Effects of Enhanced Somatosensory Information on Postural Stability in Older People and People with Parkinson’s Disease

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Submitted by

Dr. Feng Qiu
MD; MMed; BM

Movement Neuroscience Program
Institute of Health and Biomedical Innovation
School of Exercise and Nutrition Sciences
Queensland University of Technology
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Postural sway, Postural stability, Standing balance, Gait, Ageing, Parkinson’s disease,
Somatosensory information, Textured surfaces
Abstract

The somatosensory system plays an important role in balance control and age-related changes to this system have been implicated in falls. Parkinson’s disease (PD) is a chronic and progressive disease of the brain, characterized by postural instability and gait disturbance. Previous research has shown that deficiencies in somatosensory feedback may contribute to the poorer postural control demonstrated by PD individuals. However, few studies have comprehensively explored differences in somatosensory function and postural control between PD participants and healthy older individuals.

The soles of the feet contain many cutaneous mechanoreceptors that provide important somatosensory information sources for postural control. Different types of insole devices have been developed to enhance this somatosensory information and improve postural stability, but these devices are often too complex and expensive to integrate into daily life. Textured insoles provide a more passive intervention that may be an inexpensive and accessible means to enhance the somatosensory input from the plantar surface of the feet. However, to date, there has been little work conducted to test the efficacy of enhanced somatosensory input induced by textured insoles in both healthy and PD populations during standing and walking.

Therefore, the aims of this thesis were to determine: 1) whether textured insole surfaces can improve postural stability by enhancing somatosensory information in younger and older adults, 2) the differences between healthy older participants and
PD participants for measures of physiological function and postural stability during standing and walking, 3) how changes in somatosensory information affect postural stability in both groups during standing and walking; and 4), whether textured insoles can improve postural stability in both groups during standing and walking.

To address these aims, Study 1 recruited seven older individuals and ten healthy young controls to investigate the effects of two textured insole surfaces on postural stability while performing standing balance tests on a force plate. Participants were tested under three insole surface conditions: 1) barefoot; 2) standing on a hard textured insole surface; and 3), standing on a soft textured insole surface. Measurements derived from the centre of pressure displacement included the range of anterior-posterior and medial-lateral displacement, path length and the 90% confidence elliptical area (C90 area). Results of study 1 revealed a significant Group*Surface*Insole interaction for the four measures. Both textured insole surfaces reduced postural sway for the older group, especially in the eyes closed condition on the foam surface. However, participants reported that the soft textured insole surface was more comfortable and, hence, the soft textured insoles were adopted for Studies 2 and 3.

For Study 2, 20 healthy older adults (controls) and 20 participants with Parkinson’s disease were recruited. Participants were evaluated using a series of physiological assessments that included touch sensitivity, vibratory perception, and pain and temperature threshold detection. Furthermore, nerve function and somatosensory evoked potentials tests were utilized to provide detailed information regarding
peripheral nerve function for these participants. Standing balance and walking were assessed on different surfaces using a force plate and the 3D Vicon motion analysis system, respectively. Data derived from the force plate included the range of anterior-posterior and medial-lateral sway, while measures of stride length, stride period, cadence, double support time, stance phase, velocity and stride timing variability were reported for the walking assessment. The results of this study demonstrated that the PD group had decrements in somatosensory function compared to the healthy older control group. For electrodiagnosis, PD participants had poorer nerve function than controls, as evidenced by slower nerve conduction velocities and longer latencies in sural nerve and prolonged latency in the P37 somatosensory evoked potential. Furthermore, the PD group displayed more postural sway in both the anterior-posterior and medial-lateral directions relative to controls and these differences were increased when standing on a foam surface. With respect to the gait assessment, the PD group took shorter strides and had a reduced stride period compared with the control group. Furthermore, the PD group spent more time in the stance phase and had increased cadence and stride timing variability than the controls. Compared with walking on the firm surface, the two groups demonstrated different gait adaptations while walking on the uneven surface. Controls increased their stride length and stride period and decreased their cadence, which resulted in a consistent walking velocity on both surfaces. Conversely, while the PD patients also increased their stride period and decreased their cadence and stance period on the uneven surface, they did not increase their stride length and, hence walked slower on the uneven surface. In the PD group, there was a strong positive association between decreased somatosensory function and decreased clinical balance, as assessed by the Tinetti test. Poorer somatosensory
function was also strongly positively correlated with the temporospatial gait parameters, especially shorter stride length.

Study 3 evaluated the effects of manipulating the somatosensory information from the plantar surface of the feet using textured insoles in the same populations assessed in Study 2. For this study, participants performed the standing and walking balance tests under three footwear conditions: 1) barefoot; 2) with smooth insoles; and 3), with textured insoles. Standing balance and walking were evaluated using a force plate and a Vicon motion analysis system and the data were analysed in the same way outlined for Study 2. The findings showed that the smooth and textured insoles caused different effects on postural control during both the standing and walking trials. Both insoles decreased medial-lateral sway to the same level on the firm surface. The greatest benefits were observed in the PD group while wearing the textured insole. When standing under a more challenging condition on the foam surface with eyes closed, only the textured insole decreased medial-lateral sway in the PD group. With respect to the gait trials, both insoles increased walking velocity, stride length and stride time and decreased cadence, but these changes were more pronounced for the textured insoles. The effects of the textured insoles were evident under challenging conditions in the PD group and increased walking velocity and stride length, while decreasing cadence. Textured insoles were also effective in reducing the time spent in the double support and stance phases of the gait cycle and did not increase stride timing variability, as was the case for the smooth insoles for the PD group. The results of this study suggest that textured insoles, such as those evaluated in this research,
may provide a low-cost means of improving postural stability in high-risk groups, such as people with PD, which may act as an important intervention to prevent falls.
List of Abbreviations

PD Parkinson’s disease
SEP somatosensory evoked potential
COP centre of pressure
AP anterior-posterior
ML medial-lateral
PL path length
ACE Addenbrooke’s Cognitive Examination
MMSE Mini Mental State Exam
MFES Modified Falls Efficacy Scale
ABC Activities-specific Balance Confidence
ADL activities of daily living
UPDRS Unified Parkinson’s Disease Rating Scale
GFQ Gait and Falls Questionnaire
FOG freezing of gait
BMI body mass index
PPA Physiological Profile Assessment
MET Melbourne Edge Test
TSA thermal sensory analyser
VSA vibratory sensory analyser
QST quantitative thermal sensory tests
VPT vibratory perception thresholds
NCV nerve conduction velocity
<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>AFH</td>
<td>ankle-fibular head</td>
</tr>
<tr>
<td>FP</td>
<td>fibular head-popliteal fossa</td>
</tr>
<tr>
<td>AMP</td>
<td>amplitude</td>
</tr>
<tr>
<td>FH</td>
<td>fibular head</td>
</tr>
<tr>
<td>PF</td>
<td>popliteal fossa</td>
</tr>
<tr>
<td>SNAP</td>
<td>sensory nerve action potential</td>
</tr>
<tr>
<td>WS</td>
<td>warm sensation</td>
</tr>
<tr>
<td>CS</td>
<td>cold sensation</td>
</tr>
<tr>
<td>HP</td>
<td>heat-induced pain</td>
</tr>
<tr>
<td>CP</td>
<td>cold-induced pain</td>
</tr>
<tr>
<td>EDB</td>
<td>extensor digitorum brevis</td>
</tr>
<tr>
<td>AH</td>
<td>abductor hallucis muscle</td>
</tr>
<tr>
<td>ASIS</td>
<td>anterior superior iliac spine</td>
</tr>
<tr>
<td>CGM</td>
<td>conventional gait model</td>
</tr>
<tr>
<td>GCV</td>
<td>general cross validation</td>
</tr>
<tr>
<td>ANOVA</td>
<td>analysis of variance</td>
</tr>
<tr>
<td>RMS</td>
<td>root mean square</td>
</tr>
<tr>
<td>SD</td>
<td>standard deviation</td>
</tr>
<tr>
<td>LSD</td>
<td>least significant difference</td>
</tr>
<tr>
<td>N/A</td>
<td>not applicable</td>
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</tbody>
</table>
Statement of Original Authorship

The work contained in this thesis has not been previously submitted to meet requirements for an award at this or any other higher education institution. To the best of my knowledge and belief, the thesis contains no material previously published or written by another person except where due reference is made.

Signature: __________________________

Date: __________________________
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Chapter 1

Introduction

Falls are a common, dangerous and frequently incapacitating problem that leads to a higher risk of admission into hospitals or nursing homes. Research indicates that 30% of people aged 65 years and over will experience a fall each year (Tinetti et al., 1994), as compared to 46-70% of patients with Parkinson’s disease (Pickering et al., 2007; Wood et al., 2002). As the average age of the population continues to increase in many countries, including Australia, the costs associated with the consequences of falling are of increasing concern and highlight the importance of developing falls prevention strategies across the world.

Among the factors implicated in falls, diminished somatosensory capacity is associated with postural instability and falling because the somatosensory system plays an important role in human balance control (Campbell et al., 1981; Horak, 2006; Maurer et al., 2000). The distal lower extremities, especially the feet, act as key receptors of somatosensory information gained through articulation with support surfaces during standing and walking. When the quality of somatosensory information is decreased, there may be a subsequent loss of balance control. Both older people and people with peripheral nervous system disorders, such as diabetic neuropathy (Inglis et al., 1994; Simoneau et al., 1995), account for the majority of the population with decrements in postural stability due to somatosensory feedback problems.

In an attempt to enhance the somatosensory information received from the feet and improve postural stability, different types of textured surface and insole devices have been developed. These include: 1) vibrating insoles comprising electromechanical
devices that stimulate the plantar surface of the feet (Priplata et al., 2006; Lalita et al., 2003; Liu, 2002; Priplata et al., 2002); 2) footwear inserts with a raised compliant ridge around the perimeter (Maki et al., 2008) used to stimulate the cutaneous mechanoreceptors near the soles of the feet; and 3), textured surfaces or textured insoles fitted in shoes that seem to improve motion discrimination through enhancing cutaneous feedback (Davids et al., 2004; Dixon et al., 2011; Hatton et al., 2011; Waddington & Adams, 2000). However, some electromechanical insole devices tend to be complex and expensive, and not suitable for use in daily life. Furthermore, compared to investigations of diabetic neuropathy (Lalita et al., 2003; Liu, 2002), few studies (Novak & Novak, 2006; Pratorius et al., 2003) have examined whether the enhancement of feedback from the soles of the feet can improve gait steadiness in patients with Parkinson’s disease. This is an important area of research, given that such patients are known to have serious problems with postural control.
Chapter 2

Literature Review

2.1 Falls

Following cardiovascular disease, cancer, stroke and pulmonary disorder, unintentional injury is the fifth leading cause of death in older adults, with falls constituting two-thirds of these deaths (Rubenstein et al., 1994). According to the Kellogg International Working Group on the Prevention of Falls in the Elderly (1987), a fall is defined as ‘unintentionally coming to the ground or some lower level and other than as a consequence of sustaining a violent blow, loss of consciousness, sudden onset of paralysis as in stroke or an epileptic seizure’. Recently, the Prevention of Falls Network European Consensus defined a fall as ‘an unexpected event in which the participant comes to rest on the ground, floor, or lower level’ (Lamb et al., 2005). In developed countries, life expectancy continues to rise, with the average life expectancy in Australia now 77 years for men and 83 years for women. Indeed, 1 in 3 people aged 65 years and over (Tinetti et al., 1994) and 2 in 3 people with Parkinson’s disease (Gray & Hildebrand, 2000) will experience at least one fall each year. Consequently, the health and well being of our ageing population has become a considerable social and economic issue.

2.1.1 Falls Occurrence

In the older population, the risk of experiencing at least one fall in a given year is about 28% for people aged 65-70 years (Prudham & Evans, 1981), but reaches 32% per year among those aged between 75 and 80 years (Tinetti et al., 1988). After the age of 75, the rates of falls-related injuries increase significantly compared to those in
the former group and continue to rise in the 95 years and older age group (Bradley & Pointer, 2009) (Figure 2-1).

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Figure 2-1: Age-specific rates of fall-related injuries for the Australian population between 2005 and 2006 (Bradley & Pointer, 2009)

Furthermore, people who have previously fallen have a greater risk of falling again (Bradley & Pointer, 2009; Lord et al., 2001; Lord et al., 1993) (Figure 2-2), highlighting that falls are a common, dangerous and frequently incapacitating problem in the ageing population (Tinetti et al., 1994). After following up with a group of community-dwelling persons aged 60 years or older on a weekly basis, Nevitt et al. (1989) reported that 31% had two or more falls over a 12 month period.
Figure removed for copyright reasons

Figure 2-2: Proportion of individuals who retrospectively reported falling over a 12 month period. (Lord et al., 2001)

Fall rates remain particularly high in older people who have a medical history, such as Parkinson’s disease (Pickering et al., 2007), stroke (Jorgensen et al., 2002) and diabetes (Wallace et al., 2002). Research involving the retrospective evaluation of falls in people with PD have demonstrated that as many as 60% of patients will experience one fall in a given year (Ashburn et al., 2001; Gray & Hildebrand, 2000), and around 50% will fall more than once (Balash et al., 2005; Bloem et al., 2001). Recently, a prospective study undertaken by Kerr et al. (2010) noted that approximately 48% of participants with early-stage PD reported a fall and 24% of these fallers were recurrent fallers.

Previous research has shown that falls frequency was strongly positively associated with disease severity and disease state, as measured using the Hoehn and Yahr scale (Balash et al., 2005). Furthermore, disease severity has been identified as a significant predictor of recurrent falls in people with PD (Bloem et al., 2001), with recurrent fallers having significantly more advanced symptoms than non-fallers. During the
early stages of the disease (Hoehn and Yahr stage 2), research shows that about 41% of patients reported falling, but as the disease progresses, the rate of falls increases to approximately 64% for patients mild to moderate symptoms (stage 3). Of those patients who report falling once or less, approximately 25% were classified as having a Hoehn and Yahr stage of 3, 4 or 5, but the number of falls recorded at stage 5 tended to be less than the early stages. The reduced rate of falling for patients in the latter stages of the disease (stage 5) might be explained by limitations on mobility for these individuals, which would limit their exposure to risky situations that may promote falls (Bloem et al., 2001; Gray & Hildebrand, 2000; Wood et al., 2002). Additionally, a prospective research demonstrated that 42% of participants reported falling in the previous year, which showed that the incidence of falling in optimally-medicated early-stage PD individuals was also a significant problem (Kerr et al., 2010).

An increased incidence of falls is also evident in people with other conditions, with around 51.3% of stroke patients suffering falls repeatedly (Psarakis et al., 2007). Similarly, research has shown that 54% of people with diabetes reported at least one fall a year, 64% reported at least one fall during the 2-year study period and 41% reported two or more falls (Wallace et al., 2002).

2.1.2 Falls Location and Circumstances

In a 10-year follow-up study, Saari et al. (2007) concluded that the number of falls occurring outside the home decreased as a function of age, while more falls were reported to occur inside the home on a level surface, reflecting the amount of time older people spend at home. Approximately 50% of falls in people aged 65 years and older in 2005-2006 occurred in the home (Bradley & Pointer, 2008). Berg et al. (1997) also reported that falls were more likely to occur at home and further analysis highlighted that falls were more likely to occur outside the home than indoors (Figure 2-3). With respect to fall circumstances, Berg et al. (1997) pointed out that trips and
slips were the most prevalent causes of falls, accounting for approximately 59%, followed by misplaced steps and loss of balance. Previous studies (Berg et al., 1997; Luukinen et al., 1994) have also shown that falls most often occurred when walking on level or uneven surfaces. Walking surfaces are reported as being the main cause of over 25% of falls-related injuries (Cohen & Compton, 1982). Many studies have illustrated that the risk of falling due to poor balance is increased on soft or uneven surfaces (Choy et al., 2003; Horak et al., 1990; Teasdale et al., 1991), while the risk of falls-related fractures is lower on wooden and carpeted floors (Simpson et al., 2004). These findings indicate that more attention should be given to better understanding the impact of different walking surfaces on changes to postural stability in elderly people. This analysis would facilitate the development of better preventative measures to minimise the incidence of falls in the ageing population.

Figure removed for copyright reasons

Figure 2-3: The percentage of falls occurred at home and away from home (Berg et al., 1997)
2.1.3 Consequences of Falls

Falls are the leading cause of injury-related hospital admissions in older people aged 65 years and over (Fobes & Aisbett, 2003). In Australia, falls account for 62% of accidental death in older citizens aged 75 years and over (Pointer et al., 2003) and explain at least 1% and 4% of all hospital admissions for older males and females respectively (Bradley & Harrison, 2007; Hindmarsh et al., 2009). However, it is important to consider that non-fatal falls, which include injuries such as bruises, cuts and abrasions, are quite common in older people and are often treated without hospitalisation. Following a study of 238 falls-related injuries, Sadigh (2004) reported that the most common types of injuries were bruises and swelling, followed by abrasions and cuts, with 33.8% of older fallers suffering from a previously fall-related injury. Berg et al. (1997) studied 91 falls, observing that 5% of the falls resulted in fractures, and 9% resulted in soft tissue injuries, such as cuts, abrasions or bruises. Furthermore, 31% of falls resulted in pain that lasted 2 days or more, while 21% resulted in pain that lasted for 5 days or longer, with major complications including limb and hip fractures. Although the proportion of falls-related fractures account for only about 10% of injuries (Sadigh et al., 2004), the consequences of these injuries can be serious. Fractures lead to muscle atrophy and reduced muscle strength, increased postural sway and an increased psychological fear of falling, which can lead to future falls and an impaired quality of life (Sherrington & Lord, 1998). Of the injuries commonly sustained as a result of falling in the elderly, hip fracture is considered to be one of the most severe. Many people who have recovered from a hip fracture may have subsequent falls, which can result in a subsequent hip or pelvic fracture that can lead to fatality (McKee et al., 2002; Shumway-Cook et al., 2005).

Falls are also considered as the leading cause of accidental death at home, accounting for two-thirds of such fatalities in older people over 65 years of age (King & Tinetti, 1995). A report from Statistics Canada (1991) indicated that the 60-69 year old age
group had about 6.1 falls-related deaths per 100000 people compared with 27.3 and 185.8 deaths in the 70-79 and 80+ yrs age groups, respectively. An 8-year post-fall follow-up illustrated that the risk of mortality was significantly greater for recurrent fallers and people who fell indoors (Bath & Morgan, 1999). Similarly, a 9 year follow-up of 300 older women (Sylliaas et al., 2009) reported a mortality rate of 41.7% and demonstrated that frequent falling was an independent predictor of serious falls-related injury. Frequent falls were significantly associated with mortality during that follow-up period and the relative risk of death when experiencing at least 2 falls was much higher in individuals who had experienced no falls.

An often overlooked consequence of falling is the psychological impact of the incident, as research shows that about 50% of people who fall are afraid of falling again (Tinetti et al., 1988). Excessive fear is an important falls-related consequence, which can limit physical functioning and prevent individuals from performing normal activities of daily living (Tinetti et al., 1994). Fear of falling is also common for people with PD, with approximately 50% of patients expressing a fear of falling that restricts their daily activities and contributes to social isolation (Bloem et al., 2001). The limitation of activity and social isolation that stems from a fear of falling, in turn, may lead to decreased cardiorespiratory function, reduced muscle strength, immobility, loss of independence, anxiety and depression. Therefore, the impact of experiencing a fall can have long lasting effects that impact both psychological and physiological aspects of the faller and the people around them.

2.1.4 Cost of Falls

Fallers have significantly higher long-term care and hospital costs than non-fallers (Carroll et al., 2008) and account for almost 70% of all injury-related costs in the older population. However, while the economic burden of falls is significant, the costs associated with falls are likely to continue to increase, as our society continues to age.
According to a summary provided by Heinrich et al. (2009), the main contributors to the direct costs of falls-related injuries were: 1) people in the older age classes, particularly females; 2) people in hospitals and/or long-term care facilities; and 3), people with fractures. The costs attributable to each fall are dependent on the severity of the fall and range from; 1) $2,044 to $25,955USD for each fall victim; 2) $1,059 to $10,913USD for each fall; and 3) $5,654 to $42,840 USD for each falls-related hospitalisation. Falls-related costs are expected to exceed $32 billion by the year 2020 in America (Englander et al., 1996), which is a huge financial burden to society.

Stevens et al. (2006) illustrated that the estimated direct medical costs reached $178 million USD for fatal and $19 billion USD for non-fatal injuries in 2000 in America. The fatal injury cost increased with age, but differed between genders, with males aged 65-74 years having 44% higher medical costs for fatal injuries than females. Despite this, both genders showed similar medical costs for fatal injuries in the 75-84 year age bracket, but the costs attributed to female fallers increased significantly in the 85+ age group. The change in trend for the older age bracket was likely related to the greater life expectancy of females (on average), which would result in fewer men in the 85+ yrs group. This study also pointed out that falls-related costs differed when injuries occurred in distinct body regions and varied from different types of injuries. Injuries to the lower extremities and the brain were the most frequent and costly, accounting for 79% of the costs. With respect to type of injury, fractures and internal injuries were the main types of falls-related injuries, which absorbed 73% of the costs. Unspecified types of falls-related injuries accounted for 25% of the fatal injury costs. Non-fatal injury costs from falling reached a total cost of $19 billion USD. Sixty three percent of the costs ($12 billion USD) were for hospitalizations, 21% ($4 billion USD) were for emergency department visits, and 16% ($3 billion USD) were for treatment in outpatient settings. Compared with the 65 to 74 year age bracket, the cost of non-fatal injuries doubled for those aged 75 to 84 years, but remained consistent for the older age brackets. As for females, medical costs were 2.8 times higher than for males.
Injuries to the lower and upper extremities accounted for 48% and 13% of the direct medical cost associated with non-fatal fall-related injuries, respectively. Fractures and superficial injuries were responsible for 75% of the non-fatal fall injury costs.

Falls represented the leading cause with respect to national injury costs in older groups in the USA. Australia displays the same tendency, since falls injuries were the most costly of the thirteen most significant injury categories reported in 1998-1999 in New South Wales (Fobes & Aisbett, 2003). The money spent on falls-related injuries reached $644 million AUD, and amongst the expenditure approximately $333 million AUD was in direct costs for the falls-related injuries (Figure 2-4). For the Australian population, the total cost of falls-related care for older people reached $566 million AUD in 2003-2004 (Bradley & Harrison, 2007). Moller (2003) also predicted that the total health cost spent on falls-related injury will increase three-fold to $1375 million AUD per annum by 2051. Since Queensland has a younger population demographic, it can be forecasted that there will be a steady rise in these costs over that period. Also due to the high levels of migration to the warmer climate in Queensland, there will be an increased number of retirees and ageing arrivals. Thus, the cost of falls is likely to experience the biggest rise in Queensland, compared to other states (Moller, 2003). The total health cost attributable to falls-related injury among persons aged 65 years and over is expected to reach $317.6 million AUD by 2051 in Queensland (Figure 2-5). Since the cost of falls-related injuries is expected to rise dramatically over the next 50 years, research into the possible causes of falls and falls-related injuries is urgently needed to help reduce the incidence of falls and to develop more effective preventative strategies.
Figure 2-4: Total cost of the thirteen identified mechanisms from 1998-1999 in New South Wales (Fobes & Aisbett, 2003)

Figure 2-5: Total fall related health cost (millions) trends for the Australian population by jurisdiction from 2001 to 2051 (Moller, 2003)
2.1.5 Risk Factors for Falls

In a systematic review of relevant literature, Deandrea et al. (2010) found that risk factors, such as history of falls, vertigo, Parkinson’s disease, fear of falling, gait problems, use of walking aids and use of antiepileptic drugs were closely associated with the incidence of falls. Of these risk factors, the strongest associations were reported for vertigo, Parkinson’s disease and fear of falling. While a history of falls, fear of falling and the use of walking aids are all factors that cannot be prevented, they might help identify persons at a higher risk of falling in the future and, hence, help identify suitable candidates for preventive interventions. However, this study did not calculate the associations between some other important factors such as muscle weakness, balance impairment and environmental hazards, which were measured in non-comparable ways. In a separate study, the main risk factors for falling in a group of elderly individuals were intrinsic risk factors, such as cognitive impairment, balance and gait disorders, as well as the use of sedative medications and hypnotics (Stalenhoef et al., 1997).

Similar research has been conducted by Lord et al. (2001), illustrating that risk factors for falling include reduced muscle strength and muscle tone, impaired motor coordination, reaction time and sensory systems, and the walking environment. In addition to these individual risk factors, it is widely recognized that the function and integration of sensory and motor systems declines significantly with age, which poses another major risk factor for falls in older people.

2.1.5.1 Musculoskeletal System

The musculoskeletal system plays an important role in maintaining one's capacity to complete normal activities of daily living. However, longitudinal and cross-sectional studies of muscle performance and structure have shown that muscle fibres are lost with increased age, which leads to a significant decrease in muscle mass and
physiological strength (Evans & Grimby, 1995). Musculoskeletal pathology, such as muscle weakness, is known to be a risk factor for falls in older people, as adequate muscle strength is required to maintain balance when the base of support is perturbed (Tinetti et al., 1988). Earlier research showed that elderly fallers demonstrated significantly reduced isometric knee extension and ankle dorsiflexion strength compared with their non-falling counterparts (Lord et al., 1994) and these findings were supported by a recent meta-analysis (Moreland et al., 2004). Collectively, these studies indicate that lower extremity muscle weakness is a significant factor when considering falls prevention in older people and improvements in these deficits might play an important role in improving the stability of the body during the performance of everyday activities.

Musculoskeletal pain, especially widespread pain may also be an important risk factor for falls in older people with disabilities (Leveille et al., 2002). Leveille et al. (2002) reported that the presence of pain in a group of older women lead to poorer physical performance, including slower gait, slower times for rising from a chair and lower levels of knee extension strength. Furthermore, the risk of falls in the pain group was more than 60% higher than that of women who experienced mild to no pain. Similar results were observed in a separate prospective study of 749 older people, which showed that chronic musculoskeletal pain was associated with a higher rate of falling (Leveille et al., 2009). The risk of falls was greater in people who experienced pain in multiple areas and in those who experienced a higher level of pain that interfered with activities (Leveille et al., 2009).

2.1.5.2 Reaction Time

Reaction time is the elapsed time between the presentation of a sensory stimulus and the subsequent behavioural response. A quick reaction to a loss of balance is very important to prevent falls. Lajoie et al. (2002) showed that reaction time was associated with sensory and motor functions and fallers had significantly longer
reaction times than non-fallers. Such an increase in reaction time is associated with increased postural sway in both lateral and anterior-posterior directions. It has also been shown that choice stepping reaction time can discriminate between multiple fallers and non-fallers, since a slow response to a perturbation that requires supraspinal processing in daily life would likely increase the risk of falling (Pijnappels et al., 2010). Reaction time also increases with ageing, which suggests that older people may not have enough time to react appropriately to a perturbation to prevent an injurious fall (Lajoie & Gallagher, 2004). Tucker et al. (2008) showed that compared to younger people, older participants exhibited significantly slower reaction times during both static and dynamic tasks. Furthermore, this study indicated that older participants had slower reaction times when asked to rapidly change direction during whole body motion, which led to increased postural rigidity and postural instability in these individuals. When faced with more challenging conditions such as standing on compliant surfaces, reaction time may play a more significant role in preventing falls (Lord & Ward, 1994), as participants are required to allocate attention to cope with the challenging condition during static postural control.

2.1.5.3 The Sensory Systems

Postural control is dependent on feedback from the somatosensory, vestibular and visual systems (Maurer et al., 2000), all of which provide information that allows body position and movement to be coordinated in three-dimensional space. However, due to an age-related decline in the functioning of these sensory systems (Lord & Ward, 1994; Peterka & Black, 1990), older people may show performance deficits in tasks that involve maintaining postural control in both static and dynamic movements.

Of these sensory systems, vision plays a key role in maintaining postural stability by providing the nervous system with synchronous information about body position and the environment. Compared with a stable visual environment, the magnitude of body sway will increase by approximately 36% when the visual surroundings become
unstable (Ehrenfried et al., 2003; Vidal et al., 1982). Hence, accurate perception of visual stimuli and depth are important in balance control as it provides a precise reference frame for controlling postural stability relative to one's surroundings. However, ageing is associated with a wide variety of degenerative changes to the visual system, which can lead to an inability to detect low contrast hazards, judge distances and perceive spatial relationships. Research has shown that poor visual function is an independent risk factor for falling, demonstrating that impaired visual acuity, poor contrast sensitivity and reduced depth perception contribute to increased postural sway, which increases the risk of falling (Lord, 2006).

