fallers compared to healthy younger adults has been reported using approximate entropy analysis (Karmakar, Khandoker, Begg, Palaniswami, & Taylor, 2007). Despite each of these studies screening participants for neurological or other impairments that might affect their gait, it is possible that the age-related changes in stride dynamics are due to subclinical neurological pathology. This theory will be explored further in the following section.

2.3.4 Gait variability, stride dynamics and falls risk

Some averaged spatial and temporal gait parameters have been shown to be different between older fallers and non-fallers (Besser, et al., 2000; Guimaraes & Isaacs, 1980; Montero-Odasso, Schapira, Duque, et al., 2005; VanSwearing, et al., 1998; Woo, et al., 1995), and a small number of studies have shown that changes in some of these parameters might result in a slightly increased falls risk (Hill, et al., 1999; Lord, et al., 1996; Verghese, et al., 2009). Interestingly, when examined further, many of these altered mean gait parameters were actually associated with a fear of falling, and when adjusted for this measure were no longer predictive of

future fallers (Maki, 1997). Consequently, changes in mean gait parameters may not reflect instability within the locomotor system and instead may be a conscious or unconscious adaptation to fear.

Conversely, measures of gait variability and stride dynamics provide an indication of the stride-to-stride fluctuations in gait parameters and as such are more likely to indicate instability during walking (Hausdorff, Nelson, et al., 2001). It is reasonable therefore to expect that these measures might be better able to distinguish between older fallers and non-fallers and predict individuals at an increased risk of falling. To date, the majority of studies investigating age-related changes in the gait variability of fallers have employed retrospective study designs, with only a small number exploring the ability of altered gait variability to prospectively identify future fallers. No study has investigated stride dynamics in healthy older fallers. This section will review the literature employing both of these experimental designs to examine gait variability and dynamics.

Compared to healthy older adult non-fallers, an increase in the magnitude of variability of fallers based on retrospective self reporting has been shown for several gait parameters. For instance, work has found that fallers walk with an increase in the variability of stride time (Hausdorff, Edelberg, et al., 1997; Hausdorff, Rios, et al., 2001; Herman, et al., 2005), swing time (Hausdorff, Edelberg, et al., 1997; Hausdorff, Rios, et al., 2001; Springer, et al., 2006), stance time (Hausdorff, Edelberg, et al., 1997), and stride width (Brach, et al., 2005). Similar to conflicting reports of age-related differences in gait variability however, other work has not found altered gait variability in fallers. For instance, Brach and colleagues (2005) reported no difference in the variability of step length, stance time or step time for fallers and non-fallers. Similarly, Heitman et al. (1989) reported statistically similar levels of step width variability between fallers and non-fallers.

There are differences in the walking protocol employed to collect gait variability data in the above studies. For example, much of the research reporting differences in the variability of fallers compared to non-fallers employed continuous walking protocols (Hausdorff, Edelberg, et al., 1997; Hausdorff, Rios, et al., 2001; Herman, et al., 2005). Conversely, those studies reporting no difference used a repeated single trial protocol (Brach, et al., 2005; Heitman, et al., 1989). Despite the possibility of walking protocol influencing gait variability data, no study has examined the relationship between the two. The issue of gait protocol and its effects on walking variability will be explored further in Chapter 2.3.6.

To date, three studies have employed a prospective study design to examine gait variability in fallers. In a study of 75 residents living a supported care facility, increased stride length, velocity and double-support time variability were predictive (odds ratios between 1.95 and 2.30) of future falls in a 12 month follow up period (Maki, 1997). Of note, whilst averaged gait measures (e.g. mean stride length and velocity) were also significantly altered in the fallers, after adjusting for fear of falling, only the variability measures were predictive of falling independent of fear. This would suggest that average gait parameters are more likely be an adaptation to fear, whereas increased walking variability might indicate increased walking instability and heightened risk of a fall (Maki, 1997).

Hausdorff and colleagues (2001) recorded the SD of 52 community-dwelling older adults to investigate whether altered gait variability predicts older adult fallers in a 12 month prospective study. Using logistic regression, it was reported that a 1 SD increase in stride time variability at baseline was associated with a fivefold increased risk of falling in the subsequent 12 months. Survival analysis also revealed that those older adults with greater gait variability were more likely (p=0.002) to fall sooner than those with less variability. Surprisingly, the groups did not differ with respect to many other parameters of physical health (age, gender, height, weight, activity levels, ability to perform activities of daily living , number of medications), mental health

(mini mental state examination, geriatric depression scale), functional balance (timed up and go, functional reach, performance oriented mobility assessment) or average gait parameters (speed and walk duration). Similar findings were also reported by Verghese et al. (2009), who found increased swing time and stride length variability predicted future falls in older adults aged over 70 years. Unlike Hausdorff et al. (2001) however, other gait measures were also predictive of future falls, including reduced walking speed and swing phase time, and increased double support phase time. It is possible that the increased prospective period in the Verghese study (up to 42 months with a mean of 20 months) contributed to the different predictive ability of the mean gait parameters between the studies.

Although the findings from prospective gait variability studies show that measures of gait variability may be useful in identifying future fallers, it is possible that these studies were comprised of older adults with some mobility limitations. For example, nearly a third of participants (29.3%) in the Maki (1997) study reported sometimes using a cane and one fifth (18.7%) walked outside less than once per week. Similarly, participants in the Verghese (2009) study were taking an average of five medications and 35% reported some kind of gait abnormality. The average walking speed of participants in each of these studies (0.74-0.93 m/s) was also well below the average velocity of the majority of studies on older adults listed in Table 2.2 previously. Finally, in each of the three studies, the inclusion and exclusion criteria did not specifically exclude participants with medical conditions that might impact upon their walking or balance ability (see Table 2.3). Consequently, apart from the confounding influence of walking speed upon measures of gait variability (Moe-Nilssen & Helbostad, 2004; Moe-Nilssen & Helbostad, 2005), it is also possible that underlying pathology might have contributed to the findings. It would be of interest to evaluate whether gait variability is a useful marker of falls risk in more a rigorously screened and active sample of healthy older adults. Given the high frequency of falls occurring during walking in active older adults (Chapter 2.1.6.1), and the increased further risk of falls, morbidity and institutionalisation from an initial fall (chapter 2.1.5), a sensitive

marker of falls risk in this population would have considerable clinical value.

Table 2.3. Inclusion and exclusion criteria of prospective studies examining gait variability in older adult fallers.

	Inclusion criteria	Exclusion criteria
Maki (1997)	• Able to walk 10 m without walking aid, able to stand unaided, able to understand verbal instruction, no falls in previous month	• NA
Hausdorff et al. (2001)	• Ambulatory	• Severe cognitive impairment , nursing home residents, or <1 year life expectancy
Verghese et al. (2009)	• NA	• Severe auditory or visual loss, bedbound or institutionalised

Note: NA, not available.

The ability of stride dynamics to identify older adult fallers has only been explored in one study, albeit using patients with a walking pathology in a retrospective study design. Herman and colleagues (2005) initially compared the stride dynamics and magnitude of variability in a sample of 25 patients classified as having a "higher level gait disturbance" (HLGD) with 28 healthy age-matched control. Older adults with a HLGD have an altered gait pattern that is diagnosed by clinical presentation and is not directly attributable to a specific motor or sensory pathology, or to a clinically diagnosed condition such as stroke or Parkinson's disease (Elble, 2007; Nutt, et al., 1993; Snijders, van de Warrenburg, Giladi, & Bloem, 2007). Analysis of fractal stride dynamics in this population therefore may provide insight into the neuromotor changes behind these subclinical gait disturbances. The authors found that all measures of variability were significantly different (p<0.03) in the two groups and that the altered gait variability was not associated with changes in other physical or cognitive measures. Further stratification of the HLGD patients into fallers and non-fallers (based on self report of a fall in the previous 12 months) revealed that the only difference between the two groups was a significantly reduced (p<0.009) fractal scaling index in the fallers. Other clinical measures, including variability, were unchanged. Therefore, while measures of the variability and stride dynamics were useful in differentiating between

individuals with a subclinical gait disorder and healthy controls, only an analysis of stride dynamics revealed further information regarding fall history. It would be of interest to determine whether stride dynamics also prove useful in prospectively identifying future fallers, and if so, whether the technique offers superior predictive power over measures of variability. Currently, no study has investigated the use of stride dynamics to prospectively identify older adult fallers.

2.3.5 Neural origins of gait variability and stride dynamics

Age-related changes in a number of systems contributing to locomotion have been proposed to affect gait variability and fall risk (Figure 2.13). For instance, work has shown increased variability to be associated with sensory loss (Dingwell & Cavanagh, 2001), spinal stenosis (Papadakis, et al., 2009), strength and range of motion deficits (Hausdorff, Rios, et al., 2001; Kang & Dingwell, 2008), altered muscle activation patterns (Kang & Dingwell, 2009), increased cognitive requirements (Beauchet, Dubost, Herrmann, & Kressig, 2005; Dubost, et al., 2008; Dubost, et al., 2006; Springer, et al., 2006), cardiovascular pathology (Hausdorff, et al., 1994; Hausdorff, Herman, Baltadjieva, Gurevich, & Giladi, 2003), peripheral vascular disease (Myers, et al., 2009), psychological and psychiatric conditions (Hausdorff, Peng, Goldberger, & Stoll, 2004; Herman, et al., 2005; Maki, 1997), balance deficits (Hausdorff, Rios, et al., 2001) and various central nervous system disorders (Hausdorff, et al., 1998; Hausdorff, et al., 2000; Hausdorff, Mitchell, et al., 1997; Nakamura, Meguro, & Sasaki, 1996; Webster, Merory, & Wittwer, 2006). Consequently, it is apparent that many physiological and psychological factors contribute to gait variability. However, whilst these studies support the idea that variability is particularly influenced by both central and peripheral influences (Hausdorff, 2007), the origin of the control of stride dynamics is unclear. The studies that have explored this aspect of gait, and others investigating the neural correlates of locomotion, seemingly suggest that central neural influences are of greater importance.

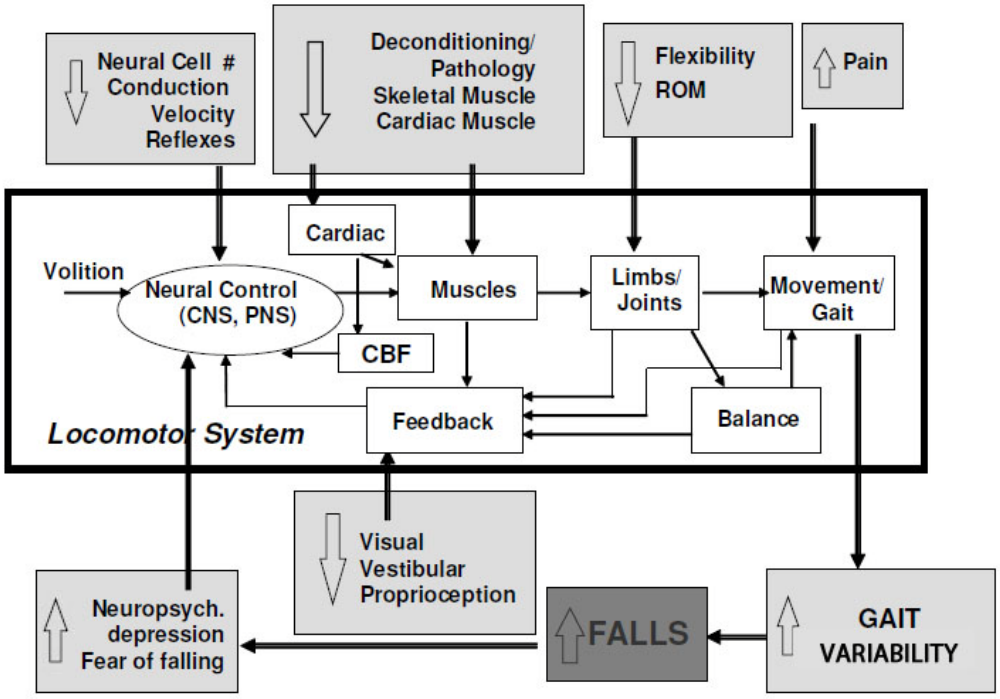


Figure 2.13. The inter-relationship between ageing, physiological systems that contribute to locomotion, gait variability and falls. CBF: cerebral blood flow; CNS: central nervous system; PNS: peripheral nervous system. Modified from Hausdorff (2005).

Initial work exploring the origins of stride dynamics investigated the question of whether central feed-forward or peripheral feedback mechanisms were responsible for the long-range correlations in walking. Hausdorff and colleagues (1996) compared the stride dynamics of healthy young adults whilst free walking and walking in time to a metronome that was set at the participant's average stride interval. If stride dynamics were centrally generated, these long-range correlations would be expected to breakdown during the metronomic condition, as the external rhythm of the metronome would override central internal pace generators. Conversely, given that both walking conditions employ the same central and peripheral neuromotor mechanisms, long-range correlations would be expected to be unchanged in the metronome condition if feedback from peripheral afferents were responsible for fractal gait dynamics. The results revealed a breakdown ($\alpha \approx 0.50$) in the long-range correlations whilst metronomic walking, suggesting that the mechanisms producing these long-term fractal properties in stride dynamics are likely to be supraspinal (Hausdorff, et al., 1996). The suggestion of higher central control over

the fractal properties of gait dynamics has since been supported by the reduction in long-term correlations in Huntington's disease (α=0.60) (Hausdorff, Mitchell, et al., 1997) and Parkinson's disease (α=0.65) (Bartsch, et al., 2007; Frenkel-Toledo, et al., 2005), and persistence in patients with peripheral neuropathy (α=0.88) who displayed significantly increased (p<0.04) gait variability (Gates & Dingwell, 2007).

If central neurological mechanisms contribute to long range correlations in walking, the findings of altered stride dynamics in healthy older adults (Chapter 2.3.3.2) and older fallers (chapter 2.3.4) may in part be explained by subclinical neurological pathology. This is consistent with reports of subclinical age-related changes in cerebral grey and white matter (Rosano, et al., 2006), and findings of an association between these neural changes and alterations in some temporal (Rosano, et al., 2008; Rosano, Aizenstein, et al., 2007; Starr, et al., 2003) and variability (Rosano, Brach, Studenski, Longstreth Jr, & Newman, 2007) measures of gait reported in Chapter 2.2.3.5. However, it is unclear which higher neurological centres might be responsible for generating stride dynamics, and thus which are altered in ageing. For instance, nuclei located within the basal ganglia (Takakusaki, et al., 2003) and cerebellum (Ilg, et al., 2007; Jahn, et al., 2004; Mori, et al., 2001) have been shown to be associated with temporal components of locomotion (Chapetr 2.2.1.5.2). Specifically, basal ganglia nuclei such as the substantia nigra and the globus pallidus are generally accepted to be responsible for selecting and initiating motor programs (Grillner, et al., 2005; Grillner, et al., 2008), including locomotion (Brudzynski, et al., 1993; Grillner, et al., 2005), and have been shown to influence temporal locomotor parameters (Takakusaki, et al., 2003). Similarly, the cerebellum is widely believed to act as a comparator for coordination during motor tasks (Marple-Horvat & Criado, 1999; Pardoe, et al., 2004), and work has also shown an association between cerebellar vermis activation and locomotor speed (Jahn, et al., 2004). Recently, other authors have argued that neural circuits located in a component of the mescencephalic locomotor region known as the pedunculopontine nucleus, connect these two structures with the spinal central pattern generators, and thus could also play a role in

initiating and modulating gait rhythmicity (Pahapill & Lozano, 2000). This latter hypothesis is strengthened with reports of an association between severity of subclinical brainstem white matter lesions and temporal gait measures in older adults free from neurological pathology (Starr, et al., 2003).

Although the specific origin of neural centres influencing stride dynamics is yet to be determined, it is apparent that measures of gait dynamics have the potential to provide novel insight into subclinical neurological changes within the locomotor system. In light of the link between sub clinical neural pathology and gait changes in older adults, and between gait changes and falling, measures of gait variability and dynamics might provide a sensitive marker of sub clinical pathology and falls risk in an otherwise healthy and active older adult population. Given the portable, relatively simple and inexpensive nature of the technique, these measures offer an exciting clinical application for the evaluation of gait instability and falls risk older in adults.

2.3.6 Methodological considerations for gait variability and stride dynamics

Studies investigating gait variability in older adults have employed a diverse range of research methodologies. As discussed however, the effect of experimental factors upon walking variability is unclear. The differences between studies make comparison difficult and can also preclude the generalisation of research outcomes to alternate populations or other forms of walking. As such, it is necessary to explore the effects of walking methodology upon measures of gait variability.

The main difference between many studies investigating gait variability lies in the use of either a treadmill [e.g. (Kang & Dingwell, 2008; Owings & Grabiner, 2004a)] or an over-ground walking protocol [e.g. (Hausdorff, Edelberg, et al., 1997; Hausdorff, Nelson, et al., 2001; Hausdorff, Rios, et al., 2001)]. Treadmills are favoured by many researchers due to space restrictions, the ability to control speed and the ease of use in collecting the required number of steps to analyse gait variability data (Owings & Grabiner, 2003; Parvataneni, Ploeg, Olney, &

Brouwer, 2009; Riley, et al., 2007; Wass, et al., 2005). The incorporation of safety features such as a body harness may also favour the use of a treadmill over over-ground walking (Crompton, et al., 2001; Parvataneni, et al., 2009), although use of a safety harness itself may result in alterations to an individual's gait pattern either directly or by decreasing fear of falling (Aaslund & Moe-Nilssen, 2008; Pillar, Dickstein, & Smolinski, 1991).

Despite these benefits however, differences in many gait parameters have been reported between continuous over-ground and treadmill walking (Marsh, et al., 2006; Riley, et al., 2007; Stolze, et al., 1997) and running (Nelson, et al., 1972). Although a study has reported that some measures of gait variability are unchanged by treadmill walking (Chang, Shaikh, & Chau, 2009), others have found that both gait variability and stride dynamics are altered during treadmill walking (Dingwell, et al., 2001; Frenkel-Toledo, et al., 2005). Further, as outlined in Chapter 2.3.2.2, older adults require greater than 15 minutes to familiarise to treadmill walking (Wass, et al., 2005), and more than 400 steps are needed to accurately estimate the gait variability of younger adults (Owings & Grabiner, 2003). Consequently, it is unclear if the alterations in gait variability between younger and older adults (Owings & Grabiner, 2004a; Owings & Grabiner, 2004b) and between fallers and non-fallers (Barak, Wagenaar, & Holt, 2006) reported by studies using treadmills are due to intrinsic differences between the populations, or to the walking protocol employed to evaluate gait variability. Given the findings of altered gait variability and stride dynamics whilst treadmill walking therefore, and the lack of ecological validity of this data collection technique, it is difficult to generalise these findings to over-ground walking. As such, employing a treadmill might not be the preferred protocol to collect and evaluate gait variability data in older adults.

A corollary issue to the influence of over-ground walking protocol upon gait variability therefore, is whether to collect consecutive steps whilst walking continuously or non-consecutive steps during repeated single trials. Whereas the majority of studies investigating stride dynamics

have necessarily employed a continuous walking protocol [e.g. (Hausdorff, Mitchell, et al., 1997; Hausdorff, et al., 1995; Hausdorff, et al., 1996; Hausdorff, et al., 1999; Herman, et al., 2005)], those recording gait variability have employed both continuous [e.g. (Owings & Grabiner, 2004a; Owings & Grabiner, 2004b; Springer, et al., 2006; Thies, Richardson, DeMott, et al., 2005)] and repeated single [e.g. (Grabiner, et al., 2001; Stolze, et al., 2000)] walking protocols. Given the frequent disruptions to spatial and temporal aspects of gait that occur with repeated single walking trials, it is possible that fluctuations in many stride parameters are altered using this protocol. Indeed, the validity of recording gait parameters such as variability from non-consecutive steps has been questioned by some authors (Dingwell, et al., 2001; Owings & Grabiner, 2003).

Support for this theory can be found in the different age and fall-related outcomes of the studies employing these two walking protocols. For instance, the majority of studies that have reported no age-related changes in gait variability magnitude have employed a continuous over-ground walking protocol (Hausdorff, Edelberg, et al., 1997; Springer, et al., 2006), whereas those that have found age differences have used a repeated single over-ground (Grabiner, et al., 2001; Stolze, et al., 2000) or continuous treadmill (Owings & Grabiner, 2004a; Owings & Grabiner, 2004b) walking protocol. Further, studies that have found altered gait variability magnitude between healthy older adult non-fallers and older fallers (Hausdorff, Rios, et al., 2001) and those with a higher level gait disorder (Herman, et al., 2005) have used a continuous over-ground walking protocol, whereas studies reporting no differences between fit and frail older adults (Moe-Nilssen & Helbostad, 2005) and older fallers and non fallers (Brach, et al., 2005; Heitman, et al., 1989) have employed a repeated single trial protocol. Although there are many other differences in the methodologies of these studies, it is plausible that at least in part, using consecutive steps whilst walking continuously versus non-consecutive steps during repeated single trials contributed to the different findings of these studies. The effect of over-ground walking protocols upon gait variability data has not been investigated.

2.4 Chapter summary

Falls among older adults have important financial, behavioural, psychological and physical consequences. Such consequences are expected to increase as the population grows and as the proportion of adults aged over 65 increases. Consequently, an important health care challenge is the identification of markers of falls risk in older adults. In particular, identifying older adults prior to falling will help prevent subsequent falls, inactivity, morbidity and institutionalisation. A sensitive marker of falls risk in active and otherwise healthy older adults therefore is of considerable clinical value.

Research has identified a number of intrinsic and extrinsic falls risk factors. However, with the high incidence of falls occurring outdoors whilst walking, markers of gait decline show particular promise in identifying falls risk. As such, kinematic, kinetic, and EMG measures are frequently employed to investigate age-related changes in gait. A number of alterations in the walking pattern of older adults have been shown in such gait analysis studies, the most consistent being reduced walking speed and shorter steps in older adults. With parallel declines in many neural systems, age-related gait changes have also been associated with sub clinical pathology. Other work however has suggested that the slower speed and shorter steps of older adults is related more to fear than pathology. It was instead proposed that measures of stride to stride fluctuations such as gait variability and stride dynamics might better reflect gait instability thereby providing a more useful marker of future falls.

To date, there is some evidence that gait variability could be a more sensitive marker of falls risk. Additionally, preliminary research suggests that stride dynamics could offer further promise as a subtle measure of walking instability in the absence of other gait or physical changes. Importantly, recent medical imaging and clinical studies have associated alterations in both gait variability and stride dynamics with clinical and sub clinical pathology in important neural locomotor regions. However, there are a number of outstanding issues that require further

investigation before the clinical utility of gait variability and dynamics is understood.

Of prime importance, no study has prospectively investigated gait variability and stride dynamics in a sample of active and healthy older adults free from recent falls. Past prospective gait variability studies have been limited to older adults already showing some evidence of mobility problems, and stride dynamics have only been studied in healthy younger adults or clinical populations. A greater understanding of variability and dynamics in active and healthy older adults might provide a sensitive marker of gait decline, thereby identifying falls risk before other locomotor or physical changes are evident. Additionally, the conflicting outcomes of many gait variability studies using different populations, and the different walking protocols employed by these studies, has made understanding the clinical value of gait variability problematic. By exploring the influence of walking protocol upon gait variability in a sample of active and healthy older adults, an understanding of both protocol and population influences upon gait variability will be gained. Therefore, the general aim of this investigation was to advance knowledge about the role of gait variability and stride dynamics in an active older adult population, and to assess whether these measures predict falls in active and otherwise healthy older women.

3 Methods

3 Methods

This thesis is comprised of three studies. The following chapter provides an overview of the progression of these studies and describes methodological aspects common to each. Specific methodological issues for each study are described in Chapters 4 to 6.

3.1 Thesis overview

The main objective of this thesis was to evaluate the contribution of stride dynamics and gait variability to predicting falls in active community-dwelling older women. Two common over-ground walking protocols are often employed to collect variability data, however the reliability of one of these, where the participant walks continuously for a fixed period of time, had not been explored. Consequently, study 1 of this thesis investigated the test-retest reliability of a continuous walking protocol to collect basic gait parameters in older and younger women. Additionally, the influence of walking protocol upon measures of gait variability was also unknown. Therefore study 2 explored whether measures of gait variability were altered by employing either the continuous walking protocol or a more traditional "single trial" walking protocol in younger and older adults. Finally, using each of these walking protocols, the ability to predict active older fallers using measures of gait variability and dynamics was investigated. These three stages form the basis of this thesis as outlined in Figure 3.1 and Table 3.1 below.

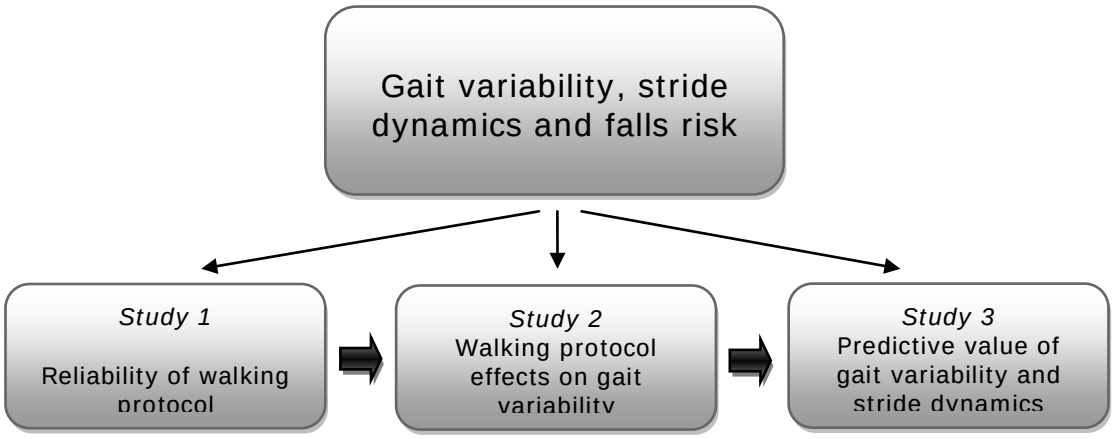


Figure 3.1. Overview of thesis design.

Table 3.1. Study objectives and statistical treatments.

	Objective	Key Statistical Treatment
Study 1		
Test-retest reliability of walking protocol	To examine the test-retest reliability and systematic bias in spatiotemporal gait parameters recorded in healthy younger and older women during repeated single and continuous over-ground walking trials.	• Measures of test-retest reliability between testing sessions. *Statistical analysis: intra-class correlation coefficients, standard errors of measurements and coefficients of variation.* • Systematic differences between testing sessions for younger and older women. *Statistical analysis: dependent t tests*
Study 2		
Walking protocol effects on gait variability	To investigate the effect of single and continuous over-ground walk protocols upon measures of gait variability commonly used to assess instability and falls risk in older women.	• Differences between spatial and temporal gait parameters collected during single and continuous walking trials with associated effect sizes. *Statistical analysis: dependent t tests* • Differences between the variability of spatial and temporal gait parameters collected during single and continuous walking trials with associated effect sizes. *Statistical analysis: dependent t tests.*
Study 3		
Predictive value of gait variability and stride dynamics	To evaluate measures of gait instability to predict falling in active community dwelling older women.	• Differences in gait variability and stride dynamic measures between fallers and non-fallers, and prediction of older fallers using these gait variables. *Statistical analysis: multiple analyses of variance, independent t tests, dependent t tests, logistic regression.*

3.2 Participants

3.2.1 Ethical approval

The Human Research Ethics Committee at Australian Catholic University approved all procedures for this study. All participants completed informed consent forms (Appendix 1).

3.2.2 Participant recruitment

The older women were recruited through advertisements distributed to local senior social groups and senior newspapers and bulletins, whereas the young women were recruited through

advertisements placed on University notice boards. A lower age band of 55 years was chosen for the older sample as research has shown alterations in common gait parameters such as walking speed and step length for women aged between 50 and 59 years (Oberg, Karsznia, & Oberg, 1993). Women were chosen because of the higher incidence of falls and fall-related morbidity in this population (Stevens & Sogolow, 2005), and because gender effects have been reported for some spatial and temporal gait measures (Laufer, 2003; Moe-Nilssen & Helbostad, 2005). Interested persons then contacted the principal investigator (KP) by phone to volunteer for the study. If inclusion criteria were met, information letters (see Appendix 2) were sent to the prospective participants and an appointment time was made. Study-specific participant information, such as exclusions and sample characteristics, are provided in Chapters 4 to 6.

3.2.2.1 Young women inclusion criteria

The inclusion criteria for the young women were:

- Aged between 18 and 35;
- In good health with no recent illness or injury (6 weeks);
- No recent hospitalisation or surgery (6 months);
- No medical condition or medication with a known detrimental effect on gait; and
- Able to walk unassisted (including no gait aids) and pain free for 10 minutes.
- Active for a minimum of 30 minutes at least one day per week (see Chapter 3.4.1.6 for definition of active)

3.2.2.2 Older women inclusion criteria

The inclusion criteria for the older women were:

- Aged between 55 and 90;

- In good health with no recent illness or injury (6 weeks);
- No recent hospitalisation or surgery (6 months);
- No falls in the previous month;
- No medical condition or medication with a known detrimental effect on gait; and
- Able to walk unassisted (including no gait aids) and pain free for 10 minutes.
- Active for a minimum of 30 minutes at least one day per week (see Chapter 3.4.1.6 for definition of active)

3.3 Instrumentation

3.3.1 GAITRite®

The GAITRite® walkway system (CIR Systems, Inc, Havertown, PA) was used to collect spatial and temporal walking data for studies 1 to 3. The reliability and validity of the walkway has been established for a variety of populations including children (Thorpe, Dusing, & Moore, 2005), healthy younger adults (Bilney, Morris, & Webster, 2003; Cutlip, Mancinelli, Huber, & DiPasquale, 2000; Gretz, et al., 1998; McDonough, Batavia, Chen, Kwon, & Ziai, 2001; Menz, Latt, Tiedemann, Kwan, & Lord, 2004; van Uden & Besser, 2004), healthy older adults (Menz, et al., 2004), people with Huntington's disease (Rao, Quinn, & Marder, 2005), cerebral palsy (Sorsdahl, Moe-Nilssen, & Strand, 2008), Parkinson's disease (Nelson, et al., 2002), Alzheimer's disease (Wittwer, Webster, Andrews, & Menz, 2008), Down syndrome (Gretz, et al., 1998) and recent knee surgery (Webster, Wittwer, & Feller, 2005). It is important to note however that each of these studies used a single walking trial protocol. Details of these studies are outlined in Table 3.2 below.

Table 3.2. Summary of studies reporting reliability and/or validity of the GAITRite® walkway system.

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics
Bilney et al. (2003)	Convenience sample of 25 healthy adults (13 male, 12 female, mean age 40.5 years, range 21 – 71 years) able to walk 100 m independently without aids or orthoses	ICCs (3,1) ≥ 0.84 for velocity, cadence, stride length, right and left single support and right and left double support	ICCs (2,1) ≥ 0.75 for velocity, cadence, stride length, left single limb support at preferred speed	Mean differences between CSA® and GAITRite® of 0.006 m/s, 0.18 steps per minute and 0.003 m for velocity, cadence and stride length respectively at neutral walking speed	Reliability tested over three trials/sessions 3 minutes apart. Validity compared to CSA®	1 single trial per session each at slow, preferred and fast speeds, 3 m before and 2m after a 13.3 m GAITRite®
Cutlip et al. (2000)	Convenience sample of 10 healthy adults (4 male, 6 female, mean age 22.1 years, range 21 – 26 years) able to walk independently without aids or orthoses	Not reported	Pearson correlations of $R \geq 0.936$ for step length, step time, stride velocity and stance and swing duration	Mean differences between GAITRite® and video system of 3.8 cm, 0.103 m/s and 0.01 s for step length, velocity and swing duration respectively at neutral walking speed. No difference between step time and stance duration	Validity assessed by agreement with manually digitised video-based system. GAITRite® sampling frequency was 30 Hz. Metronome used to standardised subject-selected fast and slow walking speeds	Participants walked at slow, normal and fast speeds on a 4.6 m GAITRite® starting 4 m before the mat. Type of walking trial, finishing position and number of recorded trials were not reported
Gretz et al. (1998)	20 healthy younger adults (9 male, 11 female, mean age 40 years, range 20 – 56 years) and 21 adults with Down Syndrome (9 male, 12 female, mean age 41 years, range 23 – 51 years)	ICCs (2, *k*) ≥ 0.91for velocity, left and right step length and left and right step time			Reliability tested over 2 sessions 2 weeks apart	2 single trials per session at normal walking speed on a 4.57 m GAITRite®.

Table 3.2. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics
McDonough et al. (2001)	One healthy woman (aged 27 years) and a "stride simulator"	No significant differences ($p \geq 0.05$) between paper and pencil and GAITRite® values for cadence, step length and step time. ICCs (2, 1) ≥ 0.94 for left and right step times	Agreement with paper and pencil revealed ICCs (2, 1) 0.96 for velocity, 0.31 for cadence, 0.97 and 0.99 for right and left step lengths, and 0.67 and 0.61 for right and left step times	Agreement with a video-based system revealed ICCs (2, 1) 0.95 for velocity, 0.96 for cadence, 0.44 and 0.85 for right and left step lengths, and 0.97 and 0.96 for right and left step times	GAITRite® sampling frequency of 30 Hz. Validity assessed by agreement with a video-based system (24 Hz) and paper and pencil method at various speeds (unspecified). Reliability of spatial values was assessed by *t* test comparison with the fixed dimensions of the "stride simulator". Reliability of temporal values was assessed by *t* test comparison with video measurements at 3 speeds (unspecified), and correlation between GAITRite® and video-based measures for step time. Number of sessions and time intervals between sessions was not reported	8 single trials on a 3.6m GAITRite®, one each at very slow, slow, preferred and fast walking speeds and with wide and narrow base of gait and in and out foot angles. Start and finish positions were not reported
Menz et al. (2004)	Convenience sample of 30 healthy adults (12 male, 18 female, mean age 28.5 years, range 22 – 40 years) and 32 community dwelling older adults (13 male, 19 female, mean age 80.8, range 76 – 87)	ICCs (3, 1) ≥ 0.83, for velocity, cadence, left and right step length, left and right base of support and left and right foot angle for the younger participants. ICCs (3, 1) ≥ 0.71 for velocity, cadence, left and right step length and left and right foot angles of older participants	Not reported	No systematic differences between the 2 testing sessions (paired *t*-tests). CVs ≤ 1.9% for the younger participants and ≤ 3.5% for the older participants for velocity, cadence and left and right step length	Reliability tested over 2 sessions approximately 2 weeks apart. Actual time interval not provided. Standardised shoes were provided to participants	3 single walking trials per session at normal velocity, starting and finishing 2 m before and after a 4.6 m GAITRite®

Table 3.2. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics
Nelson et al. (2002)	11 healthy older adults (4 males, 7 females, mean age 70.3) with no known neurological disorders and 11 volunteers with idiopathic Parkinson's disease, stage I-III on the Hoehn and Yahr scale (8 males, 3 females, mean age 74.3)	Not reported	Significant differences (p<0.05) between Parkinson's and non-impaired participants at preferred walking speeds for left and right step length, step time, base of support, single support (%), and double support		Validity assessed using discriminate analysis (method not specified) between Parkinson's and non-impaired participants	3 single walking trials (1 familiarisation, 2 recorded) each at preferred and fast speeds, starting and finishing 3 m before and after a 4.6 m GAITRite®
Rao et al. (2005)	12 adults with Huntington's disease (7 male, 5 female, mean age 50 years, range 34 – 59 years) and 12 aged-matched healthy adults. Gender matching information not provided.	ICCs (3, 2) ≥ 0.86 for velocity, cycle time, stride length, cadence and base of support for participants with Huntington's disease. ICC values not reported for the healthy participants	Not reported	CVs ≤ 10% for velocity, cycle time, stride length, cadence and base of support for participants with Huntington's disease. CV values not reported for the healthy participants. Significant differences (p < 0.005) between the 2 groups existed for all recorded gait variables	Reliability tested over 2 sessions between 30 and 45 minutes apart	2 single walking trials per session at preferred speed, starting and finishing 2 m before and after a 4.6 m GAITRite®
Sorsdahl et al. (2008)	17 children with cerebral palsy (8 male, 10 female, mean age 7.2 years, range 3 – 13 years, GMFCS levels 1 and 2)	ICCs (3, 1) ≥ 0.83 and ICCs (1, 1) ≥ 0.82 for cadence, step length (most and least affected legs), and left and right stride length	Not reported	S_w ≤ 4.0 for cadence, step length (most and least affected legs), left and right stride lengths and left and right step widths	Reliability tested over 2 sessions between 10 and 46 minutes apart (mean 25 minutes)	2 single trials each at slow and fast walking speeds and 4 single trials at normal waking speed. Children started and finished 1.5m before and after a 5.2 m GAITRite®

Table 3.2. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics
Thorpe et al. (2005)	57 healthy children (mean age 6.1, range 1.3 – 10.9 years), stratified into 3 age groups (1 – 4 years, 4 – 8 years, 8 – 11 years)	ICCs (1, 1) between 0.62 and 0.93 for velocity, cadence, step length and stride length for each age group, with the exception of step length (0.40) and stride length (0.41) for the 8 – 11 age group	Not reported	CV ≤ 16.6% for velocity, cadence, step length and stride length. ICCs (1, 1) for base of support, single and double support and foot angle ranged between 0.05 and 0.86	Reliability assess over 2 sessions, both recorded in the same testing session (i.e. 4 walks recorded, compared as 2 testing sessions of 2 trials each)	2 single walking trials per session each at preferred walking speed, starting and finishing 2m before and after a 3.66 m GAITRite®
van Uden and Besser (2004)	21 healthy adults (12 males, 9 females, mean age 34 years, range 19 – 59 years) free from disorders affecting gait	ICCs (2, *k*) ≥ 0.79 for velocity, step and stride length, step and stride time, swing and stance time, base of support, single and double support time and foot angle at preferred and fast walking speeds	Not reported	Differences between the 2 testing sessions are discussed but significance not reported	Reliability tested over 2 sessions 1 week apart	8 single trials per session each at preferred and fast walking speed, starting and finishing 2 m before and after a 6 m GAITRite®
Webster et al. (2005)	10 adults (5 male, 5 female, mean age 66.5 years, range 54 – 83) who had undergone unicompartmental knee surgery	Not reported	For values averaged from one walk, ICCs (2, 1) ≥ 0.92 for velocity, cadence, and left and right step length and time at preferred and fast speeds. For values from individual footfalls, ICCs (2,1) ≥ 0.91 for left and right step length and time at preferred and fast speeds	Small repeatability coefficients were also reported	Validity assessed by agreement with 6 camera infrared motion analysis system with sampling rate of 50 Hz	Four single trials each at preferred and fast walking speeds, starting and finishing 2 m before and after a 8.2 m GAITRite®

Table 3.2. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics
Wittwer et al. (2008)	20 participants with Alzheimer's disease (10 males, 10 females, mean age 80.6 years, range 70 – 91 years) able to walk 100 m independently without aids and able to follow testing instructions	ICC (3, 1) ≥ 0.88 for velocity, cadence, left and right step and stride length, left and right step and stance and swing time, left and right base of support and left and right foot angle	Not reported	CVs ≤ 3.82% velocity, cadence, left and right step and stride lengths and left and right swing and stance times. Similar ICC and CV vales were reported when only the first 3 trials were used (ICC ≥ 0.86 and CVs ≤ 4.20%)	Reliability tested over 2 sessions 1 week apart	10 single trials per session at preferred walking speed, starting on the GAITRite® and finishing 2 m after the mat

Note: CV, coefficient of variation; S_w, within subject standard deviation; ICC, intraclass correlation coefficients; CSA®, clinical stride analyser; GMFCS, gross motor function classification system.

The GAITRite® system used in this investigation consists of an 810 x 89 x 0.625 cm (length x width x height) instrumented mat connected to a personal computer via an interface cable (see Figure 3.2). The walkway contains 12 sensor pads encapsulated in a roll-up carpet with an active area of 720 x 60 cm (length x width). Within this area, there are 27,648 sensors arranged in a 48 x 576 grid. The sensors are 1 cm square in size, and there is a distance of 1.27 cm between sensor centres. This has been shown to provide a spatial accuracy of between 0.51 to 0.66 cm for measures of step and stride length respectively (Selby-Silverstein & Besser, 1999). The system scans the active region at a rate of 80 Hz providing a temporal resolution of 12.5 msec. Basic spatial and temporal gait parameters such as stride length and time are then extracted using specific gait analysis software. Descriptions of each parameter and their method of calculation are described in Chapter 3.6.1 below.

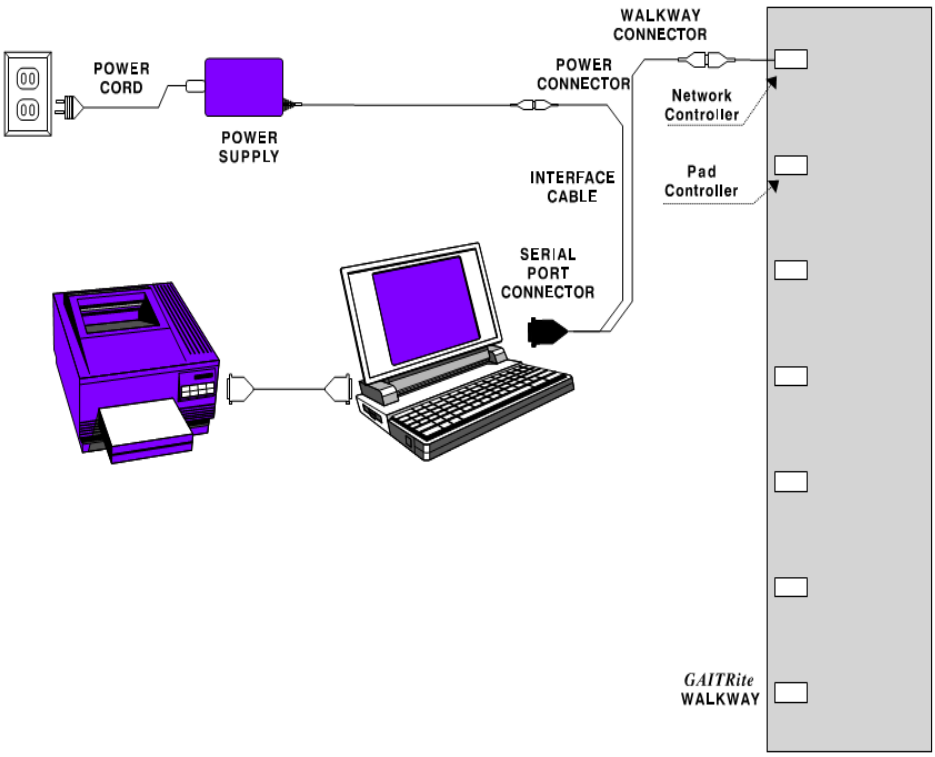
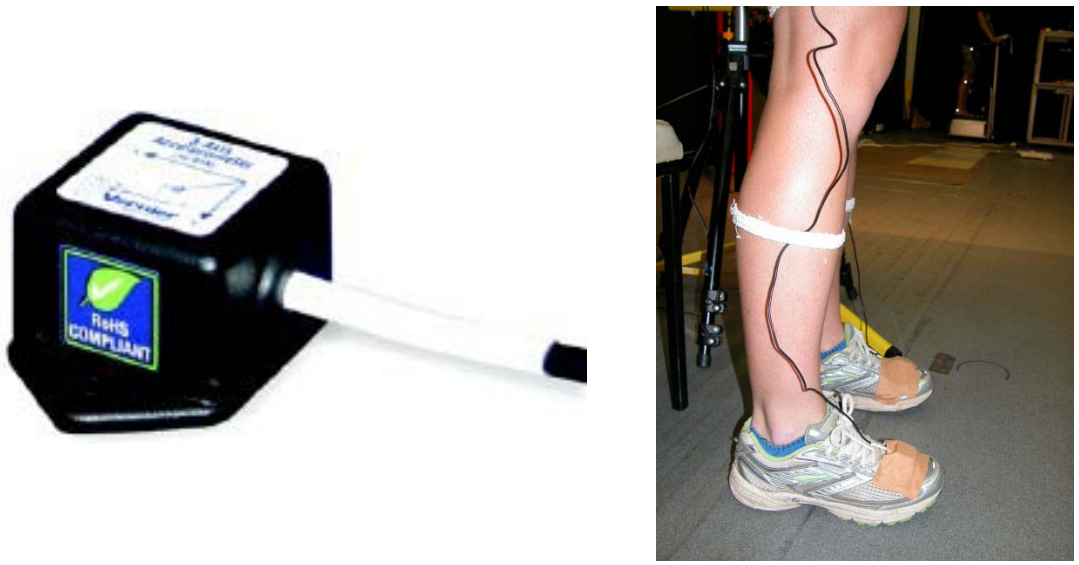


Figure 3.2. The GAITRite® walkway system.

3.3.2 Accelerometers

For study 3, foot accelerations were measured using two CXL10LP3 triaxial Crossbow® accelerometers (Crossbow Technology Inc, San Jose, CA), each with a sensitivity of 200 mV/*g*, and a range of ±10 *g* (see Figure 3.3a). The accelerometers were fixed to the dorsal aspect of each participant's left and right shoes using Elastoplast® rigid strapping tape, in the approximate region overlying the second metatarsal (see Figure 3.3b).



(a) (b)

Figure 3.3. Crossbow® tri-axial accelerometer (a), and fixation to a participant's foot (b).

The accelerometer leads were secured at the distal shank and thigh segments, and the proximal thigh segment, using looped elastic crepe bandages. Lead tension was enough to allow free movement of each joint whilst ensuring the leads did not move. Accelerometer leads were then connected to a Crossbow® AD2012 ReadyDAQ portable data logger that was attached to belt worn around the participant's waist. The belt did not interfere with free movement (see Figure 3.4).

(a) (b)

Figure 3.4. Anterior (a) and posterior (b) view of accelerometer placement and data logger attachment.

The CXL10LP3 is a 68 g capacitive accelerometer containing a silicon micro machined sensor housed in a 2.41 x 5.05 x 3.05 cm aluminium casing (see Figure 3.5). The sensor reacts to the accelerations occurring during movement and generates an electrical signal proportional to these accelerations (Kavanagh & Menz, 2008). To convert this raw voltage into acceleration patterns, the accelerometers were connected to a 226 g Crossbow® AD2012 ReadyDAQ data acquisition system that was attached to a belt worn around the participant's waist. The ReadyDAQ is a 14.7 x 9.2 x 3.3 cm portable data logger and real-time data acquisition system. At the completion of the walking trials, the data logger was connected to a personal computer via an RS-232 interface cable, and the acceleration data was transferred using specific data acquisition software. Based on the limited storage capacity of the data logger (130,000 samples), the sampling rate was set at 125 Hz and acceleration signals were only recorded from the anteroposterior (y) axis of each accelerometer. This permitted the collection of 520 seconds of acceleration signals from two accelerometers at a temporal accuracy of 8 msec, providing consistency with previous work that

has recorded continuous footfall information to retrospectively (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997), and prospectively (Hausdorff, Rios, & Edelberg, 2001) identify community dwelling older adult fallers.

