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Effects of household footwear-surface interactions on the gait of older arthritic females

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EFFECTS OF HOUSEHOLD FOOTWEAR-SURFACE INTERACTIONS ON THE GAIT OF OLDER ARTHRITIC FEMALES

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Declaration

This thesis is submitted in fulfilment of the requirements for the award of Doctor of Philosophy in the Department of Biomedical Science at the University of Wollongong. The information presented in this thesis is my original work unless otherwise referenced or acknowledged. I hereby declare that this document has not been and will not be submitted to qualify for another degree at this or any other academic institution. In accordance with the National Health and Medical Research Council Statement on Human Experimentation[1] and the National Statement on Ethical Conduct in Research Involving Humans[2], the original data are held by the Department of Biomedical Science at the University of Wollongong.

Bridget Jean Munro

13 March 2005

Dedication

To Mum and Dad

Thank you for your support throughout this thesis and my academic career. You have been there through both the good and bad times and this thesis would not be what it is today without your love, support, and acknowledgement of my hard work. Thank you for giving me the chance to experience what I've wanted, as well as some of what I haven't wanted. However, most of all, thank you for giving me the opportunity to live my life as I wished and to excel in areas I didn't think possible.

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Publications

The following publications and presentations have been produced as a result of the research conducted for this thesis.

Publications in Refereed Journals

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Munro BJ & Steele JR. Footcare awareness: A survey of persons aged 65 years and above. *Journal of the American Podiatric Medical Association* 1998, **88**(5): 242-248.

Conference Proceedings and Presentations

Munro BJ, Steele JR & Gilleard WL. Sloppy slippers vs slippery surfaces: How do older arthritic women prepare for initial foot ground contact? *Australian Falls Prevention Conference*, Manly, Australia, November 21-23, 2004.

Munro BJ, Steele JR & Gilleard WL. Sloppy slippers vs slippery surfaces: How do older arthritic women prepare for initial foot ground contact? *Showcase of Excellence - University of Wollongong Research Students*, University of Wollongong, Australia, October 9, 2003 **(1st Prize - Best Poster)**.

Munro BJ, Steele JR & Gilleard WL. Do older women with RA display slip avoidance behaviour when walking in different types of household footwear on typical household surfaces? In: Milburn P, Wilson B & Yanai T. (Eds.) *Proceedings of the International Society of Biomechanics XIXth Congress*, Dunedin, New Zealand, July 6-11, 2003: 281.

Munro BJ, Steele JR & Gilleard WL. Slippers, surfaces and the gait of older arthritic females: The final episode? In: Milburn PD. (Ed.) *Proceedings of the Sixth Symposium on Footwear Biomechanics*, Queenstown, New Zealand, July 3-5, 2003: 68-69 **(Young Investigators Prize - Winner)**.

Munro BJ, Steele JR & Bashford GM. Effects of rheumatoid arthritis on foot function: Implications for slipper design. **(Invited Presentation)**. In: Bach TM, Orr D, Baker R & Sparrow WA. (Eds.) *Proceedings of the Fourth Australasian Biomechanics Conference*, Campus Graphics: LaTrobe University, Australia, November 28-30, 2002: 40-41.

Munro BJ, Steele JR & Gilleard WL. Preparing for foot-ground contact during gait: What are the effects of slippers and arthritis? In: Bach TM, Orr D, Baker R & Sparrow WA. (Eds.) *Proceedings of the Fourth Australasian Biomechanics Conference (ABC4)*, Campus Graphics: LaTrobe University, Australia, November 28-30, 2002: 16-17.

Munro BJ, Steele JR & Bashford GM. Effects of rheumatoid arthritis on foot function: Implications for slipper design. In: Nigg BM & Nurse MA. (Eds.) *Proceedings of the VIII emed Scientific Meeting*, Kananaskis, Canada, July 31-August 3, 2002: 60.

Munro BJ. Slippers and slippery floors: What can be done to prevent falls? In: Barrett R, Simeoni R & D'Helon C. (Eds.) *Proceedings of the Third Australasian Biomechanics Conference (ABC3)*, Griffith University, Australia, January 30-February 1, 2000.

Munro BJ & Steele JR. Sippers, surfaces and the gait of older arthritic females: Where are we now? In: Hennig EM & Stefanyshyn DJ. (Eds.) *Proceedings of the Fourth Symposium on Footwear Biomechanics*, International Society of Biomechanics Working Group on Functional Footwear, Canmore, Canada, August 5-7, 1999, 72.

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Abstract

Inappropriate footwear, poor footwear-surface interactions and gait adaptations resulting from musculoskeletal problems, such as arthritis and foot problems, are all factors that contribute to home slips. However, no research was located which identified the effects of interactions between household footwear and household surface characteristics on variables that may predispose older individuals to fall in the home. Therefore, the purpose of this thesis was to identify how different household shoe-surface interactions affected the gait of older people, particularly those with foot problems, to recommend a "safe" household shoe for older people. To achieve this purpose, this thesis was completed in two experimental sections.

Experimental Section A comprised surveying 60 men and 68 women aged 65 years and above who lived independently in the community to identify the requirements and attitudes of older people living independently in the community in relation to their household footwear wearing and purchasing habits. Statistical analyses were conducted using chi-square tests, independent *t*-tests and z-scores to determine whether any relationships existed between the variables as well as the effects of gender. The main health condition reported by both men (37%) and women (50%) was arthritis. Women reported significantly more foot problems (2.3 per woman) than men (1.4 per man) as well as significantly greater foot pain and/or discomfort (59% women, 45% men). Furthermore, a greater number of women (25%) had fallen in the 12 months before the survey compared to men (17%). Shoes were worn in and around the house by 79% respondents, with the most popular household shoe type being the slipper, with 56% respondents indicating they wore both closed back and toe slippers around the home. The type of household shoe worn was significantly related to gender such that women predominantly wore closure-free, non-rigid household shoe types compared to men who tended to wear shoes with closures around the home. Only 30% of respondents did not wear household shoes, with most walking barefoot around the home (89.5%). The most slippery surface reported was that of smooth tiles (40.6%) with several respondents indicating they felt uneasy on wet surfaces, particularly tiles. It was concluded that Experimental Section B should focus upon the walking patterns displayed by older rheumatoid arthritic (RA) females when wearing toe and closed back slippers and walking on typical household surfaces, particularly wet surfaces.

Experimental Section B examined the effects of household footwear-surface interactions on the biomechanical parameters characterising initial foot-ground contact, as this phase of the gait cycle most commonly results in slips. Subjective perceptions, kinematic, kinetic and neuromuscular data were collected at initial foot-ground contact as eight community-dwelling older women with RA and eight matched controls walked at a self-selected

pace along a 6 m walkway under three footwear conditions (barefoot, closed back slippers, toe slippers) and three surface conditions (carpet, dry vinyl tile, wet vinyl tile). Mixed repeated measures three-way ANOVA were then completed to determine whether subject group, shoe type or surface type significantly ($p \leq 0.05$) influenced the gait patterns at initial foot-ground contact.

Compared to the control subjects, RA subjects displayed similar activity levels, segmental proportionality and plantar sensation, although increased foot pain and knee flexibility, decreased knee and ankle muscle strength and altered static and dynamic plantar pressure patterns. Despite displaying similar gait characteristics, the RA subjects estimated the walking trials to be significantly more difficult and experienced significantly more pain compared to the control subjects. The within-footwear main effects revealed that when subjects walked in toe slippers they found them to be uncomfortable and slippery, requiring significantly altered muscle activation patterns (earlier vastus lateralis (VL), tibialis anterior (TA) and peroneus longus (PL) onset; earlier gastrocnemius (G) offset; longer PL duration; increased TA intensity) compared to walking in closed back slippers. Within-footwear changes to muscle activation strategies (earlier rectus femoris (RF), VL and PL onset; earlier G offset; longer biceps femoris duration; decreased TA intensity) and kinetic profiles (decreased peak vertical and anteroposterior braking forces) were also evident when subjects walked barefoot compared to walking shod. Furthermore, the within-surface main effects indicated that when subjects walked on the wet vinyl-tile surface, they perceived the surface as more slippery, more difficult to walk upon and less comfortable, requiring significantly altered muscle activation patterns (later RF offset; earlier TA offset; longer RF duration; shorter VL duration; increased RF and semitendinosus (S) intensity), kinetic profiles (decreased peak anteroposterior braking forces) and kinematic profiles (decreased foot/shoe angle and angular velocity). Interestingly, footwear x surface interactions indicated that the subjects displayed significantly altered movement control strategies (longer RF and G duration; increased RF, VL and S intensity; decreased anteroposterior braking forces and foot/shoe angular velocity) and subjective perceptions (increased task difficulty and shoe/surface slipperiness; decreased shoe comfort) when subjects walked barefoot on the wet vinyl tile surface compared to all other conditions.

It was concluded that older women, particularly those with RA should wear closed back slippers, in preference to toe slippers, around the home, to reduce their incidence of foot pain and/or discomfort as well as to reduce their slip risk at initial foot-ground contact, particularly when walking on slippery surfaces. Further research is recommended to determine whether the prescription of closed back slippers, considered to be “safe” in the present thesis, does in fact reduce falls, particularly slips, in the homes of older women.

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Chapter 1

The Problem

1.1 Introduction

Falls and their resultant injuries are documented as the leading cause of unintentional injury, disability, hospitalisation and death in the world for older people[3]. In Australia in 1998, 50,000 people over 65 years of age were hospitalised for injuries received in a fall[4]. Furthermore, it has been estimated that 89 per 100,000 people aged 75 years and over will die annually in Australia as a direct result of falls[5]. Falls were also responsible for 40% of admissions to nursing homes in people aged 60 years and over[3,6], contributing to escalating health care costs estimated to be approximately $2.5 billion per annum in Australia[7]. Consequently, falls will place an ever-increasing strain on the community's financial resources and health care system[8-10]. Therefore, strategies to reduce the incidence of falls in older people are urgently required.

As a person moves through the ageing process, and are faced with declining sensory and motor performance, disease and mobility loss (see Section 2.3.1), they tend to spend more time in their homes, thereby increasing the likelihood of home falls. It has been estimated that over half of the falls experienced by older people occur within their own homes or immediate home surroundings (see Section 2.2.2). Therefore, the home environment becomes increasingly important to older adults such that it is prudent to focus attention on the environmental causes of instability within the home[11,12].

Although multifactorial in causation, falls within the home predominantly result from slips and trips which result in losses of balance when people are engaged in their usual daily activities such as walking[3,13-15]. Tripping accidents or stumbles are defined as a sudden loss of footing[16] whereby inadequate toe clearance during the swing phase of gait causes an individual's footwear/foot to catch obstacles on the supporting surface[17]. Conversely, slipping accidents occur when there is a loss of traction between an individual's footwear/foot and the surface beneath their feet[18] and, although slips can occur at any time during the gait cycle, an estimated 90% of slips leading to falls occur during level walking at initial foot-ground contact (see Section 2.3.2). Interestingly,

slips have been associated with an increased risk of falls-related injury compared to tripping accidents[15] and therefore warrant urgent investigation so that effective strategies to prevent falls in the home that occur from slipping can be developed. However, before falls in the home can be reduced in older people, an understanding of the factors predisposing older people to falls, particularly slip-related falls, is essential.

Of the many risk factors implicated in home falls (see Section 2.3), inappropriate household footwear, particularly when worn on common household surfaces, has been cited as a major contributory environmental factor in home slips[19-22]. In a study of 107 older people (mean age, 77 years) who presented to hospital with a fractured neck of femur as a result of a fall, over half were wearing slippers or footwear characterised as unsafe at the time of the fall[23] (see Section 2.3.2(A)). Therefore, a safe household shoe needs to be designed for older people to reduce the risk of falls from slips in the home.

For a safe household shoe to be the shoe of choice for older people it must meet the demands of older people, accommodating common foot pathologies which they typically develop, such as bunions, hammer toes and arthritis[21]. The shoe must also interact appropriately with a variety of common household surfaces, such as carpet, tiles and floorboards[19,24-26], to reduce the risk of slipping (see Section 2.3.2(C)). Despite the complex interplay between footwear-surface interactions, walking and balance, few researchers have focused on these relationships when developing recommendations for selecting safe shoes[19,24,26], instead tending to discuss these items in isolation. However, it is information pertaining to how the three-component system, namely that of the person, the shoe and the supporting surface, interact that is vital in order to design safe household shoes for older people which can assist in reducing slips in the home.

1.2 Statement of the Problem

The purpose of this thesis was to identify how different household shoe-surface interactions affected the gait of older people, particularly those with foot problems, in order to recommend a “safe” household shoe older people. To achieve this purpose, this thesis was completed in two parts. Experimental Section A was designed to characterise the requirements and attitudes of older people living independently in the community in relation to their household footwear wearing and purchasing habits. It

also identified characteristics of older people who suffer falls, foot problems and require specialised footwear, in order to provide the subject base for the second part of this thesis, as well as the specific household footwear and surface types to be examined. Experimental Section B then examined the effects of household footwear-surface interactions on the biomechanical parameters characterising initial foot-ground contact in older women with rheumatoid arthritis[*] in order to identify which gait modifications made by older women may reduce the likelihood of slips and to provide information about which household slipper would be safe for older people.

Hypotheses, limitations and delimitations specific to Experimental Section A and Experimental Section B are discussed in Section 3.1.2 and Section 4.3, respectively.

1.3 Significance of the Study

Over half of all falls occur within the home or surrounding home environment when people are involved in their normal daily activities, frequently when wearing slippers or shoes characterised as unsafe. Even if no physical injury results from a fall, falls may lead to a loss of confidence and self- or other-imposed restrictions on independence, social functioning and daily activities[27-38]. That is, after experiencing a fall, an older person is less likely to go outside and may be less active within the home due to anxiety and a fear of repeated falling[3,37]. The psychological impact of falling may also contribute to morbidity, institutionalisation and death[15,39]. Providing a safe household shoe that can better interact with common household surfaces may reduce the falls risk in this population of older people, breaking the cycle of falls, post-fall syndrome and repeat falls, in turn, improving quality of life, promoting independence and reducing the financial cost of falls to the community. However, before safe household shoes can be designed and used as a strategy to prevent falls in older people, it is essential to understand how older people adapt their walking patterns when wearing different household footwear while walking on common household surfaces.

[*] Older women with rheumatoid arthritis comprised the sample for Experimental Section B based on the conclusions from Experimental Section A (see Section 3.4).

Chapter 2

Review of Related Literature

2.1 Introduction

A review of the literature pertaining to falls, footwear and surfaces in older people reveals countless articles for one to surmise safe household footwear design possibilities to reduce slips in older people. However, before effective guides for safe household shoes can be developed it is necessary to gain a general understanding of the risk factors, circumstances and consequences of falls in the community-dwelling older population; parameters affecting footwear design for older people; initiation of stance; and rheumatoid arthritis, a musculoskeletal disease that commonly affects older people and the footwear choices that they make. Therefore, before focusing on the experimental sections contained within this thesis, literature related to the following sections were reviewed and presented in this chapter:

(1) Falls

(2) Falls Risk Factors

(3) Footwear Needs of Older People

(4) The Control of Initial Foot-Ground Contact

(5) Rheumatoid Arthritis

2.2 Falls

A fall is characterised by an accidental loss of balance where the individual ends up on the floor or ground[13,40,41]. Accidental falls occur in persons of all ages[42]. However, both the incidence of adult falls and the severity of complications associated with falls rise steadily after middle age[11], such that for people aged 60 years and above, falls present a significant health problem[3,11,20,27,42-53].

2.2.1 Consequences of Falls

Internationally, approximately one third of people aged 65 years and above living in the community fall at least once a year with many individuals suffering multiple falls[3,13,44,52,54]. For example, Nevitt *et al.*[15] found that approximately half of 325 community-dwelling fallers aged 60 years and above experienced more than one fall over 12 months. However, as most studies have relied upon self-reporting techniques to determine falls incidence, techniques which are known to be problematic, falls may actually be under-reported in these studies and the problem may be broader than anticipated[11,55].

Irrespective of the actual incidence, falls in older people are documented as the leading cause of unintentional injury, disability, hospitalisation and death in the world[3,11,27,32,48,56,57], with multiple falls associated with an increased risk of death[29,58]. In the local Illawarra region (New South Wales, Australia), falls account for 37% of all injury deaths, 40% of all injury admissions to hospital, and contribute to 40% of admissions to nursing homes in people aged 60 years and above[6]. The total direct health care costs resulting from accidental falls for those aged 60 years and over in Australia in 1989 was estimated at $468 million, with $180 million incurred by the hospital sector alone[9]. In 1995/96, this cost had grown to exceed $688 million for persons aged 65 years and above[59]. As 24% of Australia's population is projected to be aged 65 years and over by the year 2051[60], falls will place an ever-increasing strain on the community's financial resources and health care system[8-10,20,61,62]. Therefore, strategies focused upon reducing the incidence of falls in the elderly are urgently required. However, before effective falls prevention programs can be developed it is imperative to understand factors contributing to the occurrence of falls.

2.2.2 Where Falls Occur

It has been estimated that over half of the falls experienced by older people occur within their own homes or immediate home surroundings[11,42,45,63-68]. For example, Luukinen *et al.*[69] reported on all the falls that occurred over 7 years in 808 semi-rural community-dwelling older people aged 70 years and

above, 62% who were female. The authors found that 1,722 falls (62%) occurred inside the home compared to 1,040 falls (38%) occurring outside the home. In addition, approximately 60% of fatal falls in adults aged 65 years and older are reported to occur inside the home[70,71]. In contrast, Blake *et al.*[44] reported that 62% of 1,042 older respondents, aged 65 to 85 years, reported falling outdoors compared to only 38% who reportedly fell indoors. However, when examining the statistics for those aged 85 years and over, 70% reported falling inside the home compared to only 30% who fell outside the home[44].

Older individuals may have an increased incidence of chronic medical conditions (see Section 2.3.1(D)), greater limitations on their mobility and an increased dependence on others (see Section 2.3.1(G)). As a result, these individuals may spend more time inside the home compared to young-old individuals[22,66,72,73]. Consequently, these older adults may fall inside the home more frequently than their younger counterparts.

Falls within the home tend to occur on level surfaces, particularly in rooms often frequented by residents, such as the lounge room, kitchen, bedroom and bathroom[47,61,67,68]. Campbell[45] found that 21% of falls reported by older community-dwelling adults (aged 70 years and above) occurred in the bedroom, 27% in the lounge and dining area and 19% in the kitchen. As the author did not report where the remaining one third of the reported falls occurred, it is postulated that some of these unaccounted falls may have occurred in the bathroom. A similar study conducted by the Injury Control Council of Western Australia[74] found that 1,091 community-dwelling people, aged 70 years and above, experienced 286 falls inside the home, most often in the bedroom (23%), lounge or family room (23%), followed by the bathroom (13%) and kitchen (12%). Although lower in number, it is the falls that occur on hard floor surfaces, such as those typically found in the kitchen and bathroom, that are of particular concern as they have been associated with an increased risk of falls-related injury[15]. Furthermore, most falls reportedly occur during periods of maximum activity in the morning or afternoon, particularly when older people are performing daily tasks of toileting, showering, dressing and cooking[3,45,63,67,68]. These tasks commonly involve walking and changing

position and are habitually performed in the bathroom and kitchen[3,45,63,67,68]. Therefore, it is important to understand the mechanisms that contribute to falls in these rooms that typically have hard, non-compliant floor surfaces (see Section 2.3.2(B)).

2.2.3 Causes of Falls

Regardless of the location, falls predominantly result from trips, slips and losses of balance[44,45,52,65,75-79]. Lloyd & Stevenson[80] reported that slips and trips caused 67% of falls sustained by older people and Berg *et al.*[68] reported that 59% of 50 fallers (mean age, 72 years) sustained their falls because of slips and trips. Furthermore, Tideiksaar[81] reported that of 102 falls experienced by 25 older people, 75% were due to slips and 25% were the result of trips. However, no further information was provided as to what factors caused the slips, trips and losses of balance.

When examining 91 falls sustained by 50 older people (mean age, 72 years), Berg *et al.*[68] found that falls resulted when subjects tripped over obstacles (19%), slipped on wet or slippery surfaces (19%), tripped over their own feet or for no obvious reason (10%) or slipped while wearing slippery shoes or slippers (9%). Morfitt[66] examined 335 older residents (aged 65 to 74 years) presenting to an accident and emergency department of a hospital following a fall. The author found that when indoors, older people fell as a result of tripping on floor coverings or catching their heels (21%), slipping on wet or polished floors while wearing inadequate footwear (19%) and losing balance from participating in inappropriate activity (10%; see Figure 2.1). Unpublished data from the Illawarra Health Promotion Unit[82] found that their sample of 107 older, predominantly female, people who presented to hospital with a falls-related fractured neck of femur, tripped on uneven surfaces (42%) and slipped on slippery surfaces (41%), with water directly contributing to 17% of slipping falls. Further studies have associated indoor falls with overbalancing when older persons lean forward in chairs, transfers in and out of bed or chairs,

attempts to get into the bath, trips on hazards/obstacles within the room or slips on slippery surfaces[3,45,51,61,63,66,67,70,83,84].

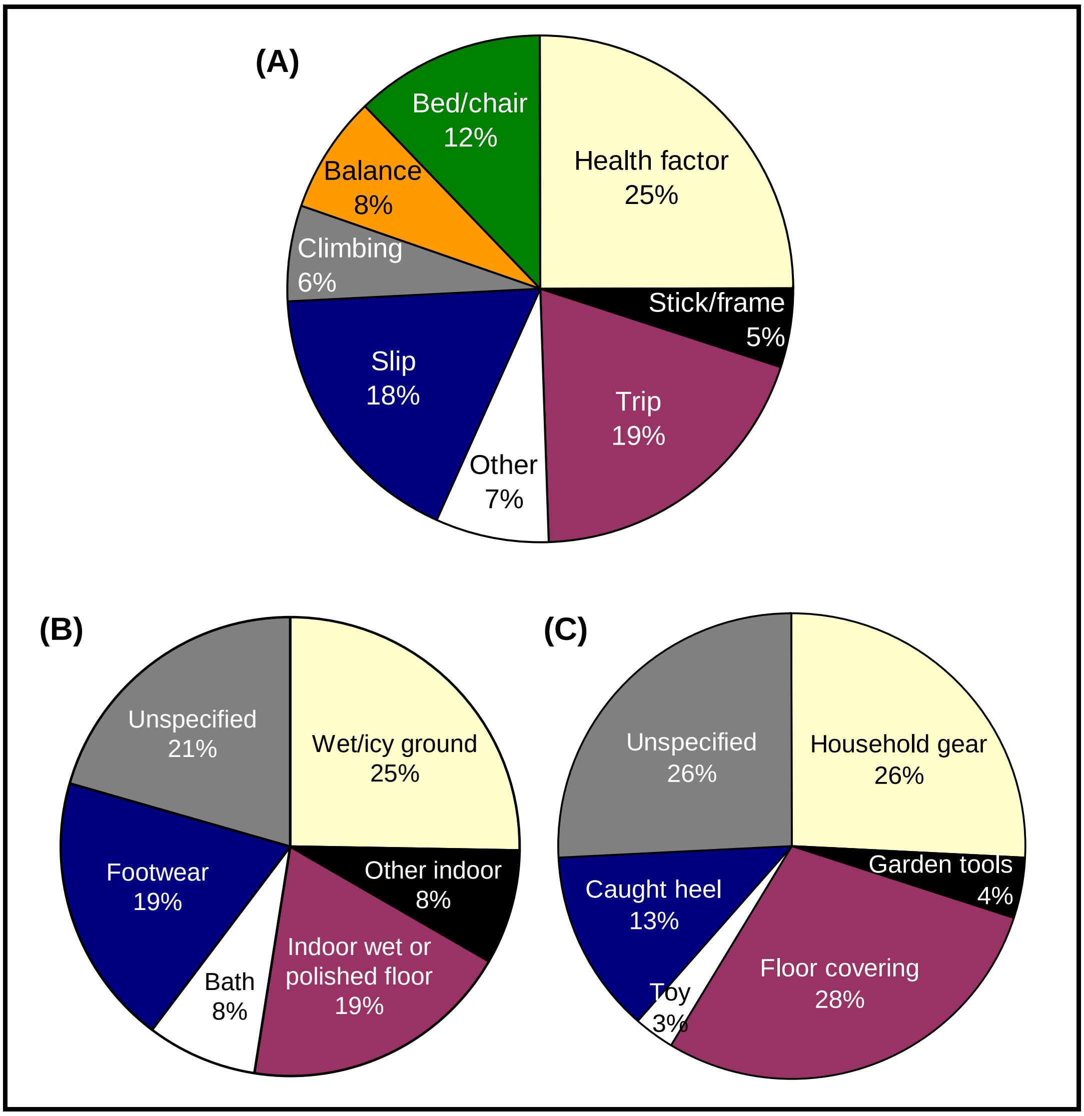

Figure 2.1: The causes of falls (A) and isolated causes of slips (B) and trips (C) for 335 older residents presenting to an accident and emergency department of a hospital following a fall[66].

Any activity that relocates an individual's total body centre of gravity with respect to their base of support, demanding an adequate response by their balancing systems (see Section 2.3.1(C)), has the potential to contribute to falls in older people[45]. Reporting specifically on the mechanisms of falls occurring

inside the home, Luukinen *et al.*[69] found that, although subjects sustained a greater number of trips (279 falls) leading to falls than slips (94 falls), those falls caused by slips resulted in a greater number of fractures (12.8%) compared to the falls caused by trips (8.2%). However, the severity of the outcome of these falls appears dependent on the surface or hardness of the flooring upon which the fall occurs[70] and, as slips commonly occur on harder surfaces, it is important to reduce the incidence of slips in older people.

Slips are usually precipitated by a loss of traction between an individual's footwear and/or foot and the supporting surface[18,83,85]. Studies on the biomechanics of gait and falls have shown that the critical point during level walking, where an estimated 90% of slips leading to falls occur, is initial foot-ground contact[86-89]. Slips at initial foot-ground contact typically occur during the first 5% to 14% of stance[90], because of misjudgements of floor surface slipperiness and unadjusted gait and/or posture patterns[91]. Furthermore, up to 26% of falls-related hip fractures in older adults reportedly result from slips compared to 13% which could be attributed to trips[92]. For these reasons, it is regarded as both more important and more possible to prevent slips rather than trips in older people through intervention, reducing the number of home falls and falls-related injuries occurring because of slips.

2.3 Falls Risk Factors

The multifactorial nature of falls indicates that a single fall is rarely due to a single cause but instead results from the convergence of multiple small, inter-related causative factors that compromise the individual's situation[3,13,15,20,29,44,93-96]. Duthie & Gambert[97] estimated that the exact aetiology of falls may be indeterminate in about 50% of cases, although there is general consensus that the risk of falling increases with the number of risk factors an individual is exposed to[73]. Furthermore, it is often difficult for an older person to recall the circumstances of the fall due to memory loss or emotional distress[72]. While acknowledging the multitude of falls risk factors, the ensuing discussion is limited to research pertaining to those falls risk factors in community-dwelling older populations and those considered directly relevant to the

present thesis. The reader is directed to a report by Hindmarsh & Estes[94] for information pertaining to falls risk factors in hostel or institutional populations.

2.3.1 Personal Falls Risk Factors

Personal risk factors, or those intrinsic to the individual, have been implicated as factors that increase the risk of falls in older people[3,7,11,13,15,20,22,35,42,56,58,65,66,96,98,99]. For example, Morfitt[66] found that personal risk factors could account for 32% of all falls in older people. Furthermore, Rubenstein *et al.*[11] reported that 55% of falls were related to medically diagnosed conditions and Waller[65] determined that an acute or chronic health problem was a contributory factor in 42% of 150 falls sustained by people aged 60 years and above. Much research has been directed towards isolating such risk factors and then intervening to reduce the falls risk[100]. However, although not all of these factors are amenable to change, all are important in identifying individuals or groups of individuals who are at a high risk of falling.

(A) Gender

The life expectancy at birth for Australian men is 75.8 years and 81.6 years for women[101]. Therefore, women predominate among older people, particularly after the age of 75 years[101-103]. In addition to outliving their male counterparts, older women report a greater number of falls than men. Even when adjusted for age, the incidence rates for falls-related hip fractures are approximately twice as high for women as for men[8,104,105].

This gender difference in falls rate has been attributed to a variety of factors such as differing bone compositions, a greater incidence of osteoporosis and a higher overall risk of falling in women[104,106,107]. In addition, women on average are physically weaker and have lower bone mineral densities but a higher percentage of body fat in proportion to their musculature compared to men[108-112]. As a result, women are more disadvantaged when performing weight bearing activities such as walking and stair climbing and are more likely than

older men to have non-lethal disabilities, particularly those associated with musculoskeletal disorders[103,113].

The consequence of this gender imbalance in falls incidence is that older women are more likely to require both informal care and State provided health and welfare services[102,114,115] compared to older men, who are usually able to rely on support and care provided by their wife. In terms of financial cost, in 1995/96 the direct cost of falls for women was in excess of $500 million, comprising 72.7% of the direct costs of falls for men and women aged 65 years and above[59]. Therefore, of older people as a whole, older women place a greater demand on society's social and economic resources compared to their male counterparts[102,111].

(B) Decline in the Musculoskeletal System with Age

All components of the musculoskeletal system, including bones, muscles and soft-tissue structures, decline with ageing. For example, increased collagen and elastin cross-linking after 35 years of age results in stiffened cartilage, tendon and ligament, which become more rigid, in turn, contributing to the onset of osteoarthritis and leading to mobility limitations, walking difficulties and abnormal movements[104,116-118]. Bone mass begins to decrease at age 30 to 35 years at a rate of approximately 1% per year[119], accelerating after menopause in women and at age 50 to 55 years in men[112]. Loss of bone mass is thought to be related to reduced bone mineral density with increased age[112] and is the cause of diminished bone strength with ageing[111,120]. Furthermore, bone diseases such as osteoporosis and osteomalacia can produce extreme bone tissue destruction, contributing to spontaneous fractures that occur during sudden or excessive movement, predisposing individuals to fall[104]. Muscle force is also an important determinant of bone density by being the major contributor to mechanical stresses imposed on the bones[121] and, together with muscle bulk, may absorb much of the energy during a fall, protecting the bones from fracture[122].

Research investigating both static and dynamic muscle strength by means of cross-sectional studies[108,109,123-126] and longitudinal studies[127,128] have

documented that both the proximal and peripheral muscles display an age-related decline in strength. This decline appears to proceed at about 1.5% per year from 45 years of age and may accelerate from age 70 years to 5% or more per year depending upon disease, mobility and the dependence on others to perform daily living activities[104,109,110,126,129-132]. Therefore, by 80 years of age, an individual's muscle strength may have deteriorated by between 20% and 40% relative to their strength at 30 years of age[133]. This loss of muscle strength correlates to a decrease in overall lean body mass[109,111,131-133] and a loss of muscle contractility[109], occurring primarily from decreased fibre number rather than fibre atrophy[109,116,124,125] and an eventual loss of functional motor units[125]. Furthermore, these changes result in reduced muscle power and endurance[111] such that the muscles of older individuals are, on average, slower in contracting and more easily fatigued relative to their younger counterparts[134].