However, while it has been well established that the visual and vestibular systems both play important roles in maintaining postural stability (Horak, 2006; Maurer et al., 2000; Redfern et al., 2001), the role of the somatosensory system in postural control is still not well understood.

The Somatosensory System

The somatosensory system is a diverse sensory system that comprises the receptors and processing centres involved in facilitating three different sensory modalities: 1) discriminative touch, which includes light touch, pressure, vibration and texture perception; 2) proprioception, which relies on sensory receptors in the joints and muscles to provide information on body position; and 3) temperature and nociception that receives information from perceived changes in temperature and pain peptides. Mechanoreceptors, which are sensitive to physical deformation due to touch, pressure or tensile strain are the main receptors of the somatosensory system (Bear, 2007). There are four kinds of mechanoreceptors in the skin that are responsible for providing information on touch, pressure and vibration and include the Meissner’s corpuscles, Pacinian corpuscles, Merkel’s disks and Ruffini endings (Figure 2-6). Alternatively, proprioceptive information, including joint position, muscle length and tendon strain is provided by; 1) muscle spindles, which are sensitive to muscle stretch;
2) Golgi tendon organs, which provide feedback on tendon tension; and 3), joint afferents. The sensation of pain depends on free nerve endings (nociceptors) that respond to potentially damaging stimuli, while thermoreceptors monitor changes in temperature.

Figure removed for copyright reasons

Figure 2-6: Four mechanoreceptors in the skin that are responsible for providing information on touch, pressure and vibration (Meissner’s corpuscles, Pacinian corpuscles, Merkel’s disks and Ruffini endings) (Bear et al., 2007)

Two pathways are responsible for passing somatosensory information from the sensory receptors in the peripheral nervous system to the central nervous system (Figure 2-7). The dorsal column-medial lemniscus pathway is responsible for relaying information related to discriminative touch, vibration and proprioception, while the spinothalamic pathway transmits information derived from pain and temperature sensation. With the integration of the visual and vestibular systems, the somatosensory system provides information on the position or motion of the body, which is used when controlling balance.
There are four types of sensory fibres involved in transmitting peripheral information to the central nervous system: 1) Aα-fibres; 2) Aβ-fibres; 3) Aδ-fibres; and 4) C-fibres. Aα-fibres have the greatest diameter and transfer speeds and are used for signal exchanges between the proprioceptors located within the skeletal muscles (muscle spindles) and tendons (golgi tendon organs) and the spinal cord. Aβ-fibres are also large in diameter and highly myelinated, allowing them to rapidly conduct important information pertaining to light touch and tactile sensation from the peripheral nerves to the central terminals. The Aδ-fibres have a smaller diameter, a thinner myelin sheath and higher activation thresholds and respond to thermal, mechanical and fast pain stimuli. Finally, C-fibres transmit slow pain to the central nervous system and have the slowest conduction velocity because they are the smallest diameter afferent and they are unmyelinated (Figure 2-8) (Bear et al., 2007; D'Mello & Dickenson, 2008; Yaksh & Hammond, 1982).
The four stratifications of primary afferent axons shown in order of fastest (Aα) to slowest conduction velocity (C) (Bear et al., 2007)

The somatosensory system, in both young and older adults, is foremost among the three sensory systems used for maintaining static balance (Bronseyn & Brandt, 2004). Peterka (2002) reported that in a well-lit environment with a firm base of support, the somatosensory information provided almost 70% of the information needed to regulate postural stability in healthy persons, while visual information provided 10%, and the vestibular system provided 20%. Even in the absence of any visual or vestibular deficits, reduced information from the somatosensory system is associated with increased postural instability during standing and walking, which may lead to falls in older people (Bergin et al., 1995; Lord et al., 1991; Lord & Ward, 1994).

During walking, cutaneous receptors on the plantar surface of the foot provide information about the initial contact between the foot and the ground (Fitzpatrick et al., 1994). This information is used to determine the site of contact on the foot and the forces associated with the activity, which subsequently affects the muscle activity in the lower extremities and helps maintain postural control (Burke et al., 1989;
Kennedy & Inglis, 2002; Perry, 2006). The sensory receptors located around the feet include; 1) slow adapting (Receptors that slowly return to their normal firing rate) mechanoreceptors that consist of Merkel and Ruffini corpuscle end-organs, and free nerve endings; and 2) fast adapting (Receptors that quickly return to their normal firing rate) mechanoreceptors that consist of Meissner corpuscles, Pacinian corpuscles, hair follicle receptors and free nerve endings (Iggo & Andres, 1982). In an earlier study, Kennedy and Inglis (2002) mapped the innervation territories and distributions of cutaneous mechanoreceptors on the plantar surface of the foot (Figure 2-9). As the soles of the feet are densely populated with mechanoreceptors, individuals are able to detect very slight changes in contact pressure under the feet, which facilitates the fine adjustment of postural control and locomotion. However, according to Inglis et al. (2002), the cutaneous sensitivity of the plantar surface of the foot is greater in the heel and toe regions and less for the ball and arch. A separate study has also shown that the sensitivity of each area of the foot is correlated with the heel to toe motion during walking (Kimmeskamp & Hennig, 2001). Thus, the somatosensory information received from the lower limbs, especially the sole of feet, is of utmost importance for maintaining dynamic postural stability in both the healthy and pathological aging populations (Kristinsdottir et al., 2001; Perry et al., 2001; Peterka, 2002). However, the performance of the somatosensory system is impaired in a distal to proximal direction in older people, largely due to a reduction in the rate of axonal transport (Uchida et al., 2004; Verd et al., 2000). These findings demonstrate the importance of maintaining and/or improving somatosensory function in healthy and pathological ageing populations so as to reduce their risk of falling.
Figure 2-9: Distribution of cutaneous mechanoreceptors in the foot sole (Kennedy & Inglis, 2002).

A. The receptive field for each receptor type (SA: slowly adapting, FA: fast adapting)
B. The approximate position of the afferent unit in the foot sole for all receptor types
C. Distribution of the total number of receptors and the accompanying threshold levels per unit

2.2 Older People and Falls

Postural stability is defined as an individual's ability to maintain the centre of gravity over the base of support, in keeping body balance while transitioning between dynamic and static states. Postural stability relies on the coordination of the sensory, motor and central nervous systems, but age-related changes in the neural, sensory and musculoskeletal systems can lead to postural instability, which increases the risk of falling (Maki & Mcilroy, 1996).
2.2.1 Standing Balance in Older People

Normal ageing leads to degeneration of sensory functions, which can increase postural sway during quiet standing. Reduction in visual acuity and other visual field defects can lead to poorer postural stability, in particular for lateral sway during standing (Paulus et al., 1984). Previous research has demonstrated that, compared to younger adults, postural sway increases to a significantly greater extent in older people when visual and somatosensory input is restricted (Ring et al., 1989). Age-related decreases in muscle strength is associated with poorer standing balance in older individuals (Bohannon et al., 1984).

Previous research has demonstrated that the average speed of postural sway was significantly greater for healthy older people who fell one or more times in a year compared with those who did not fall (Fernie et al., 1982). Furthermore, Melzer et al. (2004) reported that, compared to non-fallers, elderly fallers had a significantly longer centre of pressure path length and a higher centre of pressure velocity and mediolateral sway when standing on a firm surface with their eyes open. Postural stability was worse for fallers when standing on a foam surface, especially with eyes closed (Melzer et al., 2004; Shumway-Cook et al., 1997). Maki et al. (1994) reached a similar conclusion that postural sway amplitudes were much greater in those older adults who experienced a fall in a one-year period following standing balance measurements.

Standing balance is considered to depend on the coordination of visual, vestibular and somatosensory inputs. The collective contribution of these systems allows healthy individuals to correct the small amounts of postural sway that characterise quiet stance and come from constant small deviations from the vertical to maintain postural stability (Sheldon, 1963). However, excessive postural sway is the main feature of
instability during quiet standing, which has been defined by Daleiden (1990) as the ‘tendency for the centre of mass of the body to shift forward and back and from side to side during quiet stance’.

Various risk factors can lead to a loss of postural stability in elderly people. Using dynamic posturography, Wolfson et al. (1992) found that the standing balance of older people was particularly impaired compared to younger individuals when visual and tactile-propioreceptive inputs were occluded or distorted. Similar findings were presented in separate research involving the assessment of postural control in people ranging from 7 to 81 years of age. This research demonstrated that, compared with younger individuals, people aged 55 years or older showed larger increases in sway while standing on a fixed support surface with visual or somatosensory cues reduced (Peterka & Black, 1990). Furthermore, older people have higher frequency trunk accelerations that increase body sway and reduce postural stability in both the anterior-posterior and mediolateral directions during standing (Moe-Nilssen & Helbostad, 2002).

Somatosensory information from lower limbs, in particular the feet and ankles, plays an important role in the regulation of postural sway during quiet standing (Fitzpatrick et al., 1994; Kavounoudias et al., 2001; Meyer et al., 2004). The importance of the somatosensory system to balance control has been demonstrated in research where postural stability has been shown to decrease in both fallers and non-fallers when standing on a foam surface (Shumway-Cook et al., 1997).

2.2.2 Walking Stability in Older People

Maintaining postural stability requires not only the control of the body’s centre of mass position during quiet standing but also control of its balance during dynamic movement. During walking, individuals should accelerate the centre of gravity in a forward direction to initiate stepping and progress forward. Through placement of the
leading limb, the whole body can repeatedly initiate a forward lean and then recapture this momentum (Lord et al., 2001; Winter, 1995). Compared to standing, walking is more challenging for the human nervous system, due to initiating and terminating gait, turning and avoiding obstacles. It has been reported that more falls occur as a result of dynamic movements compared to static standing. Over half the falls in older individuals occur during transferring body segments, such as walking, stepping over and avoiding obstacles and stair walking (Berg et al., 1997; Cali & Kiel, 1995; Lord et al., 2001; Norton et al., 1997).

During walking, older individuals walk more slowly, take shorter strides and spend increased time in double limb support compared to younger individuals (Maki, 1997; Menant et al., 2009). Age-related changes in gait performance, such as decreased stride length and speed and prolonged periods of double support are believed to minimise postural instability during dynamic movements and create a more stable gait pattern (Maki et al., 1997). Recently, accelerometry has been used to assess dynamic postural stability during walking (Menz et al., 2003c) (Figure 2-10). Temporospatial gait parameters calculated from the acceleration signals demonstrated that older people displayed reduced velocity and increased step timing variability than younger adults and these differences were exacerbated when walking on an irregular surface. These findings suggest that irregular surfaces pose a more challenging walking condition for older people, which may require an even more conservative gait pattern to maintain postural stability during walking. Furthermore, during walking older individuals exhibited significantly smaller root mean square (RMS) acceleration values at the pelvis in all axes and at the head in the vertical axes on both firm and irregular surfaces, this decreased RMS was an adjustment for older participants to maintain a stable walking pattern and was closely associated with decreased walking speed (Menz et al., 2003c). Due to the slow walking speed, the peak positive anterior-posterior trunk acceleration value around initial foot contact was significantly lower for older people to maintain stability. Head motions were significantly slower and
smaller in elderly individuals, and trunk-pelvis coordination was more greatly affected, than in younger individuals during walking (Paquette et al., 2006). These differences are thought to comprise a compensatory strategy to maintain dynamic stability in the presence of age-related deficits in physiological functions. For example, elderly individuals may limit the amplitude of their movements to minimize the effects of the ensuing perturbations on postural stability. This may be particularly important given that muscle weakness is common in older people and, therefore, they are likely to have a reduced capacity to rectify their posture following a perturbation during walking. Research into the gait of older people has found that leg strength is strongly positively correlated with walking performance, such that stronger individuals take longer steps and walk faster (Ringsberg et al., 1999).

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Figure 2-10: Application of accelerometers during walking on different surfaces
(Menz et al., 2003c)

As with their effects on standing balance, age-related declines in the sensory and motor systems can lead to postural instability during walking. An earlier study investigating difference in walking stability for older people who fall demonstrated
that stride length, walking speed, arm swing and upper-lower extremity synchrony were all significantly impaired in older fallers (Wolfson et al., 1990). Furthermore, research indicates that multiple fallers have reduced and more variable cadence than non-fallers and spend significantly more time in the stance phase of the gait cycle (Hausdorff et al., 2001; Lord et al., 1996). According to Lord et al. (2001), there are a number of changes to the gait pattern that are associated with an increased risk of losing postural control and these include arrhythmic vertical and anterior-posterior accelerations of the body, reduced peak hip extension and increased cadence variability (Figure 2-11). Menz et al. (2003a) measured changes in walking stability using accelerometers placed on the pelvis and head during walking on different surfaces. Older people, with a high risk of falling, exhibited more variable step timing and less rhythmic acceleration patterns of pelvis and head. Furthermore, these individuals exhibited reduced harmonic ratios (an indicator of the rhythm of the acceleration signal) for the head and pelvis in the vertical and anterior-posterior directions, which is representative of a less stable walking pattern (Menz et al., 2003b). These changes were more apparent when walking on irregular surfaces (Menz et al., 2003a). Similar findings were presented by Yack and Berger (1993), who also found that older individuals with stability problems walked slower and had lower peak accelerations and less rhythmic trunk motion in the vertical and anterior-posterior directions compared with younger people while walking on a level surface.
Figure 2-11: Falls-related changes in gait patterns of older people during walking.
(Lord et al., 2001)

**Somatosensory Feedback in Older People**

While it is well understood that somatosensory feedback plays an important role in balance control during quiet stance, the role of somatosensory information in maintaining postural stability during walking has not been investigated extensively. Sensory declines affect postural stability during walking (Lord et al., 1996; Sakari-Rantala et al., 1998) and can adversely influence head and pelvis stability, which can reduce sensory input during dynamic movement (Menz et al., 2007).

Somatosensory function declines with age, leading to diminished proprioception and cutaneous sensation and impacting the ability of the central nervous system to integrate different sensory information sources to maintain postural control (Bugnariu & Fung, 2007). As such, age-related decrements in somatosensory perception have often been related to impaired mobility and falls in older people (Lord et al., 2001).
By combining some basic research findings with clinical evidence (Shaffer & Harrison, 2007), it has been demonstrated that advanced aging results in a decline in somatosensory structure and physiological function. Specifically, these age-related changes may be due to: 1) a decreased number of Pacinian corpuscles and Meissner’s corpuscles, which significantly reduces cutaneous sensation; 2) a reduced vibration perception threshold; 3) an increased capsular thickness and decreased diameter of intrafusal muscle fibres that comprise the muscle spindle leading to reduced sensitivity to changes in muscle length; and 4), a general decline in the number of joint receptors, reducing the sensation of accurate joint movements. Research has demonstrated that joint position sense is significantly affected in the great toe and ankle areas for older adults and they also demonstrate increased sensory thresholds for temperature, pain, tactile stimuli and two-point discrimination (Heft & Robinson, 2010).

Assessments of nerve function in older adults have shown declines in nerve conduction velocity and amplitude compared to younger participants and this may be due to the reduced number of nerve fibres that are evident in older individuals (Rivner et al., 2001). In addition to this, a separate study has also pointed out that older participants have longer latencies, smaller amplitudes, and slower velocities compared with a younger age group (Huang et al., 2009). Similarly, 36.3% of older participants with diabetes were shown to have abnormalities in F-wave (the waveform during a supramaximal stimulation of a motor nerve) latency for the peroneal nerve, while the abnormality rates for the sural nerve response latency and amplitude were slightly higher (58.3% and 62.7%, respectively) (Kong et al., 2008).

Research also revealed that older participants relied predominantly on the slower C-fibres for the transfer of painful sensations, while younger participants utilized additional input from the faster Aδ fibres. Furthermore older participants exhibited a significant elevation in thermal pain threshold compared to younger participants.
(Chakour et al., 1996). For a standard quantitative sensory testing protocol devised to assess somatosensory function, it was shown that older participants were significantly less sensitive than younger subjects, particularly in the foot area (Rolke et al., 2006). Predominantly, the decline in somatosensory function observed with aging and peripheral nerve disease occur in a distal to proximal fashion, such that the sensory loss will begin in the feet and hands and extend up the limb toward the torso. Studies have shown that two-point discrimination (Kenshalo et al., 1986; Shaffer & Harrison, 2007) and the perception of joint motion at the first metatarsophalangeal joint (Kokmen et al., 1978) are significantly decreased in older adults compared with their younger counterparts.

2.3 Parkinson’s Disease and Falls

Older people with medical conditions, such as diabetes mellitus, stroke patients and people with Parkinson’s disease have a higher risk of falls (Psarakis et al., 2007; Wallace et al., 2002). Somatosensory deficits in these groups are known to be more pronounced, leading to increased postural instability and a higher risk of recurrent falls (Priplata et al., 2003; 2006; Tyson et al., 2008). Somatosensory loss due to neuropathy decreases one’s ability to re-weight postural sensory dependence, which increases the risk of falling in both static and dynamic movements (Horak, 2006). Cavanagh (1992) found that peripheral neuropathy decreased postural stability during standing and walking. Furthermore, somatosensory deficits of the distal lower extremities resulting from diabetic neuropathy can lead to a marked decrease in the ability to maintain stable stance positions (Inglis et al., 1994; Simoneau et al., 1995). While it is well understood that motor dysfunction in people with Parkinson’s disease is due to the death of dopamine-generating cells in the substantia nigra, leading to postural instability (Connell et al., 2008; Inglis et al., 1994; Simoneau et al., 1995), little research has examined the effects of somatosensory information loss in people with PD. Given the important role that the somatosensory system plays in balance control, research focusing on better understanding the effect of PD on somatosensory
function and its relationship to postural balance during standing and walking is desperately needed.

2.3.1 Parkinson’s Disease

Parkinson's disease (PD) is a chronic and progressive degenerative disease of the brain that impairs motor control, speech, and other functions, caused by a loss of dopamine-producing cells in the substantia nigra pars compacta. PD is characterized by muscle rigidity, resting tremor, bradykinesia and loss of postural reflexes. Other symptoms may include high levels of cognitive dysfunction, poor sleep quality, subtle language problems and sensory abnormalities (Jankovic, 2008). In addition to these symptoms, a meta-analysis of six prospective studies of falling in Parkinson’s disease concluded that the 3-month fall rate for PD patients was 46%, with 21% of those PD individuals who had not previously fallen reporting an incident (Pickering et al., 2007). More than half of PD patients fall at least twice in a given year and 20% of them experience trauma including bone fractures and intracranial hematomas as a result of a fall (Balash et al., 2005; Koller et al., 1989; Yekutiel, 1993). Furthermore, other studies have demonstrated that people with PD have a much greater risk of hip fracture than age-matched controls, with proximal femur fractures being particularly prevalent (Johnell et al., 1992; Voaklander et al., 2010).

2.3.2 PD Patients and Standing Balance

People with Parkinson’s disease performed more poorly than healthy individuals of a similar age on tests of standing balance, particularly those patients who had a history of falls (Morris et al., 2000). During quiet stance, PD patients demonstrate increased postural sway, greater average postural sway area and increased length of sway path compared with healthy older adults while standing with eyes open and eyes closed (Schieppati & Nardone, 1991). Postural instability in such populations is partly
considered to be due to the increased abnormal postural reflexes or stiffness in lower limb muscles (Bloem et al., 1992; Hayashi et al., 1997). Many PD patients also display stooped posture with a moderate flexion of the knees and trunk, with elbows bent and arms adducted (Anette et al., 2001). Such postures can cause the centre-of-gravity to move outside the base of support, which promotes postural instability during standing balance. Other factors such as orthostatic hypotension, age-related sensory changes and the ability to integrate sensory inputs also contribute to changes in postural sway in people with PD during standing (Benatru et al., 2008).

2.3.3 PD Patients and Walking Stability

Gait disturbances in older people with Parkinson’s disease are considered to be associated with coordination problems and muscular rigidity (Van den Berg et al., 2000; Van Emmerik et al., 1999) and can lead to postural instability during walking. For example, deficits in movement coordination, such as reduced recruitment in leg extensor muscles, can contribute to the impaired walking in PD participants (Dietz et al., 1995) and are more obvious in gait initiation and switching of gait patterns. Schaafsma et al. (2003) reported that PD participants with a history of falling had increased stride-to-stride variability for stride time compared with controls. The authors postulated that their results suggested that the locomotor control system responsible for regulating stride time was significantly impaired among PD, which would increase their risk of falling. Furthermore, PD participants demonstrate more co-activation of antagonistic leg muscles during the support phase of the gait cycle and reduced motor unit recruitment in the leg extensors, which may contribute to impaired walking in this population (Dietz et al., 1995). Similarly, research demonstrates that the preparatory postural adjustments that typically precede stepping movements are significantly impaired in patients with Parkinson’s disease, which has significant implications for postural control during gait initiation (Gantchev et al., 1996).
People with Parkinson’s disease walk with reduced gait speed, shorter stride length, stooped posture, and reduced arm swing (Morris et al., 1994b; Rogers, 1996; Ueno et al., 1993). Additionally, Cole et al. (2010) indicated that PD patients had increased stride timing variability and spent more time in double support phase compared to controls. Furthermore, PD patients who prospectively reported falling demonstrated increased head motion in the medial-lateral direction (relative to walking speed) compared with those who did not fall and age-matched controls. These changes were reported to be exacerbated when the PD participants walked on a less stable and compliant surface, which would potentially increase the risk of falling (Cole et al., 2011). Using accelerometry, Latt et al. (2009) also studied the temporospatial gait parameters and acceleration patterns of the head and pelvis for a group of people with PD. The findings of this study revealed that, compared to healthy people, PD participants showed significant declines in walking speed and step length, and increased step timing variability. Additionally, those PD participants who retrospectively reported falling showed reduced RMS accelerations for the pelvis in all three directions and reduced head accelerations in the vertical and mediolateral directions compared to PD non-fallers and controls. Furthermore, PD fallers had less rhythmic accelerations (lower harmonic ratios) at the pelvis in the vertical and anterior-posterior directions than PD non-fallers and controls. PD non-fallers had significantly reduced RMS accelerations for the pelvis and head in all three directions compared to the controls. These findings highlight the dynamic stability deficits evident in people with PD and provide insight into the difficulties that are faced when attempting to maintain postural stability when walking. Walking slower contributed to reduced accelerations of the head and pelvis, but according to the harmonic ratios data, the smaller accelerations did not contribute to a more stable gait pattern for fallers.

During walking, abnormal gait in PD patients has been shown to be closely associated with the severity of the disease (Hoehn & Yahr, 1967; Sekine et al., 2004). In the early-stage of PD, postural instability is characterised by decreased speed, arm swing
and amplitude of leg movements (Brown & Steiger, 1996). However, as the disease progresses, freezing and start hesitation are more common, which can lead to an increased incidence of falls and complex gait disturbances in PD patients (Giladi et al., 1992). Gray and Hildebrand (2000) also found that many of the specific factors associated with falls were related to disease severity, including freezing, poor gait, postural hypotension and involuntary movements. Sudden freezing of gait is likely to disturb balance and contribute to falls. As for the symptoms and complications present at the time of falling, freezing was considered to be the most common, which was reported in connection with 36% of falls, followed by lightheadedness (23%), involuntary movements (21%), and tremor (18%). From the above, it provides support for the notion that the symptoms of PD, which can be difficult to manage via traditional methods, may impair one’s capacity to maintain postural control during movement.

In-shoe pressure distribution also has been studied during walking in PD individuals (Kimmeskamp & Hennig, 2001). Ten anatomical areas were chosen to determine the difference of sole pressures (Figure 2-12).

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Figure 2-12: Ten anatomical areas utilized in a pressure distribution study (Kimmeskamp & Hennig, 2001)

It was revealed that people with PD had significantly reduced peak pressures and relative loads in the lateral heel area (M2) and significantly higher loads in the medial
midfoot areas (M3) compare with controls. Intra-individual variability of the relative loads distribution showed increases in the heel area and decreases in medial mid- and forefoot areas in people with PD (Figure 2-13). This study demonstrated that, due to the different pressure distribution of soles, there were differences in the heel to toe motion of the foot during walking.

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Figure 2-13: Relative loads (Left) and variability of the relative loads (Right) for PD and control (*P<0.05, **P<0.01) (Kimmeskamp & Hennig, 2001)

**Somatosensory Feedback in PD**

Motor deficits have always been considered to play a major role in postural problems in Parkinson’s disease. However, recently researchers have focused more attention on the role of somatosensory feedback in adjusting posture control in PD individuals (Carpenter & Bloem, 2010; Konczak et al., 2009; Van Wegen et al., 2006). Compared to healthy people, PD patients, especially older individuals, suffer more from a loss of somatosensory information. Recent reviews have described a link between somatosensory disorders and poor postural control in patients with PD (Carpenter & Bloem, 2010; Konczak et al., 2009). Evidence for such a relationship was provided by Kerr et al. (2010), who showed that PD fallers had poorer peripheral sensation than PD non-fallers and that touch sensitivity was negatively correlated with postural sway in people with PD. Proprioception and related sensory information progressively diminish in PD, resulting in a decline in motor functioning that is likely due to an impaired capacity to evaluate proprioceptive feedback and coordinate voluntary and reflexive motor commands (Carpenter & Bloem, 2010).
Peripheral neuropathy has been shown to be very common in idiopathic Parkinson disease, with almost 55% of patients presenting with this condition (Toth et al., 2010). This study also reported that there were abnormalities in both sensory and motor nerve function of the lower extremities between idiopathic Parkinson disease and healthy age-matched controls. However, other studies showed that nerve conduction assessments were typically normal for PD, although some reported abnormal feelings of numbness, cold, burning or pain (Koller, 1984; Nolano et al., 2008). However, the different findings presented in these studies may have been influenced by the relatively small sample sizes and by the fact that the samples were comprised of participants who were more similar to the subgroup without neuropathy studied by Toth et al. (2010).

Parkinson’s disease patients experience difficulties in regulating the amplitude of movements, particularly when visual information is absent and patients are reliant on input from other sensory systems (Morris et al., 2000; 2001). This situation is further complicated by the fact that many participants with PD usually have disturbances in proprioceptive regulation of movements. Rickards and Cody (1997) used muscle vibration to investigate the effect of proprioceptive feedback on slow voluntary wrist movements. They found that, compared to healthy individuals, abnormally reduced degree of undershooting of slow voluntary wrist extension movements presented in patients with PD. There were impairments in the integration process of proprioceptive feedback, which resulted in abnormalities in sensory and motor information transfer in PD. A similar study applied a plantarflexor muscle vibration at the ankle (Khudados et al., 1999), revealing that there was a smaller degree of undershooting of slow, practiced, voluntary dorsiflexor movements in PD than age-matched controls. These results indicated that people with Parkinson’s disease may suffer from impaired
proprioceptive function, which may lead to postural instability. A recent study (Wright et al., 2010) used axial twisting tests to examine the kinesthetic sensitivity of PD patients during multiple joints movements. The results showed that PD participants were significantly less accurate in determining the direction of axial rotation than healthy controls and these axial symptoms may contribute to impairments in gait and balance (Wright et al., 2010). For people with PD, there is a degeneration of peripheral cutaneous receptors, which significantly impairs an individual’s capacity to discriminate two-point touch (Jobst et al., 1997; Nolano et al., 2008; Weder et al., 1999), tactile discrimination (Bruno et al., 1999), and static joint position and movement (Jobst et al., 1997; Nolano et al., 2008). Bloem et al. (1999) suggested that such sensory processing abnormalities may be contributory factors to balance impairments. Somatosensory input from the plantar surface of the foot was decreased in people with Parkinson’s disease, which impairs postural control (Pratorius et al., 2003). This somatosensory loss may be partially explained by a study conducted by Nolano (2008), who found that people with PD presented with significant epidermal nerve fibre loss resulting from nerve fibre swelling, increased nerve branching and nerve remodeling. Furthermore, separate research has shown that PD participants have a loss in free and encapsulated nerve endings, which contribute to increased thresholds for sensations, such as temperature, pain and touch. Schneider et al. (1986) concluded that proprioception was unquestionably impaired by PD and proposed that abnormal sensorimotor integration, by the defective basal ganglia, was an important causal factor in the genesis of Parkinsonian movement (Moore, 1987; Schneider et al., 1986).