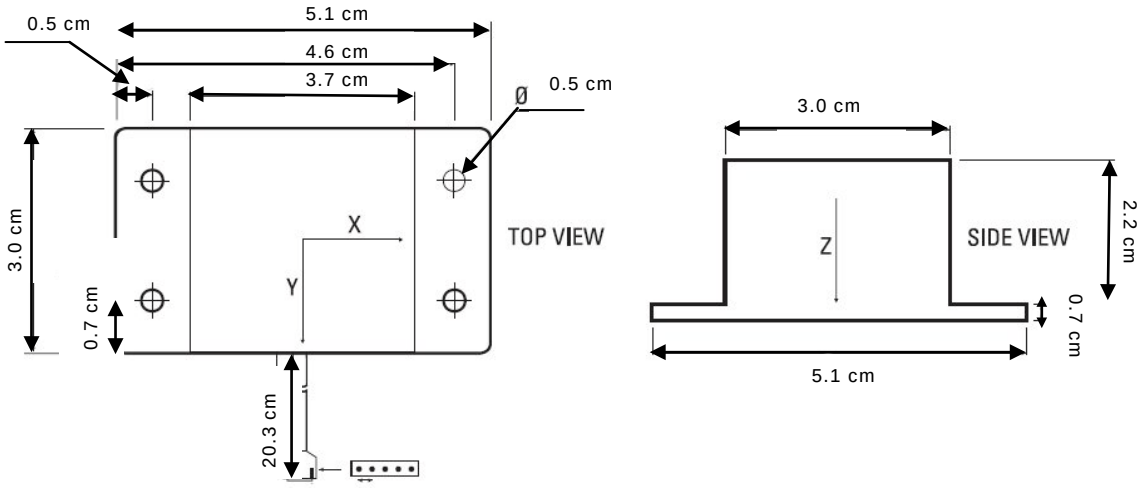


Figure 3.5. Schematic diagram showing dimensions and tri-axial orientations of the Crossbow® accelerometer.

Gait events were determined from the pedal acceleration patterns using a customised software program (see Chapter 3.5.1), and these events were used to calculate temporal locomotor parameters. Previous work has determined temporal parameters from gait cycle events using a single accelerometer fixed to the trunk (Mansfield & Lyons, 2003; Moe-Nilssen & Helbostad, 2004; Zijlstra, 2004; Zijlstra & Hof, 2003), thigh (Aminian, et al., 1999), the dorsum of one foot (Sabatini, Martelloni, Scapellato, & Cavallo, 2005), and the lateral aspect of both lower legs (Selles, Formanoy, Bussmann, Janssens, & Stam, 2005). Excluding activity monitors, reliability has only been investigated for accelerometers attached to the upper body during single trial walking (Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskiod-Samsøe, 2004; Moe-Nilssen, 1998) or using an integrated system of accelerometers such as the IDEEA during single trial walking (Maffiuletti, et al., 2008; Saremi, et al., 2006). Details of these studies are outlined in Table 3.3 below.

Table 3.3. Summary of studies investigating reliability and validity of accelerometry to collect temporal gait data.

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics	Accelerometer characteristics
Hartmann et al. (2009)	23 community dwelling older adults (7 males, 16 females, mean age 77.2 years)	Not reported	ICCs ≥ 0.99 for velocity, cadence, step duration and step length at slow, preferred and fast speeds	RLOA ≤ 3.3% for cadence, step duration and step length at slow, preferred and fast speeds	Validity assessed by agreement with GAITRite®	4 single walking trials each at slow, preferred and fast walking speeds on a 13 m long walkway	75 g DynaPort® MiniMod tri-axial accelerometer (6.4 x 6.2 x 1.4 cm), sampling frequency of 100 Hz, fixed with sports tape at the level of the second sacral vertebra. Data stored locally on a digital memory card
Henriksen et al. (2004)	Convenience sample of 20 healthy adults (6 males, 14 females, mean age 35.2 years, range 18 – 57 years)	ICCs (3, 1) ≥ 0.82 for AP, ML and vertical accelerations, cadence and step and stride length	Not reported	ICCs (1, 1) ≥ 0.82 and CVs ≤ 6.79% for AP, ML and vertical accelerations, cadence and step and stride length	Reliability assessed over 2 sessions 24 hours apart	2 continuous walking trials at slow, preferred and fast walking speeds on a 10 m walkway	30 g piezoresistant Mega® tri-axial accelerometer, sampling at 250 Hz, secured to an elastic belt at the level of the third lumbar vertebra. Data stored on a portable data logger
Kavanagh et al. (2006)	8 healthy males (mean age 23 years)	Velocity ICCs (method not reported) ≥ 0.84 for inter and intra examiner. Shank CMD values ≥ 0.91 for inter and intra examiner reliability of vertical, AP and ML accelerations at preferred speed	Not reported	Slightly lower CMD values for neck and trunk accelerations and at other walking speeds	Inter-examiner reliability assessed by repeating protocol following reattachment of accelerometers by another examiner. Intra-examiner reliability assessed over 2 sessions 24 hours apart	5 single walking trials, each at self selected slow, preferred and fast walking speeds on a 30 m walkway	4 accelerometer nodes, each consisting of 2 biaxial Analog Devices 6g accelerometers (1.5 x 2.6 cm), sapling at 250 Hz. One node was fixed over the occipital pole of the head with a firm elastic headband, whereas the other 3 were fixed at the C7 spinous process (neck), the L3 spinous process (trunk) and 3cm proximal to the lateral malleolus (shank) using rigid sports tape. Each node was connected to a processor box that sent data to a personal computer with a Bluetooth Personal Area Network Device (range 200 m)

Table 3.3. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics	Accelerometer characteristics
Maffiuletti et al. (2008)	A convenient sample of 10 healthy adults (5 male, 5 female, mean age 34 years)	ICCs (3, 1) ≥ 0.961 and CVs ≤ 5.72% for velocity, cadence, left and right single support time, and left and right step and stride length	ICCs (2, 1) ≥ 0.784 for velocity, cadence, left and right single support time, and left and right step and stride length	Significant differences ($p \leq 0.01$) between force platform and accelerometer data for each of the gait parameters was reported	Concurrent validity assessed by agreement with force plates. Reliability assessed intra-session. Participant's walked barefoot	9 – 12 single walking trials (average not reported) at self selected speed. 10 steps minimum per walking trial (4 - 5 steps per trial used for analysis). Walking length not provided	5 biaxial capacitive 2 g IDEEA® accelerometers (1.8 x 1.5 mm), sampling at 32 Hz. One was fixed on the sternum (4 cm below the jugular notch), each thigh (midway between the patellar and anterior superior iliac spine), and each plantar foot (2 cm proximal to the head of the fourth metatarsal). Sensors were connected to a processor box (7.0 x 4.4 x 1.8 cm; 59 g) fixed to the waistband of the participant's clothes
Mayagoiti et al. (2002)	10 male adults (age range 23 – 27 years)	Not reported	CMC values ≤ 0.9363 for knee linear and 0.9861 for shank angular accelerations at all speeds	RMS ≤ 14.8% for knee linear acceleration and 6.3% for shank angular accelerations at all speeds	Concurrent validity assessed by agreement with Vicon® motion analysis system (50 Hz).	2 continuous 10 or 12 second treadmill walking trials, each at five speeds (very slow = 1.4 km/h, slow = 2.1 km/h, average = 2.7 km/h, fast = 3.6 km/h and very fast = 4.6 km/h)	4 pairs of uniaxial accelerometers mounted on 2 aluminium strips (30 x 2 cm), sampling at 100 Hz. Each strip was secured to the frontal, medial aspect of the thigh and shank segments using elasticised velcro® straps
Menz et al. (2003)	10 healthy young adults (3 males, 7 females, age range 22 – 31 years)	ICCs (2, 1) between 0.84 and 0.97 (variables not reported)	Not reported	CV values between 1 and 21% (variables not reported)	Reliability assessed over 2 sessions one week apart	2 single trials at self selected comfortable walking speed. Distance not reported	2 triaxial piezo-resistant accelerometers, one mounted on a foam helmet, the other fixed to a plate firmly strapped at the level of the sacrum using a belt. Accelerometers were connected to a portable laptop carried by the participant in a backpack. Weight of entire apparatus was 2.5 kg

Table 3.3. *Continued.*

Study	Sample	Retest reliability	Concurrent validity	Other results	Comments	Walking trial characteristics	Accelerometer characteristics
Moe-Nilssen (1998)	A convenient sample of 19 healthy adults (4 males, 15 females, mean age 22.9 years, range 21 – 26 years)	ICCs (3, 1) ≥ 0.79 for vertical, AP and ML accelerations on even and uneven ground	Not reported	ICCs (1, 1) ≥ 0.79, S_w ≤ 0.0106 g and CVs ≤ 6.8% for vertical, AP and ML accelerations on even and uneven ground	Reliability assessed over 2 sessions 2 days apart	One single "up and back" walks (2 continuous trials), each at five self selected speeds (slowest to fastest), and each on even and uneven walking surface	Triaxial piezoresistant accelerometer, sampling at 128 Hz, fixed over L3 using a fixation belt. Size and weight of device not provided. Device connected to a portable data logger
Saremi et al. (2006)	Reliability study: 12 healthy adults (7 males, 5 females, mean age 31 years, range 18 – 69 years). Validity study: 8 healthy adults (from the reliability study. Demographics not reported) and 6 adults with stroke (5 males, 1 female, mean age 64 years, range 56 – 70 years)	No significant differences ($p \geq 0.10$) between spatial parameters or over the 2 testing sessions	No significant differences ($p \geq 0.5$) between SAS and accelerometer values (values not reported)	Comparisons between SAS and accelerometry CV values also revealed predominantly non-significant differences ($p \geq 0.3$) with the exception of double limb support ($p = 0.01$)	Reliability assessed over 2 sessions (between session times not reported). Validity assessed by agreement with the SAS	Reliability study: 2 – 3 single walking trials at five speeds ("faster, fastest, usual, slower and slowest") on a 30 m walkway (middle 20 strides used). Validity study: 3 single walking trials on a 15 m walkway. Healthy participants walked at "usual comfortable" and "somewhat slower" speeds and stroke participants walked at "usual comfortable" and "faster but comfortably safe" speeds	5 biaxial capacitive 2g IDEEA® accelerometers, sampling at 32 Hz. One was fixed on the upper chest (4 cm below the top of the sternum), each anterior thigh (midpoint between the knee and anterior superior iliac spine), and each medial forefoot (2 cm below the head of the fourth metatarsal). Sensors were connected to a processor box (60 g) attached to the participant's belt

Abbreviations: AP, anterior posterior , ML, mediolateral; ICC, intraclass correlation coefficients; RLOA, ratio limits of agreement; CMD, coefficient of multiple determination; S_w, within subject standard deviation; SAS, clinical stride analyser system; CMC, coefficient of multiple correlation; RMS, root of the mean squared differences.

3.4 Procedure

Upon entrance into the study, participants attended the Advanced Research Laboratory at the School of Exercise Science, Australian Catholic University, wearing their own comfortable clothing and walking shoes. Comfortable walking shoes were those with a heel less than 2.5 cm, were not dress shoes, and were used by the participant for extended periods of "everyday" walking. For each study, participants first completed a series of screening tests followed by the walking assessment. In addition, for study 3 participants also completed a laterality questionnaire and a series of balance assessments. Each of these is outlined in the following section.

3.4.1 Participant screening

Participants were screened using tests based on studies by Hill (1997), Condron and Hill (2002) and ElHaber et al. (2006), each of which is outlined below. A positive result or impairment on any of the screening tests resulted in either exclusion or was recorded for further analysis. Participant screening took approximately 40 minutes to complete.

3.4.1.1 Self reported medical and surgical history

Participants were asked to record all medical conditions, hospitalisations and surgeries. Self report of any health problems impacting upon balance or mobility, such as stroke or moderately severe arthritis, resulted in exclusion.

3.4.1.2 Medication use

All medication use, including non-prescribed medications such as vitamins and supplements, was recorded. Previous research has identified an increased falls risk with use of medications such as antipsychotics, sedatives and hypnotics (Ensrud, et al., 2002; Mustard & Mayer, 1997; Neutel, Perry, & Maxwell, 2002; Stenbacka, Jansson, Leifman, & Romelsjo, 2002), and decreased falls risk with reduced use of these medications (Campbell, Robertson, Gardner, Norton, & Buchner, 1999). Risk has also been shown to increase with polypharmacy (Leipzig, Cumming, & Tinetti, 1999; Neutel, et al., 2002). Consequently, participants taking one or more

benzodiazepines, or using more than four prescription medications for extended periods were excluded.

3.4.1.3 Self reported fall history

Number of falls in the previous 12 months was recorded. Participants who reported falling in the previous month, or more than twice in the previous year were excluded.

3.4.1.4 Anthropometric data

Height, mass and leg length were recorded for standardisation purposes.

3.4.1.5 Walking pain

Participants were asked if they were currently experiencing, or regularly experienced, pain during walking. Walking pain was recorded, and self report of current pain resulted in exclusion.

3.4.1.6 Activity level

Current level of activity for each participant was recorded as inactive (no exercise), slightly active (exercise one to times per week), active (exercise three to four times per week), or very active (exercise five to seven times per week). Activities qualifying as exercise were "vigorous activities which made them sweat, puff or pant" (Booth, Owen, Bauman, Clavisi, & Leslie, 2000, p. 18). Additionally, a minimum of 30 minutes was also required to classify activity as exercise consistent with Australian Government guidelines (Armstrong, Bauman, & Davies, 2000). Participants who were inactive were excluded.

3.4.1.7 Pulse and blood pressure

Pulse and blood pressure were recorded whilst supine and 1 and 3 minutes following standing. A reduction in systolic blood pressure of 20 mmHg or diastolic blood pressure of 10 mmHg upon standing was considered a sign of orthostatic hypotension and resulted in exclusion (The Consensus Committee of the American Autonomic Society and the American Academy of Neurology, 1996).

3.4.1.8 Vibration sense

Vibration sense was assessed at the medial aspect of the head of the left and right first metatarsal using a 64 Hz Rydel-Seiffer graduated tuning fork. The tuning fork has a nine point arbitrary scale from 0 to 8, with past work showing average vibration threshold values of 5.3/8.0 (Whitton, Johnson, & Lovell, 2005) and 5.9/8.0 (Kästenbauer, Sauseng, Brath, Abrahamian, & Irsigler, 2004) for healthy older adults with no known sensory loss. Participants were seated with eyes closed and the vibrating tuning fork was applied. The participant was required to indicate when they could no longer feel the vibrating stimulus. Based on previously reported 5% lower limit threshold values in healthy controls, left or right vibration threshold values of less than 3.0/8.0 were used to indicate impaired sensation (Kästenbauer, et al., 2004; Martina, van Koningsveld, Schmitz, van der Meche, & van Doorn, 1998). Participants were excluded if they could not sense the stimulus.

3.4.1.9 Lower limb joint proprioception

Lower limb joint proprioception was assessed by movement of the left and right hallux at the first metatarsophalangeal joints (MTPJ). Participants were seated with eyes closed and three trials of dorsiflexion and plantarflexion, presented in random order, were completed at each first MTPJ. Following a demonstration trial, the participant was required to verbally indicate whether the toe was moved upwards (dorsiflexion) or downwards (plantarflexion). One or more incorrect responses was considered suggestive of impaired proprioception (ElHaber, et al., 2006).

3.4.1.10 Visual acuity

A snellen eye chart was used to assess participant's visual acuity in a room illuminated by both natural and artificial lighting. Individuals were permitted to wear their usual glasses for the test if they reported they regularly used corrective vision whilst walking. Participants were required to stand behind a line 6 meters from the chart, cover one eye and read the letters in each line. The lowest letter correctly read was converted to a logmar score and recorded. The test was then repeated using the other eye. Logmar scores of greater than 0.4 were used to indicate impaired

vision (Hill, Schwarz, Flicker, & Carroll, 1999).

3.4.1.11 Visual contrast sensitivity

Visual contrast sensitivity was evaluated using the Melbourne Edge Test (MET) (Verbaken & Johnston, 1986) in a room illuminated by both natural and artificial lighting. The MET is a portable chart consisting of four rows of five circles, with each circle containing a contrasting edge randomly presented in one of four possible orientations (vertical, horizontal, 45° to the left or 45° to the right; see Figure 3.6). As the test progresses, the line of contrast edge decreases from high to low contrast, making edge detection increasingly difficult. A response key card is used to cover the chart and expose only a single circle as the participant proceeds. From a distance of 40 cm, participants indicated which of the four edge orientations they saw in the exposed circle. If correct, the key card was shifted to expose the next circle and the participant continued until they were unable to distinguish the contrasting edge. The last correct response was recorded, and a MET score of less than 16 was used to indicate impaired contrast sensitivity (Verbaken & Johnston, 1986).

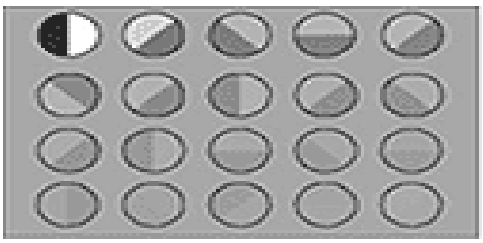


Figure 3.6. The Melbourne Edge Test.

3.4.1.12 Vestibular function

To evaluate vestibular function, each participant completed the vestibular stepping test in their own footwear as described by Peitersen (1967). Briefly, participants stood facing forwards on a 152 x 102 cm (L x W) mat, in a starting position 50 cm from the rear border. The mat was

marked with lines at 10˚ increments out from this starting position (zero degrees and zero cm) to the edge of the mat for a full 360˚ (see Figure 3.7). On each of these lines, 10 cm intervals were marked. On command, the participant commenced stepping on the spot with their eyes closed. At the completion of 50 steps the participant was told to stop, and the distance from the starting position and degree and direction of rotation was recorded. Rotation of greater than 60˚, forward displacement of greater than 100 cm or lateral displacement of greater than 25 cm was considered a positive result (Peitersen, 1967).

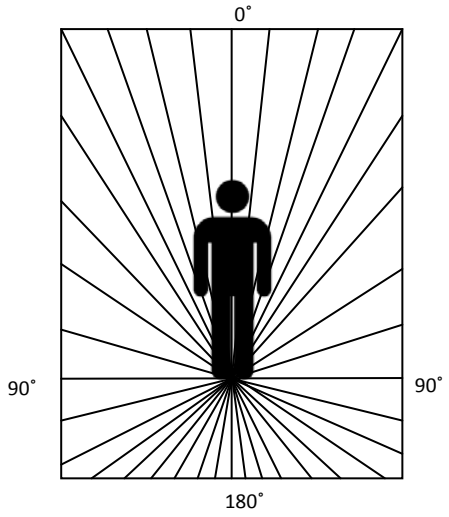


Figure 3.7. The vestibular stepping test.

3.4.1.13 Cognitive function

The Mini Mental State Examination (MMSE) was used to evaluate cognitive function (Folstein, Folstein, & McHugh, 1975). The MMSE is an 11 item 30 point questionnaire assessing functions such as arithmetic, reading, memory, and comprehension. Values greater than 26 indicate normal cognitive function, whereas values between 20 and 26 indicate mild cognitive impairment, between 10 and 19 indicate moderate to severe impairment and below 10 indicate severe impairment (Crum, Anthony, Bassett, & Folstein, 1993). Participants scoring below 26 were classified as having impaired cognition and those below 24 were excluded.

3.4.1.14 Romberg's test

Romberg's test is described in Chapter 3.4.3.4 below. Participants stepping or opening their eyes before 30 seconds were classified as having a positive Romberg's result in studies 1 and 2. Scoring for study 3 is outlined in Chapter 3.4.3.4 below.

3.4.2 Laterality assessment

3.4.2.1 Lateral preference inventory

The lateral preference inventory (LPI) was used to determine the hand and foot lateral preference of each participant in study 3. The LPI has four questions for each of these regions. A point is added for each "right" preference response and subtracted for each "left" preference response. Consequently, for each region, lateral preference may range from +4 (each of the four actions is performed on the right side) to -4 (each of the four actions is performed on the left side). The inventory has been shown to be both reliable and valid (Coren, 1993; Coren, Porac, & Duncan, 1979). An overall score of between 1 and 4 was considered as a right sided preference.

3.4.3 Balance assessments

Postural stability, or balance control, is a complex task requiring steady state, anticipatory and/or reactive adjustments, employed whilst the individual is stationary or moving (Shumway-Cook & Woollacott, 2007). Due to the inherent multidimensional nature of balance therefore, it is difficult to evaluate the construct using a single test. Consequently, the following section describes five tools that were used to evaluate components of each participant's static and dynamic balance abilities for study 3. The scores from four of these tests were then aggregated to provide an overall single balance score, the Balance Outcome Measure for Elder Rehabilitation (BOOMER) (Haines, et al., 2007). All balance tests were completed in the participant's own comfortable shoes, as described in Chapter 3.4.

3.4.3.1 Step test

The step test is a reliable (ICCs between 0.91 and 0.94) method of determining single-limb dynamic balance in healthy older adults (Hill, Berhardt, McGann, Maltese, & Berkovits, 1996). The test is significantly correlated ($p \leq 0.001$) with the functional reach test (r=0.68 and 0.73), walking speed (r=0.83) and stride length (r=0.82 and 0.83), and has also been shown to discriminate between healthy older adults and stroke patients ($p \leq 0.001$) (Hill, 1997).

Participants stood 5cm in front of a step (H x L x W = 7.5 x 58.5 x 40.5 cm) with feet shoulder width apart and arms by their side. The experimenter stood to the side of participant to provide support in case of overbalancing. Upon command, participants were instructed to place their foot completely on the top of the step and back on to the ground again as many times as possible in 15 seconds. The contralateral foot remained fixed to the ground throughout the trial. The test was performed using each foot separately, presented in random order, and the average number of steps completed in each of the two trials was recorded. If a participant overbalanced during a test, the person was steadied by the instructor and the number of steps before overbalancing was recorded. The result from the leg that performed the least number of steps ("worst leg") was used in subsequent analyses.

3.4.3.2 Timed up and go

The timed up and go test (Podsiadlo & Richardson, 1991) was used as another method of evaluating dynamic balance ability. The test has high inter-rater and intra-rater reliability (ICCs between 0.97 and 0.99) for healthy older adults (Podsiadlo & Richardson, 1991; Shumay-Cook, Brauer, & Woollacott, 2000; Steffen, Hacker, & Mollinger, 2002). It also has established validity, with strong correlations to the Berg Balance Scale (r=-0.81), walking speed (r=-0.61) and the Barthel Index (r=-0.78) (Podsiadlo & Richardson, 1991), and is able identify older adult fallers with sensitivity and specificity of 87% (Shumay-Cook, et al., 2000).

On a "go" command, participants stood up from a chair (seat height = 43 cm), walked 3

meters, turned around a marker, and returned to their original position and sat down again. Timing commenced by pressing a stopwatch on the "go" command and stopped when the participant was seated again with their back fully against the back rest of the chair. Time was recorded in seconds.

3.4.3.3 Functional reach

The functional reach test (Duncan, Weiner, Chandler, & Studenski, 1990; Podsiadlo & Richardson, 1991) was used to evaluate dynamic bilateral stance balance ability of older women. The test has excellent test-retest (ICC=0.92) and intra-rater reliability (ICC=0.98), and performance correlates well (r=0.71) to centre of pressure excursion in older males (Duncan, et al., 1990). Past work has also shown the test is able to predict multiple fallers, with older adults who scored zero (i.e. unable to reach any distance) eight times more likely to experience two or more falls in the subsequent six months (Duncan, Studenski, Chandler, & Prescott, 1992).

Participant stood in an upright position parallel to a wall, with their arms outstretched to 90° of shoulder flexion and with feet in a comfortable stance width. From this starting position the participant reached forward as far as safely possible whilst keeping their feet in place. The examiner stood next to the participant and if they lost their balance or stepped, they were steadied and instructed to begin again. The distance between the starting position and the maximum reach position was measured in centimeters by the examiner using a tape measure mounted at shoulder height on a wall next to the participant. Given there is high correlation (r=0.90) between the first trial and the average of three (Billek-Sawhney & Gay, 2005), only one trial was performed.

3.4.3.4 Romberg's test

Romberg's test is a measure of static balance ability under reduced sensory feedback conditions, and is commonly used to assess dorsal spinal column integrity in suspected ataxic patients (Khasnis & Gokula, 2003). A version of the test, also referred to as "timed static stance

with eyes closed", is a component of the clinical test of sensory interaction and balance (CTSIB) and was the version used in this study. The test has high test-retest and intra-tester reliability (*r*=0.99) (Cohen, Blatchly, & Gombash, 1993), and values have been shown to be predicted by age and correlated with scores on the Get Up and Go Test in community dwelling fallers (*r*=-0.67) and non-fallers (*r*=-0.44) (Anacker, Di Fabio, & Horak, 1992). Abnormal scores on the CTSIB have also been shown to increase falls risk (odd ratio = 8.67) in community dwelling older adults (Di Fabio & Anacker, 1996).

The test was completed as originally described by Anaker et al. (1992). Participants stood with their feet together and on command were instructed to close their eyes. Timing was commenced when the participant's eyes were closed and was stopped after 30 seconds, or if the participant stepped, opened their eyes or otherwise lost their balance. Three trials were completed, and the total time for each trial was summed and recorded. However, if a participant reached 30 seconds on their first trial, the other two trials were automatically scored as 30 seconds. The average of the three trials was recorded in seconds.

3.4.3.5 Sharpened Romberg

Due to a possible ceiling effect for healthy older adults performing Romberg's test (Cohen, et al., 1993; Morris, Iansek, Smithson, & Huxham, 2000), static balance was also assessed using the Sharpened Romberg's test (Graybiel & Fregly, 1966). The test is performed in the same manner as Romberg's test, however one foot (whichever the participant prefers) is placed heel to toe in front of the other. Timing was commenced when the participant's eyes were closed and was stopped after five seconds, or if the participant stepped, opened their eyes or otherwise lost their balance. Time was recorded in seconds and only one trial time was performed.

The Sharpened Romberg test has high intra-rater (ICCs between 0.95 and 0.99) and inter rater reliability (ICCs between 0.73 and 0.93, *r*=0.99) (Briggs, Gossman, Birch, Drews, & Shaddeau, 1989; Franchignoni, Tesio, Martino, & Ricupero, 1998). Using the test, significant differences

(p<0.05) have been found between healthy younger and older women (Wiksten, Perrin, Hartman, Gieck, & Weltman, 1996), fallers and non-fallers (Heitman, Gossman, Shaddeau, & Jackson, 1989) and between 60-64 and 75-79 year old women (Briggs, et al., 1989).

3.4.3.6 Balance Outcome Measure for Elder Rehabilitation

Whilst the above measures of static and dynamic balance provide useful information regarding specific dimensional components of balance, a collective global measure of balance ability may be preferable for comparison across different clinical and research environments. Such a tool may also have the advantage of enabling comparison between adults of differing balance abilities. Consequently, scores on four of the five balance tasks were aggregated to provide a global outcome score, the Balance Outcome Measure for Elder Rehabilitation (BOOMER) (Haines, et al., 2007). Each balance item was scaled as described by Haines et al, and these scores were summed giving the BOOMER score as shown in Table 3.4.

Table 3.4. Scoring criteria for static and dynamic balance assessments to provide BOOMER score. Adapted from (Haines, et al., 2007). See text for explanation of balance tests.

Test	0	1	2	3	4
Step test (average number of steps)	Unable	>0-5	>5-8	>8-12	>12
Timed up and go (sec)	Unable	≤30	<30-20	<20-10	<10
Functional reach test (cm)	0	>0-15	>15-20	>20-30	>30
Romberg's test (sec)	Unable	>0-30	>30-60	>60-<90	90

3.4.4 Walking assessments

Participants completed two walking protocols in their own comfortable footwear. The first protocol involved repeated single walking trials and the second protocol involved continuous laps of a walking circuit. The protocol order was randomised for each participant. Besser et al. (1999) has shown that a minimum of eight gait cycles, collected from single walking trials of at least two

to four gait cycles per trial, are required to provide stable spatiotemporal measures of gait. Thus, for studies 1 and 2, 10 walks of three to five gait cycles per trial were recorded for each protocol. For study 3, the equivalent of seven minutes of walking was recorded, consistent with previous research that has collected continuous walking data using older adult participants (Hausdorff, et al., 1997; Hausdorff, et al., 2001). Two familiarisation trials were performed before data collection for each walking protocol.

3.4.4.1 Single walking trials

The single trial protocol required participants to walk at self-selected speed along a flat walkway containing the 8.1 meter instrumented GAITRite® mat (see Figure 3.8). Research in healthy and impaired young (Miff, Childress, Gard, Meier, & Hansen, 2005) and frail older adults (Lindemann, et al., 2008) has shown a distance of two steps or 2.5 meters, respectively, are needed to achieve steady-state gait. Thus, participants commenced and finished a minimum of two body lengths from the start and end of the walkway, indicated by a marker, ensuring steady state walking across the GAITRite®. At the beginning of each trial, a verbal command was given to commence walking. At the completion of the trial, data were saved, and the software was primed for the next trial. The participant waited at the alternate end and commenced walking again on verbal instruction from the experimenter. This between-trial period took approximately 15 seconds.

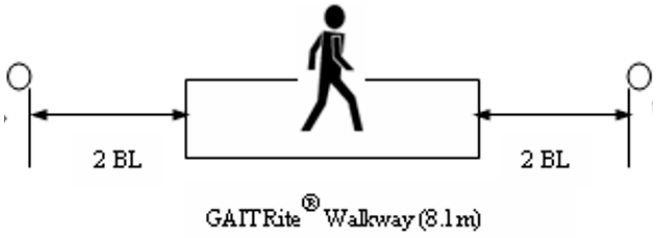


Figure 3.8. Single walk protocol walkway. BL, body length.

3.4.4.2 Continuous walking trials

The continuous walking protocol consisted of a curvilinear circuit that incorporated two straight sections that were the same length as the single trial walking condition (see Figure 3.9). The straight sections were distanced 3 metres apart laterally and separated by markers. The GAITRite® mat was positioned along one of the straight sections. On a "go" command, participants commenced walking the circuit at self selected speed, with each lap taking approximately 20 seconds to complete. The GAITRite® system was re-set when each participant was on the opposite side of the circuit. As outlined in Chapter 3.4.4 above, participants completed the same number of passes on the GAITRite® for each walking protocol ensuring the equivalent numbers of steps were recorded in each of the three studies.

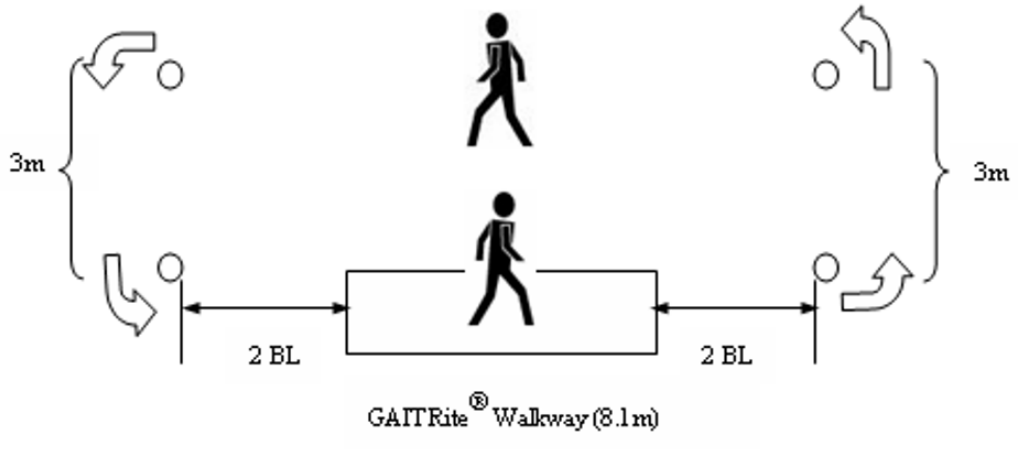


Figure 3.9. Continuous walk circuit. BL, body length.

3.4.5 Falls assessment

In study 3, all participants were provided with a falls pack to collect falls data based on the methods of Hill et al. (1999). The pack included a falls calendar (see appendix 3), a falls questionnaire (see appendix 4), and stamp self-addressed envelopes. This was used to determine prospectively whether participants experienced a fall in the subsequent 12 month period following testing.

Participants were asked to tick each day on the calendar that no fall occurred or cross the day if they did have a fall. Additionally, if a fall occurred, they were also required to complete the short fall-related questionnaire querying how and where the fall occurred and if any injuries were experienced. At the end of each month, participants were required to remove the monthly sheet from the calendar, enclose it in the stamped self-addressed envelope along with any completed fall questionnaires and mail it to the principal investigator (KP). If a calendar sheet was not received by a particular participant, the individual was contacted and reminded to send in their monthly sheet.

3.5 Data analysis

Individual left and right spatial and temporal footfall data were extracted from the GAITRite® application software (version 3.8), and left and right temporal data were derived from the Crossbow® accelerometer application software (DataReady, version 6.2) (Chapter 3.3.2 and 3.5.1). Left and right data were not pooled because some studies have shown spatiotemporal gait data to be asymmetric (Sadeghi, Allard, Prince, & Labelle, 2000), which may affect the variability of the data. In addition, pooling left and right data may create a bimodal distribution, affecting normality. Data filtering is outlined in Chapter 6.2.3.

3.5.1 Determination of stride time from accelerometer data

To determine stride time, raw antero-posterior (AP) voltage signals were transferred from the portable data logger and imported into a custom designed software program written in Igor Pro (Wave Metrics, Version 6). The program initially calibrated the signal and converted the voltage into raw AP accelerations. A “near zero” acceleration threshold was then established to use as the threshold from which positive or negative acceleration spikes arose during walking. This zero threshold was determined by searching for the most common voltage level within specific parameter values that were able to be modified by the user. User modification to the zero threshold parameter values was necessary as minor zero drift was observed in some data series.

Next, toe-off events were identified by firstly locating extended periods of zero accelerations (i.e. foot flat) of approximately 0.02 to 0.2 seconds in length. To account for individual differences in walking speed, these time windows were also altered by the user. As toe-off was typically found to be noisy, the program then searched for a maximal negative spike surrounded by smaller positive ones at the end of this zero acceleration period. The highest negative acceleration was taken as toe-off, and temporal distances between these acceleration peaks were used to calculate stride time (Figure 3.10). In some instances, toe-off was particularly noisy and there was difficulty identifying gait events. Consequently, each data file was visually inspected for accuracy, and where identification was problematic, manual identification of toe-off was made using mouse cursor position. Left and right stride times were then saved as text files.

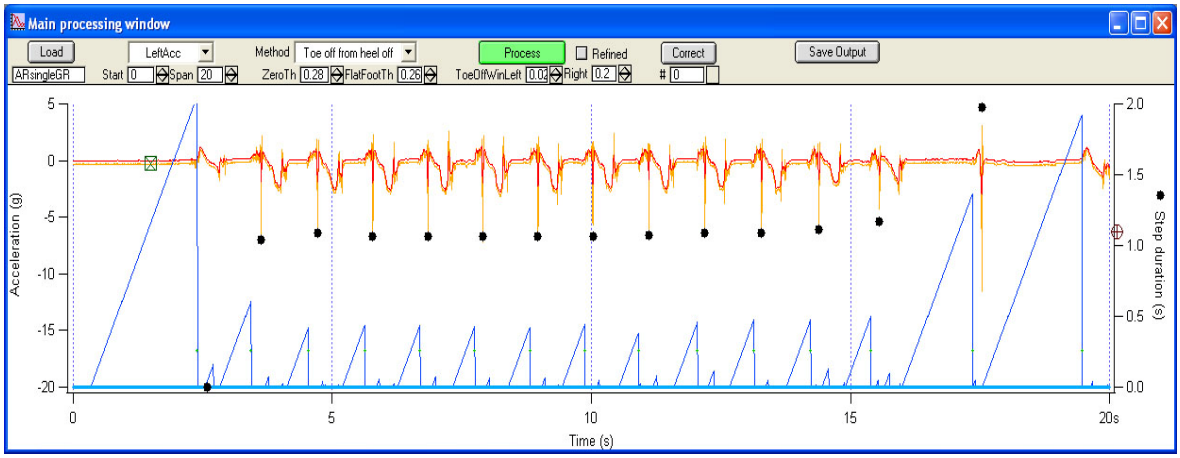


Figure 3.10. Identification of toe-off gait event (black dots) via identification of negative acceleration peak.

Post-processing of the accelerometer stride times was completed with additional custom software (Igor Pro Version 6). For the continuous protocol files, left and right data were imported and the series was plotted (Figure 3.11). A visual inspection of the series was made and minimum and maximum stride time values were set. Extreme values outside of this range were found to be initiation and termination steps and thus were deleted. The file was then saved and an accumulated continuous walking protocol file was created in which the participant's code, their processed stride time data, and a left/right designation were contained.

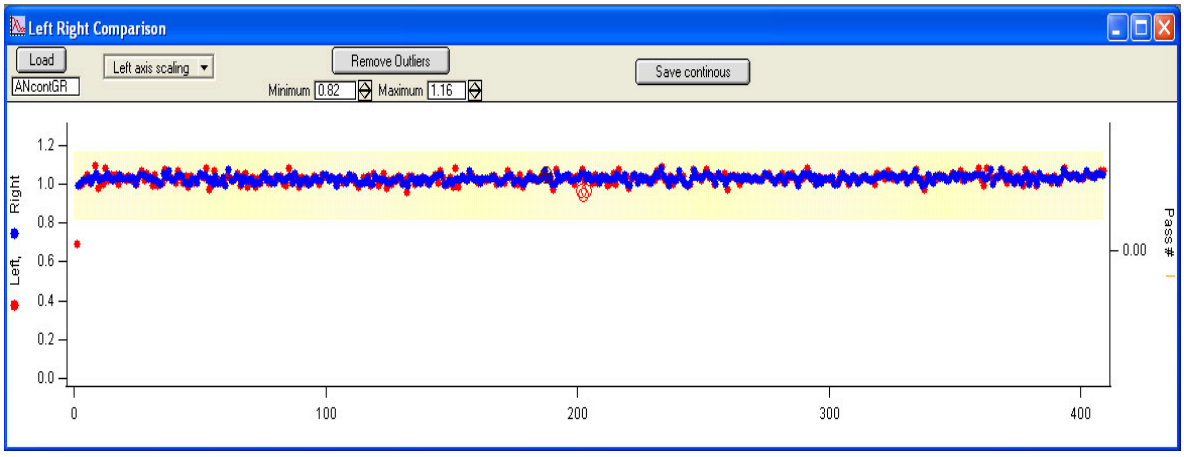


Figure 3.11. Continuous walking trial stride times. Note initial outlier at start of data file on extreme left.

For the single walking trial files, left and right data were found by specifying a maximum and minimum range based on a visual inspection of a plot of the data series and deleting any outliers (Figure 3.12). These outermost values were also caused by the first and last steps at the start and end of each of the walking trials, as well as by small movements by the participant while they were waiting in subsequent trials. To identify the appropriate start and end data points of each trial, the program searched for clusters of data points that were followed by pauses longer than 1.2 seconds. Each of these was identified as a trial and an incremental plot was made over the data series showing each walking trial (Figure 3.12).

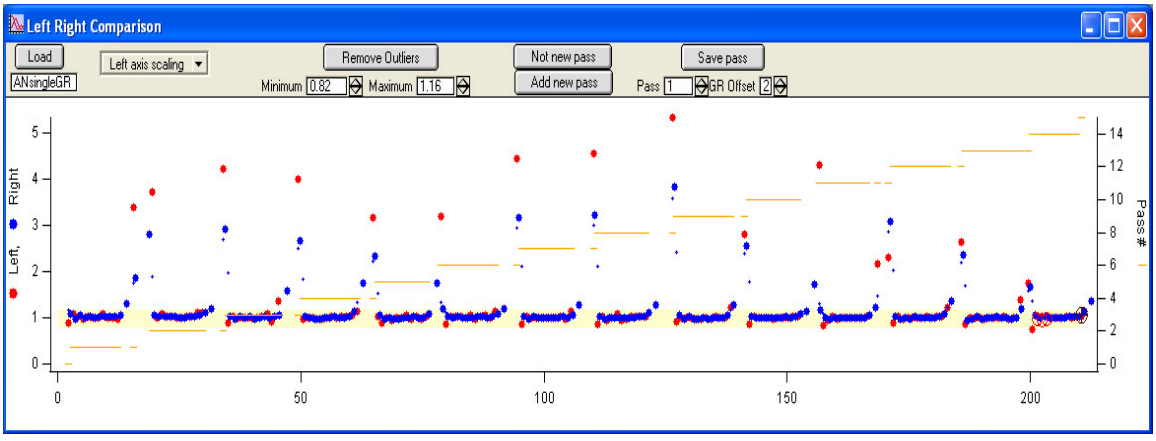


Figure 3.12. Repeated single walking trial stride times. Note the increase in outliers due to the repetitive initiation, acceleration, deceleration and termination phases. Stepped solid lines indicate incremental walking trials (passes).

To validate the data, the corresponding GAITRite® stride time data was imported and overlaid on the accelerometer data. Each trial ("pass") was visually inspected and an offset value was added to the GAITRite® data series in order to align it with the corresponding accelerometer data (Figure 3.13). A good fit between data points from the two series indicated agreement between the two systems. When a best fit between the two series was made, the data was saved, creating an accumulated single trial walking protocol listing the participant's code, the processed accelerometer and corresponding GAITRite® stride time data, left/right designation, and the trial number.

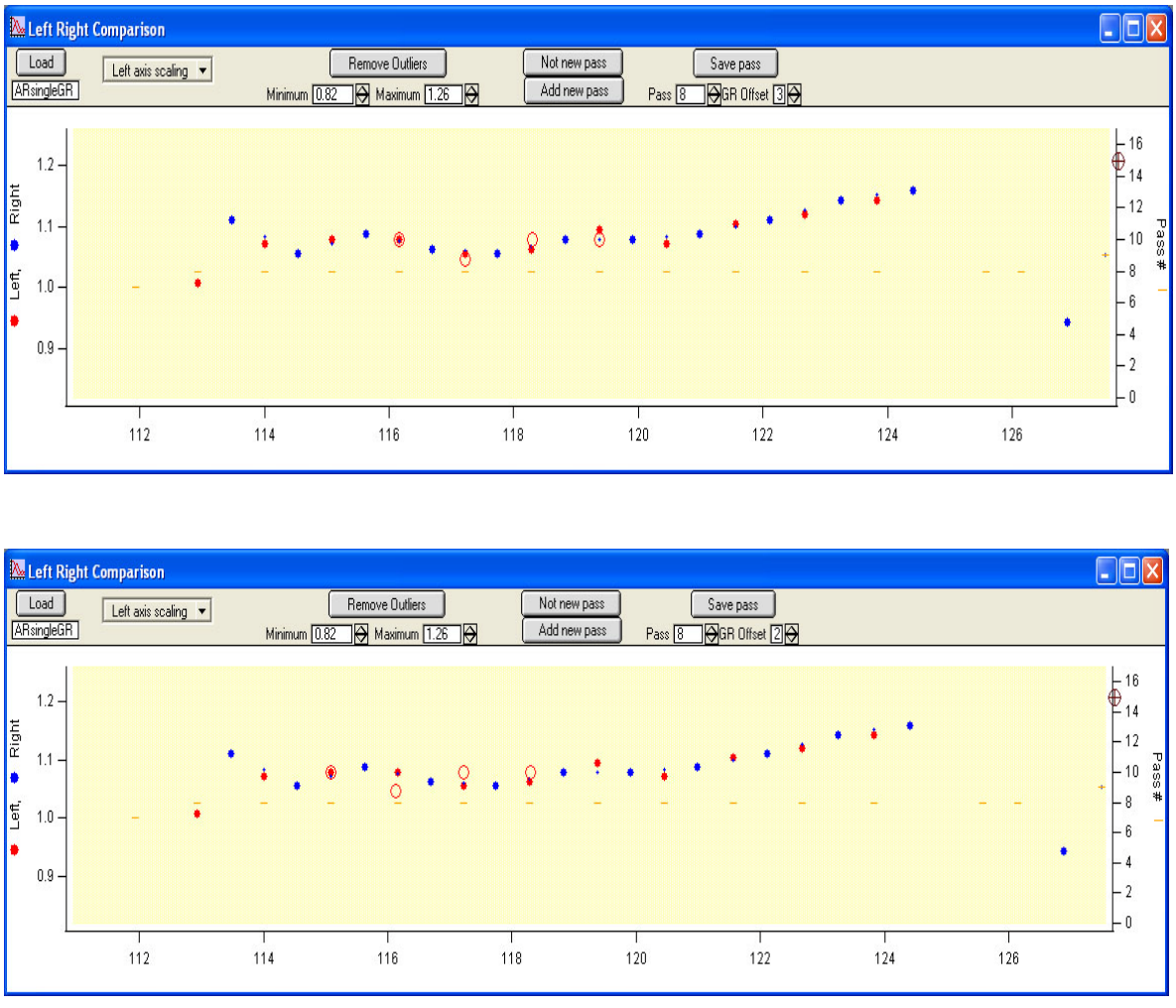


Figure 3.13. Matching of accelerometer data (filled circles) with GAITRite® data (open circles). The top diagram shows a good fit between the two data files for the loaded walking trial using an offset value of three, whereas the bottom diagram shows a poor fit of the data using an offset value of two.

3.6 Statistical analysis

Mean values for each of the spatial and temporal gait variables (see Chapter 3.6.1 below for description). To quantify gait variability in each study, standard deviation (SD) and coefficient of variation (CV) were calculated for each of the gait parameters (see Chapter 3.6.1 below for description). Additionally, to quantify stride dynamics in study 3, the fractal scaling index of stride time was also calculated and is described in Chapter 6.2.3. Unless stated, all statistical analyses were conducted with the Statistical Package for the Social Sciences (SPSS) version 17. Specific statistical analyses for each study are described in Chapters 4 to 6.

3.6.1 Dependent variables

The dependent variables under investigation were the spatial and temporal gait parameters and their associated variability, and are listed and described below in Table 3.5. Additionally, for study 3 stride dynamics was also used as a dependent variable (see Chapter 6.2.3). Spatial parameters collected were step length (cm), stride length (cm), foot angle (°) and base of support (cm), whereas temporal parameters were walking velocity (m/s), step time (sec), stride time (sec), stance time (sec) and swing time (sec). To quantify gait variability, standard deviation (SD) and coefficient of variation (CV) were calculated for each of these gait parameters. The units of milliseconds (msec) were used for the SD values of the temporal gait parameters and the units of centimetres per second (cm/s) were used for the SD values of velocity. These spatiotemporal parameters and variability statistics are commonly reported in studies of gait variability and falls (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005; Brach, Studenski, Perera, VanSwearingen, & Newman, 2008; Hausdorff, et al., 1997; Maki, 1997; Owings & Grabiner, 2004b). Each spatial and temporal gait variable is defined in the following chapter based on descriptions provided by the GAITRite® operating manual ("GAITRite Operating Manual," CIR Systems Inc.). Note that these dependent variables were not used in each of the studies. For specific dependent variables used in each study, see Chapters 4 to 6.

Table 3.5. Description of dependent variable from studies 1 to 3.