Progressive decline of the musculoskeletal system will have negative effects on balance ability, proprioception and reaction time in older individuals[116]. For example, muscle strength and power have been associated with functional status in older people[135-137]. Consequently, deficits in either strength or power have been found to affect chair rising ability[138,139], walking and stair climbing speed[138,140-142] and the ability to perform daily living activities[3,10,135,143]. Furthermore, strong associations exist between impaired quadriceps[144-146] and ankle dorsiflexor muscle strength and power[145,147] and the risk of recurrent falls[3,13,15,20,48,145,148,149] and fractures[50,150-152]. Adequate functioning of the musculoskeletal system is therefore required for posture maintenance[153] and efficient movement[104,153], as well as to compensate for unstable joints[104,154]. Consequently, a decline in musculoskeletal function may leave the older individual without the reserves to meet emergencies precipitated by the environment, such as occurs during falls[47,104,155] and, without the protective effect of muscle bulk, are more likely to sustain a serious injury following a fall[36]. However, as ageing is a variable process, with biological age not necessarily equating with chronological age[156], the extent to which these changes are directly attributable to the ageing process and not to disuse, reduced physical activity or disease is largely unknown[11,28,154]. Irrespective of the

mechanism, declines in musculoskeletal integrity and function apparent in older individuals predispose them to an increased risk of falls.

(C) Balance and Postural Stability

Balance is defined as the ability to maintain the centre of gravity of the body over the base of support[157,158] such that an individual has the ability to maintain a static position, to move within their environment and to react to a perturbation[28,158-160]. Therefore, the ability to balance relies on an individual's musculoskeletal, sensory and central nervous systems interacting appropriately[159,161-164]. This ability can deteriorate when any of these systems fail, either individually or collectively[165], and is therefore considered a personal falls risk factor.

In addition to age-related deterioration of the musculoskeletal system outlined previously, progressive deterioration of the three sensory systems that control balance, namely the visual, vestibular and somatosensory systems, have also been reported to occur with normal ageing and disease[28,119,166-171]. Vision provides information about the body's position with respect to the environment[28,161,163]. Older persons appear to depend more on visual input than other sensory inputs when controlling balance[172,173] as vision is estimated to contribute approximately 50% of the sensory input needed for balance. In fact, older people may become dependent upon vision when walking, such that they watch their feet[104], particularly if there is impairment of another sensory system.

Age-related visual impairment, particularly from cataracts, glaucoma and macular degeneration[174], may contribute to reductions in visual acuity, contrast sensitivity, night vision, depth perception, peripheral vision and glare tolerance[28,104,168,175,176]. Therefore, older people may require increased time to recover from glare[104], have reduced ability to detect and discriminate objects in the environment, such as steps, gutters, tree roots, pavement cracks and pavement misalignments[176], and have difficulty perceiving slowly moving objects[28,177,178]. If visual information is distorted and unreliable the older individual is predisposed to instability[10], particularly as the visual control of

posture is slow, so that correcting for changes in posture is delayed and righting responses may not be initiated quickly enough to prevent a fall[172]. Therefore, impaired vision is considered a predisposing factor for postural instability and falls in older people[3,10,15,58,96,175,176,179-185].

The vestibular system provides information about linear and angular accelerations of the head as well as the head's orientation with respect to gravity[161,163]. Age-related changes in vestibular function have been well documented[104,119,165] with a 40% reduction in sensory cells within the vestibular system reported in subjects above 70 years of age[186]. These changes are thought to have an effect on the capacity to resolve inter-sensory conflict and therefore interfere with the ability to balance effectively[28]. That is, as any sudden movements or directional changes that are controlled by the vestibular system may result in unsteadiness and cause the older person to stagger and fall[104], deficits to this system can be considered a personal falls risk factor. However, only a few studies have found a strong correlation between falls and vestibular dysfunction[183,187], and therefore the vestibular system does not seem to contribute as largely to postural stability as the visual or somatosensory systems. Therefore, research into the vestibular system and its effects on balance are still inconclusive and methods to assess this system are still being refined.

The somatosensory system controls the ability to receive input from articular and cutaneous mechanoreceptors and proprioceptors (for example, muscles spindles and Golgi Tendon Organs) pertaining to muscular action, body orientation and the environment. This system then processes that information in a meaningful way in the central nervous system for adequate postural control[161-163]. Cutaneous sensation and proprioception show increased thresholds for excitability with increased age[104,119,166,186,188-190] and have been significantly correlated with decreased postural stability and falls in older adults[149,189,191-194]. Furthermore, abnormalities of articular mechanoreceptors have also been related to disturbances of balance[195]. Changes to any of these receptors due to age, disease or injury may contribute to postural instability by giving inappropriate or sub-threshold cues[28].

The central nervous system receives signals from the sensory systems, processing and reacting to these signals to initiate appropriate responses to changes in the internal and external environments[196]. However, age reportedly affects the central and peripheral nervous systems by lengthening reaction time and slowing nerve conduction velocity by between 10% and 15%[119,149,177,178,183,197,198]. These age-related disturbances in neurological function may decrease the speed effectiveness and reliability of postural reflexes[188,199,200]. In fact, Elia[119] reported that changes in the central and peripheral nervous systems were related to a 35% to 40% increase in falls sustained by persons aged 60 years and above. Furthermore, Adelsberg[198] reported a relationship between delayed reaction time and longer coordination time with older people who suffered lower limb fractures and older people who were less active compared to those not suffering lower limb fractures and who were more active.

In general, the age-related decline in all these systems has a negative effect on postural stability[171,201-203] such that increased postural sway has been associated with an increased incidence of falls in older adults[13-15,48,63,77,147,158,159,173,183,204-208]. Age-related impairments to the sensory, musculoskeletal, central and peripheral nervous systems[158,209] may impede an older individual's ability to identify a loss of balance or to recover their balance after a loss of footing, thus impairing their ability to avoid a fall and increasing the likelihood of falls[43,210]. In addition, it is the alertness of the neuromuscular system that makes it possible to distribute the energy of the fall to more than one area of the body[122], possibly reducing injury risk. Fortunately, training muscle strength and/or the systems responsible for sensory organisation, have been found to significantly improve balance in older people[211-213], significantly reduce falling frequency and significantly improve stability under changing sensory conditions in groups of community-dwelling older people[212]. Therefore, the risk of falling may decrease if an older individual is provided with strategies that enable them to remain active and mobile.

(D) Chronic Medical Conditions

The biological, physiological and sociological ageing changes that occur in a person over time are associated with a gradual decline in the body's functional capacities and a reduction in the system's resistance to stress and disease[119]. As a result, older people generally have an increased incidence of infection and a greater number of acute illnesses, as well as multiple chronic diseases[101,111,214-217]. Many of these conditions have been associated with falls in older people, with one study stating that approximately 55% of falls could be directly related to medically diagnosed conditions[11]. Furthermore, fallers have been reported to have more active medical problems than non-fallers[95,218]. The chronic medical conditions that have been associated with an increased risk of falls and fall injury events are detailed in Table 2.1.

Chronic medical conditions usually lead to impairment of sensory, physical, emotional and/or cognitive abilities. Impairment can then lead to functional limitations in the ability to perform daily activities, with consequent disability[219]. Therefore, older people with chronic medical conditions are more likely to give themselves a poor subjective health rating, which has been further associated with falls in the home[10,220,221].

There is also the additional problem of medication use. Although the older population can benefit immensely from psychotropic drugs and other medications in treating their chronic medical conditions, their use has also been strongly associated with an increased risk of falls and femoral neck fractures in older people[3,11,15,41,52,106,222-228]. Medications as such can therefore also be considered a personal falls risk factor.

(E) Foot Problems

The foot forms the body's base of support and is therefore essential during all upright human stance and locomotion, serving to cushion the musculoskeletal system during impact, support the body during ground contact, transmit forces between the ground and the leg, adapt to uneven surfaces, keep

Table 2.1: Chronic medical conditions associated with falls in community-dwelling older adults.

Medical Condition	References
• Parkinson's disease / Stroke	Boult *et al.*[219], Campbell *et al.*[13], Grisso *et al.*[225], Herndon *et al.*[30], Lipsitz *et al.*[229], Nevitt *et al.*[15], O'Loughlin *et al.*[230], Prudham & Grimley Evans[52], Robbins *et al.*[95], Rubenstein *et al.*[11], Sheldon[76]
• Cognitive disease	Campbell[222], Campbell *et al.*[54], Gabell *et al.*[22], Lachs *et al.*[174], Prudham & Grimley Evans[52], Ray *et al.*[106], Sheldon[76], Tinetti *et al.*[3], Tinetti *et al.*[96], Wild *et al.*[38]
• Arthritis	Australian Bureau of Statistics[42], Barbieri[20], Blake *et al.*[44], Campbell *et al.*[13], Connell & Wolf[231], Dunn *et al.*[58], Nevitt *et al.*[15], Robbins *et al.*[95], Tinetti *et al.*[3], Vellas *et al.*[232]
• Diabetes	Rubenstein *et al.*[11]
• Neuromuscular disease	Duncan *et al.*[233], Lipsitz *et al.*[229], Nevitt *et al.*[15], Robbins *et al.*[95], Sheldon[76], Teno *et al.*[234], Vellas *et al.*[232]
• Low body mass	Dunn *et al.*[58]; O'Loughlin *et al.*[230]
• Hearing loss	Lachs *et al.*[174]
• Urinary / Bladder dysfunction	Lachs *et al.*[174]
• Depression	Cwickel *et al.*[220], Lachs *et al.*[174], Robbins *et al.*[95], Tinetti & Williams[218], Vellas *et al.*[232]
• Anaemia	Herndon *et al.*[30]
• High blood pressure	Dunn *et al.*[58], Robbins *et al.*[95], Rubenstein *et al.*[11]

the body in balance and provide a system for sensory input[235-241]. However, in serving so many roles, the feet are particularly vulnerable to the age-related changes occurring throughout the rest of the body[25,242]. These changes are further compounded by the foot's large workload, taking between 8,000 and 10,000 steps during an average day[243]. For a 75 kg individual walking at a stride length of 1.4 m, this equates to approximately 1,016 tons of force being absorbed by the feet[244]. Therefore, it is not surprising that epidemiological studies have shown that the percentage of older people said to have foot problems is as high as 50% to 90%[40,245-256]. In fact, the most important finding from the New South Wales Podiatry Survey[40] was that the rate of foot problems among older people was nearly double the rate of foot problems for the general population.

Women report significantly higher rates of foot problems compared to men[242,248,256-258]. This is perhaps due to the high proportion of women who habitually wear high-heeled shoes with pointed, shallow toe boxes[25,259]. These shoe types place increased pressure on the forefoot compared to the flat shoes or work boots characterised by rounded toe boxes traditionally worn by men[25,255,259].

Foot problems are often classified into categories including nail, dermatologic, neuropathic or orthopaedic problems[256]. Nail problems, such as hard, thickened, ingrown or infected toenails can result from incorrect cutting of the nails as well as persistent trauma, such as external pressure exerted by wearing a shoe that is too short[25,242]. In severe cases, nail problems may prevent an individual from wearing ordinary shoes[19,250,260]. Cartwright & Henderson[246] reported that between 25% to 50% of older people had thickened nails or other nail problems whereas White & Mulley[256] found that 56% of older people (mean age, 84 years) reported nail pathologies, such as ingrown toenails, fungal infection and discoloured nails. If nail problems persist for extended periods of time, they can lead to a severe disturbance of nail growth and destruction of the nail plate[242], resulting in brittle toenails and, in turn, making pedicure more difficult and risky[25]. The high incidence of nail problems in older people is predominantly due to the inability of many older individuals to properly trim their nails as they are unable to bend down to reach their nails due to decreased mobility, poor eyesight, other impairments or due to a lack of appropriate nail trimming equipment[25,42,242,250,252,255,256]. In fact, White & Mulley[256] found that of 25 men (mean age, 84 years) and 71 women (mean age, 84 years), 77% had difficulty cutting their toenails.

Dermatologic issues, such as dry, inelastic and fragile skin, can lead to fissure[*] formation which, together with the breakdown of epidermal integrity, allow bacterial invasion and subsequent infection[25,242,260-262]. Furthermore, when combined with the repetitive shear stresses sustained during daily living activities, fat pad atrophy, elevated plantar pressure distributions, foot malalignment and/or irritation from ill-fitting footwear, poor epidermal integrity

allows focal thickening of the outer keratinised layers, with subsequent development of corns and callosities[25,240,242,250,263]. Corns and callosities most often appear at those sites on the foot which sustain the highest plantar pressures, namely under the heel and the metatarsal heads[249,252,260,264]. Helfand[265] found that dermatological conditions were the most commonly reported foot problems in 417 patients (mean age, 74 years) seen at senior centres over 3 years, particularly hyperkeratosis (48%). White & Mulley[256] found that the most common foot problems reported by 25 men and 71 women (mean age, 84 years) were corns or callosities (68%). Similar findings have also been reported by other researchers[246,250,253,266].

The foot has an abundant network of afferent nerve endings which, via the central nervous system, can respond to pain, temperature, pressure and proprioception, thereby constantly supplying information about the mechanical environment[240,267-269]. In fact, the central nervous system relies on sensory input from muscle and cutaneous receptors in the lower limbs, including the feet, to generate effective motor patterns for human posture and locomotion[270]. Diminished sensory acuity, such that occurs with age (see Section 2.3.1(C)), results in neuropathic foot problems which, in turn, can be detrimental to posture, gait and ultimately, independent living. For example, in older people with adequate foot sensation, corns and callosities may be protective[262]. However, in older people with impaired foot sensation, who comprise close to one third of the population over 60 years of age[265], corn or callus formation may be damaging to the foot, as may be undue pressure from an ill-fitting shoe, wrinkled socks or hosiery or the presence of a foreign object in the shoe[25]. These foreign bodies elevate skin pressures[262,271-273] and, because they are not sensed by the patient, produce a potential site of skin breakdown, leading to ulceration, infection and eventual amputation, particularly for those individuals with diabetes mellitus and peripheral neuropathy[262,271-274].

The constant loads placed on the feet, when combined with restrictive footwear, socks or hosiery and disease, can also contribute to foot deformity or orthopaedic conditions. Common deformities reported by between 25% and

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[*] A fissure is a crack or crevice of the skin[260].

50% of older people[246] include hallux valgus[*], hallux rigidus[†], claw and hammer toes (see Figure 2.2) and bunions[19,242,253,256,264,265,275]. These deformities lead to greater forefoot height and, together with oedema, excessive weight gain and disease, usually widen the older foot[250,276], changing the foot's contour[25]. Therefore, few adults will remain at the same shoe size throughout maturity[25]. However, one study which compared foot size to shoe size found that 88% of 356 women aged 20 to 60 years were wearing shoes that were smaller than their feet (average 1.2 cm smaller), thereby further contributing to foot pain and deformity[257].

Foot impairment can result in considerable pain and disability[231,250,253,254,256,265], gait abnormalities and shoe wearing difficulties, potentially having profound consequences on an individual's mobility, independence[25,245,246,263,277] and falls risk[40,248]. In their prospective community study, Tinetti *et al.*[3] found serious foot problems were associated with an increased relative risk of falling of 1.4 when compared to those individuals without foot problems. Blake *et al.*[44] reported that the incidence of foot problems was a discriminate variable to explain why 356 out of 1,042 older adults, aged 65 years and above, reported one or more falls in 12 months. Similarly, Vellas *et al.*[232] found that 29% of 98 subjects (mean age, 73 years) who reported a fall over 6 months, reported foot problems. In addition, Menz & Lord[253] compared the balance performance of 135 older adults (mean age, 79 years) with and without foot problems, reporting a detrimental effect of foot problems on balance. This was particularly the case when subjects were required to cope with large excursions of their total body centre of gravity, shifting their body weight to the outer perimeters of the stability limits provided by their feet[253]. This association was also consistent with the results of Tanaka *et al.*[278].

To the older adult who is unable to perform sustained activities because of acutely painful feet, foot care is the key to increased mobility, productivity, independence, freedom from pain and general increased quality of life[248,255,265].

[*] Hallux valgus occurs with fibular deviation of the hallux leading to a bunion deformity[275].
[†] Hallux rigidus occurs with progressive loss of dorsiflexion in the first metatarsophalangeal joint[276].

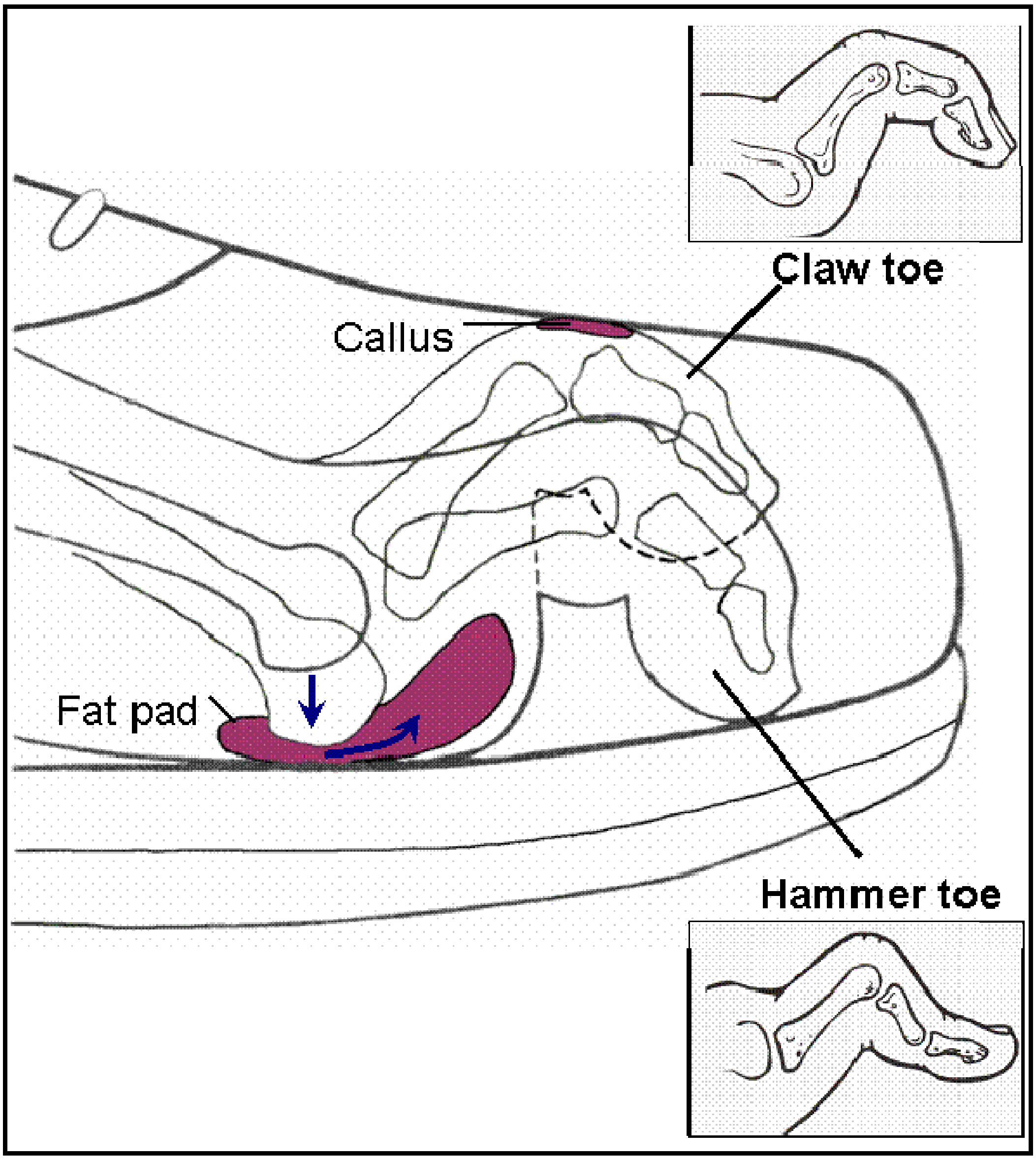

Figure 2.2: Examples of toe deformities commonly seen in the older foot (adapted from Shrader[279] and Coady *et al.*[275]). Claw toes occur when the extensor tendon hyperextends the metatarsophalangeal joint and flexes the proximal interphalangeal joint[276]. Hammer toes occur when the distal interphalangeal joint is pulled into flexion without metatarsophalangeal joint extension such that the metatarsal head drops and the fat pad migrates distally[25]. Also shown is the effect that both toe deformities have on increasing toe height as well as how incorrect footwear types may lead to callus formation.

In fact, Cartwright *et al.*[246] reported that many older people who were housebound attributed their immobility to foot problems. Many foot problems, and associated foot pain, can be avoided or minimised by increased attention to hygienic foot-care practices, treatment in the early stages of pathology by a podiatrist and appropriate footwear[280]. Therefore, older individuals require education pertaining to the causes, effects and severity of specific foot problems to encourage them to consult with appropriate medical personnel about their feet and to allow the prescription of safe and well-fitting footwear in order to decrease pain and/or discomfort, further foot problems and falls risk[25,281].

(F) Gait and Ageing

Changes to an older individual's walking pattern, as well as gait abnormalities commonly found in older people, have been strongly associated with falls[3,15,20,45,47,48,63,95,282-286], functional disability[283,284,287], nursing home placements[283,284] and death[283,284]. Epidemiological data have implicated some aspects of locomotion, such as initiation of walking, turning, walking over uneven surfaces or stopping, in almost all falls[11,29,52,76,82].

The natural processes of ageing that have been discussed in the previous sections may often interfere with an older individual's walking pattern[288-292]. For example, the combination of reduced muscle strength and power, increased joint stiffness, impaired vision, disrupted coordination, poor balance and pain may reduce gait efficiency, leading to an unsteady gait pattern[11,20,104,111,293,294]. Unsteady gait patterns have also been associated with chronic medical conditions, particularly neurological disorders such as Parkinson's disease[295,296], hemiplegia/hemiparesis[104,295,297,298], diabetic neuropathy[299,300] and musculoskeletal impairment, such as arthritis[104,116,294,296,301-308] (see Section 2.3.1(D)). Therefore, most older people recognise that they must walk more slowly, turn more carefully and expect their balance to be less steady than it was when they were younger[179].

In healthy individuals, natural walking speed has been reported to decline by 0.2% each year until 63 years of age, where this decline accelerates to 1.6%

per year[290,291]. Leiper & Craik[309] performed a multiple regression analysis on data obtained from 81 women (aged 64 to 94.5 years) at five different walking speeds and found that age accounted for 30% to 45% of the variability present in walking velocity. Greater reductions in gait speed have been reported in individuals with lower physical function[310,311], reduced lower limb muscle strength and power[138,141,288-290,296,297,312-318], decreased limb excursions[297,319], increased arthritis activity[142,296,297,315,320-324] as well as other musculoskeletal, neurological and cardiorespiratory symptoms[142,321,322]. Furthermore, females are documented to walk more slowly than males[289,290,313] and fallers record slower walking velocities than nonfallers[325-327]. Consequently, walking speed is commonly used to determine functional status[288,314,328,329], readiness for hospital discharge[330] and falls risk[229]. However, no research was located that could attribute those changes in gait due directly to the ageing process compared to those that may be due to disuse, disease or other factors. Instead, research has focused on the kinematic, kinetic and neuromuscular changes that older people make to allow and/or compensate for reduced walking velocity and possible gait instability.

Studies comparing the walking patterns of older individuals to younger people report that older people have reduced step and stride lengths[111,260,297,299,307,315,318,319,321-324,331-335], typically resulting from decreased ankle[299,315,335-338] and knee ranges of motion[20,297,306,319,332,335] and a broader walking base[161,297,307,314,315,319,333,334,339-341]. When combined, these factors generally contribute to an increased stance and double support time[161,260,288,290,297,307,308,313,319,332,333,335], together with decreased swing duration[297,319] and slower cadence[25,308,319,342,343]. However, some studies have reported older individuals to have faster cadences than their younger counterparts[297,332]. These gait adaptations ensure a more stable walking pattern as the amount of time an individual spends supporting their body mass on only one limb is reduced[344] and the total body centre of gravity is relatively easily maintained over the base of support[332]. In addition, reductions in the ground reaction forces generated during gait[299,325], reduced ankle and knee moments[299,337,343,345] and reduced ankle and knee joint powers[299,318,346,347]

contribute to a decrease in the friction required to walk safely across the ground. Therefore, there is a lessened risk of a slip occurring and less load is placed on the joints of the lower limb[260,306], assisting in pain reduction. Furthermore, some authors have reported that the reductions in joint moments and powers may actually reduce muscle activation[306,339,348,349]. Reduced muscle activity would decrease energy expenditure[342,343] when walking, forestalling fatigue[339] and reducing fatigue-related falls.

Although these gait alterations may be advantageous in terms of increasing stability and reducing both pain and fatigue, they also lessen the amount of toe clearance during the swing phase[307,332,350], predisposing the older individual to falls (see Section 2.2.3). Therefore, older individuals may be forced to compensate by increasing hip flexion at initial foot-ground contact to ensure successful toe clearance[335]. It has been postulated that increased knee flexion may also assist the individual to accommodate to surface changes[351]. However, to achieve these gait modifications, greater neuromuscular control may be required relative to an older individual’s normal gait[332,352,353]. Consequently, when combined with decreased gait speed, this modified walking pattern requires greater energy expenditure as the process of momentum transfer and forward progression is disrupted such that each step becomes an effort[353] and, in turn, the older person will fatigue earlier and be at a greater risk of fatigue-related falls compared to their younger counterparts.

Although discussion so far has focused on the changes that older individuals make to ensure a more stable walking pattern, some studies have examined the gait patterns of older people classified as fallers and nonfallers in an attempt to identify those gait parameters that place an older individual at the greatest risk of a fall. For example, in a population of hospitalised older people, Guimaraes & Isaacs[326] found that the fallers walked more slowly and with shorter and more variable step lengths and cadences than the nonfallers. The authors speculated that the fallers were more anxious about falling than the nonfallers, thereby attempting to keep their total body centre of gravity within their base of support for as long as possible[326]. Although this study examined a hospitalised population, the findings were supported by Kerrigan *et al*.[327], who

found that independently living fallers had significantly reduced walking speed compared to nonfallers, due to a reduced stride length. In addition, Hausdorff *et al*.[354] reported that fallers had increased stride-to-stride variability compared to nonfallers. Apart from these differences, few other kinematic or kinetic variables have been able to distinguish between the gait of fallers compared to nonfallers when examining older individuals who are relatively healthy, who live in the community and are not hospitalised[355-357].

Maintaining the ability to walk efficiently is important to older individuals who are constantly faced with the prospect of losing their independence, as walking is a fundamental daily living activity[39]. Furthermore, as many older people may function at a level only just above that needed to maintain independent living[358], changes in their normal gait pattern may be self-perpetuating, leading to a cyclic deterioration whereby, when an older individual feels unsteady, they will exercise less, contributing to further unsteadiness, mobility problems and increasing their falls risk. In very old age this deterioration becomes limiting as walking speeds fall below a safe level for ambulation outside the home[111] and, consequently, the individual ceases to remain living independently. Already, an estimated 8% to 9% of noninstitutionalised older adults have difficulty walking and/or require the assistance of another person or special equipment to walk[296], a number which quadruples when the older individual is diagnosed with arthritis[113]. However, the provision of a walking aid is not without hazard[10] as they may be inappropriate for the user[359] or may restrict activities. Therefore, it is important to identify strategies by which the normal gait patterns of older people can be maintained, such as supplying appropriate footwear[104] and providing exercise programs[316] to ensure older people are exposed to a reduced falls risk when walking so that they can maintain their independent living status.

(G) Mobility and Functional Decline

Non-disabled community-dwelling adults aged 75 years and older reportedly lose approximately 10% of independence in basic daily living

activities each year[325]. However, this progressive decline in daily function may be accelerated by disease, such that there appears to be associations among mobility limitation and functional degradation, advancing age and lower perceived states of health[217,360,361]. Furthermore, due to their longer life expectancies, women tend to live longer with more chronic disabilities than men[362], in turn, being more at risk of falls due to functional decline compared to their male counterparts[363-365].

Decreased mobility[35,47,147,229,233,234] and dependency in daily living activities such as bathing, dressing, toileting, transferring and eating activities[15,47,174] have been implicated in falls in older people. For example, Robbins *et al.*[95] found that functional status scores were significantly lower for older fallers (mean age, 88 years), and a greater percentage of fallers depended on assistive devices for ambulation compared to older non-fallers (mean age, 88 years). Furthermore, Tinetti & Williams[218], who investigated 1,103 community-dwelling people over 71 years of age, reported that declines in the ability to perform daily living activities were associated with both noninjurious and injurious falls over a 3 year period. Earlier studies have also confirmed that older people who ventured outside their neighbourhood less than three times a week or whose longest walk was less than one block in length were at an increased risk of falling when compared to those who ventured out daily or could walk for longer[3,13,15,44,366]. Consequently, tests and/or scales which measure components of functional ability, independence and mobility are used to predict falls in older people[3,15,219,233,361,367].

The relationships among mobility, function, falls and injurious falls, however, is not without conflict[67]. It is generally considered that older individuals who are physically active are better able to maintain good muscle strength, balance and neuromuscular control and are therefore more physically functional compared to sedentary older people[67,230,368]. This increased functionality may reduce their risk of falls and the effects of post-fall syndrome[67,230,368]. However, older people who are mobile and independent may in fact be at an increased risk of falls as they more often place themselves in situations and expose themselves to environmental hazards that could contribute

to a fall[67,230,368]. For example, voluntary daily living movements such as reaching, turning and walking, place people in unstable positions and therefore increase their risk of falls[3,233]. Conversely, self-imposed decreased activity[230,369], such as that which occurs with a fear of falling or a low health self-perception, may contribute to decreased mobility and increased functional dependence, perhaps being used as a strategy by older people to decrease their potential for future fall events[218,370]. Therefore, while unfit and/or bed-ridden elders may fall, older people who are both mobile and unstable, may in fact be most at risk of falling[3] and strategies to minimise the falls risks for older mobile people are therefore urgently warranted.

2.3.2 Environmental Falls Risk Factors

Environmental hazards within the home have been implicated as contributing to falls in community-dwelling populations[3,11,29,45,65,66,68,208,371-376]. Lipsitz *et al.*[229] reported that falls due to environmental factors only accounted for 10% to 30% of falls whereas Rubenstein *et al.*[11] reported that 37% of falls were related to environmental hazards. Common environmental hazards identified as contributors to falls include throw rugs, loose carpets, slippery and shiny surfaces, steps, cords and wires on the floor, cluttered hallways and rooms, low-lying objects such as toys or pets, low beds and toilet seats, poorly maintained walking aids and equipment, unstable furniture, poor or unsafe footwear and dim lighting[3,11,29,45,65,66,68,208,371-376]. These items, however, only become fall hazards if an individual has a reduced ability to interact appropriately with their environment due to personal limitations (see Section 2.3.1). Therefore, it is important to consider the interaction between personal risk factors contributing to compromised physical function and environmental hazards when formulating strategies to reduce the risk of falls in older people. Two main environmental factors associated with home falls in community-dwelling older people are footwear and surface type and, in particular, the interaction between these two factors.