Somatosensory evoked potentials (SEP) is a electrophysiological procedure that can be used to assess the sensory changes in peripheral neuropathy and other sensory-related diseases (Giblin, 1964). SEPs consist of a series of waves that reflect sequential activation of neural structures along the somatosensory pathways after peripheral nerve stimulation (Legatt, 2009). As the frontal cortex is closely connected
with the basal ganglia, it receives a vast array of sensory information from the peripheral nervous system (Cavada & Goldman-Rakic, 1989; Nauta, 1971) and many scalp sites have been used to evaluate this information in PD (Cheron et al., 1994; Rossini et al., 1993). Rossini et al. (1989) reported abnormalities in PD patients using SEP, indicating a significant decrease in response to sensory input at the frontal lobe area (frontal wave). The somatosensory waveforms used N and P, respectively to designate the negative and positive polarity of the recorded signals. Stimulations of SEP can be made from both the upper extremities and lower extremities (Gupta & Dorfman, 1981; Legatt, 2009), and reflect the peripheral distribution of the nerve being stimulated. In the upper limbs, the somatosensory evoked potentials are recorded by stimulating either the median or ulnar nerves, while the spinal segments (lumbar) and lower limbs are recorded after stimulation of the peroneal nerve or the posterior tibial nerve at the ankle. Due to the obvious amplitudes in lower limbs, lumbar N20 (at a latency of around 20 ms) and P37 (at a latency of around 37 ms) sites are often used to record the spinal response potential and cortical response, respectively (Toleikis, 2005; Walsh et al., 2005; Yiannikas & Vucic, 2008). Most of the above studies concentrated on cutaneous sensation of the hand, however little is known about the feedback transmission from the cutaneous afferents of the feet in people with PD.

Pratorius (2003) also pointed out that the sensitivity of the plantar surface of the feet reduces with the progression of Parkinson’s disease and suggested that the resultant decrease in proprioceptive awareness contributes to Parkinsonian gait. As a result, postural stability during walking may be affected due to the reduced sensitivity of the foot. Collectively, these findings suggest that somatosensory and proprioceptive input from the feet is impaired in people with PD, which may have serious implications on the safe performance of voluntary movements and the incidence of falls in this population. Despite this, little research has examined the role of the somatosensory
system in balance control or the relationship between somatosensory function and motor control.

2.4 Prevention of Falls

2.4.1 Artificial Somatosensory Information and Postural Stability

In a study conducted by Vuillerme and Pinsault (2007), it was reported that the somatosensory information received from the foot and the ankle was degraded when participants stood on a foam surface. They observed that the maintenance of balance on this more challenging surface was facilitated via increased cutaneous feedback from the foot and ankle. Postural stability will change according to these somatosensory inputs from the foot and ankle, which suggests that the central nervous system is able to integrate the somatosensory input from cutaneous mechanoreceptors in the foot to limit the postural destabilisation.

Artificial somatosensory changes can also affect postural stability, as shown by Diener et al. (1984), who decreased somatosensory input by using pressure cuffs placed around the calcaneus that induced ischemic hypoxia of the foot. The loss of somatosensory sensation from skin, pressure and joint receptors of the foot caused greater hip sway and larger excursion of the centre of foot pressure when the eyes were closed. Anesthesia of the feet via ischemic hypoxia also resulted in an increased use of the “hip strategy” for postural corrections, which involves increased shear forces on the standing surface, excessive hip movements and increased hip muscle activation (Horak et al., 1990). Furthermore, decreased somatosensory input by immersion of the feet in iced water increased the velocity of postural sway significantly, especially when individuals were faced with concomitant disturbances of visual or vestibular input (Magnusson et al., 1990a; 1990b; Perry et al., 2000). Vuillerme and Pinsault (2007) also recognized that somatosensory inputs from the
foot and the ankle were degraded by standing on a foam surface. Other studies have also shown that postural stability decreases when standing on a foam surface, which was due to altered somatosensory feedback caused by the degradation of information received from the plantar cutaneous mechanoreceptors (Horlings et al., 2008; Wu & Chiang, 1997). Collectively these findings signified that somatosensory information from the feet plays an important role in determining postural strategies, which may be subject to decrements when standing on uneven surfaces.

2.4.2 Enhanced Somatosensory Intervention for Changing Postural Stability

Postural instability is a major risk factor for falling, and numerous interventions have been attempted to prevent postural instability in an ageing population as well as populations with neuropathic symptoms. Techniques of biofeedback and enhanced sensory feedback have been used to improve postural stability by addressing internal factors that are thought to contribute to balance.

Several interventions have been undertaken to increase the postural stability by enhancing the somatosensory information received from the feet (Priplata et al., 2002; Wu et al., 2007). Kavounoudias et al. (1999) applied mechanical vibrations to different muscle tendons at the ankle and neck levels. They found that manipulating somatosensory information could induce whole-body postural responses that could be used to maintain standing balance. Increased cutaneous feedback provided by athletic tape adhered to the ankle joints has also been shown to benefit foot and ankle stability by ensuring proper foot placement (Simoneau et al., 1997; Vuillerme & Pinsault, 2007). Thus, it can be concluded that adjustment of somatosensory input from the foot and ankle can help control posture. Somatosensory stimulation may thus prove to be an effective way to improve balance control in older people and overcome age- and
disease-related losses in sensorimotor function. Certain levels of input can enhance the detection and transmission of weak signals in the sensory system, via a mechanism known as stochastic resonance. It has been shown that sub-sensory mechanical input applied to the feet of quietly standing participants can increase the detection of pressure changes on the sole of the feet to reduce postural sway (Moss et al., 2004; Priplata et al., 2002).

Anacker and Di Fabio (1992) demonstrated that artificially-enhancing the somatosensory input from the lower extremities by changing standing surface during balance tasks can enhance sensory feedback, which assisted with maintaining stance and controlling body sway in elderly people who fall. Priplata et al. (2002) also showed that postural sway can be decreased significantly during quiet standing through applying vibrating stimuli to feet (Figure 2-14). This finding indicated how noise-based devices, such as vibrating shoes inserts could be used to improve motor control and prevent falls in older people, making it an innovative intervention in prevention of falls.

Figure removed for copyright reasons

Figure 2-14: Actuators transmitted the displacement and force to the indentors located under each foot, inputting somatosensory information to the feet (Priplata et al., 2002)
2.4.3 Insole Devices to enhance Postural Stability

During walking, the cutaneous mechanoreceptors on the sole of the foot sense the state of contact between the foot and the ground and provide peripheral sensory input to the central nervous system for balance control (Do et al., 1994; Isamu et al., 1981). Hence, enhancing the quality of the peripheral sensation received from the feet may significantly contribute to balance improvement in high risk populations. Such somatosensory intervention devices have been tested, for example through randomly vibrating shoe inserts, which may enable people to overcome functional difficulties due to age or disease-related sensory loss. Priplata (2002) used vibrating surfaces to increase feedback and reduce postural sway during quiet standing and reported that older people gained more improvement in motor function with the added stimulation than younger participants. Similar results were found by Costa et al. (2007) who concluded that postural stability was improved when they applied vibrating noise to the sole of each foot. In addition, shoes with greater soles and surface contact area can provide better stability by preventing older adults falling (Allan et al., 2004).

In fact, insole gradient and texture is also an important factor to balance control. The declined function of balance control was associated with decreased somatosensory feedback related with sole shoes. Watanabe and Okubo (1981) found that postural sway was significantly reduced when standing on a textured surface in comparison to a smooth surface. Another study has illustrated that soft and thick soles can relieve pain by better distributing pressure under the feet, but softer and/or thicker insoles may also reduce postural stability by impairing one’s detection of foot position and pressure changes (Robbins et al., 1992). The same conclusion was made in athletes that hard-soled shoes provide more stability that the thick and soft midsoles (Robbins et al., 1994). On the other hand, greater sole contact area also can prevent older people from falling (Allan et al., 2004). These data suggest that textured insole may be an alternative way to improve postural stability by enhancing the somatosensory
feedback of the feet. Nevertheless, it is remains unclear what structural characteristics of an insole (e.g. material density) would provide the best benefits for postural control. As such, there is a clear need for further research to explore the efficacy of different texture and gradient designs on falls risks in older adults.

Textured insoles placed in athletic shoes have also been shown to improve movement control in athletes and lower the rate of ankle injuries (Waddington & Adams, 2000). A possible explanation for this finding is that enhanced feedback received from the plantar surface of the feet provides more accurate information about foot position. Currently, the noise-input applications have also been undertaken with clinical samples, especially those with peripheral nervous system disorders, for example diabetes mellitus with neuropathy. Through short-term mechanical feedback stimulation to the feet, the vibration and tactile perception in diabetic patients with moderate to severe neuropathy improved significantly during quiet standing (Lalita et al., 2003; Liu, 2002). Priplata and colleagues (2006) also studied the use of insoles to propagate vibration to the plantar foot surface, which caused similar improvements in balance control in diabetes patients. Leather insoles with sandpaper attached also changed the muscle activity in the tibialis anterior and soleus muscles, and induced more regular ankle kinematics and kinetics, which improved postural control in participants with multiple sclerosis (Kelleher et al., 2010). All these studies have suggested that insole devices might improve postural stability by enhancing sensory feedback into the foot area in populations with somatosensory deficits.

However, very little research (Jenkins et al., 2009; Novak & Novak, 2006) has been conducted to test the efficacy of noise-input application in PD patients. Given that reduced sensitivity of the foot may contribute to impaired balance control in PD patients (Pratorius et al., 2003), the use of insoles to enhance somatosensory stimulation could be an efficient way to improve postural balance in people with PD. The few studies conducted so far have found that stride variability was decreased and
walking speed, stride duration, stride length, cadence and single-limb support time were all increased. Furthermore, EMG analysis has also shown that the muscle activation of the tibialis anterior was normalized in PD participants with stimulation of the plantar surface of the feet. These data indicate that insole devices could make an improvement in balance control in the PD population, but research on treating these gait disturbances and reducing fall risk in PD is still quite limited.

To date, the majority of research conducted using insole devices have involved static standing tasks. Priplata et al. (2002) and Wu et al. (2007) both investigated participants when they stood comfortably on a platform with their hands at their side in two visual conditions: eyes closed and open (Figure 2-15). During their studies, Priplata et al. (2002) used a camera-based motion analysis system to collect data on parameters of postural sway such as range of the anterior-posterior and mediolateral shoulder displacement. However, falls often occur during dynamic movements in the home environment, such as walking on slippery surfaces (e.g. in the bathroom) or when suddenly distracted while walking. Despite this, only a limited number of studies have looked at the gait changes that occur when participants are fitted with shoe insert devices. In some of these studies, researchers have used a pressure-sensitive walkway (GAITRite system) to record measures of dynamic postural stability, such as the variability of the base of support, as well as stride length and velocity during walking (Novak & Novak, 2006; Wilson et al., 2008). Therefore, further research is needed to test whether insole applications can really improve the postural control during dynamic motion engaged in everyday tasks.
In addition, most of the insole devices assessed in earlier research have been electromechanical devices, which require a noise signal generator and a power supply to convey electric stimulation or vibration to the feet. Given that these devices are often expensive and too complex to be effectively applied to daily life, there is a real need to develop a more affordable and less complex insole device to improve the somatosensory information from the feet. In an attempt to address this need, several researchers have trialed adding different gradients and/or textures to the surface of an insole to examine the efficacy of this equipment in improving balance. Watanabe and Okubo (1981) found that postural sway was significantly reduced in healthy young participants when standing on a textured surface in comparison to a smooth surface. Similar results were reported by Hatton et al. (2011) who showed that medial-lateral sway was significantly decreased in younger adults when standing on a textured surface.

Postural performance has been measured before and after standing/walking with sandals with firm rubber nodules on the upper surface. The results showed that wearing the sandals led to significant improvement in standing balance due to the
temporarily enhanced information received from cutaneous sensory receptors in the plantar surface of the feet (Palluel et al., 2008; 2009). Corbin (2007) made some textured insoles by using plastic floor matting material, which had small rounded nubs raised about 1/4 cm off the plastic surface. A group of healthy young participants were asked to wear these textured insoles while their standing balance was assessed and the results demonstrated that the insoles decreased postural sway in bilateral stance, possibly due to enhanced somatosensory feedback from the feet. Maki et al. (2008) used footwear inserts with a raised compliant ridge around the perimeter of the foot to stimulate the cutaneous mechanoreceptors near lateral borders of the sole. The results of this study demonstrated that lateral stability was improved during stepping while wearing the insole, which compensated for the loss of plantar cutaneous sensation. Similarly, a study involving the use of a 3D motion analysis system, showed that injury-related excessive foot pronation was decreased during walking when wearing shoes with nodules located on the plantar-medial side of the foot (Ritchie et al., 2011). Despite these studies that support the efficacy of light-weight insole devices, Hatton et al. (2009) reported no changes in postural stability or lower limb muscle activity while standing on a textured surface with eyes open. However, it is important to note that the healthy older participants assessed by Hatton et al. (2009) were only examined under conditions that could be considered less challenging than used in other studies. As such, it is possible that the benefits of artificially-enhancing the somatosensory information from the feet may only be evident in situations where a greater emphasis is placed on the somatosensory system. Similarly, a study undertaken by Wilson et al. (2008) evaluated the influence of three different textured foot orthoses on postural stability and reported no significant effects. However, this study did not assess the barefoot condition and examined the effect of the textured foot orthotic after wearing it for 4 weeks, which may have caused some adaption to the stimulation of the foot.
2.5 Aims and Hypotheses

For the purposes of this research programme, three separate studies were conducted to evaluate somatosensory function and postural stability in healthy older adults and people with Parkinson’s disease. To date, there have been several studies that have demonstrated that older people and people with Parkinson’s disease suffer more from somatosensory loss than younger adults. However, there has been little research that has examined differences in somatosensory function in ageing people and patients with Parkinson’s disease. Furthermore, it remains unclear whether a relationship exists between the loss of somatosensory information and poor postural stability. Therefore, the primary objective of this research was to gain a better understanding of these two aspects through three studies:

Study 1

Aim 1: This was a feasibility study to determine whether textured insole surfaces can improve postural control by enhancing somatosensory information in young people and older people.

Hypothesis 1: Textured insole surfaces would improve postural control by enhancing somatosensory information, especially in older people.

Study 2

Aim 2: To determine what differences exist in physiological function and somatosensory function between older participants and people with Parkinson’s disease.

Hypothesis 2: People with Parkinson’s disease would perform poorer on tests of physiological function than healthy older participants. Furthermore, somatosensory function would be impaired to a greater extent in people with Parkinson’s disease compared with their healthy older counterparts.
**Aim 3:** To determine how changes in somatosensory information affect postural stability in older people and people with Parkinson’s disease during standing and walking.

**Hypothesis 3:** A decrease in somatosensory function will decrease postural stability in older people and people with PD people during standing and walking.

**Study 3**
Study 3 aimed to utilize inexpensive textured insoles to determine the effect of altering somatosensory information on postural stability during standing and walking in older people and people with Parkinson’s disease. Previous research suggests that artificially enhancing somatosensory feedback from the plantar surface of the feet, either via textured surfaces or footwear can reduce postural sway, which may significantly reduce falls risk in older people and people with Parkinson’s disease. Currently, many of the insole devices used to enhance feedback to the feet are too complex and expensive to integrate into daily life. Alternatively, insoles fitted within the shoe could provide an inexpensive and simple means of improving balance in high-risk populations. Despite the potential benefits of such insole devices, studies that have examined their efficacy in older populations and people with Parkinson’s disease are limited. Therefore, the objective of Study 3 was to determine the efficacy of textured insoles to artificially enhance somatosensory input and improve postural stability. Specifically, the aims of this study were: -

**Aim 4:** To determine whether textured insoles can improve postural stability in older people and people with Parkinson’s disease during standing and walking.

**Hypothesis 4:** Textured insoles will lead to better balance and postural control than smooth insoles and barefoot conditions in older people, and especially older people with Parkinson’s disease.
Chapter 3

Study 1

This feasibility study examined the efficacy of a newly designed textured insole surface for reducing postural sway in healthy younger and older adults during standing balance on two textured surface of different densities.

3.1 Introduction

Age-related declines in sensory and motor function can result in postural instability and an increased risk of falls leading to injury, hospitalization and mortality (Forbes & Aisbett, 2003). One third of community-dwelling older people over 65 years fall at least once a year and many suffer multiple falls (Tinetti et al., 1994). Accurate detection and integration of somatosensory information from the feet is important for balance control (Bronstein & Brandt, 2004). Degeneration of peripheral sensory receptors, exemplified in diabetic peripheral neuropathy (Simoneau et al., 1995), can lead to a diminished capacity to detect information from the soles of the feet during interactions with the external environment (Inglis et al., 1994; Perry, 2006). Diminished somatosensory function has also been identified as a significant age-related change and is believed to be a significant contributor to postural instability and falls in older people (Lord et al., 2001). Older participants have reduced sensitivity for the plantar surface of the foot than younger individuals (Inglis et al., 1994; Kenshalo, 1986), which can increase postural sway (Pyykko et al., 1990).

Artificially reducing somatosensory information, by cooling (Magnusson et al., 1990a) or local anaesthetic ischemia, induced by hypoxic anaesthesia of the feet and ankles (Horak et al., 1990), can increase postural sway. Standard balance assessments which
require participants to stand on a foam surface also reduce the reliability of somatosensory information and increase postural sway. These effects are exacerbated when vision is excluded and greater reliance is placed on somatosensory information (Lord et al., 1991). The effects of standing on a foam surface have been equated to diabetic peripheral neuropathy (Inglis et al., 1994; Simoneau et al., 1995). More recently, Patel et al. (2011) reported that standing on a foam surface with eyes closed decreased the reliability of somatosensory information of feet. This observation was also supported by findings of Vuillerme and Pinsault (2007) who recognized that somatosensory inputs from the feet were degraded by standing on a foam surface.

Previous research has provided some evidence that artificially enhancing cutaneous information can change postural sway and potentially improve postural stability (Priplata et al., 2002; Wu et al., 2007). Kavounoudias et al. (2001) showed that supra-threshold vibratory stimulation of feet during quiet stance altered postural sway; bilateral stimulation of the forefoot resulted in backward leaning. Similarly, sub-threshold mechanical vibration applied to the soles of feet increased the detection of plantar pressure changes, with a consequent reduction in postural sway in older people (Priplata et al., 2002) and peripheral neuropathy patients (Priplata et al., 2006). However, practically, vibratory devices can be expensive and complex to adopt as effective interventions to decrease postural sway. Clearly there is a need to develop and evaluate simple and inexpensive interventions that can enhance somatosensory feedback from the feet and diminish postural sway.

Recent research has suggested that passive devices may provide an inexpensive and effective alternative to decrease postural sway. Palluel et al. (2009) reported reduced postural sway during quiet stance for older people while wearing sandals with firm rubber nodules. However, sandals may not be suitable footwear for all individuals and their use can be limited by environmental, work and social constraints. Furthermore, sandals and other footwear have been suggested to introduce different confounding
effects due to differences in shoe design and construction (Hatton et al., 2011). Additionally, Palluel et al. (2009) only evaluated postural sway on a firm surface and did not randomize the order of testing conditions, which may have introduced a learning effect into their results. Assessing postural sway while standing on a foam surface may decrease the reliability of somatosensory information from the feet and provide a more useful way to evaluate the effect of somatosensory changes on postural sway, especially without visual input. Similarly, Corbin et al. (2007) reported reduced postural sway in younger participants while wearing insoles which had a textured pattern; but their effectiveness in older people was not assessed. Recently Hatton et al. (2011) noticed that mediolateral sway was decreased when standing on textured surfaces in older people. However, the performance of a younger control group was not evaluated in their study.

The aim of this study was to examine the efficacy of a newly designed textured insole surface for reducing postural sway in healthy younger and older adults during standing balance with two different texture surface levels. The results of this study will also be used to determine which kind of textured insole surface would be most suitable for insertion into shoes for Study 3. Due to ageing effects on the peripheral nervous system it was expected that insole surface attenuation effects on postural sway were likely to be greater in the older groups, compared to the younger groups, especially under conditions where peripheral somatosensory information was more important in maintaining postural stability.

3.2 Methods

3.2.1 Participants

Seven elderly adults (4 males and 3 females; mean age 72±4 years; Body Mass Index (BMI) 25.6±2.2 kg/m²) and ten healthy young adults (6 males and 4 females; mean age 27±3 years; BMI 22.3±2.4 kg/m²) participated in this study. Elderly participants
were randomly selected from a pre-existing database of healthy older adults who had expressed an interest in being involved in this type of research. All participants were free of significant cognitive impairment (Mini Mental State Examination total score ≥24) and other illnesses that may have interfered with static standing or dynamic motion.

Prior to their involvement, participants were briefed on the benefits and risks of this study and all gave written informed consent to participate in this research program. The testing procedures were approved by the Queensland University of Technology Human Research Ethics Committee.

3.2.2 Test Protocol

To examine the influence of altering somatosensory information on postural stability, participants performed standing balance tests under three insole surface conditions: 1) barefoot; 2) hard textured insole surface (320 density ethylene-vinyl acetate); and 3), soft textured insole surface (270 density ethylene-vinyl acetate). Both insole surfaces (International Children’s Orthotic Laboratory, Australia) were 1.5mm thick and had granulations with a diameter of 5.0mm and a height of 3.1mm that were distributed evenly across the upper surface (Figure 3-1). The order of insole surface conditions and assessments were randomized for each participant.

Figure 3-1: Hard insole surface (Black) and soft insole surface (White).
For each of the insole surface conditions, participants were tested under two vision conditions (eyes open, closed) on two standing surfaces (firm, foam). During the experiments, participants stood as still as possible on a force plate (HUR Labs OY, Finland), looking straight ahead to fixate a cross positioned at eye level and 1.5m away, with their feet 10cm apart and their hands at their sides. Data for four 30s trials were collected at 100Hz.

In accordance with previous research (Hatton et al., 2011; Melzer et al., 2004; Priplata et al., 2002; Priplata et al., 2006), this study used measurements derived from the displacement of centre of pressure (COP) and included the range of anterior-posterior (AP) and medial-lateral (ML) COP displacement, AP and ML sway standard deviation (SD), path length (PL) and the 90% confidence elliptical area (C90 area).

3.2.3 Statistical analyses

A mixed model Analysis of Variance (ANOVA) with one between-participant (younger; older) and three within-participant factors, including insole surface (barefoot; hard; soft insole surface), vision (eyes open; closed) and standing surface (firm; foam) was used to compare postural control. A separate analysis examining the potential interaction of age (younger, older) and insole surface (barefoot; hard; soft insole surface) was undertaken in an ‘eyes-closed’ condition standing on a foam surface. Violations of the sphericity assumption for repeated measures variables were checked using Mauchley’s test of sphericity. When a violation of this assumption was apparent, the Greenhouse-Geisser correction procedure was used to adjust the degrees of freedom of the error term for the F ratios. Post-hoc comparisons were undertaken using Fisher’s Least Significant Different (LSD) test. Statistical significance was set at the 95% confidence level (P<0.05). Data were analyzed using the Statistical Package for Social Sciences (SPSS V17.0, Chicago, IL, USA).
3.3 Results

Clear differences in postural sway as a function of age, insole surface and standing surface were revealed by a significant Group*Surface*Insole interaction for C90 area, PL, AP and ML sway and ML sway SD (P<0.05).

**Postural Sway (C90) Area.** The older group revealed a greater postural sway area than the younger group in the barefoot condition on firm and foam surfaces (P<0.05). Both insole surfaces reduced the C90 area for the older group to an area equivalent to that observed in the younger group on the firm surface. However, when standing on the foam surface, only the soft insole surfaces reduced the C90 area of the older group to be equivalent to that observed in the younger group. Overall, the measure of postural sway area revealed that only the older group benefitted from the use of different insole surfaces. No significant differences were observed for the younger participants between the barefoot, hard and soft insole surface conditions on either the firm or foam surfaces (Figure 3-2).

![Graphs showing postural sway](image-url)

Figure 3-2: Mean (+1 SD) C90 area for the older (black) and young (grey) participants during the four standing conditions.
Path Length (PL). On the firm and foam surfaces, PL for the older group was greater than the younger group under all three insole surface conditions (P<0.001) (Figure 3-3). There was a significant and progressive decrease in PL from the barefoot to hard to soft insole surface conditions for the older group, but only when standing on the foam surface (P<0.05), and this trend was more pronounced under the eyes closed condition. The only beneficial effect for the young group, relative to the barefoot condition, emerged when standing on the hard insole surface on a firm surface (P<0.05).

Anterior-Posterior (AP) Postural Sway and AP Sway SD. The older participants demonstrated increased AP sway relative to the younger group under the three insole surface conditions on the firm surface (P<0.05). The older group also demonstrated increased AP sway on the foam surface in the barefoot and soft insole surface conditions (P<0.05), but not with the hard insole surface (p=0.081). Both insole surfaces significantly decreased AP sway relative to the barefoot condition for the

Figure 3-3: Mean (+1 SD) Path length for the older (black) and young (grey) participants during the four standing conditions.
older group when standing on the foam surface (P<0.05). For the younger group, only the hard insole surface decreased AP postural sway relative to the barefoot condition when standing on the firm surface (P<0.05) (Figure 3-4). There were no significant differences observed in the Group*Surface*Insole interaction for AP sway SD.

![Graphs showing AP sway for older and young participants](image)

Figure 3-4: Mean (+1 SD) Anterior-posterior sway for the older (black) and young (grey) participants during the four standing conditions.

**Medial-Lateral (ML) Postural Sway and ML Sway SD.** Both ML sway and ML sway SD were greater for the older group compared to the young group in the barefoot and hard insole surface conditions (P<0.05), but had reduced to a level equivalent to performance of the younger group under the soft insole surface condition (Figure 3-5). For the older group there was a significant reduction in ML sway from the barefoot condition to the hard insole to the soft insole surface on both firm and foam surfaces (P<0.05). For the younger group, the hard and soft insole surfaces were equally effective in decreasing ML sway relative to the barefoot condition on both surfaces. For both groups, the hard and soft insoles decreased the ML sway SD values more than in the barefoot condition on the firm surface (P<0.05). Only the older group
demonstrated reduced ML sway SD on the foam surface (P<0.001). However, no significant changes in ML sway variability were noticed between the two textured surfaces (P>0.05).

Figure 3-5: Mean (+1 SD) Mediolateral sway for the older (black) and young (grey) participants during the four standing conditions.

**Foam Eyes-Closed Condition**

Figure 3-6 depicts differences in COP for a representative older and younger participant under each insole surface condition while standing on a foam surface with eyes closed.
Figure 3-6: Representative data for the old and young participants while standing with eyes closed on the foam surface. Data portray postural sway during the barefoot condition (a, d); on the hard textured surface (b, e); and on the soft textured surface (c, f).

When standing on the foam surface with eyes closed, the older group showed a significant reduction in ML sway, PL and C90 area from barefoot to the hard to the soft insole surface (P<0.05). It was observed that ML sway SD values were significantly decreased by standing on both insole surfaces compared to the barefoot condition (P<0.001). AP sway and PL were greater for the older participants compared to the younger group in the three insole surfaces conditions (P<0.05). ML sway and C90 area were greater for the older group in the barefoot and hard insole surface conditions (P<0.05), but their postural sway had reduced to an equivalent level to the young group under the soft insole surface condition. ML sway SD was greater for the older group in the barefoot condition (P<0.001), then was reduced to a similar level as observed in the younger group in both insole surface conditions (Figure 3-7).
Figure 3-7: Mean (+1 SD) Postural sway parameters for the older (black) and young (grey) participants during the foam eyes closed condition.

### 3.4 Discussion

This study examined the efficacy of inexpensive textured insole surfaces in reducing postural sway under conditions that challenged the somatosensory system in younger and older participants. Study 1 also determined the suitable insole surface for inserting into shoes in future planned studies.

Consistent with previous research (Moe-Nilssen & Helbostad, 2002), the current study demonstrated that, overall, older participants displayed greater postural sway than younger participants during standing with bare feet. However, the older group demonstrated a significant and progressive decrease in postural sway from the barefoot to the hard and the soft insole surfaces. A possible mechanism is that the textured insole surfaces may have produced higher plantar pressures at the elevated
parts of the textured sole, providing stronger sensory stimulation to the
mechanoreceptors. Additionally, increased pressure gradients may have been present
between the hills and valleys across the textured sole pattern, creating additional
stimulation to the mechanoreceptors. This effect may have resulted in an overall
increased neural feedback from the cutaneous receptors to the central nervous system
(Hidaka et al., 2000; Manjarrez et al., 2003), possibly contributing to improved
postural control.

Smaller improvements in postural sway were observed for the younger participants
when standing on the textured surfaces. While the overall area of postural sway was
unchanged by the textured insole surfaces, there were some small but significant
decreases in measures of path length, AP and ML sway and ML sway SD. Importantly, the improvements observed in the younger participants were
predominantly recorded while standing on the firm surface with the hard insole
surface. The soft insole surface only reduced ML sway on both surfaces. It is unclear
why the hard insole surface reduced postural sway in the younger group and not the
older group, but impaired ability in the latter to scale the postural response due to age-
related loss of peripheral cutaneous sensory function may have been a contributing
factor (Inglis et al., 1994).