Dependent variable	Description
Step length (cm)	Horizontal distance from one heel centre to the heel centre of the following footfall on the opposite foot.
Stride length (cm)	Horizontal distance from heel centre of one foot to the heel centre of the same foot's next footfall.
Foot angle (°)	Angle between the line of progression and a bisection of the foot.
Base of support (cm)	Perpendicular distance from heel centre of one foot to the line of progression of the opposite foot.
Walking velocity (m/s or cm/s)	Distance divided by time.
Step time (sec or msec)	Time between first contact of one footfall to the first contact of the next opposite footfall.
Stride time (sec or msec)	Time between the first contact of one footfall and first contact of the next footfall from the same foot.
Stance time (sec or msec)	Time between the first and last contact of one foot.
Swing time (sec or msec)	Time between last contact of one foot and the following first contact of the same foot.
Standard deviation	A variability measure calculating the distribution of individual scores around the mean value.
Coefficient of variation	An absolute and dimensionless measure of variability calculated by dividing the standard deviation by the mean and multiplying by 100.

3.6.1.1 Step length

Step length (cm) was calculated by determining the horizontal distance from the geometric heel centre of one footfall to the geometric heel centre of the following footfall on the opposite foot (i.e. left foot to the next right foot or vice versa). In Figure 3.14, the length of line AX is the step length of the right foot and the length of line YG is the step length of the left foot.

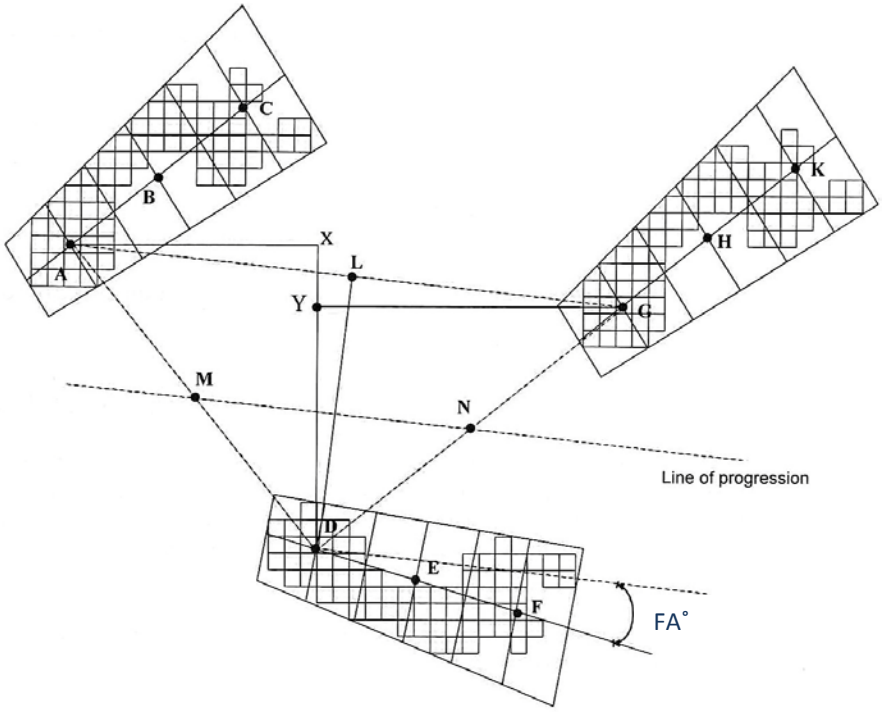


Figure 3.14. Schematic diagram showing the calculation of spatial measurements. Adapted from the GAITRite® Operating Manual (GAITRite Operating Manual, CIR Systems Inc.). FA, foot angle. See text for explanation of gait measurement determination.

3.6.1.2 Stride length

Stride length (cm) was calculated by determining the horizontal distance from the geometric heel centre of one footfall to the geometric heel centre of the next footfall of the same foot (i.e. left foot to the next left foot). In Figure 3.14, the length AG is the stride length of the left foot.

3.6.1.3 Foot angle

Foot angle (˚) was calculated by the angle formed between the line of progression and a line bisecting the geometric heel and forefoot centres. Positive values indicated toe out and negative values indicated toe in. In Figure 3.14, the right foot angle is shown by the angle FA˚.

3.6.1.4 Base of support

The base of support (cm), sometimes termed the step or stride width, was calculated by determining the perpendicular distance from the geometric heel centre of one foot to the line of progression of the following foot. In Figure 3.14, the line DL is the base of support for the right foot.

3.6.1.5 Walking velocity

Velocity (m/s) was calculated by dividing the horizontal distance between the first and last geometric heel centres recorded on the GAITRite® by the time taken to walk this distance. For the SD of velocity, the units of cm/s were used.

3.6.1.6 Step time

Step time (sec) was calculated as the time between the first contact of one footfall to the first contact of the next opposite footfall (e.g. left foot contact to the next right foot contact). For step time SD, the units of milliseconds (msec) were used. Left step time is displayed in Figure 3.15 below.

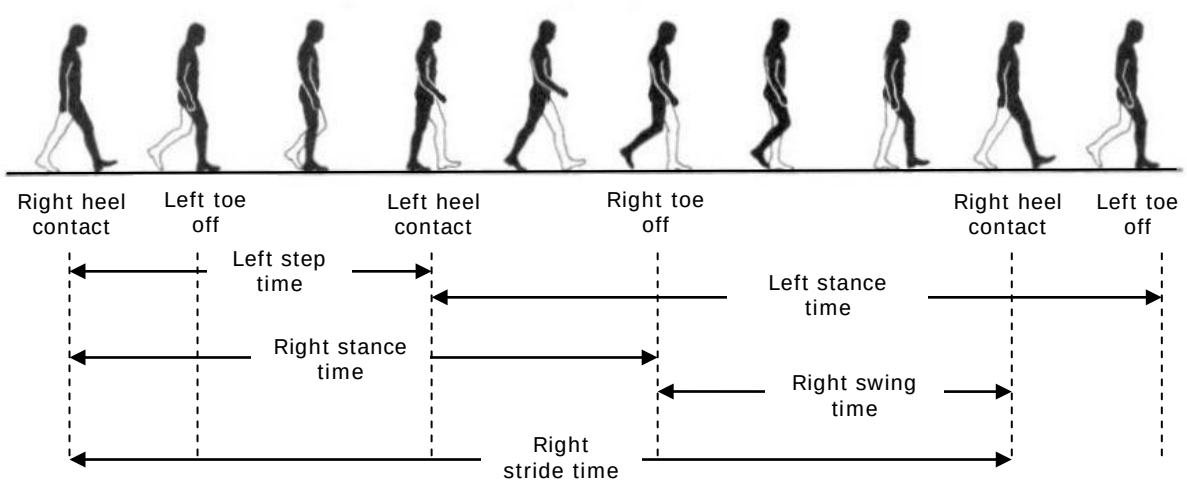


Figure 3.15. Temporal gait variables. Adapted from Inman, Ralston and Todd (1981, p. 26).

3.6.1.7 Stride time

Stride time (sec) was calculated as the time between the first contact of one footfall to the first contact of the next footfall of the same foot (e.g. the time between two consecutive left foot contacts). For stride time SD, the units of milliseconds (msec) were used. Right stride time is displayed in Figure 3.15 above.

3.6.1.8 Stance time

Stance time (sec) was calculated as the time between the first and last contact of one foot. For stance time SD, the units of milliseconds (msec) were used. Left and right stance times are shown in Figure 3.15 above.

3.6.1.9 Swing time

Swing time (sec) was calculated as the time between the last contact of one foot and the following first contact of the same foot and is equivalent to single support time. For swing time SD, the units of milliseconds (msec) were used. Right swing time is shown in Figure 3.15 above.

3.6.1.10 Standard deviation

The standard deviation (SD) is a relative measure of variability expressed in the units of the parameter being studied. It provides an indication of the distribution of values around the mean, and was calculated using the following formula:

$$SD = \sqrt{\frac{\sum\left(X - \overline{X}^2\right)}{N - 1}} \qquad \text{Equation 3.1}$$

3.6.1.11 Coefficient of variation

The coefficient of variation (CV) is an absolute and dimensionless measure of variability therefore enabling comparison between gait parameters with different units. It was calculated as follows:

$$CV = \left(\frac{SD}{\overline{X}}\right)100\% \qquad \text{Equation 3.2}$$

CV values closer to zero indicate reduced variability, thus the statistic is only applicable to ratio data (data bounded by zero) (Atkinson & Nevill, 1998). Consequently, CVs were not calculated for the parameters of step width and foot angle (interval measurements).

3.6.2 Independent variables

For studies 1 and 2, the independent variables were the walking protocol and the age group of participants. For study 3, the independent variables were the walking protocol and prospective fall status of participants. Each of these are described below.

3.6.2.1 Walking protocol

To investigate the effect of walking protocol on gait data (studies 1 and 2) and the evaluation of falls risk (study 3), each participant completed the single and continuous walking protocols, presented in random order, as described in Chapter 3.4.4.1 and 3.4.4.2.

3.6.2.2 Participant age

To investigate age-related differences in gait data, two participant age groups were studied. The young women participants were aged between 18 and 35 and met the inclusion criteria described in Chapter 3.2.2.1. The older women participants were aged between 55 and 90 and met the inclusion criteria described in Chapter 3.2.2.2. Studies 1 and 2 included both age groups whereas study 3 only included the older women.

3.6.2.3 Future falls

To investigate differences in the gait of fallers and non-fallers, and to explore walking-related falls risk factors, the older cohort of women included in study 3 were categorised as non-fallers (women who did not experience a fall in the subsequent 12 months) or fallers (women who experienced one or more falls in the subsequent 12 months), as recorded by the 12 month falls calendar described in Chapter 3.4.5. Additionally, further analyses also stratified the sample into non-fallers, single fallers (women who experienced one fall in the subsequent 12 months), multiple fallers (women who experienced two or more falls in the subsequent 12 months) and a combined group of single and non-fallers.

3.6.3 Statistical assumptions

Data were first assessed for normality by using skewness and kurtosis statistics and were considered normal if the value fell between $-2 \geq z \leq 2$ (Altman & Bland, 1996). Further normality tests are described in Chapters 4 to 6. Additionally, specific statistical assumptions for multivariate testing conducted in study 3 are described in Chapter 6.

To test for the presence of heteroscedasticity in study 1, the average test-retest score was plotted against the absolute test-retest difference, and the zero-order correlation coefficient between the values was calculated (Bland & Altman, 1986). Heteroscedasticity was considered present if the parameters with larger mean values (e.g. velocity compared with step time) displayed greater mean differences between the two sessions when compared with parameters with smaller mean values. This was characterized by a significant positive correlation coefficient (Atkinson & Nevill, 1998; Henriksen, et al., 2004).

3.6.4 Statistical power

Group sample size for study 1 was based on estimates provided by Walter, Eliasziw and Donner (1998) for use in reliability studies. Using published intra-class correlation coefficient ranges of between 0.5 and 0.9 for spatial and temporal gait data (Menz, et al., 2004; van Uden & Besser, 2004), a sample size of 8.8 for each group was required to detect differences at a significance level of 0.05 and with statistical power of 80%. Consequently, to be conservative, 13 younger and 14 older participants were recruited in study 1.

The required sample size for study 2 was estimated using between-group velocity mean (158 cm/s) and SD (19 cm/s) values recorded from study 1. Spatial parameters such as step length generated lower sample size values than velocity, and values from temporal parameters such as step time and stance time were highly variable, thus these were not used in the estimation. Based on the velocity values, a significance level of 0.05 and statistical power of 80%, 26 participants were required for each group. Therefore, 28 younger and 32 older participants were

recruited in study 2.

Sample size estimation for study 3 was made using previously reported between-group mean difference (0.20 m/s) and standard deviation (0.29 m/s) velocity values of fallers and non-fallers (Hausdorff, et al., 2001). Other spatial or temporal gait parameters were not used in the sample size estimation for the reasons outlined above. Using these velocity values, a sample size of 26 was required for each sample to detect differences between groups at a significance level of 0.05 and with 80% power. However, given the likelihood of participant drop-out due to the prospective nature of the study, and the need to include at least 26 fallers within the sample to ensure statistical power, a total of 108 participants were recruited for study 3.

4 Study 1

The test-retest reliability of spatiotemporal gait data for young and older women during continuous over-ground walking

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4.1 Introduction

Falls are a major cause of morbidity and hospitalisation in the elderly population. For example, approximately one in three people over the age of 65 fall each year (Dolinis, Harrison, & Andrews, 1997; Lord, Ward, Williams, & Anstey, 1993), and 54% of hospitalisations in this age group are the result of a fall (Cripps & Carman, 2001). In 2001, the medical cost of fall-related injuries was approximately $AUD 498 million, and the figure is expected to triple by 2051 (Moller, 2003). Research has shown that 50% to 60% of falls occur during locomotion (Bradley & Harrison, 2007). This suggests that age-related changes in dynamic balance and gait may be a major risk factor. Consequently, to identify early markers of gait decline, studies into alterations in the walking pattern of older adults have been the focus of much research in the physical sciences.

Many of these studies have shown that the mean values of a number of gait parameters are altered with increasing age [e.g. (Menz, Latt, Tiedemann, Kwan, & Lord, 2004; Stolze, Friedrich, Steinauer, & Vieregge, 2000)] and are further changed in older fallers [e.g. (Besser, et al., 2000; Montero-Odasso, et al., 2005)]. The most commonly cited age-related changes include reduced walking speed (Prince, Corriveau, Hébert, & Winter, 1997) and a shorter stride length (Watelain, Barbier, Allard, Thevenon, & Angue, 2000). These changes are traditionally assessed by a protocol involving repeated single walks along a fixed length walkway. This protocol has established reliability and validity for a variety of populations including healthy young (Bilney, Morris, & Webster, 2003; Menz, Latt, Tiedemann, Kwan, & Lord, 2004; van Uden & Besser, 2004) and older adults (Menz, et al., 2004), and people with Huntington's disease (Rao, Quinn, & Marder, 2005), cerebral palsy (Sorsdahl, Moe-Nilssen, & Strand, 2008), Parkinson's disease (Nelson, et al., 2002) and Down Syndrome (Gretz, et al., 1998).

It has been suggested that using independently sampled strides from repeated single walks may neglect the continuous nature of everyday walking (Dingwell, Cusumano, Cavanagh, & Sternad, 2001). Indeed, research has found that previous strides can influence multiple future strides during continuous walking (Hausdorff, et al., 1996). That is, fluctuations in stride time variability at one point in time were shown to be related to multiple further fluctuations in healthy adult gait. Using a continuous walking protocol, researchers have demonstrated that these long range correlations in gait variability are significantly reduced with age (Hausdorff, et al., 1997) and in people who fall (Hausdorff, Rios, & Edelberg, 2001; Herman, Giladi, Gurevich, & Hausdorff, 2005), providing a possible marker of age-related declines in gait.

Accordingly, many studies now employ a continuous treadmill or over-ground walking protocol to collect gait data. However, a common criticism of treadmill walking is that gait parameters, including gait variability, differ between treadmill and over-ground walking (Marsh, et al., 2006; Owings & Grabiner, 2004b). For example, research has shown that stride time variability, and the variability of lower limb kinematics, is significantly reduced (p<0.05) in treadmill compared to over-ground walking (Dingwell, et al., 2001). Additionally, treadmill walking has been reported to require excessive familiarization times for older people, with many still using a handrail and showing changing reliability and absolute difference scores for gait parameters after 14 minutes of treadmill walking (Wass, Taylor, & Matsas, 2005). Differences between walking parameters recorded during treadmill compared to over-ground walking were also still apparent at this time.

Therefore, due to the alterations in gait parameters, and excessive familiarization times with treadmill walking, a continuous over-ground walking protocol may be a more optimal data collection method for investigating gait changes with age. To date, no study has examined the test-retest reliability of a continuous over-ground walking protocol. Given that repeated single walks are the most common method used to collect gait data, it is also important to examine the

reliability of spatiotemporal gait parameters recorded during both single and continuous walking protocols.

4.1.1 Aims and Hypotheses

The aims of this investigation were:

- To determine the test-retest reliability of gait data captured during repeated single and continuous over-ground walking.
- To examine the systematic bias in gait data collected during a continuous over-ground walking protocol and a repeated single over-ground protocol.

The hypotheses were:

- Spatiotemporal gait data collected during single and continuous walking trials will be reliable.
- Greater systematic bias will be found during spatiotemporal gait data collected during the single walking trials than during continuous gait trials.

4.2 Methods

4.2.1 Participants

Thirteen young and 14 older women volunteered to participate in the study. Group characteristics are displayed in Table 4.1. The wider age-range for the older sample reflects the ages commonly investigated for gait changes in the literature. This higher range was not expected to influence the outcomes of the study given a within subject design has been employed to investigate the reliability of two different walking protocols. Detailed information regarding participant recruitment and inclusion criteria are outlined in Chapter 3.2.

Table 4.1. Characteristics of younger and older participants (Mean [± SD, range]).

	Younger (n = 13)	Older (n = 14)
Age (years)	20.08 (0.76, 19-21)	67.93 (7.77, 57-79)
Height (m)	1.64 (0.08, 1.48-1.83)	1.62 (0.06, 1.54-1.73)
Mass (kg)	62.18 (8.67, 42.70-73.00)	67.96 (14.39, 41.10-90.60)

4.2.2 Procedure

To ensure suitability to the study aims and testing protocol, participants were first screened according to the procedures outlined in Chapter 3.4.1. Following screening, participants then completed two test sessions seven days apart (median ± SD, 7±1.58 days). Due to availability, one participant completed their second session three days later and another 14 days later. Within each testing session a single and a continuous walking protocol was completed, presented in a random order, as described in Chapter 3.4.4. Spatiotemporal gait data was collected using an 8.1m GAITRite® walkway as discussed in Chapter 3.3.1. Consequently, the screening and testing procedures were completed as follows:

- self reported medical and surgical history
- medication use
- self reported fall history
- height, weight and leg length
- pulse and blood pressure supine and one and three minutes after standing
- vibration sense using a Rydel-Seiffer graduated tuning fork
- lower limb joint proprioception via movement of the left and right hallux
- visual acuity using snellen eye chart

- visual contrast sensitivity using the Melbourne Edge Test
- vestibular function using the vestibular stepping test
- cognitive function using the Mini Mental State Examination
- single and continuous walking trials using a GAITRite® walkway
- single and continuous walking trials repeated following seven days

4.2.3 Statistical Analyses

Individual left and right spatial and temporal footfall data were extracted and statistical assumptions were assessed as described in Chapter 3.5 and 3.6. The following gait mean spatial and temporal gait parameters from Chapter 3.6.1 were used as dependent variables:

- step length (cm)
- foot angle (°)
- step width (cm)
- walking velocity (m/s)
- step time (sec)
- stance time (sec)
- swing time (sec)

To investigate systematic bias across sessions, paired *t* tests ($p<0.05$) were conducted. A significant *t* test result indicates a statistical difference between the two sessions. As no change in performance was expected this would suggest the presence of systematic bias in the data. Bonferroni corrections were not performed in the *t* test analysis because the analysis of each

variable in its own right was of interest, not a generalized null hypothesis of systematic bias in gait (Perneger, 1998).

To assess relative reliability, intraclass correlation coefficients of type (3,1) ($ICC_{3,1}$)were calculated (Portney & Watkins, 2000). Values closer to one demonstrate higher reliability. Given that ICC values may be affected by a heterogeneous sample (Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskiod-Samsøe, 2004), two absolute measures of reliability were also calculated. The first, the standard error of measurement (SEM), was determined using the standard deviation (SD) of subject values and the $ICC_{3,1}$ reliability coefficient for the two testing sessions (Portney & Watkins, 2000; Weir, 2005):

$$SEM = SD\sqrt{(1 - ICC_{3,1})} \qquad \text{Equation 4.1}$$

The SEM expresses the error in the units of the measured variable, and is applicable only to data which is homoscedastic (Atkinson & Nevill, 1998). That is, there is the potential for larger scores to have an SEM which underestimates the amount of error, while smaller scores may have errors over-estimated by the statistic. Thus, while the SEM was still reported for parameters displaying heteroscedasticity, these should be interpreted with caution.

The coefficient of variation (CV) was also used to assess absolute reliability. It is a dimensionless measure unaffected by heteroscedasticity, and is calculated by dividing the SEM by the mean, expressed as a percentage (Hopkins, 2000):

$$CV = \left(\frac{SEM}{\overline{X}}\right) 100\% \qquad \text{Equation 4.2}$$

Coefficient of variation values closer to zero indicate better reliability, thus the statistic is only applicable to ratio data (data bounded by zero) (Atkinson & Nevill, 1998). Consequently, CVs were not calculated for the parameters of step width and foot angle (interval measurements). All statistical analyses were conducted with SPSS (version 12).

4.3 Results

4.3.1 Participant characteristics

Six younger adults reported a medical condition, only one reported taking medication and the sample was active a median of three times per week. The younger sample also had good vision (median visual acuity 0.03, median contrast sensitivity 22.61), and with the exception of three participants with mildly impaired proprioception, also had excellent sensory and balance function. Older adults were healthy, with a median of three past or current medical conditions, and were taking a median number of two medications. The older sample was also active (median of three to four activity session per week), and the majority had good cognitive, sensory and balance function (Table 4.2).

Table 4.2. Medical profile and screening test results for older adults.

Parameter	
Median number of medical conditions (range)	3 (1 to 9)
Most common medical problems (% of sample)	
Hypertension	25.00
Osteoarthritis	15.63
Hyperlipidemia	15.63
Reflux	12.50
Hysterectomy	9.38
Breast cancer	9.38
Median number of medications (range)	2 (0 to 8)
Most common medications (% of sample)	
Antihypertensive agents	28.13
Hypolipidemic agents	15.63
Hyperacidity, reflux and ulcers agents	12.50
Anti-coagulant, anti-thrombotic	9.38
Minerals	6.25
Analgesic - simple and antipyretics	6.25

Table 4.2. *Continued*.

Parameter	
Mean number of falls in previous 12 months (± SD, [range])	0.27 ± 0.59 (0 to 2)
Physical measures	
Pain during walking (% of sample)	3
Median activity level (times per week)	3 to 4
Mean MMSE (± SD)	28.67 ± 2.39
Median VA (logmar, median [range])	0.08 (0 to 0.6)
Mean contrast sensitivity (± SD)	18.2
Impaired vibration (%)	3
Impaired proprioception (%)	15.63
Positive vestibular stepping test (%)	6.25
Positive Romberg (%)	12.5

Note: NSAIDs, non-steroidal anti inflammatory drugs; MMSE, mini mental state examination; VA, visual acuity.

4.3.2 Systematic bias across test sessions

All data exhibited normality because skewness and kurtosis values fell within $-2 \geq z \leq 2$ (Altman & Bland, 1996). Mean values for days one and two, mean difference values and associated standard deviation and 95% confidence interval values are presented below for younger (Table 4.3) and older (Table 4.4) adults. These values were similar to previously reported values from other studies on younger and older women (Hageman & Blanke, 1986; Stolze, et al., 2000) and between-day testing sessions (Bilney, et al., 2003; Menz, et al., 2004; van Uden & Besser, 2004).

In the single walking condition, systematic bias (p<0.05) was found in both groups of women for the measures of velocity, step length, step time, and stance time (see Table 4.3 and Table 4.4). The older adults also exhibited systematic bias (p<0.05) in swing time. In the continuous walking condition, systematic bias (p<0.05) was only found in the younger group for the measures of step time (right limb), swing time (left limb), and stance time (both limbs). Overall,

more systematic bias across the test sessions was found for the single walking trial condition. For example, 7 (54%) of 13 variables showed systematic bias in the younger group, and 10 (77%) of 13 variables showed systematic bias in the older group. In the continuous walking condition, only 5 (38%) of 13 variables showed systematic bias in the younger group, whereas none of the variables showed systematic bias in the older group. For the group as a whole, 17 (65%) of 26 gait parameters showed systematic bias during single trial walking, whereas only 5 (19%) of 26 exhibited systematic bias for the continuous trials.

Table 4.3. Mean (SD) for days one and two, mean difference (SD) and confidence intervals of each variable for the younger participants.

	Single walking trials				Continuous walking trials			
	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals
Velocity (m/s)	1.58 (0.19)	1.63 (0.18)	0.05 (0.11)*	-0.16 to 0. 25	1.51 (0.13)	1.52 (0.17)	0.01 (0.10)	-0.17 to 0.20
Step length (L) (cm)	78.66 (7.56)	79.97 (8.08)	1.31 (3.32)*	-4.20 to 6.81	76.98 (6.80)	77.47 (7.74)	0.49 (3.67)	-3.96 to 4.85
Step length (R) (cm)	78.64 (7.44)	80.13 (7.83)	1.49 (3.20)*	-4.16 to 7.13	77.34 (6.90)	77.72 (7.74)	0.38 (3.25)	-4.72 to 5.43
Step time (L) (sec)	0.502 (0.034)	0.500 (0.035)	-0.002 (0.018)*	-0.039 to 0.030	0.511 (0.032)	0.512 (0.037)	0.001 (0.018)	-0.033 to 0.034
Step time (R) (sec)	0.502 (0.039)	0.496 (0.035)	-0.006 (0.019)*	-0.044 to 0.031	0.510 (0.036)	0.507 (0.038)	-0.003 (0.016)*	-0.034 to 0.028
Step width (L) (cm)	8.81 (2.17)	8.66 (2.13)	-0.15 (1.53)	-3.18 to 2.82	8.45 (2.22)	8.22 (2.14)	-0.23 (1.61)	-3.44 to 2.89
Step width (R) (cm)	8.71 (2.09)	8.68 (2.20)	-0.03 (1.52)	-3.04 to 2.90	8.38 (2.07)	8.22 (2.07)	-0.16 (1.61)	-3.33 to 2.98
Swing time (L) (sec)	0.401 (0.021)	0.402 (0.022)	0.001 (0.013)	-0.025 to 0.026	0.407 (0.023)	0.409 (0.022)	0.002 (0.014)*	-0.025 to 0.030
Swing time (R) (sec)	0.398 (0.024)	0.397 (0.024)	-0.001 (0.015)	-0.031 to 0.027	0.402 (0.023)	0.403 (0.023)	0.002 (0.014)	-0.027 to 0.030

Table 4.3. C*ontinued.*

	Single walking trials				Continuous walking trials			
	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals
Stance time (L) (sec)	0.600 (0.055)	0.590 (0.053)	-0.010 (0.028)*	-0.068 to 0.043	0.612 (0.048)	0.607 (0.054)	-0.005 (0.022)*	-0.050 to 0.039
Stance time (R) (sec)	0.604 (0.054)	0.597 (0.054)	-0.007 (0.028)*	-0.065 to 0.045	0.618 (0.049)	0.613 (0.054)	-0.005 (0.024)*	-0.052 to 0.043
Foot angle (L) (°)	0.35 (4.50)	0.25 (4.66)	-0.13 (2.25)	-4.52 to 4.26	-0.09 (4.80)	-0.15 (4.57)	-0.06 (2.31)	-4.54 to 4.48
Foot angle (R) (°)	3.04 (4.37)	3.10 (4.42)	0.06 (1.86)	-3.58 to 3.71	3.61 (4.06)	3.15 (4.04)	-0.46 (2.13)*	-4.61 to 3.76

Note: L, left side; NA, not applicable; R, right side; * Indicates significant difference between days one and two at the 0.05 level.

Table 4.4. Mean (S.D) for days one and two, mean difference (S.D.) and confidence intervals of each variable for the older participants.

	Single walking trials				Continuous walking trials			
	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals
Velocity (m/s)	1.37 (0.22)	1.41 (0.24)	0.04 (0.09)*	-0.15 to 0.22	1.36 (0.21)	1.37 (0.25)	0.01 (0.09)	-0.16 to 0.18
Step length (L) (cm)	68.91 (7.32)	69.63 (8.32)	0.72 (2.77)*	-4.75 to 6.14	68.42 (7.39)	68.77 (8.18)	0.35 (2.54)	-4.33 to 5.05
Step length (R) (cm)	69.17 (6.50)	70.06 (7.34)	0.89 (2.57)*	-4.17 to 5.91	68.80 (6.51)	68.95 (7.02)	0.15 (2.89)	-4.51 to 4.81
Step time (L) (sec)	0.510 (0.049)	0.503 (0.047)	-0.007 (0.026)*	-0.058 to 0.043	0.511 (0.042)	0.512 (0.053)	0.001 (0.022)	-0.046 to 0.049
Step time (R) (sec)	0.505 (0.048)	0.498 (0.045)	-0.007 (0.025)*	-0.056 to 0.043	0.504 (0.044)	0.505 (0.054)	0.001 (0.026)	-0.053 to 0.054
Step width (L) (cm)	7.90 (1.85)	7.69 (1.77)	-0.21 (1.52)	-3.20 to 2.76	8.00 (2.01)	7.98 (1.70)	-0.02 (1.65)	-3.05 to 2.97
Step width (R) (cm)	7.76 (1.83)	7.55 (1.81)	-0.21 (1.39)	-2.94 to 2.52	7.95 (1.95)	7.82 (1.70)	-0.13 (1.48)	-2.94 to 2.66
Swing time (L) (sec)	0.386 (0.024)	0.382 (0.024)	-0.004 (0.018)*	-0.040 to 0.032	0.385 (0.023)	0.384 (0.027)	-0.001 (0.018)	-0.036 to 0.034
Swing time (R) (sec)	0.382 (0.023)	0.378 (0.023)	-0.004 (0.017)*	-0.038 to 0.030	0.379 (0.025)	0.379 (0.032)	0.000 (0.021)	-0.043 to 0.043

Table 4.4. *Continued*.

	Single walking trials				Continuous walking trials			
	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals	Mean (SD) day 1	Mean (SD) day 2	Mean Difference (SD)	95% Confidence Intervals
Stance time (L) (sec)	0.630 (0.084)	0.620 (0.077)	-0.010 (0.035)*	-0.079 to 0.060	0.630 (0.075)	0.633 (0.086)	0.003 (0.036)	-0.068 to 0.073
Stance time (R) (sec)	0.634 (0.084)	0.624 (0.080)	-0.010 (0.033)*	-0.076 to 0.056	0.636 (0.074)	0.638 (0.086)	0.002 (0.032)	-0.061 to 0.063
Foot angle (L) (°)	3.43 (4.14)	3.44 (4.39)	0.01 (2.42)	-4.73 to 4.74	3.50 (4.18)	3.83 (4.55)	0.33 (2.39)	-4.37 to 4.78
Foot angle (R) (°)	5.16 (3.86)	5.56 (4.19)	0.40 (2.26)*	-4.06 to 4.82	5.34 (4.03)	5.50 (4.53)	0.16 (2.24)	-4.21 to 4.54

Note: L, left side; NA, not applicable; R, right side; * Indicates significant difference between days one and two at the 0.05 level.

4.3.3 Test-retest reliability of single and continuous walking trials

The SEM, CV, and ICCs were used to examine the reliability of each of the gait variables (see Table 4.5). In the single walking trial condition, heteroscedasticity was found in 2 (15%) of 13 gait variables for the younger participants and 8 (62%) of 13 variables for the older participants. In the continuous trials, 8 (62%) of 13 gait variables exhibited heteroscedasticity for the older participants, whereas no heteroscedasticity was observed for the younger participants. For the sample as a whole, heteroscedasticity was found in 10 (38%) of 26 gait variables for the single walking trial condition and 8 (31%) of 26 gait variables for the continuous condition. The majority of the heteroscedastic values were found for the parameters of velocity, step length, step time, swing time, and stance time.

Measures of absolute reliability (SEM, CV) were good for each of the gait parameters. For example, step length and velocity SEMs ranged from 1.50cm to 2.04cm and 0.06m/s to 0.08m/s, respectively. CVs ranged from 2.06% to 4.77%, with velocity exhibiting the highest CVs (4.48–4.77%) and step length exhibiting the lowest coefficients of variation (2.06%–2.84%). No trend for differences in reliability between the single and continuous walking protocols was observed for any of the gait parameters.

The highest ICC values were found for velocity, step length, step time, stance time, and foot angle (0.81–0.95), whereas swing time ICC values were slightly lower (0.70 –0.82). Consistent with past research (Menz, et al., 2004), the lowest ICC values were found for step width (0.66 – 0.75). No systematic changes in ICC values were observed between the single and continuous walking protocols.

Table 4.5. Standard error or measurement (SEM), coefficient of variation (CV), and intraclass correlation coefficient (ICC) values for single and continuous walking trials for younger and older females.

	Younger						Older					
	Single walking trials			Continuous walking trials			Single walking trials			Continuous walking trials		
	SEM	CV	ICC	SEM	CV	ICC	SEM	CV	ICC	SEM	CV	ICC
Velocity (m/s)	0.08	4.68	0.85	0.07	4.50	0.81	0.07†	4.77	0.92	0.06†	4.48	0.93
Step length (L) (cm)	1.99†	2.50	0.94	1.59	2.06	0.95	1.96†	2.84	0.94	1.69	2.47	0.95
Step length (R) (cm)	2.04†	2.56	0.93	1.83	2.36	0.94	1.82†	2.61	0.93	1.68	2.44	0.94
Step time (L) (sec)	0.012	2.50	0.87	0.012	2.43	0.86	0.018†	3.56	0.87	0.017†	3.34	0.87
Step time (R) (sec)	0.013	2.71	0.87	0.011	2.21	0.90	0.018†	3.56	0.86	0.019†	3.78	0.86
Step width (L) (cm)	1.08	NA	0.74	1.14	NA	0.74	1.07	NA	0.66	1.09	NA	0.66
Step width (R) (cm)	1.07	NA	0.75	1.14	NA	0.71	0.99	NA	0.71	1.01†	NA	0.70
Swing time (L) (sec)	0.009	2.33	0.82	0.010	2.43	0.80	0.013	3.39	0.70	0.013†	3.30	0.74
Swing time (R) (sec)	0.010	2.63	0.80	0.010	2.53	0.82	0.012†	3.21	0.73	0.015†	4.08	0.70

Table 4.5. *Continued*.

	Younger						Older					
	Single walking trials			Continuous walking trials			Single walking trials			Continuous walking trials		
	SEM	CV	ICC	SEM	CV	ICC	SEM	CV	ICC	SEM	CV	ICC
Stance time (L) (sec)	0.020	3.40	0.86	0.016	2.60	0.90	0.025†	3.97	0.91	0.025†	4.02	0.90
Stance time (R) (sec)	0.020	3.31	0.87	0.017	2.76	0.89	0.024†	3.77	0.92	0.022†	3.51	0.92
Foot angle (L) (°)	1.59	NA	0.88	1.63	NA	0.88	1.71	NA	0.84	1.69	NA	0.85
Foot angle (R) (°)	1.32	NA	0.91	1.51	NA	0.86	1.60	NA	0.84	1.58	NA	0.86

Note: L, left side; NA, not applicable; R, right side; † indicates data is heteroscedastic.

4.4 Discussion

This study investigated the test-retest reliability of spatiotemporal gait data recorded during steady-state walking using single and continuous walking trial protocols. Gait parameter values obtained were similar to previously reported values in older women (Hageman & Blanke, 1986; Stolze, et al., 2000) suggesting the representativeness of the sample. In general, the spatiotemporal gait data were found to be reliable for both conditions in young and older participants. Interestingly, there was also some evidence of systematic bias across the test sessions for many of the spatiotemporal variables, with the majority of the bias occurring for the single walking trial condition. This suggests that a continuous over-ground walking protocol may be a better method to assess gait changes over time because this protocol showed less systematic bias, particularly for older women.

4.4.1 Systematic bias across test sessions

This investigation found some evidence for the presence of systematic bias in the gait data recorded during both walking protocols. That is, significant differences between some test-retest mean values were found for both walking protocols. Similar findings of bias have been previously reported in both pathological (Evans, Goldie, & Hill, 1997; Hill, Goldie, Baker, & Greenwood, 1994) and healthy younger populations, although other studies have found no difference in healthy older adults (Menz, et al., 2004). Interestingly, more bias was found in the single trial condition, particularly for the older women. It must be acknowledged however that the magnitude of these mean differences or biases across the sessions was generally small. This raises an important issue in the clinical assessment of gait; that is, what magnitude of change in gait data truly indicates gait improvement or decline as opposed to gait fluctuations or familiarisation (van Uden & Besser, 2004).

Clinicians require a method to determine whether small but statistically significant differences in values across test sessions represent genuine changes in gait. To determine clinical relevance,

some researchers have suggested the use of confidence limits for the mean difference value (Evans, et al., 1997; Hill, et al., 1994; Roebroeck, Harlaar, & Lankhorst, 1993; Weir, 2005). The lower and upper 95% confidence limits for the difference score incorporate both systematic and random errors, and provide minimal detectable difference values (Roebroeck, et al., 1993; Weir, 2005). For a change in gait performance between repeated tests to be considered real (i.e., exceeding the systematic and random errors inherent in the test), it needs to fall outside these bounds.

For example, Table 4.3 and Table 4.4 shows statistically significant bias (p<0.05), for velocity changes of 0.05 m/s and 0.04 m/s respectively for the young and older women during the single walking trial condition. These values are in the range of meaningful change of between 0.04 and 0.06 m/s reported previously for older adults (Perera, Mody, Woodman, & Studenski, 2006) based upon effect size and SEM calculations. Similar values have also been reported in other clinical populations using confidence intervals (Evans, et al., 1997) and effect size statistics (Palombaro, Craik, Mangione, & Tomlinson, 2006). However, calculation of the upper bound for the 95% CI in our study shows that a velocity increase greater than 0.25 m/s for young women and 0.22 m/s for older women would indicate with 95% confidence that a genuine velocity increase or improvement had occurred. Thus, although we found statistical differences or bias, and this change is consistent with some studies reporting meaningful change, based on our small sample it might be argued that these small changes may not be important clinically. Interestingly, some intervention studies have reported mean improvement values below these levels for healthy older adults and clinical populations (Helbostad, Sletvold, & Moe-Nilssen, 2004; Yang, Wang, Lin, Chu, & Chan, 2006). Clinicians therefore should be cautious when interpreting significantly different scores independent of minimal detectable difference values.

It has also been proposed that systematic bias in test-retest gait data may be a statistical artifact that has no clinical relevance (van Uden & Besser, 2004). For example, the probability of

finding a significant difference in repeat measurements of a variable is likely when either its variability is small or differences across sessions is large (Hill & Lewicki, 2006). Interestingly, in our study low variability in the gait data was found in both the single and continuous conditions. However, larger mean differences were found during the single walk condition than the continuous walk condition, particularly for the older women. This accounts for the greater number of gait parameters showing bias in the single walk condition for the older group. Therefore, it is likely that this outcome is not the product of statistical artifact.

On average, walking velocity increased across sessions by 0.045 m/s (p<0.05) for the single walk condition and 0.01 m/s for the continuous walk condition. This is the likely cause of the greater number of systematic biases found in the gait variables for the single walk condition, because of the association of some of the spatiotemporal measures with gait velocity (Finley, Cody, & Finizie, 1969; Winter, 1991). From a clinical perspective, this change in walking speed could be interpreted as being indicative of gait improvement. The idea of gait function improving across test sessions however is somewhat implausible because no intervention was employed in a population with a stable health state. Nonetheless, improvements in second session spatiotemporal gait parameters have been reported previously in reliability studies (Evans, et al., 1997; Hill, et al., 1994), and may be partly explained by greater familiarity with the testing protocol (Hill, et al., 1994; Hopkins, 2000). The single trial protocol involves frequent waiting and steady-state walking periods, and phases of gait initiation and termination where the participant accelerates from rest in response to a “go” command and decelerates to stop once past the end of the walkway. The participants may have been more familiar with these requirements by the second testing session, thereby improving performance, albeit minimally. During the continuous walking protocol however, there is a lack of both the disruptions to the gait cycle and fewer phases of gait initiation and termination than occur during single trial walking. This may have resulted in better familiarisation and therefore less systematic bias in the continuous walking condition. Thus, single trial walking may produce more changes in test-retest mean differences as

a consequence of familiarisation, suggesting a continuous walking protocol may be more stable and therefore may more readily detect changes in gait. This could have important clinical implications where gait changes are used to indicate clinical improvements.

4.4.2 Test-retest reliability of single and continuous walking trials

With the exception of the step width variable, moderately high reliability values were found for all of the spatiotemporal data for both walking conditions. Reduced reliability for step width has been reported previously in healthy adults, and has been attributed to the low spatial resolution of the instrumented walkway and the inherent variability of the parameter (Menz, et al., 2004; van Uden & Besser, 2004). No reliability trends were found for age or walking condition.

Based on ICCs, relative reliability was highest for velocity, step length, step time, stance time, and foot angle, whereas based on SEM and CV values, the variables of step length, step time, swing time, and stance time were found to have the greatest absolute reliability. The small discrepancy between the measures highlights an important difference between these statistics. Relative reliability examines the consistency of the position, or rank, of a participant's score within a group over repeated sessions (Weir, 2005). However, this can be misleading because a heterogeneous sample where participants tend to hold their rank may be considered reliable despite large changes in participant values. In contrast, absolute reliability statistics are unaffected by sample heterogeneity and are more sensitive to changes in the participant's results over repeated tests (Hopkins, 2000). By reporting each measure, an understanding of changes in both position and participant variation of retest means is gained.

4.4.3 Study Limitations

Further research with larger participant numbers is needed to explore the extent of differences between spatiotemporal data collected during single and continuous walking protocols. An increase in sample size would also improve external validity and may further support the bias differences between the walk protocols found in this study. Future studies

should also include men in the sample, enabling sex differences in reliability and systematic bias to be explored. Indeed, an investigation by Menz et al. (2004) found no systematic differences in test-retest spatiotemporal data collected during single walks in a mixed sample of men and women. Aside from the methodological differences between the two studies, it is possible that the inclusion of men in the Menz study negated the presence of systematic bias in the women's data. This would be consistent with previous studies that have demonstrated differences in the gait variables of men and women (Kerrigan, Todd, & Croce, 1998). Standardization of walking direction between the two protocols may also improve the methodology. That is, whilst participants completed all ten walks in the same direction during our continuous trials; every second walk in the single trials was in the opposite direction. Although it is unlikely that these return single walks affected the results, future studies may benefit by having participants walk in one direction for both protocols. Finally, at this stage it is unclear as to whether the difference in systematic bias found between the single and continuous walking protocols extends to differences in gait measures such as walking variability, and whether these differences are also found in clinical populations such as older adult fallers. Between-protocol differences in measure of gait variability forms the basis of study two (Chapter 5) and gait measures are compared between walking protocols in older fallers and non-fallers in study three (Chapter 6).

4.4.4 Conclusions

Measuring changes in walking requires an instrument and data collection protocol that provides accurate results over repeat test sessions. This study found both the single and continuous over-ground walking conditions to be reliable collection methods for gait data. There was however some evidence of systematic bias between the sessions. More bias was found for trials where participants performed repeated single walks than when the same amount of information was collected from continuous walking. Although the magnitude of these significant test-retest differences was generally small in both walk conditions, it is reasonable to conclude that the continuous walk condition may be more stable over repeated testing session and

therefore may more readily detect gait changes in older women. Future studies with participants who are capable of walking between 200 and 300m without stopping may consider the adoption of a continuous walking protocol in order to minimize systematic bias across repeat test sessions. This may assist researchers and clinicians to better identify gait changes especially when working with elderly populations.

5 Study 2

Gait variability in younger and older women is altered by over-ground walking protocol

5 Gait variability in younger and older women is altered by over-ground walking protocol

5.1 Introduction

Measures of gait variability are commonly used to study age-related gait changes (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005; Hausdorff, et al., 1997; Hausdorff, Rios, & Edelberg, 2001; Heitman, Gossman, Shaddeau, & Jackson, 1989; Herman, Giladi, Gurevich, & Hausdorff, 2005; Moe-Nilssen & Helbostad, 2005; Stolze, Friedrich, Steinauer, & Vieregge, 2000). Some studies have shown that older adult fallers, and those identified with an increased falls risk, exhibit greater variability in basic spatiotemporal measures of gait when compared to both older adult non-fallers (Hausdorff, et al., 1997; Hausdorff, et al., 2001; Herman, et al., 2005) and younger adults (Hausdorff, et al., 1997). However, other work has reported no differences in measures of gait variability between healthy younger and older adults (Stolze, et al., 2000), healthy and frail older adults (Moe-Nilssen & Helbostad, 2005), and older adult fallers and non-fallers (Brach, et al., 2005; Heitman, et al., 1989).

A possible reason for the ambiguous findings of the previous studies may lie in the different walking protocols used to collect the gait data. Studies that have reported no effect of age on measures of gait variability commonly used repeated single over-ground walking protocols (Brach, et al., 2005; Heitman, et al., 1989; Moe-Nilssen & Helbostad, 2005; Stolze, et al., 2000). Conversely, studies using continuous over-ground walking protocols have reported significant age effects (Hausdorff, et al., 1997; Hausdorff, et al., 2001; Herman, et al., 2005; Kang & Dingwell, 2008). Repeated single trial walking protocols generally involve repetitive periods of waiting, gait initiation in response to an auditory command, steady state walking for several strides, followed by gait termination at the end of a short walkway. In contrast, continuous walking protocols typically involve walking without interruption over longer distances. It is possible that the frequent disruptions to the temporal locomotor rhythm of gait experienced during repeated

single walking protocols may affect gait variability and thereby contribute to the ambiguous findings reported by gait variability studies (Dingwell, Cusumano, Cavanagh, & Sternad, 2001).

To our knowledge, no study has investigated the effect of repeated single and continuous over-ground walking protocols upon measures of gait variability. As such, little is known about the effect of these protocols upon measures commonly used to investigate gait changes associated with ageing.

5.1.1 Aims and Hypotheses

The aim of this investigation was:

- To determine whether gait variability data captured during repeated single over-ground walking differs from variability data captured during continuous over-ground walking in younger and older women.

The hypothesis was:

- That gait variability data will be higher when collected using a single walking trial protocol than a continuous walking protocol.

5.2 Methods

5.2.1 Participants

Twenty two younger female volunteers (age: 21.2 ± 2.5 years, height: 1.66 ±0.08 m; mass: 62.6 ± 9.8 kg) and 32 older female volunteers (age 67.4 ± 6.3 years, height: 1.62 ± 0.07 m; mass: 65.1 ± 13.2 kg) participated in this study. Recruitment and selection criteria are described in Chapter 3.2.

5.2.2 Procedure

Participants were initially screened (see below) to ensure they did not have any overt health problems impacting on balance and mobility. If suitable, participants then completed ten repeated single walking trials and ten continuous laps of a walking circuit, presented in random order. Screening procedures and walking protocol are discussed in detail in Chapter 3.4.1 and 3.4.4 respectively. In summary, these procedures were completed as follows:

- self reported medical and surgical history
- medication use
- self reported fall history
- height, weight and leg length
- pulse and blood pressure whilst supine and one and three minutes after standing
- vibration sense using a Rydel-Seiffer graduated tuning fork
- lower limb joint proprioception via movement of the left and right hallux
- visual acuity using snellen eye chart
- visual contrast sensitivity using the Melbourne Edge Test
- vestibular function using the vestibular stepping test
- cognitive function using the Mini Mental State Examination
- single and continuous walking trials using a GAITRite® walkway

5.2.3 Statistical Analyses

Walking velocity, step and stride length, step and stride time, and step width were collected by the GAITRite® system and analysed with SPSS (Version 12). To quantify gait variability, standard deviation (SD) and coefficient of variation (CV) were calculated. It is recommended that CVs should only be calculated for ratio data (Atkinson & Nevill, 1998). Therefore since step width is interval data, CVs were not calculated for this parameter.