(A) Footwear

Inappropriate footwear has been cited as a major contributory factor in falls in older people[3,11,19,22,61,281,375,377,378]. The Australian Bureau of Statistics[42] completed a report on 139,500 people aged 65 years and over who reported a fall in 1995 in New South Wales. It was reported that 4% of these falls could be directly related to the footwear worn at the time of the fall. However, 35% of respondents stated that a surface contributed to their fall and so it would appear, when considering footwear-surface interactions, that footwear might have been a contributory factor in a greater number of falls than specified. Barbieri[20] retrospectively reviewed 420 fall incidents over 1 year that had been reported by people aged 20 years and above whilst hospitalised. The results revealed that 51% of falls could be related to poorly fitting shoes. Similarly, Sehested & Severin-Nielsen[379] identified 13 out of 25 older hospitalised patients (52%) fell because of ill-fitting shoes or slippers and Marr[21] reported that, when asking older patients to reveal what they were wearing when they had a slip or fall, the answer was constantly old, worn slippers or shoes with minimal support around the ankle and midfoot.

In a prospective study of 100 subjects aged 65 years and above, Gabell *et al.*[22] found that 45% of the 22 recorded falls were due to subjects wearing footwear classified as "unhelpful" at the time of the fall. These footwear types included heavy boots, slip-on shoes, slippers with worn soles and slippers with excessively slip-resistant soles. A study in the Illawarra region (New South Wales, Australia) reported that over half of 107 subjects (mean age, 77 years) who presented to hospital with a fractured neck of femur as a result of a fall were wearing slippers (33%) or footwear characterised as unsafe (31%) at the time of the fall, with a further 22% not wearing anything on their feet when they fell[23]. These slippers and unsafe footwear types were worn every day (70%) because they were comfortable (73%) and 80% of respondents felt that the footwear they fell in was safe. A more recent study conducted on 95 older people (mean age, 78 years) who lived independently in Sydney (New South Wales) and who had suffered a falls-related hip fracture again confirmed slippers to be the most commonly worn footwear type at the time of the fall

(22%)[378]. However, when interviewing older people presenting to emergency departments as a result of a fall, Morfitt[66] found that footwear only contributed to 6% of falls in 339 persons aged 65 years and above and Waller[65] found no relationship between the footwear worn and the mechanism of the fall for 150 people aged 60 years and above. Upon further analysis, the study by Morfitt[66] attributed the falls to other environmental factors, although the footwear worn could have also contributed to these falls by influencing surface frictional characteristics. Furthermore, in the study by Waller[65], 30% of falls occurred because of slips, 12% occurred because of trips and 13% of all falls occurred when the person was not wearing any shoes. Therefore, it would again appear that footwear, or the lack of it, might have contributed to the reported falls.

The significance of footwear in falls incidence was noted by Marr[21] who advised all hospital personnel to record information about the footwear worn by patients preceding a fall during clinical history taking. Furthermore, providing appropriate footwear to older individuals is currently a strategy used to enable them to remain active and mobile, thereby reducing falls risk[19,380]. However, more research is warranted to investigate the specific characteristics of the shoes worn by older individuals when falling, the reasons as to why these shoes are worn as well as how these shoes interact with common household surfaces.

(B) Surface Type

There is little research on the role that surface type plays in relation to falls and falls-related injury. This is generally because the surface that is walked upon cannot be readily chosen and has therefore traditionally been ignored in studies assessing falls risk[381]. Regardless, past literature has implicated ill-repaired steps and cracked sidewalks as common outdoor falls hazards[11,22,65] and loose carpets, slippery floors and reflective surfaces as frequent environmental hazards inside the home[11,47,51,65,375]. The Australian Bureau of Statistics[42], in their report of 139,500 people aged 65 years and above who reported a fall, found that 48,800 (34.9%) people implicated uneven or cracked man-made surfaces and “slippery” surfaces as contributing to their fall. Furthermore,

Vellas *et al.*[232] reported that 47% of 114 falls incurred by 488 people (mean age, 74 years) occurred on a smooth surface; Wild *et al.*[228] recorded that 7% of 125 fallers seen in physician offices after a fall, fell due to a specific hazard such as a wet surface; and Norton *et al.*[382] found that 66% of falls-related hip fractures were sustained on wet or slippery surfaces. However, no further details pertaining to surface type, lubricant or surface condition were reported by the authors of these studies and none of these studies distinguished between falls that occurred on indoor surface types to those which occurred on outdoor surface types.

Few studies have been published which have examined indoor floor surfaces and falls incidence, specifically in relation to community-dwelling older people. Morfitt[66], who examined 335 older residents (aged 65 to 74 years) presenting to an accident and emergency department of a hospital following a fall, reported that when indoors, older people fell as a result of trips on floor coverings or catching their shoes (21%) and slips on wet or polished floors when wearing inadequate footwear (19%). The Injury Control Council of Western Australia[74] reported on 280 falls sustained by 1,091 community-dwelling older people (aged 70 years and above) inside the home. The Council found that wet, slippery floors in the bathroom, laundry, toilet and kitchen were directly responsible for 22 (8%) falls, trips and slips on floor rugs and mats were directly responsible for 15 (5%) falls and a further 2 (1%) falls directly resulted from dry, slippery floors in the bathroom. The results of this study found that, compared to all other environmental hazards, slippery and wet floors were the third most prevalent falling risk factor accounting for falls inside the home[74].

The Illawarra Health Promotion Unit (1997, unpublished data) found that 37% of the falls sustained by 107 older community-dwelling people (mean age, 77 years) occurred on a carpeted surface whereas a further 24% occurred on linoleum and tile floorings. Furthermore, this survey found that when the respondents were wearing slippers they fell on low-pile carpet (38%), tiles (21%) and linoleum (18%). However, those falls that occurred because of a slip were predominantly caused by slippery surfaces (62%; Illawarra Health Promotion Unit, 1997, unpublished data). Healey[383] analysed a sample of 213

accident forms selected at random from all accident forms completed in an elderly care unit (mean age, 86 years) over 4 years. The analysis revealed that significantly more residents fell on vinyl flooring (86 women and 100 men) compared to carpet flooring (16 women and 11 men), with 91% of those patients who fell on the vinyl floor sustaining an injury compared to 15% of patients who fell on the carpet surface. However, the author was unable to determine whether patients were less likely to fall when walking on carpet compared to vinyl tile because it was unknown how long each patient spent in carpeted rather than vinyl-floored areas[15]. Therefore, although the results of the study supported previous literature, which has found that hard surfaces lead to a greater risk of major injuries when compared to falling on softer surfaces (see Section 2.2.3), this reduced risk of injury when falling on carpet would not be of benefit to patients if it were paired with an increased likelihood of falls. Despite the importance of knowing how these typical household surfaces affect the gait of older people, and in turn, the incidence of falls and falls-related injuries, only three studies[384-386] were located which investigated the effects of typical household surfaces on the gait of older people.

Willmott[384] compared the walking patterns of 58 hospital patients (mean age, 76 years) as the patients walked on carpeted and reflective vinyl tiled floors. When walking on the vinyl tile surface, subjects displayed a significantly shorter step length, which contributed to a significantly slower gait speed, compared to when walking on the carpet surface. In fact, step length decreased by up to 30% in a quarter of all trials when subjects walked on the vinyl tile floors. The authors speculated that the shortened step length evident when the older subjects walked on vinyl tile floors may have been caused by a fear of falling with subjects displaying more efficient and confident gait when walking on the carpet floor compared to the vinyl tile floor[384]. These results were supported by Bunterngchit *et al.*[385], who investigated the gait of 10 college students (mean age, 26 years) and 10 older individuals (mean age, 72 years) walking on carpet and vinyl tile and transitioning between these surfaces.

Dickinson[386] assessed whether various residential carpet and underlay pad combinations contributed to balance and gait problems among 25 healthy,

community-dwelling older adults (mean age, 73 years). In contrast to the findings by Willmott[384], Dickinson[386] found that older adults walked significantly slower on the carpet compared to a vinyl tile surface. Furthermore, this author suggested that subjects were more hesitant when they encountered the more compliant* surface. Stephens & Goldie[387] reported a similar finding in that patients walked significantly slower on a carpeted surface compared to a wooden surface at both a self-selected comfortable and fast walking pace. However, this study analysed data for 24 stroke patients, who are known to have a shuffling gait pattern[307] and, therefore, these stroke patients may find a carpet surface more challenging to walk on compared to a wooden floor[387].

Although not examining an older population, Whittle *et al*.[388] analysed young subjects walking barefoot at a self-selected comfortable speed across a carpeted and an uncarpeted surface, the specific type of which was unspecified. Despite the expectation that carpet, being a slightly resilient surface, would modify ground reaction forces[389], there were no statistically significant differences between the carpeted and uncarpeted conditions for either the temporal-spatial gait parameters or the ground reaction forces. However, significant differences were reported between the two surface conditions for five measurements of sagittal plane joint angles. Maximum hip extension and maximum ankle dorsiflexion during swing significantly decreased whereas knee flexion at heel strike, knee flexion at swing and maximum ankle dorsiflexion during stance all displayed significant increases when subjects walked on the carpeted surface compared to the uncarpeted surface. However, although statistically significant, the observed differences were only small and therefore the author concluded that they might have been due to random variation and recommended further research. A similar study by Augsburger *et al*.[390] confirmed these findings, reporting no differences in force platform recordings when healthy adults walked on carpeted and uncarpeted laboratory floors.

The surface covering the floor of most people's homes in developed countries has traditionally been carpet, particularly in the bedroom and lounge

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* Compliance is a function of the thickness and density of the different floor surfaces, such that a more compliant surface is thicker and softer[387].

room, and chosen predominantly for warmth and comfort[383]. Being a more compliant surface, carpet is often reported as comfortable to walk on[384,388,390,391]. Its use has also been associated with preventing oedema, possibly due to reduced forces in joints, muscles and ligaments; reduced energy expenditure; and a reduced presence (or absence) of microtrauma as a result of its compliant properties[388,392]. Furthermore, fewer and less severe injuries often result from falls that occur on carpet (see Section 2.2.3). However, carpet, particularly high pile and shag carpets[387], may provide a less firm base during gait and require greater toe clearance compared to harder, flatter surfaces[393]. Consequently, subjects, particularly those with sensory impairment, muscle weakness or musculoskeletal abnormalities or inadequate footwear, may not be able to detect and respond to these soft and uneven surface types, to make necessary compensatory motor output adjustments accurately or efficiently to avoid a loss of balance and an ensuing fall[393,394]. This soft surface phenomenon has also been reported when older subjects complete balance tests on altered or softer surfaces[149,188,190,194,395] or walk across uneven surfaces[393,396,397].

In contrast to the more compliant carpet surfaces, harder surfaces such as ceramic tile, vinyl tile or linoleum are typically laid in the kitchen, laundry and bathroom areas, predominantly for cleaning ease and hygiene[383]. Although, these surface types may be easier to traverse for the older individual who walks using a shuffling type gait, they are typically smooth, shiny and may be slippery, particularly when combined with inadequate footwear and/or covered in a contaminant, such as water. Therefore, these surfaces may lead to fear-of-falling evoked gait adaptations[385], increasing the risk of falls in these individuals (see Section 2.3.1(F)). Furthermore, a greater number of injuries, together with more severe injuries, are recorded when older people fall on these harder surfaces (see Section 2.2.3).

Information with respect to the most appropriate surface to lay inside the homes of older people still remains unanswered. This is particularly due to the effects that surface type has on gait and falls and, is dependent upon the biomechanical factors involved in walking as well as the psychophysical factors of how walkers adjust their gait based on their perception of floor slipperiness

and/or softness, shoe characteristics, shoe and surface composition and condition and surface contaminants[87,90,398-402]. Therefore, further research is warranted to ascertain appropriate household surfaces for older people, particularly focussing on how various shoe-surface combinations influence the gait of older people.

(C) Shoe-Surface Interaction

Studies on the biomechanics of gait and falls show that the critical point during level walking where an estimated 90% of slips leading to falls occur is initial foot-ground contact, which typically occurs at the heel[86-88]. Slips may also occur if the toe slides backwards during terminal stance or sideways when turning on the ball of the outer foot[88,353,399,403,404]. However, these falls typically result in injuries to the upper limbs as the individual falls forwards. In contrast, more devastating lower limb injuries, such as a fractured neck of femur, predominate with backward falls resulting from slips at initial foot-ground contact.

To identify hazardous slip conditions and to assist in designing safer environments, research has concentrated on an objective measure of slip resistance, namely the coefficient of friction[87]. The coefficient of friction is the ratio of the foot's horizontal shear forces divided by the foot's vertical normal force when interacting with the supporting surface. The coefficient of friction is often divided into a static component[*] and a dynamic component[†], although due to the many differences in coefficient of friction testing methods[85,405], there is much debate as to which provides a better estimate of the degree of shoe-surface slipperiness[85-87,90,400,405-407]. As a result, most researchers analyse both the static and dynamic coefficient of friction to quantitatively determine classifications of safe and hazardous shoe-surface conditions[85,87,400,406,408].

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[*] Static coefficient of friction is defined as the shear force required to initiate sliding of the shoe material over the supporting surface material divided by the vertical force on the material, that is, the resisting force at the instant relative motion begins between the sole and floor[87,353].

[†] Dynamic coefficient of friction is the shear force required to sustain movement of the shoe material divided by the vertical force, that is, the resisting force when movement is occurring without interruption[87,353].

The static coefficient of friction threshold value required for traction safety during normal, level walking is generally accepted to be 0.50, with values of less than 0.50 becoming gradually more hazardous and commonly resulting in slips and falls[407,409]. However, both Lin *et al*.[353] and Sherman[410] recommended that a static coefficient of friction of 0.60 should be considered a minimum for safe walking surfaces. With respect to the dynamic coefficient of friction and traction safety, Perkins[411] reported a threshold value of 0.28 at initial foot-ground contact for normal, level walking and Strandberg[86] suggested that this value would normally not exceed 0.25. These values reportedly increase to 0.40 with faster walking[90] and have been found to be significantly greater for mobility impaired individuals (0.64) than for able-bodied individuals (0.31) near initial foot-ground contact, regardless of the walking speed[412]. Coefficient of friction values on either side of these threshold values may predispose an individual to slips, falls and falls-related injury. For example, dynamic coefficient of friction values between 0.15 and 0.19 will usually evoke slips with a recoverable loss of balance. At values below 0.15, imminent slips generally result in unrecoverable losses of balance and consequent falls[398,399,411]. However, coefficient of friction values higher than the threshold values may result in foot fixation and, due to excessive torque, may also contribute to falls[413,414]. Both Gabell *et al*.[22] and Connell & Wolf[231] have described incidences in which excessive friction between the shoe and the supporting surface resulted in falls in older people.

Lubrication can add to the risk of slips by providing a thin layer of fluid between the shoe and the surface[415]. The lubricant, therefore, lowers the coefficient of friction, particularly during motion when it can best work between the two surfaces[400]. Previous studies have shown that, regardless of shoe sole material, wet, dusty or icy surfaces impart a risk for older people in terms of slips and falls[22,65,66,404]. Interestingly, Gard & Lundberg[416] reported that a lot more pedestrians chose to stay inside during wet weather compared to dry weather, possibly because of the perception of slippery surfaces and the increased likelihood of falls. The effect that a lubricant has on the calculated coefficient of friction depends upon the area of the contacting surfaces, the

roughness of the shoe and the surface, the velocity of the contacting surfaces, the vertical loads, the lubricant viscosity and combinations of these conditions[90]. For example, a larger contact area, a higher fluid viscosity and a lighter compressive force acting on the lubricant, makes it harder for the lubricant to squeeze out, thereby lowering the coefficient of friction[90]. In comparison, the size, shape and number of irregularities on the interacting surfaces, together with very slow velocities may allow a fluid to drain more effectively, hence improving the coefficient of friction values[90]. Therefore, it is only with knowledge of both the shoe and surface properties that the effect of a lubricant can be determined.

In addition to influencing frictional properties, different shoe-surface combinations will introduce different types of sensory input into an individual's physiological balance control mechanisms[393]. As such, these different combinations may considerably influence an individual's movement technique, altering the muscle activation patterns, loads on the body and energy expenditure required to perform the task[417-419]. When a slippery floor surface is perceived, an individual typically adjusts their gait to elicit slip-avoidance behaviour or a more stable gait pattern[85,90]. This adaptation usually occurs approximately four or five paces before the surface is reached so that the hazard can be avoided[353]. Commonly reported gait modifications in response to a slippery floor include a shortened stride length, which will generally decrease the required coefficient of friction[411]. This is because the required coefficient of friction has been related to a tangent of the angle between the leg and a line perpendicular to the foot[411]. Shortened step/stride lengths will lead to reduced walking velocity, reduced foot velocity and smaller foot shear forces, as an individual attempts to maintain their total body centre of gravity over their base of support[85,89,353,385,420,421]. Furthermore, co-contraction of the lower limb musculature occurs in response to a slippery floor to prevent foot slippage and an ensuing fall[85,89,353,385,420,421]. Interestingly, the gait changes made in response to a perceived slippery surface are similar to those displayed by ageing individuals, particularly those with disease or a fear of falling (see Section 2.3.1(F)).

It has been suggested that perceived floor slipperiness is subconsciously estimated and memorised by individuals from preceding steps or previous experiences, forming a mental slipperiness model, which can be updated whenever perceptions of surface conditions differ from what was expected[400,422]. This updated information is maintained by the visual, proprioceptive and vestibular systems[172,423,424]. If, however, the slippery surface is not detected within an individual's effective visual field[*], or the surface slipperiness is misinterpreted, there is limited time available to make immediate compensatory gait adjustments to accommodate the slippery surface and the likelihood of a slip is significantly increased[340,353,399,416,425-429]. Therefore, correct subjective perceptions of shoe-surface slipperiness are vital for older individuals with age-related declines in muscle strength, flexibility and proprioception (see Section 2.3.1(B) and Section 2.3.1(C)) to ensure falls are prevented.

A study by Swenson *et al*.[85] found a strong correlation between static coefficient of friction values and subjective ratings of perceived slipperiness of steel beams by both professional steel workers and students (aged 20 to 60 years). Similarly, Myung and colleagues[419] found that subjects appeared able to relate their sensations to a function of the static coefficient of friction. However, despite examining subjects of a similar age range (17 to 50 years), Cohen & Cohen[426] found that subjects were poor at perceiving floor tile slipperiness, being unable to reach agreement with respect to the static coefficient of friction measurement. This latter study used different methods to analyse subjective perceptions of surface friction, finding that tactile cues were the most sensitive to the static coefficient of friction, although visual and auditory senses at times appeared to override the tactile sensation, giving the individual false impressions about a surface's slipperiness[426]. As the foot's tactile senses are often inhibited by footwear and may be impaired in older individuals[430], people will often rely upon visual and auditory cues to assess surface slipperiness, despite the risk of making an incorrect decision about the degree of slipperiness[426]. When one considers the age-related deterioration in the visual, auditory and proprioceptive

[*] An individual's effective visual field is usually between 3 m and 5 m away[353].

systems (see Section 2.3.1(C)), older individuals may have an even more reduced ability to correctly perceive surface slipperiness[91,340,422], compounding the effect of slippery surfaces and escalating the falls risk in these people.

Despite the volume of research investigating surface slipperiness, very little research has investigated slipperiness associated with typical household shoe-surface combinations. A slip will occur if the frictional requirements of the task being performed, such as walking, exceed the available frictional capabilities of the shoe-surface interface[431]. However, slips in the home also occur from misinterpreted slipperiness perceptions as well as undetected surface irregularities, such as a drop of water on a bathroom floor. Therefore, there is a need to determine how older people perceive different household shoe-surface combinations and how these perceptions affect the walking patterns of older people if we are to better understand the complex relationship between gait biomechanics and slip avoidance.

Summarising the information presented on falls risk factors, older people form an extremely heterogenous group with a large proportion having at least one or two disabilities[432]. Consequently, an older person prone to falls is likely to have multiple factors contributing to this risk. As falls are produced by random situations[433], preventing a fall depends upon the timely initiation of an appropriate postural response to control the body's centre of gravity once a displacement occurs[434]. Of the many risk factors implicated in home falls, the major factors cited include inappropriate footwear, slippery surfaces and gait modifications caused by musculoskeletal problems such as arthritis and foot pathologies[19,44,63,65-67,208,377]. Most progress in preventing falls will come from identifying and mitigating environmental causes of instability[12]. Therefore, it is prudent to focus research attention on household shoe-surface interactions and how these affect the gait of older people, particularly those with musculoskeletal problems. However, before examining such interactions it is important to understand the unique footwear needs of older people.

2.4 Footwear Needs of Older People

Historically, footwear was devised purely as a means of protecting the feet from the environment[26,435,436]. In later years, shoes became more ornate with heel height and colour ornamentation being used to indicate social class[26,435]. Shoes were also so expensive that people bequeathed their footwear to family and loved ones, giving rise to the saying "following in your father's footsteps" [435]. With further developments, footwear was used to indicate fashion with society dictating what was deemed appropriate shoe fit and function[435,436]. For the older individual whose feet have borne static and dynamic loads for many years, and who may have foot pain and/or deformity (see Section 2.3.1(E)), shoes may mean the difference between painful and non-painful gait[262,437] and, consequently, independent living or institutionalisation[19,24,250]. Therefore, shoes are considered the most essential element in an older person's wardrobe[24,250,280].

Shoes have six main roles, all of which attempt to be achieved within the realms of fashion. Shoes should protect the feet from the environment, provide friction between the shoe and supporting surface, stabilise the foot and ankle, attenuate impact forces, treat and/or accommodate foot deformities and provide a foundation for foot orthoses[19,24,242,438,439]. For older people with orthopaedic disorders, footwear serves the additional purpose of stress reduction by redistributing weight from sensitive and deformed structures to pain-free areas[24]. If all these functions are satisfied, the shoe should be comfortable and safe for the older wearer. However, incorrect footwear selection may be detrimental to the older individual, in that many foot problems and deformities have been documented to be created by shoes that are improperly manufactured, fitted, or both[26,252,257,262,436,439-448]. Published studies regarding societies that do not wear shoes demonstrate feet which are free from the disabilities commonly noted among shod populations, such as blisters, bunions, corns, ingrown toenails and callosities[26,439-441,445,449,450]. Therefore, one might ask why older individuals are advised to wear shoes.

With age, people's skin becomes less pliant and, together with dryness, leads to fissures, providing an excellent avenue for bacterial invasion. Although skin does repair, it does so slowly, increasing the risk of infection[264]. Furthermore, diminished

sensory acuity and circulation mean older individuals may be unable to feel pressure, pain or temperature (see Section 2.3.1(C)). Consequently, wearing shoes is usually recommended for older individuals to protect the insensate or older foot and avoid skin breakdown[451]. The literature also shows that older subjects display better stability when performing balance tests when wearing shoes compared to barefeet[210,452-454], and that older subjects have faster self-selected walking speeds when wearing shoes compared to barefoot walking[291,452]. Furthermore, older people still record falls when not wearing shoes (see Section 2.3.2(A)). For example, Hourihan *et al.*[23] reported that 22% of older people who fractured their neck of femur following a fall were barefoot at the time of the fall. Therefore, correct footwear can protect the older foot as well as assist in increasing mobility, independence, freedom from pain and general well being[24,255,257,438,439]. In contrast, incorrect shoes may deprive older individuals of mobility and contribute to foot problems[250-252,254] and falls[3,19,46,377].

2.4.1 Household Footwear

Only three studies[19,46,255] were located which specifically investigated the footwear types worn by older people throughout the day. Robinson[255] assessed what footwear 95 older men and women (mean age, 73 years), who lived independently in South Australia, wore at the time of a face-to-face interview. The results revealed that 42 respondents (44%) were wearing slippers with 80% citing slippers as their principal daily footwear. Respondents claimed that they wore slippers because of foot problems, donning ease, low expense and convenience. The author concluded that, when combined with the other main shoe types worn, namely slip-on and court shoes, inappropriate footwear was a significant factor in reducing the mobility of older people, possibly compounding the necessity for walking frames and potentially lowering the optimum exercise level of older people. However, Robinson[255] did not collect any information regarding the actual effects of these footwear types on the gait of the respondents or on their falls history to substantiate her claims. Nevertheless, a similar result was reported by Finlay[19] who found that 55% of 206 older women and 21% of 68 older men, who were admitted to a geriatric

medical unit over 3 months, were wearing shoes that the authors classified as unsatisfactory or potentially dangerous. These footwear types included slippers (28%), high-heeled shoes (25%) and slip-on shoes (24%), with an additional 9% of both men and women not wearing any form of footwear upon admission[19]. Despite all patients reporting that they suffered foot problems, most of these were minor nail problems. However, 69% of those wearing slippers or slip-on shoes presented with claw or hammer toes compared to 52% of people who wore shoes with adequate fastenings. Unfortunately, the only information provided by this author in relation to falls, functionality and footwear was that half of those who wore slippers had a recorded history of falls[19].

A comprehensive assessment of what community-dwelling older people, aged 65 years and above, wore at home was undertaken by Dunne *et al*.[46]. Of the 492 women (75%) and 160 men (25%) who responded to the survey, only 167 respondents (26%) were wearing shoes classified as sturdy at the time they were contacted by telephone. The next most frequent responses were persons going barefoot or in socks (20%) followed by persons wearing house slippers (18%); laced, canvas shoes (15%); loafers or slip-on shoes (10%); thongs (8%) and dress shoes (3%). However, 69% reported wearing sturdy shoes at some time during the week, with 64% wearing them daily. Expense (6%), style (6%), donning problems (6%) and foot problems (13%) were raised as reasons for not wearing sturdy shoes, although close to half of the respondents (47%) felt that their regular shoes, despite not fitting the definition of a sturdy shoe, were adequate for their needs. Although the study did not intend to find an association between falls, functionality and footwear, the authors reported that 28% of respondents had fallen in the 12 months prior to participating in the survey and fallers were just as likely as non-fallers to be wearing sturdy shoes at the time of their fall. Further research is therefore required to identify how indoor footwear affects the gait of the wearers to enable us to better understand which shoe characteristics may predispose older people to fall in the home. Only then can safe household footwear be designed and manufactured, which not only protects the older individual from falls but also allows for freedom from pain, mobility and, therefore, independent living.

2.4.2 Components of a Safe Outdoor Shoe

Before discussing what factors are currently deemed components of a safe shoe, it is necessary to review the structure of a typical shoe. Figure 2.3 displays the components of a shoe, the main sections of which include the upper, the portion above the sole that covers the dorsum of the foot; and the sole, the layer of material between the plantar surface of the foot and the supporting surface. The upper functions to support and protect the foot and ankle, passively control foot temperature and humidity levels, and determine shoe comfort of fit[26,455,456]. The shoe upper can be divided into an anterior portion, namely the vamp which surrounds the toes and instep, comprising the toe box, instep, throat and shoe closure; and a posterior portion called the quarter, which surrounds the heel and ankle, comprising the heel counter and collar. The sole, composed of a midsole, outsole and heel, functions to absorb impact, provide traction, enable proper foot function, provide foot comfort and protect the foot from the ground. Table 2.2 documents a list of features, compiled from past literature, currently recommended as guidelines when selecting shoes for older people[19,24-26,61,86,87,231,242,250,260,275,276,301,307,375,381,392,400,413,414,436,438,454-490].

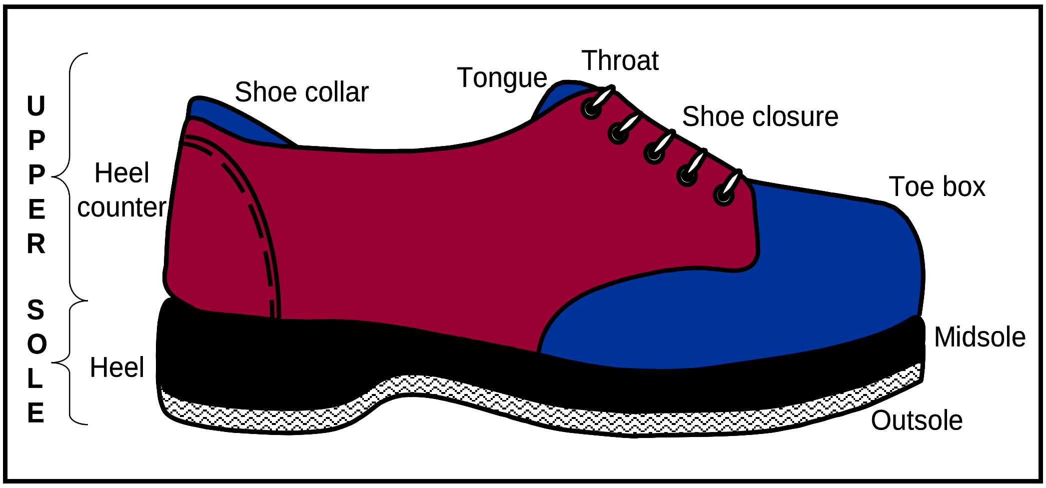

Figure 2.3: Components of a shoe (adapted from Lord *et al.*[72]).

(Continued on next page)

Table 2.2: Recommended guidelines to assist older people when selecting a well-designed, safe slip or trip) that is suitable to be worn outside the home.

Shoe Component	Recommended Guidelines
Total Shoe	
• Shoe fit	• Overall shoe length should be 13-16 mm longer than the individual's longest toe and the ball of the foot should fit in the widest part of the shoe, ensuring that the shoe bends at the metatarsophalangeal joints. • Close fit should replicate the effects of a support stocking on circulation in the foot.
Upper	
• Material	• Soft, lightweight and firm to support shoe shape, yet be easy to wear and gentle on older, less elastic skin. • Flexible and elastic to accommodate many foot shapes and allow both unhampered and complementary flexibility of the foot as well as prevent oedema. • Porous to allow the foot to breathe to prevent burning and unusual perspiration and reduce the possibility of skin irritation, yet warm to protect the foot with circulatory problems. • Moisture resistant but washable, particularly in the case of incontinent older adults. • Lined to provide padding to the foot and prevent irritation, providing additional comfort and pain relief to the older foot, and replacing the need to wear socks or nylons[a].
• Vamp	• Plain so as to cover the forefoot and toes in one piece, reducing the number of seams and decorative stitching that could cause irritation to delicate skin on the dorsum of the foot, particularly the toes.
• Toe box	• Wide and deep to allow the toes and forefoot to spread comfortably inside the shoe during weight bearing, permitting normal toe motion. • Accommodating to forefoot deformities, such as hallux valgus, claw and hammer toes and nail anomalies as well as corns and callosities, bunions, oedema, bulky foot dressings and/or ulcers.
• Throat	• Wide opening to permit the fitting of a foot of large girth, ensuring the shoe is easy to don and greater adjustment over the dorsum of the foot by means of shoe closures compared to narrov

[a] If not lined, socks and stockings should be worn to form an interface between the foot and the shoe to prevent irritation from the stitching and dyes in the shoe materials and keep the feet in the cleanest environment possible. Socks and stockings should not constrict the foot or be wrinkled and should be porous to reduce the risk of excessive perspiration.