In agreement with our findings, Corbin et al. (2007) reported decreased postural sway
during quiet stance for younger participants wearing textured insoles and Palluel et al.,
(2008; 2009) reported significant reductions in ML sway for younger and older
participants wearing sandals comprising textured rubber nodules. Furthermore, recent
research by Hatton et al. (2011) reported significant reductions in ML sway for older
participants while standing on textured surfaces. Given that Maki et al. (1996)
reported that a loss of lateral stability was closely associated with increased risk of
falling, these results may indicate that reducing ML sway may be of benefit to falls
prevention in older people. Taken together, these data suggest that ML sway may be
an important parameter to consider when appraising the efficacy of insole interventions in improving standing balance in future research. Furthermore, the study by Hatton et al. (2011) has indicated that standing on textured surfaces may provide different effects on postural stability compared to footwear, due to the possible confounding effects of different shoe construction characteristics.

Relative to the firm surface, the results of our study showed that postural sway was increased while standing on the foam surface and that this increase was more pronounced in older participants. However, the soft textured insole surface reduced ML sway and the C90 area for older participants to an equivalent level in the younger group. This observation was particularly evident for the balance tests performed on the foam surface with eyes closed, where there was a greater reliance on somatosensory information for maintaining balance. It is well understood that standing balance depends on the integration of visual, vestibular and somatosensory inputs (Maurer et al., 2000). When standing on a foam surface, the reliability of plantar cutaneous information is decreased (Patel et al., 2011) and closing the eyes negates the contribution of the visual system to balance control. Therefore, when standing on the foam surface with eyes closed, participants could have become more dependent on their vestibular and somatosensory inputs, which may have exposed age-related sensory deficits in the older participants. The present findings suggest that textured insole surfaces may be effective in ameliorating age-related deficits in somatosensory function.

While the findings of this study demonstrated that both textured insole surfaces reduced postural sway, it is important to note that most participants anecdotally reported that the harder insoles were uncomfortable to stand on for an extended period of time. Discomfort experienced by the participants while wearing the hard textured insoles may provide an explanation why the soft textured insoles were more effective at reducing postural sway in the older group. Perhaps the harder textured insoles
elicited greater discomfort in some people that may have caused a distraction or may have influenced their ability to control balance effectively. Furthermore, harder insole surfaces would most likely be more problematic for people with peripheral neuropathy, who often have ulcers and wounds on their feet. As such, it is recommended that a softer material be used for future falls prevention interventions.

The results of Study 1 indicated that a simple and inexpensive textured insole surface can decrease postural sway in older people, presumably due to the enhancement of the somatosensory information received from the feet. Given that postural sway is more common in clinical patient groups (e.g. people with diabetic peripheral neuropathy or Parkinson’s disease), textured insole surfaces may also provide potential benefits to these high risk populations. Further work involving a larger sample of older participants is also needed to confirm our findings and, given that a large percentage of falls occur during locomotion (Berg et al., 1997; Lord et al., 2001), it would be of interest to evaluate the efficacy of these textured insole surfaces on postural sway during walking. Therefore, Study 1 establishes the feasibility for the second phase of studies examining the effects of textured insoles on standing balance and gait in people with Parkinson’s disease.
Chapter 4

Study 2

Study 2 examined whether there were differences in physical function and the pick up of somatosensory information in older participants and people with PD. It also examined whether changes in available somatosensory information would affect postural stability in older participants and people with PD during standing and walking.

4.1 Introduction

Parkinson’s disease is a chronic and progressive degenerative disease of the brain, which is characterized by postural instability and gait disturbance (Adkin et al., 2005; Cole et al., 2010; Hanakawa et al., 1999; Latt et al., 2009; Morris et al., 2000; Schieppati & Nardone, 1991). Previous studies have reported that PD patients demonstrate more postural sway than controls during standing and often develop problems with gait initiation and gait adaptation (Adkin et al., 2005; Schieppati & Nardone, 1991). Furthermore, it is well established that people with PD walk with more trunk flexion and at a slower speeds than controls and have reduced step length and arm swing and increased stride timing variability. A recent study by Cole et al. (2011) reported that these deficits were exacerbated in patients when walking on a less stable and compliant surface, which may contribute to an increased risk of falling in this population.

Due to the loss of dopamine-producing cells in the substantia nigra pars compacta, Parkinson’s disease is characterized by motor deficits that play a major role in postural problems (Van den Berg et al., 2000; Van Emmerik et al., 1999). However,
the basal ganglia area is also connected to the somatosensory cortex (Albin et al., 1989), receiving a large amount of sensory information. With the stimulation of lower limb nerves, the cortical somatosensory evoked potential reflects the medial location of the foot areas of the somatosensory homunculus (Kievit & Kuypers, 1977; Legatt, 2009; Schell & Strick, 1984). Recent research has also pointed out that somatosensory feedback played an important role in adjusting postural control in people with PD (Carpenter & Bloem, 2010; Konczak et al., 2009; Van Wegen et al., 2006). The somatosensory system comprises the receptors and processing centres involved in facilitating three different sensory modalities: 1) discriminative touch; 2) proprioception; and 3), temperature and nociception. Evidence for the relationship between the somatosensory system and postural control was provided by Kerr et al. (2010), who showed that PD fallers had poorer peripheral sensation than PD non-fallers and that touch sensitivity was positively correlated with postural sway in patients with PD. Proprioception and related sensory information progressively diminish with severity of illness in PD individuals, resulting in a decline in motor functioning that is likely due to an impaired capacity to evaluate proprioceptive feedback and coordinate voluntary and reflexive motor commands (Carpenter & Bloem, 2010). However, while these studies have provided insight into the potential importance of the somatosensory system to balance control, they have not specifically explored somatosensory differences between people with PD and age-matched controls. Therefore, there is a clear need for research aiming to evaluate the relationship between somatosensory feedback and postural control in ageing individuals and people with PD.

To address this gap in the current research, Study 2 utilised a battery of physiological assessments to evaluate somatosensory function in people with PD and healthy age-matched controls. To comprehensively evaluate somatosensory function in these two groups, the assessments included measures of touch sensitivity, vibratory perception, and pain and temperature detection, as well as tests of nerve function and
somatosensory evoked potentials. Furthermore, both standing and walking balance were assessed in the two groups to explore the association between the measures of somatosensory function and postural control.

The aims of study 2 were to determine; 1) what differences exist in physical function and somatosensory information between older participants and people with PD, and 2), how changes in somatosensory information affect postural stability in older participants and people with PD during standing and walking.

4.2 General Methodology

4.2.1 Participants

Twenty people (13 males and 7 females; mean age 65±9 yrs) with Parkinson’s disease were recruited from: 1) a database of patients who had previously participated in falls prevention research, 2) interested patients from a Brisbane-based neurology practice (Neurosciences Queensland) at St Andrew’s War Memorial hospital, and 3), members of the Parkinson’s Community in Queensland who expressed an interest in being involved. Participants had a clinical diagnosis of idiopathic PD and were on a stable medication regime according to the determination of their treating physicians. The Unified Parkinson’s Disease Rating Scale (UPDRS) (Fahn et al., 1987) and Hoehn and Yahr Scale (1967) were undertaken to quantify disease severity.

An age and gender-matched control group of 20 healthy older people (13 males and 7 females; mean age 69±5 yrs) who participated in the study were recruited from a database developed from previous research projects on falls prevention.

Participants from both groups were required to be free of signs of dementia according to the Addenbrooke’s Cognitive Examination (ACE) (Mathuranath et al., 2000) (ACE total score <82: provides 84% sensitivity, 100% specificity for dementia (Reyes et al.,
2009)). They were also free from serious co-morbidities, such as unstable cardiovascular disease, rheumatologic disease, orthopedic disturbances, pain while walking, or acute illnesses that would interfere with static standing or dynamic motion. They were also required to be able to walk unaided for at least 15 minutes at a self-selected speed without interruption. All participants gave written informed consent after the nature and risks of the study were fully explained. The testing procedures were approved by the Queensland University of Technology Human Research Ethics Committee (Ethics approval number: 0900000118).

4.2.2 Experimental Protocol

4.2.2.1 Baseline Assessment

All participants in both groups undertook questionnaires that addressed specific characteristics, including demographic details, current medications, health status, history of falls, fear of falling (Modified Falls Efficacy Scale; Activities-specific Balance Confidence (ABC) scale) and general cognitive functioning (ACE).

Addenbrooke’s Cognitive Examination (ACE)
The ACE, which comprises a battery of cognitive tasks that provide a quantitative measure of cognitive status, was used to test for cognitive impairment in both groups. It has the considerable benefit of containing all items of the nearly-universally recognized Mini Mental State Exam (MMSE) (Folstein et al., 1975), with a number of added items to increase its sensitivity to specific dementing diseases, such as Alzheimer's disease and fronto-temporal dementia (Mathuranath et al., 2000). The ACE is a valid tool for dementia evaluation (Mathuranath et al., 2000; Reyes et al., 2009) and assesses five cognitive domains, including attention/orientation, memory, verbal fluency, language and visuospatial abilities (Appendix A.1). The total ACE score is a score out of 100, with higher scores indicating better cognitive functioning.
Modified Falls Efficacy Scale (MFES)

The modified falls efficacy scale (Hill et al., 1996) consists of 14 questions related to how confident an individual is that they will not fall while performing a series of activities of daily living (e.g. dressing and undressing, preparing a simple meal, crossing the road etc.) (Appendix A.2). All participants rated their confidence on a scale ranging from 0 (not confident at all) to 10 (completely confident) and the average score was used as a measure of fear of falling.

Activities-specific Balance Confidence (ABC) scale

In addition to the modified-falls efficacy scale, fear of falling was also assessed using the activities-specific balance confidence (ABC) scale. Like the MFES, this questionnaire assesses the confidence level in performing activities of daily living (ADLs) without falling and has been used to assess elderly individuals (Powell & Myers, 1995) and people with Parkinson disease (Oude Nijhuis et al., 2007). It includes 16 items covering walking and reaching-oriented activities which challenge postural balance and activities that are performed both indoors and outdoors. Participants were asked to score their confidence on a scale from 0% (no confidence) to 100% (complete confidence) and, again, the average score provided a measure of fear of falling for these participants (Appendix A.3).

Parkinson’s Disease Specific Questionnaires

In addition to these tests, participants in the Parkinson’s disease group undertook a number of additional assessments to evaluate the severity of their condition and the impact of common symptoms of the disease. Specifically, participants were assessed using the Unified Parkinson Disease Rating Scale (UPDRS), the Hoehn and Yahr scale (1967), the Schwab and England Activities of Daily Living scale (1969) and the Freezing of Gait Questionnaire (Giladi et al., 2000).
Unified Parkinson’s Disease Rating Scale (UPDRS)

The UPDRS was undertaken to provide a measurement of disease severity (Fahn et al., 1987). This assessment consists of three parts: 1) Mentation, behaviour, and mood; 2) Activities of daily life (ADLs); and 3), Motor. Some sections of the UPDRS require multiple grades assigned to each extremity and the possible total scores can range from 0 (least severe) to 199 (most severe). In addition to the UPDRS, the Hoehn and Yahr (1967) and the Schwab and England ADL scales (1969) were conducted to provide an indication of disease progression and functional capacity, respectively (Appendix A.4).

Gait and Falls Questionnaire (GFQ)

The gait and falls questionnaire was used to provided an indication of the severity of gait disability and falls-related motor function in the Parkinson's disease patients during the previous week (Giladi et al., 2000; 2009) (Appendix A.5). Specifically, the GFQ consists of 16 questions that broadly covered four areas: 1) gait in daily living; 2) frequency and severity of freezing of gait; 3) frequency of festinating gait and its relationship to falls; and 4), frequency and severity of falls. Scores for each item range from 0 (best) to 4 (worst) and the total score was used as a measure of gait and falls-related function.

4.2.2.2 Physiological Function Assessment

A comprehensive physiological assessment of sensory and motor functions was conducted to determine whether there were any deficits that might have contributed to balance and gait problems in these populations. These assessments included tests from the Physiological Profile Assessment (PPA) (Lord et al., 2003), which consists of a series of simple tests of vision, peripheral sensation, muscle force, reaction time, and postural sway. Additional tests were undertaken to more fully evaluate peripheral sensory function and included tests for the perception of temperature, pain, vibration and touch.
Vision Tests

Contrast sensitivity

The Melbourne Edge Test (MET) is a validated contrast sensitivity test (Haymes & Chen, 2004) that consists of 20 circular patches that have light and dark halves separated by a line through the middle (Figure 4-1). The contrast between the light and dark portions is progressively reduced from the first to the last circle, such that it becomes increasingly difficult to identify the margin between the two halves. Participants were required to correctly identify the direction of the line that divided the dark and light halves of the circles, starting from the circle with the highest contrast (top left in Figure 4-1). Participants continued across the chart and then down until they could no longer identify the direction of the line separating the two halves. The highest number correctly identified was recorded.

![Figure 4-1: Melbourne Edge Test chart (Left) and MET response key card (Right)](image)

Visual Acuity (Low and High Contrast)

The Bailey-Lovie Visual Acuity chart is a valid, reliable and rapid method of measuring threshold visual acuity for research purposes (Lovie-Kitchin, 1988). High- and low-contrast visual acuity was measured by using a chart with high contrast visual acuity letters and low (10%) contrast letters (Figure 4-2). Acuity was assessed binocularly at a distance of 3.2m with participants wearing their own corrective distance glasses (if needed). During the test, participants read aloud the letters on the high contrast chart first. The lowest line correctly read and the numbers of letters correctly identified on that line were recorded. The same procedure was followed for
the low contrast chart. Visual acuity was scored letter-by-letter (each letter corresponded to 0.02 logMAR units).

![Low contrast chart](image)

Figure 4-2: Bailey-Lovie High and Low Contrast Visual Acuity Test

**Strength Tests**

**Knee Extension and Knee Flexion**

Knee extension (quadriceps muscles) and flexion (hamstring muscles) strength was measured according to the experimental protocol described by Gandevia (2001) using a strain gauge attached to the lower leg 10 cm above the ankle. Participants were seated with the hip at 90 degrees and the knee at 90 degrees. Participants performed 3 maximal isometric contractions of the knee extensors and flexors of each leg and each of these exertions lasted approximately 3 seconds. After each trial, participants were given a brief rest period to minimize the effects of fatigue and during each performance. The participants were given strong verbal encouragement to ensure that they were exerting maximally. The best score of the three attempts for each side was recorded.
Ankle Dorsiflexion
Participants were seated with their bare foot securely fastened to a foot rest, by a strap positioned over the top of the foot. During the performance of the test, the lateral malleolus of the ankle was aligned with the hinge of the device and the knee was positioned at approximately 120 degrees (Figure 4-3). Participants were encouraged to raise the front of their foot (dorsiflexion) as forcefully as possible for 3 seconds on each attempt. During each performance, participants were given strong verbal encouragement to ensure that they were exerting maximally. This was repeated three times for each side and a rest period was given between attempts to minimise fatigue. The maximum value achieved for each side was recorded.

Figure removed for copyright reasons

Figure 4-3: Measurement of ankle dorsiflexion strength: the lateral malleolus of the ankle was aligned with the hinge of the device and the knee was positioned at 120 degrees (Lord et al., 2003)

Grip Strength
While performing this test, participants were seated with their shoulder adducted and neutrally rotated, with their elbow flexed to 90 degrees and their forearm and wrist in a neutral position. During the test, participants grasped the handle of the grip strength dynamometer and were encouraged to squeeze. During each trial, participants were given strong verbal encouragement to ensure that they were exerting maximally. The
maximum scores from three attempts for each hand were recorded and a rest period was given between attempts to minimize fatigue.

**Reaction time**

Hand and foot reaction times were measured bilaterally using a light as the stimulus and a rapid depression of a switch by the hand or foot as the response (Figure 4-4). The participants were presented with the light stimulus after a random (1 to 3s) foreperiod, which followed initiation of the trial by the experimenter. The reaction time of each hand and foot were assessed separately and the testing procedures included five practice trials, which were undertaken to ensure that the participants understood the procedure and ten ‘test’ trials, which were averaged.

Figure removed for copyright reasons

Figure 4-4: Reaction time test for the hands and feet (Lord et al., 2003)

**Peripheral Sensation Tests**

**Touch Sensation (Ankle and Foot)**

This test measured sensory reception using a set of Semmes-Weinstein-type pressure aesthesiometers, which contain eight different filaments of equal length, but varying diameters. The lateral malleolus of the ankle was tested because of its extensive use in previous ageing and falls studies (Lord et al., 2003; Simoneau et al., 1996), while nine
other positions on the soles of the feet were also assessed based on previously reported measures of tactile sensitivity (Hills et al., 2001). The nine test sites examined on the plantar surface of the feet were specifically chosen as the represented areas innervated by different peripheral nerves (5 sites innervated by medial plantar nerve, 2 sites by lateral plantar nerve, 1 each for the tibial and sural nerve) (Figure 4-5) (Delfaut et al., 2003). The testing procedure was conducted in accordance with the protocol described by Lord et al. (2003) and required the examiner to start with the thickest filament (#8) and progressively decrease the size of the filament until the participant could not detect the touch. Once the threshold was reached, the smallest filament that the participant could detect was tested again to ensure that the participant had not guessed the correct answer the first time. If the participants correctly identified the touch a second time, then the next smallest filament (previously undetected) was tried one last time. During the test, the examiner would say “A” and “B” with equal vocal emphasis and would touch the filament to the participant’s skin after one of these vocalisations. The participant was required to indicate whether they felt the monofilament touch their skin after the examiner said “A” or after the examiner said “B”; it was always one or the other. The finest filament detected provided an estimate of the participant's touch threshold.

Figures removed for copyright reasons

Figure 4-5: Semmes-Weinstein-type pressure aesthesiometer used on 9 foot stimulation points and ankle stimulation point (Lord et al., 2003) (http://en.wikipedia.org/wiki/File:Gray834.svg)
**Perception of Temperature, Vibration and Nociception**

The Thermal Sensory Analyser (TSA-2001 neurosensory analyser model TSA-II) and Vibratory Sensory Analyser (VSA-3000) (Medoc Advanced Medical Systems Ltd., Ramat Yishai, Israel) were used to evaluate nociception (pain), temperature and vibratory perception of the feet. These devices are advanced, computerised stimulators that are designed for advanced neurological and pain research. Assessments included quantitative thermal sensory tests (QST) and vibratory perception thresholds (VPT), which have been widely used in clinical and research studies (Lowenstein et al., 2008; Shy et al., 2003).

A Thermode (30 × 30 mm² contact probe) was attached to the participant on the dorsilateral surface of the dominant foot over the fifth metatarsal head. This device was capable of heating or cooling the skin as needed and during each of the trials, the temperature was either increased (heating) or decreased (cooling) until a change in the stimulus was perceived (Yarnitsky & Ochoa, 1991). Four sensory sub-modalities were tested: 1) Warm sensation (WS), a sensation mediated by C fibres; 2) Cold sensation (CS), mediated by Aδ fibres; 3) Heat-induced pain (HP), which is mostly a C fibre mediated sensation with some involvement of Aδ fibres; and 4) Cold-induced pain (CP), mediated by a combination of both C and Aδ fibres. During the WS and CS tests, the thermode started at a baseline temperature of 32°C and increased or decreased by 0.3°C per second. When the participants perceived a change in temperature, they were asked to press a button on a computer mouse that they held in their hand as quickly as possible. Once the button had been pressed, the temperature of the thermode remained constant and the participants were required to tell the experimenter whether the temperature increased or decreased. To familiarise themselves with the protocol, participants were allowed a number of practice trials, after which six trials were undertaken in a random order (3 WS trials and 3 CS trials). During the HP and CP assessments, participants were asked to respond using the computer mouse as soon as they felt the temperature was getting too hot or cold,
making them feel uncomfortable. Again, the baseline temperature of the thermode was 32°C and increased or decreased at 1.5°C per second and the participant used the computer mouse to respond to the painful sensation. Once the button was pressed, the thermode returned to the baseline temperature at a rate of 10°C per second and participants were allowed a 4-seconds interval between trials. Several practice trials were given to familiarise the participants with the protocol, then six trials were undertaken in a random order (3 hot-induced pain sensation trials and 3 cold-induced pain sensation trials). To prevent participants from any personal harm or injury during data collection, upper and lower limits of 50°C and -10°C were entered. These limits were designed to ensure the safety of participants who could not feel the temperature change and the system stopped automatically once it reached these limits.

**Vibration Test**

The purpose of this test was to assess the participant’s sensitivity to vibration, which is mediated by the larger fibres responsible for perception of touch, mild pressure and joint position. Participants were seated with their dominant foot on a vibratory device with their big toe resting on the vibrating nodule of the device (Figure 4-6). During the data collection phase, this nodule vibrated at 100Hz, starting at a amplitude of 0 microns and increasing at a rate of 0.8 microns per second. Once the participants perceived the vibratory stimulus, they were asked to press a button on the computer mouse as quickly as possible. After several practice trials, each participant completed four experimental trials. To prevent participants from personal harm due to the inability to perceive changes in vibration, an upper limit of 130 microns was set and the equipment automatically stopped when this limit was reached.
Proprioception

Peripheral sensation measurement was also undertaken through an assessment of lower-limb proprioception to determine the participants’ ability to know where their body is in space. Participants were seated on a high stool with a 60cm x 60cm x 1cm thick clear acrylic sheet positioned between their feet. The vertical surface of the acrylic sheet was inscribed with a protractor that showed angular increments of 2° (Figure 4-7). Prior to testing, a small cross was drawn bilaterally on the medial aspect of the first metatarsal head. Participants were blindfolded and asked to attempt to match the position of these crosses on either side of the sheet. The experimenter passively rotated the lower limb and moved one of the feet to the board and asked the participant to match this by moving their other leg and placing their foot in the same perceived position. Each foot was tested in 3 randomly presented positions on the board; 1) high (knee angle approximately 160°); 2) medium (knee angle approximately 130°); and 3) low (knee angle approximately 100°). Following these trials, participants remained blindfolded and were asked to move both feet to the same position on the board six times, twice at each of the levels described previously, which were presented in a random order.
Figure 4-7: Participant matching the feet on the acrylic sheet during the proprioception test (Lord et al., 2003)

4.2.2.3 Peripheral Nerve Electrodiagnosis

Nerve conduction
Nerve function was evaluated using the Neuropack S1 system (MEB-9400, Nihon Kohden, Tokyo, Japan). For the purposes of these assessments, nerve conduction tests were conducted on three nerves of the participant’s dominant leg: 1) the peroneal nerve; 2) the tibial nerve; and 3), the sural nerve. For the assessment of the peroneal and tibial nerves, the amplifier sensitivity was set to 5mV, while the high- and low-pass filters had cut-off frequencies of 5kHz and 10Hz, respectively. For the sural nerve, the sensitivity of the amplifier was set to 20µV, while the cut-off frequencies of the high- and low-pass filters were set at 2kHz and 20Hz, respectively. The analysis time was set at 2ms per division, while the stimulation rate and the intensity of the electrical stimulation signal was set to 1Hz and the duration of electrical stimulations signal was 0.2ms. For assessing the F-Wave, the amplifier’s sensitivity was set to 5mV and the high-and low-pass filters respectively had cut-off frequencies of 5kHz and 20Hz. The analysis time for the F-Wave was set on 5ms per division and the stimulation occurred at a rate of 1Hz for duration of 0.2ms. The parameters used for each of these assessments are summarised in Appendix A.6.

For the purposes of assessing nerve function, participants were asked to remove their
shoes and socks and to lie in a supine position on a therapy bed. Recording electrodes (Skintact RT34, Innsbruck, Austria) were attached in conjunction with conductive adhesive Aqua-Tac hydrogel over the optimal recording areas for each site.

**Peroneal Nerve**

For assessment of the peroneal nerve, the ground electrode (shown in green in Figure 4-8) was placed on the dorsal surface of the foot, while the active electrode (shown in black in Figure 4-8) was placed over the belly of the extensor digitorum brevis (EDB). The reference electrode (shown in red in Figure 4-8) was attached distal to the active electrode just below the head of the fifth metatarsal. Peroneal nerve functioning was assessed at three sites, the most distal of which was between the tendons of the tibialis anterior and extensor hallucis: 8 cm superior to the active electrode. The second stimulation site was located 2 cm below the fibula head, while the third was 10 cm above the fibula head, just medial to the end of the biceps femoris tendon in the lateral popliteal fossa (Figure 4-8). During the tests, stimulation was gradually increased until the maximum amplitude of the wave was reached, at which point the M-wave, which is indicative of muscle contraction, was recorded. After marking the peak and baseline values on the waveforms, the amplitude of the M-wave was derived by calculating the difference between these two values. Additionally, the latency was calculated as the time interval between the onset of the stimulus and the corresponding waveform and nerve conduction velocity was derived by dividing the measured distance between the site of stimulation and the detecting electrode by the elapsed time.

The F-wave is only present in the waveform during a supramaximal stimulation of a motor nerve. In this situation, the supramaximal impulse travels proximally to the motor neuron cell bodies located in the spinal cord, but a small portion of the impulse backfires and travels back down the nerve towards the muscle, resulting in the small
F-waves. The F-wave of peroneal nerve was assessed by placing the stimulation electrode between the tendons of the tibialis anterior and extensor hallucis (8 cm superior to the active electrode) with the negative (-) contact placed closer to the spinal cord and the positive (+) contact closer to the foot. Stimulation intensity was gradually increased to a level that was 20 to 25% greater than the intensity which obtained the maximal compound muscle action potential that presents a group of almost simultaneous action potentials from several muscle fibers in the same area. After achieving this intensity, ten waveforms were recorded and used to calculate the timing, magnitude and latency of the F-wave.

Figure removed for copyright reasons

Figure 4-8: Left: Three stimulation positions (Blue-negative, Red-positive) and three recording electrodes (Black-active, Red-reference and Green-ground) during the test of the peroneal nerve; Right: Examples of peroneal nerve waves during stimulation. (Neupack S1 Measuring System-MEB9400 User Manual)

**Tibial Nerve**

For assessing activity of the tibial nerve, the ground electrode (shown in green in Figure 4-9) was placed on dorsal surface of the foot. The active electrode (shown in black in Figure 4-9) was placed over the belly of the abductor hallucis muscle (AH) and the reference electrode (shown in red in Figure 4-9) was attached to base of the hallux near the big toe. Two stimulation sites were chosen, the most distal of which was 10 cm proximal to the active electrode on the middle portion of the medial
malleolus. The second site was more proximal and was located at the central portion of the popliteal fossa with the anode proximal in position (Figure 4-9). The procedures used to measure the M-wave and the F-wave for the tibial nerve were the same as those described for the peroneal nerve, with the exception that the stimulation electrode was positioned on the middle portion of the medial malleolus while assessing the F-Wave.

Figure removed for copyright reasons

Figure 4-9: Left: Two stimulation positions (Blue-negative, Red-positive) and three recording electrodes (Black-active, Red-reference and Green-ground) during the test of the peroneal nerve; Right: Examples of tibial nerve waves during stimulation.

(Neuropack S1 Measuring System-MEB9400 User Manual)

Sural Nerve
Finally, the assessment of the sural nerve required the ground electrode (shown in green in Figure 4-10) to be placed on the dorsal surface of the foot. The active electrode (shown in black in Figure 4-10) was placed posterior to the lateral malleolus one third of the distance along a line drawn between the heel and the malleolus. The reference electrode (shown in red in Figure 4-10) was attached 3 cm distal to the active electrode. Stimulation was applied to the posterior shank, 14 cm proximal to the active electrode and 1 to 2 cm lateral to the Achilles tendon (the junction of gastrocnemius muscle and the Achilles tendon) (Figure 4-10). In accordance with the
procedures outlined for the peroneal and tibial nerves, the intensity of the stimulation was gradually increased until 3-5 waveforms were obtained. The same techniques were used to calculate the latency, amplitude and velocity of nerve conduction as those described previously.

Figure removed for copyright reasons

Figure 4-10: Left: Stimulation position (Blue-negative, Red-positive) and three recording electrodes (Black-active, Red-reference and Green-ground) during the test of the sural nerve; Right: Examples of sural nerve waves during stimulation. (Neuropack S1 Measuring System-MEB9400 User Manual)

Somatosensory Evoked Potentials (SEP)
The Neuropack S1 evoked potential measurement system was also used to evaluate somatosensory evoked potentials for the two participant groups. This included an assessment of the dominant lower extremity somatosensory evoked potentials in terms of latency between two input marks (N20-Lumbar and P37-Cranial). Activity at the N20-Lumbar (at a latency of around 20 ms) and P37-Cranial (at a latency of around 37 ms) regions were recorded after stimulation of the posterior tibial nerve at the ankle, indicating the spinal response potential and cortical response, respectively (Figure 4-11). A pair of electrode channels was chosen: The positive (+) contact of the probe was channel X2, Cz, while the negative (-) contact was X1, Fz. The amplifier
sensitivity was set at 10µV and the high- and low-pass filters had cut-off frequencies of 3kHz and 20Hz, respectively. The analysis time was set at 10ms per division and the stimulation signal occurred at a rate of 2Hz for a period of 0.2ms.