Paired *t* tests were used to compare variability measures across protocols. Bonferroni adjustments were not made because this study was interested in detecting differences in each individual gait parameter (Perneger, 1998). Instead, effect sizes were calculated to determine the importance of statistical differences found between the protocols (Portney & Watkins, 2000). Based on Cohen's (1988) suggestions, a small effect size was defined as 0.2, a medium effect as 0.5, and a large effect as 0.8.

5.3 Results

5.3.1 Participant characteristics

Younger adults were active (two to three times per week), had good vision (median visual acuity 0.0, median contrast sensitivity 23), and intact vibration perception, proprioception, vestibular function and balance as measured using the Romberg test (i.e. no participants showed impairments). Older adults were also healthy, with a median of three past or current medical conditions and taking a median number of three medications. Despite being active (median of three to four activity session per week), nearly a fifth of the sample (17%) occasionally experienced some level of pain during walking. Results from medical history and screening assessments for the older adults are displayed in Table 5.1 below.

Table 5.1. Medical profile and screening test results for older adults.

Parameter	
Median number of medical conditions (range)	3 (0 to 9)
Most common medical problems (% of sample)	
Hypertension	50.00
Osteoarthritis	37.50
Hyperlipidemia	25.00
Hysterectomy	21.88
Reflux	18.75
Breast cancer	12.50
Osteoporosis	12.50
Median number of medications (range)	3 (0 to 8)
Most common medications (% of sample)	
Antihypertensive agents	62.50
Minerals	28.13
Anti-coagulant, anti-thrombotic	25.00
Hypolipidemic agents	25.00
Hyperacidity, reflux and ulcers agents	21.88
NSAIDs	15.63
Analgesic - simple and antipyretics	12.50
Mean number of falls in previous 12 months ($\pm$ SD, [range])	0.38 $\pm$ 0.66 (0 to 2)

Table 5.1. *Continued.*

Parameter	
Physical measures	
Pain during walking (% of sample)	17
Median activity level (times per week)	3-4
Mean MMSE (± SD)	28.47 ± 2.33
Median VA (logmar, median [range])	0.1 (-0.08 to 0.2)
Mean contrast sensitivity (± SD)	18.6 ± 2.11
Impaired vibration (%)	9
Impaired proprioception (%)	19
Positive vestibular stepping test (%)	19
Positive Romberg (%)	9

Note: NSAIDs, non-steroidal anti inflammatory drugs; MMSE, mini mental state examination; VA, visual acuity.

5.3.2 Differences in mean spatial and temporal gait parameters between walking protocols

Mean values for the single and continuous walking protocols, mean difference values and the associated 95% confidence interval values, and calculated effect sizes for younger and older adults are displayed below in Table 5.2. Average values are similar to those reported previously on similar samples (Hageman & Blanke, 1986; Stolze, Friedrich, Steinauer, & Vieregge, 2000). Significant differences ($p<0.001$) were found for all the spatiotemporal gait parameters for both the young and older adults. Of these differences however, only velocity for the young adults had a medium effect size (ES = 0.45), whilst the effect sizes for step width, step and stride length and step and stride times were small (values ranging from 0.06 to 0.29).

Table 5.2. Mean values for the single and continuous walking protocols, between-protocol mean difference values, 95% confidence intervals of mean difference values, and effect sizes of between protocol comparisons for each gait variable for the younger and older participants.

	Younger					Older				
	Single	Continuous	Mean Difference	95% Confidence Intervals	Effect size	Single	Continuous	Mean Difference	95% Confidence Intervals	Effect size
Velocity (m/s)	1.55	1.48	0.07*	0.04 to 1.0	0.45	1.36	1.31	0.05*	-0.02 to 0.05	0.19
Step length (cm)	77.59	76.01	1.6*	1.2 to 2.1	0.25	67.49	66.60	0.9*	0.7 to 1.1	0.13
Stride Length (cm)	155.18	152.18	3.0*	2.2 to 4.5	0.24	134.89	133.18	1.7*	1.4 to 2.1	0.13
Step width (cm)	8.62	8.81	-0.2*	-0.6 to -0.1	0.14	7.81	8.01	-0.2**	-0.4 to -0.1	0.06
Step time (sec)	0.506	0.515	-0.009*	-0.012 to -0.007	0.27	0.510	0.517	-0.007*	-0.008 to -0.005	0.15
Stride time (sec)	1.012	1.030	-0.018*	-0.026 to -0.014	0.29	1.0219	1.035	-0.014*	-0.017 to -0.011	0.15

Note: * Indicates significant difference between the walking protocol at the 0.001 level; ** indicates significant difference between the walking protocols at the 0.01 level.

5.3.3 Differences in gait variability between walking protocols

Variability data for each group and walking protocol are listed in Tables 5.3 and 5.4. For the younger adult group, significant differences between the single and continuous walking protocols were found for velocity (p=0.01), step length (p=0.04) and stride time (p=0.02) SDs. Additionally, velocity (p=0.02), step time (p=0.02) and stride time (p=0.01) CVs were also found to be significantly different between the repeated single and continuous walking protocols. Most of the effect sizes (ES) for these differences were medium to large, ranging from 0.46 to 0.79. For the older adults, significant differences between the repeated single and continuous walking protocols were found for velocity, step length and stride length SDs and CVs (p<0.01). Medium effect sizes were observed for these differences, with values ranging from 0.34 to 0.58.

Table 5.3. Standard deviations, coefficients of variation and effect size of each variable for the repeated single and continuous walking protocols for younger participants.

	SD				CV			
	Single	Continuous	p value	Effect size	Single	Continuous	p value	Effect size
Velocity (cm/s)	6.0	4.6	0.01	0.79	3.9	3.2	0.02	0.63
Step length (cm)	2.5	2.2	0.04	0.46	3.1	2. 9	0.07	0.34
Stride Length (cm)	4.2	3.7	0.05	0.43	2.7	2.4	0.07	0.33
Step width (cm)	1.8	1.9	0.43	0.05	NA	NA	NA	NA
Step time (msec)	1.6	1.5	0.08	0.41	3.2	2.9	0.02	0.61
Stride time (msec)	2.2	1.9	0.02	0.67	2.2	1.8	0.01	0.78

Note: SD indicates standard deviations; CV indicates coefficient of variation; NA indicates not applicable.

Table 5.4. Standard deviations, coefficients of variation and effect size of each variable for the repeated single and continuous walking protocols for older participants.

	SD				CV			
	Single	Continuous	*p* value	Effect size	Single	Continuous	*p* value	Effect size
Velocity (cm/s)	5.0	3.8	<0.01	0.58	3.8	2.9	<0.01	0.51
Step length (cm)	2.5	2.2	<0.01	0.46	3.6	3.3	<0.01	0.34
Stride Length (cm)	3.7	3.2	<0.01	0.48	2.8	2.4	<0.01	0.40
Step width (cm)	2.2	2.5	0.05	0.34	NA	NA	NA	NA
Step time (msec)	1.9	1.9	0.47	0.02	3.6	3.6	0.44	0.04
Stride time (msec)	2.6	2.4	0.14	0.24	2.6	2.3	0.09	0.31

Note: SD indicates standard deviations; CV indicates coefficient of variation; NA indicates not applicable.

5.4 Discussion

This study investigated the effect of over-ground walking protocol on the gait variability of younger and older women. The repeated single walking protocol resulted in significantly higher SDs for velocity, step length and stride time, and significantly higher CVs for velocity, step time and stride time in the younger adult sample. For the older participants, the repeated single walking protocol resulted in significantly higher SDs and CVs for velocity, step length and stride length. Medium to large effect sizes were found for the majority of these increases in variability.

The higher gait variability found with repeated single walking may be due to the repeated stoppages in the protocol. Although recent work (Orendurff, Schoen, Bernatz, Segal, & Klute, 2008) suggests this protocol might better reflect everyday walking, recording variability in this manner assumes that any given stride is unaffected by a previous stride (Dingwell, et al., 2001). This assumption is questionable since other studies have shown that a given stride is affected by previous strides, demonstrating dependency between consecutive gait cycles (Dingwell, et al., 2001; Griffin, West, & West, 2000; Hausdorff, et al., 1996). Hence it is possible that any inter-relationship between strides during continuous walking may be perturbed by the repeated stoppages encountered in the single walking protocol. The presence of this perturbation may partly explain the higher gait variability exhibited by the participants in this protocol.

The significant increases found in the gait variability measures for the repeated single walking protocol may have important clinical implications. Previous work has shown that increased variability in the measures of stride length, stride time, stride width and walking velocity are associated with future falls in older adults and pathological populations (Hausdorff, et al., 2001; Herman, Giladi, Gurevich, & Hausdorff, 2005). Consequently, many researchers and clinicians now use the presence of increased gait variability as a marker of gait instability or impairment. The results of this investigation however show that a repeated single walking protocol may increase gait variability. It would be of interest to investigate whether the increased variability

found during the repeated single trials as opposed to the continuous trials affects the capacity of clinicians and researchers to identify gait impairments or predict older adults at risk of future falls. This question is addressed in study 3 (Chapter 6).

Despite findings of altered walking variability during the repeated single trials, a continuous walking protocol might not be feasible in certain clinical settings. In this study, the continuous circuit was approximately 11 x 3 m, which some clinicians might not have available for gait evaluation. However, other studies employing a continuous walking protocol in confined spaces using clinical populations have instructed participants to walk back and forth without stopping along a walkway or corridor [e.g. (Bartsch, et al., 2007; Herman, et al., 2005)]. Consequently, clinicians wishing to employ a continuous walking protocol in a setting with reduced space might employ similar methods.

5.4.1 Study Limitations

A potential limitation of this study was the collection of gait data over a discrete section of the continuous walking circuit. Recording gait data in this manner may result in a series of gait trials that are similar to a repeated single walk protocol. This protocol however was chosen in order to reflect the increasing use of instrumented mats in clinical gait research (Besser, Selby-Silverstein, & Prickett, 2001; Brach, et al., 2005; Moe-Nilssen & Helbostad, 2005). Furthermore, strides from discrete gait trials collected during continuous over-ground walking do not have the same spatial and temporal perturbations that result from the repeated stoppages during single walking protocols. This ensures maintenance of the dependency relationship between gait cycles during a continuous walking protocol.

5.4.2 Conclusions

In conclusion, this study found that measures of gait variability are altered by the over-ground walking protocol adopted. In comparison to a continuous over-ground walking protocol, a repeated single over-ground walking protocol significantly increased the variability of several

basic spatiotemporal measures of gait. Future research should consider single trial or continuous walking protocol when investigating gait variability.

6 Study 3

A prospective study of stride dynamics, gait variability and falls risk in active community dwelling older women

6 A prospective study of stride dynamics, gait variability and falls risk in active community dwelling older women

6.1 Introduction

Falls are the leading cause of injury deaths in older adults (Kochanek, Murphy, Anderson, & Scott, 2004). Falling is also associated with high rates of serious injuries such as hip fractures (Rapp, Becker, Lamb, Icks, & Klenk, 2008). Following a fall, older adults are more likely to have decreased levels of physical activity (Murphy, Williams, & Gill, 2002), an increased fear of falling (Jørstad, Hauer, Becker, & Lamb, 2005) and reduced confidence in physical abilities (Yardley & Smith, 2002). Moreover, after an initial fall, risk of further falls is also heightened, further increasing the likelihood of secondary complications such as morbidity, institutionalisation and death (Clough-Gorr, et al., 2008). Given the majority of falls occur outdoors, and outdoor falls occur more often in active rather than frail older adults (Bergland, Jarnlo, & Laake, 2003; Li, et al., 2006), there is merit in identifying an early marker of falls risk in active community dwelling older adults before secondary complications such as fear of falling and disability arise.

The majority of outdoor falls have been shown to occur during walking (Li, et al., 2006) and as such, age-related alterations in gait parameters have been extensively studied. For example, a small number of prospective studies have shown that decreased average walking speed is associated with future falls in community dwelling older adults (Montero-Odasso, et al., 2005; Verghese, Holtzer, Lipton, & Wang, 2009). However, other work has suggested that this may be an adaptation to fear and not a cause of walking instability (Maki, 1997). Consequently, some falls researchers have looked beyond the average or mean value of gait parameters and studied the stride-to-stride fluctuations that occur during walking. It has been suggested that quantifying stride fluctuations through measures of gait variability might provide additional insights into the neuromotor control of walking and assist in identifying gait instability and falls risk in older adults

(Hausdorff, Nelson, et al., 2001; Hausdorff, 2007).

To date, there is some evidence supporting a link between gait variability, instability and falls. For example, some studies have shown that the magnitude of gait variability is altered in older adult fallers compared to non-fallers (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005; Herman, Giladi, Gurevich, & Hausdorff, 2005; Springer, et al., 2006). Importantly, some measures of gait variability have also been shown to be stronger and more sensitive predictors of falls than some averaged gait measures, including walking speed (Hausdorff, Rios, & Edelberg, 2001; Maki, 1997; Verghese, et al., 2009). However, other work has reported similar magnitudes of gait variability between fallers and non-fallers (Brach, et al., 2005; Heitman, Gossman, Shaddeau, & Jackson, 1989; Menz, Lord, & Fitzpatrick, 2003). In light of the inconsistent findings of these studies, the clinical usefulness of gait variability is currently unclear. It is possible however that the inclusion of older adults already demonstrating signs of mobility impairments, particularly in the prospective gait variability studies, contributed to the inconsistent findings. Therefore, further work on active community dwelling older adults is required to explore the clinical utility of measures of gait variability in predicting falls in higher functioning populations.

Another possible explanation for the conflicting findings in the gait variability studies is the different over-ground walking protocols used to collect variability data. For instance, many of the studies that reported increased gait variability in older fallers used a continuous walking protocol, where participants walked for a fixed period of time (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Hausdorff, Rios, & Edelberg, 2001; Herman, et al., 2005). In contrast, studies that have reported no difference used a repeated single trial protocol that involved repetitive periods of waiting, gait initiation, steady state walking and gait termination (Brach, et al., 2005; Heitman, et al., 1989; Menz, et al., 2003). Given there is evidence that gait variability is altered during continuous treadmill compared to over-ground walking (Dingwell, Cusumano, Cavanagh, & Sternad, 2001), it is also possible that gait variability is altered by the over-ground walking

protocol. The influence of walking protocol on gait variability and falls prediction is yet to be explored.

In addition to quantifying the magnitude of gait fluctuations, researchers have also explored the underlying dynamic structure of stride fluctuations and their changes over longer time periods, termed stride dynamics. In healthy young adults, fluctuations in stride time at any one moment have been shown to be statistically related to fluctuations occurring hundreds of strides earlier in a fractal-like manner (Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995; Hausdorff, et al., 1996; Jordan, Challis, Cusumano, & Newell, 2009; Jordan, Challis, & Newell, 2007; Pierrynowski, et al., 2005; Terrier, Turner, & Schutz, 2005). In older adults (Hausdorff, Mitchell, et al., 1997) and individuals with neurological disorders (Frenkel-Toledo, et al., 2005; Hausdorff, et al., 2000; Hausdorff, Mitchell, et al., 1997; Herman, et al., 2005), this long-term fractal organisation has been shown to break down, suggesting ageing and disease might disturb the control of stride dynamics. It is possible therefore that evaluating stride dynamics could provide further information regarding walking stability and falls risk in older adult populations.

To date, only one retrospective study has explored stride dynamics in older adult fallers, albeit using a clinical population. In a sample of patients diagnosed with a higher level gait disorder, Herman et al. (2005) reported that a measure of stride dynamics, the fractal scaling index, was significantly (p<0.009) reduced in fallers compared to non-fallers. This study also included a control sample of age-matched healthy older adults, however no report was made as to whether the differences between fallers and non-fallers remained when the controls were included in the analysis. Therefore, it is unclear whether stride dynamics are altered in active community dwelling older adult fallers.

Additionally, of the two studies that have investigated stride dynamics in older adults, data recorded from only one limb was analysed (Hausdorff, Mitchell, et al., 1997; Herman, et al., 2005). However, there is some evidence that inter-limb temporal control is altered in older

fallers. For example, Yogev et al. (2007) found greater asymmetry in the swing time of older fallers compared to non-fallers, and other work showed that stance time asymmetries can predict future falls in older adults (Hill, Schwarz, Flicker, & Carroll, 1999). Given this evidence of altered inter-limb temporal control in fallers therefore, further investigation of the influence of inter-limb dynamics on falls is also warranted in this population.

In summary, measures of stride dynamics and gait variability have the potential to provide clinically useful markers of early falls risk in older adults. However, a number of important questions need to be addressed before the clinical utility of these measures is known. For instance, previous prospective falls studies investigating gait variability in older adults, and the only falls study investigating the stride dynamics of older adults, evaluated participants already showing signs of mobility limitations, or who were diagnosed with a mobility disorder. As such the usefulness of stride dynamics and gait variability to predict falls in active community dwelling older adults is unknown. It would be of considerable clinical value to develop an early marker of falls risk in an active and otherwise healthy older adult population before other secondary changes such as fear of falling and mobility problems become apparent. Additionally, the control of inter-limb stride dynamics in active community dwelling older fallers and non-fallers has not been explored. However, given there is evidence of altered between-limb temporal control in older fallers, it would be of interest to examine the influence of inter-limb dynamics on walking stability and falls in older adults. Finally, no prospective study has investigated the effects of walking protocol on discriminating and predicting older adult fallers and non-fallers. As such it is unknown as to whether walking protocol influences gait variability and falls prediction.

6.1.1 Aims and Hypotheses

The aims of this investigation were:

- To determine whether stride dynamics, as measured by the fractal scaling index, is different between active older fallers and non-fallers.

- To evaluate differences in inter-limb gait dynamics between active older fallers and non-fallers.
- To determine whether gait variability is different between active older fallers and non-fallers and to evaluate the influence of walking protocol upon these differences.
- To explore the usefulness of stride dynamics and gait variability in predicting falls in active older adults, and to determine the influence of walking protocol upon prediction accuracy.

The hypotheses were:

- The fractal scaling index will be reduced in fallers compared to non-fallers.
- Inter-limb dynamics will be more asymmetrical in fallers compared to non-fallers.
- Gait variability data will be greater in fallers compared to non-fallers.
- Measures of stride dynamics and gait variability will predict future fallers.
- Walking protocol will influence between-group differences and the prediction accuracy of gait variability data.

6.2 Methods

6.2.1 Participants

One hundred and eight older females (age: 68.64 ± 7.44 years, height: 1.61 ±0.06 m; mass: 70.53 ± 16.89 kg) volunteered to participate in the study. Recruitment and selection criteria are described in Chapter 3.2. Five participants were excluded during the screening process due to the presence of medical conditions known to impact upon walking. These included three participants who reported having had a stroke, one participant who was unable to complete the continuous walking protocol and another participant with Parkinson's disease. Additionally, a further six participants were lost during the subsequent 12 month follow up period due to loss of contact from either extended hospitalisation or institutionalisation (three participants) or moving house

with no forwarding contact details (three participants). Consequently, a final sample of 97 older women completed all testing and follow-up procedures (Figure 6.1).

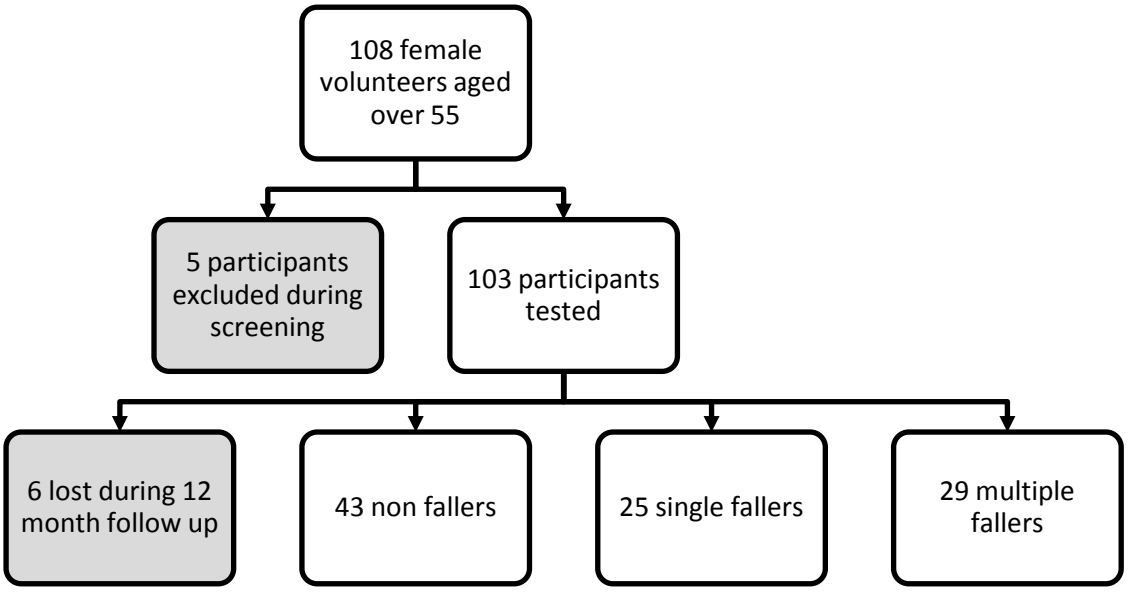


Figure 6.1. Flowchart of sample composition.

6.2.2 Procedure

6.2.2.1 Screening and initial assessment

Prior to data collection, participants were first screened to ensure suitability to the aims of the study (Chapter 3.4.1). Following screening, laterality and balance assessments were completed as outlined in Chapters 3.4.2 and 3.4.3 respectively. Two CXL10LP3 tri-axial Crossbow® accelerometers were then attached to each participant's left and right feet and connected to a portable data logger attached to a belt worn around the waist, as outlined in Chapter 3.3.2. All leads were secured and free movement was maintained.

6.2.2.2 Familiarisation

Participants completed two practice trials each of a repeated single walking protocol and a continuous walking circuit to ensure familiarisation to the testing apparatus and walking assessment. Each participant's average walking velocity was recorded during familiarisation, and

the approximate number of walking trials was determined to enable the collection of similar quantities of data during each walking condition.

6.2.2.3 Walking protocol and falls assessment

Following familiarisation, participants completed seven minutes of continuous walking around the circuit, and the equivalent number of repeated single walking trials, as outlined in Chapter 3.4.4. A GAITRite® walkway recorded spatial and temporal gait data for each pass during both walking protocols. Foot-mounted accelerometers were also employed to continuously record anteroposterior accelerations and derive continuous temporal gait data for each walking protocol (see Chapter 3.5.1). Walking protocol was presented in random order. Falls assessment packs were then provided, which included a falls calendar and questionnaires, and participants were required to send in a monthly sheet for 12 months indicating whether they experienced a fall during that month (Chapter 3.4.5). If a calendar sheet was not returned, or was completed incorrectly, the participant was contacted by phone by the principal investigator (KP).

6.2.2.4 Reliability testing

A small subset (n=12) of participants attended a second visit one week following their initial testing session to evaluate the test-retest reliability of the accelerometer data. These participants were the first 12 volunteers that were willing and available to attend the laboratory at the same time on the following week from their initial testing session. Only the gait assessment was completed at this second testing session.

6.2.2.5 Summary of methodology

With the exception of the walking assessments which were presented in a randomised order, all assessments were presented in a fixed order. All screening and assessment procedures are described in Chapter 3.4. In summary, screening, balance, laterality, gait and falls assessments were completed as follows:

- self reported medical and surgical history
- medication use
- self reported fall history
- height, weight and leg length
- pulse and blood pressure whilst supine and one and three minutes after standing
- vibration sense using a Rydel-Seiffer graduated tuning fork
- lower limb joint proprioception via movement of the left and right hallux
- visual acuity using a snellen eye chart
- visual contrast sensitivity using the Melbourne Edge Test
- vestibular function using the vestibular stepping test
- cognitive function using the Mini Mental State Examination
- laterality assessment using the lateral preference inventory
- balance assessments using the step test, the timed up and go test, the functional reach test, Romberg's test and sharpened Romberg's test. Scores from the first four of these tests were also aggregated to provide a measure of global balance termed the BOOMER score
- gait variability and stride dynamics recorded using two Crossbow® accelerometers and a GAITRite® walkway during continuous and repeated single walking trials
- re-testing of gait data one week following initial testing for a small sub sample of participants to evaluate test-retest reliability
- falls status through the use of a 12 month falls diary

6.2.3 Analysis of gait data

Gait data were recorded by the GAITRite® walkway for each pass over the mat during both walking protocols. Additionally, foot-mounted accelerometers recorded continuous anteroposterior accelerations during both walking protocols. Spatial and temporal gait data was

extracted from the GAITRite® walkway system as described in Chapters 3.5 and 3.6.1. To enable comparison with previous studies, only the variables of walking velocity, stride length, foot angle, base of support, stride time, stance time and swing time were extracted. Additionally, stride time data was extracted from the accelerometers as described in Chapter 3.5.1. Except where indicated, ***data collected during the continuous walking circuit was used for the majority of analyses*** as the assessment of walking dynamics requires longer data sets recorded during continuous walking. Additionally, walking variability data was shown to be altered when collected during repeated single walking trials during an earlier study (Chapter 5) further supporting the use of continuous walking data. However, to investigate the influence of walking protocol on gait variability and falls, between-group analyses were repeated using data obtained during the repeated single walking trials (see Chapters 6.2.4.7 and 6.3.10).

Outliers from stride time data derived from the accelerometers were initially removed by visual inspection of the raw data series using custom designed software (Chapter 3.5.1). The majority of these points were due to noisy acceleration signals or stopping and starting steps. However, in four instances one of the accelerometers was faulty resulting in corrupt data for the entire trial for that foot. In these cases only unilateral data from the working accelerometer was used in further analysis. Next, acceleration data was filtered using a median filter whereby data points greater than three standard deviations from the median value were removed (Hausdorff, Edelberg, et al., 1997). This helped remove points that were due to turning at the ends of the circuit. A final check of data was made by examining a boxplot and excluding any extreme values extending more than three lengths from the edge of the box (Tabachnick & Fidell, 2001). No extreme values were observed however hence no further data was removed.

To quantify gait variability, the standard deviation (SD) was calculated for each gait parameter. The SD was used in preference to the coefficient of variation (CV) as the CV is only applicable to ratio data (data bounded by zero) (Atkinson & Nevill, 1998) and therefore should not be

calculated for gait variables that may have negative values, such as the base of support and foot angle (see Chapter 3.6.1.11). Additionally, Pearson product moment correlations between the CV and SD values for the ratio gait variables were highly correlated (r>0.91, p<0.01) and therefore the use of the SD in preference to the CV was not expected to influence the results.

To examine stride dynamics, the processed stride time acceleration signals were imported into a modified version of Biomedical Workbench (National Instruments®) written in LabVIEW (2009), and the stride time fractal scaling index was calculated using detrended fluctuation analysis (DFA) (Hausdorff, et al., 1995; Peng, Havlin, Stanley, & Goldberger, 1995). DFA is unaffected by local trends or artifacts in the data that can arise due to non-stationarities, and hence is favoured in non-linear analysis of many physiologic time series such as stride time (Chen, Ivanov, Hu, & Stanley, 2002). To calculate the fractal scaling index, the entire time series, *N*, is initially integrated and then divided into equal windows of length *n*. This new integrated series, *y(k)*, is detrended by fitting a least squares line to the window and then subtracting the local trend, $y_n(k)$. Detrended fluctuations, *F(n)* , are then calculated using a modified root mean square analysis, repeated over increasing window sizes:

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}\left[y(k) - y_n(k)\right]^2} \qquad \text{Equation 6.1}$$

A log of *F(n)*is plotted against a log of *n* and the slope of the line on this graph provides the fractal scaling index (α). Increasing fluctuations with larger windows suggests a power-law relationship and the presence of scaling behaviour (Hausdorff, et al., 1995). For uncorrelated random fluctuations α = 0.5, for Brownian noise α = 1.5, and for long-range power law correlations with fractal scaling, 0.5 > α≤ 1.0 (Goldberger, et al., 2000; Goldberger, et al., 2002; Hausdorff, 2007). Based on recommendations by Hu et al. (2001), α was calculated between 4 and *N*/10 as larger box sizes could under-sample *F(n)*.

6.2.4 Statistical analyses

6.2.4.1 Statistical assumptions

All statistical analyses were conducted with SPSS (version 17). Normality was assessed using skewness, kurtosis and Shapiro-Wilk statistics, and by visually examining the normal QQ plot and comparing the distribution to a normal histogram (Tabachnick & Fidell, 2001). Multivariate normality was also assessed by calculating Mahalanobis distance values. Each of these normality measures demonstrated normality, consequently parametric statistics were used to analyse the data.

To avoid multicollinearity, dependent variables were grouped and analysed separately as gait, balance and physical (demographic and screening) measures, and relationships amongst the remaining dependent variables in each grouping were checked to ensure moderate correlations. The assumption of homogeneity of variance-covariance matrices was tested by Box's M Test of Equality of Covariance Matrices, and equality of variance for each variable was verified by inspecting the significance value from Levene's Test of Error Variances. No violations of normality, linearity, univariate or multivariate outliers, equality of variance, homogeneity of variance-covariance matrices and multicollinearity were noted unless otherwise stated.

6.2.4.2 Test-retest reliability of accelerometer and stride dynamic data

Test-retest reliability of the accelerometer stride time data and the stride time scaling exponent was determined by examining systematic bias and measures of relative and absolute reliability. To investigate systematic bias across the two testing sessions, paired *t* tests ($p<0.05$) were conducted and associated effect sizes were calculated. To assess relative reliability, intra-class correlation coefficients of type (3,1) ($ICC_{3,1}$) were calculated (Portney & Watkins, 2000). As ICC values can be influenced by heterogeneous samples (Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskiod-Samsøe, 2004), two absolute measures of reliability were also calculated. The first, the standard error of measurement (SEM), expresses reliability in the same units as the variables

of interest (Atkinson & Nevill, 1998), and was determined using the following formula (Portney & Watkins, 2000; Weir, 2005):

$$SEM = SD\sqrt{(1 - ICC_{3,1})} \qquad \text{Equation 6.2}$$

As discussed in Chapter 4.2.3 however, the SEM may be influenced by heteroscedasticity. Therefore, absolute reliability was also assessed using the coefficient of variation (CV), a dimensionless measure unaffected by heteroscedasticity. The statistic is calculated by dividing the SEM by the mean and expressing the value as a percentage (Hopkins, 2000). As outlined above, the CV is only applicable to ratio data (Atkinson & Nevill, 1998) and thus was not calculated for the gait parameters of base of support and foot angle.

6.2.4.3 Validity of accelerometer data

To evaluate validity of stride time data recorded by the Crossbow® accelerometers, comparisons were made to stride time data concurrently recorded by the GAITRite® walkway. Excellent agreement has been shown between GAITRite® data and other spatiotemporal gait data collection methods including pencil and paper (McDonough, Batavia, Chen, Kwon, & Ziai, 2001), a single video camera system (Cutlip, Mancinelli, Huber, & DiPasquale, 2000; McDonough, et al., 2001; Wilson, Lorenzen, & Lythgo, 2002), a three-dimensional motion analysis system (Webster, Wittwer, & Feller, 2005) and the clinical stride analyzer (Bilney, Morris, & Webster, 2003).

Firstly, to explore differences between the two systems, a paired *t* test ($p<0.05$) was performed, and the associated effect size was calculated. Next, concurrent validity was assessed using an intra-class correlation coefficient of type (2,1) ($ICC_{2,1}$) (Portney & Watkins, 2000). Finally, level of agreement was evaluated by calculating absolute (AbsRC) and percentage mean repeatability coefficients (MeanRC) (Bland & Altman, 1986; Webster, et al., 2005). The absolute repeatability coefficient was calculated by multiplying 1.96 by the standard deviation of the difference between the GAITRite® and Crossbow® stride times. The percentage of this value was

derived by dividing by the average stride time of the two systems and multiplying by 100%. Smaller values indicate greater agreement between the two systems.

6.2.4.4 Between group differences

The sample was stratified into fallers (participants who fell at least once) and non-fallers (participants who did not fall) based on responses to the 12 monthly falls diary assessment (Chapter 3.4.5). To compare differences in sample characteristics between the groups, two one-way between subjects Multivariate Analyses of Variance (MANOVAs) were performed using the physical (demographic and screening) and balance measures listed in Table 6.1 as dependent variables. Independent samples *t* tests were then used to compare differences in the fractal scaling index between fallers and non-fallers, and where indicated, differences in individual gait variables such as walking speed. Between-leg differences in stride dynamics were assessed using a paired samples *t* test. Differences in the magnitude of gait variability between fallers and non-fallers were also evaluated using a one-way between subjects MANOVA with the gait variability measures as dependent variables (Table 6.1). Bonferroni adjustments were performed as appropriate to protect against Type I error, and effect sizes (ES) were calculated to ascertain the statistical strength of any observed difference. Using Cohen's (1988) suggestions for interpreting the strength of effect size values, statistical strength of differences was interpreted based upon 0.2 indicating a small effect, 0.5 a medium effect and 0.8 a large effect. In addition, partial eta squared value were also provided for multivariate analyses, with 0.01 indicating a small effect, 0.06 a moderate effect and 0.14 a large effect (Cohen, 1988). Finally, as outlined in Chapter 3.6.4, a minimum sample size of 26 was required for each sample to detect differences between groups at a significance level of 0.05 and with 80% power.

Table 6.1. Physical (demographic and screening), balance and gait variability measures used in the MANOVA analyses.

	Physical measures	Balance measures	Gait variability measures
Purpose	To investigate physical differences between fallers and non-fallers	To investigate balance differences between fallers and non-fallers	To investigate gait differences between fallers and non-fallers
Independent Variable	Faller and non-faller	Faller and non-faller	Faller and non-faller
Dependent Variables	1. Age 2. Height 3. Weight 4. Previous falls 5. Pain during walking 6. Activity level 7. Handedness 8. Footedness 9. Left vibration 10. Right vibration 11. Proprioception 12. Left VA 13. Right VA 14. Contrast sensitivity 15. MMSE 16. VST distance 17. VST angle	1. ST (worst leg score) 2. TUG 3. FR 4. Romberg 5. Sharpened Romberg 6. BOOMER score	1. Velocity SD 2. Stride length SD 3. Foot angle SD 4. Base of support SD 5. Stride time SD 6. Stance time SD 7. Swing time SD

Note: See Chapter 3.4.1 and 3.4.2 for description of physical measures, Chapter 3.4.3 for description of balance assessments and Chapter 3.6.1 for description of gait measures. VA, visual acuity; MMSE, mini mental state examination; VST, vestibular stepping test; ST, step test; TUG, timed up and go; FR, functional reach; BOOMER, balance outcome measure for elder rehabilitation.

6.2.4.5 Prediction of fallers

To evaluate the ability of stride dynamics and gait variability to predict future fallers, direct logistic regression was performed. Logistic regression was chosen over discriminant analysis to facilitate comparison to previous studies [e.g. (Hausdorff, Rios, et al., 2001; Maki, 1997)]. In addition to the gait variability measures listed in Table 6.1, their corresponding average values and the fractal scaling index were also included as predictor variables. Significant predictors of falling were determined using Wald's criterion, and where identified, odds ratios were calculated to evaluate the importance of risk.

6.2.4.6 Influence of fall group stratification

Research has shown differences in fall characteristics for older people who are single fallers and those who have fallen multiple times (Morris, et al., 2004). For example, multiple fallers were more likely to report intrinsic factors such as mobility problems and reduced balance as predisposing factors for their fall compared to single fallers. As such, the identification of a sensitive marker of gait decline in this high risk population could help prevent the greater probability of injury and other secondary complications arising from falls. No gait variability study has compared multiple fallers to a group of single or non-falling group. Consequently, to evaluate the influence of multiple falls as an independent variable, all analyses were repeated with the sample stratified into non-fallers, single fallers and multiple fallers, and into multiple fallers and a combined group of single and non-fallers. The analyses were completed using one-way between groups MANOVAs, independent and dependent *t* tests and logistic regression as appropriate.

6.2.4.7 Influence of walking protocol

To confirm findings, and to evaluate whether walking protocol influences the study outcomes, all between-group and prediction analyses were repeated using data collected from the repeated single walking protocol. Due to dependence of α on data length (Hu, et al., 2001; Kantelhardt, Koscielny-Bunde, Rego, Havlin, & Bunde, 2001), the fractal scaling index was unable to be calculated using data from repeated single trials and hence this comparison was not performed.

6.3 Results

6.3.1 Sample Characteristics

One hundred and three participants were initially tested on all physical, balance and gait measures, however six were lost during the 12 month follow-up period. Consequently, 97 seven older women (age: 68.73 ± 7.07 years, height: 1.61 ±0.06 m; mass: 69.78 ± 16.00 kg) underwent screening procedures, completed all gait assessments and were followed for 12 months to record fall incidence. The participants that were lost during the follow up period were similar to those

that completed the study in most measures, with the exception of increased weight (p=0.02) of the participants that were lost. Baseline demographic, screening and balance results for the two groups are presented below in Table 6.2 and Table 6.3.

Table 6.2. Demographic information for participants that completed the study and participants that were lost during the follow up period (mean ± SD unless otherwise stated).

	Completed participants (n=97)	Participants lost during follow up (n=6)
Age (years)	68.73 ± 7.07	65.83 ± 10.07
Height (cm)	161.07 ± 6.29	161.52 ± 9.66
Weight (kg)	69.78 ± 16.01	85.82 ± 24.34*
Median number of falls in previous 12 months (± SD, [range])	0 ± 0.62 (0 to 2)	0.5 ± 0.98 (0 to 2)
Pain during walking (% of sample)	27.84	16.67
Median activity level (times per week, ± SD)	2 ± 0.70	2 ± 0.75
Right handedness (% of sample)	87.63	100
Right footedness (% of sample)	86.60	100
Median number of medical conditions (range)	4 (0 to 11)	2.5 (0 to 3)
Most common medical problems (% of sample)		
Osteoarthritis	42.27	33.33
Hypertension	38.14	0
Hyperlipidemia	26.80	0
Hysterectomy	22.68	0
Tonsillectomy	19.59	16.67
Appendectomy	12.37	33.33

Table 6.2. *Continued*.

	Completed participants (n=97)	Participants lost during follow up (n=6)
Median number of medications (range)	2 (0 to 9)	1 (0 to 6)
Most common medications (% of sample)		
Antihypertensive agents	40.21	0
Hypolipidemic agents	26.80	0
Anti-coagulant, anti-thrombotic	19.59	0
Herbal analgesic and anti-inflammatory	13.40	16.67
Hyperacidity, reflux and ulcers agents	13.40	16.67
Gonadal hormones	7.22	33.33

Note: See Chapter 3.4.1 for assessment description. * indicates significant differences between the two groups (p=0.02).

Table 6.3. Screening and balance assessment data for participants that completed the study and participants that were lost during the follow up period (mean ± SD unless otherwise stated).

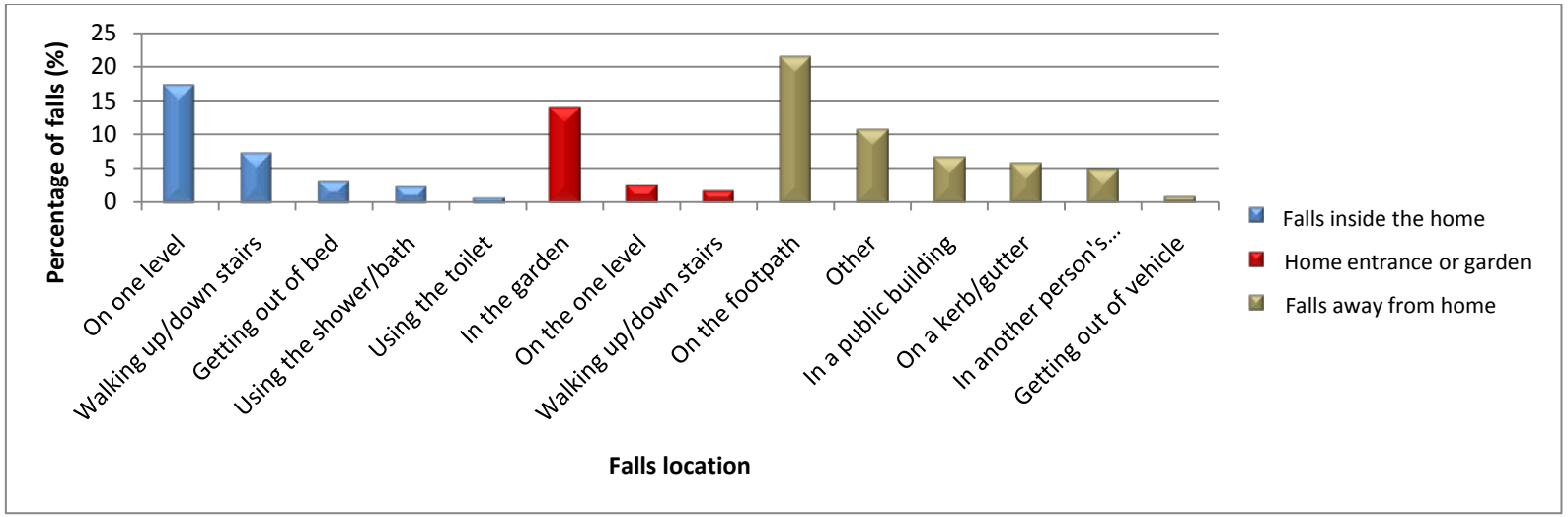
	Completed participants (n=97)	Participants lost during follow up (n=6)
Screening assessments		
Left vibration (Hz, ± SD)	6.11 ± 1.50	5.32 ± 1.94
Right vibration (Hz, ± SD)	6.15 ± 1.68	5.65 ± 1.97
Impaired proprioception (% of sample)	5.15	16.67
Left median VA (range)	0.18 (-0.4 to 1)	0.27 (0 to 0.58)
Right median VA (range)	0.18 (0 to 0.8)	0.38 (0 to 0.8)
Mean contrast sensitivity (± SD)	20.30 ± 2.07	21.50 ± 1.52
Mean MMSE (± SD)	28.01 ± 2.14	27.50 ± 2.51
VST distance (cm, ± SD)	63.13 ± 30.77	39.25 ± 29.94
VST angle (°, ± SD)	23.35 ± 24.28	20.83 ± 21.31
Balance assessments		
Worst ST (number of steps in 15 sec, ± SD)	14.68 ± 4.05	16.5 ± 5.43
TUG (sec, ± SD)	7.70 ± 1.93	8.28 ± 1.44
FR (cm, ± SD)	32.62 ± 6.51	34.58 ± 8.81
Romberg (sec, ± SD)	29.77 ± 1.67	27 ± 6.00
Sharpened Romberg (sec, ± SD)	4.11 ± 1.53	4.17 ± 2.04
BOOMER (% of sample <12)	1	16.67

Note: See Chapter 3.4.1 for screening description and 3.4.3 for description of balance assessments. Hz, hertz; VA, visual acuity; MMSE, mini mental state examination; VST, vestibular stepping test; ST, step test; TUG, timed up and go; FR, functional reach; BOOMER, balance outcome measure for elder rehabilitation.

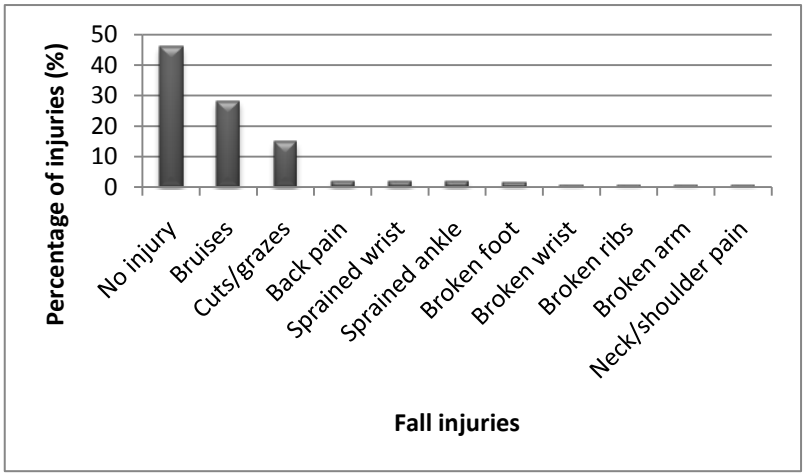
6.3.2 Descriptive statistics of fallers and non fallers

The following section describes fall details and the results from the physical (demographic and screening) and balance measures for the fallers and non-fallers. Differences between non-fallers, single fallers and multiple fallers, and between multiple fallers and a combined group of single and non-fallers were examined separately in Chapter 6.3.9.

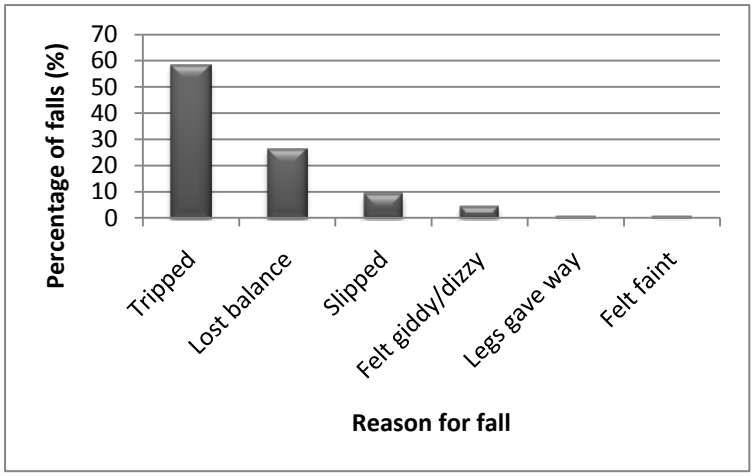
During the subsequent year, 54 participants (55.67%) reported at least one fall, of which 25 (25.77%) fell once and 29 (29.90%) fell more than once. Details regarding fall location, the cause of fall and injuries sustained are listed below in Figure 6.2. Outdoor falls away from the home were more prevalent (50.41%) than either indoor falls (31.40%) or falls occurring around the home (18.18%). The majority of outdoor falls happened on the footpath (21.49%) or in other locations such as on public transport or in parks (10.74%). Most indoor falls happened on the one level (17.36%) or on stairs (7.44%). Falls in the garden accounted for the majority of falls around the home (14.05%). Most falls were reported to be due to tripping (58.46%) or loss of balance (26.15%), and most fallers reported no injury as a result of the fall (46.10%). Of the injuries that were sustained on account of the fall, the most common were bruises (28.37%) and cuts or grazes (14.89%). Only 3.55% of falls resulted in a fractured bone in the sample.



(a)



(b)



(c)

Figure 6.2. Details regarding fall location (a) (n=121), injuries suffered from falls (b) (n=141) and the reason for falling (c) (n=130).