Table 2.2: Recommended guidelines to assist older people when selecting a well-designed, safe sho or trip) that is suitable to be worn outside the home (continued).

(Continued on next page)

Shoe Component	Recommended Guidelines
• Shoe closure	• Lace using cotton laces and have a minimum of 4-5 eyelets with space remaining between the lace stays to allow for expanding foot volumes, proper fit and reduced dorsal pressures, thereby keeping the foot firm against the heel counter and preventing the forefoot from rubbing during walking. • If impaired hand or wrist function, visual loss or inability to reach down to tie the laces, eliminate manipulation of cotton laces with elastic laces which remain tied; Velcro flaps, which can be managed with gross movement of the hand, opposite foot, or cane; or zippers, to reduce the likelihood of slips and trips resulting from trailing laces or a loose fitting shoe. Ensure when using other closure types that the shoe is held snugly to the foot. • In extreme cases, slip-on shoes may be worn, as they are easy to don and doff and may provide a similar fit to shoes with closures that are incorrectly fastened. However, care needs to be taken to ensure these shoe types do not cause the wearer's heel to withdraw from the shoe, requiring increased toe flexion during the swing phase of gait to retain the shoe on the foot, predisposing the wearer to slips, trips and falls.
• Heel counter	• Firm to stabilise the rearfoot and ankle; particularly important for the excessive pronator.
• Shoe collar	• High to provide mechanical stability to the ankle such that the movements of inversion and eversion, unable to be controlled by weak musculature surrounding the ankle, are restricted by the shoe. • High to provide greater surrounding pressure on the skin to enhance firing of cutaneous receptors and joint mechanoreceptors, improving proprioceptive feedback of ankle position and resulting in earlier and enhanced muscle stimulation and improved balance as measured by body sway and coordinated stability tasks. • Low for older people who experience problems with donning and doffing their shoes, and to prevent oedema. • Padded to improve shoe fit against the heel and around the ankle, as well as assist in pain relief and further comfort for the older foot.
• Insole	• Required to assist in shock attenuation, perspiration absorption and hypersensitive/insensitive foot protection • Material should be covered to decrease friction between the insole board and the plantar s leading to shoes that are more comfortable. • Individually shaped to comply with the special demands of specific foot deformities and padded to manage areas of high plantar pressure, both in aid of increasing wearer comfort and reducing injury risk.

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Table 2.2: Recommended guidelines to assist older people when selecting a well-designed, safe sho or trip) that is suitable to be worn outside the home (continued).

(Continued on next page)

Literature Review

Shoe Component	Recommended Guidelines

Sole	
• Midsole	• Present to provide foot stability and attenuate the peak forces at initial foot-ground contact to enhance the level of comfort and pain relief provided to the wearer, in particular those affected by degenerative joint changes, plantar foot pain or deformed feet. • Soft and thick to offer greater shock absorption and perhaps prevent oedema. • Thin and hard to provide more stability, due to the reduced distance from the support surface to the foot and greater proprioceptive feedback to the wearer.
• Outsole	• Synthetic, thick and durable to cushion impact, protect the foot against irregularities in terrain, while still able to flex at the metatarsophalangeal joints during late stance for comfort of fit and reduced need to expend extra energy to flex the shoe against additional resistance, which contributes to local muscle fatigue and injury. • Non-uniform sole pattern that presents good frictional properties when walking on typical surfaces in an individual's environment with more abrasive-resistant materials placed in areas of high wear or in areas where greater traction is necessary; yet not too much traction, particularly for Parkinson's patients who require movement facilitation, as excessive traction may result in foot fixation and, due to excessive torque, may contribute to joint injuries and falls.Patterned, bevelled sole to provide suction gripping on wet surfaces and assist in slip resistance of shoes by increasing the surface contact area at heel strike.Hard to increase the available tread pattern after usage by increasing the resistance to wear and prolonging the service life of shoes by slowing deterioration of the tread. • Slipping protection at the back of the heel to prevent slipping when body weight is transferred to the heel during straight line walking. • Rubber, as these soles generally have the largest dynamic coefficient of friction in both dry and wet conditions.
• Heel	• Low (< 2.5 cm) and broad to increase the base of support and thereby stability, reducing the risk of a fall as well as to decrease the stress applied to the forefoot compared to higher heels, in turn, decreasing the incidence of foot pain, discomfort and/or foot problems.

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2.4.3 Safe Indoor Shoes: The Slipper Dilemma

There is a multitude of research conducted on shoes that are worn outside the home (see Table 2.2), such that there are many recommendations as to the design characteristics that should be included in these shoes types so that they are safe for older people. However, there is very little specific information available on shoes that are worn inside the home, shoes that are traditionally worn for comfort, warmth and donning ease. Therefore, before adequate recommendations with respect to what design characteristics should be included in indoor shoes can be made, the available literature pertaining to these household shoe types, in conjunction with how they may predispose older individuals to slips, should be assessed.

Shoe comfort and/or fit is attained when shoe shape is matched to foot shape[26,275,436,455,466,467,485,491]. The average older foot increases both in width, as the transverse arch flattens with age[19], and in length by at least half a size with lifetime weight bearing[26,436]. Furthermore, the foot may increase approximately 4% in volume from morning to night due to daily weight bearing[26]. Therefore, as the last[*] determines the shoe's shape, size, inner volume, fit, style and specific stress points during locomotion, last manufacture is the critical element in constructing comfortable footwear for older people[438,455,467,485,492-494]. Unfortunately, variations in the needs of the older foot suggest that one last shape will not represent the human foot closely enough to satisfy comfort of fit requirements[455,491]. Therefore, older persons, particularly those with foot problems and deformities, often have difficulty finding mass-produced shoes to fit their feet[21] and may need to have their shoes custom-made from a cast of their foot. This increased expense may be warranted to achieve shoe comfort in the short term but does not guarantee reduced pain upon walking over extended time periods[26]. As a result, shoes such as slippers are frequently the most popular choice of household footwear as their lack of structure can mould to the shape of any foot, particularly the changing shape of the older foot.

[*] The last is a three-dimensional wooden or plastic form shaped to represent the general outline of the foot, being made from a minimum of 30 measurements[439] about which the upper material is then stretched and stitched or glued to the sole of the shoe, giving it permanent shape[456].

Slippers generally lack rigidity in the sole and will therefore easily flex at the metatarsophalangeal joints, facilitating normal foot function in activities such as walking[307,438,465]. Slippers also traditionally have hard, thin soles made of rubber or synthetic materials, which research shows may assist stability for older people (see Table 2.2). However, slippers are commonly described as old, “sloppy”, ill-fitting shoes with slippery soles, which provide an insecure base for gait, particularly for the older individual who is unable to grasp with their toes to keep the slipper on their foot during the swing phase of gait. Finlay[19] found that toe deformities were common among individuals who wore slippers because slippers compelled the wearer to grasp the footwear with clenched toes to keep it from falling off the foot during the swing phase of gait. As a result, slippers may alter gait patterns in older people, predisposing them to slips, trips and falls. Consequently, older people are usually discouraged from wearing slippers[21].

Although guidelines are available to assist older people on how to select a well-designed, safe shoe suitable to be worn outside the home (see Table 2.2), only two articles were located which looked at specific design characteristics of a safe shoe to be worn inside the home. Robinson[255] suggested older people should use “Ugg” or sheepskin boots in winter and sandals in summer, but gave no reasons as to why these shoe types should be worn. Finlay[19] suggested that slippers were only acceptable if they had adequate fastenings and firm heel counters. The recommendations of these two authors appear to contradict each other as sheepskin boots rarely have fastenings or firm heel counters and neither study appeared to take into account changes that occur as a result of ageing in both the older foot and the older individual (see Section 2.3.1). However, no other specific published guidelines for the design of safe household footwear were located.

It has been reported that a shoe could be considered safe for the older wearer if it was comfortable[495]. Therefore, regardless of concerns pertaining to slipper safety, slippers are still frequently worn, possibly because they provide comfort to the wearer and, as such, may be considered safe for older people. The fact that, despite educational campaigns discouraging slipper use, slippers

are still being worn was recognised by Rubenstein[11] who recommended that older people need to "obtain safe slippers" [(p 267)]. The need to wear safe shoes throughout the day should be emphasised because most falls, particularly in older women, occur in the home when persons are engaged in their usual activities such as walking or changing position while wearing household shoes[68,382,496]. Therefore, greater attention needs to be paid to the design of shoes for indoor use, particularly safe slippers, as they are frequently the household shoes of choice. It is also important that the needs of older individuals are understood before these design characteristics can be formulated, in particular those older individuals who may have to modify their already altered gait pattern to compensate for wearing household shoes to prevent a slip, particularly at initial foot-ground contact.

2.5 The Control of Initial Foot-Ground Contact

Walking results when the total body centre of gravity is continuously moved both outside and inside a changing base of support in a pendulum type action[334]. When one considers: a) the small base of the support; b) that 60% of the body's mass is located approximately 60% above the ground; c) that 80% of the gait cycle is spent supported on only one limb[344,497]; and d) that the toe of the swinging leg clears the ground by less than 1 cm as it travels at its highest forward velocity[497], it is easy to see why walking has been thought of as "a unique activity during which the body, step by step, teeters on the edge of catastrophe" [498 (p 56)]. However, before reviewing the strategies by which slips during walking can be prevented, the gait cycle must firstly be understood.

2.5.1 The Gait Cycle

The main events occurring during the gait cycle, time normalised to 100%, are depicted in Figure 2.4. Slips in older people typically occur during initial foot-ground contact (see Section 2.2.3). As the present thesis is predominantly focused upon factors affecting slips in older people, the ensuing

discussion of gait will be limited to those events leading up to and including the foot of the stance limb contacting the supporting surface, namely terminal swing, initial foot-ground contact and loading response (see Figure 2.4). The reader is referred to papers by Whittle[499], Hagemen[500] and Winter[497] for further information concerning other phases of the gait cycle.

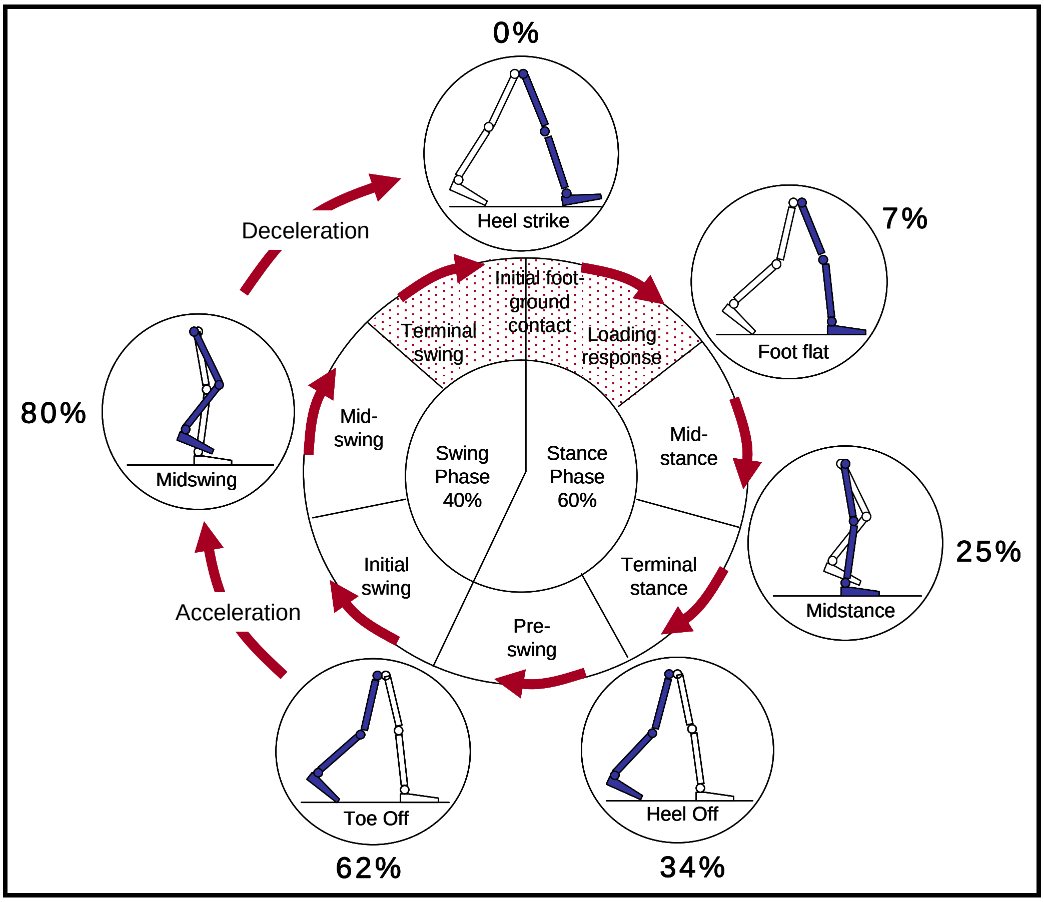

Figure 2.4: The gait cycle, normalised to 100%, consisting of events that take place between initial foot-ground contact and successive contact of the ipsilateral lower limb (events refer to motion of the blue lower limb depicted in the diagram; adapted from Inman[342] and Winter[497]). Percentage figures may vary with factors such as changes in gait speed.

2.5.2 Kinematics of Initial Foot-Ground Contact

Initial foot-ground contact, the critical event in the gait cycle during which most slips occur, traditionally begins as the heel of the leading limb

contacts the supporting surface (see Figure 2.4). However, older people, particularly those with abnormal gait patterns, may contact the supporting surface with the whole foot or with the toes rather than the heel. In healthy individuals, as the knee is extended in preparation for initial foot-ground contact, the ankle is in a neutral position and the shoe-surface angle is between 10° and 35° [18,86,399,411,501-503]. Loading of up to 90% of body weight[399] takes place at initial foot-ground contact, typically via the small surface area of the heel. This places the individual in a precarious position such that if they were to contact an unexpected surface lubricant, a slip would be inevitable. However, slips and falls have been recorded when vertical force levels during weight bearing are anywhere between 35% and 90% of body weight[399] and shoe-surface angles are between 5° and 30° [87,399,504]. Consequently, contacting the supporting surface with a flatter foot, or transferring a greater percentage of body weight to the contacting lower limb does not appear to always prevent a slip.

Heel velocity at initial foot-ground contact for healthy individuals has been reported to be between 10 $cm.s^{-1}$ and 20 $cm.s^{-1}$ [399], having been decelerated from approximately 100 $cm.s^{-1}$ during the preceding swing phase[18]. However, Chaffin *et al.*[90], using a pull-sled test rig, reported that heel velocities could exceed 50 $cm.s^{-1}$ to 100 $cm.s^{-1}$ and studies analysing the conditions under which slips occur report highly variable heel velocities, increasing above 40 $cm.s^{-1}$ or 50 $cm.s^{-1}$ [399]. Furthermore, Winter[497] found that, despite slower gait velocities, older subjects had significantly higher heel contact skid velocities than younger subjects and, as such, were at an increased risk of incurring a slip-induced fall. Despite individuals contracting the lower limb muscles in an attempt to arrest forward motion of the foot at initial foot-ground contact (see Section 2.5.4), many studies report that the foot is still moving when it impacts the supporting surface[90,399,406,503], such that there is foot slide. For example, in their review of the biomechanics of gait relevant to slips, most of which has been completed using young, healthy subjects, Redfern and colleagues[502] found that, although heel velocity was directed forwards at impact, it then either came to a stop or reversed its sliding direction before ceasing its motion. Cham & Redfern[503,505] reported a significant number of walking trials where the impact velocity of the

heel was negative, such that the heel was moving backwards at impact. Furthermore, studies on both older individuals and their younger counterparts have reported that, although heel slide values of between 1 cm and 3 cm appear normal[411,505,506], values above these may incur slips, with a slip that results in a fall likely if the slip distance exceeds 10 cm[399,411,412].

Lower limb joint angle data displayed by both younger and older individuals during gait have been commonly reported[296,323,324,338,497,503,507-509]. In general, at initial foot-ground contact, the knee approaches full extension[510] and then flexes 15° to 20° during the loading response to assist in shock and energy absorption[497,511,512] until full body weight is accepted by the stance limb at midstance (see Figure 2.4). Knee flexion also aids in shortening the relative limb length and preventing excessive vertical translation of the total body centre of gravity[512]. Simultaneously, the ankle, which is in slight dorsiflexion at initial foot-ground contact, rapidly[*] plantar flexes 15° to 30° from approximately 10% of stance until the foot is flat on the surface[238,239,513,514]. Furthermore, the subtalar joint is supinated to provide a rigid lever for impact absorption[238,239,479] and body weight acceptance[318,514], after which pronation occurs[515], producing a flexible unit to accommodate environmental variations and to further absorb the shock of impact, until foot flatness is achieved at 15% of the cycle[238,239,514].

2.5.3 Ground Reaction Forces at Initial Foot-Ground Contact

The ground reaction forces generated by older individuals at initial foot-ground contact and loading response (see Figure 2.4) are imperative as it is at this time that the shear forces are the highest and, therefore, the frictional capabilities of the shoe/floor interface will be most important (see Figure 5.7) [87,399]. If the shear forces generated by the foot at contact are less that the opposing frictional forces the foot will not slip[90]. If, however, the shear forces generated during a particular step exceed the frictional capabilities of the shoe/floor interface, then a slip is inevitable[502].

[*] Cham & Redfern[503] reported the foot angular velocity at initial foot-ground contact to be 224 $°.s^{-1}$ for a group of 16 healthy, young individuals aged 19 to 30 years.

Several researchers have examined the ground reaction forces generated by older individuals during normal gait on a level surface[308,317,327,497,500,516]. Similar to their younger counterparts, upon initial foot-ground contact a high frequency impact force peak is initiated that occurs in all components of the ground reaction force trace during the first 50 ms of ground contact[517-519]. This impact peak propagates across the joints of the ankle, knee and hip and into the spine[520] and must be attenuated by the body's anatomical structures, such as the calcaneal heel pad, the cartilage and subchondral bone at the joints, the tissues of the lower limb, together with coordinated motion of the lower limb (see Section 2.5.2) and learned anticipatory muscular actions[520] (see Section 2.5.4). In healthy individuals 50% to 90% of the heel strike impact is attenuated via these structures and mechanisms operating at the knee[490,521-523]. However, if the efficiency of these natural shock absorption mechanisms is reduced because of musculoskeletal disease[490], an individual will experience pain and adapt their gait to elicit pain-avoidance gait strategies, perhaps increasing their falls risk (see Section 2.3.1(F) and Section 2.6.3). Therefore, augmenting an individual's anatomical resources by providing external shock absorbing materials, such as those commonly seen in footwear, may be beneficial for individuals with musculoskeletal disease[520].

Researchers have speculated that the anteroposterior shear force component of the impact peak may indicate heel slide that occurs as the foot impacts the supporting surface[411,524] (see Section 2.5.2). A second anteroposterior shear force peak also reportedly occurs in the opposite direction to the initial peak during the loading response[86,411,420] and is thought to be due to rearward movement of the heel during the early loading phase[420], a finding consistent with the kinematic results indicating heel slide (see Figure 5.7 and Section 2.5.2).

At the end of the loading phase, at approximately 25% of stance[502,503], the normal ground reaction forces then peak[*] as body weight is transferred to the

[*] The first force peak of the vertical ground reaction force for normal gait is typically between 0.75 times body weight to 1.80 times body weight for older individuals[327,517].

supporting foot. Similarly, the anteroposterior shear forces peak* at approximately 19% of stance, or 90 ms to 150 ms after initial foot-ground contact[502,503]. These ground reaction force variables may[525,526] or may not change with altered surface types[503], although the variables appear to be influenced by changes in footwear[524,527-529] and the resulting shoe-surface interaction (see Section 2.3.2(C)).

2.5.4 Neuromuscular Control of Initial Foot-Ground Contact

Although the quadriceps muscles work concentrically to extend the knee for initial foot-ground contact, deceleration of the swinging lower limb at initial foot-ground contact is achieved by eccentric contraction of the hamstring muscles, namely semimembranosus, semitendinosus and biceps femoris[335,530-533]. Consequently, delayed and/or reduced hamstring muscle activation during this critical period is associated with increased horizontal heel contact skid velocity[534].

Following initial foot-ground contact, the hamstring and quadriceps muscles co-contract throughout the loading response, allowing knee flexion for shock absorption but maintaining lower limb stability[335,348,512,535] and consequent balance control[348]. However, at between 5% and 10% of stance, the knee extensors provide the dominant activity responsible for slowing knee flexion as full weight bearing takes place[512,536,537], assisting to shorten relative limb length and prevent excessive vertical translation of the total body centre of gravity[512].

This model of knee flexor and extensor muscle activity during gait described above is confirmed by knee moment profiles during the loading response which move from a knee flexor moment just prior to and at initial foot-ground contact, into a knee extensor moment during the loading response[352,497,502,537,538]. Therefore, as the knee is flexing under the control of the knee extensors, this leads to the first major energy absorption phase of gait[497]. Reduced knee moments have been noted when individuals encounter

* The first force peak of the anteroposterior ground reaction force for normal gait is typically between 0.24 times body weight to 0.50 times body weight for older individuals[327,517].

slippery surfaces[538], perhaps reflecting greater co-contraction between the knee flexors and extensors to allow the individual to adapt to the more “dangerous” situation.

With respect to ankle and foot motion control, the ankle dorsiflexor muscles play a major role at initial foot-ground contact, changing their concentric action, which was required to ensure adequate foot clearance during swing, to an isometric contraction which allows correct ankle alignment for impact[514,539]. Then, during the loading response, the ankle dorsiflexor muscles eccentrically control plantar flexion of the foot as it rapidly rotates to the foot flat position[238,239,514,539]. Tibialis anterior also assists to control foot pronation[238,239,514] and prevent buckling of the stance limb[335,535]. The ankle dorsiflexor muscles are assisted in their action by the peroneal muscles[238], which also control the amount of ankle dorsiflexion and supination at initial foot-ground contact until approximately 10% of the stride[539,540]. Furthermore, as the peroneal muscles are primarily pronators of the foot, being predominantly active during stance[540], these muscles may also assist to control medial to lateral balance during walking[540,541]. In contrast to the synergistic action of tibialis anterior and peroneus longus, the triceps surae muscle group are not active until foot flat[535], displaying eccentric behaviour as the body rotates forward over the stance limb before reaching peak concentric activity just before pre-swing. Therefore, the role of the ankle plantar flexors is to help support the knee, decelerate tibial advancement and then plantar flex the foot during terminal stance and preswing.

Moment profiles of the ankle, confirm that the ankle dorsiflexor muscles are dominant in controlling the plantar flexion required at initial foot-ground contact, assisting to absorb the energy of impact[327,497,502]. Furthermore, similar to the knee moments, ankle moments have been found to be reduced when individuals encounter slippery surfaces, with additional reductions seen in the incidence of an actual fall event[538]. However, at foot flat the dorsiflexor moment rapidly becomes a plantar flexor moment[327,497,502] to assist in energy absorption during stance, increasing in magnitude until the ankle plantar flexes

at terminal stance and preswing. It is at this point, when the plantar flexor muscles are controlling ankle plantar flexion that "the single most important energy generation phase" [497 (p 45)] results, which is responsible for 80% to 85% of the power generated across the entire gait cycle[345]. Therefore, although the plantar flexor muscles have a very small role at initial foot-ground contact, they provide most of the power for walking[238].

Initial foot-ground contact is a critical event for slip-related falls in older people. Successful contact during gait arises from adequate deceleration of the swinging lower limb so that its velocity nears zero at impact[399,406,501,530] combined with precise foot alignment[238] onto the supporting surface. As successful foot-ground contact is dependent upon an individual's functional capabilities, the incidence of disease or deformity as well as the shoes being worn and the surface being traversed, individual perceptions of the surrounding environment become increasingly important[542] in order to detect and avoid potentially hazardous situations prior to actual contact[543] (see Section 2.3.2). Furthermore, gait alterations that significantly reduce the frictional forces required at contact[505], will reduce this slip risk further. It is therefore necessary to investigate how older individuals detect and respond to changes in the frictional demands of different footwear-surface interactions to reduce their risk of slips at initial foot-ground contact, particularly those with musculoskeletal disease such as rheumatoid arthritis.

2.6 Rheumatoid Arthritis

Rheumatoid arthritis (RA) is a chronic inflammatory articular and systemic disorder[116,118,544-546] which was first recognised in the mid 18th century[546,547]. A common disease, RA affects approximately 1% of the population worldwide[479,547-551] with an annual incidence of between two and four people per every 10,000 adults[548,549]. In gender terms, RA affects women two to three times more often than men during the typical years of onset, which are between 20 and 60 years of age[116,118,216,479,545-547,549,552]. There is, however, a general increase in prevalence for both sexes[545,549] with some suggestion of a declining sex ratio with advanced age[545,547,553]. Regardless, women

appear to fare much worse than men to the extent that sex hormones have been strongly implicated in the aetiopathogenesis of RA[554]. These implications have arisen as the peak age of onset in women is around the time of menopause when they are seemingly no longer protected by the effects of female hormones such as oestrogen, oral contraceptives and pregnancy, or are yet to be involved in oestrogen replacement therapy[216,547,554]. Furthermore, women require surgical intervention in response to their RA more frequently than men, suggesting that women could be less resistant to tissue damage than men such that a lower degree of inflammation would destroy their joint architecture and lead to malalignment[555].

The aetiologic factor responsible for RA remains largely unknown. However, based on the level of disease among monozygotic twins and an inflated prevalence among highly inbred populations[547], there appears to be a hereditary predisposition to RA[216,556]. In fact, genetic factors have been attributed to cause between 20% and 30% of RA cases with the rest of the cause remaining unexplained[216,547]. The aetiology of RA has implicated endocrine, metabolic and nutritional factors as well as geographic, occupational and psychosocial variables[546,547]. However, none of these factors have been found responsible for RA incidence, instead appearing to influence the course of the disease[546,547].

The progression of RA would suggest that the aetiologic factor responsible is carried by the circulatory system, as the disease affects not only the joints but also has many extra-articular features[546]. Rheumatoid (Rh) factor, an immunoglobin that circulates in the blood serum, has been found in approximately 80% of RA patients and was therefore thought to be causative[549]. However, as Rh factor has also been found in unaffected individuals and some RA patients have an absence of Rh factor, this theory has been refuted[546]. Despite this, RA patients usually have more Rh factor than unaffected individuals[549] or those with other diseases. Furthermore, high amounts of Rh factor are generally associated with a more severe and active joint disease, the presence of nodules, a greater frequency of systemic complications of RA and a poorer outcome[557,558]. Consequently, positive tests for Rh factor form the basis of RA diagnosis with further diagnosis based on radiographic findings of bone demineralisation and erosion around affected joints.

The clinical course of RA is highly inconsistent, ranging from a mild illness of brief duration to a relentlessly progressive, destructive polyarthritis associated with systemic vasculitis[116,547,550,559-561]. However, due to age-related alterations in the normal immune response[562], the individual course that RA follows for each patient cannot be predicted at disease onset[563]. The most common mode of onset is the insidious development of symptoms over a period of several weeks[549], frequently heralded by pain, swelling, fatigue, weight loss, weakness and generalised aching and stiffness[547] but with little deformity[116,549]. As the disease progresses, brief remittent episodes of articular involvement may occur before persistent arthritis develops[547]. Joint deformity appears because of the disease's inflammatory processes, namely those of synovitis and pannus formation. Synovitis or inflammation of the synovium and changes in synovial fluid composition compromise lubrication and permeability of cartilage leading to cartilaginous destruction and increased joint laxity[550,564-566]. Pannus, an abnormal tissue growth consisting of proliferating fibroblasts, small blood vessels and inflammatory cells, forms on the synovial membrane of affected joints, expanding to invade the joint[564,565]. Both synovitis and pannus gradually destroy the joint surface and underlying bone structure[565,567], weaken the joint capsule and surrounding ligaments, and cause inflammatory changes in tendons, tendon sheaths and often skeletal muscle[116,545,549,568]. It is these pathological changes, in conjunction with the mechanical forces of weight bearing and muscular action, which produce the characteristic deformities of RA[547] and, ultimately, varied degrees of incapacitation[547,564,569]. Joint position is often dictated by comfort and the patient's attempt to avoid pain by posturing joints in least painful positions[564]. However, these joint positions are often far from ideal in terms of achieving efficient joint action and can therefore exacerbate joint degeneration[116].

Chronic RA has assumed its more characteristic clinical features by the end of 1 year to 2 years having a bilateral, symmetrical and polyarticular pattern of joint involvement[116,564]. Several studies have indicated that radiographic erosion, joint space narrowing and structural damage are seen in more than 60% of RA patients within the first 2 years of disease[549,561,569-571]. However, radiographic malalignment is not usually seen before 5 years of the disease[569]. During the subsequent 10 years, radiographic progression[569], deterioration in functional status[116,544,545,547,549,559,571-573], an inability to

perform daily living activities[545,562,574], progressive work disability[572] and early death[575,576] characterise the disease in most patients.

As a systemic disease, RA affected individuals may also have associated extra-articular features involving several other tissues[547]. For example, rheumatoid subcutaneous nodules are a common feature, appearing at some time in approximately 20% to 25% of RA patients. Nodules are lumps of tissue that form under the skin, often over bony areas and usually at nodes proximal to inflamed joints, although they may also be present in areas remote from articular inflammation such as the internal organs. Furthermore, RA patients may develop inflammation of the membrane that lines the heart, lungs and spleen, as well as inflammation of the lung tissue itself. Therefore, there are often complications of cardiac lesions, respiratory symptoms, ocular abnormalities and anaemia, as well as peripheral and autonomic sensory neuropathy[577-580]. These pathological changes may occur in conjunction with loss of appetite, weight loss, fever, pain, fatigue and fragile skin that is easily torn and bruised[216]. General morning stiffness lasting more than 30 minutes, and frequently several hours, is another characteristic of RA and is thought to result from congestion in the synovium, joint capsule thickening and, frequently, from an increase in the volume of synovial fluid[216,546,547].

Characteristic damage to articular and periarticular structures often results in pain and discomfort, joint stiffness, limited motion, inflammation, muscle atrophy, reduced muscle strength and ensuing functional limitations[116,179,544,545,559,571,581,582]. This damage and the associated complications disrupt the biomechanics of the joints, alter the pattern of joint loading[116] and increase energy expenditure when performing daily living activities. Consequently, the RA patient may develop balance difficulties[546,582,583] and unsteady gait (see Section 2.6.3) with an increased risk of falls, further aggravating the symptoms of arthritis and impairing coordinated joint movement[116,583]. Deconditioning due to lack of physical exercise can also lead to a vicious cycle whereby joint stiffness and decreased strength lead to less activity which, in turn, compounds musculoskeletal weakness and stiffness[179] and poor cardiopulmonary fitness[544,584]. Therefore, activity maintenance can prove beneficial for RA patients in terms of increasing physical and functional capacity[116,585,586], decreasing pain[586], improving psychosocial factors[585,586] and even decreasing radiological progression[586]. For these

reasons, impediments to RA patients participating in physical activity must be identified and removed, particularly impediments from lower limb dysfunction.