Figure removed for copyright reasons

Figure 4-11: Recording N20 and P37 during stimulation from posterior tibial nerve (Modified From (Walsh et al., 2005))

Prior to lying on a therapy bed, participants were asked to remove their shoes and socks and to roll their trouser leg above the knee. The disc electrodes were placed on four positions with conductive paste (Ten 20; Weaver and Co., CO, USA) and held in place using tape. Specifically, the first site used for this assessment (N20-Lumbar) required an active electrode to be placed on the spinous process of the first lumbar vertebra (L1) and a reference electrode to be placed on the contralateral anterior superior iliac spine (ASIS). By dividing the head into halves in the transverse and sagittal planes, the vertex (Cz point) was derived as the point where the two lines intersected. The second site prepared for this assessment (P37-Cranial) required an active electrode to be placed at Cz1 (1cm posterior to the Cz point). The Fz point was identified by dividing the distance between Cz and the nasion in proportions and placing the reference electrode within the anterior 2/5ths of the distance (Figure 4-12). The ground electrode was placed on the tibialis anterior muscle of the dominant side.
The stimulator was positioned over the posterior tibial nerve on the middle portion of the medial malleolus with the negative (-) contact placed towards the spinal cord, proximal to the positive (+) contact. The stimulation intensity was gradually increased approximately 3 times above threshold until the muscle contracted. Once clear waveforms for the N20 (Lumbar) and P37 (Cranial) sites were displayed on the screen, a minimum of 150 sweeps were recorded for averaging. The same marking skills, as those described for the nerve conduction assessments were used to calculate the latency of the N20 and P37 measures. The amplitude of waveforms and interval between the N20 and P37 waveforms were calculated automatically by the software.

Figure removed for copyright reasons

Figure 4-12: Electrode position for P37 (cranial) site for the Somatosensory Evoked Potential measurements (Neuropack S1 Measuring System-MEB9400 User Manual)

4.2.2.4 Postural Stability Assessment

Tinetti Balance and Gait

The Tinetti Balance and Gait assessment (Tinetti et al., 1986) was developed to determine the risk for falls in older adults. This test is a reliable tool that involves an assessor evaluating a participant’s performance on sixteen predefined tasks that are
common to normal daily activities (Raîche et al., 2000). Specifically, this assessment evaluates sitting and standing balance, rising from a chair, turning on the spot, as well as posture and walking speed during gait. The scores derived from these tests are used to calculate two sub-scores that include, a balance sub-score (26 points maximum) and a gait sub-score (9 points maximum), which are summed to give the total score (35 points maximum).

**3D Gait Assessment**

To evaluate group differences in postural control during dynamic tasks, a comprehensive assessment of gait was conducted using a 3D motion analysis system (Vicon, Oxford, UK). Participants completed a series of walking trials on two different walking surfaces: 1) a flat and firm surface; and 2), an uneven surface (Figure 4-13). The uneven surface was comprised of wooden blocks (10.5x7x2cm) placed beneath a layer of foam (900x92x1cm) and artificial grass. Participants were required to complete 6 barefoot walking trials at a self-selected comfortable pace on each surface and completed these trials in a random order.

![Firm Surface](image1.png) ![Uneven Surface](image2.png)

Figure 4-13: Two walking surfaces (firm and uneven surface)

During these tests, 3D kinematic data were recorded bilaterally using an eleven camera (200Hz) optoelectronic motion analysis system (Vicon Nexus, Vicon, Oxford, UK). Prior to data collection, the Vicon motion analysis system was calibrated using a
3-marker wand (240mm) and a 4-marker ErgoCal L-frame (14mm). To facilitate a number of anthropometric measurements and the accurate placement of reflective joint markers, participants were asked to wear short pants and a singlet with a hole cut in the back for the placement of markers on the thorax. Prior to data collection, anthropometric measures, including body mass, height, ASIS breadth, leg length, knee and ankle width, hand thickness, shoulder offset, and elbow and wrist width were taken for each participant and entered into the Vicon Nexus software (Appendix A.7). Following this, reflective markers were positioned on specific anatomical landmarks of the body in accordance with the Vicon Plug-In Gait Full Body model, which is the validated Vicon implementation of the Conventional Gait Model (CGM). The CGM was developed based on the Newington-Helen Hayes gait model and has been widely used in research and clinical practice for assessing gait (Davis et al., 1991; Kadaba et al., 1990; Woltring, 1986). This model required the placement of 34 retro-reflective markers on: the head (bilateral on the front and back), trunk (jugular notch, C7 and T10 spinous processes, xiphoid process of the sternum, right scapula), pelvis (sacrum, bilateral ASIS), arms and hands (bilaterally on the lateral border of the acromion, olecranon process of the humerus, radial and ulnar styloids, 3rd metacarpal head), and legs and feet (bilaterally on the lower lateral 1/3 of the thigh and shank, lateral aspect of the femoral condyle, lateral malleolus, calcaneus, 2nd metatarsal head) (Appendix A.8). Where possible, markers were attached directly to the skin using double-sided tape, while the markers positioned on the head and wrists were attached to a firm-fitting headband and wrist bands, respectively. The markers positioned on the lower lateral 1/3 of the thigh and shank were positioned on 10cm wands and were firmly held in place via Velcro straps. Prior to performing the gait assessment, a static calibration trial was collected, which involved each participant standing stationary in the anatomical position in the middle of the 2.6m (W) x 9.3m (L) x 2.5m (H) capture volume (Figure 4-14). Once it was established that all markers could be clearly seen in the 3D workspace, participants completed the previously described gait tasks while the three-dimensional trajectories of the markers were recorded.
Standing balance assessment

To examine postural stability under static conditions, postural sway was assessed for each participant using a force plate (OR6-6-2000, AMTI, USA). Centre of pressure data were recorded at 1000Hz for two 30 second trials performed under each of the following conditions: 1) on a firm surface with eyes open; 2) on a firm surface with eyes closed; 3) on a foam surface with eyes open; and 4), on a foam surface with eyes closed. For the assessment of postural sway on the foam surface, participants were asked to position themselves in the centre of a medium density foam block (74.5x62x15.7cm) that was positioned over the surface of the force platform. While standing on the force plate, participants were asked to look straight ahead at a cross positioned at eye level, with their feet 10cm apart and their hands at their sides. Participants were permitted a rest between trials, if required, and if participants over-balanced during data collection, the trial was repeated.
Data Analysis

Neurological Assessments:
In addition to the total score, the ACE also comprises five sub-scores that assess: 1) attention and orientation; 2) memory; 3) fluency; 4) language; and 5), visuospatial perception providing significant insight into specific deficits in cognition for a demographic group. Similarly, the UPDRS includes 3 sub-scores that relate to; 1) mentation, behaviour, and mood; 2) activities of daily life (ADLs); and 3) motor symptoms and also incorporates the Hoehn & Yahr (1967) and Schwab & England ADL scales (1969). To examine group differences between the older individuals and the people with PD, group averages were calculated for all the test scores derived from the baseline assessments and the physiological function tests.

Electrodiagnosis:
A number of variables were derived from the electrodiagnosis assessment, including the motor and sensory amplitudes, which were calculated as the difference between the peak positive and the peak negative values recorded in these tests (values between P1 and P2 shown in Figure 4-15, Figure 4-16 and Figure 4-17). The sensory nerve action was reported to the nearest 1 mV. To assess motor latency of the waveforms, it was measured from the stimulation point (S in Figure 4-15 and Figure 4-16) to the onset latency at the take-off of the negative component of the M-wave (L in Figure 4-15 and Figure 4-16). It was calculated at the display sensitivity at which the response was obtained, and was recorded to the nearest millisecond (values between S and L). The sensory latency was measured from stimulation point (S in Figure 4-17) and the onset latency from the take-off of the negative component of the sensory nerve action potential (L in Figure 4-17). A minimum of 10 F-waves values were measured using supramaximal stimuli and the minimum F-wave latency was recorded to the nearest millisecond (values from S to L in Figure 4-15 and Figure 4-16). Also the ratio between the number of F-waves and the number of stimulations was used to present F-wave occurrence. All conduction velocities of the waves were automatically
calculated using onset latencies and the distances between the two sites of stimulation.

For the somatosensory evoked potential, the amplitude and the latency of the early cortical evoked potential waves of the N20 (Lumbar) and P37 (Cranial) sites were recorded similar to nerve conduction tests (values from P1 to P2 showed in Figure 4-18), following stimulation of the posterior tibial nerve. Then the latency interval between N20 and P37 were calculated according to the methods described for the onset latencies (values from S to L showed in Figure 4-18).

Figure 4-15: M waves (Left) and F-waves (Right) of peroneal nerve. S- stimulation point, L- latency; P1- Positive peak of the wave; P2- Negative peak of the wave.
Figure 4-16: M waves (Left) and F-waves (Right) of tibial nerve. S- stimulation point, L- latency; P1- Positive peak of the wave; P2- Negative peak of the wave.

Figure 4-17: Sensory Nerve Action Potential (SNAP) of Sural nerve. S- stimulation point, L- latency; P1- Positive peak of the wave; P2- Negative peak of the wave.
Figure 4-18: Results example of the N20 (Lumbar) and P37 (Cranial) Somatosensory evoked potentials. S- stimulation point; P1- Positive peak of the wave; P2- Negative peak of the wave.

**Temperature, Vibration and Nociception Analysis**

To examine group differences between the older individuals and the people with PD, group averages were calculated for all test scores derived from quantitative thermal sensory tests (QST) and vibratory perception thresholds. The accuracy of correctly identifying the direction of the temperature change (e.g. identified temperature as hotter when the temperature was increased) was also recorded and included as an outcome measure.

**Gait Analysis**

Following data collection, the raw 3D coordinates were labelled within the Vicon Nexus software and filtered using the Woltring filtering routine (Woltring, 1986), which uses a cubic spline function to smooth the data and fill any gaps in the marker trajectories. When applying the Woltring filter to the 3D trajectories, the General Cross Validation (GCV) (Davis et al., 1991; Kadaba et al., 1990) option was used to account for any noise in the marker trajectories. After the data had been labeled, three
heel contact and toe-off events were visually identified for each foot, yielding a total of 2 complete gait cycles per leg. Based on the 3D trajectories of the markers and the timing of these events, it was possible to calculate a number of temporospatial gait characteristics. Specifically, stride length, stride frequency (cadence), double support time (percentage of time with both feet on the ground), walking velocity and stride timing variability (SD of stride period) (Hausdorff et al., 1998) were calculated.

**Force Plate**

Following data collection, the raw force plate data for each of the standing balance trials were filtered using a Butterworth filter with high and low pass settings and the COP data were exported to Microsoft Excel. Using the COP data, medial-lateral sway was calculated as the range of movement along the X-axis of the force plate, while the anterior-posterior sway was calculated as the range of movement along the Y-axis.

**Correlation**

To assess the correlation between somatosensory function and the temporospatial gait characteristics, the results were analysed using the Pearson's correlation coefficient and the Spearman's Rho statistical procedures.

**4.2.3 Statistical Analysis**

The student’s t-test was used to analyze differences in physiological function between the people with PD and the control group, while the Chi-Square test was used to analyze the accuracy of temperature and pain detection between the two groups. To compare the groups with respect to the gait measurements, a mixed model Analysis of Variance (ANOVA) with 1 between group factor (group: PD, older) and 1 within group factor (surface: firm, uneven) was used. Similarly, the static balance data were also analysed using a mixed model ANOVA, which had 1 between group factor
(group: PD, older) and 2 within group factors (vision: eyes open, eyes closed; and surface: firm, foam). Violations of the sphericity assumption for repeated measures variables were checked using Mauchley’s test of sphericity and when a violation of this assumption was apparent, the Greenhouse-Geisser correction procedure was used. When a significant main effect was reported for a comparison, post-hoc analyses were undertaken using the Fisher’s Least Significant Difference (LSD) test. Finally, to evaluate the potential relationship between the standard physiological function assessments and other measures of postural stability, Pearson’s and Spearman’s correlations coefficients were used. All data analyses were completed using the Statistical Package for Social Sciences (SPSS V17.0, Chicago, IL, USA) and the level of statistical significance was set at 95% (P<0.05).
4.3 Results

A total of 20 control participants (13 male, 7 female, mean age 69.0±4.8 years) and 20 people with PD (13 male, 7 female, mean age 64.9±9.0 years) completed all assessments. The PD participants were predominantly in the early stages of the disease (Hoehn & Yahr score 1.4±0.9) and had an average UPDRS total score of 40.9±17.9. For the three sub-scales of the UPDRS, the mean scores were 2.8(±2.2), 11.5(±7.4) and 26.6(±10.8) for parts 1, 2 and 3, respectively. Finally, the results of the Schwab & England ADL scale (87.3±7.5), the GFQ (8.4±10.7) and the FOG questionnaire (4.3±4.9) demonstrated that the PD patients’ daily activities were only mildly affected by their symptoms (Table 4-1).

Table 4-1: Disease characteristics of Parkinson’s participants

<table>
<thead>
<tr>
<th>Measure</th>
<th>Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>UPDRS I</td>
<td>2.8 (2.2)</td>
</tr>
<tr>
<td>UPDRS II</td>
<td>11.5 (7.4)</td>
</tr>
<tr>
<td>UPDRS III</td>
<td>26.6 (10.8)</td>
</tr>
<tr>
<td>UPDRS Total</td>
<td>40.9 (17.9)</td>
</tr>
<tr>
<td>Hoehn &amp; Yahr</td>
<td>1.4 (0.9)</td>
</tr>
<tr>
<td>Schwab &amp; England ADL</td>
<td>87.3 (7.5)</td>
</tr>
<tr>
<td>GFQ</td>
<td>8.4 (10.7)</td>
</tr>
<tr>
<td>FOG</td>
<td>4.3 (4.9)</td>
</tr>
</tbody>
</table>

Abbreviations: UPDRS _ Unified Parkinson’s Disease Rating Scale; GFQ _ Gait and Fall Questionnaire; FOG _ Freezing of Gait Questionnaire

a Data are mean (SD)

Comparison of demographics between control and PD groups

For the purposes of this study, the control group and the PD group were age and gender-matched and there were no significant differences recorded between the groups for height, weight or BMI (P>0.05).
Both groups reported a similar number of falls during the previous year (P>0.05), but the Modified Falls Efficacy Scale and the ABC scale demonstrated that the control group had more confidence undertaking activities without falling (P<0.05). It is worth noting, however, that the ABC scores collected for the PD group were more variable than the controls, with scores ranging from 46.3 to 100, compared with 74.4 to 100 for the control participants.

With respect to cognitive function, the results demonstrated that neither group showed signs of significant cognitive impairment (ACE score >83) and the groups did not differ significantly for the ACE Total score. However, while the groups recorded similar total scores for the ACE, the PD participants scored significantly lower in the attention and orientation sub-scales than the control group (P<0.05). There were no significant differences recorded for the ACE sub-scores of memory, fluency, language or visuospatial function (Table 4-2).

Table 4-2: Demographic and disease characteristics of both groups

<table>
<thead>
<tr>
<th></th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Demographic</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Age, y</td>
<td>69.0 (4.8)</td>
<td>64.9 (9.0)</td>
<td>0.082</td>
</tr>
<tr>
<td>Male, %</td>
<td>65%</td>
<td>65%</td>
<td></td>
</tr>
<tr>
<td>Height, cm</td>
<td>171.4 (7.8)</td>
<td>171.1 (7.5)</td>
<td>0.612</td>
</tr>
<tr>
<td>Weight, kg</td>
<td>83.6 (15.3)</td>
<td>75.4 (12.7)</td>
<td>0.071</td>
</tr>
<tr>
<td>BMI, kg/m²</td>
<td>28.4 (4.6)</td>
<td>25.8 (4.4)</td>
<td>0.068</td>
</tr>
<tr>
<td><strong>Fear of fall and history of fall</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Modified Falls Efficacy Scale</td>
<td>9.9 (0.1)</td>
<td>8.8 (1.9)</td>
<td>0.016</td>
</tr>
<tr>
<td>ABC</td>
<td>95.9 (6.1)</td>
<td>85.5 (18.6)</td>
<td>0.029</td>
</tr>
<tr>
<td>Falls in previous 12 months</td>
<td>0.5 (1.0)</td>
<td>0.8 (1.6)</td>
<td>0.477</td>
</tr>
<tr>
<td><strong>Cognitive Function</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ACE</td>
<td>91.4 (5.3)</td>
<td>88.2 (6.7)</td>
<td>0.097</td>
</tr>
<tr>
<td>Attention and Orientation</td>
<td>17.7 (0.5)</td>
<td>17.2 (0.9)</td>
<td>0.024</td>
</tr>
<tr>
<td>Memory</td>
<td>21.6 (3.2)</td>
<td>20.3 (3.9)</td>
<td>0.280</td>
</tr>
<tr>
<td>Fluency</td>
<td>11.5 (2.2)</td>
<td>11.0 (2.3)</td>
<td>0.531</td>
</tr>
<tr>
<td>Language</td>
<td>25.1 (1.1)</td>
<td>24.6 (1.0)</td>
<td>0.144</td>
</tr>
<tr>
<td>Visuospatial</td>
<td>15.6 (0.6)</td>
<td>15.1 (1.3)</td>
<td>0.115</td>
</tr>
</tbody>
</table>

Abbreviations: PD_ Parkinson disease; BMI_ body mass index; ABC_ Activities-specific Balance Confidence; ACE_ Addenbrooke’s Cognitive Examination

a Data are mean (SD)
Comparison of physiological function assessment between control and PD

For the vision tests, there were no significant differences between the control group or the participants with PD with respect to contrast sensitivity (MET) or high and low contrast visual acuity (P>0.05).

Similarly, there were no significant differences shown in the measures of knee extension, ankle dorsiflexion or grip strength (P>0.05). However, the PD participants had significantly poorer knee flexion strength than the control group (P<0.05).

PD participants performed significantly more poorly than controls on the hand reaction time test (P<0.01), but there was no significant difference found between the groups for foot reaction time (P>0.05) (Table 4-3)

Table 4-3: Physiological function assessment between two groups a

<table>
<thead>
<tr>
<th></th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Vision</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Contrast sensitivity: MET</td>
<td>20.2 (1.9)</td>
<td>19.4 (2.5)</td>
<td>0.263</td>
</tr>
<tr>
<td>Visual acuity: high contrast</td>
<td>0.1 (0.1)</td>
<td>0.1 (0.1)</td>
<td>0.959</td>
</tr>
<tr>
<td>Visual acuity: low contrast</td>
<td>0.4 (0.1)</td>
<td>0.4 (0.2)</td>
<td>0.649</td>
</tr>
<tr>
<td><strong>Strength, kg</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee flexion</td>
<td>19.1 (6.8)</td>
<td>14.3 (7.0)</td>
<td><strong>0.038</strong></td>
</tr>
<tr>
<td>Knee extension</td>
<td>33.3 (10.6)</td>
<td>29.9 (11.4)</td>
<td>0.348</td>
</tr>
<tr>
<td>Ankle dorsiflexion</td>
<td>19.4 (3.5)</td>
<td>17.0 (4.8)</td>
<td>0.077</td>
</tr>
<tr>
<td>Hand grip</td>
<td>36.9 (11.9)</td>
<td>32.9 (8.3)</td>
<td>0.239</td>
</tr>
<tr>
<td><strong>Reaction time, ms</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hand</td>
<td>228.2 (27.8)</td>
<td>264.5 (46.0)</td>
<td><strong>0.005</strong></td>
</tr>
<tr>
<td>Foot</td>
<td>314.3 (43.4)</td>
<td>349.9 (77.9)</td>
<td>0.084</td>
</tr>
</tbody>
</table>

Abbreviations: PD _ Parkinson disease; MET _ Melbourne Edge Test
a Data are mean (SD)
Comparison of Peripheral Sensation between control and PD

Compared to controls, the participants with PD had significantly poorer touch thresholds at each of the nine sites on the plantar surface of the foot and the lateral malleolus, yielding poorer tactile sensitivity scores overall (4.0±1.5 vs. 2.5±0.7, P<0.01). Similarly, vibratory sensitivity was also poorer for the PD patients compared to the controls (P<0.01), with two people from the PD group reporting no detection of the stimulus even at the maximum setting. Furthermore, the range of vibratory sensitivity scores was considerably wider for the PD group, ranging from 1.6 to 130 microns compared with 2.2 to 58.6 microns for controls. However, despite the differences observed in tactile sensitivity and vibratory perception, the groups did not differ with respect to lower limb proprioception (P>0.05) (Table 4-4).

Table 4-4: Sensory of Touch, Vibration, Lower Limb Proprioception in both groups

<table>
<thead>
<tr>
<th></th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Touch</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total</td>
<td>2.5 (0.7)</td>
<td>4.0 (1.5)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Ankle</td>
<td>2.5 (1.1)</td>
<td>4.1 (1.8)</td>
<td>0.003</td>
</tr>
<tr>
<td>Position 1 (Medial Plantar)</td>
<td>2.6 (1.1)</td>
<td>4.0 (1.8)</td>
<td>0.006</td>
</tr>
<tr>
<td>Position 2 (Medial Plantar)</td>
<td>2.1 (0.9)</td>
<td>3.8 (2.2)</td>
<td>0.004</td>
</tr>
<tr>
<td>Position 3 (Medial Plantar)</td>
<td>2.4 (0.7)</td>
<td>4.4 (1.8)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Position 4 (Medial Plantar)</td>
<td>2.3 (0.8)</td>
<td>3.6 (1.9)</td>
<td>0.009</td>
</tr>
<tr>
<td>Position 5 (Lateral Plantar)</td>
<td>2.4 (0.7)</td>
<td>3.6 (1.7)</td>
<td>0.007</td>
</tr>
<tr>
<td>Position 6 (Medial Plantar)</td>
<td>1.7 (0.7)</td>
<td>2.9 (1.4)</td>
<td>0.003</td>
</tr>
<tr>
<td>Position 7 (Lateral Plantar)</td>
<td>2.1 (0.6)</td>
<td>3.3 (1.5)</td>
<td>0.002</td>
</tr>
<tr>
<td>Position 8 (Tibial)</td>
<td>3.3 (1.0)</td>
<td>5.4 (1.5)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Position 9 (Sural)</td>
<td>3.3 (0.9)</td>
<td>5.4 (1.6)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td><strong>Vibration</strong></td>
<td>17.0 (15.8)</td>
<td>34.8 (39.9)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td><strong>Lower Limb Proprioception</strong></td>
<td>3.6 (1.4)</td>
<td>3.2 (1.2)</td>
<td>0.431</td>
</tr>
</tbody>
</table>

Abbreviations: PD = Parkinson disease
a Data are mean (SD)

In addition to reduced tactile sensitivity, the PD participants also demonstrated a reduced capacity to detect changes in temperature compared with the controls. From the standardised starting temperature of 32°C, the older participants were able to correctly identify both a reduction (26.6°C vs. 22.2°C, P<0.01) and an increase...
(39.4°C vs. 41.1°C, P<0.05) in temperature significantly sooner than the PD participants. Unlike the cold and warm sensation tests, there were no significant differences observed between the two groups for the cold and hot induced pain tests (P>0.05). According to the accuracy results for the temperature tests, there were 10 PD participants and 4 controls who could not sense the change in temperature (cold or warm) and, hence their temperature values reached maximum. With respect to the temperature-induced pain tests, 13 and 11 PD participants did not report experiencing cold induced pain and hot induced pain, respectively, before the temperature reached its maximum setting, while all controls reported experiencing discomfort during these tests (Table 4-5).

Table 4-5: Temperature and Pain Sensory in both groups

<table>
<thead>
<tr>
<th></th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Temperature</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cold, degree</td>
<td>26.6 (4.2)</td>
<td>22.2 (11.1)</td>
<td><strong>0.006</strong></td>
</tr>
<tr>
<td>Cold Accuracy</td>
<td></td>
<td></td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>True</td>
<td>54</td>
<td>28</td>
<td></td>
</tr>
<tr>
<td>False</td>
<td>6</td>
<td>22</td>
<td></td>
</tr>
<tr>
<td>N/A</td>
<td>0</td>
<td>10</td>
<td></td>
</tr>
<tr>
<td>Warm, degree</td>
<td>39.4 (3.5)</td>
<td>41.1 (5.3)</td>
<td>0.046</td>
</tr>
<tr>
<td>Warm Accuracy</td>
<td></td>
<td></td>
<td>0.019</td>
</tr>
<tr>
<td>True</td>
<td>58</td>
<td>49</td>
<td></td>
</tr>
<tr>
<td>False</td>
<td>2</td>
<td>7</td>
<td></td>
</tr>
<tr>
<td>N/A</td>
<td>0</td>
<td>4</td>
<td></td>
</tr>
<tr>
<td><strong>Pain (Temperature induced)</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cold Pain, degree</td>
<td>19.4 (8.7)</td>
<td>15.9 (11.1)</td>
<td>0.056</td>
</tr>
<tr>
<td>Cold Pain Accuracy</td>
<td></td>
<td></td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>True</td>
<td>43</td>
<td>20</td>
<td></td>
</tr>
<tr>
<td>False</td>
<td>17</td>
<td>27</td>
<td></td>
</tr>
<tr>
<td>N/A</td>
<td>0</td>
<td>13</td>
<td></td>
</tr>
<tr>
<td>Hot Pain, degree</td>
<td>46.8 (2.3)</td>
<td>47.3 (3.8)</td>
<td>0.362</td>
</tr>
<tr>
<td>Hot Pain Accuracy</td>
<td></td>
<td></td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>True</td>
<td>59</td>
<td>46</td>
<td></td>
</tr>
<tr>
<td>False</td>
<td>1</td>
<td>3</td>
<td></td>
</tr>
<tr>
<td>N/A</td>
<td>0</td>
<td>11</td>
<td></td>
</tr>
</tbody>
</table>

Abbreviations: PD _ Parkinson disease
a Data are mean (SD)
Comparison of electrodiagnosis results between control and PD groups

Nerve Conduction

Compared with the PD group, the control group had a faster motor nerve conduction velocity for the tibial nerve (P<0.01) and the peroneal nerve that was calculated through both the ankle-fibular head and fibular head-popliteal fossa (P<0.05). Similarly, the conduction velocity of the sural nerve was significantly slower in the PD group compared to the control group (P<0.01). For the sensory nerve, the response of the sural nerve was absent for 30% of controls compared to 60% of the PD group and the PD patients had prolonged latency of onset (P<0.05). There were no differences in the amplitude of action potentials between the two groups for any of the three nerves (P>0.05). A F-wave for the peroneal nerve was obtained in 65% of controls compared with only 40% of the PD group and similar results were observed for the tibial nerve, where 80% of controls displayed F-waves, compared with approximately 50% of the PD participants. For the F-waves frequency in 10 stimulations, controls had a significantly higher frequency of occurrence of F-wave than the PD participants for both the peroneal and tibial nerves (P<0.05). However, no differences were observed in the latency of the F-wave changes between the two groups (P>0.05).