Fallers walked with similar speeds to non-fallers (1.37 cm/s and 1.38 cm/s respectively, p=0.79). Demographic information and results from the screening and balance assessments for the fallers and non-fallers are listed below in Table 6.4 and Table 6.5. Fallers and non-fallers were similar in all physical measures with no significant MANOVA differences on any of the combined test groupings between the two samples (F(15, 81)=0.67, p=0.809; Wilks' Lambda=0.89, partial eta squared=0.110). Although a statistically significant difference was found between fallers and non-fallers on the combined balance measures (F(8, 88)=2.371, p=0.023; Wilks' Lambda=0.823, partial eta squared=0.177), further examination of between-subjects effects revealed no significant differences when each dependent variable was considered separately (p>0.10). To confirm the outcome, independent t tests, with a Bonferroni adjusted significance level of 0.008 (0.05 divided by 6 comparisons), were also performed to individually compare fallers and non-fallers on each of the balance measures. These revealed no significant differences between the groups on any of the measures.

Table 6.4. Demographic information for older fallers and non-fallers (mean ± SD unless otherwise stated).

	Fallers (n=54)	Non-fallers (n=43)
Age (years)	69.00 ± 6.93	68.40 ± 7.31
Height (cm)	161.27 ± 7.06	160.83 ± 5.23
Weight (kg)	70.31 ± 13.89	69.12 ± 18.47
Median number of falls in previous 12 months (± SD, [range])	0 ± 0.66 (0 to 2)	0 ± 0.55 (0 to 2)
Pain during walking (% of sample)	32.41	22.09
Median activity level (times per week, ± SD)	2 ± 0.74	2 ± 0.64
Right handedness (% of sample)	88.89	86.05
Right footedness (% of sample)	87.04	86.05
Median number of medical conditions (range)	5 (1 to 11)	4 (0 to 9)
Most common medical problems (% of sample)		
Osteoarthritis	50	32.56
Hypertension	40.74	34.88
Tonsillectomy	25.93	11.63
Hysterectomy	24.07	20.93
Hyperlipidemia	22.22	32.56
Appendectomy	18.52	4.65

Table 6.4. *Continued*.

	Fallers (n=54)	Non-fallers (n=43)
Median number of medications (range)	2 (0 to 9)	2 (0 to 9)
Most common medications (% of sample)		
Antihypertensive agents	37.04	44.19
Hypolipidemic agents	22.22	32.56
Anti-coagulant, anti-thrombotic	18.52	20.93
Hyperacidity, reflux and ulcers agents	12.96	13.95
Minerals	12.96	11.63
Herbal analgesic and anti-inflammatory	9.26	18.60

Note: See Chapter 3.4.1 for assessment description.

Table 6.5. Screening and balance assessment data for fallers and non-fallers (mean and SD unless otherwise stated).

	Fallers (n=54)	Non-fallers (n=43)
Screening assessments		
Left vibration (Hz, ± SD))	6.21 ± 1.41	5.99 ± 1.62
Right vibration (Hz, ± SD)	6.15 ± 1.68	6.16 ± 1.69
Impaired proprioception (% of sample)	7.41	2.33
Left median VA (range)	0.18 (-0.4 to 0.82)	0.18 (-0.02 to 1)
Right median VA (range)	0.18 (0 to 0.8)	0.26 (0 to 0.8)
Mean contrast sensitivity (± SD)	20.31 ± 2.19	20.23 ± 1.93
Mean MMSE (± SD)	28.04 ± 2.18	27.98 ± 2.11
VST distance (cm, ± SD)	58.11 ± 31.23	69.30 ± 29.39
VST angle (°, ± SD)	21.46 ± 18.40	25.67 ± 30.07
Balance assessments		
Worst ST (number of steps in 15 sec, ± SD)	14.37 ± 4.28	15.07 ± 3.74
TUG (sec, ± SD)	7.98 ± 2.15	7.36 ± 1.56
FR (cm, ± SD)	33.36 ± 7.05	31.67 ± 5.67
Romberg (sec, ± SD)	29.87 ± .95	29.65 ± 2.29
Sharpened Romberg (sec, ± SD)	4.00 ± 1.60	4.24 ± 1.44
BOOMER (% of sample <12)	1.85	0

Note: See Chapter 3.4.1 for screening description and 3.4.3 for description of balance assessments. Hz, hertz; VA, visual acuity; MMSE, mini mental state examination; VST, vestibular stepping test; ST, step test; TUG, timed up and go; FR, functional reach; BOOMER, balance outcome measure for elder rehabilitation.

6.3.3 Test-retest reliability of accelerometer and stride dynamic data

To evaluate the test-retest reliability of the accelerometer and stride dynamic data, a small sub sample of older adults (n=12) were retested on the gait assessment tasks one week following the initial assessment. Sample characteristics of the participants tested in the reliability study are listed in Table 6.6 below.

Table 6.6. Physical (demographic and screening) and balance measures at initial assessment for older fallers participating in the reliability analysis (mean ± SD unless otherwise stated).

	Reliability participants (n=12)
Demographic	
Age (years)	67.17 ± 5.27
Height (cm)	162.73 ± 7.31
Weight (kg)	61.89 ± 22.95
Average walking speed (m/s)	1.46 ± 0.23
Median number of falls in previous 12 months (± SD, [range])	0 ± 0.45 (0 to 1)
Pain during walking (% of sample)	29.17
Median activity level (times per week, ± SD)	2 ± 0.69
Right handedness (% of sample)	91.67
Right footedness (% of sample)	100
Screening assessments	
Left vibration (Hz, ± SD)	5.89 ± 2.68
Right vibration (Hz, ± SD)	6.21 ± 2.42
Impaired proprioception (% of sample)	0

Table 6.6. *Continued*.

	Reliability participants (n=12)
Screening assessments (*Continued*)	
Left median VA (range)	0.15 (0.04 to 0.4)
Right median VA (range)	0.13 (0.08 to 0.4)
Mean contrast sensitivity (± SD)	19.58 ± 2.54
Mean MMSE (± SD)	28.33 ± 1.15
VST distance (cm, ± SD)	56.67 ± 33.80
VST angle (°, ± SD)	23.33 ± 19.69
Balance assessments	
Worst ST (number of steps in 15 sec, ± SD)	15.25 ± 4.75
TUG (sec, ± SD)	7.06 ± 1.36
FR (cm, ± SD)	33.54 ± 6.63
Romberg (sec, ± SD)	28.75 ± 4.33
Sharpened Romberg (sec, ± SD)	3.75 ± 1.46
BOOMER (% of sample <12)	0

Note: See Chapter 3.4.1 for screening description and 3.4.3 for description of balance assessments. Hz, hertz; VA, visual acuity; MMSE, mini mental state examination; VST, vestibular stepping test; ST, step test; TUG, timed up and go; FR, functional reach; BOOMER, balance outcome measure for elder rehabilitation.

Table 6.7 lists the results from the reliability analysis. The stride time fractal scaling index was shown to have excellent test-retest reliability with an $ICC_{3,1}$ of 0.93, SEM of 0.04 and CV of 5.77%. Stride time data derived from acceleration signals recorded during the continuous walking trials were also found to have excellent test-retest reliability based on relative ($ICC_{3,1}$=0.98) and absolute (SEM=0.01 sec and CV=1.06%) measures. Similar results were found when data from the repeated single walking trials were used ($ICC_{3,1}$=0.98, SEM=0.01 sec, CV=0.89%). Although the

mean stride time value was found to significantly differ (p<0.01) between the two testing sessions, the effect size for this difference was small (ES=0.24).

Table 6.7. Results of the reliability analysis for the fractal scaling index and the stride time accelerometer data recorded during continuous and single walking trials.

	Day 1	Day 2	Mean Difference (± SD)	p value	Effect size	SEM	CV (%)	ICC ($_{3,1}$)
α	0.71	0.69	0.02 ± 0.08	0.27	0.17	0.04	5.77	0.93
Continuous ST (sec)	1.02	1.01	0.01 ± 0.02	0.01	0.24	0.01	1.05	0.98
Single ST (sec)	0.96	0.95	0.01 ± 0.02	0.17	0.13	0.01	0.89	0.98

Note: α, fractal scaling index; ST, stride time; SEM, standard error of measurement; CV, coefficient of variation; ICC, intraclass correlation coefficient.

6.3.4 Validity of accelerometer data

Excellent agreement between stride time data recorded from the GAITRite® walkway and Crossbow® accelerometers was found. For the continuous trials, the $ICC_{2,1}$ value of 0.95 was high, whereas the absolute (1.2 msec) and mean (0.37%) repeatability coefficients were very low, indicating excellent agreement between the two systems. Similar results were found for the repeated single trials ($ICC_{2,1}$=0.98, AbsRC=0.34 msec and MeanRC=0.09%). Whilst the mean stride times were found to be significantly different (p<0.01) between the two systems for each walking protocol, this is likely to be a statistical artifact as effect sizes were small (ES=0.22 for the continuous walking data and ES=0.00 for the single trial data).

6.3.5 Differences in intra-limb stride dynamics between fallers and non-fallers

To investigate stride dynamics, the fractal scaling index of stride time, α, was calculated. Long range correlations (α>0.50) were found in all except five participants (two fallers and three non-fallers), and the mean value of the sample (α=0.73) was similar to previously reported values for

healthy older (Hausdorff, Mitchell, et al., 1997) and younger adults (Hausdorff, et al., 1995; Jordan, et al., 2007; Terrier, et al., 2005). Figure 6.3 illustrates a typical log-log plot and calculated fractal scaling index for a non-faller and faller. As can be seen, there is a linear relationship between log *n* and log F(*n*), indicating power-law scaling and therefore the presence of long-range correlations in the stride time data, shown by the fractal scaling exponent, α>0.50.

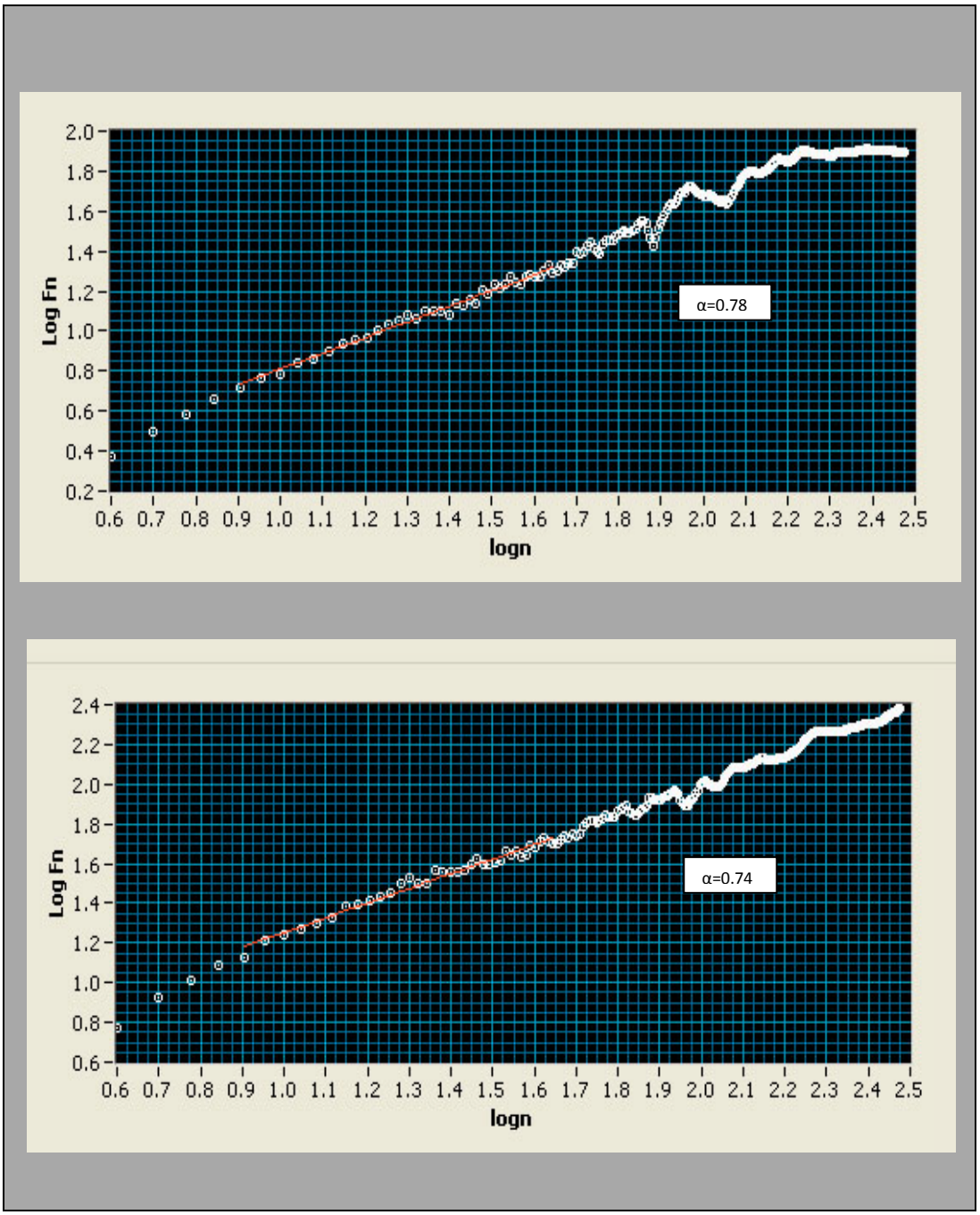


Figure 6.3. Typical log-log plot with scaling exponent for a non-faller (top) and faller (bottom).

To determine whether long range correlations differed between fallers and non-fallers, the right limb α was compared between the two groups using an independent *t* test. No significant differences between fallers (α=0.75, SD=0.11) and non-fallers (α=0.72, SD=0.14) were found (*t*(95)=-1.127, *p*=0.262). Given that the sample of older adults were relatively young and walked with a comparatively fast walking speed, analyses were repeated using only the participants aged over 70 years (N: 38; age: 75.95 ± 4.80 years, height: 1.60 ±0.07 m; mass: 61.23 ± 17.05 kg; average velocity: 1.30 $\pm$ 0.15 cm/s). No significant difference was again found between fallers (α=0.79, SD=0.07) and non-fallers (α=0.75, SD=0.13) using the older cohort (*t*(36)=-1.215, *p*=0.232).

Calculation of α is highly sensitive to the region of line fit on the log-log graph (Hu, et al., 2001). Consequently, to determine whether the lack of difference between the two groups is influenced by the region of line fit, α was also calculated in the region $10 \leq n \leq 20$, consistent with other work investigating stride dynamics in shorter (≤8 minutes) data samples (Hausdorff, Mitchell, et al., 1997; Hausdorff, Zemany, Peng, & Goldberger, 1999; Herman, et al., 2005). Although the difference in α was slightly greater between the two groups (fallers=0.88, non-fallers=0.83), this difference was not statistically significant (*t*(95)=-1.118, *p*=0.266).

6.3.6 Differences in inter-limb stride dynamics between fallers and non-fallers

To investigate inter-limb dynamics and the impact upon falling, a paired samples *t* test was conducted to evaluate left and right α difference in each group (Table 6.8). No differences were found between limbs in either the fallers (*t*(21)=0.4334, *p*=0.669) or non-fallers (*t*(40)=0.851, *p*=0.400). Interestingly, when the inter-limb comparisons were repeated on the participants aged over 70 years, significant left-right differences were found in the scaling exponent of fallers (*t*(21)=3.767, *p*=0.001) but not in non-fallers (*t*(14)=1.064, *p*=0.305). The magnitude of this difference between the left and right scaling values in fallers aged over 70 was large (ES=0.80).

Table 6.8. Left and right fractal scaling index values for the entire sample of fallers and non-fallers and for fallers and non-fallers aged over 70 years.

	Entire sample				Participants aged over 70 years			
	Fallers (n=54)		Non-fallers (n=43)		Fallers (n=23)		Non-fallers (n=15)	
	Left	Right	Left	Right	Left	Right	Left	Right
α	0.73	0.74	0.71	0.72	0.73	0.78*	0.72	0.75

Note: α, fractal scaling index; * indicates significant difference between the left and right limbs (*p*=0.001).

6.3.7 Differences in gait variability between fallers and non-fallers

Figure 6.4 shows the SD values for each gait variable for the fallers and non-fallers. As can be seen, fallers walked with comparable variability magnitudes to non-fallers, with no statistically significant differences noted between the groups on any of the gait variability measures: $F(7, 89)=0.773$, $p=0.612$; Wilks' Lambda=0.943, partial eta squared=0.057.

As the older sample was slightly younger and walked somewhat faster than previous prospective gait variability studies (Hausdorff, Rios, et al., 2001; Maki, 1997; Verghese, et al., 2009), group differences in gait variability were also compared with only those participants over the age of 70 years. Once again, no significant differences were found in the measures of gait variability magnitude between fallers and non-fallers ($F(7, 30)=0.82$, $p=0.562$; Wilks' Lambda=0.836, partial eta squared=0.164). Descriptive statistics for fallers and non-fallers for each of the gait variability measures are listed in Table 6.9 below.

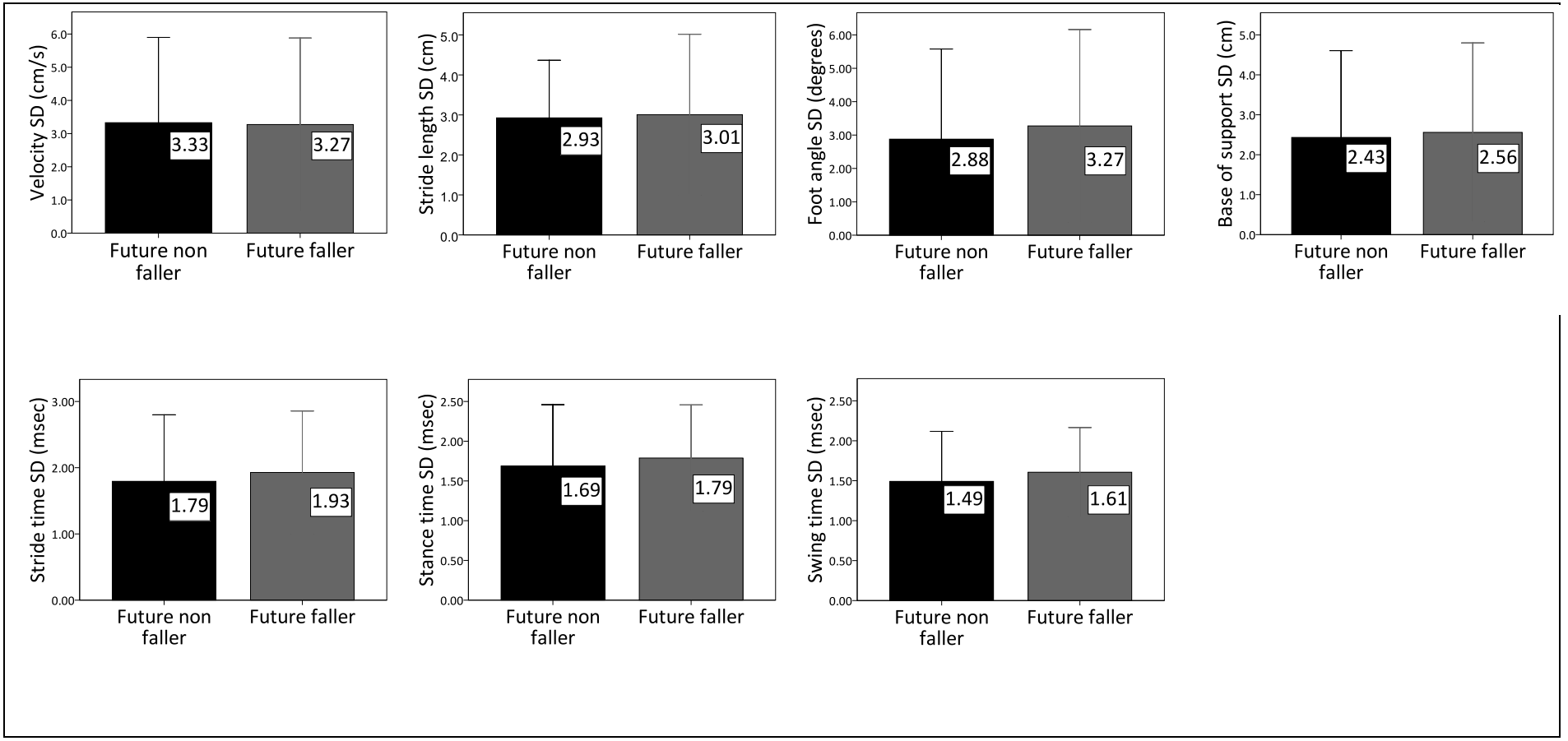


Figure 6.4. Variability (SD) for the fallers and non-fallers for each of the gait parameters. All differences between fallers and non-fallers were not significant.

Table 6.9. Gait variability data (SD) for the entire sample of older fallers and non-fallers and for fallers and non-fallers aged over 70 years.

	Entire sample			Sample aged over 70 years		
	Non-fallers (n=43)	Fallers (n=54)	*p* value	Non-fallers (n=15)	Fallers (n=23)	*p* value
Velocity (cm/s)	3.33 ± 1.29	3.27 ± 1.30	0.83	3.10 ± 1.21	3.20 ± 1.49	0.82
SL (cm)	2.93 ± 0.71	3.01 ± 1.01	0.66	3.00 ± 0.67	3.11 ± 1.15	0.74
Foot angle (°)	2.88 ± 1.35	3.27 ± 1.44	0.17	2.90 ± 1.58	3.31 ± 0.94	0.32
BOS (cm)	2.43 ± 1.09	2.56 ± 1.12	0.58	2.31 ± 0.87	2.88 ± 1.29	0.14
ST (msec)	1.79 ± 0.50	1.93 ± 0.46	0.18	1.89 ± 0.48	2.12 ± 0.51	0.18
StT (msec)	1.69 ± 0.39	1.79 ± 0.33	0.17	1.78 ± 0.40	1.96 ± 0.33	0.15
SwT (msec)	1.49 ± 0.31	1.61 ± 0.28	0.06	1.62 ± 0.41	1.75 ± 0.31	0.27

Note: SL, stride length; BOS, base of support; ST, stride time; StT, stance time; SwT, swing time.

6.3.8 Stride dynamics, gait variability and falls risk

Direct logistic regression was performed to predict future falls using all gait measures, including stride dynamics, gait parameter means and the measures of gait variability. Due to a violation of the assumption of multicollinearity, the gait variables of velocity and stance time were not included in the analysis. Model fit was poor for classification of faller versus non-faller, with Omnibus Tests of Model Coefficients $X^2(11)=9.573$, $p=0.569$, however the Hosmer and Lemeshow Test showed support for the model with $X^2(8)=3.883$, $p=0.868$. The gait model explained between only 9% and 13% of the variance in fall status, with Cox & Snell *R* Square and Nagelkerke *R* Square values of 0.094 and 0.126 respectively. These values would suggest that the model did not contribute well to the prediction of falling in the sample. The overall prediction success of 67% was moderate, with a sensitivity of 74.1% of fallers correctly classified and a specificity of 58.1% of non-fallers correctly predicted. This provided a positive predictive value (percentage of individuals the model classified as falling who actually fell) of 69.0% and a

negative predictive value (percentage of individuals that the model classified as not falling who did not fall) of 64.1%. According to the Wald criterion no gait variable was able to correctly predict group membership as a faller or non-faller. The variables that were closest to achieving significant predictive power are listed in Table 6.10.

Table 6.10. The four gait variables with *p* values closest to significance, determined using logistic regression to predict future falling.

	Wald value	*p* value
SwT SD	3.18	0.08
FA SD	1.42	0.23
Fractal scaling index	1.40	0.24
SL mean	1.24	0.27

Note: SwT, swing time; FA, foot angle; SL, stride length; NA, not applicable.

6.3.9 Influence of fall group stratification

The sample was stratified into non-fallers, single fallers, multiple fallers (two or more falls) and a combined group of single and non-fallers, and all analyses were repeated. Results on the physical (demographic and screening) and balance assessments for each group are presented in Table 6.11 below. Average spatial and temporal gait parameter values for each group are presented in Table 6.12 below. No significant differences ($p>0.05$) were found between groups for any of the variables.

Table 6.11. Physical (demographic and screening) and balance measures at initial assessment for older non-fallers, single fallers, multiple fallers and the combined group of single and non-fallers (mean ± SD unless otherwise stated).

	Non-fallers (n=43)	Single fallers (n=25)	Multiple fallers (n=29)	Single and non-fallers (n=68)
Demographic				
Age (years)	68.40 ± 7.31	70.04 ± 7.25	68.10 ± 6.65	69.00 ± 7.28
Height (cm)	160.83 ± 5.23	161.82 ± 7.38	160.79 ± 6.86	161.19 ± 6.07
Weight (kg)	69.12 ± 18.47	69.84 ± 13.82	70.72 ± 14.18	69.38 ± 16.81
Median number of falls in previous 12 months (± SD, [range])	0 ± 0.55 (0 to 2)	0 ± 0.68 (0 to 2)	0 ± 0.64 (0 to 2)	0 ± 0.59 (0 to 2)
Pain during walking (% of sample)	22.09	32	32.76	25.74
Median activity level (times per week, ± SD)	2 ± 0.64	2 ± 0.81	2 ± 0.67	2 ± 0.71
Right handedness (% of sample)	86.04	84	93.10	85.29
Right footedness (% of sample)	86.05	56	86.21	86.76
Screening assessments				
Left vibration (Hz, ± SD)	5.99 ± 1.62	6.43 ± 1.30	6.02 ± 1.48	6.15 ± 1.52
Right vibration (Hz, ± SD)	6.16 ± 1.69	6.05 ± 1.78	6.23 ± 1.62	6.12 ± 1.71
Impaired proprioception (% of sample)	2.33	8	6.90	4.41
Left median VA (range)	0.18 (-0.02 to 1)	0.12 (-0.04 to 0.68)	0.24 (-0.4 to 0.82)	0.18 (-0.04 to 1)
Right median VA (range)	0.26 (0 to 0.8)	0.18 (0.04 to 0.64)	0.14 (0 to 0.8)	0.21 (0 to 0.8)
Mean contrast sensitivity (± SD)	20.23 ± 1.93	19.84 ± 2.08	20.72 ± 2.23	20.09 ± 1.98
Mean MMSE (± SD)	27.98 ± 2.11	28.00 ± 2.53	28.07 ± 1.87	27.99 ± 2.26

Table 6.11. *Continued*.

	Non-fallers (n=43)	Single fallers (n=25)	Multiple fallers (n=29)	Single and non-fallers (n=68)
Screening assessments (*Continued*)				
VST distance (cm, ± SD)	69.30 ± 29.39	53.60 ± 33.78	62.14 ± 28.78	63.53 ± 31.75
VST angle (°, ± SD)	25.67 ± 30.07	22.08 ± 23.11	20.91 ± 13.30	24.35 ± 27.59
Balance assessments				
Worst ST (number of steps in 15 sec, ± SD)	15.07 ± 3.74	14.4 ± 3.66	14.34 ± 4.81	14.82 ± 3.70
TUG (sec, ± SD)	7.36 ± 1.56	7.77 ± 2.08	8.16 ± 2.23	7.51 ± 1.76
FR (cm, ± SD)	31.67 ± 5.67	33.88 ± 5.99	32.91 ± 7.93	32.49 ± 5.85
Romberg (sec, ± SD)	29.65 ± 2.29	30.00 ± 0.00	29.76 ± 1.30	29.78 ± 1.82
Sharpened Romberg (sec, ± SD)	4.24 ± 1.44	3.81 ± 1.79	4.17 ± 1.43	4.08 ± 1.58
BOOMER (% of sample <12)	0	0	3.45	0

Note: See Chapter 3.4.1 for screening description and 3.4.3 for description of balance assessments. Hz, hertz; VA, visual acuity; MMSE, mini mental state examination; VST, vestibular stepping test; ST, step test; TUG, timed up and go; FR, functional reach; BOOMER, balance outcome measure for elder rehabilitation.

Table 6.12. Mean spatial and temporal gait parameters for non-fallers, single fallers, multiple fallers and the combined group of single and non-fallers.

	Non-fallers (n=43)	Single fallers (n=25)	Multiple fallers (n=29)	Single and non-fallers (n=68)
Velocity (cm/s)	137.61	139.56	133.96	138.33
SL (cm)	136.81	139.17	136.47	137.68
Foot angle (°)	5.57	4.91	5.75	5.33
BOS (cm)	8.69	8.55	8.48	8.64
ST (sec)	1.00	1.00	1.03	1.00
StT (sec)	0.63	0.63	0.64	0.63
SwT (sec)	0.37	0.38	0.39	0.37

Note: SL, stride length; BOS, base of support; ST, stride time; StT, stance time; SwT, swing time.

To determine whether stride dynamics differed between non-fallers (α=0.72, SD=0.14), single fallers (α=0.74, SD=0.13) and multiple fallers (α=0.75, SD=0.09), the right limb fractal scaling index was compared between the three groups using a one-way between groups ANOVA. No significant differences between groups were found ($F(2, 94)=0.726$, $p=0.487$). Outcomes were similar when the fractal scaling index of multiple fallers (α=0.75, SD=0.09) was compared to the combined group of single and non-fallers (α=0.73, SD=0.13) using an independent *t* test ($t(95)=-1.012$, $p=0.314$). There were also no between-group differences when α was calculated in the region $10 \leq n \leq 20$ and when only the participants aged over 70 years were compared (data not shown).

To compare inter-limb dynamics between multiple fallers and the combined group of single and non-fallers, a paired samples *t* test was conducted to evaluate differences in left and right limb α between the two groups (Table 6.13). No differences were found between limbs in the single and non-fallers ($t(62)=0.957$, $p=0.342$), however significant differences were found

between the left and right α of multiple fallers (t(27)=2.146, p=0.041). The magnitude of this difference was moderate (ES=0.44).

Table 6.13. Left and right fractal scaling index values of multiple fallers and the combined group of single and non-fallers for the entire sample and for participants aged over 70 years.

	Entire sample				Participants aged over 70 years			
	Multiple fallers (n=29)		Single or non-fallers (n=68)		Multiple fallers (n=10)		Single or non-fallers (n=28)	
	Left	Right	Left	Right	Left	Right	Left	Right
α	0.71	0.75*	0.72	0.73	0.70	0.77*	0.73	0.76

Note: α, fractal scaling index; * indicates significant difference between the left and right limbs (p=0.04).

Similar to comparisons between the over 70 year old fallers and non-fallers, significant left-right differences were also found in the scaling exponent of multiple fallers aged over 70 years (t(9)=2.420, p=0.039) but not in the combined group of single and non-fallers aged over 70 years (t(26)=2.039, p=0.052) (Table 6.13). Although the magnitude of this difference between the left and right scaling index in multiple fallers was large (ES=0.77), the small sample size of the over 70 year old multiple fallers (n=10) could reduce the statistical power of the finding.

To investigate differences in gait variability between non-fallers, single fallers and multiple fallers, a one-way between subjects MANOVA was conducted (Table 6.14). No statistically significant differences were noted between the groups for any of the gait variability measures: F(14, 176)=0.745, p=0.727; Wilks' Lambda=0.891, partial eta squared=0.056. Additionally, no differences were found for any of the gait variability measures between multiple fallers compared to the combined group of single and non-fallers (F(7, 89)=1.149, p=0.340; Wilks' Lambda=0.917, partial eta squared=0.083). The results were also similar when analyses were repeated using participants aged over 70 years (data not shown).

Table 6.14. Gait variability (SD) for non-fallers, single fallers, multiple fallers and the combined group of single and non-fallers.

	Non-fallers (n=43)	Single fallers (n=25)	Multiple fallers (n=29)	Single and non-fallers (n=68)
Velocity (cm/s)	3.33	2.97	3.53	3.20
SL (cm)	2.92	2.81	3.18	2.88
Foot angle (°)	2.88	3.22	3.32	3.00
BOS (cm)	2.43	2.44	2.66	2.44
ST (msec)	1.79	1.85	1.99	1.82
StT (msec)	1.69	1.73	1.83	1.71
SwT (msec)	1.49	1.55	1.65	1.51

Note: SL, stride length; BOS, base of support; ST, stride time; StT, stance time; SwT, swing time.

Direct logistic regression was repeated using multiple fallers and the combined group of single and non-fallers as the dependent variable and using stride dynamics, the mean gait variables, and the gait variability measures as the predictor variables. Due to a violation of the assumption of multicollinearity, the gait variables of velocity and stance time were not included in the analysis. Similar to the prediction model using the faller and non-faller group membership, model fit was poor based on Omnibus Tests of Model Coefficients [$X^2(11)=8.232$, $p=0.692$], although again the Hosmer and Lemeshow Test results showed support for the model [$X^2(8)=2.547$, $p=0.960$]. Explanation of the variance in fall status by the model was small, with Cox & Snell *R* Square and Nagelkerke *R* Square values of 0.081 and 0.115 respectively. Overall prediction success (73.2%) was marginally improved compared to the non-fallers and fallers stratification. This improvement was predominantly due to a greater sensitivity in correctly classifying members in the group of combined single and non-fallers (97.1%), however the specificity of correctly classifying multiple fallers (17.2%) was much worse than using the non-fallers and fallers stratification. Positive and negative predictive values based on these values were 71.4% and 73.3% respectively. Based upon

the Wald criterion, no gait variable was able to independently predict multiple faller or single/non-faller group membership. The four variables that were closest to significant predictive power are listed in Table 6.15.

Table 6.15. The four gait variables with *p* values closest to significance, determined using logistic regression to predict multiple faller or single/non-faller group membership.

	Wald value	*p* value
SwT SD	1.24	0.27
SL SD	0.72	0.40
Fractal scaling index	0.49	0.49
FA mean	0.06	0.51

Note: SwT, swing time; SL, stride length; FA, foot angle; NA, not applicable.

6.3.10 Influence of walking protocol

To investigate whether differences between groups exist when gait data is collected using an alternate walking protocol, a MANOVA was repeated using variability data recorded from the repeated single walking trials. Similar to the outcomes from the continuous walking trial analyses, no significant differences were found between the gait data of fallers and non-fallers ($F(7, 89)=0.391$, $p=0.905$; Wilks' Lambda=0.970, partial eta squared=0.030), between non-fallers, single fallers and multiple fallers ($F(14, 176)=1.012$, $p=0.443$; Wilks' Lambda=0.856, partial eta squared=0.075), or between multiple fallers and the combined group of single and non-fallers ($F(7, 30)=0.697$, $p=0.674$; Wilks' Lambda=0.860, partial eta squared=0.140) (descriptive data not shown).

Direct logistic regression was again performed using gait data from the repeated single walking trials to determine if the predictive value of the combined gait measures is altered by walking protocol. Due to a violation of the assumption of multicollinearity, the gait variables of velocity and stance time were not included in the analysis. Model fit was poor based on the Omnibus Tests of Model Coefficients [$X^2(10)=3.113$, $p=0.979$], however similar to previous

models, results from the Hosmer and Lemeshow Test suggest support [$X^2(8)=9.952$, $p=0.268$]. The single trial gait model explained less of the variance in fall status compared to the continuous data gait model, with Cox & Snell *R* Square and Nagelkerke *R* Square values of 0.032 and 0.042 respectively. These values would suggest that the model does not meaningfully contribute to the prediction of falling in the sample. This was further supported with an overall prediction success of only 58.8%, with a sensitivity of 79.6% of fallers correctly classified and a specificity of 32.6% of non-fallers correctly predicted. These values provided a positive predictive value of 59.7% and a negative predictive value of 56.0%. Based upon the Wald criterion, no gait variable recorded from the repeated single walking trials was able to independently predict future fallers. The four variables that were closest to significant predictive power are listed in Table 6.16.

Table 6.16. The four variables with *p* values closest to significance, determined using logistic regression to predicting future falling using gait variables from repeated single walking trials.

	Wald value	*p* value
Sw T mean	0.61	0.44
ST SD	0.51	0.48
SL SD	0.33	0.57
FA SD	0.09	0.77

Note: SwT, swing time; ST, stride time; SL, stride length; FA, foot angle; NA, not applicable.

6.4 Discussion

This is the first prospective study to explore stride dynamics and gait variability in active older adult fallers and non-fallers. The aim was to identify an early marker of gait decline in otherwise healthy older adults. Although no significant differences were found in either gait variability magnitude or within-limb stride dynamics between the two groups, or in other measures of physical and balance ability, fallers but not non-fallers aged over 70 years showed evidence of altered inter-limb gait dynamics. Moreover, when adults who experienced two or more falls

were examined, inter-limb differences in gait dynamics were found for the entire sample of fallers, in addition to the subset of fallers aged over 70 years. Effect sizes for these findings were moderate to large suggesting the differences were statistically meaningful.

6.4.1 Stride dynamics and falls

Significant differences were found between the left and right fractal scaling index values in active community dwelling adults aged over 55 years who experienced two or more falls ("multiple fallers") and in active adults aged over 70 years who experienced one or more falls ("fallers"). These differences were not observed in the non-fallers. The fallers, multiple fallers and non-fallers were comprised of well screened and relatively high functioning older adults who were statistically similar on a range of other gait, balance and physical measures. Consequently, these findings could indicate that reduced inter-limb coordination decreases stability, potentially providing a marker of early gait decline and falls risk in an active and otherwise healthy older adult population. The clinical importance of this finding in the absence of other discernable change is strengthened considering the relatively young average age (multiple fallers = 68.10 years, fallers = 68.00) and the comparatively fast average walking speed (multiple fallers = 1.34 cm/s, fallers = 1.37 cm/s) of the sample.

Inter-limb dynamics have not been investigated in older adults however past work has shown that asymmetrical temporal gait parameters predict future older fallers. In a study of 96 community dwelling older adults, Hill and colleagues (1999) reported that single leg stance phase asymmetry increased with age, and was one of only two variables with the strongest prediction accuracy of future falls. Other studies have also shown that older fallers exhibit more asymmetrical swing times than healthy controls during two minutes of normal walking (Yogev, et al., 2007). These authors suggested that walking asymmetry could perturb mediolateral balance, possibly resulting in instability and falls. As gait dynamics have been said to reflect stability within the locomotor system (Hausdorff, 2007), it is possible that reduced coordination of inter-limb

dynamics might decrease stability and subsequently increase falls in older adults.

Previous findings of symmetrical inter-limb coordination in the gait dynamics of healthy young adults would appear to confirm that asymmetries in the stride time scaling exponent of older fallers indicates altered inter-limb gait dynamics. For example, in a single subject design, Taylor and colleagues (2001) reported no significant differences between the left and right foot clearance time scaling exponents during 30 minutes of treadmill walking. More recently, other authors have also reported symmetrical left and right stride time scaling exponents in small samples of healthy young adults tested both as a group (Jordan, et al., 2007) and on a case-by-case basis (Echeverria, Rodriguez, Velasco, & Alvarez-Ramirez, 2010). Similarly, Pierrynowski et al. (2005) reported only minor left (α=0.688) and right (α=0.664) differences in the stride time scaling exponent, which although significant (p=0.04), had low power (0.55) and therefore was likely to be a statistical artifact.

Interestingly, many of these authors reported minor asymmetries in scaling exponents calculated using other gait parameters. For instance, left-right differences were found in the step interval (Jordan, et al., 2007) and swing and stance time (Echeverria, et al., 2010) scaling exponents, although these differences were minor and statistical power or effect size was not reported. Assuming these differences are real and not simply the result of small mean values and high variance, this could suggest the presence of asymmetries in sub phase gait dynamics, whilst the global dynamic output, the stride time scaling index, is symmetrical in healthy populations. This is consistent with other work reporting symmetrical “outcome” or global variables such as stride length and time, and asymmetrical local variables such as joint moments or powers in healthy young adults (Sadeghi, 2003). Of note, inter-limb alterations in these local variables have been shown with healthy ageing (Prince, Sadeghi, Zabjek, & Allard, 2000; Sadeghi, Allard, Prince, & Labelle, 2003; Sadeghi, Prince, Zabjek, & Allard, 2001; Sadeghi, Prince, Zabjek, & Labelle, 2004). It would be of interest to evaluate whether the local sub phase dynamics are also altered in

healthy older adults or in older fallers compared to healthy younger adults.

6.4.2 Gait variability and falls

This study found no differences in gait variability magnitude between fallers and non-fallers. Findings from past work exploring variability differences in fallers and non-fallers are mixed. For example, some retrospective studies have reported that fallers walk with an increase in the variability of stride time (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Hausdorff, Rios, et al., 2001), swing time (Hausdorff, Edelberg, et al., 1997; Hausdorff, Rios, et al., 2001; Springer, et al., 2006), stance time (Hausdorff, Edelberg, et al., 1997), and stride width (Brach, et al., 2005). Conversely, other retrospective studies have shown similar magnitudes of variability between fallers and non-fallers for the gait parameters of step length (Brach, et al., 2005), stance time (Brach, et al., 2005), step time (Brach, et al., 2005; Menz, et al., 2003), and step width (Brach, et al., 2005; Heitman, et al., 1989). Further, although this study found poor predictive ability of gait variability measures, other prospective studies have reported that increased baseline levels of walking variability independently predicted falls in older adults (Hausdorff, Rios, et al., 2001; Maki, 1997; Verghese, et al., 2009).

A likely explanation for the different outcomes of these studies lies in the different characteristics of the samples under investigation. Table 6.17 below displays the sample characteristics and reported outcomes from a number of studies over the past 20 years that have investigated walking variability in older fallers and non-fallers. Apart from the smaller sample sizes and older participants in the studies reporting differences in gait variability between fallers and non-fallers, a major distinction is also the slower walking speed of participants in these studies. The average walking speed in studies reporting increased variability in older fallers was 0.86 m/s, whereas the average speed in the studies where there were no differences between groups was 1.12 m/s. The active community dwelling participants in the present study walked with an average speed of 1.37 m/s.

Table 6.17. Participant characteristics and reported outcomes of studies investigating gait variability in healthy older fallers and non-fallers.

Author	Difference reported	Participant characteristics	Study outcomes
Heitman et al. (1989)	No	• N = 110 older women aged between 60 and 89 years • Age = 73.6 years • Height = NA • Weight = NA • Walking speed = NA • Falls ascertainment = retrospective • Inclusion: able to walk 27 metres without an assistive device, independent in ADLs • Exclusion: institutionalised, primary balance disorder such as Parkinson's disease, multiple sclerosis or stroke	• No difference (*p* value NA) between fallers and non-fallers for step width variability
Hausdorff et al. (1997)	Yes	• N = 35 community dwelling adults aged over 70 years • Age = 82.2 (fallers), 76.5 (non-fallers) years • Height = 1.56 m (fallers), 1.63 m (non-fallers) • Weight = 64.6 kg (fallers), 71.5 kg (non-fallers) • Walking speed = 1.13 m/s (fallers and non-fallers) • Falls ascertainment = retrospective • Inclusion: independent walking for 6 minutes and medically stable • Exclusion: NA	• Fallers had significantly greater variability ($p < 0.001$) in stride time, stance time and swing time compared to non-fallers
Maki (1997)	Yes	• N = 75 older adults aged 62 to 96 years living in self care residencies with on site nursing care • Age = 82.05 years • Height = NA • Weight = 61.05 kg • Walking speed = 0.74 m/s • Falls ascertainment = prospective • Inclusion: able to walk 10 m without walking aid, able to stand unaided, able to understand verbal instruction, no falls in previous month • Exclusion: NA • Other: 29.33% of sample sometimes used a cane, 18.67 walk outside less than once weekly	• Stride length variability (AOR=1.95), stride velocity variability (AOR=2.30) and double support variability (AOR=2.05) predictive of fallers • Between group differences not reported

Table 6.17. *Continued.*

Author	Difference reported	Participant characteristics	Study outcomes
Hausdorff et al. (2001)	Yes	• N = 64 older adults aged over 70 years • Age = 77.7 years • Height = NA • Weight = NA • Walking speed = 0.98 m/s (fallers and non-fallers) • Falls ascertainment = retrospective • Inclusion: at least two functional limitations • Exclusion: unstable cardiovascular disease, psychiatric disorders, neurological or muscular disease, terminal illness, or cognitive impairment	• Multiple fallers had significantly greater variability (p<0.02) in stride time compared to single fallers and non-fallers
Hausdorff et al. (2001)	Yes	• N = 52 older adults aged over 70 years who presented to geriatric clinic • Age = 80.3 years • Height = 1.61 m • Weight = 66.9 kg • Walking speed = 0.84 m/s • Falls ascertainment = prospective • Inclusion: ambulatory • Exclusion: severe cognitive impairment , nursing home residents, or <1 year life expectancy	• Fallers had significantly greater variability in stride time (p<0.04) and swing time (p<0.02) compared to non-fallers • Stride time variability (OR=5.3) and swing time variability (OR=2.2) predictive of fallers
Menz et al. (2003)	No	• N = 100 community dwelling older adults aged between 75 and 93 years • Age = 79.9 years • Height = NA • Weight = NA • Walking speed = 1.16 m/s (low falls risk), 1.08 m/s (moderate risk) and 0.98 m/s (high risk) • Falls ascertainment = risk score • Inclusion: NA • Exclusion: Parkinson's disease or cognitive impairment	• No difference (p value NA) between high, medium or low falls risk groups for step time variability
Brach et al. (2005)	No	• N = 503 older adults aged over 65 years • Age = 79 years • Height = 1.64 m • Weight = 70.26 kg • Walking speed = 1.03 m/s • Falls ascertainment = retrospective • Inclusion: independent ambulation without assistive devices, non-institutionalised • Exclusion: wheelchair-bound in the home, receiving hospice care, radiation therapy or chemotherapy for cancer	• No difference (p>0.06) between fallers and non-fallers for step length, stance time, step width or step time variability

Table 6.17. *Continued*.

Author	Difference reported	Participant characteristics	Study outcomes
Springer et al. (2006)	Yes	• N = 41 older adults aged between 65 and 85 years • Age = 71 (non-fallers), 76.1 (fallers) years • Height = 1.68 m (non-fallers), 1.65 m (fallers) • Weight = 73.3 kg (non-fallers), 70.6 kg (fallers) • Walking speed = 0.77 m/s (non-fallers) and 0.60 m/s (fallers) • Falls ascertainment = retrospective • Inclusion: independent and walk without assistance • Exclusion: cognitive decline, depression or neurological, affective, orthopaedic or other co-morbidities likely to affect gait	• Fallers had significantly greater variability in swing time (p<0.001) compared to non-fallers
Verghese et al. (2009)	Yes	• N = 597 community dwelling older adults aged over 70 years • Age = 80.5 years • Height = NA • Weight = NA • Walking speed = 92.8 m/s • Falls ascertainment = prospective • Inclusion: NA • Exclusion: severe auditory or visual loss, bedbound or institutionalised • Other: sample had an average of 5 medications and 35% reported gait abnormalities, followed for longer than 12 months (mean of 20 months)	• Swing time variability (RR=1.007) and stride length variability (RR=1.076) predictive of fallers • Between group differences not reported
Paterson (2010)	No	• N = 97 community dwelling older women aged between 55 and 90 years • Age = 68.73 years • Height = 1.61 m • Weight = 69.78 kg • Walking speed = 1.37m/s • Falls ascertainment = prospective • Inclusion: Able to walk unassisted and pain free for 10 minutes, no walking aids • Exclusion: Medical conditions known to impact mobility or balance	• No difference (p>0.05) between fallers and non-fallers for stride length, stride time, stance time, swing time, velocity, base of support or foot angle variability • Gait variability not predictive of fallers

Note: Where possible average values for the entire sample have been used; NA, not available/not applicable; ADL, activities of daily living; m/s, metres per second; AOR, adjusted (for fear of falling) odds ratio; OR, odds ratio; RR, relative risk. Shaded rows are prospective studies.