2.6.1 Rheumatoid Arthritis and the Lower Limb

Rheumatoid arthritis appears to affect smaller joints earlier than larger joints and lower limb joints earlier than upper limb joints[550,560]. Therefore, the joints of the hands, the wrists, knees and feet are commonly involved early in the disease process. Then, as the disease becomes established, the arthritis spreads to the elbows, shoulders, sternoclavicular joints, hips, ankles and subtalar joints[279,546,569]. Although both vulnerable, the foot is more commonly the initial site of RA compared to the hand (15.7% versus 14.7%, respectively)[466,587] with foot problems being only second to those of the knee during the later stages of the disease[279]. In a random survey of 995 RA patients, Vainio[588] reported 91% of women and 85% of men had significant foot and ankle problems. Similarly, Michelson *et al.*[550] reported a prevalence of significant foot and ankle complaints in 94% of 99 RA outpatients. However, the frequency and degree of foot and ankle problems appear directly proportional to the severity and duration of the disease[479,549,567,588]. That is, Michelson *et al.*[550] found that only 55% of patients reporting RA for less than 10 years displayed foot and ankle symptoms compared to 76% of patients who reported RA for more than 20 years.

The forefoot is one of the most frequently affected areas of the foot in RA patients[466,479,549,560,588-591], with virtually all RA patients displaying some radiological evidence of metatarsophalangeal joint erosion within 1 to 3 years of incurring the disease[279,592]. Synovitis renders the supporting ligamentous and capsular structures incapable of stabilising the metatarsophalangeal joints[279,587,591,593,594] leading to hypermobility, a flattened transverse arch, splaying of the forefoot and a dorsal migration of the proximal phalanges[216,279,466,560,587,593-596] (see Figure 2.5). In addition, proteolytic enzymes and pannus formation progressively destroy the articular cartilage and subchondral bone[279,587,591,593,594]. Therefore, weight bearing, together with muscular contractions, pull on the unstable joints resulting in dorsal subluxation

and ensuing dislocation of the metatarsophalangeal joints with accompanying toe deformities[216,479,560,587,590-592,594,596] (see Figure 2.5).

These forefoot deformities characteristic of the RA patient, together with anterior migration and atrophy of the fibrofatty pads (see Figure 2.2) that normally bear weight under the metatarsal heads, expose already inflamed metatarsophalangeal joints to a less protected position adjacent to the ground. Therefore, the skin under the metatarsal heads is overloaded giving rise to pain[466,560,591,593] such that patients often describe that they feel as if they are "walking on stones" [593 (p 87)] as one of their symptoms. This has been confirmed by studies that report consistently higher peak plantar pressures across the metatarsal heads in RA patients compared to normal values[592], values sometimes exceeding two to three times that of normal subjects[590,597]. As a protective response to these high plantar pressures, skin callosities, bursae and rheumatoid

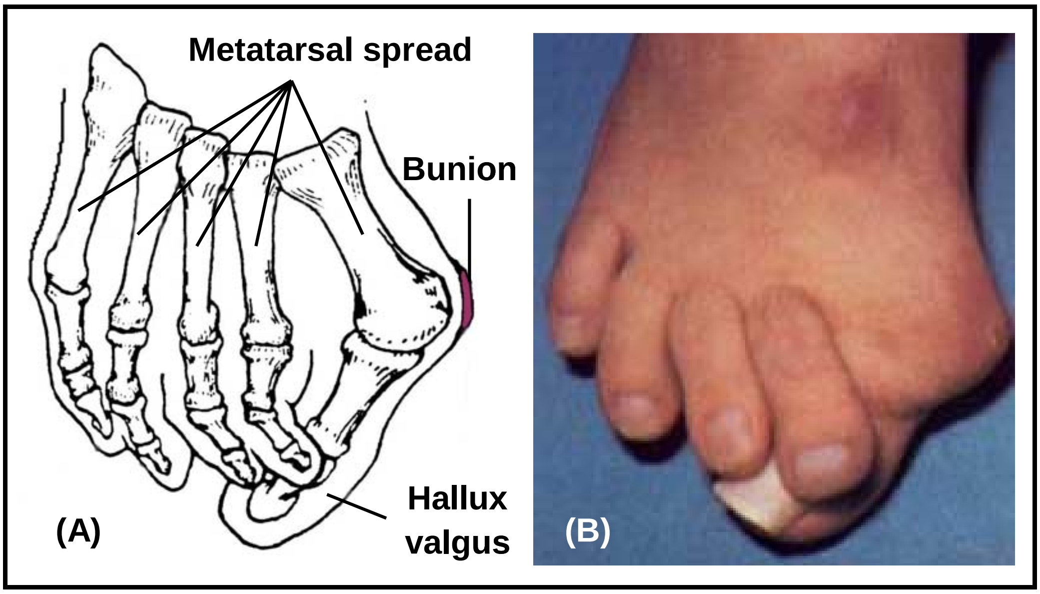

Figure 2.5: Forefoot involvement in RA indicating forefoot splay with associated hallux valgus and bunion deformities resulting in dorsal subluxation and dislocation of the metatarsophalangeal joints in both diagrammatic (A) and photographic (B) forms (adapted from Smyth & Janson[560]).

nodules often develop beneath the metatarsal heads on the plantar surface of the foot[479,560,591]. However, in time, these protective responses may actually further increase local pressures, exacerbating the pain symptomatic of the RA foot[592].

In addition to the forefoot deformities, rearfoot synovitis associated with RA can lead to subsequent weakness and laxity in supporting soft tissues, cartilage and bone destruction, as well as inflammation of the posterior tibial, peroneal and Achilles tendon sheaths[303,479,549,560,567,588,596,598]. These symptoms destabilise the talonavicular joint, in turn, causing the talar head to drift medially and inferiorly[599,600], leading to excessive pronation at the subtalar joint and flattening of the longitudinal arch[303,479,549,567,591,594,596,598]. Consequently, the midfoot and forefoot drift into abduction and the calcaneus drifts into valgus[598], resulting in a rearfoot valgus deformity that is seen in between 30% and 80% of patients with rearfoot involvement[303,479,592,596,598]. Valgus deformity may place high pressures on the medial heel and affect functioning of the posterior tibial tendons as medial stabilisers of the rearfoot[598].

Midfoot, rearfoot and ankle involvement tends to occur between 5 years and 10 years post RA onset[279,550,560,567,588,598]. Their involvement can redistribute the load medially to the forefoot, altering the normal pressure distribution during gait and heightening the medial peak pressures[592]. Furthermore, incorrect and increased mechanical loading within the ankle and knee joints during gait[116,303,466,479,549,594,598], as well as reduced ankle dorsi- and plantar flexion and subtalar inversion and eversion[594], may result. Therefore, as RA disease duration increases, it is deformities of the rearfoot, as opposed to the initial forefoot deformities, that begin to determine function and disability in other joints, particularly the knee and ankle[567].

Involvement of the knee as a result of RA occurs in a similar manner to other joints but is exacerbated by foot and ankle malalignment and altered mechanical stresses. Inflammation, synovitis as well as cartilage and subchondral bone loss at the knee lead to a valgus-flexion knee deformity[595] with external knee rotation also occurring in severe cases. Therefore, these biplanar or triplanar knee joint deformities further contribute to excessive pronation of the rearfoot[595]. Destruction of the collateral ligaments and cartilage

may cause the knee joint to become fixed in a deformed or "altered" position[595]. Atrophy of the quadriceps muscles and hamstring muscle spasms often follow, producing flexion contractures and limited knee joint ranges of motion which, in turn, increase the risk of falls[595].

Foot involvement typically signals the onset of ankle, knee and hip deformities due to the altered loading patterns that occur at these joints as a result of RA[279,479]. As a consequence, RA alters the mechanical linkages between the lower limb joints, producing significant pain, deformity, joint stiffness, abnormal weight bearing and gait disturbance[116,216,279,560,589,595,598,601]. If left untreated, foot problems may lead to a decline in ambulatory ability, predisposing the individual to a sedentary lifestyle, and an increased risk of osteoporosis and cardiovascular disease[279]. In fact, Cimino & O'Malley[549] found that, even in those individuals with little or no involvement of the hip and knee joints, foot involvement seriously limited functional capacity. Therefore, non-operative foot care that prevents or manages lower limb involvement, including appropriate footwear, is critical in preserving joint function for RA patients and, in turn, enabling them to remain mobile and physically active[479,601]. However, when non-operative techniques are no longer effective and patients have significant deformity, surgical intervention[*] is the only way of restoring functional integrity to the affected joints and reducing pain[479,516,593,595,602]. Although usually successful, there is risk and pain associated with any surgery, and the results of any surgery may only be successful in the short term[380]. Therefore, non-operative interventions, such as appropriate footwear and foot care, are preferred to allow the RA patient to maintain an active independent lifestyle[380,601,603].

[*] Surgical procedures include ankle fusion, between the tibia and talus; subtalar joint fusion, between the talus and calcaneus; triple arthrodesis, a fusion of the subtalar, talonavicular and calcaneocuboid joints; and forefoot reconstruction, which excises the metatarsal heads to place the joints in a more aligned position[480].

2.6.2 Footwear and the Rheumatoid Arthritic Foot

The progressive lower limb deformities that result from RA (see Section 2.6.1) can lead to marked limitations in shoe wear[466,604]. As such, RA patients are encouraged to wear shoes that are comfortable, supportive to hypermobile joints and accommodating to foot deformities, while avoiding the temptation to sacrifice comfort for style[242,479,596,598]. Early in the disease process, standard shoes can be worn by RA sufferers. However, the insidious nature of foot and ankle deformities characteristic of RA change the shape of the feet of patients, creating areas of high pressure and pain[590], such that standard shoe types no longer fulfil the requirements of a comfortable, supportive shoe. In a study of 200 RA patients, Vidigal *et al.*[590] found that only 40% were able to buy shoes that they deemed comfortable at an ordinary shoe shop. Consequently, once deformity occurs, it is more difficult to obtain mass-produced footwear sold at retail stores, necessitating the RA patient to visit health professionals to obtain orthoses and/or surgical footwear[216,587].

Surgical shoes, with wide and deep toe boxes, moulded insoles and sole wedges; and rigid or flexible orthoses are designed to cushion the foot as well as to alter foot mechanics to redistribute loads within the joints[479,596,605]. Such shoes thereby provide considerable relief even when an individual's feet are extremely deformed[216,250,587]. Fransen & Edmonds[606] found that, compared to control subjects, RA patients who had been fitted with extra-depth orthopaedic footwear demonstrated significant improvements in weight-bearing pain scores, physical function, gait velocity and stride length without an increase in the use of arthritis medications and walking aids. Similarly, Budiman *et al.*[607] reported that the progression of hallux valgus deformity was 73% less likely to occur in 41 patients with RA who were treated with rigid functional foot orthoses. Shrader & Siegel[608] described a 73 year old woman with longstanding RA, who was given foot orthoses together with modified shoes. After 6 weeks this patient displayed a more symmetrical gait pattern with an increased stride length, meaning that for a given number of steps per minute, the patient could walk a greater distance, increasing her walking speed by 33%[608]. However, with further pain and deformity, orthoses may not provide adequate relief and custom-made

footwear may be required to improve ambulation and reduce the need for surgery[466,590].

Custom-made footwear are moulded from a cast of an individual's foot, incorporating changes to existing parts of the shoe or integrating new elements so that the shoe conforms to the shape of the individual's foot. Therefore, these shoe types can distribute pressure evenly over the whole foot, relieving pressure from deformed joints as well as providing pain relief and joint stability[24,242,466]. However, for long term pain relief, these shoes have to be frequently inspected and continually modified as the foot continues to change shape[216,587]. Consequently, a considerable proportion of custom-made shoes are never worn by the patient because they fail to meet expectations of comfort, appearance and function[216,479,609] for such high cost[*]. Instead, RA patients tend to make their own modifications to their footwear or find shoes that provide sufficient pain relief to enable them to be mobile[590].

Possible footwear solutions for common RA foot problems (see Section 2.6.2) are described in Table 2.3. However, as this population is at a high risk of falls, where possible, footwear for the older RA patient should also adhere to the guidelines documented for safe footwear in Table 2.2, although this may not always be possible. For example, a flexible sole facilitates foot motion; however, in the RA patient a flexible sole may place more pressure on already painful metatarsal heads during walking. In this situation a rocker sole, which has a curved sole, facilitates the rollover process during gait but because the sole is rigid, it also decreases pressure on painful joints[610].

As appropriate footwear, together with functional orthoses, may delay or preclude the need for surgical intervention to restore function to the joints of the lower limb[587], both researchers and clinicians have advocated proper shoe wear to be the most important aspect of non-operative care in RA patients[279,380,466,590,596,603,611]. However, a recent study of 93 RA outpatients with

[*] Custom-made shoes can retail between $400 and $750 in Australia[611].

Table 2.3: Footwear therapy recommended for the RA patient.

Problem[a]	Therapy	Footwear Solution
Pain on weight-bearing	Support unstable joints, restrict the motion of painful joints, cushion the foot and redistribute the loads within the joints for reduced pressure.	Shoes may require low heels; soft, shock absorbing soles; polyethylene or foam inserts and/or orthoses; toe slings and sleeves; felt pads and rigid metatarsal bars. Rocker bottom soles may also assist in pain reduction.
Deformed toes	Allow toes and the forefoot to spread during weight bearing to permit normal toe motion.	Shoes may require an extra wide and deep toe box to allow for possible padding for bony prominences as well as inserts or orthoses.
Excessive pressure on the metatarsal heads	Relieve forefoot pain and facilitate roll off during gait.	Shoes may require a long steel shank, external metatarsal bar or stiffer soles. Midfoot, ball and toe gait rocker soles may also be advantageous.
Rearfoot valgus and instability	Provide rearfoot stability and maintain rearfoot in neutral position on weight bearing.	Shoes may require firm, padded heel counters, outward flared heels, heel wedges, firm medial counters, longitudinal arch support as well as polyethylene or foam inserts.
Rearfoot rigidity	Cushion heel and facilitate roll off during gait.	Shoes may require cushioning in the heel together with rocker bottom soles.
Excessive perspiration, thin and fragile skin	Prevent friction of shoe against skin and provide a dry shoe environment.	Shoes may require lined, soft, pliable and lightweight uppers which conform to the foot.
Donning difficulty and upper limb pain and deformity	Facilitate donning ease for deformed foot and ensure easy to manage closures for good shoe fit.	Shoes may require Velcro closures, a wider throat opening or extended toe lacing to ensure correct toe placement.
Oedema, pain and poor foot circulation	Enable mechanism of venous pump.	Shoes may require soft and pliable soles with low collars to facilitate ankle plantar flexion and metatarsophalangeal dorsiflexion. The shoe should also fit the foot well.

[a] Table compiled from Burra & Katchis[587], Coady *et al.*[275], Cracchiolo[598], Edelstein[24], Gilchrist[242], Gould[26], Grifka[466], Janisse[612], Janisse[613], Mann & Horton[479], McPoil[438], Rubenstein *et al.*[11] and Simon *et al.*[614].

foot and/or ankle involvement found that only four patients had received footwear-related interventions, such as shoe inserts, foot orthoses, prescription footwear or shoe modifications[550]. Furthermore, Vidigal *et al*.[590] reported that only 19 out of 200 RA patients had been supplied with special shoes, with 40% of the patients wearing shoes considered unsatisfactory, such as sandals or extra large shoes, which were padded or cut to accommodate misshapen feet. Therefore, although it is recognised that correct footwear is an essential component for foot and ankle management strategies, RA patients may not prioritise footwear as a necessary therapy, instead adapting their current shoes for comfort and pain relief rather than function[590]. Alternatively, appropriate safe household footwear, which meets the needs of specific groups of the population, such as older RA patients, may not be readily available to the people who would benefit so greatly from such footwear. Inappropriate footwear may also increase the risk of falls in this population, particularly if their shoes do not fit their feet properly. Therefore, it is important to understand the basic requirements of footwear construction, design and fit for older people with RA[438,612,613], particularly with respect to how changes in footwear alter walking characteristics on different surfaces, to reduce falls in this population. However, before focusing on footwear effects, it is imperative to understand the effects of RA itself on the gait of older people.

2.6.3 Effects of Rheumatoid Arthritis on Gait

The lower limb joint deformities, instabilities and associated pain characteristic of the RA disease process (see Section 2.6.1) lead to changes in the normal walking pattern of RA patients[279,466,587], such that they often have difficulty with community ambulation[479]. In fact, if a joint deformity becomes too severe or painful, ambulation may cease[479], thereby contributing to institutionalisation and severely affecting the quality of life of RA patients[305].

A normal gait pattern is characterised by a heel-toe progression, whereby the main function of the lower limb at initial foot-ground contact is to absorb ground impact, adapt to the terrain and provide a supporting base for the body

(see Section 2.5). These roles are usually carried out by combined plantar flexion and eversion of the ankle, subtalar joints and transverse tarsal joints, enabling a flexible foot at impact[479]. However, problems common to RA patients, such as progressive forefoot deformity, collapse of the rearfoot or marked valgus deformity of the ankle joint with associated pain (see Section 2.6.1), lead the RA patient to adopt an apropulsive gait pattern, commonly referred to as the "rheumatoid shuffle" [303,587,589,594,615].

The rheumatoid shuffle is typified by a very limited or absent push off that is replaced by a lift off with reduced ground clearance, similar to a plodding gait style[479]. Delaying heel rise allows the loads on the lower limbs to be redistributed so that weight-bearing is avoided on painful joints, such as the metatarsal heads, enabling comfortable, pain-free and efficient ambulation for the RA patient[466,479,596]. Teixiera & Olney[308] suggested that subjects with osteoarthritis displayed similar pain-avoidance gait strategies to the RA patients. However, the apropulsive gait pattern may also result from weak calf muscles[303,596] and it has been suggested that increased gastrocnemius and soleus muscle activity may be required to compensate for this weakness[303]. Irrespective of the cause, reduced push-off leads to a shorter stride length, reduced single limb support, increased double limb support and a slower gait velocity[179,303,589,592,594,596,615,616]. Furthermore, the vertical and shear components of the ground reaction forces and peak pressures recorded during the stance phase of RA patients are lower compared to non-arthritic subjects[592]. These reduced ground reaction forces may be advantageous in terms of reducing the required coefficient of friction (see Section 2.5.3) and consequent slip prevention.

O'Connell *et al*.[589] studied the gait parameters displayed by 10 RA patients (mean age, 54 years; mean arthritis, 12 years) with forefoot involvement and compared them to individuals with no incidence of arthritis. When walking at a self-selected pace, the RA patients recorded gait velocities and stride lengths 71% and 77%, respectively, of the control subject values. Furthermore, RA subjects displayed significantly lower total ankle joint range of motion and late stance ankle plantar flexor muscle moments (24° and 1.13 N.m/BW.foot length,

respectively) compared to the control subjects (39° and 1.43 N.m/BW.foot length, respectively). Furthermore, heel rise was delayed in the RA patients until 70% of stance compared to 53% of stance in control subjects[589]. Therefore, whereas the control subjects were undergoing rapid ankle plantar flexion in late stance, the RA subjects maintained ankle dorsiflexion, again displaying the characteristic apropulsive gait pattern.

Similar results were reported by Keenan *et al.*[303] who analysed the gait of 20 RA patients either with normal rearfoot alignment (mean age, 63 years; mean arthritis, 14 years) or clinical evidence of a valgus rearfoot deformity (mean age, 60 years; mean arthritis, 25 years). These authors found that, independent of rearfoot deformity, none of the RA patients used all of their available ankle plantar flexion range of motion while walking[303]. Instead, compared to non-arthritic subjects[244], RA patients recorded significantly increased ankle dorsiflexion combined with limited ankle plantar flexion. Based on this result, the authors speculated that the RA patients failed to roll over the forefoot during late stance, postponing heel rise and diminishing walking efficiency, stride length and gait velocity[303]. Rearfoot valgus deformities have been found to emphasise these effects[303,596].

In addition to delaying heel rise during gait to protect the feet, RA patients with painful heels often avoid initially contacting the ground with the heel[596]. Katoh *et al.*[617] demonstrated that RA subjects with painful heels demonstrated reduced vertical and anterior shear ground reaction forces at initial foot-ground contact. The authors concluded that subjects with heel pain walked flatfooted in an attempt to avoid high stresses on the painful heel during initial foot-ground contact[617]. However, unfortunately this gait adaptation may not reduce the risk of sustaining a slip as increasing the foot contact area at initial foot-ground contact has not been shown to always prevent a slip (see Section 2.5.2).

Those RA patients with resulting joint hypermobility may also develop excessive and prolonged pronation with weight bearing which, in turn, imparts mobility to the midtarsal joint[592,596] to assist in cushioning the impact forces[596]. Increased midtarsal joint mobility, however, may also diminish balance ability in

single limb support[592,596], as well as prevent the foot from becoming rigid during propulsion, reducing the effectiveness of the foot as a lever for push off[596]. Furthermore, the resulting pronation can preclude the apropulsive gait pattern by making the individual roll off the medial aspect of their foot at terminal stance, exacerbating the forces under the forefoot, particularly across the medial metatarsal heads[279]. Consequently, the valgus forces on the rearfoot are increased and the windlass action of the metatarsal heads and plantar fascia, which normally provide support for the longitudinal arch as body weight rolls over the metatarsal heads during the terminal stance, is eliminated[303]. Therefore, to assist in further reducing forefoot pain, RA patients have been found to walk with their lower limbs externally rotated, shortening the effective lever arms that are provided by the feet and decreasing the pressures on the painful metatarsal heads[303,479,594].

During walking, healthy individuals usually flex their knee after initial foot-ground contact to assist with shock and energy absorption (see Section 2.5.2). However, compared to straight-legged gait, knee flexion requires increased activation of the quadriceps muscles and thereby increased compressive joint forces, leading to greater knee joint pain. Consequently, patients with painful knees attempt to reduce these compressive force by decreasing the range of knee motion during walking[306]. Therefore, as expected and conducive to the typical lower limb problems seen in the RA patient (see Section 2.6.1), knee flexion during the stance phase of gait has been reported to decrease in RA individuals when compared to age-matched control subjects[305,306,308,616,618,619]. Regrettably, reduced knee flexion, particularly when combined with impaired corrective responses (see Section 2.3.1) and unstable joints, leads to subsequent secondary muscle weakness and quadriceps wasting from disuse[305], such that the patient displays a limp or attempts to lock their knees for gait stability[179,308]. Impaired muscle strength or motor control patterns may leave the RA patients without the reserves to adequately respond to changes in shoe-surface interactions, in turn, increasing their slip risk.

The altered gait patterns seen in the RA patient appear dependent upon the lower limb deformities experienced, as well as the severity of the disease, and occur as a direct result of the disease process. However, further deformities and/or gait modifications may be acquired, indirectly as a result of pain-avoidance gait strategies used by these patients[596]. Irrespective of whether the gait changes are direct or indirect, the RA patient will require altered patterns of muscular activity to compensate for redistributed loading patterns as well as to ensure adequate gait stability[592]. Therefore, footwear design characteristics (see Table 2.2 and Table 2.3), which can reduce the need for these pain-avoidance gait modifications, will assist the RA patient to have pain free gait, allowing them to remain independent and mobile for longer, thereby improving their quality of life. However, whether these design characteristics can be integrated into household shoes, such that the household shoes do not lose their desirable features (see Section 2.4.1), is unknown. Furthermore, if such indoor shoes did exist for the RA patient, further research is warranted before they are prescribed to ensure that these footwear types did not negatively affect the gait of RA patients when walking on common household surfaces, particularly with respect to the strategies used for initial foot-ground contact and thereby slip prevention.

2.7 Summary

After reviewing the extensive literature pertaining to falls in older individuals, it is evident that this population of people, particularly older women, are at risk of falling in their own home environment. Furthermore, these falls often result from slips that typically occur when individuals are walking on level household surfaces. Of the many risk factors implicated in home slips, the major factors cited include inappropriate household footwear, such as slippers, slippery surfaces and gait modifications caused by musculoskeletal problems such as arthritis and foot pathologies. However, due to their lack of structure, it can be difficult to keep household shoes on the foot and, when coupled with common household surfaces, household shoes can provide an insecure base for gait, thereby predisposing older people to slips[21,25].

In walking, it is usually unseen hazards in one's route of travel or unexpected changes in the walking surface and how this surface interacts with the footwear being

worn, that places an older person in danger of slipping. When a potentially hazardous condition is perceived or expected to exist in the walking person's perceptual field, the walking pattern is adjusted to elicit slip-avoidance behaviour. Research currently exists as to how walking patterns change with age, on different surfaces and with different outdoor footwear types. However, no study was located which has investigated whether the kinematic, kinetic and neuromuscular variables which describe an older individual's gait pattern, are indicative of slip-avoidance evoked walking changes when older individuals, particularly those with musculoskeletal disease, wear different household shoes and walk on typical household surfaces. Therefore, this thesis aimed to identify how different household shoe-surface interactions affected the gait of older people, particularly those with foot problems, to set boundaries upon which recommendations for designing a household shoe characterised as safe for older people could be based.

EXPERIMENTAL SECTION A:

Identifying older people who suffer falls, foot problems and require specialised footwear

Chapter 3

The Survey

3.1 Introduction

The need for safe footwear for older people is well documented (see Section 2.4). However, current recommendations for safe footwear appear limited to outdoor shoe types. Therefore, there is an urgent need to develop recommendations for safe indoor footwear for older adults, particularly as most falls sustained by older people who live independently in the community occur inside their own home (see Section 2.2.2), with inappropriate household footwear having been identified as a major risk factor in home falls. However, to ensure that guidelines for safe indoor footwear will be applicable to the target population it is vital to identify the specific footwear needs of independently living older people who suffer foot pathologies and who are at risk of falling in the home. Only then may specific footwear design characteristics be recommended to aid in developing a safe household shoe to decrease the risk of slips and eventual falls in older people. However, as no one shoe will satisfy all older people, there is also a need to firstly identify which specific sub-group of older people is prone to falls, have foot problems and therefore require specialised footwear. To achieve these goals, a survey was designed, validated and implemented in Experimental Section A.

3.1.1 Statement of the Problem

The primary purpose of the survey was to characterise requirements and attitudes of older people living independently in the community in relation to their household footwear wearing and purchasing habits. The survey also identified characteristics of older people who suffer falls, foot problems and require specialised footwear, to provide selection criteria for the subject base as well as the specific household footwear and surface types to be examined, in Experimental Section B.

3.1.2 Hypotheses

Based on the reviewed literature, the following hypotheses were formulated for Experimental Section A:

(1) Medical conditions that affect the musculoskeletal system, such as arthritis, would be the most frequently reported medical condition by the respondents.

(2) Foot pathologies that are associated with foot pain and foot deformity and that may require specialised footwear would be the most frequently reported foot pathologies.

(3) Shoes that lack structure would be the most frequently reported shoe type to be worn around the home by the respondents.

(4) Hard, smooth surfaces, such as those commonly found inside the home, would be the most frequently reported surfaces perceived as slippery.

(5) Female respondents would report a greater number of falls, musculoskeletal medical conditions, foot pathologies and foot pain compared to male respondents.

3.1.3 Limitations and Delimitations

(A) Limitations

The following factors were acknowledged as limitations of Experimental Section A:

(1) Subjects were limited to volunteers aged 65 years and above who were living independently in the community. Therefore, the survey results may not represent all people aged 65 years and above in the Kiama, Shellharbour and Wollongong local government areas.

(2) Although the validity of the questionnaire was determined (see Section 3.2.3(B)), validity of response items could not be ensured as each respondent completed the questionnaire in the privacy of their own home.

(3) The questionnaire included some items that required retrospective responses. Response accuracy could therefore have varied, as the information retained by respondents could not be controlled.

(4) The initial survey response rate could not be altered due to restrictions specified by the University of Wollongong Human Research Ethics Committee that disallowed any follow up.

(B) Delimitations

The following delimitations were imposed on Experimental Section A:

(1) Subjects were restricted to independent community-dwelling people aged 65 years and above who had telephone numbers listed in the 1996 Kiama, Shellharbour, Wollongong White Pages™ [620]. Therefore, the results of the survey are specific to this population.

(2) Data were collected over summer and therefore may have excluded those respondents who only wear household shoes during the cooler seasons of the year.

(3) Respondents who failed to comprehend the telephone conversation did not participate in the survey. As these respondents are at a high risk of falling, omitting these subjects may have reduced the total number of fallers.

(4) Telephone calls were made to households during the day. Therefore, any prospective respondent who was not at home during the day was not included in the study.

3.2 Survey Methods

3.2.1 Survey Design

To characterise the household footwear wearing and purchasing habits of the participants, a self-administered mail questionnaire was designed. This format was selected for the present survey as it is more time and cost efficient

than other questionnaire formats and avoids interviewer or respondent bias for topics that require information of a sensitive nature as respondents remain anonymous[621-624]. Furthermore, this format is preferred for longer questionnaires as it gives respondents time to answer, is well suited to large and geographically diverse study populations[622,625] and is thought to provide higher quality data when compared to telephone and interview survey techniques[621]. Finally, the survey was implemented using a cross-sectional design, which assumes good validity for questionnaires that are completed only once by a sample who are considered representative of the population[626].

The response rates of mail surveys usually range from 22% to 75%[627-631], although increased age has been reported to reduce these response rates[621,632]. Prenotification of an impending survey via a telephone call, mailed letter, postcard or personal contact has been shown to significantly increase response rates in a variety of mail surveys[627-631]. Linsky[624] reported that maximum improvements in mail-survey response rates were gained by prenotifying potential respondents by telephone. Therefore, in the present study, households were prenotified by telephone, firstly to establish whether an eligible subject resided within the contacted household and, secondly, to ask consent of eligible subjects in an attempt to the increase response rate. Subjects who gave their consent were then mailed a questionnaire to self-administer. Details of sample selection, sample size, telephone procedure and survey distribution are described in the following sections.

(A) Sample Selection

People aged 65 years and above who resided independently in the Kiama, Shellharbour and Wollongong local government areas of the Illawarra (New South Wales) were selected as the survey population. This age group was selected because falls-related hospital admissions are reported to dramatically increase in people aged 65 years and above[633] (see Section 2.2.3). Furthermore, older people living in the Illawarra have been identified as a population who

wore slippers and/or footwear characterised as unsafe at the time of a major fall, which resulted in a fractured neck of femur[23].

(B) Sample Size

At the time the survey was implemented, 249,540 people lived in the Kiama, Shellharbour and Wollongong local government areas of which 29,268 people (11.7%) were aged 65 years and above[634]. Approximately 93% of this older population was living independently in the community at the time of survey completion[635] and only an estimated 3.7% of these individuals did not have a telephone connected at their home[636,637]. Approximately 87% of private households in the community have telephone numbers listed in telephone directories[636] with the Kiama, Shellharbour, Wollongong Telstra White Pages™ [620] telephone directory including most of the telephone numbers of private households in the areas selected to survey. Therefore, this telephone directory constituted the sampling frame for the survey. Based on these preceding statistics, it was anticipated that 27,748 people aged 65 years and above living independently in the Kiama, Shellharbour and Wollongong local government areas (9.1% of the total population) had an equal opportunity of being contacted using the sampling frame, thereby forming an unbiased sample.