Somatosensory Evoked Potential

With respect to the SEP assessments, the N20 and P37 values could not be detected in approximately 5% of the control group and 30% of the PD group. However, based on the available data, the PD patients were observed to have prolonged latencies at both the N20 and P37 sites, but based on the statistical analyses, only the P37 latencies were statistically significant (P<0.01). In accordance with this finding, the time interval between N20 and P37 was also prolonged in participants with PD (P<0.01), but no differences were found between the groups with respect to the amplitude of the N20 and P37 measurements (P>0.05) (Table 4-6).
Table 4-6: Electrodiagnosis results of two groups

<table>
<thead>
<tr>
<th>Nerve Conduction Assessment</th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneal Nerve</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Motor NCV (AFH), m/s</td>
<td>48.4 (4.0)</td>
<td>42.9 (9.4)</td>
<td>0.023</td>
</tr>
<tr>
<td>Motor NCV (FP), m/s</td>
<td>48.5 (6.3)</td>
<td>42.6 (9.5)</td>
<td>0.028</td>
</tr>
<tr>
<td>M-wave AMP (A), mV</td>
<td>4.1 (2.3)</td>
<td>3.6 (1.7)</td>
<td>0.395</td>
</tr>
<tr>
<td>M-wave AMP (FH), mV</td>
<td>3.6 (1.7)</td>
<td>3.2 (1.7)</td>
<td>0.402</td>
</tr>
<tr>
<td>M-wave AMP (PF), mV</td>
<td>3.8 (1.8)</td>
<td>3.2 (1.8)</td>
<td>0.287</td>
</tr>
<tr>
<td>F-wave latency, ms</td>
<td>51.3 (4.0)</td>
<td>51.3 (3.5)</td>
<td>0.991</td>
</tr>
<tr>
<td>F-wave frequency, %</td>
<td>33.5 (34.1)</td>
<td>15.0 (20.4)</td>
<td>0.046</td>
</tr>
<tr>
<td>Tibial Nerve</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Motor NCV, m/s</td>
<td>48.6 (7.7)</td>
<td>40.7 (10.0)</td>
<td>0.008</td>
</tr>
<tr>
<td>M-wave AMP (Ankle), mV</td>
<td>6.3 (2.5)</td>
<td>6.1 (4.3)</td>
<td>0.831</td>
</tr>
<tr>
<td>M-wave AMP (PF), mV</td>
<td>3.7 (2.4)</td>
<td>2.6 (2.8)</td>
<td>0.200</td>
</tr>
<tr>
<td>F-wave latency, ms</td>
<td>54.1 (4.6)</td>
<td>57.6 (7.9)</td>
<td>0.155</td>
</tr>
<tr>
<td>F-wave frequency, %</td>
<td>70.0 (40.3)</td>
<td>41.0 (42.9)</td>
<td>0.034</td>
</tr>
<tr>
<td>Sural Nerve</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sensory NCV, m/s</td>
<td>46.9 (3.9)</td>
<td>41.9 (4.1)</td>
<td>0.013</td>
</tr>
<tr>
<td>SNAP AMP, µV</td>
<td>12.5 (4.1)</td>
<td>14.5 (2.4)</td>
<td>0.255</td>
</tr>
<tr>
<td>Latency to onset, ms</td>
<td>3.0 (0.3)</td>
<td>3.4 (0.4)</td>
<td>0.013</td>
</tr>
<tr>
<td>SEP</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Latency of N20, ms</td>
<td>21.7 (2.1)</td>
<td>22.2 (3.7)</td>
<td>0.622</td>
</tr>
<tr>
<td>Latency of P37, ms</td>
<td>39.2 (2.3)</td>
<td>45.7 (5.8)</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>N20 AMP, µV</td>
<td>1.1 (0.3)</td>
<td>1.1 (0.7)</td>
<td>0.831</td>
</tr>
<tr>
<td>P37 AMP, µV</td>
<td>0.9 (0.4)</td>
<td>0.9 (0.5)</td>
<td>0.974</td>
</tr>
<tr>
<td>N20-P37 Interval, ms</td>
<td>17.4 (2.4)</td>
<td>23.5 (5.8)</td>
<td>0.002</td>
</tr>
</tbody>
</table>

Abbreviations: PD _ Parkinson disease; NCV _ Nerve Conduction Velocity; AFH _ Ankle-Fibular Head; FP _ Fibular Head-Popliteal Fossa; AMP _ Amplitude; FH _ Fibular Head; PF _ Popliteal Fossa; SNAP: Sensory Nerve Action Potential

Results of Clinical Balance Assessments

In the clinical tests of balance, the PD group performed more poorly than the control group for the Tinetti (balance, gait, total) assessments (P<0.05) (Table 4-7).
Table 4-7: Clinical tests of balance in two groups

<table>
<thead>
<tr>
<th>Clinical Assessments of Balance</th>
<th>Control (n=20)</th>
<th>PD (n=20)</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Balance</td>
<td>16.1 (0.5)</td>
<td>14.0 (2.6)</td>
<td>0.002</td>
</tr>
<tr>
<td>Gait</td>
<td>11.2 (1.0)</td>
<td>9.9 (2.0)</td>
<td>0.016</td>
</tr>
<tr>
<td>Total</td>
<td>27.3 (1.1)</td>
<td>24.0 (4.3)</td>
<td>0.003</td>
</tr>
</tbody>
</table>

Abbreviations: PD _ Parkinson disease

Standing Balance Assessments

Main Effects

The PD group demonstrated more postural sway in both the anterior-posterior and medial-lateral directions than controls (P<0.01). The postural sway was more obvious when standing on the foam surface (P<0.001) and anterior-posterior sway was increased without visual input (P<0.001) (Figure 4-19).

Figure 4-19: Main effects in anterior-posterior (Black) and medial-lateral (Grey) sway during standing balance

Two way interactions

There was an increase in AP and ML sway when standing on the foam surface with eyes closed relative to eyes open (P<0.05). In contrast, there was no difference in postural sway when standing on a firm surface with eyes open or closed (Figure 4-20).
Gait Assessment

Main Effect of Group

For the main effect of group, PD patients were observed to take shorter strides and to have reduced stride periods compared with the controls (P<0.01). Furthermore, the participants with PD spent more time in the stance phase, and had increased cadence and stride timing variability than the control group (P<0.05). The PD participants also tended to spend an increased percentage of the gait cycle in double support; however this trend did not achieve statistical significance. There were no differences observed for walking velocity between the two groups (P>0.05) (Figure 4-21).
Figure 4-21: Main effects of group (Black- Control group, Grey- PD group) in temporospatial gait assessments during walking on firm and uneven surfaces with barefoot.

**Main Effect of Surface**

Participants took longer strides, had longer stride times and displayed increased stride timing variability when walking on the uneven surface compared to the firm surface (P<0.01). Furthermore, participants walked slower, had reduced cadence, and spent less time in the stance phase on the uneven surface compared with the firm surface (P<0.05). No changes were recorded for double support time between the two walking surfaces (P>0.05) (Figure 4-22).
Two Way Interactions

The results of the mixed-model ANOVA identified significant group*surface interactions for stride length, velocity, cadence, stride time and stance phase (P<0.05).

The results demonstrated that the PD and control groups walked at a similar speed on the firm surface. The PD participants were observed to reduce their walking speed while walking on the uneven surface compared to the firm surface, while the control group maintained the same walking velocity on both walkways. This lead to a tendency for the PD group to walk slower than the control group on the uneven surface (P=0.059). (Figure 4-23).
The PD participants had significantly higher cadence than the control group while walking on the firm surface (P<0.01). However, while walking on the uneven surface, both groups reduced cadence, but this was more pronounced for the PD participants, which resulted in similar values for the two groups on the uneven surface (Figure 4-24).

With respect to stride length, there were no significant differences between the groups on the firm surface. However, while the control group increased their stride length on the uneven surface (P<0.001), the PD group did not (Figure 4-25).
Figure 4-25: The mean (+1SD) stride length for the Control (Black) and PD (Grey) groups while walking on the firm and uneven surfaces.

On the firm surface, the PD group had shorter stride periods than the control group (P<0.01), but they increased their stride period to a greater extent on the uneven surface, which resulted in similar values for the two groups on the more difficult surface (P<0.001) (Figure 4-26).

Figure 4-26: The mean (+1SD) stride period of the Control (Firm surface) and PD (Grey) groups while walking on the firm and uneven surfaces.

Both groups had similar stance periods on the firm surface, but the PD group had significantly longer stance periods than the control group on the uneven surface (P<0.001). Compared with the firm surface, both groups spent a decreased percentage of the gait cycle in the stance phase on the uneven surface (P<0.05), but this decrease was greater for the control group. (Figure 4-27).
Figure 4-27: The mean (+1SD) stance phase for the Control (Black) and PD (Grey) groups while walking on the firm and uneven surfaces.

**Correlations**

The results of correlations between somatosensory function and the temporospatial gait characteristics are shown in Table 4-8. The results of these analyses suggested that the clinical balance tests (Tinetti balance, gait and total scores) were all negatively associated with peripheral sensory (touch, vibration thresholds) in the PD group (P<0.05) (Figure 4-28). Given that larger Tinetti scores represent better function and lower touch and vibration sensitivity scores represent better sensitivity, the negative association between these measures indicates that better mobility scores are correlated with improved sensitivity. With respect to temperature and pain sensation, participants with higher Tinetti gait and total scores tended to have better cold sensitivities in the PD group. While PD participants with increased warm and hot-induced pain thresholds tended to have lower scores in all Tinetti clinical balance tests. No significant correlations were observed for controls.

For the nerve function tests, both the Tinetti gait and total scores increased with the peroneal and tibial nerve velocity (P<0.05), while the Tinetti balance score was only positively associated with the peroneal nerve velocity at the ankle-fibular head site (P<0.05). No significant correlations were observed for the control group.
Table 4-8: Correlations between sensory functions and postural assessments.

<table>
<thead>
<tr>
<th>Sensory Function</th>
<th>Temperature and Pain Sensory</th>
<th>Nerve Function</th>
<th>Somatosensory Evoked Potential</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peripheral Sensory</td>
<td>Temperaturte and Pain Sensory</td>
<td>Nerve Function</td>
<td>Somatosensory Evoked Potential</td>
</tr>
<tr>
<td>Tinetti Balance Score</td>
<td>R=0.485, P=0.01</td>
<td>R=0.48, P=0.012</td>
<td>R=0.47, P=0.017</td>
</tr>
<tr>
<td>Tinetti Gait Score</td>
<td>R=0.485, P=0.01</td>
<td>R=0.536, P=0.046</td>
<td>R=0.47, P=0.033</td>
</tr>
<tr>
<td>Tinetti Total</td>
<td>R=0.485, P=0.01</td>
<td>R=0.508, P=0.009</td>
<td>R=0.44, P=0.032</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Clinical</th>
<th>Quiet Stance</th>
<th>Foam Surface</th>
<th>Uneven Surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>FIRM SURFACE</td>
<td>Eyes Open: AP</td>
<td>R=0.442, P=0.051</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Eyes Open: ML</td>
<td>R=0.471, P=0.036</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Eyes Closed: AP</td>
<td>R=0.451, P=0.046</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Eyes Closed: ML</td>
<td>R=0.434, P=0.056</td>
<td></td>
</tr>
<tr>
<td>FOAM SURFACE</td>
<td>Eyes Open: AP</td>
<td>R=0.44, P=0.052</td>
<td>R=0.621, P=0.003</td>
</tr>
<tr>
<td></td>
<td>Eyes Open: ML</td>
<td>R=0.471, P=0.036</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Eyes Closed: AP</td>
<td>R=0.468, P=0.038</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Eyes Closed: ML</td>
<td>R=0.451, P=0.046</td>
<td></td>
</tr>
<tr>
<td>GAIT</td>
<td>Stride Length</td>
<td>R=0.436, P=0.015</td>
<td>R=0.512, P=0.021</td>
</tr>
<tr>
<td></td>
<td>Stride Timing Variability</td>
<td>R=0.477, P=0.033</td>
<td>R=0.464, P=0.039</td>
</tr>
<tr>
<td></td>
<td>Double Support Time</td>
<td>R=0.477, P=0.033</td>
<td>R=0.464, P=0.039</td>
</tr>
<tr>
<td></td>
<td>Uneven Surface</td>
<td>Stride Length</td>
<td>R=0.547, P=0.013</td>
</tr>
<tr>
<td></td>
<td>Stride Timing Variability</td>
<td>R=0.436, P=0.055</td>
<td>R=0.539, P=0.014</td>
</tr>
<tr>
<td>Abbreviations: AP _ Anterior-posterior; ML _ Medial-lateral; AFH _ Ankle-Fibular Head; FP _ Fibular Head-Popliteal Fossa; Red: PD group; Green: Control group.</td>
<td></td>
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</table>
Figure 4-28: Relationship between peripheral sensation (touch and vibration) and the Tinetti clinical assessments (balance, gait and total scores) for the Control (Black) and PD (Grey) participants.
For standing balance, only vibration threshold was positively associated with medial-lateral sway on the firm surface with eyes open for the PD group (R=0.467, P<0.05). PD participants with quicker nerve transfer speeds for the peroneal nerve (measured at the ankle-fibular head) and the tibial nerve seemed to have less medial-lateral sway on the firm surface with eyes open. The control group showed a similar trend for anterior-posterior sway on the foam surface with eyes open. Conversely, medial-lateral sway tended to be negatively correlated with peroneal nerve velocity (measured at the ankle-fibular head) for the PD group when standing on firm surface without visual input (P=0.056). When standing on the firm surface, PD participants with slower tibial nerve velocity tended to have greater anterior-posterior sway, while anterior-posterior sway was negatively correlated with tibial nerve velocity on the foam surface for the control group with both eyes open (R=-0.621, P<0.05) and eyes closed (R=-0.468, P<0.05) conditions.

When PD participants were walking on the firm surface, stride length was negatively correlated with vibration sense, warm sensitivity and hot-induced pain thresholds and positively correlated with cold sensitivity (P<0.05). There was also a tendency for stride length to be negatively correlated with touch threshold and positively correlated with peroneal nerve velocity (as measured at the fibular head-popliteal fossa) (P=0.055). On the uneven surface, stride length was significantly negatively correlated with touch sensitivity, vibration sense, warm sensitivity and hot-induced pain thresholds and positively correlated with cold sensitivity, cold-induced pain threshold, peroneal nerve velocity (as measured at the fibular head-popliteal fossa) and tibial nerve velocity (P<0.05). There was also a tendency for a negative correlation between stride length and N20 Latency (P=0.055).

With respect to stride timing variability on the firm surface, higher values were associated with increased warm threshold, slower peroneal nerve velocity (Ankle-Fibular Head part) and tibial nerve velocity (P<0.05).
Double support time values demonstrated positive correlations with warm and hot-induced pain thresholds on both surfaces, however the correlation with warm thresholds on the uneven surface was not significant (P=0.055).

### 4.4 Discussion

This study examined whether there were differences in somatosensory function between people with PD and healthy age-matched controls and also examined how changes in somatosensory information may influence balance and gait. Overall, the PD group demonstrated a declined in somatosensory function compared to the control group and there was a significant association between decreased somatosensory function and deficits of postural control in the former.

**Peripheral Sensory**

The increased touch and vibrations thresholds for the PD participants were consistent with previous research (Konczak et al., 2009; Pratorius et al., 2003; Weder et al., 1999) and demonstrated reduced perception of somatosensory stimuli.

The increased touch thresholds across all stimulation positions of the foot and ankle corresponded to innervation regions of the medial plantar, lateral plantar, tibial and sural nerves. Other studies have also reported that cutaneous mechanoreceptors were spread through the foot sole, including both slowly adapting (SAI, SAIi) and fast adapting (FAI, FAII) receptors. Over 70% of the skin receptors are rapidly adapting receptors according to monofilaments stimulation (Kennedy & Inglis, 2002). These receptors are preferentially affected in PD participants (Nolano et al., 2008). This finding indicates that there was an overall decline in peripheral sensation for the PD group. This result was supported by previous studies that have reported diminished proprioception, tactile discrimination and vibration sensitivity for the soles of the feet.
in people with PD (Konczak et al., 2009; Pratorius et al., 2003; Weder et al., 1999). Furthermore, a series of skin biopsies performed by Nolano et al. (2008) for the fingertip, thigh and distal leg identified cutaneous denervation and sensory impairment in people with PD. The decreased peripheral sensation was associated with increased postural instability and was considered an important predictor of falls in people with Parkinson’s disease (Carpenter & Bloem, 2010; Kerr et al., 2010).

In contrast to the previously reported declines in proprioception in people with PD (Konczak et al., 2009; Tagliabue et al., 2009), no differences were observed in this study. This may have been because most proprioception tests that have reported a decline in PD participants have targeted the upper limbs, whereas the present study evaluated lower limb proprioception sensitivity and focused more on the knee joint position sensation.

**Temperature and Pain Sensory**

The impairments in thermal sensation and nociception demonstrated by the PD participants in the current study are supported by previous research, which demonstrated that thermal thresholds were increased significantly in PD participants (Nolano et al., 2008).

Pain sensation thresholds for warm and cold stimuli were also increased in the present study, with over 50% of the PD participants unable to report experiencing pain before the temperature (hot and cold) reached the predefined maximum limit. Therefore, given that the data presented for the two groups only calculated the maximum values for those who were unable to detect the painful stimulus, the reported pain thresholds represent a conservative estimate of thermally-induced nociceptive thresholds. In
summary, the PD participants performed worse in all temperature and pain tests than the control group, which indicated that there was deficit in thermal and nociception sensation in people with PD.

A previous study provides support for this notion, indicating that PD patients had a significant increase in thermal threshold and a reduction in mechanical pain perception (Nolano et al., 2008). The authors postulated that these deficits were due to a loss in free and encapsulated nerve endings (Nolano et al., 2008), which would affect the Aδ-fibres that respond to thermal, mechanical, fast pain stimuli and the C-fibres that transmit slow pain sensation to the central nervous system. Thus, these findings suggest that signal transmission was diminished on the spinothalamic and dorsal column-medial lemniscus pathway. If the function of receptors is impaired, people with PD cannot receive normal information from pain and temperature stimulus.

In contrast, other studies (Brefel-Courbon et al., 2005; Djaldetti et al., 2004; Mylius et al., 2011; Mylius et al., 2009) have reported that PD patients demonstrated greater heat pain sensitivity than controls. This difference may be due to the fact that some participants in these studies reported having some clinical pain prior to their involvement in the research, which may have caused them to be more sensitive to painful stimuli (Mylius et al., 2009). Furthermore Mylius et al. (2009) showed that the decreased pain threshold only occurred in PD participants during those periods where patients were in a medication free period. The PD participants assessed in the current research were all optimally-medicated at the time of testing or did not require
medication for their symptoms, which may account for the differences observed in the sensory responses.

Electrodiagnosis

Nerve Conduction

The nerve conduction results demonstrated that the PD group had significant decreases in velocity and F-wave frequency for the motor component of the tibial and peroneal nerves around the sole area compared to the control group. Similar results were observed by Toth et al. (2010), who reported that PD participants had slower conduction velocities for the motor component of the peroneal and tibial nerves compared with controls. The comparable M-wave amplitudes recorded for the PD and control groups in the present study indicated that both groups demonstrated similar muscle activation for the lower extremities. Abbruzzese et al. (1985) evaluated the electrophysiological function of the nerves of the upper limbs, while the present study assessed the lower limbs, which may have contributed to some differences observed between these two studies: There were no differences reported for the latency of the F-wave for the peroneal and tibial nerves, which was in accordance with the results from Abbruzzese et al. (1985). However, the PD sample in the present study comprised fewer people for whom the F wave values of the motor nerves were obtained compared to the control group. This finding indicated that there is a breakdown along the pathway responsible for transmitting the impulse to the spinal cord and back for the PD group. Thus, transmission deficits of motor nerve in PD group cannot be ruled out. Conversely, Abbruzzese et al. (1985) showed that the amplitude of the F-wave was significantly increased in people with PD. A possible explanation for this disparity may be that the participants assessed by Abbruzzese et
al (1985) had prominent symptoms of rigidity and akinesia, which would increase the amplitude of the F-wave by recruiting additional motor units. In contrast the participants assessed in the current study were all in the early stages of the disease (Hoehn & Yahr: 1.4±0.9) and showed no obvious signs of rigidity (average rigidity for neck, wrists and ankles: 1.0±0.9).

With respect to sensory nerve function, the conduction velocity and latency of the sural nerve was decreased in the PD group compared to control group, data in agreement with the findings presented by Toth (2010). Nolano et al. (2008) reported that the conduction velocity of the sural nerve was normal in a group of 18 PD participants, despite the presence of peripheral deafferentation and sensory dysfunction. However, Nolano et al. (2008) did not include a control group for comparison. In fact, according to the standardized normal values of nerve conduction tests (Weiss et al., 2004), the nerve conduction velocities and latencies for the peroneal, tibial and sural nerves were all within the normal range for the PD group in the present study. This is despite the PD group performing worse than the control group. Sural nerve velocity was also within the normal range for Nolano’s study (2008), indicating the consistent results of sural nerve function in present study. Therefore, while Nolano’s study (2008) provided evidence of impaired sensory receptor functioning in people with PD based on biopsy, the potential for sensory nerve fibre degeneration cannot be ruled out in the present study. This in turn may lead to decreased nerve conduction velocity in people with PD when compared to controls. In the present study, also it was noticed that it was difficult to record sural nerve responses (only 8 PD participants) due to difficulties with stimulating this nerve to elicit an appropriate response. Therefore caution should be taken when interpreting the explanation of these findings. Finally, no differences were noticed in the
amplitude of the sensory nerve action potentials in present study, which was similar to the results presented by Nolano et al. (2008).

In summary, the electrophysiological tests demonstrated that the PD participants had poorer nerve conduction than age-matched controls. These changes in nerve function are likely to contribute to a higher incidence of peripheral neuropathy in people with PD. Based on a random sample of 1000 people with PD, Toth et al. (2010) reported that 43% of PD participants presented with peripheral neuropathy. Given that this prevalence is higher than that of age-matched controls (Chovancova et al., 2008), these findings suggest that the neurodegenerative symptoms associated with PD may affect the nervous system peripherally as well as centrally. A possible mechanism for the decline in nerve conduction velocity evident in the PD group is the degeneration of the myelin sheath surrounding the nerves (Waxman, 1980). This explanation is supported by the biopsy results presented by Nolano et al. (2008), which demonstrated axonal swelling and demyelination at the distal ends of the peripheral nerves in PD participants.

**Somatosensory Evoked Potential**

The SEP results showed that, compared with controls, PD participants demonstrated no differences for the Lumbar N20 recording, but that there was a significant delay in the P37 cranial measure. The lumbar N20 potential is a measure of the sensory reflex received at the spinal area, which originates from the postsynaptic neurons in the gray matter of the spinal cord (Legatt, 2009). Absence or prolonged lumbar N20 potentials is indicative of proximal demyelination in peripheral nerve disorders (Chawla, 2010; Yiannikas & Vucic, 2008). The similar lumbar N20 potentials recorded in this study
for the two groups suggested that the transmission of information from the ankle to the spinal cord was not significantly delayed in people with PD. However, it is important to reiterate that the PD group had significantly slower nerve conduction velocities and increased latencies for the sural nerve compared to controls. This conflict may have resulted due to the difficulties experienced while assessing the sural nerve, which limited the data for these measures to 8 PD participants. Furthermore, the sural nerve was only evaluated over a relatively short distance (14cm) from the ankle according to the stimulation and recording positions, while the N20 assessments were recorded between the posterior tibial nerve and the spinal cord. Given the greater distance of measurement for the N20 assessments, these tests would be likely to cover a mixture of nerves that include the sural nerve. Nonetheless, the normality of the N20 recording cannot rule out the presence of potential deficits in somatosensory function for the PD group, particularly given the evident declines in cutaneous function compared to control group.

There was an increase in P37 latency in this study, indicating the dysfunction within the somatosensory information transfer from the spinal to the cerebral cortex in PD participants. With respect to P37, it is a primary cortical SEP following lower limb stimulation and representative of the sensory information transferred to the lower limb areas of the somatosensory cortex (Legatt, 2009). The ventral posterior nucleus of the thalamus is a key somatosensory relay that receives somatosensory information from the peripheral nerves and sends it to the primary somatosensory cortex. The primary somatosensory cortex is located in the post-central gyrus (Künzle, 1977; Saladin, 2004). P37 arises from independent cortical generators, located in the pre-rolandic cortex, which may be selectively affected by basal ganglia dysfunction (Tinazzi et al.,
Thus dysfunction of the basal ganglia area might account for the observed P37 abnormality (Kievit & Kuypers, 1977; Legatt, 2009; Lidsky et al., 1985; Schell & Strick, 1984). The prolonged P37 responses evident in the PD group may have represented demyelination of the dorsal lumbar nerve rootlets, which would contribute to distal peripheral neuropathy in this population (Yiannikas & Vucic, 2008). Other abnormalities occurred during the transmission of information, which may have also contributed to this increased latency, including demyelination of the nerve projections that resulted in a delay in transmission of information from the lumbar area to thalamus and somatosensory cortex, or basal ganglia deficits that affected the central transmission. The results of the current study were supported by Rossini et al. (1989) who reported a significant decrease in the frontal wave for people with PD, as the frontal wave also reflects frontal cortex where a large amount of sensory information is received. Given that both groups showed similar latencies for the N20 measure, the longer latency recorded for the P37 suggests that the conduction time between the spinal cord and the cerebral cortex is longer for people with PD. This indicated transmission of somatosensory information was affected on the pathway from spinal cord to central nervous system in people with PD. Furthermore, we noticed that fewer controls were missing N20 and P37 values than people with Parkinson’s disease, indicating transmission of somatosensory information was worse in PD group.

**Balance**

Both clinical and force plate assessments of balance revealed that the PD group performed worse than the control group. The Tinetti clinical test has been used previously to predict falls in older people (Raîche et al., 2000) and people with
Parkinson’s disease (Kegelmeyer et al., 2007; Kerr et al., 2010). The results presented in this study, for the Tinetti balance sub-score, were in accordance with the scores previously reported by Kerr et al. (2010) for PD participants who prospectively reported falling. Similarly, the increased COP values reported for the PD group in both the AP and ML directions were consistent with previous research (Adkin et al., 2005; Mitchell et al., 1995) and have been linked to an increased risk of falling in this population (Kerr et al., 2010).

The increased postural sway observed when standing on the foam surface was consistent with previous research (Horlings et al., 2008; Patel et al., 2011; Vuillerme & Pinsault, 2007; Wu & Chiang, 1997). These studies suggested that the increased postural sway may be attributed to decreased information from the cutaneous mechanoreceptors in the soles of the feet due to the softer foam surface, which alters somatosensory feedback.

Similarly, the PD participants demonstrated increased AP postural sway when standing on the firm and foam surfaces with eyes closed, which is in agreement with previous studies conducted by Kerr et al. (2010) and Latt et al. (2009). The visual system is known to play an important role in maintaining balance control by receiving visual information from the surrounding environment and provides the nervous system with synchronous information about body position and the environment. The importance of the visual system in balance control has been demonstrated in humans, with research showing that body sway can increase by around 36% when vision is excluded (Ehrenfried et al., 2003; Vidal et al., 1982). While ML sway in this study did not show a significant increase, other studies have reported that both AP and ML
sway are increased when visual input is deprived in both PD and older participants (Adkin et al., 2005; Moe-Nilssen & Helbostad, 2002).

In summary, both AP and ML sway were increased significantly when standing on the foam surface compared to the firm surface. Under the most challenging condition, foam surface with eyes closed, the greatest increase, relative to the foam surface eyes open condition, was observed in the AP direction; there was no increase in ML sway. In this situation, closing the eyes provided a heavier emphasis on vestibular and somatosensory inputs to maintain postural control. However, given that the foam surface diminishes the reliability of the feedback received from the feet, standing on this surface exposed participants to a more challenging environment. As such, the participants had to adjust their posture to maintain their balance control, which resulted in more postural sway in the AP direction.

**Gait Assessments**

Both the clinical and motion analysis-based assessments of gait revealed that people with PD had altered gait relative to the control group. PD patients had lower scores for the Tinetti gait subscale which were comparable to those previously reported for PD patients who fell (Kerr et al. 2010). In agreement with previous research, there were also differences in the temporospatial gait parameters for the PD group compared to the control group (Almeida et al., 2007; Del Olmo & Cudeiro, 2005; Morris et al., 2005; Yang et al., 2008).

Compared to the decreased velocity reported in many studies (Almeida et al., 2007; Del Olmo & Cudeiro, 2005; Hausdorff et al., 1998; Knutsson, 1972), walking speed
was not significantly different between the PD and control groups, but the PD group took shorter strides than controls. Similar decreases in stride length were reported by many other studies in PD (Knutsson, 1972; Rogers, 1996; Ueno et al., 1993) and research suggests that regulation of stride length is a fundamental problem in gait adjustment in PD participants (Morris et al., 1994b).

Stride timing variability was increased in the PD group, which was consistent with previous research (Cole et al., 2010; Hausdorff et al., 1998). Hausdorff et al. (1998) pointed out that increased stride timing variability indicated an inability to accurately regulate variations in gait cycle timing. Hausdorff et al (1998) also indicated that the increased stride timing variability may be reflective of a discontinuous status of walking, which was due to the impairment in basal ganglia function and directly related to the impaired reflexes responsible for the postural instability in PD.

**Difference When Walking on An Uneven Surface**

Walking on an uneven surface presents a challenge for participants, which requires significant alterations in gait (Menz et al., 2003b; Thies et al., 2005). In the present study the uneven surface caused alterations in walking velocity, cadence, stride length, stride period, stride timing variability and stance phase. However, these alterations were different for the PD and control groups. On the uneven surface, the controls maintained their walking speed by increasing their stride length and stride period and decreasing their cadence and stance period. Conversely, while the PD group also had a longer stride period and decreased cadence and stance periods, they did not increase their stride length on the uneven surface (Morris et al., 1994a; 1994b). While these gait adaptations may be reflective of a more cautious gait pattern, the ability to regulate stride length in the PD population has long been considered to be a
characteristic of gait impairment in this population (Lewis et al., 2000; Morris et al., 1996).