The speed-dependent nature of walking variability has been reported previously for some averaged gait parameters (Kang & Dingwell, 2008) and some authors have suggested that walking speed is a confounding factor for other measures such as walking variability (Moe-Nilssen & Helbostad, 2004; Moe-Nilssen & Helbostad, 2005). However, as explained by Hausdorff (2007), averaged measures (such as mean walking speed) and variability measures (such as the standard deviation of stride time) are first and second moments respectively. As such, these measures are statistically independent, at least theoretically. Hence it is unlikely that the different outcomes in these studies can be explained by walking speed alone. Supporting this, a number of studies have shown associations between walking variability and falls but not between walking speed and falls in older adults (Hausdorff, Edelberg, et al., 1997; Hausdorff, Nelson, et al., 2001; Hausdorff, Rios, et al., 2001; Maki, 1997). Consequently, it is likely that in addition to speed, other factors such as aging, health and underlying pathology contribute to the conflicting findings to a greater degree.

For example, older adults in this study were active, relatively high functioning and were well screened based on strict inclusion and exclusion criteria. Therefore it is likely that participants with sub clinical pathology, manifesting as minor mobility problems, were excluded during the recruitment and screening process. Other prospective variability studies however (shaded rows, Table 6.17) have recruited older adults with a wider range of physical abilities and health outcomes. Indeed, of the three previous prospective gait variability studies, none reported that they excluded participants based upon medical conditions that might affect gait or balance, nor did they exclude participants using an assistive device such as a cane for distances over 10m. In fact, in two of these studies (Maki, 1997; Verghese, et al., 2009), more frail older adults already showing signs of mobility problems were included in the sample (e.g. many participants used a cane, walked outdoors infrequently, reported gait abnormalities and had high average medication use). It is possible therefore, that in each of these studies, the sample comprised older adults with either clinical or sub clinical pathology that could impact upon walking. Given the greater likelihood that that these participants would have been excluded in this study, the

presence of underlying pathology in past studies could partly explain the conflicting outcomes on the ability of gait variability to predict falls in healthy older adults.

Further support for the role of health and underlying pathology confounding the relationship between variability and falls risk comes from recent MRI studies in overtly healthy older adults. Two such studies have shown that increased variability in some gait measures were associated with sub clinical vascular infarctions in the basal ganglia and cerebral white matter (Rosano, Brach, Studenski, Longstreth Jr, & Newman, 2007) and with white matter lesion volume (Srikanth, et al., 2009) in community-dwelling older adults. Interestingly, this latter study also showed that risk of falls associated with white matter lesion volume was higher in adults with greater gait variability. These findings strongly support a link between falling, gait variability and sub clinical pathology.

Consequently, whilst it is of considerable clinical value to predict falls before risk becomes overt, it is generally more difficult to identify such markers in the absence of other clinical signs. This is clearly shown by the lack of predictive power of gait variability in this study whilst past studies using a broader range of older adults have shown differences. It is also supported by the lack of difference in other measures of gait, balance and physical function reported in this study. In this context therefore, the clinical importance of the findings of altered inter-limb gait dynamics is further strengthened.

The high fall rate (55.7%) found in the group of active older adults that participated in this study was unexpected. More active older adults may be at greater risk of a fall simply because of the greater activity level (Cummings & Nevitt, 1994) and exposure to risk. For instance, O'Loughlin and colleagues (O'Loughlin, Robitaille, Boivin, & Suissa, 1993) reported an increased rate of falls to be associated with frequent physical activity (IRR = 2.0) in community dwelling older adults. And although Speechley and Tinetti (Speechley & Tinetti, 1991) reported that vigorous older adults had fewer falls compared to frail community-dwelling older adults, both

Freiberger and Menz (Freiberger & Menz, 2006) and Hill et al (Hill, et al., 1999) found similarly high fall rates in their active older adult samples (42% and 49%). The study by Hill et al also reported that temporal asymmetry in single leg stance was one of only two falls prediction variables (Hill, et al., 1999). Again, these findings lend support to the possible role of asymmetrical temporal coordination increasing instability and falls in active and otherwise healthy adults.

6.4.2.1 Influence of fall group stratification

The sample in this study was a well screened and active older adult population. It is likely therefore that the group of fallers (i.e. participants who had one or more falls) contained a cohort of healthy older adults who experienced a single accidental fall unrelated to intrinsic risk. If gait variability was a true marker of gait decline and falls risk, the inclusion of these participants in the falling group could diminish the ability of variability to differentiate between samples and predict fallers. Additionally, sampling older participants at the healthy to mild falls-risk end of the spectrum could also reduce the likelihood of finding between-group differences due to the homogeneous nature and high functional abilities of the entire cohort. Consequently, the sample of fallers was further stratified into multiple fallers (two or more falls), in which past research has shown there is a greater likelihood that falling is due to the presence of intrinsic risk (Morris, et al., 2004), and the analyses were repeated.

Once again gait variability was found to be similar between multiple fallers, single fallers and non-fallers, and between multiple fallers and a combined group of single and non-fallers. Additionally, the predictive ability of gait variability was also unchanged with this stratification. Therefore, the lack of observed differences, even in multiple fallers where intrinsic factors have been shown to be the major predictors of falls (Morris, et al., 2004), further strengthens the argument that gait variability might not be an early marker of falls risk in active and higher functioning older adults.

6.4.2.2 Influence of walking protocol

It was hypothesised that the different outcomes of studies investigating gait variability in older fallers and non-fallers might also be due in part to the different protocols employed to collect gait variability data. Altered gait variability in older fallers has been reported in a number of studies employing a continuous walking protocol (Hausdorff, Edelberg, et al., 1997; Hausdorff, Rios, et al., 2001; Herman, et al., 2005), whereas some studies employing a repeated single trial walking protocol have reported no differences in gait variability between fallers and non-fallers (Brach, et al., 2005; Heitman, et al., 1989). Therefore, all between-group analyses were also repeated using gait data collected from repeated single walking trials.

Similar to the finding of no difference between fallers and non-fallers when gait variability data were collected using a continuous walking protocol, there were no differences in gait variability data collected using the single trial protocol. Additionally, the gait data collected during the repeated single walking trials had poorer overall falls predictive ability than the continuous gait data. Findings were also similar when the sample was stratified into multiple fallers and a combined group of single and non-fallers. Thus, gait variability data collected during either walking protocol were unable to predict falls in the sample of active older adults. These findings would suggest that factors other than walking protocol contribute to the different findings of studies investigating gait variability and falls. As outlined above, it is likely that functional status and sub clinical pathology influence the association between gait variability and falls in older adults.

6.4.3 Study Limitations

This study was restricted to older females, and as such the findings cannot be extrapolated to older males. Future studies may wish to explore the influence of gender on the relationship between gait dynamics, walking variability and falls. Moreover, the sample was comprised predominantly of higher functioning healthy older adults and therefore does not reflect the wider

older adult population. We intentionally sampled a healthier cohort to eliminate confounding variables such as mobility limitations, and because a major aim of the study was to evaluate the usefulness of gait variability and walking dynamics in explaining why otherwise healthy older adults fall. The goal of this was to identify an early marker of falls risk in an active older adult population. Although it was shown that gait variability cannot predict falls in high functioning older adults, past work has supported the predictive ability of walking variability in other populations (Hausdorff, Rios, et al., 2001; Maki, 1997; Verghese, et al., 2009). As discussed, it is probable that underlying or sub clinical pathology influences the link between gait variability and falls and therefore explains in part the conflicting findings of these studies. Clarifying the degree to which underlying pathology confounds the association between gait variability and falls will aid in clarifying the populations in which gait variability measures have the greatest predictive value.

Finally, although sample size calculations were performed prior to testing to ensure adequate statistical power of the study, it is possible that greater numbers in each group would expose trends that did not reach statistical significance. For example, *p* values of around 0.10 were observed in many of the analyses. Larger sample sizes may lead to significant differences. Moreover, stratification of the group into single fallers (N=25) and into over 70 year old non-fallers (N=15), fallers (N=23), single fallers (N=13), and multiple fallers (N=10) reduced the statistical power of the between-group comparisons. Where statistically significant differences were observed in these groups however, effect sizes were calculated and interpreted based on Cohen's (1988) suggestions to ensure conclusions were robust. Nonetheless, increasing the sample size would ensure group stratification retained greater statistical power.

6.4.4 Conclusions

In conclusion, this study found greater asymmetry in the fractal scaling index of a sample of active community dwelling older women aged over 70 years who fell one or more times, and in women aged over 55 years who fell two or more times during a prospective 12 month follow up

period. Other gait measures, including gait variability, and other measures of physical function and balance were similar between groups. These findings suggest that alterations in inter-limb dynamics might be a sensitive marker of reduced stability and increased falls risk in active and otherwise healthy older adults before other changes are evident.

7 Final discussion

7 Final discussion

7.1 Introduction

This study investigated walking instability in older adults. The major aim was to evaluate whether measures of gait variability and stride dynamics were markers of falls risk in a sample of active community dwelling older women. Firstly, study 1 established the test-retest reliability of a continuous over-ground walking protocol, a common methodology employed in gait variability studies (Chapter 4). Study 2 showed that gait variability is altered by the methodology employed to collect gait data, with increased gait variability found when data were recorded using a repeated single trial protocol (Chapter 5). Finally, study 3 (Chapter 6) confirmed the test-retest reliability and validity of stride data recorded using two tri axial accelerometers. It was then found that inter-limb stride dynamics were altered in older women aged over 70 years who had fallen once or more, and in women aged over 55 who had fallen two or more times in a 12 month prospective period. All other physical, balance, gait variability and stride dynamic measures were similar between the groups investigated. Therefore, measures of inter-limb dynamics might provide a clinically useful marker of gait instability and early falls risk in active community dwelling older women before other changes are apparent.

A general discussion of the findings from each study is provided in the following sections, followed by a synthesis of the major outcomes. A brief final conclusion is then presented.

7.2 Summary of major findings

Study 1: The test-retest reliability of spatiotemporal gait data for young and older women during continuous over-ground walking

A common over-ground protocol employed to collect gait variability and stride dynamic data involves continuous walking, such as back and forth along a walkway or around a walking circuit in a laboratory. However, no study had examined the test-retest reliability of these protocols for

younger or older women. Consequently, study 1 sought to determine whether a continuous over-ground walking protocol was a reliable method of collecting gait data in these two populations. For comparison, the test-retest reliability of the more traditional repeated single walking trial protocol was also examined.

Each walking protocol was found to be reliable in both younger and older adults. However, greater difference between the two testing sessions, or systematic bias, was found in the repeated single walking trial condition, particularly for the older sample. Although the magnitude of the differences was generally small in both walk conditions, the reduced systematic bias for continuous walking protocol suggests this condition was more stable over repeated testing sessions. Therefore small alterations in gait measures such as walking variability may be more readily detected using a continuous walking protocol.

Study 2: Gait variability in younger and older women is altered by over-ground walking protocol

To further explore the association between walking protocol and gait variability, study 2 compared the magnitude of variability of several common gait parameters recorded during a continuous and a repeated single over-ground walking protocol. Most previous studies investigating gait variability in older adults and clinical populations have employed either of these protocols. However, the influence of walking protocol upon measures of variability magnitude was unknown.

The results of the study showed that gait variability significantly differed between the two protocols. In both the younger and older adults, measures of variability magnitude were significantly greater when recorded using a repeated single walking trial protocol. Calculated effect sizes (0.46 to 0.79) suggested the strength of differences were meaningful. It is likely that the frequent stoppages inherent in the repeated single trials perturb longer-term relationships amongst strides, resulting in an increase in gait variability.

Study 3: A prospective study of stride dynamics, gait variability and falls risk in community dwelling older women

The final study investigated the relationship between gait variability, stride dynamics and falls occurring in active and otherwise healthy older women, with the aim of developing an early marker of falls risk in this population. Findings from retrospective studies that explored differences in the magnitude of gait variability in fallers were inconclusive, whereas prospective studies were limited to older adults already showing signs of walking instability. Additionally, a measure of walking dynamics, the stride time fractal scaling index, had not been examined in healthy older fallers, despite previous work in other populations showing potential in identifying future fallers (Frenkel-Toledo, et al., 2005; Herman, Giladi, Gurevich, & Hausdorff, 2005).

Stride time data recorded with two tri-axial accelerometers were firstly validated with protocols developed in studies 1 and 2, following which the test-retest reliability of these data, and the calculated fractal scaling index, were established. The major results of the study showed that measures of gait variability and intra-limb stride dynamics were not predictive of future falls in the sample of healthy older adults. However, alterations were found in the inter-limb dynamics of fallers (one or more falls) aged over 70 years, and in multiple fallers (two or more falls) aged over 55 years. This difference was not observed in the non-fallers for either age group, nor has it been shown in healthy young adults in previous studies. Fallers and non-fallers were similar in all other physical, gait and balance measures. This suggests that alterations in the bilateral control of stride dynamics may decrease walking stability in active older adults. These alterations therefore could be a marker of future falls before changes in other known measures of intrinsic risk of falls are evident.

7.3 Synthesis of major findings

The following section will briefly integrate the major results of each investigation and discuss these outcomes in the context of previous research in the areas of gait variability, stride

dynamics and falls. The clinical implications of these outcomes will also be discussed and suggestions for future research in gait variability and dynamics will be presented.

7.3.1 The influence of walking protocol upon gait and gait variability

In previous studies investigating gait variability, three main walking protocols were employed; a repeated single, a continuous over-ground and a continuous treadmill walking protocol. As outlined in Chapters 2.3.2.2 and 2.3.6 however, differences in many gait parameters (Marsh, et al., 2006; Riley, Paolini, Croce, Paylo, & Kerrigan, 2007; Stolze, et al., 1997), including measures of gait variability (Dingwell, Cusumano, Cavanagh, & Sternad, 2001; Frenkel-Toledo, et al., 2005) have been reported when data recorded on a treadmill were compared to data recorded during over-ground walking. Therefore, a repeated single and a continuous over-ground walking protocol were employed to collect gait variability data in the three studies of this project.

The first two experiments investigated the influence of single and continuous over-ground walking protocols upon measures of gait variability in order to determine the optimal methodology for evaluating falls risk in study 3. A repeated single trial protocol resulted in greater systematic bias (study 1) and an increase in walking variability (study 2) when compared with a continuous over-ground walking protocol. Although past work has theorized that a repeated single trial methodology may not be optimal for evaluating measures of instability such as gait variability (Dingwell, et al., 2001), this was the first study to find that the repeated single trial methodology affected mean gait parameter data and measures of gait variability.

For the single walking trials, greater bias in study 1 was found to be primarily due to an increase in walking speed across the two testing sessions, possibly as a result of greater familiarisation with the protocol in the second session (Hill, Goldie, Baker, & Greenwood, 1994; Hopkins, 2000). Additionally, greater variability in study 2 was attributed to a disruption of inter-stride dependency (Dingwell, et al., 2001; Griffin, West, & West, 2000; Hausdorff, et al., 1996) from the frequent stoppages inherent in the repeated single walking protocol. These findings

could have important clinical implications because alterations in gait variability are often used to indicate a change in the control of locomotion in clinical intervention studies [e.g. (Ainsworth, Lamoth, Polomski, & Houdijk, 2007; Blin, Ferrandez, & Serratrice, 1990; Ebersbach, et al., 1999; Myers, et al., 2009; Öken, Yavuzer, Ergöçen, YorgancIoglu, & Stam, 2008; Papadakis, et al., 2009)]. However, it is apparent from the findings of these studies that a single trial walking protocol alters gait data across testing sessions (study 1) and increases gait variability data (study 2). Walking protocol therefore may potentially affect clinical findings due to these alterations in gait data. In contrast, a continuous over-ground walking protocol was found to be reliable (studies 1 and 3), with less systematic bias (study 1) and reduced walking variability (study 2) than the repeated single protocol. Thus, a continuous over-ground walking protocol might be more stable and more readily detect gait changes such as instability and falls risk. An investigation of this hypothesis was conducted in study 3.

7.3.2 The relationship between walking protocol, gait variability and falls

The major aim of the study 3 was to identify a marker of gait decline and falls risk in otherwise healthy older adults, and to evaluate whether walking protocol influenced the identification of this marker. Variability measures, recorded using both a repeated single and continuous over-ground walking protocol, were compared between a group of active older fallers and non-fallers. Gait variability was not found to differ between the fallers and non-fallers using data collected from either walking protocol. Moreover, it was not found to differ when analyses were repeated using multiple fallers, a population in which intrinsic risk factors have been shown to be the major predictor of falls (Morris, et al., 2004). Consequently, although findings from study 1 and 2 showed walking protocol affects gait data, study 3 showed that the use of either walking methodology did not influence the outcomes of evaluating between-group differences in gait variability in active older fallers and non-fallers.

Minor differences were found in the predictive power of the regression models calculated using gait data recorded from each walking protocol. However, both models yielded weak predictors (study 3). For example, in contrast to the gait model based on data collected during the single walking trials, the continuous data gait model explained slightly greater variance in falls status (between 9% and 13% compared to between 3% and 4%), and resulted in greater overall prediction success (67% compared to 58.8%). Although sensitivity was slightly lower in the continuous gait model compared to the single trial model (74.1% compared to 79.6%), specificity was much greater (58.1% compared to 32.6%) and the positive and negative predictive values were also better (69.0% and 64.1% compared to 59.7% and 56.0% respectively). Despite the small differences between models however, none of the gait predictor variables were found to meaningfully predict (p>0.05) to future falls in older adults independent of the walking protocol. The regression analysis was also repeated using multiple fallers, but again measures of gait variability failed to predict fallers. This shows that gait measures, including variability, are not altered in active community dwelling older adults and hence do not predict future falls in this population.

The inability of the gait variability measures to predict falls goes against the findings of other prospective studies investigating the predictive accuracy of gait variability data recorded using repeated single (Maki, 1997; Verghese, Holtzer, Lipton, & Wang, 2009) and continuous over-ground (Hausdorff, Rios, & Edelberg, 2001) walking protocols. However, further examination of differences in the sample characteristics and screening criteria help explain this contradiction. Based on less stringent selection criteria, previous prospective variability studies sampled older adults aged over 62 or 70 years (average age = 80.95 years) and included participants already showing signs of mobility problems. These criteria resulted in a cohort that included more frail older adults with reduced function. Study 3 strategically recruited and screened active and otherwise healthy older adults aged over 55 years (average age = 68.73 years) in an attempt to identify a marker of early falls risk. Indeed, the average walking speed of participants in this study

was 1.37 m/s, compared with a combined average of 0.84 m/s in the other studies, clearly showing the higher functioning of the participants in the present study. It is likely therefore, that the active and otherwise healthy older fallers and non-fallers in this study were more homogenous and hence walked with similar magnitudes of gait variability, thereby contributing to the lack of predictive success of the gait variability measures.

7.3.3 The relationship between gait dynamics and falls

A novel finding of this study was that fallers aged over 70 years, and multiple fallers aged over 55 years, exhibited reduced coordination in inter-limb dynamics. This is the first time measures of walking dynamics, such as the fractal scaling index, have been studied prospectively in healthy older adults, and the first report of altered gait dynamics in otherwise healthy older fallers. Other studies however have also shown a reduction in other measures of inter-limb temporal coordination in older fallers (Hill, Schwarz, Flicker, & Carroll, 1999; Yogev, Plotnik, Peretz, Giladi, & Hausdorff, 2007). Combined, these findings suggest that control of between-limb timing is intrinsically important to dynamic stability.

The results of this study support previous research showing that measures of gait dynamics produce new information about the neuromotor control of locomotion in clinical populations [e.g. (Frenkel-Toledo, et al., 2005; Hausdorff, et al., 2000; Hausdorff, et al., 1997; Herman, et al., 2005)]. However, this study builds upon previous work investigating gait dynamics and falls risk by showing that a subtle alteration in gait control, such as reduced inter-limb dynamics, may indicate changes in stability in active and otherwise healthy older adults. The clinical relevance of this finding is further strengthened considering the faster walking speed and relatively younger age of the sample compared with participants from previous prospective studies. Moreover, the targeted populations showed no significant changes in other measures of physical, balance or gait function, including gait variability. Thus the presence of reduced inter-limb control in an active group of older fallers prior to change in other intrinsic risk factors, shows that gait

dynamics may be a marker of early decline in locomotor control and stability.

7.4 Future directions

More knowledge about the relationship between gait variability, dynamics and falls in older adults is needed. Firstly, the influence of underlying health on gait variability and falls risk is unknown. The results of this study using an active older cohort, and the contrasting findings of previous prospective studies using more frail older adults, suggest that health and sub clinical pathology influence measures of walking stability such as gait variability. Despite evidence that sub clinical pathology, such as cerebral infarctions, may be a confounding factor when examining variability and falls (Srikanth, et al., 2009), the extent of this association requires clarification. Therefore, in exploring gait changes in higher functioning older adults, future prospective studies should include medical imaging such as MRI to examine the relationship between sub clinical neurological pathology and gait variability. Where this technology is not feasible however, employing detailed neuromotor screening assessments might further inform the link between pathology, gait variability and falls. Determining the nature of this relationship will assist clinicians and researchers in identifying the most suitable populations for which gait variability is a useful predictor of future falls.

Further exploration of gait dynamics in older adults is required, and in particular, the control of inter-limb temporal coordination. The findings from study 3 extend the results of a small body of work hypothesising an association between reduced inter-limb coordination and instability in older adults. However, the influence of this relationship on falls is currently unknown. A greater understanding of neuromotor changes contributing to altered inter-limb dynamics is therefore warranted to aid in explaining this relationship. To this end, imaging technology could be used to explore cerebral laterality in pre-clinical populations using a longitudinal study design. It would also be of value to explore whether there are inter-limb asymmetries present in other biomechanical data associated with falls risk such as foot clearance and ground reaction force

data. Similarly, whilst some research has studied age-related changes in gait dynamics using non-linear methods other than detrended fluctuation analysis (Buzzi, Stergiou, Kurz, Hageman, & Heidel, 2003; Kang & Dingwell, 2009), future work should examine whether the stride dynamic asymmetries found in study 3 remain present when calculated using these other techniques.

Finally, adjusting the fitting region (or window size) to calculate the fractal scaling index did not influence the major outcomes of the study. However, it altered the calculated values. This supports other studies showing the dependence of the index on fitting region and the potentially misleading results from simply calculating an exponent independent of potential shifts in the data (Bartsch, et al., 2007; Hu, Ivanov, Chen, Carpena, & Stanley, 2001; Seely & Macklem, 2004). Indeed, other work in cardiac physiology has identified two distinct fitting regions associated with a short and long-term scaling index (Pikkujamsa, et al., 1999), with some studies showing greater prognostic ability of the short term scaling region (Mäkikallio, et al., 2001; Mäkikallio, et al., 1999; Tapanainen, et al., 2002). Although distinct scaling regions are yet to be investigated in human gait dynamics, some attempts have made at overcoming non-linear scaling behaviour using mono fractal (Bartsch, et al., 2007) and multifractal approaches (West & Scafetta, 2005). To date, these techniques have not been used in healthy older adults or older fallers. It would be of interest to determine whether distinct scaling regions occur in the stride dynamics of older adults and whether investigation of these regions can increase prognostic ability in identifying declines in locomotor function, instability and falls. It would also be of interest to explore whether other non-linear

7.5 Final conclusion

In conclusion, this investigation found evidence that the control of inter-limb stride dynamics is altered in a sample of active and otherwise healthy community dwelling older fallers and multiple fallers. Other markers of physical, balance and gait function were not different between the groups. In particular, measures of gait variability and intra-limb stride dynamics were similar

between fallers and non-fallers, and were not predictive of future falls. Walking protocol did not influence the outcomes despite evidence of alterations in common gait parameters, including measures of variability. These findings have clinical merit given the comparatively young age and relatively high functioning nature of the older adults investigated. The findings suggest that altered inter-limb stride dynamics might provide a marker of early falls risk in the absence of other discernable change. The association between inter-limb coordination, walking instability and falls warrants further investigation.

A1

Appendix 1

Consent form from study 3

(Participant's copy)

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National
Australian Catholic University Limited
ABN 15 050 192 660
Melbourne Campus (St Patrick's)
115 Victoria Parade Fitzroy VIC 3065
Locked Bag 4115 Fitzroy MDC VIC 3065
Telephone 613 9953 3000
Facsimile 613 9953 3005
www.acu.edu.au

CONSENT FORM (PARTICIPANT'S COPY)

TITLE OF PROJECT: Changes in Gait variability and Stride Dynamics with Age

NAMES OF STAFF SUPERVISORS: Professor Geralidine Naughton

NAME OF STUDENT RESEARCHER: Mr Kade Paterson (PhD Candidate)

I ..... have read (*or, where appropriate*, have had read to me*)* and understood the information provided in the Letter to Participants. Any questions I have asked have been answered to my satisfaction. I agree to participate in this one and a half hour walking assessment, including screening tests, and I realise that I can withdraw at any time without comment or penalty, and without affecting my future studies. I agree that research data collected for the study may be published or may be provided to other researchers in a form that does not identify me in any way.

NAME OF PARTICIPANT: ……

(block letters)

SIGNATURE: ..... DATE: .....

SIGNATURE OF PRINCIPAL INVESTIGATOR or SUPERVISOR:

…… DATE: ……..

SIGNATURE OF STUDENT RESEARCHER:

…… DATE: ……..

A2

Appendix 2

Information letter to

participants from study 3

Australian Catholic University
Brisbane Sydney Canberra Ballarat Melbourne

ACU National

Australian atholic University Limited
ABN 15 050 192 660
Melbourne Campus (St Patrick's)
115 Victoria Parade Fitzroy VIC 3065
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Facsimile 613 9953 3005
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INFORMATION LETTER TO PARTICIPANTS

TITLE OF PROJECT: Changes in Gait variability and Stride Dynamics with Age

NAME OF STAFF SUPERVISORS: Professor Geraldine Naughton

NAME OF STUDENT RESEARCHER: Mr Kade Paterson

NAME OF PROGRAMME IN WHICH ENROLLED: Doctor of Philosophy

Dear Participant,
You are invited to take part in a project that will examine your walking pattern on a flat unobstructed walkway. The study will further recent work that has investigated the differences in function between the left and right legs during walking. This research demonstrated that the right leg is used more for pushing off while the left is used for controlling balance and stability. These ideas have not been examined in older adults, for whom a loss of balance is a common problem. This study hopes to establish the effects of aging upon this right-left balance and identify possible links to the high incidence of falling in older persons.

As this study requires you to complete a series of walking trials, risk is deemed minimal. However, if you choose to volunteer for this study you will be required to complete a series of screening questionnaires and tests to ensure the walking trials will not pose any risk to you.

The actual investigation will involve a single visit of approximately one and a half hours, at the Advanced Research Laboratory (room LG62), in the lower ground floor of ACU National in Fitzroy. Initially, screening questionnaires and tests will be administered, which should take approximately 45 minutes. During this time, your medical history and physical measurements (height, weight and leg length) will be taken, pulse and blood pressure recorded, and simple tests assessing your sensation (touch, vibration, sight, and position), muscle strength, balance, range of motion and cognition (mental ability) will be administered. Following this, reflective markers will be placed on the joints of your legs and you will be required to perform one block of 10 walking trials along a flat walkway circuit. This should take approximately 45 minutes. Specially designed cameras will record the movement of the joint markers (you will not be identifiable), and sensors in the walking circuit will measure your walking pattern. You will be able to rest between walks or trials as necessary.

Your participation in this study will help provide valuable information on walking changes with age, and potentially aid in the early detection of those at-risk of falling. You will also be provided with an insight into your own walking pattern. It is anticipated that results from these tests will be published in scientific and/or medical journals. However, all results will be aggregated, and all information you provide shall remain confidential. A coding system will be used to ensure that you cannot be identified. Only the Principal Investigator will know this code, which will be stored in a locked filing cabinet. Upon the completion of the study, the coding sheet will be destroyed.

As a participant in this study, it is important that you understand that you are free to refuse consent altogether without having to justify that decision, or to withdraw consent and discontinue participation in the study at any time without giving a reason. If you are a student, this will not affect your academic progress in any way. You are also able to ask any questions regarding the study and your results at any time, including if you choose to withdraw your participation. Any feedback on the results of the project will be provided to you upon your request.

Any questions you have regarding the study, or any issues raised in this information letter may be directed to the Principal Investigator or Student Researcher.

Professor Geraldine Naughton (Principal Investigator)
Tel: 9953 3034

Mr Kade Paterson (Student Researcher)
Tel: 9953 3552

School of Exercise Science
ACU National
115 Victoria parade Fitzroy Victoria 3065

This study has been approved by the Human Research Ethics Committee at Australian Catholic University. In the event that you have any complaint or concern about the way you have been treated during the study, or if you have any query that the Investigators have not been able to satisfy, you may write to the Chair of the Human Research Ethics Committee care of the address below:

Chair, HREC
C/o Research Services
Australian Catholic University
Melbourne Campus
Locked Bag 4115
FITZROY VIC 3065
Tel: 03 9953 3158
Fax: 03 9953 3315

Any complaint or concern will be treated in confidence and fully investigated. You will be fully informed of the outcome.

If you agree to participate in this project, you should sign both copies of the Consent Form, retain one copy for your records and return the other copy to the Investigator or Student Researcher.

Thank you for your cooperation with this important research

Your sincerely,

Professor Geraldine Naughton
Supervisor

Mr Kade Paterson
Student Researcher

A3 Appendix 3

Falls calendar from study 3

Falls Risk Factors Research Program

Welcome to the research program into falls risk factors coordinated by the School of Exercise Science and the National Ageing Research Institute. This is your falls diary on which to record falls and other important events daily. Please take this diary with you if you are going away on holidays or to hospital.

*Remember that a **fall** is an **accident** (including a slip or trip) where you **lose your balance and part of your body hits the ground, floor or a lower level surface** (e.g. table or chair).*

At the end of each day, please mark the appropriate date with either:

If **no fall** occurred; or

If **a fall** occurred; then complete the **Falls Survey** contained within the Falls Risk factors Research Program folder provided

At the end of each month, **detach the calendar page for that month**, place it in the addressed envelope supplied and **mail it back to us** regardless of you having had any falls or not during the month. Include the **Falls Survey if a fall did occur**. **No postage stamp** is necessary.

Please remember to **fill in your falls diary every day** to enable us to help you.

If you have any questions, please do not hesitate to contact **Kade Paterson** on **(03) 9953 3552** (Mon-Fri 8am – 5:30pm).

Thank you for your assistance.

A4

Appendix 4

Falls survey from study 3

If you experienced a fall, please complete the following survey as soon as possible. At the end of the month, **detach the calendar page for that month**, place it in the addressed envelope supplied **along with this survey** and **mail it back to us**. **No postage stamp** is necessary.

Falls Survey. Date: _____

1. Where have you fallen?		
Inside:	Please tick	
On the one level		
Getting out of bed		
Getting out of a chair		
Using the shower/bath		
Using the toilet		
Walking up or down stairs		
Home entrances or in the garden:		
Walking up or down a step/stairs		
On the one level (e.g. pathway)		
In the garden		
Away from home:		
On the footpath		
On a kerb/gutter		
In a public building		
Getting out of a vehicle		
In another person's home		
Falls not described above (please specify)		
2. How did you fall?		
(Tick more than one if necessary)		
I tripped		
I slipped		
I lost my balance		
My legs gave way		
I felt faint		
I felt giddy/dizzy		
I am not sure		
3. As a result of this fall or falls did you suffer any injuries	Yes	No
If yes, what type of injuries did you suffer?		
Bruises		
Cuts/grazes		
Broken wrist		
Broken hip		
Broken ribs		
Back pain		

Thank you for your assistance.

R References

References

Aaslund, M. K., & Moe-Nilssen, R. (2008). Treadmill walking with body weight support. Effect of treadmill, harness and body weight support system. *Gait & Posture, 28*, 303-308.

Adamo, D., Martin, B., & Brown, S. (2007). Age-related differences in upper limb proprioceptive acuity. *Perceptual Motor Skills, 104*, 1297-1309.

Ainsworth, E., Lamoth, C. J. C., Polomski, W., & Houdijk, H. (2007). Variability and stability of the gait pattern of transfemoral amputees. *Gait & Posture, 26*(1), Poster 93.

Alexander, B. H., Rivara, F. P., & Wolf, M. E. (1992). The cost and frequency of hospitalization for fall-related injuries in older adults. *American Journal of Public Health, 82*(7), 1020-1023.

Alexander, L. D., Black, S. E., Patterson, K. K., Gao, F., Danells, C. J., & McIlroy, W. E. (2009). Association between gait asymmetry and brain lesion location in stroke patients. *Stroke, 40*(2), 537-544.

Altman, D. G., & Bland, J. M. (1996). Detecting skewness from summary information. *British Medical Journal, 313*, 1200.

Amaral, L. A. N., Goldberger, A. L., Ivanov, P. C., & Stanley, H. E. (1998). Scale-independent measures and pathologic cardiac dynamics. *Physical Review Letters, 81*, 2388-2391.

Aminian, K., Rezakhanlou, K., De Andres, E., Fritsch, C., Leyvraz, P., & Robert, P. (1999). Temporal feature estimation during walking using miniature accelerometers: an analysis of gait improvement after hip arthroplasty. *Medical and Biological Engineering and Computing, 37*(6), 686-691.

Anacker, S. L., Di Fabio, R. P., & Horak, F. B. (1992). Influence of sensory inputs on standing balance in community-dwelling elders with a recent history of falling. *Physical Therapy, 72*(8), 575-585.

Andersson, O., & Grillner, S. (1983). Peripheral control of the cat's step cycle. II. Entrainment of the central pattern generators for locomotion by sinusoidal hip movements during "fictive locomotion". *Acta Physiologica Scandinavica, 118*(3), 229-239.

Anstey, K. J., & Low, L. F. (2004). Normal cognitive changes in aging. *Australian Family Physician, 33*(10), 783-787.

Arfken, C. L., Lach, H. W., Birge, S. J., & Miller, J. P. (1994). The prevalence and correlates of fear of falling in elderly persons living in the community. *American Journal of Public Health, 84*(4), 565-570.

Armstrong, D. M. (1988). The supraspinal control of mammalian locomotion. *Journal of Physiology, 405*, 1-37.

Armstrong, T., Bauman, A., & Davies, J. (2000). *Physical activity patterns of Australian adults. Results of the 1999 National Physical Activity Survey* (Vol. AIHW cat. no. CVD 10). Canberra: Australian Institute of Health and Welfare.

Atkinson, G., & Nevill, A. M. (1998). Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Medicine, 26*(4), 217-238.

Australian Bureau of Statistics (2008). Population Projections, Australia, 2006 to 2101, from http://www.abs.gov.au/AUSSTATS/abs@.nsf/mf/3222.0

Avidan, A. Y., Fries, B. E., James, M. L., Szafara, K. L., Wright, G. T., & Chervin, R. D. (2005). Insomnia and hypnotic use, recorded in the minimum data set, as predictors of falls and hip fractures in Michigan nursing homes. *Journal of the American Geriatrics Society, 53*, 955-962.

Baloh, R. W., Ying, S. H., & Jacobson, K. M. (2003). A longitudinal study of gait and balance dysfunction in normal older people. *Archives of Neurology, 60*(6), 835-839.

Baltadjieva, R., Giladi, N., Balash, Y., Herman, T., & Hausdorff, J. (2004). Gait changes in de novo Parkinson's disease patients: A force/rhythm dichotomy. *Movement Disorders, 19*, S138 - S138.

Baltadjieva, R., Giladi, N., Gruendlinger, L., Peretz, C., & Hausdorff, J. M. (2006). Marked alterations in the gait timing and rhythmicity of patients with de novo Parkinson's disease. *European Journal of Neuroscience, 24*, 1815-1820.

Barak, Y., Wagenaar, R., & Holt, K. (2006). Gait characteristics of elderly people with a history of falls: a dynamic approach. *Physical Therapy, 86*(11), 1501-1510.

Barr, E. L., Browning, C., Lord, S. R., Menz, H. B., & Kendig, H. (2005). Foot and leg problems are important determinants of functional status in community dwelling older people. *Disability & Rehabilitation, 27*(16), 917-923.

Barrière, G., Leblond, H., Provencher, J., & Rossignol, S. (2008). Prominent role of the spinal central pattern generator in the recovery of locomotion after partial spinal cord injuries. *The Journal of Neuroscience, 28*(15), 3976-3987.

Bartsch, R., Plotnik, M., Kantelhardt, J. W., Havlin, S., Giladi, N., & Hausdorff, J. M. (2007). Fluctuation and synchronization of gait intervals and gait force profiles distinguish stages of Parkinson's disease. *Physica A: Statistical Mechanics and its Applications, 383*(2), 455-465.

Bath, P. A., & Morgan, K. (1999). Differential risk factor profiles for indoor and outdoor falls in older people living at home in Nottingham, UK. *European Journal of Epidemiology, 15*, 65-73.

Baumert, M., Wessel, N., Schirdewan, A., Voss, A., & Abbott, D. (2007). Scaling characteristics of heart rate time series before the onset of ventricular tachycardia. *Annals of Biomedical Engineering, 35*(2), 201-207.

Baumgartner, T. A., Jackson, A. S., Mahar, M., & Rowe, D. (2006). *Measurement for Evaluation in Physical Education and Exercise Science* (8th ed.). New York: McGraw-Hill.

Beauchet, O., Dubost, V., Herrmann, F., & Kressig, R. (2005). Stride-to-stride variability while backward counting among healthy young adults. *Journal of NeuroEngineering and Rehabilitation, 2*(1), 26.

Beauchet, O., Dubost, V., Herrmann, F., Rabilloud, M., Gonthier, R., & Kressig, R. W. (2005). Relationship between dual-task related gait changes and intrinsic risk factors for falls among transitional frail older adults. *Aging Clinical and Experimental Research, 17*(4), 720-275.

Beckers, F., Verheyden, B., & Aubert, A. E. (2006). Aging and nonlinear heart rate control in a healthy population. *American Journal of Physiology: Heart and Circulatory Physiology, 290*(6), H2560-2570.

Beloozerova, I. N., Sirota, M. G., Swadlow, H. A., Orlovsky, G. N., Popova, L. B., & Deliagina, T. G. (2003). Activity of different classes of neurons of the motor cortex during postural corrections. *Journal of Neuroscience, 23*(21), 7844-7853.

Berg, W. P., Alessio, H. M., Mills, E. M., & Tong, C. (1997). Circumstances and consequences of falls in independent community-dwelling older adults. *Age Ageing, 26*(4), 261-268.

Bergland, A., Jarnlo, G. B., & Laake, K. (2003). Predictors of falls in the elderly by location. *Ageing Clinical Experimental Research, 15*, 43-50.

Besser, M., Selby-Silverstein, L., Oberholzer, J., Welliver, M., Christianson, A., & Carlton, T. (2000). The relationship between temporal-spatial parameters of gait and history of falls in the elderly. *Gait & Posture, 11*(2), 141-142.

Besser, M. P., Kmieczak, K., Schwartz, L., Snyderman, M., Wasko, J., & Selby-Silverstein, L. (1999). Representation of temporal spatial gait parameters using means in adults without impairment. *Gait & Posture, 9*, 113.

Besser, M. P., Selby-Silverstein, L., & Prickett, N. (2001). Predicting fall risk in the elderly using temporal-spatial parameters of gait. *Neurology Report, 25*(4), S3.

Bhala, R. P., O'Donnell, J., & Thoppil, E. (1982). Ptophobia: Phobic fear of falling and its clinical management. *Physical Therapy, 62*(2), 187-190.

Billek-Sawhney, B., & Gay, J. (2005). The functional reach test: are three trials necessary. *Topics in Geriatriatric Rehabilitation, 21*, 144-148.

Bilney, B., Morris, M. E., & Webster, K. (2003). Concurrent related validity of the GAITRite walkway system for quantification of the spatial and temporal parameters of gait. *Gait & Posture, 17*, 68-74.

Blake, A. J., Morgan, K., Bendall, M. J., Dallosso, H., Ebrahim, S. B., Arie, T. H., et al. (1988). Falls by elderly people at home: prevalence and associated factors. *Age Ageing, 17*(6), 365-372.

Bland, B. H. (2004). The power of theta: Providing insights into the role of the hippocampal formation in sensorimotor integration. *Hippocampus, 14*(5), 537-538.

Bland, J. M., & Altman, D. G. (1986). Statistical methods for assessing agreement between two methods of clinical measurement. *Lancet, 1*, 307-310.

Blanke, D. J., & Hageman, P. A. (1989). Comparison of gait of young men and elderly men. *Physical Therapy, 69*, 144-148.

Blin, O., Ferrandez, A., & Serratrice, G. (1990). Quantitative analysis of gait in Parkinson patients: increased variability of stride length. *Journal of Neurological Sciences, 98*, 91 - 97.

Booth, M. L., Owen, N., Bauman, A., Clavisi, O., & Leslie, E. (2000). Social-cognitive and perceived environment influences associated with physical activity in older Australians. *Preventive Medicine, 31*(1), 15-22.

Brach, J. S., Berlin, J. E., VanSwearingen, J. M., Newman, A. B., & Studenski, S. A. (2005). Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. *Journal of Neuroengineering Rehabilitation, 2*, 21.

Brach, J. S., Berthold, R., Craik, R., VanSwearingen, J. M., & Newman, A. B. (2001). Gait variability in community-dwelling older adults. *Journal of the American Geriatrics Society, 49*(12), 1646-1650.

Brach, J. S., Studenski, S., Perera, S., VanSwearingen, J. M., & Newman, A. B. (2008). Stance time and step width variability have unique contributing impairments in older persons. *Gait & Posture, 27*(3), 431-439.

Brach, J. S., Studenski, S. A., Perera, S., VanSwearingen, J. M., & Newman, A. B. (2007). Gait variability and the risk of incident mobility disability in community-dwelling older adults. *Journal of Gerontology Series A: Biological Sciences and Medical Sciences, 62*(9), 983-988.

Bradley, C., & Harrison, J. E. (2007). Hospitalisations due to falls in older people, Australia, 2003-04. Injury research and statistics series number 32 *AIHW cat. no. INJCAT 96*, Adelaide: Australian Institite of Health and Welfare.

Bradley, C., & Pointer, s. (2008). Hospitalisations due to falls by older people, Australia 2005-06. Injury research and statistics series number 50. *Cat. no. INJCAT 122*, Adelaide: AIHW

Brandt, T., & Dieterich, M. (1993). Vestibular falls. *Journal of Vestibular Research, 3*, 3-14.

Briggs, R. C., Gossman, M. R., Birch, R., Drews, J. E., & Shaddeau, S. A. (1989). Balance performance among noninstitutionalized elderly women. *Physical Therapy, 69*(9), 748-756.

Brocard, F., & Dubuc, R. (2003). Differential contribution of reticulospinal cells to the control of locomotion induced by the mesencephalic locomotor region. *Journal of Neurophysiology, 90*, 1714-1727.

Brown, A. P. (1999). Reducing falls in elderly people: A review of exercise interventions. *Physiotherapy Theory & Practice, 15*(2), 59-68.

Brown, J. S., Vittinghoff, E., Wyman, J. F., Stone, K. L., Nevitt, M. C., Ensrud, K. E., et al. (2000). Urinary incontinence: does it increase risk of falls and fractures? Study of Osteoporosis Research Group. *Journal of the American Geriatrics Society, 48*, 721-725.

Bruce, D. G., Devine, A., & Prince, R. L. (2002). Recreational physical activity levels in healthy older women: The importance of fear of falling. *Journal of the American Geriatrics Society, 50*(1), 84-89.

Brudzynski, S. M., Wu, M., & Mogenson, G. J. (1993). Decreases in rat locomotor activity as a result of changes in synaptic transmission to neurons within the mesencephalic locomotor region. *Canadian Journal of Physiology and Pharmacology, 71*(5-6), 394-406.

Buchman, A. S., Wilson, R. S., Leurgans, S., & Bennett, D. A. (2009). Vibratory thresholds and mobility in older persons. *Muscle & Nerve, 39*(6), 754-760.

Buzzi, U. H., Stergiou, N., Kurz, M. J., Hageman, P. A., & Heidel, J. (2003). Nonlinear dynamics indicates aging affects variability during gait. *Clinical Biomechanics, 18*(5), 435-443.

Callisaya, M. L., Blizzard, L., Schmidt, M. D., McGinley, J. L., Lord, S. R., & Srikanth, V. K. (2009). A population-based study of sensorimotor factors affecting gait in older people. *Age Ageing, 38*(3), 290-295.

Camicioli, R., Moore, M. M., Sexton, G., Howieson, D. B., & Kaye, J. A. (1999). Age-related brain changes associated with motor function in healthy older people. *Journal of the American Geriatrics Society, 47*(3), 330-334.

Campbell, A. J., Robertson, M. C., Gardner, M. M., Norton, R. N., & Buchner, D. M. (1999). Psychotropic medication withdrawal and a home-based exercise program to prevent falls: A randomized, controlled trial. *Journal of the American Geriatrics Society, 47*(7), 850-853.

Carter, S. E., Campbell, E. M., Sanson-Fisher, R. W., Redman, S., & Gillespie, W. J. (1997). Environmental hazards in the homes of older people. *Age Ageing, 26*(3), 195-202.

Chang, M. D., Shaikh, S., & Chau, T. (2009). Effect of treadmill walking on the stride interval dynamics of human gait. *Gait & Posture, 30*(4), 431-435.

Chen, Z., Ivanov, P., Hu, K., & Stanley, H. (2002). Effect of nonstationarities on detrended fluctuation analysis. *Physical Review E: Statistical, Nonlinear and Soft Matter Physics, 65*(4 pt 1), 041107.

Chu, L.-W., Chiu, A. Y. Y., & Chi, I. (2006). Impact of falls on the balance, gait, and activities of daily living functioning in community-dwelling Chinese older adults. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 61*(4), 399-404.

Chung, S. H., & Giuliani, C. A. (1997). Within- and between-session consistency of electromyographic temporal patterns of walking in non-disabled older adults. *Gait & Posture, 6*(2), 110-118.

Clough-Gorr, K. M., Erpen, T., Gillmann, T., von Renteln-Kruse, W., Iliffe, S., Beck, J., et al. (2008). Preclinical disability as a risk factor for falls in community-dwelling older adults. *The Journals of Gerontology, 63A*(3), 314-320.

Coffey, C. E., Wilkinson, W. E., Parashos, L., Soady, S. A. R., Sullivan, R. J., Patterson, L. J., et al. (1992). Quantitative cerebral anatomy of the aging human brain: A cross-sectional study using magnetic resonance imaging. *Neurology, 42*(3), 527-536.

Cohen, H., Blatchly, C. A., & Gombash, L. L. (1993). A study of the clinical test of sensory interaction and balance. *Physical Therapy, 73*, 346-351.

Cohen, H. S. (2000). Vestibular disorders and impaired path integration along a linear trajectory. *Journal of Vestibular Research, 10*(1), 7-15.

Cohen, J. (1988). *Statistical power analysis for the behavioral sciences* (2nd ed.). Hillsdale, NJ: Lawrence Earlbaum Associates.