The sample size was calculated using Equation 3.1[638]:

$$N = \frac{(Py)(Pn)}{StdError^2} \qquad \textbf{Equation 3.1}$$

where:

where:	Py	=	the proportion of eligible respondents
	Pn	=	the proportion of non-eligible respondents (1 – Py)
	$StdError$	=	standard error equal to the sampling error divided by the confidence level coefficient
	N	=	sample size

Equation 3.1 was calculated based on a sampling error of 0.05, indicating a confidence level of 95% with a corresponding coefficient of 1.96[639], and the proportion of eligible people in the population being 9.1%. Based on these data, a sample size of 127 persons 65 years and above and living independently in the

Kiama, Shellharbour and Wollongong local government areas were required for the survey.

(C) Telephone Call Procedure

Traditionally, only half of eligible people contacted by telephone participate in a survey[629]. Therefore, based on an average response rate of 65% of those who volunteer to participate in the study and the proportion of eligible people in the population being 9.1%, it was anticipated that 4,294 telephone calls would have to be made to derive the required sample size of 127 respondents.

To identify which telephone numbers to call, a simple random sample of telephone numbers was drawn from the sampling frame by means of a random generation of page number (39-308), column number (1-5) and row number (1-122) using Microsoft® Excel for Windows Version 6.0 (Microsoft Corporation, Washington, USA). Telephone numbers were dialled between 9.00 am - 11.30 am and 1.30 pm - 4.00 pm. These times were chosen as pilot testing determined that most people aged 65 years and above were home during these times, thereby reducing non-response by non-contact[636]. Four qualified research assistants, trained in survey presentation by the Chief Investigator, together with the Chief Investigator, made the telephone calls.

When the telephone was answered, the research assistant making the call introduced herself by name, department (Biomedical Science) and university (University of Wollongong), and then introduced the survey. Prospective respondents were then screened for age and residency status and, if considered ineligible, thanked for their time and the telephone call was ended. If eligible, the respondent was given further information pertaining to the survey and their expected involvement and then asked if they would like to participate. If consent was denied, the respondent was thanked and the telephone call was ended. If consent was granted, the name and address of the respondent were recorded, the respondent was thanked, and the telephone call ended. Only those respondents volunteering to participate in the survey were mailed a

questionnaire[640]. The average time of each telephone call was 6 minutes and the telephone calls were made in Summer (February and March).

As a method to decrease the number of telephone calls necessary to obtain the required sample size, and in turn the time and cost of the survey[636], an equal probability method of sampling was used. This method allowed all eligible individuals at one contacted household to participate in the survey[636]. Therefore, each eligible individual residing in any one household was personally spoken to and invited to participate in the survey before the telephone call was ended.

If an eligible respondent was unavailable at the time of the initial telephone call, but another resident answered the telephone, arrangements were made to call again at a more appropriate time. In the case of an answering machine, a message and contact telephone number were provided and a follow up telephone call made. If a respondent was unable to speak English, the research assistant asked to speak to another member of the family who could translate the information to the eligible respondent. If no one at the household spoke English, potential respondents at the household were considered ineligible and the telephone call was ended. Furthermore, potential respondents classed as unable to complete the questionnaire because of illness, physical or mental disability were also considered ineligible and not included in the survey[636]. Following common survey practice in Australia, each identified telephone number was phoned up to four times on different days at different times before it was registered as a non-answered call[636]. Furthermore, each telephone was allowed to ring at least 10 times before the unanswered call was ended, to enable sufficient time for older, more disabled people who may need longer to get to the telephone than their younger counterparts[636].

(D) Survey Distribution

The questionnaire was mailed to consenting respondents together with a stamped self-addressed envelope and a personalised introductory cover letter,

specifying the university's name*, which explained the procedure required to complete and then return the completed questionnaire. For confidentiality reasons, no numbers or marks distinguishing the respondents were placed on the questionnaire[640]. These techniques reportedly increase response rates[624,627,632].

The telephone introduction script, introductory cover letter, questionnaire and summary of findings were all approved by the University of Wollongong Human Research Ethics Committee (see Appendix A.1). All testing was conducted according to the Statement on Human Experimentation[1]. Once the data were analysed, a summary of the major findings of the survey were mailed to all respondents who originally volunteered to participate in the survey.

3.2.2 Response Rate

When the survey was completed, 1,715 households had been contacted by telephone, the details of which are displayed in Figure 3.1. The total number of telephone calls made was less than the 4,294 anticipated calls (see Section 3.2.1(D)) as 23.4% of the households contacted had residents who were eligible to participate in the survey compared to the anticipated 9.1% (see Figure 3.1). Of these 287 eligible respondents, 201 (70.0%) consented to participate in the survey and were mailed a questionnaire.

One hundred and twenty eight respondents completed and returned their questionnaires for analysis, yielding a response rate of 63.7% (of those 201 subjects initially consenting to complete the questionnaire). This number of respondents matched the required 127 sample size (see Section 3.2.1(B)). Similar response rates have been documented for mail surveys using telephone prenotification[629] and for surveys of community-dwelling older people[641].

* Indicates to the respondents that the research is sponsored by a university, which has been shown to increase response rate[624].

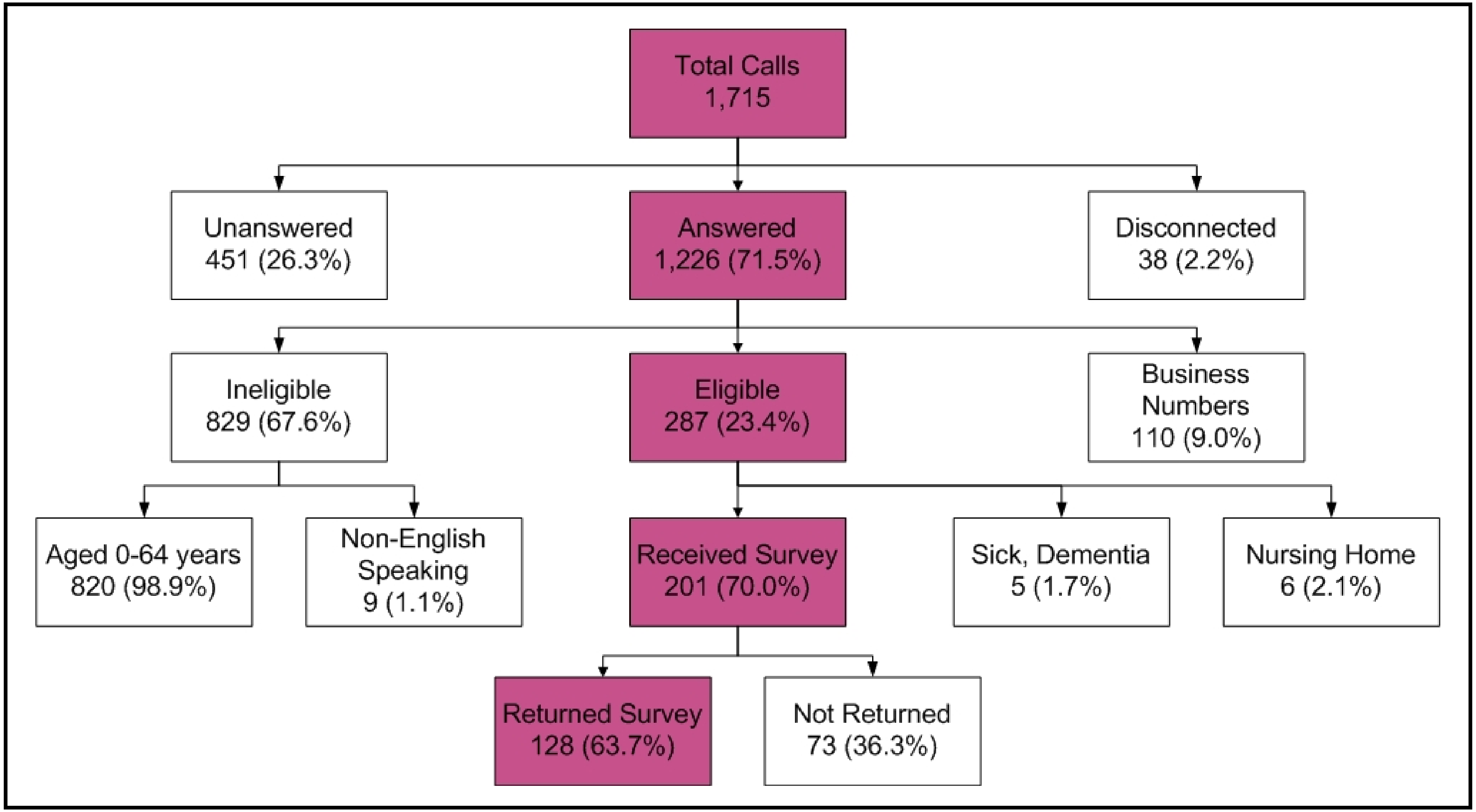

Figure 3.1: Details of telephone contact leading to the final survey response rate.

Ethical restrictions prevented any follow-up of the consenting respondents to facilitate this response rate[640] or to establish characteristics of non-respondents.

3.2.3 Survey Instrument

The survey was a 15-page questionnaire composed of 51 open- and closed-ended items written in plain English[*] (see Appendix A.2). Although it is generally believed that long questionnaires result in low response rates, evidence for this is inconclusive[625,642,643] and each item included in the questionnaire was considered necessary to achieve the survey's purpose (see Section 3.1.1). The survey was printed on both sides of A4 paper in size 14 Arial font for ease of reading in a sample likely to have deteriorated vision[167]. Each closed-ended question had a checklist of mutually exclusive categorical response options and a separate response option of "other" in case the desired answer was not available from the checklist. The open-ended items sought a written response composed by the respondent. More closed-ended items were used in preference to open-ended items as closed-ended items have been shown to be more reliable and consistent over time[626]. The open-ended items were used to provide additional information where appropriate.

Items for the questionnaire were derived from previous literature[19,25,46,257,644,645], previous questionnaires[23,40,255] and interviews with older people. Demographic questions on gender, age, postcode, citizenship, country of birth, Aboriginal or Torres Strait Islander status and annual income sought information to describe the sample. Information pertaining to each subject's medical and falls history was collected from responses to questions on general health status, presence of diagnosed medical conditions, walking ability, level of independence, medication use, doctor and hospital visitations and falls incidence. Information pertaining to the frequency and nature of foot pathologies was collected from responses to questions documenting foot pathologies, foot pain and discomfort, use of health services for feet and the need to wear various shoe appliances. Shoe wearing and purchasing habits were

determined by questioning the difficulty respondents had when donning shoes, the types of shoes worn inside the home, why these shoes were worn and purchased as well as the amount that would be spent on household and outdoor shoes. Information about surfaces deemed by respondents to be slippery in the home was also questioned.

(A) Survey Design Review

The questionnaire was piloted by six independently living older people (four females and two males; mean age, 72.3 years) to assess the gauged reaction to the survey. The questionnaire was also reviewed in terms of overall length, individual item length, logical item sequence, the placing of items into relevant sections, item and instruction wording, topic sensitivity, font size and questionnaire layout. Following this review, questionnaire items were modified and items deemed redundant were eliminated. Therefore, following the review process the potential for any forms of non-sampling error that may have arisen from errors in survey instrument design, item responses and non-response were considered minimal.

(B) Survey Instrument Validity

The validity of the current questionnaire was assumed as it was formulated from several sources already proven to be valid[23,40,255]. However, to ensure each question asked what it was designed to ask and that it was relevant to an older population, a panel[*] assessed the survey. The panel included experts in geriatric surveying, people educated in coherency of prose and members of the target population. Each panellist recorded a description of what each item asked together with an example answer, as well as whether each item was comprehensible, directly related to the purpose of the study and therefore required in the final instrument. Finally, panellists were asked whether they felt

[*] Questions were written so that they could be understood by a 14 year-old child[1].

any pertinent issues had not been covered in the questionnaire. When all comments from the panellists were collated, it was determined that the questionnaire was easy to understand and that each item measured what it was designed to measure[626]. Therefore, the questionnaire used was considered valid.

(C) Survey Instrument Reliability

The survey instrument was assessed for repeatability using the test-retest method whereby the questionnaire was administered to six independently living older people (four females and two males; mean age, 72.3 years) twice with a 1 month interval between tests[†]. The time lapse between completing each questionnaire was considered sufficient to minimise the subjects remembering their initial item responses. Therefore, as no responses were found to significantly differ in the information supplied on any of the open- or closed-ended items, the questionnaire was considered highly reliable.

3.2.4 Data Analysis

Answers to the closed-ended items were coded and counted to determine the frequency response for each item. The frequency data were then expressed as a percentage of the total cases and analysed using Chi-Square statistical tests to determine whether the observed frequencies differed significantly from the expected frequencies and whether gender was significantly related to any factor[646]. Yates' correction was used if the degrees of freedom were less than one[647]. Independent *t*-tests and *z*-scores were used to indicate differences between male and female responses for specific items. Descriptive comments from open-ended item responses were reviewed and systematically represented using "codes". These responses were then tabulated to obtain further

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[*] The panel consisted of an epidemiologist, a health promotion worker, a podiatrist, a rehabilitation geriatrician, two academic lecturers, two administrative assistants and two people aged over 65 years who were living independently in the community.

[†] Only minor changes were made to the survey following the survey design review. Therefore, the same group of older people, who were involved in the survey design review, were considered suitable to be involved in the survey instrument reliability assessment.

information on the relevant topics. The level of significance was set at $p \leq 0.05$. All statistical procedures were completed using SigmaStat for Windows Version 2.03[648].

3.3 Results and Discussion

Extensive data were collected from each questionnaire completed for the present survey. However, only data that proved to have direct relevance to Experiment Section B of this thesis (see Chapter 4) are presented in this chapter. Further survey results are presented by Munro & Steele[649,650].

3.3.1 Demographic Data

Of the 128 surveys returned for analysis, 46.9% were completed by men (mean age = 72.5 ± 5.2 years) and 53.1% by women (mean age = 71.6 ± 6.5 years). Men and women were not found to significantly differ with respect to mean age (t = 0.81; p = 0.419), number of participating respondents (t = 0.53; p = 0.597), country of birth (z = 2.95; p = 0.003) or income (χ^2 = 5.80; p = 0.122). The age characteristics of the men, women and total survey respondents are compared to data recorded in the 1993 estimated resident population (age and sex) for the Kiama, Shellharbour and Wollongong local government areas[634] in Table 3.1. Based on these data, the present sample was considered representative of the population of interest (see Section 3.2.1(A)).

There were no indigenous respondents (Aboriginal or Torres Strait islanders) in the present sample, although 4% of Australia's 327,000 indigenous people (approximately 13,080 persons) are aged 60 years and above. However, as anticipated, significantly more respondents (61.7%) were born in Australia compared to in other countries (31.3%). Of the respondents born overseas, most were born in Great Britain (52.5%) followed by other European countries (47.5%). The 31.3% of non-Australian country of birth respondents was higher than State and National estimates (23.4% and 22.8%, respectively) for people

born overseas[635]. However, as 119 respondents (93%) were Australian citizens, it is assumed that country of birth did not affect household shoe wearing or purchasing habits.

Table 3.1: Age categories of the men, women and total survey respondents compared to Kiama, Shellharbour and Wollongong local government areas estimates.

Age	**Men**			**Women**			**Total**		
(years)	**No.**[a]	**%**	**LGA**[b]	**No.**	**%**	**LGA**	**No.**	**%**	**LGA**
65-69	19	31.7	38.1	29	42.6	32.3	48	37.5	34.9
70-74	19	31.7	30.5	16	23.5	27.6	35	27.3	28.9
75-79	17	28.3	17.5	15	22.1	18.5	32	25.0	18.1
80-84	4	6.7	9.8	6	8.8	12.8	10	7.8	11.5
85 +	1	1.7	4.2	2	2.9	8.8	3	2.3	6.7
TOTAL	**60**	**46.9**	**44.3**	**68**	**53.1**	**55.7**	**128**	**100.0**	**100.0**

[a] Number of respondents.
[b] 1993 estimated resident population in the combined Kiama, Shellharbour and Wollongong local government areas (LGA)[634].

Sixty-three of 111 respondents* (56.8%) indicated that their approximate yearly income was below \$12,000† and income was not significantly related to gender ($\chi^2 = 5.80$; $p = 0.122$). Approximately 74.1% of Australians aged over 65 years derive their principal source of income from government cash benefits[60] and therefore have lower incomes than the average Australian wage‡, with a median income of \$8,200[635,651]. As over half of the respondents in the present survey had limited disposable income, any future recommendations for safe household footwear need to be cost effective if they are to be accessible by most older members of the community[650].

* This item was only answered by 111 respondents who reported wearing household footwear (see Section 3.3.6).

† All dollar amounts are recorded in Australian (AUD) dollars.

‡ \$29,010 and \$33,961 for females and males, respectively[651].

3.3.2 Falls History

Falls were reported by 27 respondents (21.1%) with 55.6% of these 27 respondents reporting more than one fall in the 12 months before completing the survey. This result was similar to State estimates, in which 20.1% of older people had fallen once in the year before being surveyed[42]. However, the figure is somewhat lower than other studies investigating community-dwelling older people[3,47,371].

Although a greater number of women (25.0%) reported a fall compared to men (16.7%), this difference was not statistically significant (z = 0.90; p = 0.366). The greater number of women reporting falls was, however, consistent with previous studies reported by the Australian Bureau of Statistics[42] and Dunne *et al*.[46]. In terms of falling rate, the higher proportion of older women compared to men (see Section 2.3.1(A)) usually results in a higher overall fall rate in women, if falling rates are not corrected for age and gender[45]. However, despite a similar mean age and number of male and female respondents, women reported 1.7 times more falls than men in the present survey. Although not statistically different, an increase in fall rate of nearly two-fold should not be dismissed and is consistent with the literature investigating falls in community-dwelling older populations, which conclude that women fall more often than men (see Section 2.3.1(A)). Therefore, falls prevention programs or interventions targeted towards women may have a more significant effect on falling rates compared to programs or interventions targeted towards men.

3.3.3 Health Characteristics

Similar to the 1995 National Health Survey[101], most respondents (88.3%) considered their overall state of health to range from fair to very good with only 12 respondents (9.4%) reporting poor health. This result was also consistent with the study by Bogle Thorbahn & Newton[93] who found that 86% of older people (mean age, 79.2 years) perceived that their overall health was good or excellent. Respondents in the present survey reported being mobile and independent with over half being able to walk for longer than 30 minutes before

they needed to rest (58.6%) and requiring no assistance to perform daily chores (62.5%). Older people who have difficulty walking are often prone to falls due to associated problems of postural instability[10,230], limited mobility, a dependence on others or a low self-perception of health[217]. However, as survey respondents were community-dwelling people (see Section 3.2.1(A)) it was assumed they would be relatively active, able to independently perform daily living activities and, therefore, consider themselves in good health, an assumption that was supported by the results. Interestingly, despite the relatively good health of the respondents, 21% had still incurred a fall (see Section 3.3.2).

Gender was not found to be significantly related to health-perception (χ^2 = 1.33; p = 0.932), the length of time able to be walked before requiring a rest (χ^2 = 1.46; p = 0.482) or the assistance required to perform daily chores, such as bathing, cooking, cleaning and shopping (χ^2 = 2.58; p = 0.630). Regardless, there were trends towards women not being able to walk for as long as men before requiring a rest and more women required assistance to perform everyday household chores compared to men. These trends are consistent with the fact that women are generally weaker than men (see Section 2.3.1(A)) and have a greater incidence of musculoskeletal conditions such as arthritis and foot pathologies compared to men, which may impose restrictions on their activity and ability to complete daily living activities (see Section 2.3.1(G) and Section 3.3.4). As most falls occur during normal daily living activities (see Section 2.2.3), respondents in the present study are still at risk of falls as they are placing themselves in situations where falls can commonly occur.

3.3.4 Medical Characteristics

The diagnosed medical conditions reported by men and women are displayed in Figure 3.2. There was no significant relationship between gender and diagnosed medical condition (χ^2 = 14.51; p = 0.339). Furthermore, despite the average number of medical conditions per women being 3.2 compared to 2.7

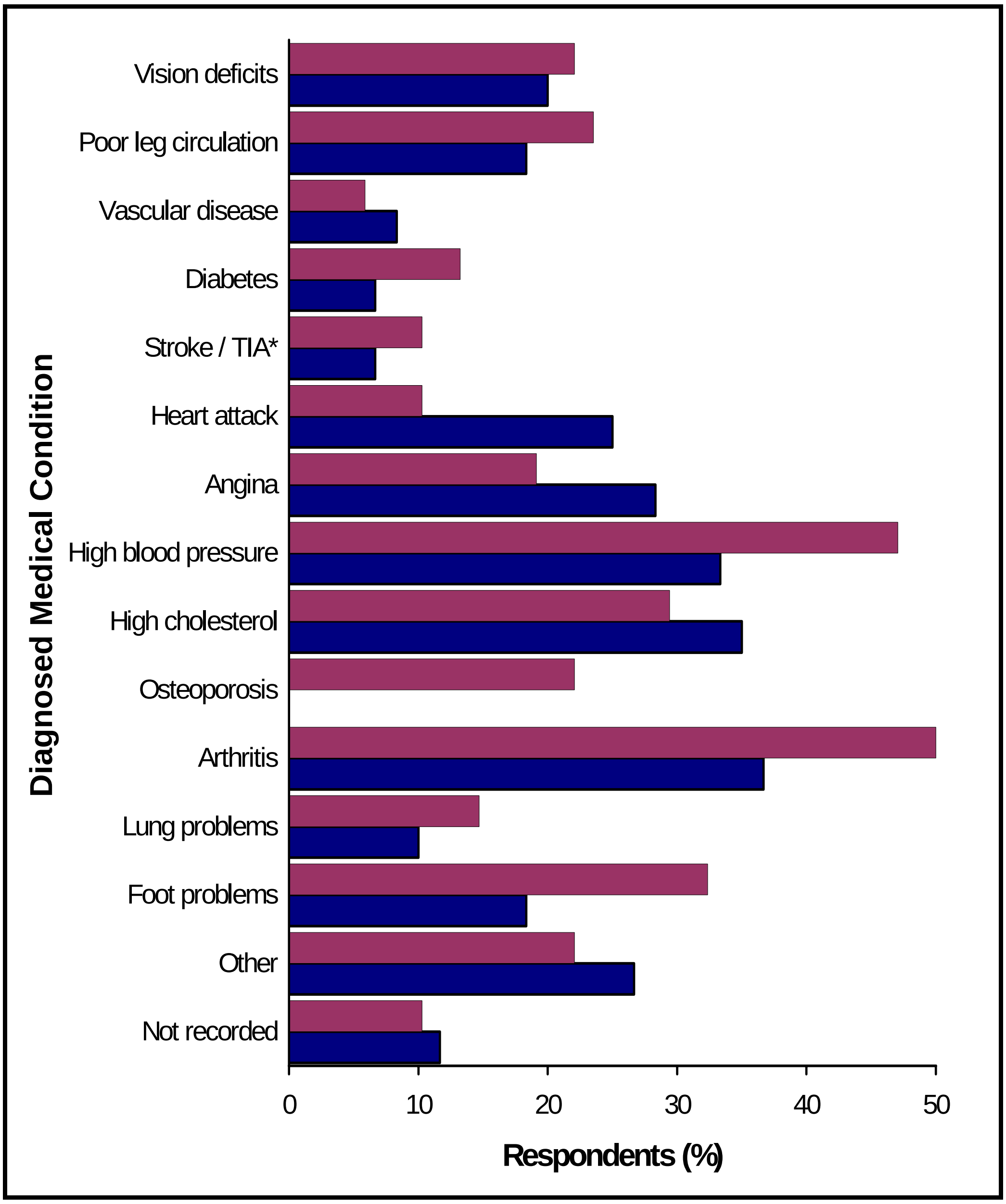

Figure 3.2: Diagnosed medical conditions reported by men (n = 60; ■) and women (n = 68; ■; *TIA = transient ischaemic attack).

medical conditions for men, women did not report significantly more medical conditions on average than men (t = -1.11; p = 0.268). The diagnosed medical conditions most commonly reported by the women were arthritis (50.0%), high blood pressure (47.1%) and foot problems (32.4%) whereas the men most commonly reported arthritis (36.7%), high cholesterol (35.0%) and high blood

pressure (33.3%; see Figure 3.2). Only seven men (11.7%) and seven women (10.3%) reported not suffering any diagnosed medical conditions.

Consistent with the present results, previous studies have indicated that between 77% and 99.4% of older people suffer chronic medical conditions, with most having multiple conditions[101,652] (see Table 2.1). Furthermore, these studies found that no person examined was pathology free, with the most commonly reported conditions including arthritis, high blood pressure and high cholesterol; conditions also commonly reported in the present survey. Factors related to the development, type or number of specific medical conditions may lead to necessary and/or self imposed restrictions in physical activity[653] as well as a diminished capacity to interact safely with environmental hazards[654]. Therefore, individuals suffering pathologies are at a greater risk of falls. Consistent with this notion, in the present survey, respondents who reported a fall in the 12 months before completing the survey also reported significantly (t = -3.75; $p < 0.001$) more diagnosed medical conditions (4.7 ± 1.6) compared to respondents who did not fall (2.9 ± 2.2). Musculoskeletal conditions, such as foot problems and arthritis, have the potential to limit mobility and provoke falls due to pain and deformity (see Section 2.3.1). Therefore, characteristics of older individuals affected by foot problems and arthritis require further consideration before safe shoe recommendations can be made.

3.3.5 Foot Problems

The specific foot problems reported by 36 men (60%) and 55 (81%) women are displayed in Figure 3.3. Women reported significantly (t = -2.68; p = 0.008) more foot problems than men, with an average of 2.3 foot problems per woman, compared to 1.4 foot problems per man. Furthermore, a total 67 respondents (52.5%) experienced foot pain and/or discomfort and, although not significant (z = 1.19; p = 0.236), more women (58.8%) experienced pain and/or discomfort in their feet compared to men (45.0%). The reporting of specific foot problems was significantly related to gender (χ^2 = 22.39; p = 0.022), such that

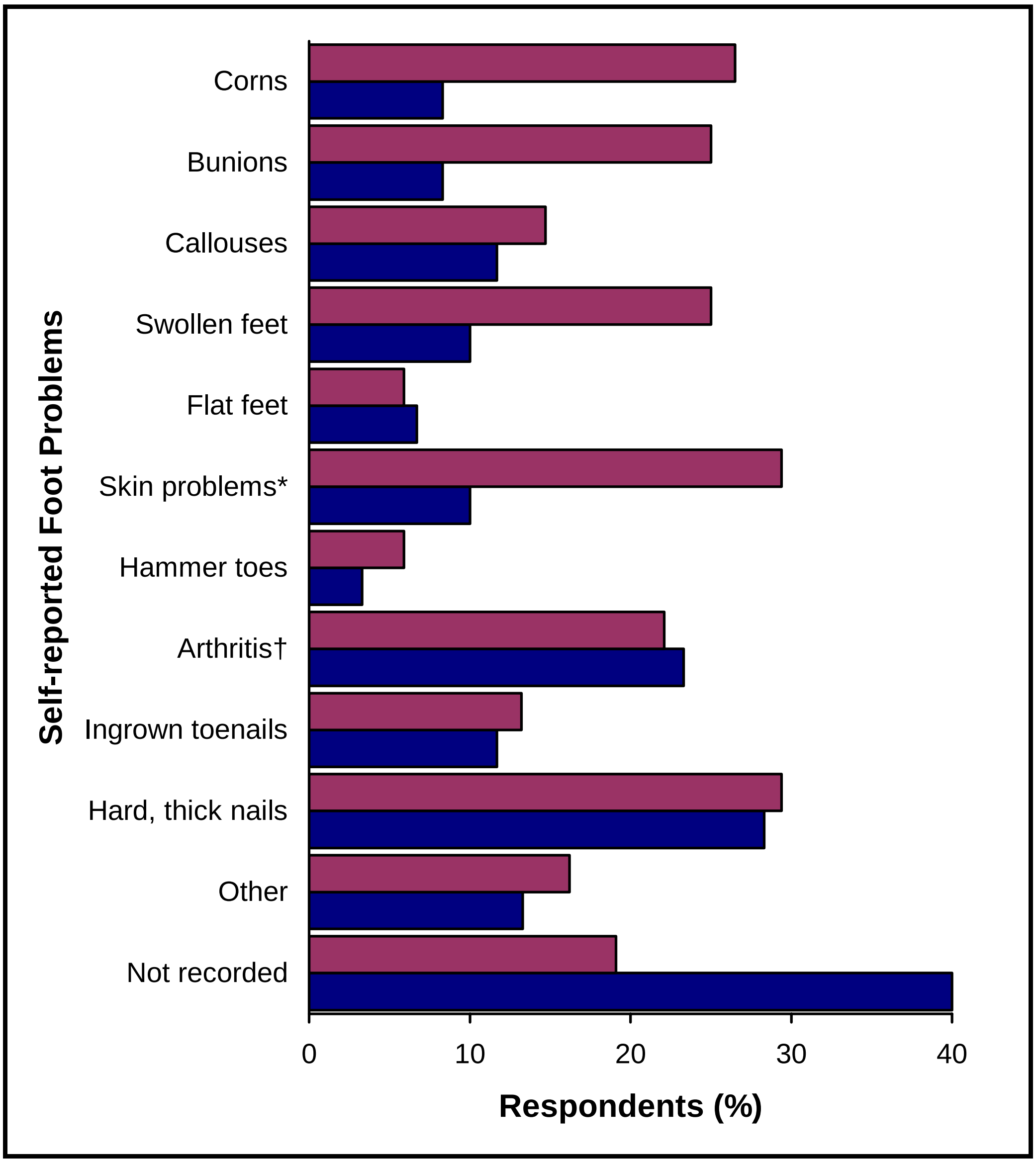

Figure 3.3: Specific foot problems reported by men (n = 60; ■) and women (n = 68; ■; *skin problems included rashes, blisters, dry skin and warts; †Arthritis included osteoarthritis, rheumatoid arthritis and gout).

women reported more conditions and men reported fewer conditions than expected (see Figure 3.3). When controlled for those respondents who reported foot problems, men and women both indicated hard thick nails (28.9%), arthritis (22.7%) and skin conditions (20.3%) as the most common foot problems. Consistently, foot pain and/or discomfort were predominantly experienced, for both genders, around the toes (47.8%), across the whole foot (25.4%) or around

the heels (14.9%), particularly when their feet were hot and swollen (see Table 3.2). In addition, foot pain and/or discomfort mainly occurred during normal everyday activities such as walking (56.7%; see Table 3.2).

Table 3.2: Foot pain and/or discomfort reported by men (n = 60) and women (n = 68).