Previous research has shown that both step length and step width are increased to ensure a large base of support to enhance stability when walking on a compliant surface (Cole et al., 2011; MacLellan & Patla, 2006). The alterations in stride length, cadence and stride timing variability when walking on an uneven surface observed for older people in this study were consistent with those reported by Menz et al (2003b). Furthermore, the controls in this study were able to maintain their walking speed on the uneven surface, similar to that reported by Menz et al (2003b), which was achieved by changing step length and cadence. However, the PD participants assessed in this study did not demonstrate the same adaptation as controls, resulting in a decreased walking speed. Furthermore the increased stride timing variability demonstrated by both groups on the uneven surface, indicated that gait may become less regular on unpredictable surfaces, as suggested by other studies (Cole et al., 2011).

**Correlations**

There were associations between some of the measures of somatosensory and peripheral nerve function and the balance and gait measures for the PD group, but not for the control group. Poorer performance on the clinical Tinetti tests were closely associated with declines in tactile, temperature and pain sensation and peripheral nerve function in the PD group. Declines in somatosensory function for people with PD have been shown to lead to abnormal postural reactions to perturbations and a reduced capacity to recognise the direction of motion (Carpenter & Bloem, 2010).
Previous research has also reported that the Tinetti total score had a moderate sensitivity and specificity for predicting falls in PD patients (Kerr et al., 2010) and older people (Tinetti et al., 1988), with poorer performance being related to poorer touch sensitivity and more AP postural sway. Due to the decline in peripheral sensory function, PD participants may be unable to transfer enough feedback to the central nervous system to adjust their posture efficiently. These deficits may influence the participants’ capacity to adjust their posture during tasks like sitting, rising from a chair and turning and may also influence posture during gait, which would be captured by the Tinetti tests.

Among the gait variables, stride length was closely associated with the somatosensory tests and nerve function in the PD group. With respect to somatosensory function, increased thresholds for touch and vibration were correlated with reduced stride length. For the electrodagnosis tests, there was a tendency for a negative correlation between postural sway and nerve conduction velocity for the peroneal and tibial nerves in both groups. These results suggest that the delays in the information transfer between the peripheral nerves and the central nervous system may affect postural control and lead to increased postural sway during standing. Furthermore, a decline in the nerve conduction velocity of the tibial nerve was also associated with reduced stride length in the PD group. This result was supported by previous research that showed reduced stride length has been an important marker of postural instability during walking, and has been associated with gait disturbance and increase risk of falls (Morris et al., 1994b; 1996). Morris et al. (1996) also indicated that an inadequate ability to activate the motor control system due to the deficits in the basal ganglia lead to gait hypokinesia in people with Parkinson’s disease, especially in term of shorter stride length. Thus, the worse was the transmission of somatosensory information to the central nervous system, the less information was obtained by the motor control system to regulate the stride length.
Study 2 revealed that there were declines of somatosensory function in people with Parkinson’s disease compared to healthy older controls. These decrements occurred in touch, vibration, temperature and nociception, nerve function and SEP. In the PD group, there was a close association between decreased somatosensory function and decreased clinical balance (Tinetti Test) score. Decline in somatosensory function also closely correlated with temporospatial gait assessments, especially shorter stride length.
Chapter 5

Study 3

5.1 Introduction

Somatosensory feedback plays an important role in balance control and declined somatosensory function due to ageing and disease has been closely associated with impaired mobility and falls in older people (Lord et al., 2001) and patients with Parkinson’s disease (Carpenter & Bloem, 2010).

Many previous studies have altered the quality of the somatosensory information received from the feet to determine how it affects postural stability. Methods have included ischemic hypoxia of foot induced by a pressure cuff placed around the calcaneus (Diener et al., 1984), immersion of feet in iced water and standing on a foam surface (Magnusson et al., 1990a; 1990b; Perry et al., 2000). All of these methods have been suggested to decrease or alter the somatosensory input from the foot and ankle and to cause postural instability. Given the apparent relationship between reduced somatosensory function and poorer postural stability, it is perhaps not surprising that numerous methods have been utilised to enhance somatosensory feedback and improve postural stability. Artificially enhancing cutaneous information feedback via mechanical vibration devices has been shown to reduce postural sway (Priplata et al., 2002; Wu et al., 2007). Similarly, separate research has shown that smooth strips of athletic tape placed across both ankle joints can provide additional somatosensory feedback to improve the adjustment of posture control (Simoneau et al., 1997; Vuillerme & Pinsault, 2007).
Recent research has also suggested that passive devices may provide an inexpensive and effective alternative to decrease postural sway in people with balance problems. In a recent study by Hatton et al. (2011) it was reported that mediolateral sway was decreased in older people when standing on textured surfaces. Similarly, Palluel et al. (2009) reported reduced postural sway during quiet stance for older people while wearing sandals with firm rubber nodules, and Corbin et al. (2007) reported reduced postural sway in younger participants while wearing insoles which had a textured pattern. While these studies suggest that artificially-enhancing somatosensory information from the feet may be effective in improving standing balance in people with balance impairments, there has been little research involving dynamic tasks, such as walking. Maki et al. (2008) used footwear inserted with raised compliant ridge around the perimeter of the foot to stimulate the cutaneous mechanoreceptors near the sole of the foot and their results showed improved lateral stability during stepping. Similarly, Ritchie et al. (2011) examined the effects of shoes with nodules that applied pressure to the plantar-medial surface of the foot. Their results showed that the shoes were effective in controlling excessive foot pronation, which was important for preventing injuries. However, little work has been conducted to test the efficacy of enhanced somatosensory input on gait and balance in PD patients (Jenkins et al., 2009; Novak & Novak, 2006). As such, there is a clear need to develop and evaluate simple and inexpensive interventions that can be used in daily activities to improve somatosensory function and benefit postural control in both healthy and PD populations.

Study 3 examined the efficacy of a newly designed textured insole for improving postural control during standing and walking in a group of people with PD and age-matched controls. Due to the deleterious effects of Parkinson's disease on somatosensory function, it was expected that textured insole attenuation effects on
postural control were likely to be greater in the PD group compared to the control group.

5.2 Method

5.2.1 Participants

For the purposes of this study, the same 20 people with Parkinson’s disease and 20 healthy controls participating in Study 2 were recruited.

5.2.2 Experimental Protocol

Postural Stability Assessment

3D Gait Assessment

To evaluate the efficacy of the textured insoles in improving postural control, a comprehensive gait assessment was conducted for all participants using the same equipment and procedures outlined in study 2. Specifically, postural stability was assessed during standing (static) and walking (dynamic) tasks under three conditions, which were completed in a random order: 1) wearing shoes fitted with textured insoles; 2) wearing shoes fitted with smooth insoles (no texture); and 3), barefoot.

The textured insoles used in this study were designed based on the results of study 1, but were slightly modified to include raised compliant edges based on recent research by Maki et al. (2008) who examined the effects of textured insoles on gait. Specifically, the insoles were 1.5mm thick and constructed using the soft insole material (270 density EVA) (International Children’s Orthotic Laboratory, Brisbane, Australia). The textured surface comprised granulations measuring 5.0mm in diameter and 3.1mm in height that were distributed evenly across the upper surface. Two raised compliant ridges measuring 3.1mm in height and 3.1mm in width were located around
the lateral perimeter of the insole and around the heel of the foot (Figure 5-1). For the purposes of the assessments, the insoles were inserted into standardized footwear (Donated by Pacific Brands Australia Pty Ltd), which were a basic construct rubber soled shankless shoe with a soft canvas upper (Figure 5-2).

Figure 5-1: Textured insole utilized in this study.

Figure 5-2: standardized footwear

For each of the footwear conditions, participants completed 6 walking trials on two walking surfaces: 1) a flat and firm surface, and 2), an uneven surface, both of which have been described in study 2 (Figure 4-13).

Following data collection, the data were labelled and processed using the same methods described previously, and the temporospatial data were derived to evaluate differences between the insole conditions.
Standing balance assessment

To examine postural stability under static conditions, postural sway was assessed for each participant using the force plate described for Study 2. Centre of pressure data were recorded at 1000Hz for two 30 second trials performed under each of the following conditions: 1) on a firm surface with eyes open; 2) on a firm surface with eyes closed; 3) on a foam surface with eyes open; and 4) on a foam surface with eyes closed. This test battery was repeated three times in random orders to allow for the assessment of the three different footwear conditions; 1) barefoot; 2) smooth insoles; and 3), textured insoles. The same procedure was followed as described for Study 2. Measurements derived from centre of pressure displacements included the range of anterior-posterior and medial-lateral displacements.

5.2.3 Data Analysis

Gait and Force Plate Analysis were the same as described in Study 2.

5.2.4 Statistical Analysis

For the gait data, a mixed model Analysis of Variance (ANOVA) with 1 between-participant (PD; older) and 2 within-participant factors, including footwear (barefoot; smooth insole; textured insole) and surface (firm; uneven) was used to compare the experimental data. For the static balance data, a mixed model Analysis of Variance (ANOVA) with 1 between-participant (PD; older) and 3 within-participant factors, including footwear (barefoot; smooth insole; textured insole), vision (eyes open; eyes closed) and surface (firm; foam) was used to compare the experimental data. Violations of the sphericity assumption for repeated measures variables were checked using Mauchley’s test of sphericity. When a violation of this assumption was apparent, the Greenhouse-Geisser correction procedure was used to adjust the degrees of freedom of the error term for the F ratios. Post hoc comparisons were undertaken...
using Least Significant Difference (LSD) test. Statistical significance was set at the 95% confidence level (P<0.05). Data were analysed using the Statistical Package for Social Sciences (SPSS V17.0, Chicago, IL, USA).

5.3 Results

Standing Balance Main Effects

The PD group displayed significantly more postural sway than the control group in both the anterior-posterior and medial-lateral directions (P<0.01). Standing on the foam surface produced significantly greater postural sway compared to the firm surface in both directions (P<0.001). Attenuation of visual feedback (eyes closed) also increased postural sway in the anterior-posterior and medial-lateral directions (P<0.01). Conversely, neither anterior-posterior nor medial-lateral sway were affected by the different insoles (P>0.05) (Figure 5-3).

Figure 5-3: Main effects of group, surface, vision and insole for postural sway in anterior-posterior (Grey) and medial-lateral (Black) directions.
Two-way interactions

With respect to anterior-posterior sway, only the surface\(^*\)vision interaction achieved statistical significance. The post-hoc analysis demonstrated that although there was no significant change in postural sway while standing on the firm surface with eyes open and closed, anterior-posterior sway was significantly increased on the foam surface without visual input (P<0.001) (Figure 5-4).

![Figure 5-4: Anterior-posterior sway in Surface\(^*\)Vision interaction (Black-Firm surface, Grey- Foam surface)](image)

For mediolateral sway there were significant surface\(^*\)vision and surface\(^*\)group interactions (P<0.01) as shown in Figure 5-5. Post-hoc analyses for the surface\(^*\)group interaction demonstrated that both groups displayed increased medial-lateral sway when standing on the foam surface (P<0.001), but that the PD group exhibited significantly more postural sway than controls on both surfaces (P<0.05). With respect to the surface\(^*\)vision interaction, a significant increase was observed on the foam surface compared with the firm surface for both groups (P<0.001). Postural sway was increased on the foam surface with eyes closed (P<0.01) (Figure 5-5).
Three-way interactions

There were also significant surface*vision*group (P=0.05) and group*vision*insole three-way interactions for medial-lateral sway (P<0.05).

Figure 5-6 illustrates the surface*vision*group interaction and shows that significant increases in medial-lateral sway were observed on the foam surface in all conditions (P<0.001). Both groups had similar postural sway on the firm surface with eyes open. However, ML postural sway was greater in the PD group than the control group with eyes closed or when standing on the foam surface (P<0.05). In the control group, postural sway was greater on both surfaces without vision (P<0.05). Conversely, in the PD group, the same levels of ML postural sway were observed with or without vision on both surfaces (P>0.05) (Figure 5-6).
Figure 5-6: Medial-lateral sway in the surface*vision*group three-way interaction (Black- Firm surface, Grey- Foam surface)

Post-hoc analysis of the group*vision*insole interaction demonstrated that the PD group always displayed more postural sway than the control group (P<0.05) as shown in Figure 5-7. Furthermore, the removal of visual input caused a significant increase in postural sway in the control group when standing barefoot and in the PD group when wearing smooth insoles (P<0.01). Relative to standing barefoot with eyes closed, the smooth insoles reduced postural sway in control group (P<0.05). For standing barefoot with eyes open, postural sway was decreased by the smooth insoles in PD group (P<0.05). During the more challenging task of standing with eyes closed, the textured insoles were effective in significantly decreasing medial-lateral postural sway relative to the barefoot and smooth insole conditions (P<0.01) (Figure 5-7).

Figure 5-7: Medial-lateral sway in group*vision*insole three-way interactions (Black-Control group, Grey- PD group)
Four-way interactions

Finally, as shown in Figure 5-8, there was also a significant group*surface*vision*insole interaction recorded for medial-lateral sway (P<0.05). On the firm surface with eyes open, the PD group had more ML postural sway than the control group while barefoot (P<0.05), but both the smooth and textured insoles reduced ML sway in the PD group to the same level as the controls (P>0.05). PD participants continually demonstrated significant greater postural sway than controls under all three insole conditions while standing on a firm surface with eyes closed and on the foam surface with eyes open (P<0.05). Standing on the foam surface caused significantly greater postural sway in all conditions, except for the controls standing with their eyes closed when wearing textured insoles. In the control group, visual deprivation affected ML postural sway only on the firm surface with textured insoles and on the foam surface while barefoot (P<0.05). With respect to the PD group, significant increases in ML postural sway due to visual deprivation, were observed on both surfaces with smooth insoles (P<0.05). For the most challenging condition, standing on the foam surface with eyes closed, the PD group showed greater ML postural sway than the control group when barefoot (P=0.051). However, relative to the barefoot and smooth insole conditions, the PD participants demonstrated significantly reduced ML sway while wearing the textured insoles (P<0.05). There were no differences between the barefoot and smooth insole conditions (P>0.05) (Figure 5-8).
Gait Assessment

Main Effect of Group
The PD group walked at a similar velocity to the control group but displayed a shorter stride length and an increased walking cadence ($P<0.01$). While there were no significant differences between the groups with respect to stride time, the PD group did have greater stride timing variability ($P<0.01$). There were no significant differences observed in the double support and stance phase of gait cycles ($P>0.05$) (Figure 5-9).
Main effect of Surface

When walking on the uneven surface, the velocity, cadence and stance phase were decreased significantly (P<0.001), while stride time (P<0.001) and stride timing variability (P<0.01) increased. No differences were observed in stride length or double support time between the two surfaces (Figure 5-10).

Main effect of Insole

Wearing smooth and textured insoles increased walking velocity relative to the barefoot condition (P<0.001). Both stride length and stride time were increased progressively from barefoot to smooth insoles to the textured insoles (P<0.05). Conversely, cadence was decreased from barefoot to smooth insoles to the textured insoles (P<0.001). Double support time decreased from smooth insoles to barefoot to the textured insoles (P<0.01) and while a similar trend was noted for the stance phase, the decrease was only statistically significant between the textured and smooth insoles (P<0.01). Stride timing variability, however, was higher when wearing the smooth
insoles compared with barefoot and the textured insoles (P<0.05), but no difference was observed between the barefoot and textured insole conditions (Figure 5-11).

Figure 5-10: Main effect of surface (Black- Firm surface, Grey- Uneven surface) in gait assessments

Figure 5-11: Main effect of insole (Black- Barefoot, Grey- Smooth insole, Cross-Textured insole) for the temporospatial gait assessments
Two-way interactions

Group*Surface Interaction

Figure 5-12 describes the Group*Surface interactions. While both groups walked at a similar velocity on the firm surface, walking on the uneven surface caused both groups to walk slower (P<0.05), but this decrease was greatest for the PD group (P<0.01). Cadence was greater for the PD group on the firm surface but both groups decreased cadence to similar levels on the uneven surface. Both groups had similar stride lengths on the firm surface (P>0.05), but when walking on the uneven surface, the PD group decreased their stride length compared to firm surface, while controls had longer stride than firm surface (P<0.001). Although the PD group demonstrated shorter stride periods on the firm surface, both groups increased stride period to a similar duration on the uneven surface (P<0.001). Stride timing variability was similar for both groups on the firm surface (P>0.05), but this was increased for the PD group on the uneven surface (P<0.001). While not statistically significant, double support time tended to be higher for the PD group on the firm surface, but was similar between the groups on the uneven surface (P=0.056) (Figure 5-12).

Figure 5-12: Group*Surface interaction for temporospatial gait assessments (Black-Control group, Grey- PD group)
**Group*Insole Interaction**

Figure 5-13 describes the Group*Insole interactions. While both groups walked at a similar velocity when barefoot, both the smooth and textured insoles increased walking speed in the two groups, but this increase was greatest for the control group. While cadence was higher and stride time was shorter for the PD group when walking barefoot (P<0.05), walking with either insole led to similar cadence and stride times between the groups. For the barefoot and smooth insole conditions, the PD group had longer stance times than controls (P<0.05). When wearing the textured insoles the stance time was significantly decreased for the PD group relative to controls (P<0.001) (Figure 5-13). In contrast, the control group increased the duration of the stance phase when using the textured insoles (P<0.001).

Figure 5-13: Group*Insole interaction for gait assessments (Black- Control group, Grey- PD group)
Insole*Surface Interactions

Figure 5-14 describes the Insole*Surface interactions. Participants walked slower on the uneven surface, relative to the firm surface, under all insole conditions. On the uneven surface the smooth insole resulted in an increased walking velocity (P<0.05), while the smooth and the textured insoles resulted in a faster walking velocity on the firm surface (P<0.01). On the firm surface, stride length increased progressively from the barefoot to the smooth insole and the textured insole conditions (P<0.01). With respect to the uneven surface, both insoles increased stride length relative to walking barefoot (P<0.001). For the barefoot condition, larger stride lengths were observed on the uneven surface compared to the firm surface, with the textured insoles resulting in a shorter stride length on the uneven surface (P<0.01). (Figure 5-14).

Figure 5-14: Insole*Surface interaction for three variables of temporospatial gait assessments (Black- Firm surface, Grey- Uneven surface)

Group*Insole*Surface Interaction

Significant Group*Insole*Surface interactions were obtained for stride timing variability, stance phase and double-support time.

Figure 5-15 describes the Group*Insole*Surface interaction for stride timing variability. On the firm surface the PD group had greater stride timing variability than the control group when barefoot (P<0.05), but decreased stride timing variability to a similar level to the control group when using the textured insoles. For the control
group, stride timing variability increased when using both insoles, relative to walking barefoot (P<0.05).

The PD group had greater stride timing variability than the control group for all footwear conditions when walking on the uneven surface (P<0.05). The smooth insole resulted in the PD group having increased stride timing variability relative to the barefoot and textured insole conditions (P<0.01). In contrast, the control group exhibited a slight, but significant, decrease in stride timing variability when wearing both insoles on the uneven surface (P<0.05) (Figure 5-15).

Figure 5-16 describes the Group*Insole*Surface interaction for double support time. On the firm surface, the PD group displayed longer double support time than the control group when barefoot and while wearing the smooth insoles. However, with the textured insoles, there was a reduction in double support time to an equivalent duration to the control group. On the uneven surface, double support time was slightly longer for the PD group relative to controls across all insole conditions. However, both groups exhibited a decrease in double support time with the textured insoles relative to the barefoot and smooth insole conditions (Figure 5-16).
Figure 5-16: Group*Insole*Surface three way interaction for double support time
(Black- Control group, Grey- PD group)

Figure 5-17 describes the Group*Insole*Surface interactions for stance period. Compared with controls, the PD group had a significantly longer stance period while walking barefoot on the uneven surface (P<0.01), while the PD group had shorter stance periods when wearing the textured insoles on both surfaces (P<0.01). For the control group, the textured insoles increased the stance phase relative to both the barefoot and smooth insoles (P<0.05) on both surfaces. On both the firm and uneven surfaces, the textured insoles significantly reduced stance time for the PD group relative to the barefoot and smooth insole conditions (P<0.001). Furthermore, walking on the uneven surface consistently lead to shorter stance phases than the firm surface (P<0.01) (Figure 5-17).

Figure 5-17: Group*Insole*Surface three way interaction for stance period (Black-
Control group, Grey- PD group)
5.4 Discussion

The aim of study 3 was to determine the effect of textured insoles on postural control during standing and walking in people with PD and age-matched controls. The results showed that both insoles caused different beneficial effects on postural control. The most significant benefits were noticed in the PD group with the textured insole, which changed performance of standing balance and gait under challenging conditions.

Standing Balance Assessments

Consistent with previous research (Adkin et al., 2005; Mitchell et al., 1995) and the results of study 2, the PD group displayed greater medial-lateral postural sway than the control group. Both insoles were effective in reducing ML postural sway in the PD group, but the efficacy of the insoles were dependent on the type of support surface and the availability of vision. Both the smooth and textured insoles reduced ML sway for the PD group to a level equivalent to that of the control group when standing on a firm surface with unconstrained visual feedback. Conversely, the control group demonstrated no changes in postural sway under any of the insole conditions. Given that both insoles contributed to performance improvements for the PD group, it could be argued that the benefits observed in this study on the firm surface may have been due to the wearing of shoes rather than to the insoles themselves. Nonetheless, these findings are important because previous research has identified increased postural
sway on a firm surface with eyes open to be a significant risk factor of falls for people with PD (Kerr et al., 2010).

Under more challenging conditions, such as standing on a foam surface with eyes closed, the textured insoles were effective in reducing ML sway for the PD group to a level that was equivalent to the control group. In accordance with the results for the firm surface, the textured insoles provided no significant benefit to the control group, with respect to ML sway. These results suggested that while standing on the foam surface without visual input, the hills and valleys of the textured insole surfaces may have been effective in applying strong pressure to the mechanoreceptors on the planar surface of the foot. This additional stimulation could have resulted in enhanced somatosensory feedback to the central nervous system and contributed to significantly improved postural control (Hidaka et al., 2000; Manjarrez et al., 2003).

In most experimental conditions, standing on the foam surface led to more ML postural sway than the firm surface, which was consistent with previous research (Patel et al., 2008; Vuillerme & Pinsault, 2007) and the results presented for studies 1 and 2. This increase in sway was likely to be related to somatosensory information being decreased on the foam surface (Patel et al., 2008; Vuillerme & Pinsault, 2007), which would have resulted in participants having difficulties detecting the ground clearly. Given that the results of the electrophysiological tests performed in Study 2 demonstrated that the PD participants also had significantly poorer somatosensory
function relative to controls, the effects of standing on the foam surface would likely have compounded poor performance. The textured insoles may be effective in ameliorating these deficits in somatosensory function for the PD participants and demonstrated the most pronounced effect under conditions where there was greater reliance on somatosensory information.

In contrast to the results presented in Study 1, the insoles produced no alterations in ML sway in the control group. A possible explanation for this could be that the insole design was altered slightly between Studies 1 and 3 to incorporate two raised compliant ridges around the lateral perimeter of the insole and around the heel of the foot to increase the area of stimulation. Finally, unlike Study 1 where participants were required to stand on textured insole surface, the data presented in Study 3 were derived from textured insoles fitted into shoes.

Overall, the textured insoles decreased postural sway and improve gait stability in the PD group due to the enhancement of somatosensory information from the feet. This finding was consistent with previous work by Hatton et al. (2011) who showed that standing on a textured surface could decrease medial-lateral sway during standing. However, it is important to note that this research study did not attempt to insert the textured surface into shoes and only studied the effects of the textured surface in a younger control group. Conversely, Corbin et al. (2007) and Palluel et al. (2008; 2009) both made textured surfaces into insoles and reported significant
reductions in postural sway during standing balance while wearing the textured insoles and sandals, respectively. However, Corbin et al. (2007) only studied the effects of their textured insoles in a younger cohort, while the sandals evaluated by Palluel et al. (2008; 2009) may not be suitable for all individuals and their use may be limited in some environments (e.g. the workplace). Furthermore, Palluel et al. (2008; 2009) did not randomize the order of testing conditions, which may have introduced a learning effect into the results.

Gait Assessments

The PD group had similar gait characteristics to those previously reported for walking on a firm surface (Knutsson, 1972; Rogers, 1996; Ueno et al., 1993). They had reduced stride length, increased cadence, shorter stride duration and increased stride timing variability relative to the control group. On the uneven surface the PD group had a longer stride period and decreased cadence and stance period. However, unlike the control group, they did not increase their stride length.

The insoles were effective in altering some aspects of gait. Both the smooth and textured insoles resulted in positive changes such as increased walking velocity. Both insoles also led to longer strides lengths and durations, and reduced cadence. Given that both insoles had a similar effect on these temporospatial characteristics, these results may suggest that the changes were due to wearing normal shoes rather than
due to the insoles themself. Despite this, the textured insoles were effective in reducing the time spent in the double support phase of the gait cycle without increasing stride timing variability, while the smooth insoles led to an increased stride timing variability. Given that increased stride timing variability has been shown to be associated with an increased risk of falling (Hausdorff et al., 1998). The results suggested that only the textured insole altered the gait pattern positively compared to the barefoot condition without causing deleterious effects. Both the smooth and textured insoles were effective in increasing stride length relative to the barefoot condition, but the textured insole was significantly better than the smooth insole. As stride length is an important variable in the regulation of gait (Lewis et al., 2000; Morris et al., 1996), the increased stride length observed in the two groups while wearing the textured insole may be associated with better postural stability. Participants may have increased their stride length due to the improved postural reflexes resulting from the foot being stimulated by the elevated textured surface of the insoles, which would transmit enhanced somatosensory information into central nervous system. In accordance with the changes that were evident in stride length, stride period was increased and cadence was decreased from barefoot to the smooth and textured insoles so that similar walking patterns were maintained. With respect to velocity, both insoles increased walking speed to the same level, indicating that participants can maintain similar walking patterns even at high speeds when wearing shoes.
Effects of insoles for two groups when walking on different surfaces

While the insoles seemed to provide beneficial effects for both groups, the effects of the insoles were dependent on the walking surface and were different for the control and the PD groups. Both the smooth and textured insoles increased stride time and walking velocity and decreased cadence in the control group, while the textured insoles also increased stance time. For the PD group, the smooth and textured insoles also resulted in similar increases in stride time and decreased cadence, but reduced stance time and did not alter walking velocity.

The effects of the insoles for each group were dependent on the walking surface as highlighted by measures of stride timing variability, stance and double support phases. Stride timing variability has been reported to be a measure of ability to accurately regulate variations in gait cycle timing (Hausdorff et al., 1998) and increases in this variable have been associated with falls (Cole et al., 2011). In the present study both insoles resulted in an increase in stride timing variability on the firm surface for the control group, but not on the uneven surface. The effects were markedly different for the PD group who demonstrated no change in stride timing variability on the firm surface for either insole, but significantly increased variability on the uneven surface for the smooth insole. These results indicated that the benefits of insoles were group dependent and may result in different effects on gait. The PD group had poorer somatosensory function than controls, so it would be reasonable to expect that the insoles would provide greater improvements in sensory feedback for this group.
Both insoles resulted in some positive effects on walking balance, such as increased velocity and stride length. However, participants wearing the smooth insole displayed increased stride timing variability. This may have been due to that the smooth insole, as a control condition equivalent to wearing a shoe only, did not provide enhanced somatosensory information to the feet, which may lead to less postural stability in gait.

The textured insole resulted in decreased stance and double support phases of gait for the PD group on both the firm and uneven surfaces, which may be indicative of a more confident gait pattern, as increases in double support time have been associated with greater instability and falls (Cole et al., 2011; Wood et al., 2009; Woollacott & Tang, 1997). In contrast, the control group had an increase in the stance phase on both surfaces while wearing the textured insoles, but demonstrated a decrease in the duration of the double support phase on the uneven surface. This finding indicated that the most beneficial effects of the textured insole were demonstrated on the uneven surface in the PD group. The participants took longer strides while on the uneven surface with the textured insoles, which would have limited the number of times that the foot contacted the more challenging surface. These results imply that the participants felt more stable when walking under these conditions, possibly due to enhanced somatosensory input received from the textured insoles.

Increased double support time was an adaption mechanism associated with greater instability and falls (Cole et al., 2011; Wood et al., 2009; Woollacott & Tang, 1997).
The increased double supported time was evident while wearing the smooth insoles compared with both the barefoot and textured insole conditions, which may have represented an attempt to minimize postural instability. Stride timing variability was increased only in the smooth insole condition, which may be due to the fact that the smooth insole was only a control of textured insole. The smooth insole condition was included to represent wearing flat-soled shoes with no additional enhancement and, hence, would not be expected to improve sensory feedback as the textured insoles did in people with deficits in somatosensory function. Therefore, compared to walking with the textured insole, the gait pattern was more irregular when walking with the smooth insoles, which may decrease postural stability during walking (Hausdorff et al., 1998).