Condron, J. E., & Hill, K. D. (2002). Reliability and validity of a dual-task force platform assessment of balance performance: effects of age, balance impairment, and cognitive task. *Journal of the American Geriatrics Society, 50*, 157-162.

Connell, B. R. (1996). Role of the environment in falls prevention. *Clinics in Geriatric Medicine, 12*(4), 859-880.

Conway, B. A., Hultborn, H., & Kiehn, O. (1987). Proprioceptive input resets central locomotor rhythm in the spinal cat. *Experimental Brain Research, 68*(3), 643-656.

Conway, B. A., Scott, D. T., Riddle, J. S., & Hadian, M. R. (1994). Effects of plantar nerve stimulation on the transmission of late flexion reflexes in the decerebrate spinal cat. *Journal of Physiology, 479*(145P).

Coren, S. (1993). The lateral preference inventory for measurement of handedness, footedness, eyedness, and earedness: Norms for young adults. *Bulletin of the Psychonomic Society, 31*(1), 1-3.

Coren, S., Porac, C., & Duncan, P. (1979). A behaviorally validated self-report inventory to assess four types of lateral preference. *Journal of Clinical and Experimental Neuropsychology, 1*(1), 55-64.

Cotter, P. E., Timmons, S., O'Connor, M., Twomey, C., & O'Mahony, D. (2006). The financial implications of falls in older people for an acute hospital. *Irish Journal of Medical Science, 175*(2), 11-13.

Craik, R. L., & Dutterer, L. (1995). Spatial and Temporal Characteristics of Foot Fall Patterns. In R. L. Craik & C. A. Otis (Eds.), *Gait Analysis: Theory and Applications*. St Louis: Mosby Year Book.

Cripps, R., & Carman, J. (2001). Falls by the elderly in Australia: Trends and data for 1998. *Injury Research and Statistics Series: Adelaide: Australian Institute of Health and Welfare*, (AIHW cat no. INJCAT 35).

Crompton, S., Khemlani, M., Batty, J., Ada, L., Dean, C., & Katrak, P. (2001). Practical issues in retraining walking in severely disabled patients using treadmill and harness support systems. *Australian Journal of Physiotherapy, 47*(3), 211-213.

Crum, R. M., Anthony, J. C., Bassett, S. S., & Folstein, M. F. (1993). Population-based norms for the Mini Mental State Examination by age and eductional level. *Journal of the American Medical Association, 12*, 2386-2391.

Cumming, R. G., Salkeld, G., Thomas, M., & Szonyi, G. (2000). Prospective study of the impact of fear of falling on activities of daily living, SF-36 scores, and nursing home admission. *The Journals of Gerontology, 55A*(5), M299-M305.

Cummings, S. R., Nevitt, M. C., Browner, W. S., Stone, K., Fox, K. M., Ensrud, K. E., et al. (1995). Risk factors for hip fracture in white women. *New England Journal of Medicine, 332*(12), 767-774.

Cummings, S. R., & Nevitt, M. C. (1994). Falls. *New England Journal of Medicine, 331*, 872-873.Cummings, S. R., Rubin, S. M., & Black, D. (1990). The future of hip fractures in the United States. Numbers, costs, and potential effects of postmenopausal estrogen. *Clinical Orthopaedics and Related Research, 252*, 163-166.

Cutlip, R., Mancinelli, C., Huber, F., & DiPasquale, J. (2000). Evaluation of an instrumented walkway for measurement of the kinematic parameters of gait. *Gait & Posture, 12*(2), 134 - 138.

Dargent-Molina, P., Favier, F., Grandjean, H., Baudoin, C., Schott, A. M., Hausherr, E., et al. (1996). Fall-related factors and risk of hip fracture: the EPIDOS prospective study. *The Lancet, 348*(9021), 145-149.

de Boer, M. R., Pluijm, S. M., Lips, P., Moll, A. C., Vöker-Dieben, H. J., Deeg, D. J., et al. (2004). Different aspects of visual impairment as risk factors for falls and fractures in older men and women. *Journal of Bone and Mineral Research, 19*(9), 1539-1547.

Dellinger, A. M., & Stevens, J. A. (2006). The injury problem among older adults: Mortality, morbidity and costs. *Journal of Safety Research, 37*(5), 519-522.

Deshpande, N., Ferrucci, L., Metter, J., Faulkner, K. A., Strotmeyer, E., Satterfield, S., et al. (2008). Association of lower limb cutaneous sensitivity with gait speed in the elderly: The Health ABC Study. *American Journal of Physical Medicine & Rehabilitation, 87*(11), 921-928.

Deshpande, N., & Patla, A. (2007). Visual–vestibular interaction during goal directed locomotion: effects of aging and blurring vision. *Experimental Brain Research, 176*(1), 43-53.

DeVita, P., & Hortobágyi, T. (2000). Age causes a redistribution of joint torques and powers during gait. *Journal of Applied Physiology, 88*, 1804-1811.

Di Fabio, R. P., & Anacker, S. L. (1996). Identifying fallers in community living elders using a clinical test of sensory interaction for balance. *European journal of physical medicine & rehabilitation, 6*(2), 61-66.

Dietz, V. (2003). Spinal cord pattern generators for locomotion. *Clinical Neurophysiology, 114*, 1379-1389.

Dingwell, J. B., & Cavanagh, P. R. (2001). Increased variability of continuous overground walking in neuropathic patients is only indirectly related to sensory loss. *Gait & Posture, 14*(1), 1-10.

Dingwell, J. B., Cusumano, J. P., Cavanagh, P. R., & Sternad, D. (2001). Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. *Journal of Biomechanical Engineering, 123*, 27-32.

Dingwell, J. B., Ulbrecht, J. S., Boch, J., Becker, M. B., O'Gorman, J. T., & Cavanagh, P. R. (1999). Neuropathic gait shows only trends towards increased variability of sagittal plane kinematics during treadmill locomotion. *Gait & Posture, 10*(1), 21-29.

Djaldetti, R., Lorberboym, M., & Melamed, E. (2006). Primary postural instability: a cause of recurrent sudden falls in the elderly. *Neurological Sciences: Official Journal Of The Italian Neurological Society And Of The Italian Society Of Clinical Neurophysiology, 27*(6), 412-416.

Dolinis, J., Harrison, J. E., & Andrews, G. R. (1997). Factors associated with falling in older Adelaide residents. *Australian and New Zealand Journal of Public Health, 21*(5), 462-468.

Drew, T., Jiang, W., Kably, B., & Lavoie, S. (1996). Role of the motor cortex in the control of visually triggered gait modifications. *Canadian Journal of Physiology and Pharmacology, 74*(4), 426-442.

Drew, T., Prentice, S., & Schepens, B. (2004). Cortical and brainstem control of locomotion. *Progress in Brain Research, 143*, 251-261.

Dubost, V., Annweiler, C., Aminian, K., Najafi, B., Herrmann, F. R., & Beauchet, O. (2008). Stride-to-stride variability while enumerating animal names among healthy young adults: Result of stride velocity or effect of attention-demanding task? *Gait & Posture, 27*(1), 138-143.

Dubost, V., Kressig, R. W., Gonthier, R., Herrmann, F. R., Aminian, K., Najafi, B., et al. (2006). Relationships between dual-task related changes in stride velocity and stride time variability in healthy older adults. *Human Movement Science, 25*(3), 372-382.

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

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

Duysens, J. (1977a). Fluctuations in sensitivity to rhythm resetting effects during the cat's step cycle. *Brain Research, 133*, 190-195.

Duysens, J. (1977b). Reflex control of locomotion as revealed by stimulation of cutaneous afferents in spontaneously walking premammillary cats. *Journal of Neurophysiology, 40*(4), 737-751.

Duysens, J., Clarac, F., & Cruse, H. (2000). Load-regulating mechanisms in gait and posture: comparative aspects. *Physiological Reviews, 80*(1), 83-133.

Duysens, J., & Pearson, K. G. (1976). The role of cutaneous afferents from the distal hindlimb in the regulation of the step cycle of thalamic cats. *Experimental Brain Research, 24*, 245–255.

Duysens, J., & Pearson, K. G. (1980). Inhibition of flexor burst generation by loading ankle extensor muscles in walking cats. *Brain Research, 187*(2), 321-332.

Duysens, J., & Van de Crommert, H. W. A. A. (1998). Neural control of locomotion; part 1: the central pattern generator from cats to humans. *Gait & Posture, 7*, 131-141.

Ebersbach, G., Sojer, M., Valldeoriola, F., Wissel, J., Müller, J., Tolosa, E., et al. (1999). Comparative analysis of gait in Parkinson's disease, cerebellar ataxia and subcortical arteriosclerotic encephalopathy. *Brain, 122*(7), 1349-1355.

Echeverria, J. C., Rodriguez, E., Velasco, A., & Alvarez-Ramirez, J. (2010). Limb dominance changes in walking evolution explored by asymmetric correlations in gait dynamics. *Physica A: Statistical Mechanics and its Applications, 389*(8), 1625-1634.

Ekstrom, A. D., Kahana, M. J., Caplan, J. B., Fields, T. A., Isham, E. A., Newman, E. L., et al. (2003). Cellular networks underlying human spatial navigation. *Nature, 425*, 184-188.

Elble, R. J. (2007). Gait and dementia: moving beyond the notion of gait apraxia. *Journal of Neural Transmission, 114*(10), 1253-1258.

ElHaber, N., Hill, K. D., Cassano, A. T., Paton, L. M., MacInnis, R. J., Cui, J. S., et al. (2006). Genetic and environmental influences on variation in balance performance among female twin pairs aged 21-82 years *American Journal of Epidemiology, 164*(3), 246-256.

Eng, J. J., & Winter, D. A. (1995). Kinetic analysis of the lower limbs during walking: What information can be gained from a three-dimensional model? *Journal of Biomechanics, 28*(6), 753-758.

Englander, F., Hodson, T. J., & Terregrossa, R. A. (1996). Economic dimensions of slip and fall injuries. *Journal of Forensic Science, 41*(5), 733-746.

Ensrud, K. E., Blackwell, T. L., Mangione, C. M., Bowman, P. J., Whooley, M. A., Bauer, D. C., et al. (2002). Central nervous system-active medications and risk for falls in older women. *Journal of the American Geriatrics Society, 50*, 1629-1637.

Evans, M. D., Goldie, P. A., & Hill, K. D. (1997). Systematic and random error in repeated measurements of temporal and distance parameters of gait after stroke. *Archives of Physical Medicine and Rehabilitation, 78*, 725-729.

Fagerstedt, P., Orlovsky, G. N., Deliagina, T. G., Grillner, S., & Ullén, F. (2001). Lateral turns in the lamprey. II. activity of reticulospinal neurons during the generation of fictive turns. *Journal of Neurophysiology, 86*(5), 2257-2265.

Felson, D. T., Anderson, J. J., & Annan, M. T. (1989). Impaired vision and hip fracture. The Framingham Study. *Journal of the American Geriatrics Society, 37*, 495-500.

Finley, F. R., Cody, K. A., & Finizie, R. V. (1969). Locomotion patterns in elderly women. *Archives of Physical Medicine and Rehabilitation, 50*(3), 140-146.

Folstein, M. F., Folstein, S. E., & McHugh, P. R. (1975). Mini-mental state: A practical method for grading the cognitive state of patients for the clinician. *Journal of Psychiatric Research, 12*, 189-198.

Franchignoni, F., Tesio, L., Martino, M. T., & Ricupero, C. (1998). Reliability of four simple, quantitative tests of balance and mobility in healthy elderly females. *Ageing (Milano), 10*(1), 26-31.

Frenkel-Toledo, S., Giladi, N., Peretz, C., Herman, T., Gruendlinger, L., & Hausdorff, J. M. (2005). Treadmill walking as an external pacemaker to improve gait rhythm and stability in Parkinson's disease. *Movement Disorders, 20*(9), 1109-1114.

Freiberger, E., & Menz, H. B. (2006). Characteristics of falls in physically active community-dwelling older people. *Zeitschrift fur Gerontologie und Geriatrie, 39*(4), 261-267.

Fukuyama, H., Ouchi, Y., Matsuzaki, S., Nagahama, Y., Yamauchi, H., Ogawa, M., et al. (1997). Brain functional activity during gait in normal subjects: a SPECT study. *Neuroscience Letters, 228*(3), 183-186.

Gabell, A., & Nayak, U. S. L. (1984). The effect of age on variability in gait. *Journal of Gerontology, 36*(6), 662-666.

GAITRite Operating Manual (CIR Systems Inc.). Havertown, PA.

Garcia-Rill, E. (1986). The basal ganglia and the locomotor regions. *Brain Research, 396*(1), 47-63.

Garcia-Rill, E., & Skinner, R. D. (1987). The mesencephalic locomotor region. II. Projections to reticulospinal neurons. *Brain Research, 411*(1), 13-20.

Gates, D. H., & Dingwell, J. B. (2007). Peripheral neuropathy does not alter the fractal dynamics of stride intervals of gait. *Journal of Applied Physiology, 102*(3), 965-971.

Gibson, S. J., & Farrell, M. (2004). A review of age differences in the neurophysiology of nociception and the perceptual experience of pain. *The Clinical Journal of Pain, 20*(4), 227-239.

Gielen, S., van Bolhuis, B., & Vrijenhoek, E. (1998). On the Number of Degrees of Freedom in Biological Limbs. In M. L. Latash (Ed.), *Progress in Motor control Volume One: Bernstein's Traditions in Movement Studies*. Champaign, Il: Human Kinetics.

Gill, T. M., Williams, C. S., Robison, J. T., & Tinetti, M. E. (1999). A population-based study of environmental hazards in the homes of older persons. *American Journal of Public Health, 89*(4), 553-556.

Goldberger, A. L., Amaral, L. A. N., Glass, L., Hausdorff, J. M., Ivanov, P. C., Mark, R. G., et al. (2000). PhysioBank, PhysioToolkit and PhysioNet: Components of a new research resource for complex physiologic signals. *Circulation, 101*(23), e215-e220.

Goldberger, A. L., Amaral, L. A. N., Hausdorff, J. M., Ivanov, P. C., Peng, C.-K., & Stanley, H. E. (2002). Fractal dynamics in physiology: Alterations with disease and aging. *Proceedings of the National Academy of Sciences of the United States of America, 99*(Suppl 1), 2466-2472.

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

Grabiner, P. C., Biswas, S. T., & Grabiner, M. D. (2001). Age-related changes in spatial and temporal gait variables. *Archives of Physical Medicine and Rehabilitation, 82*(1), 31-35.

Graham Brown, T. (1911). The intrinsic factors in the act of progression in the mammal. *Proceedings of the Royal Society of London Series B, 84*, 308-319.

Graham Brown, T. (1912). The factors in rhythmic activity of the nervous system. *Proceedings of the Royal Society of London Series B, 85*, 278-289.

Graham Brown, T. (1914). On the nature of the fundamental activity of the nervous centres; together with an analysis of the conditioning of rhythmic activity in progression, and a theory of the evolution of function in the nervous system. *Journal of Physiology, 48*(1), 18-46.

Graybiel, A, & Fregly, A. R. (1966). A new quantitative ataxia test battery. *Acta Oto-Laryngologica (Stockholm), 61*, 292-312.

Gretz, H. R., Doering, L. L., Quinn, J., Raftopolous, M., Nelson, A. J., & Zwick, D. E. (1998). Functional ambulation performance testing of adults with Down Syndrome. *Neurorehabilitation, 11*(3), 211-225.

Griffin, L., West, D. J., & West, B. J. (2000). Random stride intervals with memory. *Journal of Biological Physics, 26*, 185-202.

Grillner, S., Hellgren, J., Ménard, A., Saitoh, K., & Wikström, M. A. (2005). Mechanisms for selection of basic motor programs – roles for the striatum and pallidum. *Trends in Neurosciences, 28*(7), 364-370.

Grillner, S., & Rossignol, S. (1978). On the initiation of the swing phase of locomotion in chronic spinal cats. *Brain Research, 146*(2), 267-277.

Grillner, S., & Wallén, P. (1985). Central pattern generators for locomotion, with special reference to vertebrates. *Annual Review of Neuroscience, 8*, 233-261.

Grillner, S., Wallén, P., Saitoh, K., Kozlov, A., & Robertson, B. (2008). Neural bases of goal-directed locomotion in vertebrates—An overview. *Brain Research Reviews, 57*, 2-12.

Grillner, S., & Zangger, P. (1979). On the central generation of locomotion in the low spinal cat. *Experimental Brain Research, 34*(2), 241-261.

Guertin, P., Angel, M. J., Perreault, M. C., & McCrea, D. A. (1995). Ankle extensor group I afferents excite extensors throughout the hindlimb during fictive locomotion in the cat. *Journal of Physiology, 487*(1), 197-209.

Guimaraes, R. M., & Isaacs, B. (1980). Characteristics of the gait in old people who fall. *International Rehabilitation Medicine, 2*(4), 177-180.

Guttmann, C. R. G., Benson, R., Warfield, S. K., Wei, X., Anderson, M. C., Hall, C. B., et al. (2000). White matter abnormalities in mobility-impaired older persons. *Neurology, 54*(6), 1277-1283.

Hageman, P. A., & Blanke, D. J. (1986). Comparison of gait of young women and elderly women. *Physical Therapy, 66*(9), 1382-1387.

Haines, T., Kuys, S. S., Morrsison, G., Clarke, J., Bew, P., & McPail, S. (2007). Development and validation of the Balance Outcome Measure for Elder Rehabilitation. *Archives of Physical Medicine and Rehabilitation, 88*, 1614-1621.

Hamill, J., & Knutzen, K. M. (2003). *Biomechanical Basis of Human Movement*. PA: Lippincott, Williams & Wilkins.

Hanakawa, T., Fukuyama, H., Katsumi, Y., Honda, M., & Shibasaki, H. (1999). Enhanced lateral premotor activity during paradoxical gait in Parkinson's disease. *Annals of Neurology, 45*(3), 329-336.

Hanna, J. P., & Frank, J. I. (1995). Automatic stepping in the pontomedullary stage of central herniation. *Neurology, 45*, 985-986.

Hartmann, A., Luzi, S., Murer, K., de Bie, R., & de Bruin, E. D. (2009). Concurrent validity of a trunk tri-axial accelerometer system for gait analysis in older adults. *Gait & Posture, 29*, 444-448.

Hathout, G. M., & Bhidayasiri, R. (2005). Midbrain ataxia: an introduction to the mesencephalic locomotor region and the pedunculopontine nucleus *American Roentgen Ray Society 184*, 953-956.

Hausdorff, J., Cudkowicz, M., Firtion, R., Wei, J., & Goldberger, A. (1998). Gait variability and basal ganglia disorders: stride-to-stride variations of gait cycle timing in Parkinson's disease and Huntington's disease. *Movement Disorders, 13*(4), 428 - 437.

Hausdorff, J., Edelberg, H., Mitchell, S., Goldberger, A., & Wei, J. (1997). Increased gait unsteadiness in community-dwelling elderly fallers. *Archives of Physical Medicine and Rehabilitation, 78*, 278 - 283.

Hausdorff, J., Lertratanakul, A., Cudkowicz, M., Peterson, A., Kaliton, D., & Goldberger, A. (2000). Dynamic markers of altered gait rhythm in amyotrophic lateral sclerosis. *Journal of Applied Physiology, 88*(6), 2045 - 2053.

Hausdorff, J., Mitchell, S., Firtion, R., Peng, C., Cudkowicz, M., Wei, J., et al. (1997). Altered fractal dynamics of gait: reduced stride-interval correlations with aging and Huntington's disease. *Journal of Applied Physiology, 82*(1), 262 - 269.

Hausdorff, J., Nelson, M., Kaliton, D., Layne, J., Bernstein, M., Nuernberger, A., et al. (2001). Etiology and modification of gait instability in older adults: a randomized controlled trial of exercise. *Journal of Applied Physiology, 90*(6), 2117 - 2129.

Hausdorff, J., Peng, C.-K., Goldberger, A., & Stoll, A. (2004). Gait unsteadiness and fall risk in two affective disorders: a preliminary study. *BMC Psychiatry, 4*(1), 39.

Hausdorff, J., Peng, C., Ladin, Z., Wei, J., & Goldberger, A. (1995). Is walking a random walk? Evidence for long-range correlations in stride interval of human gait. *Journal of Applied Physiology, 78*(1), 349 - 358.

Hausdorff, J., Purdon, P., Peng, C.-K., Ladin, Z., Wei, J., & Goldberger, A. (1996). Fractal dynamics of human gait: stability of long-range correlations in stride interval fluctuations. *Journal of Applied Physiology, 80*(5), 1448 - 1457.

Hausdorff, J., Rios, D., & Edelberg, H. (2001). Gait variability and fall risk in community-living older adults: a 1-year prospective study. *Archives of Physical Medicine and Rehabilitation, 82*(8), 1050 - 1056.

Hausdorff, J., Zemany, L., Peng, C.-K., & Goldberger, A. (1999). Maturation of gait dynamics: stride-to-stride variability and its temporal organization in children. *Journal of Applied Physiology, 86*(3), 1040 - 1047.

Hausdorff, J. M. (2005). Gait variability: methods, modeling and meaning. *Journal of Neuroengineering Rehabilitation, 2*(19).

Hausdorff, J. M. (2007). Gait dynamics, fractals and falls: Finding meaning in the stride-to-stride fluctuations of human walking. *Human Movement Science, 26*(4), 555-589.

Hausdorff, J. M., Ashkenazy, Y., Peng, C.-K., Ivanov, P. C., Stanley, H. E., & Goldberger, A. L. (2001). When human walking becomes random: fractal analysis and modeling of gait rhythm fluctuations. *Physica A, 302*, 138-147.

Hausdorff, J. M., Edelberg, H. K., Mitchell, S. L., Goldberger, A. L., & Wei, J. Y. (1997). Increased gait unsteadiness in community-dwelling elderly fallers. *Archives of Physical Medicine and Rehabilitation, 78*, 278-283.

Hausdorff, J. M., Forman, D. E., Ladin, Z., Goldberger, A. L., Rigney, D. R., & Wei, J. Y. (1994). Increased walking variability in elderly persons with congestive heart failure. *Journal of the American Geriatrics Society, 42*(10), 1056-1061.

Hausdorff, J. M., Forman, D. E., Pilgrim, D. M., Rigney, D. R., & Wei, J. Y. (1992). A new technique for simultaneous monitoring of electrocardiogram and walking cadence. *The American Journal of Cardiology, 70*(11), 1064-1071.

Hausdorff, J. M., Herman, T., Baltadjieva, R., Gurevich, T., & Giladi, N. (2003). Balance and gait in older adults with systemic hypertension. *The American Journal of Cardiology, 91*(5), 643-645.

Havlin, S., Amaral, L. A. N., Ashkenazy, Y., Goldberger, A. L., Ivanov, P. C., Peng, C.-K., et al. (1999). Application of statistical physics to heartbeat diagnosis. *Physica A: Statistical Mechanics and its Applications, 274*(1-2), 99-110.

Heasley, K., Buckley, J. G., Scally, A., Twigg, P., & Elliott, D. B. (2005). Falls in older people: Effects of age and blurring vision on the dynamics of stepping. *Investigative Ophthalmololgy and Visual Science, 46*(10), 3584-3588.

Heiderscheit, B. C., Hamill, J., & van Emmerik, R. E. A. (2002). Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *Journal of Applied Biomechanics, 18*, 110-121.

Heitman, D. K., Gossman, M. R., Shaddeau, S. A., & Jackson, J. R. (1989). Balance performance and step width in noninstitutionalized, elderly, female fallers and nonfallers. *Physical Therapy, 69*(11), 923-931.

Helbostad, J. L., Sletvold, O., & Moe-Nilssen, R. (2004). Home training with and without additional group training in physically frail old people living at home: effect on health-related quality of life and ambulation. *Clinical Rehabilitation, 18*(5), 498-508.

Helbostad, J. L., Vereijken, B., Hesseberg, K., & Sletvold, O. (2009). Altered vision destabilizes gait in older persons. *Gait & Posture, 30*(2), 233-238.

Hendrie, D., Hall, S. E., Arena, G., & Legge, M. (2004). Health system costs of falls of older adults in Western Australia. *Australian Health Review, 28*(3), 363-373.

Henriksen, M., Lund, H., Moe-Nilssen, R., Bliddal, H., & Danneskiod-Samsøe, B. (2004). Test-retest reliability of trunk accelerometric gait analysis. *Gait & Posture, 19*(3), 288-297.

Herman, T., Giladi, N., Gurevich, T., & Hausdorff, J. M. (2005). Gait instability and fractal dynamics of older adults with a "cautious" gait: Why do certain older adults walk fearfully? *Gait Posture, 21*, 178-185.

Hikosaka, O., Takikawa, Y., & Kawagoe, R. (2000). Role of the basal ganglia in the control of purposive saccadic eye movements. *Physiological Reviews, 80*(3), 953-978.

Hill, K. D. (1997). *Studies of balance in older people.* Unpublished Doctoral Thesis, University of Melbourne, Melbourne.

Hill, K. D., Berhardt, J., McGann, A., Maltese, D., & Berkovits, D. (1996). A new test of dynamic standing balance for stroke patients: reliability, validity and comparison with healthy elderly. *Physiotherapy Canada, 48*, 257-262.

Hill, K. D., Goldie, P. A., Baker, P. A., & Greenwood, K. M. (1994). Retest reliability of the temporal and distance characteristics of hemiplegic gait using a footswitch system. *Archives of Physical Medicine and Rehabilitation, 75*, 577-583.

Hill, K. D., Schwarz, J., Flicker, L., & Carroll, S. (1999). Falls among healthy, community-dwelling older women: a prospective study of frequency, circumstances, consequences and prediction accuracy. *Australian and New Zealand Journal of Public Health, 23*(1), 41-48.

Hill, T., & Lewicki, P. (2006). *Statistics: methods and applications*. Tulsa, OK: StatSoft.

Ho, K. K. L., Moody, G. B., Peng, C.-K., Mietus, J. E., Larson, M. G., Levy, D., et al. (1997). Predicting survival in heart failure case and control subjects by use of fully automated methods for deriving nonlinear and conventional indices of heart rate dynamics. *Circulation, 96*(3), 842-848.

Hollman, J. H., Kovash, F. M., Kubik, J. J., & Linbo, R. A. (2007). Age-related differences in spatiotemporal markers of gait stability during dual task walking. *Gait & Posture, 26*(1), 113-119.

Hopkins, W. G. (2000). Measures of reliability in sports medicine and science. *Sports Medicine, 30*(1), 1-15.

Hortobágyi, T., & DeVita, P. (2000). Muscle pre- and coactivity during downward stepping are associated with leg stiffness in aging. *Journal of Electromyography and Kinesiology, 10*(2), 117-126.

Hortobágyi, T., Solnik, S., Gruber, A., Rider, P., Steinweg, K., Helseth, J., et al. (2009). Interaction between age and gait velocity in the amplitude and timing of antagonist muscle coactivation. *Gait & Posture, 29*(4), 558-564.

Howland, J., Lachman, M., Peterson, E., Cote, J., Kasten, L., & Jette, A. (1998). Covariates of fear of falling and associated activity curtailment. *Gerontologist, 38*(5), 549-555.

Howland, J., Peterson, E. W., Levin, W. C., Fried, L., Pordon, D., & Bak, S. (1993). Fear of falling among the community-dwelling elderly. *Journal of Aging and Health, 5*(2), 229-243.

Hsu, M. J., Wei, S. H., Yu, Y. H., & Chang, Y. J. (2007). Leg stiffness and electromyography of knee extensors/flexors: Comparison between older and younger adults during stair descent. *Journal of Rehabilitation Research and Development, 44*(3), 429-435.

Hu, K., Ivanov, P., Chen, Z., Carpena, P., & Eugene Stanley, H. (2001). Effect of trends on detrended fluctuation analysis. *Physical Review E: Statistical, Nonlinear and Soft Matter Physics, 64*(1 pt 1), 011114.

Huikuri, H. V., Mäkikallio, T. H., Peng, C.-K., Goldberger, A. L., Hintze, U., & Moller, M. (2000). Fractal correlation properties of R-R interval dynamics and mortality in patients with depressed left ventricular function after an acute myocardial infarction. *Circulation, 101*(1), 47-53.

Hultborn, H., & Nielsen, J. B. (2007). Spinal control of locomotion - from cat to man. *Acta Physiologica, 189*, 111-121.

Hurley, M., Rees, J., & Newham, D. (1998). Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle aged and elderly subjects. *Age Ageing, 27*, 55-62.

Ilg, W., Golla, H., Thier, P., & Giese, M. A. (2007). Specific influences of cerebellar dysfunctions on gait. *Brain, 130*(3), 786.

Inman, V. T., Ralston, H., & Todd, F. (1981). *Human Walking*. Baltimore: Williams & Wilkins.

Inoue, Y., Kuwahara, T., & Araki, T. (2004). Maturation- and aging-related changes in heat loss effector fnction. *Journal of Physiological Anthropology and Applied Human Science, 23*(6), 289-294.

Isberner, F., Ristzel, D., Sarvela, O., Brown, K., Hu, P., & Newbolds, D. (1998). Falls in elderly rural home health clients. *Home Health Care Service Quarterly, 17*, 41-51.

Ishikawa, K., Edo, M., Terada, N., Okamoto, Y., & Togawa, K. (1993). Gait analysis in patients with vertigo. *European Archives of Oto-Rhino-Laryngology, 250*(4), 229-232.

Ishikawa, K., Edo, M., Yokomizo, M., Terada, N., Okamoto, Y., & Togawa, K. (1994). Analysis of gait in patients with peripheral vestibular disorders. *Journal of Oto-Rhino-Laryngology and its Related Specialities, 56*(6), 325-330.

Ishikawa, K., Edo, M., Yokomizo, M., & Togawa, K. (1995). Characteristics of human gait related variables in association with vestibular system disorders. *Acta Oto-laryngologica, 115*(s520), 199-201.

Ivers, R. Q., Norton, R., Cumming, R. G., Butler, M., & Campbell, A. J. (2000). Visual impairment and risk of hip fracture. *American Journal of Epidemiology, 152*(7), 633-639.

Iyengar, N., Peng, C.-K., Morin, R., Goldberger, A. L., & Lipsitz, L. A. (1996). Age-related alterations in the fractal scaling of cardiac interbeat interval dynamics. *American Journal of Physiology, 271*, 1078-1084.

Jacobs, J., Dimitrova, D., Nutt, J., & Horak, F. (2005). Can stooped posture explain multidirectional postural instability in patients with Parkinson's disease? *Experimental Brain Research, 166*(1), 78-88.

Jahn, K., Deutschländer, A., Stephan, T., Kalla, R., Wiesmann, M., Strupp, M., et al. (2008). Imaging human supraspinal locomotor centers in brainstem and cerebellum. *Neuroimage, 39*, 786-792.

Jahn, K., Deutschlander, A., Stephan, T., Strupp, M., Wiesmann, M., & Brandt, T. (2004). Brain activation patterns during imagined stance and locomotion in functional magnetic resonance imaging. *Neuroimage, 22*, 1722-1731.

Jankowska, E., Jukes, M. G. M., Lund, S., & Lundberg, A. (1967a). The effect of DOPA on the spinal cord 5. Reciprocal organization of pathways transmitting excitatory action to alpha motoneurones of flexors and extensors. *Acta Phsyiologica Scandinavica, 70*, 369-388.

Jankowska, E., Jukes, M. G. M., Lund, H., & Lundberg, A. (1967b). The effect of DOPA on the spinal cord 6. Half-centre organization of interneurones transmitting effects from the flexor reflex afferents. *Acta Phsyiologica Scandinavica, 70*, 389-402.

Jordan, K., Challis, J. H., Cusumano, J. P., & Newell, K. M. (2009). Stability and the time-dependent structure of gait variability in walking and running. *Human Movement Science, 28*(1), 113-128.

Jordan, K., Challis, J. H., & Newell, K. M. (2007). Walking speed influences on gait cycle variability. *Gait & Posture, 26*(1), 128-134.

Jordan, L. M., Pratt, C. A., & Menzies, J. E. (1979). Locomotion evoked by brainstem stimulation: occurrence without phasic segmental afferent input. *Brain Research, 177*(1), 204-207.

Jørstad, E. C., Hauer, K., Becker, C., & Lamb, S. E. (2005). Measuring the psychological outcomes of falling: A systematic review. *Journal of the American Geriatrics Society, 53*(3), 501-510.

Judge, J., King, M., Whipple, R., Clive, J., & Wolfson, L. (1995). Dynamic balance in older persons: effects of reduced visual and proprioceptive input. *Journal of Gerontology Series A: Biological Sciences and Medical Sciences, 50*(5), M263-270.

Judge, J. O., Davis, R. B., & Ounpuu, S. (1996). Step length reductions in advanced age: the role of ankle and hip kinetics. *Journal of Gerontology Series A: Biological Sciences and Medica Sciences, 51A*, 303-312.

Kang, H. G., & Dingwell, J. B. (2008). Separating the effects of age and walking speed on gait variability. *Gait & Posture, 27*(4), 572-577.

Kang, H. G., & Dingwell, J. B. (2009). Dynamics and stability of muscle activations during walking in healthy young and older adults. *Journal of Biomechanics, 42*(14), 2231-2237.

Kantelhardt, J. W., Koscielny-Bunde, E., Rego, H. H. A., Havlin, S., & Bunde, A. (2001). Detecting long-range correlations with detrended fluctuation analysis. *Physica A: Statistical Mechanics and its Applications, 295*(3-4), 441-454.

Karmakar, C. K., Khandoker, A. H., Begg, R. K., Palaniswami, M., & Taylor, S. (2007). *Understanding Ageing Effects by Approximate Entropy Analysis of gait variability.* Paper presented at the Proceedings of the 29th Annual International Conference of the IEEE EMBS, Lyon, France.

Kästenbauer, T., Sauseng, S., Brath, H., Abrahamian, H., & Irsigler, K. (2004). The value of the Rydel-Seiffer tuning fork as a predictor of diabetic polyneuropathy compared with a neurothesiometer. *Diabetic Medicine, 21*, 563-567.

Kavanagh, J. J., & Menz, H. B. (2008). Accelerometry: a technique for quantifying movement patterns during walking. *Gait & Posture, 28*(1), 1-15.

Kavanagh, J. J., Morrison, S., James, D. A., & Barrett, R. (2006). Reliability of segmental accelerations measured using a new wireless gait analysis system. *Journal of Biomechanics, 39*, 2863-2872.

Kerrigan, D. C., Todd, M. K., & Croce, U. D. (1998). Gender differences in joint biomechanics during walking: normative study in young adults. *American Journal of Physical Medicine & Rehabilitation, 77*(1), 2-7.

Khasnis, A., & Gokula, R. M. (2003). Romberg's test. *Journal of Postgraduate Medicine, 49*(2), 169-172.

Klein, R., Klein, B. E. K., Lee, K. E., Cruickshanks, K. J., & Gangnon, R. E. (2006). Changes in visual acuity in a population over a 15-year period: The Beaver Dam Eye Study. *American Journal of Ophthalmology, 142*(4), 539-549.e532.

Kochanek, K. D., Murphy, S. L., Anderson, R. N., & Scott, C. (2004). Deaths: final data for 2002. *National Vital Statistics Report, 53*(5), 1-115.

Koepsell, T. D., Wolf, M. E., Buchner, D. M., Kukull, W. A., LaCroix, A. Z., Tencer, A. F., et al. (2004). Footwear style and risk of falls in older adults. *Journal of the American Geriatrics Society, 52*(9), 1495-1501.

Kriellaars, D. J., Brownstone, R. M., Noga, B. R., & Jordan, L. M. (1994). Mechanical entrainment of fictive locomotion in the decerebrate cat. *Journal of Neurophysiology, 71*(6), 2074-2086.

Kwon, O.-Y., Minor, S. D., Maluf, K. S., & Mueller, M. J. (2003). Comparison of muscle activity during walking in subjects with and without diabetic neuropathy. *Gait & Posture, 18*(1), 105-113.

Lach, H. W. (2005). Incidence and risk factors for developing fear of falling in older adults. *Public Health Nursing, 22*(1), 45-52.

Lachman, M. E., Howland, J., Tennstedt, S., Jette, A., Assmann, S., & Peterson, E. W. (1998). Fear of falling and activity restriction: The Survey of Activities and Fear of Falling in the Elderly (SAFE). *Journals of Gerontology Series B: Pshycological Sciences and Social Sciences, 53B*(1), P43-50.

Larsen, A. H., Puggaard, L., Hämäläinen, U., & Aagaard, P. (2008). Comparison of ground reaction forces and antagonist muscle coactivation during stair walking with ageing. *Journal of Electromyography and Kinesiology, 18*(4), 568-580.

Laufer, Y. (2003). Age- and gender-related changes in the temporal-spatial characteristics of forwards and backwards gaits. *Physiotherapy Research International, 8*(3), 131-142.

Leibson, C. L., Tosteson, A. N. A., Gabriel, S. E., Ransom, J. E., & III, L. J. M. (2002). Mortality, disability, and nursing home use for persons with and without hip fracture: A population-based study. *Journal of the American Geriatrics Society, 50*(10), 1644-1650.

Leipzig, R. M., Cumming, R. G., & Tinetti, M. E. (1999). Drugs and falls in older people: a systematic review and meta-analysis: II. Cardiac and analgesic drugs. *Journal of the American Geriatrics Society, 47*(1), 40-50.

Li, W., Keegan, T. H. M., Sternfeld, B., Sidney, S., Quesenberry, C. P., Jr, & Kelsey, J. L. (2006). Outdoor falls among middle-aged and older adults: A neglected public health problem. *American Journal of Public Health, 96*(7), 1192-1200.

Lichtenstein, M. J., Burger, M. C., Shields, S. L., & Shiavi, R. G. (1990). Comparison of biomechanics platform measures of balance and videotaped measures of gait with a clinical mobility scale in elderly women. *Journal of Gerontology, 45*(2), M49-54.

Lindemann, U., Najafi, B., Zijlstra, W., Hauer, K., Muche, R., Becker, C., et al. (2008). Distance to achieve steady state walking speed in frail elderly persons. *Gait & Posture, 27*(1), 91-96.

Lord, S. R. (2006). Visual risk factors for falls in older people. *Age Ageing, 35*(suppl_2), ii42-45.

Lord, S. R., & Dayhew, J. (2001). Visual risk factors for falls in older people. *Journal of the American Geriatrics Society, 49*(5), 508.

Lord, S. R., Lloyd, G. G., & Keung Li, S. (1996). Sensori-motor function, gait patterns and falls in community-dwelling women. *Age Ageing, 25*(4), 292-299.

Lord, S. R., & Menz, H. B. (2000). Visual contributions to postural stability in older adults. *Gerontolgy, 46*, 306-310.

Lord, S. R., & Menz, H. B. (2002). Physiologic, psychologic, and health predictors of 6-minute walk performance in older people. *Archives of Physical Medicine and Rehabilitation, 83*, 907-911.

Lord, S. R., Menz, H. B., & Sherrington, C. (2006). Home environment risk factors for falls in older people and the efficacy of home modifications. *Age Ageing, 35-S2*, ii55-ii59.

Lord, S. R., Menz, H. B., & Tiedemann, A. (2003). A physiological profile approach to falls risk assessment and prevention. *Physical Therapy, 83*(3), 237-252.

Lord, S. R., Ward, J. A., Williams, P., & Anstey, K. J. (1993). An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study. *Australian Journal of Public Health, 17*, 240-245.

Luft, A. R., Skalej, M., Schulz, J. B., Welte, D., Kolb, R., Burk, K., et al. (1999). Patterns of Age-related Shrinkage in Cerebellum and Brainstem Observed In Vivo Using Three-dimensional MRI Volumetry. *Cerebral Cortex, 9*(7), 712-721.

Lundberg, A. (1979). Multisensory control of spinal reflex pathways. *Progress in Brain Research, 50*, 11-28.

Lundberg, A., Malmgren, K., & Schomburg, E. D. (1977). Cutaneous facilitation of transmission in reflex pathways from Ib afferents to motoneurones. *Journal of Physiology, 265*(3), 763-780.

Luukinen, H., Herala, M., Koski, K., Honkanen, R., Laippala, P., & Kivelä, S. L. (2000). Fracture Risk Associated with a Fall According to Type of Fall Among the Elderly. *Osteoporosis International, 11*(7), 631-634.

MacKay-Lyons, M. (2002). Central pattern generation of locomotion: A review of the evidence. *Physical Therapy, 82*(1), 69-83.

Madhavan, S., & Shields, R. K. (2005). Influence of age on dynamic position sense: evidence using a sequential movement task. *Experimental Brain Research, 164*(1), 18-28.

Maffiuletti, N. A., Gorelick, M., de Quervain, I. K., Bizzini, M., Munzingera, J. P., S, T., et al. (2008). Concurrent validity and intrasession reliability of the IDEEA accelerometry system for the quantification of spatiotemporal gait parameters. *Gait & Posture, 27*(1), 160-163.

Magaziner, J., Hawkes, W., Hebel, J. R., Zimmerman, S. I., Fox, K. M., Dolan, M., et al. (2000). Recovery from hip fracture in eight areas of function. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 55*(9), M498-507.

Magaziner, J., Simonsick, E. M., Kashner, T. M., Hebel, J. R., & Kenzora, J. E. (1990). Predictors of functional recovery one year following hospital discharge for hip fracture: A prospective study. *Journal of Gerontology, 45*(3), M101-107.

Maki, B. E. (1997). Gait changes in older adults: predictors of falls or indicators of fear? *Journal of the American Geriatrics Society, 45*, 313-320.

Mäkikallio, T. H., Huikuri, H. V., Mäkikallio, A., Sourander, L. B., Mitrani, R. D., Castellanos, A., et al. (2001). Prediction of sudden cardiac death by fractal analysis of heart rate variability in elderly subjects. *Journal of the American College of Cardiology, 37*(5), 1395-1402.

Mäkikallio, T. H., Koistinen, J., Jordaens, L., Tulppo, M. P., Wood, N., Golosarky, B., et al. (1999). Heart rate dynamics before spontaneous onset of ventricular fibrillation in patients with healed myocardial infartcs. *American Journal of Cardiology, 83*, 880-884.

Malouin, F., Richards, C. L., Jackson, P. L., Dumas, F., & Doyon, J. (2003). Brain activations during motor imagery of locomotor-related tasks: a PET study. *Human Brain Mapping, 19*(1), 47-62.

Mansfield, A., & Lyons, G. M. (2003). The use of accelerometry to detect heel contact events for use as a sensor in FES assisted walking. *Medical Engineering & Physics, 25*(10), 879-885.

Marchetti, G. F., Whitney, S. L., Blatt, P. J., Morris, L. O., & Vance, J. M. (2008). Temporal and spatial characteristics of gait during performance of the dynamic gait index in people with and people without balance or vestibular disorders. *Physical Therapy, 88*(5), 640-651.

Marple-Horvat, D. E., & Criado, J. M. (1999). Rhythmic neuronal activity in the lateral cerebellum of the cat during visually guided stepping. *Journal of Physiology, 518*(2), 595-603.

Marsh, A. P., Katula, J. A., Pacchia, C. F., Johnsosn, L. C., Koury, K. L., & Rejeski, W. J. (2006). Effects of treadmill and overground walking on function and attitudes in older adults. *Medicine and Science in Sports and Exercise, 38*(6), 1157-1164.

Martin, F. C., Hart, D., Spector, T., Doyle, D. V., & Harari, D. (2005). Fear of falling limiting activity in young-old women is associated with reduced functional mobility rather than psychological factors. *Age Ageing, 34*(3), 281-287.

Martina, I. S. J., van Koningsveld, R., Schmitz, P. I. M., van der Meche, F. G. A., & van Doorn, P. A. (1998). Measuring vibration threshold with a graduated tuning fork in normal aging and in patients with polyneuropathy. *Jounal of Neurol Neurosurg Psychiatry, 65*, 743-747.

Masdeu, J. C., Alampur, U., Cavaliere, R., & Tavoulareas, G. (1994). Astasia and gait failure with damage of the pontomesencephalic locomotor region. *Annals of Neurology, 35*, 619-621.

Masud, T., & Morris, R. O. (2001). Epidemiology of falls. *Age Ageing, 30*(S4), 3-7.

Mathers, C., & Penm, R. (1999). *Health system costs of injury, poisoning and musculoskeletal disorders in Australia 1993–94. AIHW Cat. No. HWE 12.* Canberra: Australian Institute of Health and Welfare (Health and Welfare Expenditure Series no. 6).

Matsuyama, K., Mori, F., Nakajima, K., Drew, T., Aoki, M., & Mori, S. (2004). Locomotor role of the corticoreticular-reticulospinal-spinal interneuronal system. *Progress in Brain Research, 143*, 239-249.

Mayagoitiaa, R. E., Neneb, A. V., & Veltink, P. H. (2002). Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems. *Journal of Biomechanics, 35*, 537-542.

Mbourou, G. A., Lajoie, Y., & Teasdale, N. (2003). Step length variability at gait initiation in elderly fallers and non-fallers, and young adults. *Gerontology, 49*(1), 21-26.

McDonough, A. L., Batavia, M., Chen, F. C., Kwon, S., & Ziai, J. (2001). The validity and reliability of the GAITRite system's measurements: a preliminary evaluation. *Archives of Physical Medicine and Rehabilitation, 82*, 419-425.

McGibbon, C. A. (2003). Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. *Exercise and. Sport Sciences. Reviews, 31*(2), 102-108.

Ménard, A., Auclair, F., Bourcier-Lucas, C., Grillner, S., & Dubuc, R. (2007). Descending GABAergic projections to the mesencephalic locomotor region in the lamprey Petromyzon marinus. *The Journal of Comparitive Neurology, 501*(2), 260-273.

Mendes de Leon, C. F., Seeman, T. E., Baker, D. I., Richardson, E. D., & Tinetti, M. E. (1996). Self-efficacy, physical decline, and change in functioning in community-living elders: A prospective study. *Journals of Gerontology Series B: Pshycological Sciences and Social Sciences, 51B*(4), S183-190.

Menz, H. B., Latt, M. D., Tiedemann, A., Kwan, M. M. S., & Lord, S. R. (2004). Reliability of the GAITRite walkway system for the quantification of temporo-spatial parameters of gait in young and older people. *Gait & Posture, 20*, 20-25.

Menz, H. B., & Lord, S. R. (2001). The contribution of foot problems to mobility impairment and falls in community-dwelling older people. *Journal of the American Geriatrics Society, 49*, 1651-1656.

Menz, H. B., Lord, S. R., & Fitzpatrick, R. C. (2003). Acceleration patterns of the head and pelvis when walking are associated with risk of falling in community-dwelling older people. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 58*(5), M446-452.

Menz, H. B., Lord, S. R., & Fitzpatrick, R. C. (2003). Acceleration patterns of the head and pelvis when walking on level and irregular surfaces. *Gait & Posture, 18*(1), 35-46.

Menz, H. B., Morris, M. E., & Lord, S. R. (2006). Foot and ankle risk factors for falls in older people: A prospective study. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 61*(8), 866-870.

Menz, H. B., Morris, M. E., & Lord, S. R. (2006). Footwear characteristics and risk of indoor and outdoor falls in older people. *Gerontology, 52*(3), 174-180.

Mian, O. S., Thom, J. M., Ardigò, L. P., Narici, M. V., & Minetti, A. E. (2006). Metabolic cost, mechanical work, and efficiency during walking in young and older men. *Acta Physiologica, 186*(2), 127-139.