Question & Response	Men		Women	
	No.	%[a]	No.	%
Where do you experience pain and/or discomfort on your feet?				
Toes	11	40.7	21	52.5
Heels	3	11.1	7	17.5
Soles	4	14.8	4	10.0
Arches	3	11.1	4	10.0
Whole feet	8	29.6	9	22.5
Not recorded	1	3.7	1	2.5
When do you experience pain and/or discomfort on your feet?				
Only in the morning	3	11.1	2	5.0
Only in the afternoon	2	7.4	1	2.5
After standing	1	3.7	4	10.0
After exercise	5	18.5	5	12.5
Only in hot weather	4	14.8	11	27.5
At all times	6	22.2	15	37.5
Other time	2	7.4	9	17.5
Not recorded	7	25.9	3	7.5
What activity causes pain and/or discomfort on your feet?				
Walking	16	59.3	22	55.0
Standing	3	11.1	5	12.5
Other activity	5	18.5	10	25.0
Not recorded	7	25.9	8	20.0

[a] Percentages may not add up to 100 due to multiple answers.

The high frequency of foot problems reported in the present study supports the notion that foot problems are common, especially in older people, and that these problems lead to foot pain and/or discomfort together with foot deformity (see Section 2.3.1(E)). For example, Robinson[255] found that nearly 40% of older people living independently in the community stated that their feet hurt and that their foot pain affected their mobility. Interestingly, despite a tendency, no respondents in the present survey reported that foot problems

affected their mobility or ability to perform daily living activities (see Section 3.3.3). However, the present findings are also consistent with previous research in which women presented with more foot problems and a greater incidence of foot pain and/or discomfort than men[25,259]. It is thought that the higher number of women presenting with foot problems and foot pain and/or discomfort may be due to the high proportion of women who habitually wear high-heeled shoes with pointed, shallow toe boxes (see Section 2.3.1(E)). These shoe types place increased pressure on the forefoot, compared to the flat shoes or work boots with rounded toe boxes traditionally worn by men, leading to the foot problems characteristically reported by women[25,255,259].

The foot problem most commonly reported by the survey respondents, irrespective of gender, was that of hard thick nails (41.4%; see Figure 3.3). This high incidence of nail problems may be due to an inability of older people to properly trim their nails as they are unable to bend down to reach their nails due to decreased mobility, poor eyesight, some other impairment or due to a lack of appropriate nail trimming equipment[42,101,252,255]. Conversely, poor footwear, which impinges on the nail, may contribute to foot problems[255]. The high reporting of arthritic feet and skin problems (see Figure 3.3) was expected from this sample who had also reported a high incidence of general arthritis (see Section 3.3.4). Although most respondents reported they were active, mobile and independent (see Section 3.3.3), 52.3% also experienced pain and/or discomfort during some form of activity (see Table 3.2). Appropriate foot care may therefore provide the key to increased mobility, productivity, independence, freedom from pain and general well-being in this sample[255,655]. However, instead of seeking treatment for their foot problems, many older people are thought to select shoes that mould to the shape of their feet to provide comfort and freedom from pain, enabling them to maintain independence[255,280,655]. Therefore, it is postulated that older individuals, particularly older females with foot problems and arthritis, would benefit greatly from well designed safe footwear, which can accommodate their specific needs, in order to decrease their pain and/or discomfort, further foot problems and falls risk[25,281].

3.3.6 Household Shoe Design and Wearing Characteristics

The shoe types worn in and around the home by the survey respondents are listed in Figure 3.4. One hundred and eleven respondents (78.9%) indicated they wore shoes in and around the home with 32.4% of women and 28.3% of men not wearing household shoes and instead going barefoot or wearing socks around the home (see Figure 3.4). The number of barefoot respondents may have been inflated as the survey was conducted during the warmer months of the year (average temperature = 23.9°C)[656]. A further 10 respondents indicated they alternated between going barefoot and wearing socks or household shoes. The wearing of shoes around the home was not significantly related to gender (χ^2 = 0.20; p = 0.652). However, the type of household shoe chosen was significantly related to gender ($\chi^2 = 25.29$; $p = 0.003$) such that women predominantly wore slippers, slip-ons, thongs or sandals as their household shoes. Conversely, men predominantly wore slippers, tied non-athletic shoes, tied athletic shoes or thongs around the home (see Figure 3.4). Only those respondents who indicated they wore shoes around the home completed the full questionnaire. Therefore, the ensuing results and discussion are based on information derived from those 50 men and 61 women who indicated they wore household shoes.

Older people are usually advised to wear shoes around the home rather than going barefoot. This is so that older feet, which often have decreased plantar sensory perception[250-252,254], fragile skin[598] and reduced healing abilities[598] are protected from possible hazards in the home environment[3,13,20,23,377] (see Section 2.3.1(E)).

Slippers were the most popular type of household shoe worn by 71.3% of the survey respondents with significantly ($z = 3.56$; $p < 0.001$) more of the respondents wearing closed back slippers (74.7%) compared to toe slippers[*] (25.3%). This percentage of respondents indicating that they wore slippers around the home was much higher than has been previously documented (see Section 2.4.1). For example, Dunne *et al*.[46] reported that only 18.3% of older

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[*] Closed back slippers have an upper that consists of both a toe box and a heel counter whereas toe slippers have an upper that consists only of a toe box (see Figure 5.2).

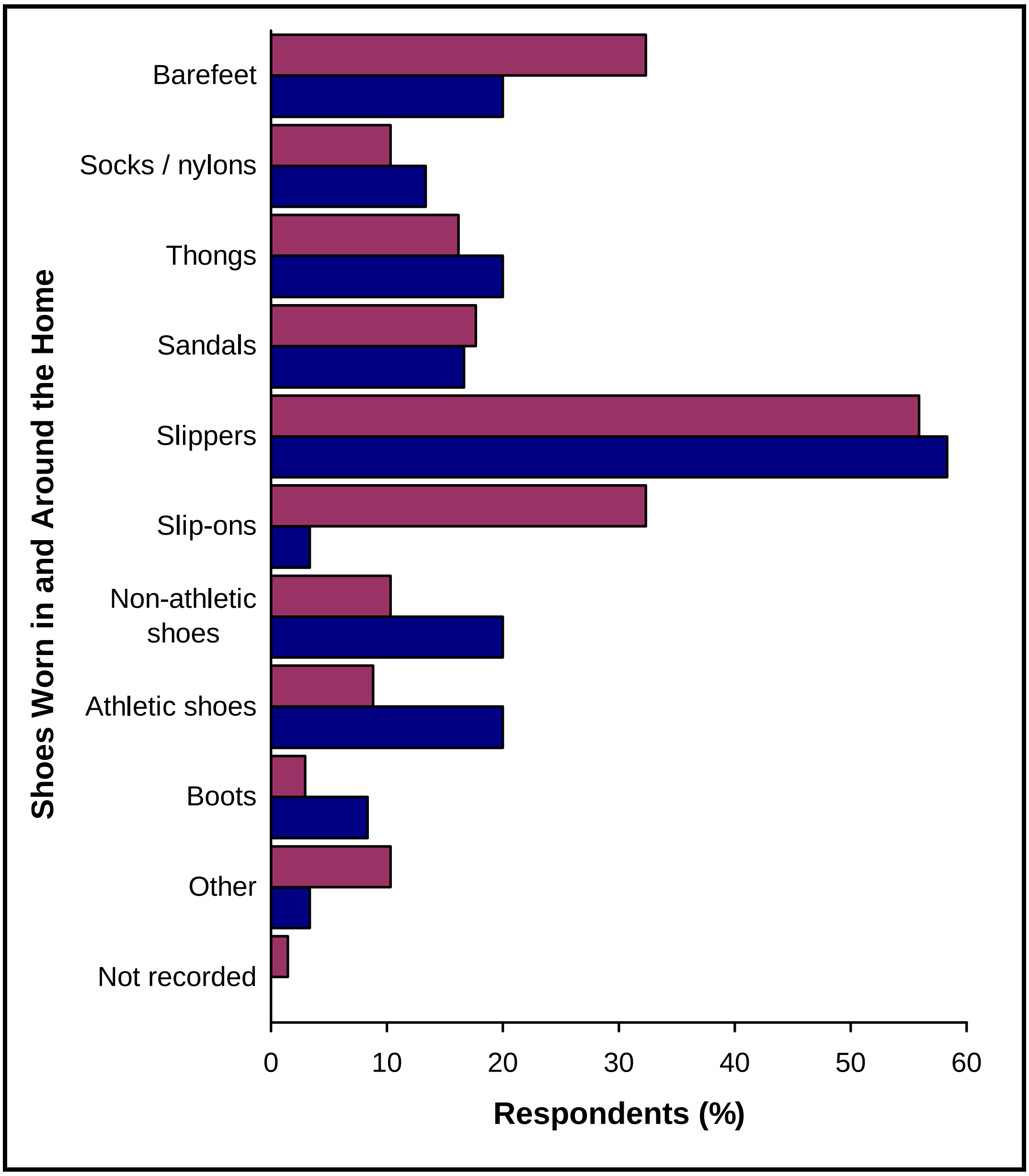

Figure 3.4: The types of footwear worn around the home by men (n = 60; ■) and women (n = 68; ■).

people were wearing slippers at the time the researchers called their homes by telephone during the day. Furthermore, Robinson[255] found that only 44.2% of older independently living residents wore slippers around the home whereas Finlay[19] reported that 28% of residents in a geriatric hospital wore footwear such as slippers. The discrepancies apparent between the present survey and previous studies may be explained by different definitions of slippers, differing methods of data collection and data coding and analysis. However, no previous studies

have documented the specific style of slipper (toe or closed back slipper) being worn by older people in and around their home. Therefore, further investigation is warranted to determine slipper types and designs commonly worn around the home by older individuals before finalising recommendations with respect to safe household shoes.

Although advised to wear shoes in and around the home, older people are usually discouraged from wearing slippers as household shoes. This is because slippers tend to become sloppy and do not fit adequately, making it difficult for older individuals, particularly those with foot problems, to walk without scuffing the ground[255]. Furthermore, slippers usually have slippery vinyl soles, which provide an insecure base for gait, and which may contribute to an increased risk of falls, especially when first getting out of bed[21,24]. However, only one study was located which provided an alternative to the wearing of slippers around the home. Robinson[255] suggested that older people should wear knee high Ugg boots instead of slippers, as these shoe types would provide the warmth of the slipper, due to their lambs-wool construction, a snug fit and support to the foot and ankle due to having a high collar (see Section 2.4.2). However, without a lacing mechanism, knee high Ugg boots may be difficult to don, particularly if foot pathologies and associated foot deformities are present. This difficulty would be compounded for those people who have limited fine motor skills in their hands due to diseases such as arthritis, or have insufficient upper body strength to pull the boots on to their feet. Ugg boots may also develop many of the problems associated with slippers such as inadequate fit and sloppiness.

Despite differences in shoe type, men and women in the present survey wore household shoes of similar composition to each other (see Table 3.3). That is, the household shoes worn by both men and women had uppers made predominantly of soft leather or fabric coupled with linings composed of fabric, leather or sheepskin. The soles of the shoes were made predominantly from synthetic materials with varied sole patterns and low heel heights coupled with soles that were easy to bend with their hands (see Table 3.3). Only 49 respondents (44.1%) indicated that their household shoes had fastenings, which included laces (69.4%), buckles (16.3%), Velcro (12.2%) and zippers (8.2%).

Table 3.3: Household shoe design features reported by men (n = 50) and women (n = 61).

Question & Response	Men		Women		χ^2	*p* value
	No.	%[a]	No.	%		
What type of material is the top of your household shoes made of?						
Leather	22	44.0	29	47.5	1.82	0.769
Fabric	14	28.0	18	29.5		
Synthetic	10	20.0	9	14.8		
Other material	11	22.0	17	27.9		
Not recorded	6	12.0	4	6.6		
What is the inside material/lining of your household shoes made of?						
Fabric	18	36.0	26	42.6	2.80	0.720
Leather	11	22.0	12	19.7		
Sheepskin	4	8.0	9	14.8		
No lining	5	10.0	6	9.8		
Other material	8	16.0	15	24.6		
Not recorded	8	16.0	6	9.8		
What sort of material is the sole of your household shoes?						
Synthetic	41	82.0	44	72.1	3.65	0.302
Leather	6	12.0	10	16.4		
Other material	2	4.0	8	13.1		
Not recorded	6	12.0	5	8.2		
What is the sole of your household shoes like?						
Smooth/flat	19	38.0	24	39.3	2.90	0.400
Patterned	18	36.0	18	29.5		
Rough	8	16.0	17	27.9		
Other sole	1	2.0	1	1.6		
Not recorded	7	15.0	5	8.2		
How high are the heels on your current household shoes?[b]						
Flat (< 1"/2.5 cm)	30	60.0	44	72.1	2.28	0.320
Medium (1"/2.5 cm)	15	30.0	16	26.2		
Other height	1	2.0	2	3.3		
Not recorded	7	14.0	4	6.6		

[a] Percentages may not add up to 100 due to multiple answers.
[b] Classification of heel height from Illawarra Health Promotion Unit[82].

There were no significant differences between men and women in the present study for any shoe wearing characteristics (see Table 3.4). That is, most respondents reported wearing their household shoes all day (52.4%) with over half of the respondents (58.1%) wearing their household shoes for more than 5 hours every day (see Table 3.4). Apart from three respondents, all considered

their household shoes to be comfortable because they were soft, light, easy to don and conformed to fit their foot shape. The three respondents who considered their household shoes to be uncomfortable did so for the reason of sloppiness, whereby they felt their shoes were too big or too loose. Although comfortable when on the foot, 33 respondents (30.5%) had difficulty placing their household shoes on their feet as they had trouble reaching down to their feet and getting the shoe on to the foot.

Table 3.4: Shoe wearing habits reported by men (n = 50) and women (n = 61).

Question & Response	**Men**		**Women**		χ^2	***p* value**
	No.	**%**[a]	**No.**	**%**		
When do you wear your household shoes?						
Morning/night only	19	38.0	17	27.9	3.28	0.657
All day	20	40.0	35	57.4		
When feet are cold	5	10.0	7	11.5		
Varies	7	14.0	7	11.5		
Other time	7	14.0	7	11.5		
Not recorded	6	12.0	5	8.2		
For how long do you wear your household shoes each day?						
< 2 hours	10	20.0	10	16.4	2.37	0.499
2 - 5 hours	11	22.0	10	16.4		
> 5 hours	23	46.0	38	62.3		
Not recorded	7	14.0	6	9.8		
Do you ever wear your household shoes outside (eg, in garden)?						
Yes	32	64.0	32	52.5	3.03	0.220
No	12	24.0	24	39.3		
Not recorded	6	12.0	5	8.2		

[a] Percentages may not add up to 100 due to multiple answers.

Slipper design characteristics considered to make the shoe unsafe for the wearer may at times be considered beneficial for some older people, particularly those with arthritis, foot problems and foot pain and/or foot discomfort. For example, although less supportive, soft and flexible uppers and lining materials allow the shape of the shoe to accommodate many foot shapes, particularly irregular foot shapes characteristic of arthritis sufferers. Furthermore, household shoes without any shoe closures to manipulate may be ideal for the older person

with impaired upper limb function, poor vision or who is unable to bend down, due to postural hypotension, musculoskeletal impairment or balance problems, to place a shoe on their foot[25]. Consequently, it would appear that current guidelines on how to select a safe household shoe (see Table 2.2) need to consider the often unique needs of older people, particularly those with specific foot problems. Results of the present survey have provided further insight into the specific needs of older people, particularly the large percentage with arthritis and associated foot problems, which can assist in developing recommendations for safe household shoes suited to the unique needs of this subgroup of older persons to be developed.

3.3.7 Household Shoe Purchasing Characteristics

No significant relationships were found between gender and any of the household shoe purchasing habits presented in Table 3.5. Household shoes were purchased when needed (59%) or every 1 or 2 years (31.4%; see Table 3.5). Similar to the results of a previous study of older individuals living in the Illawarra community[23], respondents in the present study purchased their household shoes at several locations including specialist shoe stores, variety stores without specialist shoe fitting assistance and department stores with specialist shoe fitting assistance (see Table 3.5). Furthermore, although not commonly marketed as a shoe store, many women in the present study purchased household shoes from pharmacies. As older people regularly visit pharmacies to refill prescription medications, promoting pharmacies as venues to purchase household shoes may be a strategy for future marketing of safe household shoes.

The cost of household shoes was important for survey respondents, with both genders indicating they would spend no more than $30 on a new pair of household shoes (see Table 3.5), a result which is analogous to a previous study[23]. However, over 50% were prepared to spend more than $50 on a new pair of fashion shoes to wear outside the home (see Table 3.5), despite earning

Table 3.5: Shoe purchasing habits reported by men (n = 50) and women (n = 61).

Question & Response	Men		Women		χ^2	*p* value
	No.	%[a]	No.	%		
Why did you buy the shoes that you currently wear around the house?						
Old shoes were worn out	11	22.0	5	8.2	10.90	0.141
Saw them and liked them	6	12.0	10	16.4		
Didn't buy – gift	8	16.0	8	13.1		
Medical advice/foot problem	5	10.0	9	14.8		
Comfortable	18	36.0	33	54.1		
Ease of putting them on	11	22.0	14	23.0		
Other reason	2	4.0	10	16.4		
Not recorded	7	14.0	5	8.2		
How often do you buy household shoes?						
Less than once per year	4	8.0	13	21.3	8.26	0.143
Every 1 - 2 years	4	8.0	14	23.0		
Only when I need to	31	62.0	31	50.8		
When I see something I like	5	10.0	10	16.4		
Other reason	7	14.0	8	13.1		
Not recorded	7	14.0	6	8.2		
How much money would you spend for a new pair of household shoes?						
< $30	29	58.0	28	45.9	1.09	0.780
$30 - $50	10	20.0	14	23.0		
> $50	10	20.0	15	24.6		
Not recorded	6	12.0	7	11.5		
How much money would you spend for a new pair of going out shoes?						
< $30	5	10.0	6	9.8	6.43	0.092
$30 - $50	7	14.0	21	34.4		
> $50	31	46.0	29	47.5		
Not recorded	8	16.0	6	9.8		
Where do you usually go when you are trying to buy household shoes?						
Variety store (no assistance)[b]	16	32.0	19	31.1	6.56	0.161
Department store (assistance)[b]	11	22.0	14	23.0		
Shoe store	20	40.0	26	42.6		
Other store	5	6.0	20	32.8		
Not recorded	7	14.0	5	8.2		

[a] Percentages may not add up to 100 due to multiple answers.
[b] Assistance refers to shoe fitting assistance.

less than $12,000 annually and with restricted income for essential items (see Section 3.3.1). Household shoes are typically cheaper than shoes designed to be worn out in public as household shoes are traditionally made of cheaper, lower

quality materials; require less material or are less expensive to manufacture. Therefore, household shoes would be expected to have less durability than a shoe designed for outdoor use. However, if worn for extended periods every day, as was reflected by the fact that 58.1% of the present respondents wore their slippers for more than 5 hours each day, it could be assumed that household shoes should be replaced more often than outdoor shoes. In contrast to this expectation, household shoes were infrequently replaced in the present survey (see Table 3.5), implying that respondents did not seem to place a high priority on purchasing household shoes or did not have enough spare income to regularly replace their household shoes. This result is consistent with the earlier findings of Marr[21] and Gabell *et al*.[22] who both reported that older people fell when wearing old worn slippers or slippers with worn soles (see Section 2.3.2(A)).

Regardless of place of purchase, cost or shoe type (that is, indoor or outdoor shoe), 55.8% of respondents in the present study had not had their feet measured for more than 5 years. Frey[257] reported that 95% of a sample of 356 females aged 50 to 60 years had increased foot size since 20 years of age and that 88% were wearing shoes that were on average 1.2 cm too small for their feet, often contributing to foot pain and deformity (see Section 2.3.1(E)). The static and dynamic loads that are borne by the feet year after year cause morphologic and physiological changes to the older foot, such that both foot size and foot shape change[24,25,250,257,439] (see Section 2.3.1(E)). The older foot is therefore usually wider[250] and has increased forefoot height compared to the younger foot[276]. However, as respondents in the present study considered their household shoes to fit their feet well and to be comfortable, irrespective of their foot pathologies, it is postulated that their household shoes lacked sufficient structure so that they were able to accommodate the many foot shapes typical of the older foot for each shoe size. Whether this lack of structure may also be considered unsafe by potentially contributing to falls by becoming loose on the wearer’s feet and, in turn, a hazard is not known.

A major problem commonly reported by older individuals in previous studies has been that mass-produced footwear, considered safe by health professional and researchers, does not adapt painlessly to the older foot, which is

often characterised by bunions, hammer toes, arthritis and other foot deformities[21,25,40]. The results of the present study are somewhat in conflict with this notion in that many respondents were able to purchase mass-produced household shoes, predominantly unstructured slippers that were relatively comfortable. However, whether such slippers pose an environmental hazard by altering the gait of these older people and, in turn, predisposing them to slips in the home on typical household surfaces, is unknown.

3.3.8 Surface Characteristics

The surfaces that respondents deemed slippery when wearing household shoes are presented in Figure 3.5. Most survey respondents deemed smooth tile (40.6%) and linoleum (15.2%) as the most slippery household surfaces when wearing household shoes and grass (14.9%) and unsealed surfaces (7.9%) as the most slippery surfaces outside the home (see Figure 3.5). A further five women and one man provided a written response, deeming wet surfaces as the most slippery surfaces. Twenty-three respondents (22.8%) did not find any surface slippery when wearing their household shoes (see Figure 3.5). There was no significant relationship between gender and the surface reported as most slippery ($\chi^2 = 3.66$; $p = 0.600$).

A survey conducted by the Australian Bureau of Statistics[42] of 139,500 persons aged 65 years and above who had fallen at least once in 12 months, found that slippery surfaces contributed to 10.3% of all falls (see Section 2.3.2(B)). However, no information was collected in this previous study on the types of surfaces classified as slippery. Waller[65] found that 22% of 150 people over 60 years of age, who were examined for falls, fell on icy or wet surfaces that would be classified as slippery. Therefore, although fall accidents may result from slips, trips or losses of balance (see Section 2.2.3), slips tend to be caused mainly by insufficient friction between the shoe sole and the walking surface, particularly at initial foot-ground contact. The traction between the shoe and walking surface is related to many factors, such as the adjustment of an

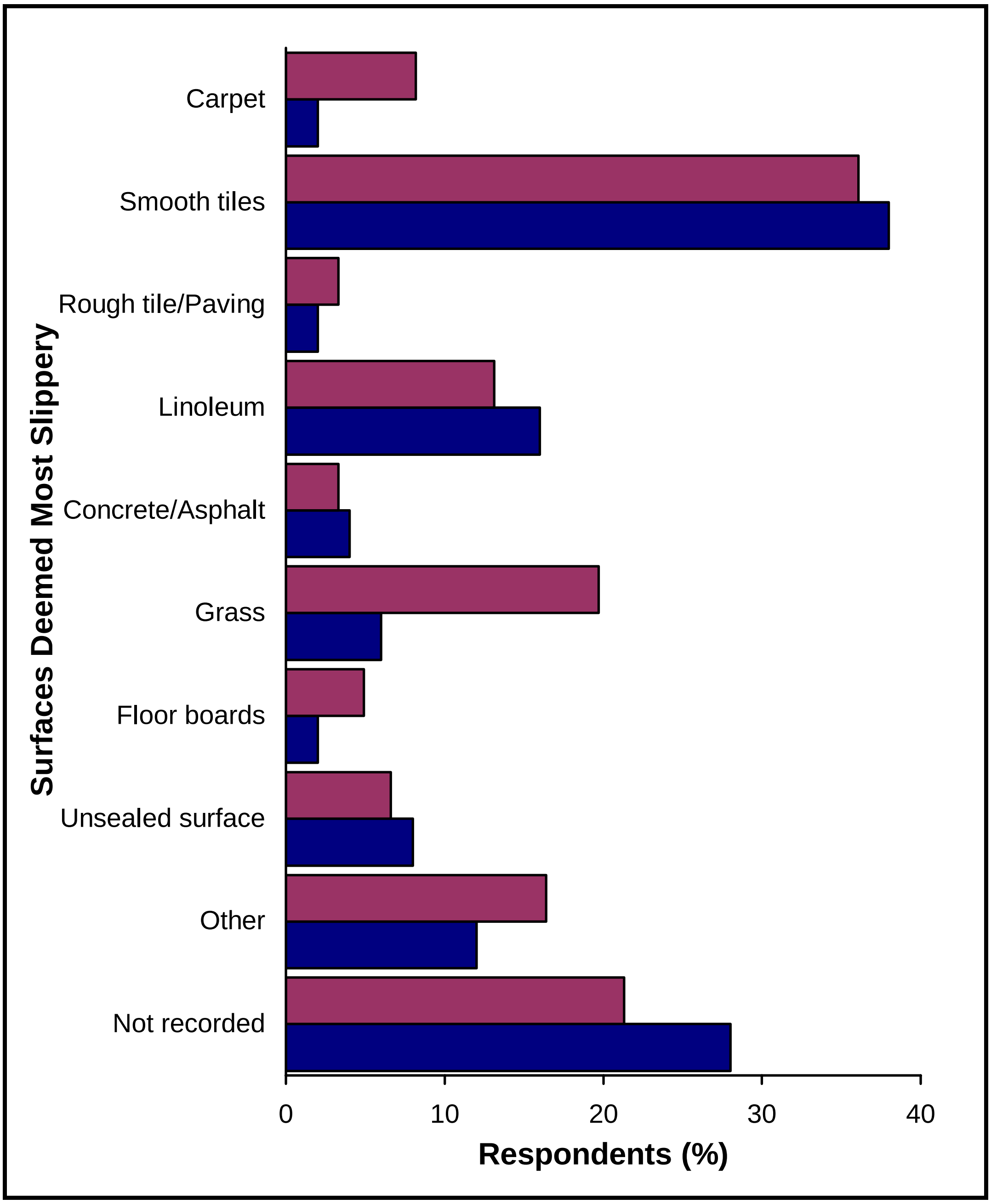

Figure 3.5: Surfaces deemed to be the most slippery when wearing household shoes as reported by men (n = 50; ■) and women (n = 61; ■).

individual's gait based on their perception of floor slipperiness[657,658], the floor surface condition, any surface contaminations (for example, water) and shoe wear pattern characteristics[461] (see Section 2.3.2(C)). Wilmott[384] found that older patients expressed fears of walking on vinyl, a smooth surface similar to linoleum and tile. Furthermore, vinyl-covered floors produced slower gait

patterns in older people compared to when these people walked on carpeted floors[384]. Vellas *et al*.[37] postulated that falls and fear of falling caused older people to reduce their gait speed on slippery surfaces in order to decrease their potential to fall.

Typically, home falls due to slips occur in the bathroom, kitchen and laundry, rooms that traditionally have tile, vinyl or linoleum floor coverings (see Section 2.2.2). When coupled with socks or shoes with low traction, linoleum or tile surfaces may be slippery when dry but are especially so if they become wet[659]. Linoleum and tile surfaces may be deemed more slippery than carpet surfaces because they are relatively smooth and slip resistance decreases as surface roughness decreases[87,489,660]. Furthermore, wetness reduces static and dynamic coefficients of friction on smooth surfaces[90] to levels potentially low enough to pose a significant slip hazard[87]. Therefore, smooth tile and linoleum surfaces pose a high risk factor for falls due to slips in older people both from decreased frictional properties and possible fear of falling evoked gait changes (see Section 2.3.2(B)). However, few researchers have investigated the interaction between different types of household footwear and common household surfaces, such as tile and linoleum, as well as how these interactions affect the gait of older persons and, in turn, their risk of slipping. Furthermore, over half of the respondents in the present study also wore their household shoes outside the home, where surface frictional properties vary to those indoors. Therefore, either household shoes will need to be designed with soles that have good coupling capabilities with both indoor and outdoor surfaces or, older people should be encouraged to change their household shoes before going outside.

3.4 Survey Summary and Conclusions

The primary purpose of this survey was to characterise requirements and attitudes of older persons living independently in the community in relation to their household footwear wearing and purchasing habits. The survey also identified characteristics of older people who suffer falls, foot problems and require specialised

footwear, to provide the subject selection criteria for Experimental Section B, as well as the specific household footwear and surface types to be examined. To achieve this purpose, a custom survey was completed by 60 men and 68 women who closely resembled the older community-dwelling population in terms of gender, age, self-perceived state of health, mobility, independence, income and falls incidence. Furthermore, consistent with a community-dwelling population, the current sample was composed predominantly of pensioners with a restricted income who perceived themselves to be in relatively good health, were mobile and were independent in terms of performing daily living activities.

In agreement with Hypothesis (1), the medical condition most frequently reported by the survey sample was arthritis, a medical condition that affects the musculoskeletal system and mobility. Furthermore, as proposed in Hypothesis (2), respondents in the present survey frequently reported foot problems and foot pain and/or discomfort, suggesting that the foot problems reported were associated with foot pain and/or discomfort. The incidence of arthritis and foot problems have the potential to limit mobility and make older individuals more dependent on others because of the consequential joint pain, joint deformity and altered joint loading, placing the individual at an increased risk of falls. Therefore, individuals with arthritis and foot problems would appear to benefit from specialised footwear, which could be designed to fit the often-deformed shape of their feet while still providing a safe base for gait within the home.

In partial agreement with Hypothesis (3), the most common footwear type worn around the home by the present respondents were less structured slippers, which were worn for most of the day or greater than 5 hours per day. However, unexpectedly a large number of respondents also went barefoot around the home, most probably reflecting the warm climatic conditions at the time of the survey. Slippers were purchased and worn because they were comfortable, fitted the varied shaped feet of the respondents, were easy to don, had no fastenings and were inexpensive. Although the wearing of slippers is usually discouraged by researchers and health professionals, it is postulated that slippers may provide some fundamental design characteristics that should be incorporated into safe household shoes designed for older individuals, particularly those suffering arthritis and associated foot problems, as more restrictive

footwear is often difficult to fit their unique foot shape. Whether such unstructured footwear cause gait changes that may predispose older people to falls requires further investigation.

In agreement with Hypothesis (4), smooth surfaces commonly found within the home, such as tiles and linoleum, were perceived as the most slippery surfaces, particularly when wet, whereas carpet was deemed the least slippery. These smooth surfaces are common household surfaces traditionally found in bathrooms and kitchens, where many slips occur. Despite being deemed slippery when the respondents performed daily living activities while wearing common types of household shoes such as slippers, no research was located investigating the interaction between these household surfaces and different slipper types. Surfaces deemed as slippery can also invoke fear of falling gait adaptations further increasing the risk of falls. Therefore, research investigating the effects of walking over smooth surfaces, particularly when wet, on the gait of older persons while wearing household slippers is recommended.

Consistent with Hypothesis (5) women in the present study reported a trend towards more falls, significantly more diagnosed medical problems and a greater amount of foot pain and/or discomfort compared to men. Furthermore, more women than men suffered from arthritis and reported foot problems, perhaps due to women wearing inappropriate footwear moderated by fashion throughout their life. Women reported wearing shoe types that could be slipped on to their feet whereas men reported wearing shoes with fastenings. Furthermore, a greater number of women reported surfaces as slippery when coupled with household shoes compared to men. Therefore, compared to older men, older women appear to have a greater need for specialised footwear to limit foot pain and deformity and ensure adequate mobility to remain living independently.