Similar effects for the insoles were observed when walking on the different surfaces, with both insoles increasing stride length relative to the barefoot condition, but the increase in stride length was greater on the firm surface with the textured insole. The textured insole altered stride cadence and stride timing in the PD group, such that both groups demonstrated similar patterns for these variables. This finding provides support for a positive effect of the textured insoles on reducing the disparity in gait measures between people with PD and controls. The improved gait characteristics evident for the PD group while wearing the textured insoles may be the result of improved somatosensory feedback contributing to improved confidence and balance control.
When walking on the more challenging uneven surface, the textured insoles were more effective than the smooth insole at decreasing stride time variability for the PD group. These findings suggested that the PD group demonstrated a more stable walking pattern when wearing the textured insoles. Walking on an uneven surface reduces feedback for all participants, but the reduced somatosensory function which has been suggested by the electrodiagnosis tests in PD patients compounds the problem. In this situation, artificial somatosensory stimulation using textured insoles seems to be an effective intervention for compensating for these deficits in somatosensory function.

Previous research by Maki et al. (2008) presented findings that suggested that textured insoles were effective in improving postural stability by controlling lateral stability during walking in older people. In accordance with the findings presented by Maki et al. (2008), the results of the current study showed a similar tendency for a pair of inexpensive textured insoles to decease postural sway and improve gait control in people with PD. This overall result showed evidence that somatosensory information can be enhanced by textured insoles to improve postural stability in both standing and walking, especially in people with PD.

Study 3 indicated that postural stability can be improved in people with PD in both standing and walking conditions, possibly due to improved somatosensory information received from plantar mechanoreceptors. Textured insoles decreased ML
postural sway when standing on foam surface and increased walking velocity, stride length and stride time. Furthermore, textured insoles reduced cadence, double support time and stance phase without increasing stride timing variability when walking on the more challenging uneven surface.

**Summary Conclusion**

There were still some limitations in this project. Firstly, the changes in postural control observed in this project may have been somewhat transient, so it is possible that the apparent benefits of the insoles may only have been temporary. To evaluate the long-term benefits of these insoles, a longitudinal study would be required. Secondly, further work involving a larger sample size would be needed to confirm our findings and to test their applicability to different patient groups who have somatosensory loss.

This project demonstrated that the PD group had decrements in somatosensory function compared to the healthy older control group. People with PD presented with a reduced sensitivity in touch, vibration, temperature and nociception and had deficits in measures of nerve conduction and SEP. Furthermore, there was a close association between poor somatosensory function and decreased clinical balance (Tinetti Test) in the PD group, while reduced somatosensory function was also correlated with temporospatial gait measures, including shorter stride length. This research demonstrates that somatosensory information can be enhanced by textured insoles,
which would serve to improve postural stability during both standing and walking
tasks in populations with somatosensory loss, such as people with PD.
# Appendix A.1

## ADDENBROOKE’S COGNITIVE EXAMINATION - ACE-R

*Final Revised Version C (2005)*

<table>
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<th>Name</th>
<th>Date of testing:</th>
<th>Date of birth:</th>
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**Addressograph**

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<tr>
<th>Handedness:</th>
<th>Tester’s name:</th>
<th>Age at leaving full-time education:</th>
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## ORIENTATION

1. Ask: What is the Day, Date, Month, Year, Season?

2. Ask: Which Building, Floor, Town, County, Country?

## REGISTRATION

1. Tell: ‘I’m going to give you three words and I’d like you to repeat after me: lemon, key and ball. After subject repeats, say ‘Try to remember them because I’m going to ask you later’. Score only the first trial (repeat 3 times if necessary).

## ATTENTION & CONCENTRATION

1. Ask the subject: ‘Could you take 7 away from a 100? After the subject responds, ask him/her to take away another 7 to a total of 5 subtractions. If subject makes a mistake, carry on and check the subsequent answer (i.e. 93, 86, 79, 72, 65).

2. Stop after five subtractions (93, 86, 79, 72, 65).

3. Ask: ‘Could you please spell WORLD for me? Then ask him/her to spell it backwards:

## MEMORY - Recall

1. Ask: ‘Which 3 words did I ask you to repeat and remember?’

## MEMORY - Anterograde Memory

1. Tell: ‘I’m going to give you a name and address and I’d like you to repeat after me. We’ll be doing that 3 times, so you have a chance to learn it. I’ll be asking you later’

## MEMORY - Retrograde Memory

1. Name of current Prime Minister

2. Name of the last Prime Minister

3. Name of the leader of the Opposition

4. Name of the president of USA

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### Verbal Fluency - Letter 'P' and Animals

**Letters**

Say: "I'm going to give you a letter of the alphabet and I'd like you to generate as many words as you can beginning with that letter, but not names of people or places. Are you ready? You've got a minute and the letter is P."

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<td>12-15</td>
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<td>16-19</td>
<td></td>
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<tr>
<td>&gt;20</td>
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</table>

**Animals**

Say: "Now can you name as many animals as possible, beginning with any letter?"

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<thead>
<tr>
<th>Score</th>
<th>Words</th>
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<td>16-19</td>
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<td>&gt;20</td>
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### Language - Comprehension

Show written instruction:

**Close your eyes**

### Language - Writing

3 stage command:

'Take the paper in your right hand. Fold the paper in half. Put the paper on the floor.'

Ask the subject to make up a sentence and write it in the space below.

Score 1 if sentence contains a subject and a verb (see guide for examples).
### LANGUAGE - Repetition

- Ask the subject to repeat: 'hippopotamus'; 'eccentricity'; 'unintelligible'; 'statistician'
  - Score 2 if all correct; 1 if 3 correct; 0 if 2 or less.

- Ask the subject to repeat: 'Above, beyond and below'

- Ask the subject to repeat: 'No ifs, ands or buts'

### LANGUAGE - Naming

- Ask the subject to name the following pictures:

  1. Pencil
  2. Watch
  3. Kangaroo
  4. Anchor
  5. Camel
  6. Harp
  7. Barrel
  8. Crown
  9. Crocodile
  10. Accordion

### LANGUAGE - Comprehension

- Using the pictures above, ask the subject to:
  - Point to the one which is associated with the monarchy
  - Point to the one which is a marsupial
  - Point to the one which is found in the Antarctic
  - Point to the one which has a nautical connection

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**LANGUAGE - Reading**

- Ask the subject to read the following words: [Score 1 only if all correct]
  - sew
  - pint
  - soot
  - dough
  - height

**VISUOSPATIAL ABILITIES**

- Overlapping pentagons: Ask the subject to copy this diagram:

![Overlapping pentagons diagram](image)

- Wire cube: Ask the subject to copy this drawing (for scoring, see instructions guide)

![Wire cube diagram](image)

- Clock: Ask the subject to draw a clock face with numbers and the hands at ten past five. (for scoring see instruction guide: circle = 1, numbers = 2, hands = 2 if all correct)

![Clock diagram](image)
Ask the subject to count the dots without pointing them.

[Score 0-4]
ADDCRE: COGNITIVE EXAMINATION - ACE-R

PERCEPTUAL ABILITIES

➤ Ask the subject to identify the letters

[Score 0-4]

RECALL

➤ Ask "Now tell me what you remember of that name and address we were repeating at the beginning."

John Marshall
24 Market Street
Spilsby
Lincolnshire

[Score 0-7]

RECOGNITION

➤ This test should be done if subject failed to recall one or more items. If all items were recalled, skip the test and score 5. If only part is recalled start by ticking items recalled in the shadowed column on the right hand side. Then test not recalled items by telling "ok, I'll give you some hints: the name X, Y or Z" and so on. Each recognised item scores one point which is added to the point gained by recalling.

<table>
<thead>
<tr>
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<tbody>
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<td>Market Street</td>
<td>High Street</td>
<td>Market Square</td>
<td>recalled</td>
</tr>
<tr>
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<td>Horncastle</td>
<td>Sleaford</td>
<td>recalled</td>
</tr>
<tr>
<td>Northamptonshire</td>
<td>Lincolnshire</td>
<td>Leicestershire</td>
<td>recalled</td>
</tr>
</tbody>
</table>

General Scores

| MMSE | 30 |
| ACER | 100 |

Subscores

| Attention and Orientation | 18 |
| Memory | 26 |
| Fluency | 6 |
| Language | 26 |
| Visuospatial | 16 |

Normative values based on 63 controls aged 52-75 and 142 dementia patients aged 46-86
Cut-off <28 gives 98% sensitivity and 89% specificity for dementia
Cut-off <22 gives 84% sensitivity and 100% specificity for dementia

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## Appendix A.2

**Modified Falls Efficacy Scale**

How confident/sure are you that you can do each of the activities *without falling*?  
(Please circle one response for each statement)

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

The Activities-specific Balance Confidence (ABC) Scale*

Instructions to Participants:
For each of the following, please indicate your level of confidence in doing the activity without losing your balance or becoming unsteady from choosing one of the percentage points on the scale form 0% to 100%. If you do not currently do the activity in question, try and imagine how confident you would be if you had to do the activity. If you normally use a walking aid to do the activity or hold onto someone, rate your confidence as if you were using these supports. If you have any questions about answering any of these items, please ask the administrator.

The Activities-specific Balance Confidence (ABC) Scale*
For each of the following activities, please indicate your level of self-confidence by choosing a corresponding number from the following rating scale:

0% 10 20 30 40 50 60 70 80 90 100%
no confidence completely confident

“How confident are you that you will not lose your balance or become unsteady when you…
1. …walk around the house? ____%
2. …walk up or down stairs? ____%
3. …bend over and pick up a slipper from the front of a closet floor ____%
4. …reach for a small can off a shelf at eye level? ____%
5. …stand on your tiptoes and reach for something above your head? ____%
6. …stand on a chair and reach for something? ____%
7. …sweep the floor? ____%
8. …walk outside the house to a car parked in the driveway? ____%
9. …get into or out of a car? ____%
10. …walk across a parking lot to the mall? ____%
11. …walk up or down a ramp? ____%
12. …walk in a crowded mall where people rapidly walk past you? ____%
13. …are bumped into by people as you walk through the mall? ____%
14. …step onto or off an escalator while you are holding onto a railing? ____%
15. …step onto or off an escalator while holding onto parcels such that you cannot hold onto the railing? ____%
16. …walk outside on icy sidewalks? ____%

Appendix A.4

UNIFIED PARKINSON'S DISEASE RATING SCALE

MENTATION, BEHAVIOR & MOOD

1. Intellectual Impairment
   a. None
   b. Mild: Can manage household tasks and personal affairs
   c. Moderate: Has difficulty in managing daily activities
   d. Severe: Requires constant supervision

2. Memory
   a. Normal
   b. Mild: Can remember recent events
   c. Moderate: Can remember past events
   d. Severe: Cannot remember past events

3. Judgement
   a. None
   b. Mild: Occasionally makes errors in judgment
   c. Moderate: Frequently makes errors in judgment
   d. Severe:常 makes errors in judgment

4. Speech
   a. Normal
   b. Mild: Slow and slurred
   c. Moderate: Slurred and difficult to understand
   d. Severe: Incoherent

5. Writing
   a. Normal
   b. Mild: Can write legibly
   c. Moderate: Can write legibly but with difficulty
   d. Severe: Cannot write legibly

6. Walking
   a. Normal
   b. Mild: Slow and difficult
   c. Moderate: Requires assistance
   d. Severe: Cannot walk or stand

7. Patient's Behavior
   a. Normal
   b. Mild: Occasional outbursts
   c. Moderate: Frequent outbursts
   d. Severe: Constantly agitated

8. Patient's Mood
   a. Normal
   b. Mild: Slight depression
   c. Moderate: Moderate depression
   d. Severe: Severe depression

9. Emotions
   a. Normal
   b. Mild: Slight depression
   c. Moderate: Moderate depression
   d. Severe: Severe depression

10. Appetite
    a. Normal
    b. Mild: Slight decrease
    c. Moderate: Moderate decrease
    d. Severe: Severe decrease

11. Activities of Daily Living
    a. Normal
    b. Mild: Slight decrease
    c. Moderate: Moderate decrease
    d. Severe: Severe decrease

12. Dressing
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

13. Grooming
      a. Normal
      b. Mild: Slight decrease
      c. Moderate: Moderate decrease
      d. Severe: Severe decrease

14. Eating
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

15. Personal Hygiene
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

16. Grooming
      a. Normal
      b. Mild: Slight decrease
      c. Moderate: Moderate decrease
      d. Severe: Severe decrease

17. Activities of Daily Living
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

18. Motor Examination
    a. Normal
    b. Mild: Slight decrease
    c. Moderate: Moderate decrease
    d. Severe: Severe decrease

19. Sensation
    a. Normal
    b. Mild: Slight decrease
    c. Moderate: Moderate decrease
    d. Severe: Severe decrease

20. Reflexes
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

21. Spasticity
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

22. Rigidity
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

23. Tremor
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

24. Posture
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

25. Speech
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

26. Swallowing
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

27. Eating
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

28. Bowel Function
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

29. Bladder Function
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

30. Cognitive Functions
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

31. Intellectual Functions
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

32. Psychiatric Symptoms
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

33. Depression
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

34. Anxiety
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

35. Agitation
     a. Normal
     b. Mild: Slight increase
     c. Moderate: Moderate increase
     d. Severe: Severe increase

36. Sleep
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

37. Appetite
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

38. Weight
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease

39. Autonomic Functions
     a. Normal
     b. Mild: Slight decrease
     c. Moderate: Moderate decrease
     d. Severe: Severe decrease
23. Finger Taps (Patient taps thumb with index finger in rapid succession.)
   0 = Normal.
   1 = Mild slowing and/or reduction in amplitude.
   2 = Moderately impaired, definite and early slapping. May have occasional
      errors in movement.
   3 = Severely impaired. Frequent hesitation in initiating movements or error
      in ongoing movements.
   4 = Can hardly perform the task.

24. Hand Movements (Patient opens and closes hands in rapid succession.)
   0 = Normal.
   1 = Mild slowing and/or reduction in amplitude.
   2 = Moderately impaired, definite and early slapping. May have occasional
      errors in movement.
   3 = Severely impaired. Frequent hesitation in initiating movements or error
      in ongoing movements.
   4 = Can hardly perform the task.

25. Rapid Alternating Movement of Hands (Prosodion expansion movements of
    hands, vertically and horizontally, with as large an amplitude as possible, both
    hands simultaneously.)
   0 = Normal.
   1 = Mild slowing and/or reduction in amplitude.
   2 = Moderately impaired, definite and early slapping. May have occasional
      errors in movement.
   3 = Severely impaired. Frequent hesitation in initiating movements or error
      in ongoing movements.
   4 = Can hardly perform the task.

26. Leg Agility (Patient taps heel on ground in rapid succession picking up an order
    leg. Amplitude should be at least 3 inches.)
   0 = Normal.
   1 = Mild slowing and/or reduction in amplitude.
   2 = Moderately impaired, definite and early slapping. May have occasional
      errors in movement.
   3 = Severely impaired. Frequent hesitation in initiating movements or error
      in ongoing movements.
   4 = Can hardly perform the task.

27. Arising from Chair (Patient attempts to rise from a straightened chair, with no
    movement allowed.)
   0 = Normal.
   1 = Slow as may need more than one start.
   2 = Paced well up from area of rest.
   3 = Tends to fall back and may have to try more than once time, but are
      able to rise without help.
   4 = Unable to rise without help.

28. Posture
   0 = Normal erect.
   1 = Slight stoop, slightly slouched posture, would be normal for older
      persons.
   2 = Moderately stooped posture, definitely abnormal, can be slighty hunched
      in one side.
   3 = Severely stooped posture, with kyphosis, can be very difficult to leave
      seated.
   4 = Marked flexion with extreme abnormality of posture.

29. Gait
   0 = Normal.
   1 = Stilts, may shuffle with short steps, but no indication of stumbling
      or freezing.
   2 = Walks with difficulty, but requires little or no assistance may have some
      freezing, short steps, or propelling.
   3 = Severe disturbance of gait, requiring assistance.
   4 = Cannot walk at all, even with assistance.

30. Postural Instability (Response to sudden, strong posterior displacement
    produced by pulling on shoulders while patient erect with eyes open and
    feet slightly apart. Patient is propelled.)
   0 = Normal.
   1 = Recumbent, but not recommended.
   2 = Absence of postural response, would fall if not caught by examiner.
   3 = Very unsteady, tends to lose balance spontaneously.
   4 = Unable to stand without assistance.

31. Body Bradkinesia and Hypokinesia (Combining slowness, hesitancy,
    decreased armwing, small amplitude, and poverty of movement in
    general.)
   0 = None.
   1 = Minimal slowing, giving movement a deliberate character; could be
      normal for some persons. Possibly related amplitude.
   2 = Mild degree of slowness and poverty of movement which is distinctly
      abnormal. Althougthness, some reduced amplitude.
   3 = Moderate slowness, poverty or small amplitude of movement.
   4 = Marked slowness, poverty or small amplitude of movement.

COMPILATIONS OF THERAPY (in brackets)

32. Excretion: What proportion of the morning day is diuresis present?
    0 = None.
    1 = 25% of day.
    2 = 50% of day.
    3 = 75% of day.
    4 = 100% of day.

33. Function: How disabling are the dyskinesias?
    0 = Not disabling.
    1 = Mildly disabling.
    2 = Moderately disabling.
    3 = Severely disabling.
    4 = Completely disabling.

34. Painful Dyskinesias: How painful are the dyskinesias?
    0 = Not painful.
    1 = Numb.
    2 = Moderate.
    3 = Severe.
    4 = Completely.

35. Presence of Early Morning Dyskinesia
    0 = No.
    1 = Yes.

36. Are 'off' periods predictable?
    0 = No.
    1 = Yes.

37. Are 'off' periods unpredictable?
    0 = No.
    1 = Yes.

38. Do 'off' periods come on suddenly, within a few seconds?
    0 = No.
    1 = Yes.

39. What proportion of the waking day is the patient 'off' in average?
    0 = None.
    1 = 25% of day.
    2 = 50% of day.
    3 = 75% of day.
    4 = 100% of day.

40. Olfactory Complications
    0 = Yes.

41. Are any sleep disturbances, such as insomnia or hyperventilation?
    0 = Yes.
    1 = No.

42. Does the patient have asymmetric, narrow or timing?
    0 = Yes.
    1 = No.

MODIFIED HOHNS AND YAMB STAGING
Stage 0 = No signs of clinical
Stage 1 = Ulcerative disease.
Stage 1 = Ulcerative plus at least involvement.
Stage 2 = Advanced disease with or without balance.
Stage 3 = Mild to moderate ulcerative disease, some minimal immobility,
    physically independent.
Stage 4 = Severe immobility; unable to walk or stand unassisted.
Stage 5 = Wheelchair bound or bedfast unless served.

SCHWARTZ AND ENGL AND ACTIVITIES OF DAILY LIVING SCALE
100% = Completely independent. Able to do all chores without assistance,
    difficulty or impairment. Essentially normal. No evidence of disability.
    90% = Completely independent. Able to do all chores with some degree
    of assistance, difficulty, and intermittently. Might take twice as long.
    Beginning to be aware of difficulty.
    80% = Completely independent in most chores. Takes twice as long.
    Some degree of difficulty and impairment. Essentially normal.
    70% = Completely independent. Moderate difficulty with some chores.
    Three in later stage as long as some. Must spend a large part of the day
    with chores. Some degree of difficulty and intermittent. Difficulty.
    60% = More dependent. Help with half, chores etc. Difficulty with
    everything.
    50% = Very dependent. Can't do much with chores, but can do
    /the chores. Some degree of difficulty and intermittently.
    40% = Very dependent. Can't do much with chores, but can do
    a few chores. Some degree of difficulty.
    30% = Nothing alone. Can do a slight help with some chores. Some
    degree of disability.
    20% = Family dependent, help. Complete invalid.
    0% = Family dependent, help. Complete invalid.
Appendix A.5

PD Gait and Falls Questionnaire

PLEASE TICK THE BOX WHICH MOST CLOSELY APPLIES TO YOU

1. During your **best** state – do you walk:
   - [ ] Normally
   - [ ] Almost normally – somewhat slow
   - [ ] Slow but fully independent
   - [ ] Need assistance or walking aid
   - [ ] Unable to walk

2. During your **worst** state – do you walk:
   - [ ] Normally
   - [ ] Almost normally – somewhat slow
   - [ ] Slow but fully independent
3. Are your gait difficulties affecting your daily activities and independence?
   - Not at all
   - Mildly
   - Moderately
   - Severely
   - Unable to walk

4. Do you feel that your feet get glued to the floor while walking, making a turn or when trying to initiate walking (freezing)?
   - Never
   - Very rarely - about once a month
   - Rarely – about once a week
   - Often – about once a day
   - Always – whenever walking

5. How long is your longest freezing episode?
   - Never happened

☐ Need assistance or walking aid
☐ Unable to walk
1. 1-2 seconds
2. 3-10 seconds
3. 11-30 seconds
4. Unable to walk for more than 30 seconds

6. How long is your typical start hesitation episode (freezing when initiating the first step)?
   - None
   - Takes longer than 1 second to start walking
   - Takes longer than 3 seconds to start walking
   - Takes longer than 10 seconds to start walking
   - Takes longer than 30 seconds to start walking

7. How long is your typical turning hesitation (freezing when turning)?
   - None
   - Resume turning in 1-2 seconds
   - Resume turning in 3-10 seconds
   - Resume turning in 11-30 seconds
   - Unable to resume turning for more than 30 seconds
8. How long is your typical destination hesitation (freezing when approaching the target, such as when stepping onto a scale or approaching a chair to sit down)?
   - None
   - Resume walking in 1-2 seconds
   - Resume walking in 3-10 seconds
   - Resume walking in 11-30 seconds
   - Unable to resume walking for more than 30 seconds

9. How long is your typical tight quarters hesitation (freezing when attempting to get through narrow space such as a doorway)?
   - None
   - Resume walking in 1-2 seconds
   - Resume walking in 3-10 seconds
   - Resume walking in 11-30 seconds
   - Unable to resume walking for more than 30 seconds

10. How long is your typical freezing episode while walking straight?
    - None
    - Resume walking in 1-2 seconds
    - Resume walking in 3-10 seconds
11. How long is your typical freezing episode during stressful time-demanding situations, such as when the telephone rings, at elevators or street crossings?

- None
- Resume walking in 1-2 seconds
- Resume walking in 3-10 seconds
- Resume walking in 11-30 seconds
- Unable to resume walking for more than 30 seconds

12. How often do you fall?

- Never
- Very rarely - about once a year
- Rarely – about once a month
- Often – about once a week
- Very often – once a day or more

13. How often do you fall when standing?

- Never
14. How often do you fall because of freezing episodes?
   - Never
   - It has happened once or twice
   - It has happened 3-12 times in the last 6 months
   - More than once a week
   - Whenever trying to walk unassisted

15. Do you experience festinating gait (Festinating gait is accelerated, short steps or gait)
   - Never
   - Very rarely - about once a month
   - Rarely – about once a week
   - Often – about once a day
   - Whenever walking

16. How often do you fall because of festinating gait?
Never
☐ It has happened once or twice
☐ It has happened 3-12 times in the last 6 months
☐ More than once a week
☐ Whenever trying to walk unassisted
# Appendix A.6

Parameter setup for the nerve conduction test

<table>
<thead>
<tr>
<th>Nerve</th>
<th>Amplifier Sensitivity</th>
<th>High/Low-pass Filter Cut-off Frequencies</th>
<th>Analysis Time</th>
<th>Stimulation Rate</th>
<th>Intensity of Stimulation</th>
<th>Duration of Stimulations</th>
<th>Amplifier Sensitivity of F-wave</th>
<th>F-wave High/Low-pass Filter Cut-off Frequencies</th>
<th>F-wave Duration of Stimulations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peroneal</td>
<td>5mv</td>
<td>5kHz/10Hz</td>
<td>2ms/div</td>
<td>1Hz</td>
<td>1Hz</td>
<td>0.2ms</td>
<td>5mv</td>
<td>5kHz/20Hz</td>
<td>0.2ms</td>
</tr>
<tr>
<td>Tibial</td>
<td>5mv</td>
<td>5kHz/10Hz</td>
<td>2ms/div</td>
<td>1Hz</td>
<td>1Hz</td>
<td>0.2ms</td>
<td>5mv</td>
<td>5kHz/20Hz</td>
<td>0.2ms</td>
</tr>
<tr>
<td>Sural</td>
<td>20uv</td>
<td>2kHz/20Hz</td>
<td>2ms/div</td>
<td>1Hz</td>
<td>1Hz</td>
<td>0.2ms</td>
<td>20µV</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
## Appendix A.7

### Vicon Nexus Full-Body Plug-In Gait Marker Set

<table>
<thead>
<tr>
<th>Body Segment</th>
<th>Anatomic Landmark</th>
<th>Bilateral</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Head</strong></td>
<td>Front Head - Temple</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>Back Head - positioned to define the transverse plane of the head, together with the frontal markers</td>
<td></td>
</tr>
<tr>
<td><strong>Trunk</strong></td>
<td>Spinous process of the 7&lt;sup&gt;th&lt;/sup&gt; cervical vertebra</td>
<td>No</td>
</tr>
<tr>
<td></td>
<td>Spinous process of the 10&lt;sup&gt;th&lt;/sup&gt; thoracic vertebra</td>
<td>No</td>
</tr>
<tr>
<td></td>
<td>Jugular notch</td>
<td>No</td>
</tr>
<tr>
<td></td>
<td>Xiphoid process of the sternum</td>
<td>No</td>
</tr>
<tr>
<td></td>
<td>Spine of the right scapula</td>
<td>No</td>
</tr>
<tr>
<td><strong>Upper arm</strong></td>
<td>Acromio-clavicular joint</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Forearm</strong></td>
<td>Olecranon process of the humerus</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>Radial and ulnar styloid processes</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Hand</strong></td>
<td>Head of the 3&lt;sup&gt;rd&lt;/sup&gt; metacarpal</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Pelvis</strong></td>
<td>Anterior superior iliac spine (ASIS)</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>Sacrum - mid-way between the posterior superior iliac spines (PSIS) and positioned to lie in the plane formed by the ASIS and PSIS points</td>
<td>No</td>
</tr>
<tr>
<td><strong>Thigh</strong></td>
<td>Over the lower lateral 1/3 surface of the thigh in line with the hip and knee joint centres</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td>Flexion/extension axis of the knee; lateral condyle of the femur.</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Shank</strong></td>
<td>Over the lower 1/3 surface of the shank in line with the knee and ankle joint centres</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Ankle</strong></td>
<td>Lateral malleolus along an imaginary line that passes through the transmalleolar axis.</td>
<td>Yes</td>
</tr>
<tr>
<td><strong>Foot</strong></td>
<td>2&lt;sup&gt;nd&lt;/sup&gt; metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot</td>
<td>Yes</td>
</tr>
<tr>
<td></td>
<td>Calcaneus at the same height above the plantar surface of the foot as the toe marker</td>
<td>Yes</td>
</tr>
</tbody>
</table>
## Appendix A.8

<table>
<thead>
<tr>
<th>Name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body mass (kg)</td>
<td>Participant mass</td>
</tr>
<tr>
<td>Height (mm)</td>
<td>Participant height</td>
</tr>
<tr>
<td>ASIS breadth (mm)</td>
<td>Inter-ASIS distance, which is the distance between the left ASIS and right ASIS.</td>
</tr>
<tr>
<td>Leg length (mm) bilateral</td>
<td>Full leg length, measured between the ASIS and the medial malleolus, via the knee joint. Measured while the participant was standing.</td>
</tr>
<tr>
<td>Knee width (mm) bilateral</td>
<td>The mediolateral width of the knee across the line of the knee axis, which is the distance between medial and lateral condyles of the femur. Measured with the participant standing.</td>
</tr>
<tr>
<td>Ankle width (mm) bilateral</td>
<td>The mediolateral distance across the malleoli. Measured with participant was standing.</td>
</tr>
<tr>
<td>Shoulder offset (mm) bilateral</td>
<td>Vertical offset from the base of the acromion to shoulder joint centre</td>
</tr>
<tr>
<td>Elbow width (mm) bilateral</td>
<td>Width of elbow along the flexion axis, which is between the medial and lateral epicondyles of the humerus</td>
</tr>
<tr>
<td>Wrist width (mm) bilateral</td>
<td>Anterior-posterior thickness of wrist at position where the wrist marker were positioned</td>
</tr>
<tr>
<td>Hand thickness (mm) bilateral</td>
<td>Anterior-posterior thickness between the dorsum and palmar surfaces of the hand.</td>
</tr>
</tbody>
</table>
References