Miff, S. C., Childress, D. S., Gard, S. A., Meier, M. R., & Hansen, A. H. (2005). Temporal symmetries during gait initiation and termination in nondisabled ambulators and in people with unilateral transtibial limb loss. *Journal of Rehabilitation Research and Development, 42*(2), 175-182.

Mishina, M., Senda, M., Ishii, K., Ohyama, M., Kitamura, S., & Katayama, Y. (1999). Cerebellar activation during ataxic gait in olivopontocerebellar atrophy: a PET study. *Acta Neurologica Scandinavica, 100*(6), 369-376.

Miyai, I., Tanabe, H. C., Sase, I., Eda, H., Oda, I., Konishi, I., et al. (2001). Cortical mapping of gait in humans: a near-infrared spectroscopic topography study. *Neuroimage, 14*(5), 1186-1192.

Moe-Nilssen, R. (1998). Test-retest reliability of trunk accelerometry during standing and walking. *Archives of Physical Medicine and Rehabilitation, 79*(11), 1377-1385.

Moe-Nilssen, R., & Helbostad, J. L. (2004). Estimation of gait cycle characteristics by trunk accelerometry. *Journal of Biomechanics, 37*(1), 121-126.

Moe-Nilssen, R., & Helbostad, J. L. (2005). Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. *Gait & Posture, 21*(2), 164-170.

Moes, E., & Lombardi, K. (2009). The relationship between contrast sensitivity, gait, and reading speed in Parkinson's disease. *Aging, Neuropsychology, and Cognition, 16*, 121-132.

Mold, J. W., Vesely, S. K., Keyl, B. A., Schenk, J. B., & Roberts, M. (2004). The prevalence, predictors, and consequences of peripheral sensory neuropathy in older patients. *Journal of the American Board of Family Practice, 17*(5), 309-318.

Moller, J. (1998). Cost of Injury. Based on Watson, WL & Ozanne-Smith, J. The Cost of Injury to Victoria. December 1997. Monash University Accident Research Centre. Report No. 124, from http://www.nisu.flinders.edu.au/pubs/injcost/

Moller, J. (2003). Projected costs of fall related injury to older persons due to demographic change in Australia. Report to the Commonweath Department of Health and Ageing, under the National Falls Prevention for Older People Initiative.

Montero-Odasso, M., Schapira, M., Duque, G., Soriano, E. R., Kaplan, R., & Camera, L. A. (2005). Gait disorders are associated with non-cardiovascular falls in elderly people: a preliminary study. *BMC Geriatrics, 5*(15).

Montero-Odasso, M., Schapira, M., Soriano, E. R., Varela, M., Kaplan, R., Camera, L. A., et al. (2005). Gait velocity as a single predictor of adverse events in healthy seniors aged 75 years and older. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 60*, 1304-1309.

Moore, D. S., & Ellis, R. (2008). Measurement of fall-related psychological constructs among independent-living older adults: A review of the research literature. *Aging & Mental Health, 12*(6), 684 - 699.

Mori, S., Matsuyama, K., Mori, F., & Nakajima, K. (2001). Supraspinal sites that induce locomotion in the vertebrate central nervous system. *Advanced Neurology, 87*, 25-40.

Morris, M. E., Iansek, R., Matyas, T. A., & Summers, J. J. (1994). The pathogenesis of gait hypokinesia in Parkinson's disease. *Brain, 117*(5), 1169-1181.

Morris, M. E., Iansek, R., Smithson, F., & Huxham, F. (2000). Postural instability in Parkinson's disease: a comparison with and without a concurrent task. *Gait & Posture, 12*(3), 205-216.

Morris, M. E., Osborne, D., Hill, K., Kendig, H., Lundgren-Lindquist, B., Browning, C., et al. (2004). Predisposing factors for occasional and multiple falls in older Australians who live at home. *Australian Journal of Physiotherapy, 50*(3), 153-159.

Mottram, S., Peat, G., Thomas, E., Wilkie, R., & Croft, P. (2008). Patterns of pain and mobility limitation in older people: cross-sectional findings from a population survey of 18,497 adults aged 50 years and over. *Quality of Life Research, 17*(4), 529-539.

Murphy, J., & Isaacs, B. (1982). The post-fall syndrome. A study of 36 elderly patients. *Gerontology, 28*(4), 265-270.

Murphy, S. L., Williams, C. S., & Gill, T. M. (2002). Characteristics associated with fear of falling and activity restriction in community-living older persons. *Journal of the American Geriatrics Society, 50*(3), 516-520.

Murray, M. P., Drought, A. B., & Kory, R. C. (1964). Walking patterns of normal men. *Journal of Bone and Joint Surgery, 46*, 335-360.

Mustard, C. A., & Mayer, T. (1997). Case-control study of exposure to medication and the risk of injurious falls requiring hospitalization among nursing home residents *American Journal of Epidemiology, 145*(8), 738-745.

Myers, S. A., Johanning, J. M., Stergiou, N., Celis, R. I., Robinson, L., & Pipinos, I. I. (2009). Gait variability is altered in patients with peripheral arterial disease. *Journal of Vascular Surgery, 49*(4), 924-931.

Nachreiner, N. M., Findorff, M. J., Wyman, J. F., & McCarthy, T. C. (2007). Circumstances and consequences of falls in community-dwelling older women. *Journal of Women's Health, 16*(10), 1437-1446.

Nakamura, T., Meguro, K., & Sasaki, H. (1996). Relationship between falls and stride length variability in senile dementia of the Alzheimer type. *Gerontology, 42*, 108 - 113.

Nallegowda, M., Singh, U., Handa, G., Khanna, M., Wadhwa, S., Yadav, S. L., et al. (2004). Role of sensory input and muscle strength in maintenance of balance, gait, and posture in Parkinson's disease: a pilot study. *American Journal of Physical Medicine and Rehabilitation, 83*(12), 898-908.

National Ageing Research Institute (2004). An analysis of research on preventing falls and falls injury in older people: Community, residential care and hospital settings (2004 update). Department of Health and Ageing, Canberra.

Nelson, A. J., Zwick, D., Brody, S., Doran, C., Pulver, L., Rooz, G., et al. (2002). The validity of the GAITRite and the Functional Ambulation Performance scoring system in the analysis of Parkinson gait. *NeuroRehabilitation, 17*(3), 255.

Nelson, R. C., Dillman, C. J., Lagasse, P., & Bickett, P. (1972). Biomechanics of overground versus treadmill running. *Medicine & Science in Sports, 4*(4), 233-240.

Neutel, C. I., Perry, S., & Maxwell, C. (2002). Medication use and risk of falls. *Pharmacoepidemiology and Drug Safety, 11*, 97-104.

Noga, B. R., Kriellaars, D. J., Brownstone, R. M., & Jordan, L. M. (2003). Mechanism for activation of locomotor centers in the spinal cord by stimulation of the mesencephalic locomotor region. *Journal of Neurophysiology, 90*, 1464-1478.

Norris, J. A., Granata, K. P., Mitros, M. R., Byrne, E. M., & Marsh, A. P. (2007). Effect of augmented platarflexion power on preferred walking speed and economy in young and older adults. *Gait & Posture, 25*, 620-627.

Northridge, M. E., Nevitt, M. C., Kelsey, J. L., & Link, B. (1995). Home hazards and falls in the elderly: the role of health and functional status. *American Journal of Public Health, 85*, 509-515.

Norton, R., Campbell, A., Lee-Joe, T., Robinson, E., & Butler, M. (1997). Circumstances of falls resulting in hip fractures among older people. *Journal of the American Geriatrics Society, 45*, 1108-1112.

Nutt, J. G., Marsden, C. D., & Thompson, P. D. (1993). Human walking and higher-level gait disorders, particularly in the elderly. *Neurology, 43*(2), 268-279.

Oberg, T., Karsznia, A., & Oberg, K. (1993). Basic gait parameters: reference data for normal subjects, 10-79 years of age. *Journal of Rehabilitation Research and Development, 30*(2), 210-223.

Oberg, T., Karsznia, A., & Oberg, K. (1994). Joint angle parameters in gait: Reference data for normal subjects. *Journal of Rehbailitation Research and Development, 31*, 199-213.

Ochi, K., & Ohashi, T. (2003). Age-related changes in the vestibular-evoked myogenic potentials. *Otolaryngology - Head and Neck Surgery, 129*(6), 655-659.

Öken, Ö., Yavuzer, G., Ergöçen, S., YorgancIoglu, Z. R., & Stam, H. J. (2008). Repeatability and variation of quantitative gait data in subgroups of patients with stroke. *Gait & Posture, 27*(3), 506-511.

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

Oliver, D., Daly, F., Martin, F. C., & McMurdo, M. E. T. (2004). Risk factors and risk assessment tools for falls in hospital in-patients: a systematic review. *Age Ageing, 33*(2), 122-130.

Orces, C. H. (2009). Trends in hospitalization for fall-related injury among older adults in the United States, 1988-2005. *Ageing Research, 1*(1).

Orendurff, M. S., Schoen, J. A., Bernatz, G. C., Segal, A. D., & Klute, G. K. (2008). How humans walk: Bout duration, steps per bout, and rest duration. *Journal of Rehabilitation Research & Development, 45*(7), 1077–1090.

Orlovsky, G. N. (1991). Cerebellum and locomotion. In M. Shimamura, S. Grillner & V. R. Edgerton (Eds.), *Neurobiological Basis of Human Locomotion*. Tokyo, Japan: Japan Scientific Societies Press.

Owings, T. M., & Grabiner, M. D. (2003). Measuring step kinematic variability on an instrumented treadmill: how many steps are enough? *Journal of Biomechanics, 36*(8), 1215-1218.

Owings, T. M., & Grabiner, M. D. (2004a). Step width variability, but not step length variability or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion. *Journal of Biomechanics, 37*(6), 935-938.

Owings, T. M., & Grabiner, M. D. (2004b). Variability of step kinematics in young and older adults. *Gait & Posture, 20*, 26-29.

Pahapill, P. A., & Lozano, A. M. (2000). The pedunculopontine nucleus and Parkinson's disease. *Brain, 123*(9), 1767-1783.

Palombaro, K. M., Craik, R. K., Mangione, K. K., & Tomlinson, J. D. (2006). Determining meaninful changes in gait speed after hip fracture. *Physical Therapy, 86*, 809-816.

Papadakis, N. C., Christakis, D. G., Tzagarakis, G. N., Chlouverakis, G. I., Kampanis, N. A., Stergiopoulos, K. N., et al. (2009). Gait variability measurements in lumbar spinal stenosis patients: part A. Comparison with healthy subjects. *Physiological Measurement*(11), 1171.

Pardoe, J., Edgley, S. A., Drew, T., & Apps, R. (2004). Changes in excitability of ascending and descending inputs to cerebellar climbing fibers during locomotion *Journal of Neurophysiology, 24*(11), 2656-2666.

Parvataneni, K., Ploeg, L., Olney, S. J., & Brouwer, B. (2009). Kinematic, kinetic and metabolic parameters of treadmill versus overground walking in healthy older adults. *Clinical Biomechanics, 24*(1), 95-100.

Patel, I., Turano, K. A., Broman, A. T., Bandeen-Roche, K., Munoz, B., & West, S. K. (2006). Measures of visual function and percentage of preferred walking speed in older adults: The Salisbury Eye Evaluation Project. *Investigative Ophthalmololgy and Visual Science, 47*(1), 65-71.

Pearson, K. G. (1995). Proprioceptive regulation of locomotion. *Current Biology, 5*, 786-791.

Pearson, K. G. (2004). Generating the walking gait: role of sensory feedback. *Progress in Brain Research, 143*, 123-129.

Peitersen, E. (1967). Vestibulospinal reflexes X. theoretical and clinical aspects of the stepping test. *Acta Oto-Laryngologica, 85*, 88-94.

Peng, C.-K., Havlin, S., Stanley, H. E., & Goldberger, A. L. (1995). Quantification of scaling exponents and crossover phenomena in nonstationary heartbeat time series. *Chaos, 5*, 82-87.

Peng, C.-K., Mietus, J., Hausdorff, J. M., Havlin, S., Stanley, H. E., & Goldberger, A. L. (1993). Long-range anticorrelations and non-Gaussian behavior of the heartbeat. *Physical Review Letters, 70*(9), 1343.

Peppe, A., Chiavalon, C., Pasqualetti, P., Crovato, D., & Caltagirone, C. (2007). Does gait analysis quantify motor rehabilitation efficacy in Parkinson's disease patients? *Gait & Posture, 26*, 452-462.

Perera, S., Mody, S. H., Woodman, R. C., & Studenski, S. A. (2006). Meaningful change and responsiveness in common physical performance measures in older adults. *Journal of the American Geriatrics Society, 54*(5), 743-749.

Perneger, T. V. (1998). What's wrong with Bonferroni adjustments. *British Medical Journal, 316*, 1236-1238.

Perry, J. (1992). *Gait analysis : normal and pathological function* NJ: Slack.

Petty, J., Hill, K., Haber, N. E., Paton, L., Lawrence, K., Berkovic, S., et al. (2010). Balance impairment in chronic antiepileptic drug users: A twin and sibling study. *Epilepsia, 51*(2), 280-288.

Pierrynowski, M. R., Gross, A., Miles, M., Galea, V., McLaughlin, L., & McPhee, C. (2005). Reliability of the long-range power-law correlations obtained from the bilateral stride intervals in asymptomatic volunteers whilst treadmill walking. *Gait & Posture, 22*(1), 46-50.

Pikkujamsa, S. M., Mäkikallio, T. H., Sourander, L. B., Raiha, I. J., Puukka, P., Skytta, J., et al. (1999). Cardiac interbeat interval dynamics from childhood to senescence. Comparison of conventional and new measures based on fractals and chaos theory. *Circulation, 100*, 393-399.

Pillar, T., Dickstein, R., & Smolinski, Z. (1991). Walking reeducation with partial relief of body weight in rehabilitation of patients with locomotor disabilities. *Journal of Rehabilitation Research and Development, 28*, 47-52.

Pit, S. W., Byles, J. E., Henry, D. A., Holt, L., Hansen, V., & Bowman, D. A. (2007). A quality use of medicines for general practitioners and older people: a cluster randomised controlled trial. *Medical Journal of Australia, 187*(1), 22-30.

Pluijm, S., Smit, J., Tromp, E., Stel, V., Deeg, D., Bouter, L., et al. (2006). A risk profile for identifying community-dwelling elderly with a high risk of recurrent falling: results of a 3-year prospective study. *Osteoporosis International, 17*(3), 417-425.

Podsiadlo, D., & Richardson, S. (1991). The Timed Up & Go: a test of basic functional mobility for frail elderly persons. *Journal of the American Geriatrics Society, 39*(2), 142-148.

Portney, L. G., & Watkins, M. P. (2000). *Foundations of clinical research: applications to practice* (2nd ed.). Upper Saddle River, NJ: Prentice Hall Health.

Prince, F., Corriveau, H., Hébert, R., & Winter, D. A. (1997). Gait in the elderly. *Gait & Posture, 5*(2), 128-135.

Prince, F., Sadeghi, H., Zabjek, K. F., & Allard, P. (2000). *Functional tasks for sagittal hip muscle power in elderly and young able-bodied gait*. Paper presented at the Gait and Clinical Movement Analysis Society.

Prochazka, A., Bennett, D. J., Stephens, M. J., Patrick, S. K., Sears-Duru, R., Roberts, T., et al. (1997). Measurement of rigidity in Parkinson's disease. *Movement Disorders, 12*(1), 24-32.

Pruessner, J. C., Collins, D. L., Pruessner, M., & Evans, A. C. (2001). Age and gender predict volume decline in the anterior and posterior hippocampus in early adulthood. *Journal of Neuroscience, 21*(1), 194-200.

Rao, A. K., Quinn, L., & Marder, K. S. (2005). Reliability of spatiotemporal gait outcome measures in Huntington's disease. *Movement Disorders, 20*(8), 1033-1037.

Rao, S. S., Hofmann, L. A., & Shakil, A. (2006). Parkinson's disease: diagnosis and treatment. *American Family Physician, 79*(12), 2046-2054.

Rapp, K., Becker, C., Lamb, S. E., Icks, A., & Klenk, J. (2008). Hip fractures in institutionalized elderly people: Incidence rates and excess mortality. *Journal of Bone and Mineral Research, 23*(11), 1825-1831.

Raz, N., Gunning, F., Head, D., Dupuis, J., McQuain, J., Briggs, S., et al. (1997). Selective aging of the human cerebral cortex observed in vivo: differential vulnerability of the prefrontal gray matter. *Cerebral Cortex, 7*(3), 268-282.

Raz, N., Lindenberger, U., Rodrigue, K. M., Kennedy, K. M., Head, D., Williamson, A., et al. (2005). Regional brain changes in aging healthy adults: General trends, individual differences and modifiers. *Cerebral Cortex, 15*(11), 1676-1689.

Resnick, H. E., Vinik, A. I., Schwartz, A. V., Leveille, S. G., Brancati, F. L., Balfour, J., et al. (2000). Independent effects of peripheral nerve dysfunction on lower-extremity physical function in old age: the Women's Health and Aging Study. *Diabetes Care, 23*(11), 1642-1647.

Resnick, S. M., Pham, D. L., Kraut, M. A., Zonderman, A. B., & Davatzikos, C. (2003). Longitudinal magnetic resonance imaging studies of older adults: A shrinking brain. *Journal of Neuroscience, 23*(8), 3295-3301.

Reyes-Ortiz, C. A., Al Snih, S., & Markides, K. S. (2005). Falls among elderly persons in Latin America and the Caribbean and among elderly Mexican-Americans. *Revista Panamericana de Salud Pública, 17*(5-6), 362-369.

Richardson, J., & Hurvitz, E. (1995). Peripheral neuropathy: a true risk factor for falls. *Journal of Gerontology, 50*, M211-M215.

Riley, P. O., Croce, U. D., & Kerrigan, D. C. (2001). Effect of age on lower extremity joint moment contributions to gait speed. *Gait and Posture, 14*, 264-270.

Riley, P. O., & Kerrigan, D. C. (2005). Laboratory-Based Evaluation of Gait Disorders: High-Tech. In J. M. Hausdorff & N. B. Alexander (Eds.), *Gait Disorders: Evaluation and Management*. FL: Taylor and Francis.

Riley, P. O., Paolini, G., Croce, U. D., Paylo, K. W., & Kerrigan, D. C. (2007). A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait & Posture, 26*(1), 17-24.

Rivas, J., Gaztelu, J. M., & García-Austt, E. (1996). Changes in hippocampal cell discharge patterns and theta rhythm spectral properties as a function of walking velocity in the guinea pig. *Experimental Brain Research, 108*(1), 113-118.

Rodriguez, E., Lerma, C., Echeverria, J. C., & Alvarez-Ramirez, J. (2008). ECG scaling properties of cardiac arrhythmias using detrended fluctuation analysis. *Physiological Measurement, 29*(11), 1255-1266.

Roebroeck, M. E., Harlaar, J., & Lankhorst, G. J. (1993). The application of generalizability theory to reliability assessment: an illustration using isometric force measurements. *Physical Therapy, 73*(6), 386-395.

Roebroeck, M. E., Harlaar, J., & Lankhorst, G. J. (1993). The application of generalizability theory to reliability assessment: an illustration using isometric force measurements. *Physical Therapy, 73*(6), 386-395.

Rosano, C., Aizenstein, H., Brach, J., Longenberger, A., Studenski, S., & Newman, A. B. (2008). Gait measures indicate underlying focal gray matter atrophy in the brain of older adults. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 63*(12), 1380-1388.

Rosano, C., Aizenstein, H. J., Studenski, S., & Newman, A. B. (2007). A regions-of-interest volumetric analysis of mobility limitations in community-dwelling older adults. *Journals of Gerontology Series A: Biological Sciences and Medical Sciences, 62*(9), 1048-1055.

Rosano, C., Brach, J., Longstreth Jr, W. T., & Newman, A. B. (2006). Quantitative measures of gait characteristics indicate prevalence of underlying subclinical structural brain abnormalities in high-functioning older adults. *Neuroepidemiology, 26*(1), 52-60.

Rosano, C., Brach, J., Studenski, S., Longstreth Jr, W. T., & Newman, A. B. (2007). Gait variability is associated with subclinical brain vascular abnormalities in high-functioning older adults. *Neuroepidemiology, 29*(3-4), 193-200.

Rose, J., & Gamble, J. G. (2006). *Human Walking* (3rd ed.). PA: Lippincott, Williams & Wilkins.

Roudsari, B. S., Ebel, B. E., Corso, P. S., Molinari, N.-A. M., & Koepsell, T. D. (2005). The acute medical care costs of fall-related injuries among the U.S. older adults. *Injury, 36*(11), 1316-1322.

Sabatini, A. M., Martelloni, C., Scapellato, S., & Cavallo, F. (2005). Assessment of walking features from foot inertial sensing. *IEEE Transactions on Biomedical Engineering, 52*(3), 486-494.

Sadeghi, H. (2003). Local or global asymmetry in gait of people without impairments. *Gait & Posture, 17*(3), 197-204.

Sadeghi, H., Allard, P., Prince, F., & Labelle, H. (2000). Symmetry and limb dominance in able-bodied gait: a review. *Gait & Posture, 12*(1), 34-45.

Sadeghi, H., Allard, P., Prince, F., & Labelle, H. (2003). *Balance control and propulsion in gait of healthy young and elderly subjects*. Paper presented at the Summer Bioengineering Conference.

Sadeghi, H., Prince, F., Zabjek, K. F., & Allard, P. (2001). Sagittal-hip-muscle power during walking in old and young able-bodied men. *Journal of Aging & Physical Activity, 9*(2), 172-183.

Sadeghi, H., Prince, F., Zabjek, K. F., & Labelle, H. (2004). Simultaneous, bilateral, and three-dimensional gait analysis of elderly people without impairments. *American Journal of Physical Medicine & Rehabilitation, 83*(2), 112-123.

Sadeghi, H., Sadeghi, S., Prince, F., Allard, P., Labelle, H., & Vaughan, C. L. (2001). Functional roles of ankle and hip sagittal muscle moments in able-bodied gait. *Clinical Biomechanics, 16*(8), 688-695.

Saitoh, K., Ménard, A., & Grillner, S. (2007). Tectal control of locomotion, steering, and eye movements in lamprey. *Journal of Neurophysiology, 97*(4), 3093-3108.

Salat, D. H., Buckner, R. L., Snyder, A. Z., Greve, D. N., Desikan, R. S. R., Busa, E., et al. (2004). Thinning of the cerebral cortex in aging. *Cerebral Cortex, 14*(7), 721-730.

Salkeld, G., Cameron, I. D., Cumming, R. G., Easter, S., Seymour, J., Kurrle, S. E., et al. (2000). Quality of life related to fear of falling and hip fracture in older women: a time trade off study. Commentary: Older people's perspectives on life after hip fractures. *British Medical Journal, 320*(7231), 341-346.

Sands, M. L., Schwartz, A. V., Brown, B. W., Nevitt, M. C., Seeley, D. G., & Kelsey, J. L. (1998). Relationship of neurological function and age in older women. *Neuroepidemiology, 17*(6), 318-329.

Saremi, K., Marehbian, J., Xiaohong, Y., Regnaux, J. P., Elashoff, R., Bussel, B., et al. (2006). Reliability and validity of bilateral thigh and foot accelerometry measures of walking in healthy and hemiparetic subjects *Neurorehabilitation and Neural Repair, 20*(2), 297-305.

Sato, S., Hashimoto, T., Nakamura, A., & Ikeda, S. (2001). Stereotyped stepping associated with lesions in the bilateral medial frontoparietal cortices. *Neurology, 57*(4), 711-713.

Sattin, R., Rodriguez, J., DeVito, C., Wingo, P., & Group, t. S. (1998). Home environmental hazards and the risk of falls injury events among community dwelling older persons. *Journal of the American Geriatrics Society, 46*, 669-676.

Sattin, R. W., Lambert Huber, D. A., DeVito, C. A., Rodriguez, J. G., Ros, A., Bacchelli, S., et al. (1990). The incidence of fall injury events among the elderly in a defined population. *American Journal of Epidemiology, 131*, 1028-1037.

Scahill, R. I., Frost, C., Jenkins, R., Whitwell, J. L., Rossor, M. N., & Fox, N. C. (2003). A longitudinal study of brain volume changes in normal aging using serial registered magnetic resonance imaging. *Archives of Neurology, 60*(7), 989-994.

Scherder, E., Eggermont, L., Swaab, D., van Heuvelen, M., Kamsma, Y., de Greef, M., et al. (2007). Gait in ageing and associated dementias; its relationship with cognition. *Neuroscience & Biobehavioral Reviews, 31*(4), 485-497.

Schneider, E., Jahn, K., Dieterich, M., Brandt, T., & Strupp, M. (2008). Gait deviations induced by visual stimulation in roll. *Experimental Brain Research, 185*(1), 21-26.

Scott, V., Pearce, M., & Pengelly, C. (2009). Technical report: Hospitalizations due to falls among Canadians aged 65 and over, from http://www.phac-aspc.gc.ca/seniors-aines/publications/pro/injury-blessure/falls-chutes/tech/hospital-eng.php

Scuffham, P., Chaplin, S., & Legood, R. (2003). Incidence and costs of unintentional falls in older people in the United Kingdom. *Journal of Epidemiology and Community Health, 57*(9), 740-744.

Seely, A. J. E., & Macklem, P. T. (2004). Complex systems and the technology of variability analysis. *Critical Care, 8*, R367-R384.

Selby-Silverstein, L., & Besser, M. P. (1999). *Accuracy of the GAITRite system for measuring temporal-spatial parameters of gait.* Paper presented at the American Physical Therapy Association Combined Annual Conference, Washington, D.C.

Selles, R. W., Formanoy, M. A. G., Bussmann, J. B. J., Janssens, P. J., & Stam, H. J. (2005). Automated estimation of initial and terminal contact timing using accelerometers; development and validation in transtibial amputees and controls. *IEEE Transactions on Neural Systems and Rehabilitation Engineering, 13*(1), 81-88.

Serrador, J. M., Lipsitz, L. A., Gopalakrishnan, G. S., Black, F. O., & Wood, S. J. (2009). Loss of otolith function with age is associated with increased postural sway measures. *Neuroscience Letters, 465*(1), 10-15.

Shefchyk, S. J., & Jordan, L. M. (1985). Excitatory and inhibitory postsynaptic potentials in alpha-motoneurons produced during fictive locomotion by stimulation of the mesencephalic locomotor region. *Journal of Neurophysiology, 53*, 1345-1355.

Shefer, V. F. (1973). Absolute number of neurons and thickness of the cerebral cortex during aging, senile and vascular dementia, and Pick's and Alzheimer's diseases. *Neuroscience and Behavioral Physiology, 6*(4), 319-324.

Sherrington, C. S. (1910). Flexion-reflex of the limb, crossed extension-reflex, and reflex stepping and standing. *Journal of Physiology, 40*, 28-121.

Shibasaki, H., Fukuyama, H., & Hanakawa, T. (2004). Neural control mechanisms for normal versus parkinsonian gait. *Progress in Brain Research, 143*, 199-205.

Shik, M. l., & Orlovsky, G. N. (1976). Neurophysiology of locomotor automatism. *Physiological Reviews, 56*(3), 465-501.

Shumay-Cook, A., Brauer, S., & Woollacott, M. H. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Physical Therapy, 80*, 896-903.

Shumway-Cook, A., Ciol, M. A., Yorkston, K. M., Hoffman, J. M., & Chan, L. (2005). Mobility limitations in the Medicare population: prevalence and sociodemographic and clinical correlates. *Journal of the American Geriatrics Society, 53*(7), 1217-1221.

Shumway-Cook, A., & Woollacott, M. H. (2007). *Motor control: translating research into clinical practice* (3rd ed.). Baltimore, MD: Lippincott Williams & Wilkins.

Sinkjaer, T., Andersen, J. B., Ladouceur, M., Christensen, L. O., & Nielsen, J. B. (2000). Major role for sensory feedback in soleus EMG activity in the stance phase of walking in man. *Journal of Physiology, 523*(3), 817-827.

Snijders, A. H., van de Warrenburg, B. P., Giladi, N., & Bloem, B. R. (2007). Neurological gait disorders in elderly people: clinical approach and classification. *The Lancet Neurology, 6*(1), 63-74.

Sorsdahl, A. B., Moe-Nilssen, R., & Strand, L. I. (2008). Test-retest reliability of spatial and temporal gait parameters in children with cerebral palsy as measureed by an electronic walkway. *Gait & Posture, 27*(1), 43-50.

Speechley, M., & Tinetti, M. E. (1991). Falls and injuries in frail and vigorous community elderly persons. *Journal of the American Geriatrics Society, 39*, 46-52.

Springer, S., Giladi, N., Peretz, C., Yogev, G., Simon, E. S., & Hausdorff, J. M. (2006). Dual-tasking effects on gait variability: the role of aging, falls, and executive function. *Movement Disorders, 21*(7), 950-957.

Srikanth, V., Beare, R., Blizzard, L., Phan, T., Stapleton, J., Chen, J., et al. (2009). Cerebral white matter lesions, gait, and the risk of incident falls: A prospective population-based study. *Stroke, 40*(1), 175-180.

Starr, J. M., Leaper, S. A., Murray, A. D., Lemmon, H. A., Staff, R. T., Deary, I. J., et al. (2003). Brain white matter lesions detected by magnetic resosnance imaging are associated with balance and gait speed. *Journal of Neurology, Neurosurgery and Psychiatry, 74*(1), 94-98.

Steffen, T. M., Hacker, T. A., & Mollinger, L. (2002). Age- and gender-related test performance in community-dwelling elderly people: Six-Minute Walk Test, Berg Balance Scale, Timed Up & Go Test, and gait speeds. *Physical Therapy, 82*, 128-137.

Stein, P. K., Barzilay, J. I., Chaves, P. H. M., Domitrovich, P. P., & Gottdiener, J. S. (2009). Heart rate variability and its changes over 5 years in older adults. *Age Ageing, 38*(2), 212-218.

Stein, P. K., Barzilay, J. I., Chaves, P. H. M., Mistretta, S. Q., Domitrovich, P. P., Gottdiener, J. S., et al. (2008). Novel measures of heart rate variability predict cardiovascular mortality in older adults independent of traditional cardiovascular risk factors: The Cardiovascular Health Study (CHS). *Journal of Cardiovascular Electrophysiology, 19*(11), 1169-1174.

Stenbacka, M., Jansson, B., Leifman, A., & Romelsjo, A. (2002). Association between use of sedatives or hypnotics, alcohol consumption, or other risk factors and a single injurious fall or multiple injurious falls: a longitudinal general population study. *Alcohol, 28*, 9-16.

Stevens, J. A. (2005). Falls among older adults: Risk factors and prevention strategies. *Journal of Safety Research, 36*(4), 409-411.

Stevens, J. A., Corso, P. S., Finkelstein, E. A., & Miller, T. R. (2006). The costs of fatal and non-fatal falls among older adults. *Injury Prevention, 12*(5), 290-295.

Stevens, J. A., & Sogolow, E. D. (2005). Gender differences for non-fatal unintentional fall related injuries among older adults. *Injury Prevention, 11*, 115-119.

Stolze, H., Friedrich, H. J., Steinauer, K., & Vieregge, P. (2000). Stride parameters in healthy young and old women: Measurement variability on a simple walkway. *Experimental Aging Research, 26*(2), 159-168.

Stolze, H., Kuhtz-Buschbeck, J. P., Mondwurf, C., Boczek-Funcke, A., Jöhnk, K., Deuschl, G., et al. (1997). Gait analysis during treadmill and overground locomotion in children and adults. *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control, 105*(6), 490-497.

Strategic Injury Prevention Partnership (2000). *National Injury Prevention Action Plan: Priorities for 2001-2003*. Canberra, Australia: DHAC.

Tabachnick, B. G., & Fidell, L. S. (2001). *Using Multivariate Statistics* (4th ed.). MA: Pearson.

Takakusaki, K., Habaguchi, T., Ohtinata-Sugimoto, J., Saitoh, K., & Sakamoto, T. (2003). Basal ganglia efferents to the brainstem centers controlling postural muscle tone and locomotion: a new concept for understanding motor disorders in basal ganglia dysfunction. *Neuroscience, 119*(1), 293-308.

Takakusaki, K., Kohyama, J., Matsuyama, K., & Mori, S. (2001). Medullary reticulospinal tract mediating the generalized motor inhibition in cats: parallel inhibitory mechanisms acting on motoneurons and on interneuronal transmission in reflex pathways. *Neuroscience, 103*(2), 511-527.

Takakusaki, K., Oohinata-Sugimoto, J., Saitoh, K., & Habaguchi, T. (2004). Role of basal ganglia-brainstem systems in the control of postural muscle tone and locomotion. *Progress in Brain Research, 143*, 231-237.

Takakusaki, K., Saitoh, K., Harada, H., & Kashiwayanagi, M. (2004). Role of basal ganglia–brainstem pathways in the control of motor behaviors. *Neuroscience Research, 50*(2), 137-151.

Talkowski, J. B., Brach, J. S., Studenski, S., & Newman, A. B. (2008). Impact of health perception, balance perception, fall history, balance performance, and gait speed on walking activity in older adults. *Physical Therapy, 88*(12), 1474-1481.

Tapanainen, J. M., Thomsen, P. E., Kober, L., Torp-Pedersen, C., Mäkikallio, T. H., Still, A. M., et al. (2002). Fractal analysis of heart rate variability and mortality after an acute myocardial infarction. *American Journal of Cardiology, 90*, 347-352.

Tashiro, M., Itoh, M., Fujimoto, T., Fujiwara, T., Ota, H., Kubota, K., et al. (2001). 18F-FDG PET mapping of regional brain activity in runners. *Journal of Sports Medicine and Physical Fitness, 41*(1), 11-17.

Taylor, S. B., Begg, R. K., & Best, R. j. (2001). *Inter and intra-limb processes of gait control.* Paper presented at the Proceedings of the IEEE Engineering in Medicine and Biology Society Conference: Biomedicial Research in 2001, Monash University, Victoria, Australia.

Terrier, P., & Schutz, Y. (2003). Variability of gait patterns during unconstrained walking assessed by satellite positioning (GPS). *European Journal of Applied Physiology, 90*(5), 554-561.

Terrier, P., Turner, V., & Schutz, Y. (2005). GPS analysis of human locomotion: further evidence for long-range correlations in stride-to-stride fluctuations of gait parameters. *Human Movement Science, 24*, 97-115.

Thach, W. T., & Bastian, A. J. (2004). Role of the cerebellum in the control and adaptation of gait in health and disease. *Progress in Brain Research, 143*, 353-366.

The Consensus Committee of the American Autonomic Society and the American Academy of Neurology (1996). Consensus statement on the definition of orthostatic hypotension, pure autonomic failure and multiple system atrophy. *Clinical Autonomic Research, 6*, 125-126.

Thies, S. B., Richardson, J. K., & Ashton-Miller, J. A. (2005). Effects of surface irregularity and lighting on step variability during gait:: A study in healthy young and older women. *Gait & Posture, 22*(1), 26-31.

Thies, S. B., Richardson, J. K., DeMott, T., & Ashton-Miller, J. A. (2005). Influence of an irregular surface and low light on the step variability of patients with peripheral neuropathy during level gait. *Gait & Posture, 22*(1), 40-45.

Thomas, K. E., Stevens, J. A., Sarmiento, K., & Wald, M. M. (2008). Fall-related traumatic brain injury deaths and hospitalizations among older adults -- United States, 2005. *Journal of Safety Research, 39*(3), 269-272.

Thornbury, J. M., & Mistretta, C. M. (1981). Tactile sensitivity as a function of age. *Journal of Gerontology, 36*(1), 34-39.

Thorpe, D. E., Dusing, S. C., & Moore, C. G. (2005). Repeatability of temporospatial gait measures in children using the GAITRite electronic walkway. *Archives of Physical Medicine and Rehabilitation, 86*(12), 2342-2346.

Tinetti, M. E., De Leon, C. F. M., Doucette, J. T., & Baker, D. I. (1994). Fear of falling and fall-related efficacy in relationship to functioning among community-living elders. *Journal of Gerontology, 49*(3), M140-147.

Tinetti, M. E., Richman, D., & Powell, L. (1990). Falls efficacy as a measure of fear of falling. *Journal of Gerontology, 45*(6), P239-243.

Tinetti, M. E., Speechley, M., & Ginter, S. F. (1988). Risk factors for falls among elderly persons living in the community. *New England Journal of Medicine, 319*, 1701-1707.

Tinetti, M. E., & Williams, C. S. (1997). Falls, injuries due to falls, and the risk of admission to a nursing home. *New England Journal of Medicine, 337*(18), 1279-1284.

Tinetti, M. E., Williams, T. F., & Mayewski, R. (1986). Fall risk index for elderly patients based on number of chronic disabilities. *The American Journal of Medicine, 80*, 429-434.

Tromp, A. M., Pluijm, S. M. F., Smit, J. H., Deeg, D. J. H., Bouter, L. M., & Lips, P. (2001). Fall-risk screening test: A prospective study on predictors for falls in community-dwelling elderly. *Journal of Clinical Epidemiology, 54*(8), 837-844.

Tsang, W., & Hui-Chan, C. (2004). Effects of exercise on joint sense and balance in elderly men: Tai Chi versus golf. *Medicine & Science in Sports and Exercise, 36*, 658-667.

Uustal, H., & Baerga, E. (2004). Prosthetics and Orthotics. In S. J. Cuccurullo (Ed.), *Physical Medicine and Rehabilitation Board Review*. New York: Demos Medical Publisher, Inc.

Van de Crommert, H. W. A. A., Mulder, T., & Duysens, J. (1998). Neural control of locomotion: sensory control of the central pattern generator and its relation to treadmill training. *Gait & Posture, 7*, 251-263.

van Uden, C. J. T., & Besser, M. P. (2004). Test-retest reliability of temporal and spatial gait characteristics measured with an instrumented walkway system (GAITRite). *BMC Musculoskeletal Disorders, 5*(13).

VanSwearing, J. M., Paschal, K. A., Bonino, P., & Chen, T. W. (1998). Assessing recurrent fall risk of community-dwelling, frail older veterans using specific tests of mobility and the physical performance test of function. *The Journals of Gerontology. Series A, Biological Sciences and Medical Sciences, 53*(6), M457-M464.

Vardaxis, V. G., Allard, P., Lachance, R., & Duhaime, M. (1998). Classification of able-bodied gait using 3-D muscle powers. *Human Movement Science, 17*(1), 121-136.

Vaughan, C. L., Davis, B. L., & O'Connor, J. C. (1999). *Dynamics of Human Gait* (2nd ed.). Cape Town: Kiboho Publishers.

Vellas, B. J., Wayne, S. J., Romero, L. J., Baumgartner, R. N., & Garry, P. J. (1997). Fear of falling and restriction of mobility in elderly fallers. *Age Ageing, 26*(3), 189-193.

Verbaken, J. H., & Johnston, A. W. (1986). Population norms for edge contrast sensitivity. *American Journal of Optometry and Physiological Optics, 63*(9), 724-732.

Verghese, J., Holtzer, R., Lipton, R. B., & Wang, C. (2009). Quantitative gait markers and incident fall risk in older adults. *The Journals of Gerontology, 64A*(8), 896-901.

Verghese, J., Wang, C., Lipton, R. B., Holtzer, R., & Xue, X. (2007). Quantitative gait dysfunction and risk of cognitive decline and dementia. *Journal of Neurology, Neurosurgery & Psychiatry, 78*(9), 929-935.

Voss, A., Schroeder, R., Vallverdu, M., Cygankiewicz, I., Vazquez, R., Bayes de Luna, A., et al. (2008). Linear and nonlinear heart rate variability risk stratification in heart failure patients. *Computers in Cardiology, 35*, 557-560.

Walter, S. D., Eliasziw, M., & Donner, A. (1998). Sample size and optimal designs for reliability studies. *Statistics in Medicine, 17*, 101-110.

Wass, E., Taylor, N. F., & Matsas, A. (2005). Familiarisation to treadmill walking in unimpaired older people. *Gait & Posture, 21*, 72-79.

Watelain, E., Barbier, F., Allard, P., Thevenon, A., & Angue, J.-C. (2000). Gait pattern classification of healthy elderly men based on biomechanical data. *Archives of Physical Medicine and Rehabilitation, 81*(5), 579-586.

Webster, K. E., Merory, J. R., & Wittwer, J. E. (2006). Gait variability in community dwelling adults with Alzheimer disease. *Alzheimer Disease & Associated Disorders, 20*(1), 37-40.

Webster, K. E., Wittwer, J. E., & Feller, J. A. (2005). Validity of the GAITRite walkway system for the measurement of averaged and individual step parameters of gait. *Gait & Posture, 22*, 317-321.

Weinberg, L. E., & Strain, L. A. (1995). Community dwelling older adults' attributions about falls. *Archives of Physical Medicine and Rehabilitation, 76*, 955-960.

Weir, J. P. (2005). Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *Journal of Strength and Conditioning Research, 19*(1), 231-240.

West, B. J., & Scafetta, N. (2005). A multifractal dynamical model of Human gait. In G. A. Losa, D. Merlini, T. F. Nonnenmacher & E. R. Weibel (Eds.), *Mathematics and Biosciences in Interaction: Fractals in Biology and Medicine* (Vol. IV). Basel: Birkhauser.

Westlake, K., Wu, Y., & Culham, E. (2007). Sensory-specific balance training in older adults: effect on position, movement and velocity sense at the ankle. *Physical Therapy, 87*, 560-568.

Whelan, P. J. (1996). Control of locomotion in the decerebrate cat. *Progress in Neurobiology, 49*, 481-515.

Whitman, G. T., Tang, T., Lin, A., & Baloh, R. W. (2001). A prospective study of cerebral white matter abnormalities in older people with gait dysfunction. *Neurology, 57*(6), 990-994.

Whitney, S. L., Hudak, M. T., & Marchetti, G. F. (2000). The dynamic gait index relates to self-reported fall history in individuals with vestibular dysfunction. *Journal of Vestibular Research, 10*(2), 99-105.

Whitney, S. L., Marchetti, G. F., Pritcher, M., & Furman, J. M. (2009). Gaze stabilization and gait performance in vestibular dysfunction. *Gait & Posture, 29*(2), 194-198.

Whittle, M. W. (2002). *Gait Analysis: An Introduction*. Oxford: Butterworth Heinmann.

Whitton, T. L., Johnson, R. W., & Lovell, A. T. (2005). Use of the Rydel-Seiffer graduated tuning fork in the assessment of vibration threshold in postherpetic neuralgia patients and healthy controls. *European Journal of Pain, 9*(2), 167-171.

Whooley, M. A., Kip, K. E., Cauley, J. A., Ensrud, K. E., Nevitt, M. C., & Browner, W. S. (1999). Depression, falls and risk of fracture in older women. Study of Osteoporotic Fractures Research Group. *Archives of Internal Medicine, 159*, 848-490.

Wichmann, T., & DeLong, M. R. (1996). Functional and pathophysiological models of the basal ganglia. *Current Opinion in Neurobiology, 6*(6), 751-758.

Wiksten, D. L., Perrin, D. H., Hartman, M. L., Gieck, J. H., & Weltman, A. (1996). The relationship between muscle and balance performance as a function of age. *Isokinetics and Exercise Science, 6*, 125-132.

Wilson, C., Lorenzen, C., & Lythgo, N. (2002). *Validation of the GAITRite walkway system using 2-dimensional motion analysis*. Paper presented at the Proceedings of the 4th Australasian Biomechanics Conference.

Winter, D. A. (1991). *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological* (2nd ed.). Ontario, Canada: Waterloo Biomechanics.

Winter, D. A. (2005). *Biomechanics and Motor Control of Human Movement* (3rd ed.). New York: John Wiley and Sons.

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

Wittwer, J. E., Webster, K. E., Andrews, P. T., & Menz, H. B. (2008). Test-retest reliability of spatial and temporal gait parameters of people with Alzheimer's disease. *Gait & Posture, 28*(3), 392-396.

Wolfson, L. (2001). Gait and balance dysfunction: A model of the interaction of age and disease. *Neuroscientist, 7*(2), 178-183.

Woo, J., Ho, S. C., Lau, J., Chan, S. G., & Yuen, Y. K. (1995). Age-associated gait changes in the elderly: Pathological or physiological? *Neuroepidemiology, 14*(2), 65-71.

Wood, J. M., Lacherez, P., Black, A., Cole, M. H., Boon, M. Y., & Kerr, G. (2009). Postural stability and gait among older adults with age-related maculopathy. *Investigative Ophthalmololgy and Visual Science, 50*, 482-487.

World Health Organisation (2007). *WHO global report on falls prevention in older adults*. Geneva, Switzerland: WHO Press.

Yang, Y., Wang, R., Lin, K., Chu, M., & Chan, R. (2006). Task-oriented progressive resistance strength training improves muscle strength and functional performance in individuals with stroke. *Clinical Rehabilitation, 20*, 860-870.

Yardley, L., & Smith, H. (2002). A prospective study of the relationship between feared consequences of falling and avoidance of activity in community-living older people. *Gerontologist, 42*(1), 17-23.

Yazawa, S., Shibasaki, H., Ikeda, A., Terada, K., Nagamine, T., & Honda, M. (1997). Cortical mechanism underlying externally cued gait initiation studied by contingent negative variation. *Electroencephalography and Clinical Neurophysiology, 105*(5), 390-399.

Yogev, G., Plotnik, M., Peretz, C., Giladi, N., & Hausdorff, J. M. (2007). Gait asymmetry in patients with Parkinson's disease and elderly fallers: when does the bilateral coordination of gait require attention? *Experimental Brain Research, 177*, 336-346.

Yoshida, H., & Kim, H. (2006). Frequency of falls and their prevention (article in Japanese). *Clinical Calcium, 16*(9), 1444-1450.

You, S. (2005). Joint position sense in elderly fallers: a preliminary investigation of the validity and reliability of the SENSERite measure. *Archives of Physical Medicine and Rehabilitation, 86*, 346-352.

Yu, P.-L., Qin, Z.-H., Shi, J., Zhang, J., Xin, M.-Z., Wu, Z.-L., et al. (2009). Prevalence and related factors of falls among the elderly in an urban community of Bejing. *Biomedical and Environmental Sciences, 22*(3), 179-187.

Zehr, E. P., & Stein, R. B. (1999). What functions do reflexes serve during human locomotion? *Progress in Neurobiology, 58*(2), 185-205.

Zelenin, P. V., Orlovsky, G. N., & Deliagina, T. G. (2007). Sensory-motor transformation by individual command neurons. *Journal of Neuroscience, 27*(5), 1024-1032.

Zijlstra, W. (2004). Assessment of spatio-temporal parameters during unconstrained walking. *European Journal of Applied Physiology, 92*, 39-44.

Zijlstra, W., & Hof, A. L. (2003). Assessment of spatio-temporal gait parameters from trunk accelerations during human walking. *Gait & Posture, 18*, 1-10.

Zimmerman, M. E., Lipton, R. B., Pan, J. W., Hetherington, H. P., & Verghese, J. (2009). MRI- and MRS-derived hippocampal correlates of quantitative locomotor function in older adults. *Brain Research, 1291*, 73-81.