In conclusion, arthritic women represented a group of older people who suffer foot problems, who wear household shoes typically characterised as unsafe and who are at risk of falls, particularly falls resulting from slips on slippery household surfaces. Therefore, research examining how older arthritic women alter their walking patterns when wearing different slippers on common household surfaces, particularly surfaces perceived as slippery, will be conducted in Experimental Section B. Information from such research is required before a slipper can be recommended as safe for older people.

EXPERIMENTAL SECTION B:

Slipper-surface interactions and the gait of older rheumatoid arthritic women

Chapter 4

The Problem

4.1 Introduction

Several studies pertaining to the biomechanics of gait and falls have been performed to identify and classify both hazardous and safe shoe-floor conditions to assist in designing safer environments (see Section 2.3.2). However, most of this research to date has had an occupational emphasis with little research focussed on older individuals, household floor surfaces or household footwear types. Based on the conclusions stated in Experimental Section A, it was recommended that the interaction between unstructured household shoes (slippers), slippery household surfaces and the gait of older women with RA, a population known to have foot problems and require specialised footwear, is conducted. As opposed to falls in general, which can result from a multitude of causative factors (see Section 2.3), slips are dependent upon the frictional characteristics of different footwear-surface interactions, the frictional demand of the movement being performed, namely gait, and the individual's perception of the footwear-surface interaction. Therefore, research specifically into how various household slipper-surface interactions affect the gait of older women with RA at initial foot-ground contact is warranted as this is the event in the gait cycle where slips are deemed most problematic (see Section 2.2.3). It is only after this relationship is properly understood that recommendations for the design of safe household footwear for older women with RA can be developed.

4.2 Statement of the Problem

The purpose of Experimental Section B was to compare the effects of different household slipper-surface interactions on the biomechanical parameters characterising initial foot-ground contact in older women with RA, to provide information about which household slipper would be safe for older people.

4.3 Hypotheses

Based on the results of Experimental Section A (see Section 3.3) and in conjunction with previous research (see Chapter 2), the following research hypotheses were formulated.

(1) In the preliminary assessment tasks, RA subjects would report a higher incidence of foot problems and foot pain together with reduced plantar sensation, knee and ankle muscle strength and knee and ankle joint range of motion but increased foot reaction times as well as increased static and dynamic plantar pressures, particularly in the forefoot region, when compared to the control subjects. These between-group differences would occur as a direct result of the RA disease process.

(2) When compared to control subjects, during the walking trials RA subjects would display the following gait differences at initial foot-ground contact as a direct result of their RA:

(i) altered kinematic variables such as decreased horizontal heel velocity, decreased foot/shoe angle, decreased knee flexion, increased ankle plantar flexion, decreased foot/shoe angular velocity and increased stance time;

(ii) altered kinetic variables such as decreased braking forces in both the vertical and anteroposterior directions, increased time between initial foot-ground contact and the braking ground reaction force peaks, reduced knee and ankle joint moments, decreased knee and ankle joint powers, increased static and peak dynamic coefficients of friction and increased time between initial foot-ground contact and the dynamic coefficient of friction;

(iii) altered neuromuscular variables such as earlier activation of the hamstring, peroneus longus and gastrocnemius muscles together with a delayed onset of the quadriceps and tibialis anterior muscles, leading to altered muscle burst durations as well as increased intensity of hamstring, peroneus longus and gastrocnemius muscle bursts and reduced intensity of the quadriceps and tibialis anterior muscle burst; and

(iv) perceptions of increased task difficulty and foot pain.

(3) Irrespective of subject group or surface type, when walking in toe slippers compared to closed back slippers and when walking in both slipper types compared to barefoot, subjects would display changes to their gait at initial foot-ground contact similar to those documented in Hypotheses (2(i)) to (2(iii)), as well as increased vertical and anteroposterior ground reaction forces and perceived decreased shoe comfort and increased task difficulty during the walking tasks. These gait adaptations are anticipated as subjects adapt to wearing slippers that both enclose the foot (closed back slippers) and expose the heel (toe slippers). It was anticipated that when barefoot, subjects would display a regular gait pattern, as described previously (see Section 2.3.1(F)).

(4) Irrespective of subject group or footwear type, when walking on a wet vinyl tile surface compared to a dry vinyl tile surface and when walking on a vinyl tile surface compared a carpet surface, subjects would display changes in their gait at initial foot-ground contact similar to those documented in Hypotheses (2(i)) to (2(iii)), together with increased horizontal heel slide, and perceived increased surface slipperiness and task difficulty during the walking trials. These gait adaptations would indicate an attempt to achieve a more stable and regular gait pattern (see Section 2.5). Changes contrary to these, which are characteristic of irregular gait patterns, would contribute to an increased slip risk.

(5) There would be significant subject group x footwear type interactions such that the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking barefoot or wearing a specific slipper would be dependent upon subject group.

(6) There would be no significant subject group x surface type interactions such that the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking on a specific surface would not be dependent upon subject group.

(7) There would be significant footwear type x surface type interactions such that the biomechanical parameters characterising initial foot-ground contact and the

subjective perceptions when walking barefoot or wearing a specific slipper would be dependent on the surface walked upon.

(8) There would be significant subject group x footwear type x surface type interactions such that the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking barefoot or wearing a specific slipper on a specific surface would be dependent upon subject group.

4.4 Limitations and Delimitations

4.4.1 Limitations

The following factors were acknowledged as limitations of Experimental Section B:

(1) Subjects were limited to independently living older women, aged 60 years and above, who volunteered to participate in the study. Therefore, the subjects did not constitute a random sample of the population and, as such, may not represent all older women in the community.

(2) As subjects volunteered to participate in the present study, the incidence of joint replacement could not be standardised across subject groups.

(3) Although no participant was in acute pain on the day of testing, the intensity of pain or disability each subject experienced on the day of testing could not be controlled and may have affected their normal walking patterns.

(4) The study was constrained to a laboratory and walking in a harness and, although familiarised with the assessment protocol before data collection, the walking patterns displayed by participants may not have been characteristic of their normal walking patterns, on different surface types within their own homes.

(5) Although all subjects were fitted with both the toe and closed back slippers on the day of testing, it was not possible to standardise whether subjects currently wore similar footwear types within their own home.

Therefore, some subjects may have become familiarised to the slipper types faster than others.

(6) Despite strict adherence to standard procedures for kinematic and electromyographical data collection, errors associated with the use of such procedures required the use of computerised data smoothing to help reduce these inaccuracies.

4.4.2 Delimitations

The following delimitations were imposed on Experimental Section B:

(1) Subjects were restricted to community-dwelling women, aged 60 years and above, who were either diagnosed with RA or had no incidence of RA. Therefore, the results of the survey are specific to this population.

(2) All subjects were informed about each footwear-surface condition and could therefore alter their gait accordingly. This may not truly represent a household situation where hazards may go unnoticed and are experienced unexpectedly.

(3) The study was limited to a two-dimensional kinematic analysis, reconstructed using external body markers, where the foot was considered a rigid link and the ankle a hinge joint; EMG analyses of seven muscles; and a kinetic analysis of the dominant lower limb, when performing the assessment task on wet and dry surfaces and when barefoot and shod with standard slipper types.

4.5 Assumptions

The following assumptions were implied in Experimental Section B:

(1) Each subject's body could be represented as a series of rigid links (link-segment model) interconnected by frictionless hinge joints and, as such, each segment had a constant length, a fixed mass located as a point mass at its centre of gravity, and a constant mass moment of inertia about its mass centre.

(2) Errors inherent in the two-dimensional sampled displacement data, resulting from marker movement, were considered minimal and were therefore adequate for use in subsequent calculations of kinematic variables such as velocity and acceleration, which were relevant components of the task.

Chapter 5

Materials and Methods

5.1 Subjects

Eight women from the Illawarra volunteered as experimental subjects in the present study. Subjects were included based on the following selection criteria:

(1) aged 60 years and above and living independently in the community;

(2) diagnosed by a rheumatologist as having adult onset RA for longer than 5 years;

(3) no surgical correction involving the bony structures of the feet as a result of the RA disease process in the past 5 years;

(4) ability to walk unassisted under all conditions in the study with no other major pathologies unassociated with arthritis which would significantly influence their gait; and

(5) medical clearance from a general practitioner to participate in the study.

People aged 65 years and above have high falling rates (see Section 2.2 and Section 3.3.2), numerous foot problems requiring specialised footwear (see Section 2.3.1(E) and Section 3.3.5), and a high incidence of arthritis (see Section 2.3.1(D) and Section 3.3.4). However, as older people are classified as those aged 60 years and above[661] and the retirement age, or age at which women qualify for the aged pension in Australia is 61 years of age (Department of Social Security, personal communication, 1995), female volunteers aged 60 years and above with diagnosed RA were included as subjects for the present study (see Section 3.4).

Patients with arthritis were selected for the present study as the results of Experimental Section A clearly identified arthritis as being the most common diagnosed medical condition reported by older community-dwelling people (see Section 3.3.4). The most common forms of arthritis are osteoarthritis, RA and gout[662]. However, compared to other arthritis types, RA is a systemic disease that affects the body bilaterally (see Section 2.6). Furthermore, RA affects women three times more often than men and, frequently involves the foot and ankle, leading to joint swelling, skin ulcers and neuropathies (see Section 2.6). These conditions contribute to widespread

foot pain, and persistent foot deformities, as were evident in the results of Experimental Section A (see Section 3.3.5), as well as reduced standing and walking ability (see Section 2.6.2 and Section 2.6.3). Patients with RA therefore have a need for specialised footwear such that the prescription of proper footwear is reported to be the most important aspect of non-operative care for RA patients[598] (see Table 2.3). For this reason, older women with RA were included as the experimental subjects in the present study.

Diagnosis of adult onset RA was confirmed by a rheumatologist using the 1987 revised criteria as stated by the American Rheumatological Association for the classification of RA[663] (see Table 5.1). Although radiographical damage has been documented to occur in RA patients after 2 years, the effects of RA are highly varied in patients who have had RA for less than 5 years (see Section 2.6). Furthermore, musculoskeletal damage, particularly in the feet has been shown to occur 5 years after RA onset (see Section 2.6.1). Surgical correction to the bony structures of the foot alters foot biomechanics and, consequently, the gait of RA individuals[593,602]. Therefore, all subjects in the present study had been diagnosed with RA for longer than 5 years, but had not had any surgical intervention to the bony structures of their feet in the 5 years, before participating in the study.

An additional eight women aged 60 years and above with no evidence of RA volunteered as control subjects. Control subjects were matched to the RA subjects for age, activity level and anthropometric characteristics. A general practitioner screened all subjects and gave or denied them permission to participate in the study based on their ability to complete the assessment tasks. All subjects were unpaid volunteers and provided written consent before testing. Ethical clearance for the study was received from the University of Wollongong Human Research Ethics Committee (see Appendix B.1) with all testing conducted according to the National Health and Medical Research Council Statement on Human Experimentation[1].

The method devised by Bach & Sharpe[664] was used to determine the sample size required to demonstrate a difference between the RA and control subjects or within RA and control subjects with adequate statistical power. Paired *t*-tests were completed using data from past studies, which have investigated older people and RA patients performing daily living activities in different footwear types[337,665-668]. Eight subjects per

group provided an estimated 80% power which, considering the complex data collection and analysis procedures, as well as the demands placed on the subjects in the present study, was considered appropriate for Experimental Section B[664].

Table 5.1 The 1987 revised criteria for classifying rheumatoid arthritis[663].

Criterion[a]	Definition
1. Morning stiffness	Morning stiffness in and around the joints, lasting at least 1 hour before maximal improvement.
2. Arthritis of three or more joint areas	At least three joint areas with simultaneous soft tissue swelling or fluid (not bony overgrowth alone) observed by a physician. Possible areas are elbow, wrist, metacarpophalangeal, proximal interphalangeal, knee, ankle or metatarsophalangeal joints, right or left side.
3. Arthritis of hand joints	At least one joint area swollen (as in 2.) in a wrist, metacarpophalangeal or interphalangeal joint.
4. Symmetric arthritis	Simultaneous involvement of the same joint areas (as in 2.) bilaterally (bilateral involvement of interphalangeals, metacarpophalangeals or metatarsophalangeals is acceptable without absolute symmetry).
5. Rheumatoid nodules	Physician observed subcutaneous nodules over bony prominences, extensor surfaces or in juxta-articular regions.
6. Serum rheumatoid factor	Demonstration of abnormal amounts of serum rheumatoid factor by any method for which the result has been positive in < 5% of normal control subjects.
7. Radiographic changes	Radiographic changes typical of rheumatoid arthritis on posteroanterior hand and wrist radiographs, which must include erosions or unequivocal bony decalcification localised in or most marked adjacent to the involved joints (osteoarthritis changes alone do not qualify).

[a] For classification purposes, a patient shall be said to have RA if he/she satisfies at least four of these seven criteria. Criteria 1. through 4. must be present for at least 6 weeks. Patients with two clinical diagnoses are not excluded. Designation as classic, definite or probable RA is *not* to be made.

5.2 Preliminary Assessment of the Subjects

5.2.1 Questionnaires

On the day of testing, each subject completed the questionnaire that was used in Experimental Section A (see Section 3.2.3 and Appendix A.2). The questionnaire was expanded to incorporate selected scales from the Arthritis

Impact Measurement Scales 2 (AIMS2)[669] to assess functional impairment and the Foot Function Index (FFI)[670] to obtain information pertaining to limitations in daily living activities due to foot problems and foot pain. Information derived from the questionnaire was used to confirm that the subjects met the selection criteria and to describe the RA and control subjects.

The AIMS2 (see Appendix B.2) was designed to assess health status in subjects with rheumatic diseases[669,671] and has been shown to be valid, reliable and accurate in assessing rheumatic patients[571,669,672-677]. The scales that form the physical function (mobility, physical activity, household tasks and self-care tasks) and pain components were selected in the present study as indicators of functional impairment and arthritis pain[676,678,679]. These scales contain four to five items with each item containing five possible responses[674]. The response options for the mobility, physical activity and pain scales ranged from “all days” to “no days” and the household and self-care task scales ranged from “always” to “never” [669]. The period of the responses was standardised by adding the phrase “During the past month…” to the beginning of each AIMS2 scale[669]. After coding, item responses were summed to produce scale scores and then standardised to a score of between 0 and 10, a higher score indicating greater impairment[674,680].

The FFI consists of 23 items grouped into three sub-scales, which measure foot pain, disability and activity limitation (see Appendix B.3) and, has been found to be valid and reliable[670]. All 23 items were measured on a 0 to 10 scale based on verbal anchors representing opposite extremes of the dimensions being measured. That is, the verbal anchors of “no pain” and “worst pain imaginable”; “no difficulty” and “so difficult, unable”; and “none of the time” and “all of the time” described the pain, disability and activity limitation sub-scales, respectively. The total item scores for each sub-scale were divided by the maximum score possible for each sub-scale and were then multiplied by 100 so the sub-scale scores ranged from 0 to 100. The average of the three sub-scale scores represented total foot function with higher total and sub-scale scores indicating greater impairment[670].

5.2.2 Physical Assessment of the Subjects

(A) Lower Limb Dominance

Data collection during the experimental protocol was restricted to an analysis of each subject's dominant lower limb. Therefore, lower limb dominance was determined for each subject according to the lower limb the subject's self-selected to kick a stationary soccer ball which was placed in front of them[313,681-683]. Limb dominance has been found to affect slip mechanisms[684] and strength measurements[683]. Furthermore, older individuals may show asymmetrical movement patterns when walking[685,686], particularly if diseased or incapacitated[687]. However, due to equipment restrictions, the risk of subject fatigue and the bilateral effects of RA (see Section 2.6), performing a full kinematic, kinetic and electromyographic analysis of the dominant limb was considered appropriate for the present study[688].

(B) Height and Body Mass

The height of each subject was measured to the nearest 0.1 mm using a Seca Model 220 stadiometer (Lafayette Instrument® Company, Indiana, USA) while the subject stood barefoot in the anatomical position. Body mass was recorded, with each subject barefoot and in minimal clothing, to the nearest 0.5 kg using calibrated BW-150 Freeweight precision balance scales (Colonial Scales, New South Wales, Australia; DC: +6 V; 150 kg x 0.5 kg capacity). Each measure was completed three times and the average data were used to describe the subject sample as well as to later assist in calculating and normalising kinetic data (see Section 5.6.3).

(C) Lower Limb Segmental Proportionality

The lengths and circumferences of each subject's thigh, leg and foot segments were recorded to 0.1 mm using a Harpenden anthropometer (Holtain Ltd, Crosswell, UK) and to 0.1 cm using a Harpenden retractable steel tape measure (Holtain Ltd, Crosswell, UK), respectively. These measures were taken

while subjects lay supine following the methods described by Zatsiorsky *et al.*[689]. Although this method was devised for younger people (mean age, 23.9 years), it can also be used for subjects with a different physical stature[690]. Each measurement was recorded three times, with the averages used later to calculate biomechanical lengths and to estimate the segmental mass and inertial parameters of each segment (see Section 5.6.1).

(D) Slipper Size

The Brannock® foot-measuring device (Brannock Device Co, New York, USA) was used to accurately measure the size of each subject's foot to determine correct slipper size. Subjects placed one foot on the device while standing with their body mass distributed equilaterally on both feet and three measurements were recorded[691]. Heel-to-toe length was measured using the numbers on the footplate while arch length (heel-to-ball) was measured by placing the pointer of the device over the 1st metatarsal joint and reading the numbers adjacent to the pointer. Shoe size was then determined by using the larger measurement from the recording of arch length and heel-to-toe length. Foot width, and thus shoe width, was recorded by sliding the width bar firmly to the edge of the lateral aspect of each subject's foot at the metatarsal head. Both feet were measured and shoe size was fitted to the larger foot[691]. However, despite being correctly fitted for their slippers, three of the eight RA patients required larger slipper sizes to accommodate their foot deformities.

(E) Lower Limb Strength

Bilateral isometric knee flexion and extension and ankle dorsi- and plantar flexion strength for each subject were measured using the Nicholas Manual Muscle Tester (MMT; Model 01160, 9 V, Lafayette Instrument® Company, Indiana, USA), a factory calibrated hand held device, which quantifies the peak force required to break an isometric contraction[692]. Strength tests using the MMT have been found to display good reliability and

validity[693,694] when assessing proximal muscle strength[95] and maximal strength effort for a specific motion[692].

For strength testing the subjects were initially seated on a plinth in 90° hip and knee flexion, no hip rotation or abduction, and their legs hanging freely. Knee flexion and extension muscle strength were then measured with the stirrup of the MMT placed 10 cm above the lateral malleolus on the dorsal and ventral aspect of the leg, respectively. Ankle dorsi- and plantar flexor muscle strength were assessed with the subjects supine in 0° hip flexion, rotation and abduction and 0° knee extension. The stirrup of the MMT was then placed 2 cm proximal to the hallux on the dorsal and plantar surfaces of the foot, respectively. Subjects were positioned so that gravity did not influence the strength measures. For each test of muscle strength, a gradual resistive force was applied to the test limb through the stirrup of the MMT over 1 second. This gradual force allowed the subject to adjust and recruit the maximum amount of muscle fibres[692]. Additional force was then applied to the limb over 2 seconds until the muscle contraction started to "break" and the limb began to move. After one familiarisation attempt, three maximum voluntary contractions were completed for each test and the peak force was then recorded as the value characterising individual muscle strength. Knee flexion and extension and ankle dorsi- and plantar flexion strength were assessed as reduced strength in these muscles have been displayed by older people who fall[45,145-147,149,190,695].

(F) Lower Limb Range of Motion

A plastic goniometer (Lafayette Instrument® Company, Indiana, USA, 2° increments) was used to measure the range of motion of both the right and left knee and ankle joint complexes for each subject. Goniometric measurements of joint range of motion display good reliability[696,697] and are advocated for measuring joint range of motion in RA patients[479]. Furthermore, it has been reported that one measurement is as reliable as taking the average of repeated measurements in one session for a single examiner[696]. All range of motion testing was conducted with subjects in a supine position with their lower limbs

extended and 0° hip rotation and abduction. Knee range of motion was recorded by having the axis of the goniometer positioned over the lateral femoral condyle with the proximal arm aligned with the lateral malleolus and the distal arm aligned with the greater trochanter. Force was then applied to the dorsal surface of the leg to move it as close to the subject's buttocks as possible to measure total knee flexion or to the ventral aspect of the subject's leg to measure total knee extension. Ankle range of motion was recorded by having the axis of the goniometer positioned over the distal aspect of the lateral malleolus with the proximal arm aligned with the midline of the head of the fibula and the distal arm aligned parallel to and above the lateral midline of the 5th metatarsal. Force was then applied just proximal to the subject's toes on the plantar surface of the foot for ankle dorsiflexion and on the dorsum of the subject's foot for ankle plantar flexion. After one familiarisation trial, all flexibility assessments were performed three times with the greatest amplitude of movement recorded as joint range of motion.

(G) Reliability of Physical Assessments

All preliminary physical assessments were recorded using the same equipment, by the same experienced researcher (the Chief Investigator) who was proficient in conducting each test. Intrarater reliability of each testing protocol was established by measuring the same physical dimensions for three people on three consecutive days. As the intraclass correlation coefficients[698] exceeded 0.93 for each of the physical assessment tests, the results obtained by the researcher were considered highly reproducible and, therefore, reliable (see Appendix B.4).

5.2.3 Assessment of Foot Functionality

(A) Foot Reaction Time

Reaction time during many daily living activities declines with age (see Section 2.3.1(C)). Older people with slower reaction times are also at a greater

risk of falls and fractures[122,149,187,190,198,695]. Foot reaction time was therefore assessed in the present study using a simple reaction time device (9 V) that used a light emitting diode (LED) as a stimulus and depression of a switch (by the foot) as a response (Prince of Wales Medical Research Institute, New South Wales, Australia). This method of assessing foot reaction time has been found to be valid and reliable in older people[149,183]. Subjects were seated with the reaction timer placed under the ball of their foot. The LED was set to illuminate at random periods and when depressed would switch off. After familiarisation, subjects performed the foot reaction time test 10 times with the mean of the 10 trials then calculated to indicate foot reaction time in milliseconds.

(B) Plantar Sensation

Sensory loss, reflecting a loss of integrity in the somatic system[185], has been implicated both in falls[699] and in the normal ageing process[166,700] (see Section 2.3.1(C)). Therefore, each subject's loss in plantar sensation was assessed in the present study.

Touch sensitivity was measured using Semmes-Weinstein monofilaments while each subject lay supine with the plantar aspect of their foot perpendicular to the floor[169]. The use of Semmes-Weinstein monofilaments has been reported as an accurate and reliable quantitative test to detect early nerve compression[701-703]. Three Semmes-Weinstein monofilaments with ratings of 4.17, 5.07 and 6.10 were applied perpendicularly to the plantar surface of each subject's right and left feet at four sites per foot and pushed to obtain a C-shaped deformation[193,702]. The monofilament ratings are expressed as the log of 10 times the buckling force in grams[185] and the ratings of 4.17, 5.07 and 6.10 were representative of normal sensation, protective sensation (the ability to perceive potential external sources of injury) and loss of protective sensation, respectively[262,704,705]. In fact, an inability to sense the 5.07 (10 g force) monofilament at any site has been defined as peripheral neuropathy[706]. The monofilaments were applied to both feet at the hallux, 1st and 5th metatarsal heads and the heel[185,193]. Subjects were required to verbally respond to three of five repetitions correctly at each site in order to be assigned that monofilament rating[193].

(C) Plantar Pressure

Static and dynamic plantar pressures generated by each subject were quantified using the emed-AT/4 pressure distribution platform (novel$_{gmbh}$, Munich, Germany, 582 x 340 x 20 mm; 110 V; 4 sensors.cm^{-2}). Plantar pressure measures quantify the actual forces and pressures applied to each region of the plantar surface of the foot and therefore provide an indication of foot deformities and variations in foot form[707]. The emed® system portrays accurate, reliable and objective dynamic plantar pressure distribution characteristics of different foot disorders[137,708] and was therefore used in the present study to characterise the effects of RA on plantar pressure distributions. Both static and dynamic plantar pressure data were analysed to provide an indication of foot structure, discomfort and function.

The emed-AT/4 platform was placed on a firm surface, levelled and surrounded by dense foam mats allowing a continuous surface. Static plantar pressures were initially recorded when subjects stood in a relaxed anatomical position with one foot on the platform and the other foot on the dense foam mat adjacent to the platform with their body weight evenly distributed over both feet. Dynamic plantar pressures were then measured using the two-step method[709] whereby subjects would contact the platform on their second step. This method was chosen as it has been found to be valid, accurate and reliable to assess dynamic plantar pressures in older people[709,710]. Two trials of data were collected at 25 Hz for both the static and dynamic tests using WinEmed 1.18e software (novel$_{gmbh}$, Munich, Germany).

Each static footprint was divided into 10 regions contained within the Cavanagh mask[711] using novel-ortho automask software (Version 9.35; novel$_{gmbh}$, Munich, Germany). The 10 masked regions of the foot (see Figure 5.1) were manually adjusted for each subject and included: the lateral heel (M01), medial heel (M02), lateral midfoot (M03), medial midfoot (M04), 1st metatarsal (M05), 2nd metatarsal (M06), 3rd-5th metatarsals (M07), hallux (M08), 2nd phalange (M09) and 3rd-5th phalanges (M10). These regions are commonly affected by RA, often requiring surgical intervention to alleviate symptoms (see Section 2.6.2). Data analysis was then performed for each static footprint to

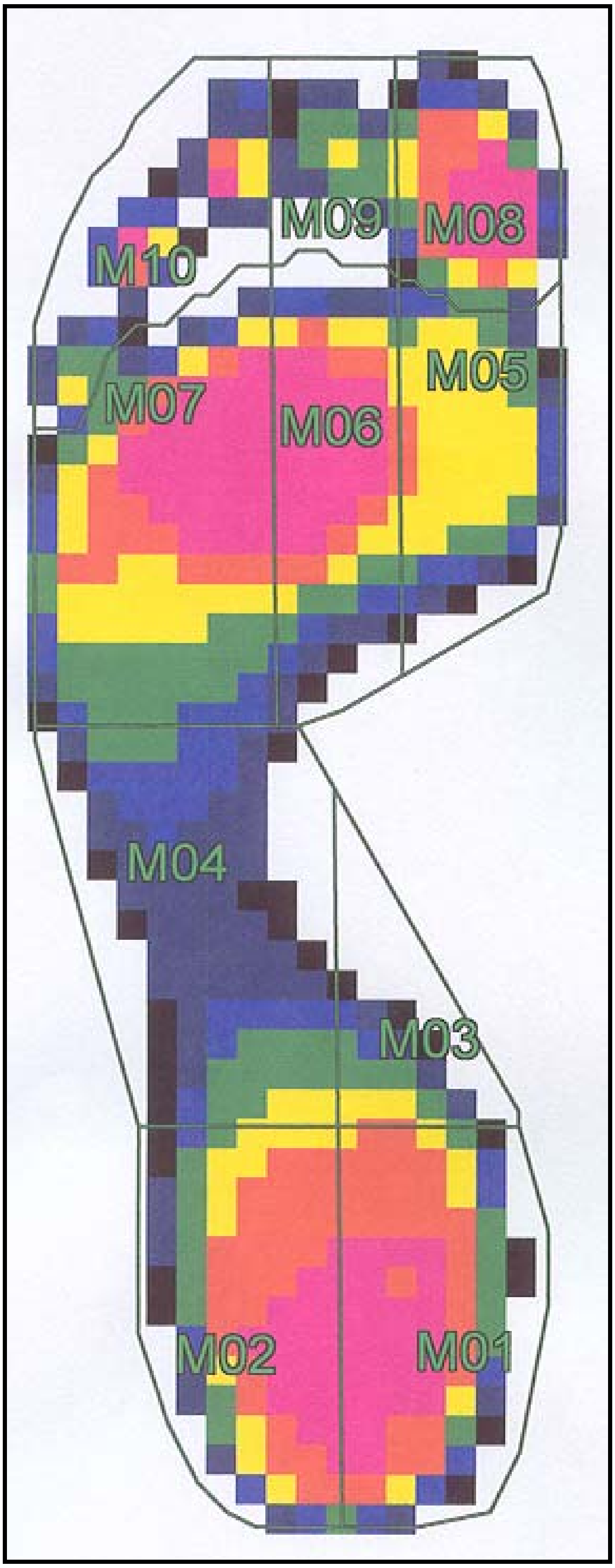

Figure 5.1 Maximum pressure picture displaying the 10 regions of the Cavanagh mask[711].

calculate the peak pressure (N/cm^2), peak force (N) and maximal active area (cm^2) for both the total and masked foot from the maximum pressure picture (MPP). Dynamic footprint analysis also used the MPP to calculate the peak pressure (N/cm^2), peak force (N) and maximal active area (cm^2) and the time at which they each occurred during the rollover process (%). The pressure-time integral (N.s/cm^2) and force-time integral (N.s) for both the total and masked MPP were also analysed[460,708,712]. Peak pressure and peak force were selected for analysis as they represented the highest pressure and force under the foot, respectively, at any time during foot contact[713]. The maximal active area was analysed as it identified the maximum area of the plantar surface of the foot in contact with the plate at any time[713]. The integrals, calculated by multiplying the pressures and forces by the times they occurred[713], were analysed because of their importance in terms of indicating skin ulceration (pressure-time integral) and bone fatigue (force-time integral)[714]. These plantar pressure variables were chosen for analysis to determine whether the feet of the RA subjects displayed the foot deformities characteristic of RA as reported in the literature (see Section 2.6.2). All data analysis was performed using novel-win multimask software (Version 8.32; novel$_{gmbh}$, Munich, Germany).

5.3 Data Collection Techniques for the Walking Trials

5.3.1 Experimental Task

For the experimental task, each subject was required to walk at a self-selected pace along an 8 m walkway under three footwear conditions (barefoot, toe slippers and closed back slippers) combined with three surface conditions (carpet, dry vinyl tile and wet vinyl tile) resulting in a total of nine conditions, which are described in the following sections. Condition order was randomly generated for each subject and the subjects knowingly encountered all conditions. Although no restrictions were placed on the subject’s walking motion, each trial was commenced when subjects had their feet together, arms by sides and eyes looking straight ahead; and completed when subjects reached the opposite end of the walkway. Subjects were also required to wear a harness

attached to a custom-designed monorail system (see Section 5.4) throughout the walking trials for safety. As a constant gait velocity is typically reached within two to three steps of gait initiation and maintained until two to three steps before gait termination[715,716], each subject walked approximately 6 m, ensuring a minimum of two steps before and after force platform contact, and therefore a consistent gait pattern. In addition to the falling practice completed to ensure confidence in the harness system (see Section 5.4), each subject completed up to three familiarisation trials, whilst wearing the full-body harness, to ensure they understood the requirements of, and were able to complete, the walking task.

Many researchers have started to investigate how subjects, both younger and older, react to, and recover from, a slip and therefore the research protocols have attempted to induce slips[431,505,538,717]. However, rather than cause slips, the present study focused on how subjects modified their gait with respect to changing footwear and surface conditions to avoid slips, specifically when preparing for initial foot-ground contact, the time of highest slip risk (see Section 2.2.3). Therefore, the experimental task was designed so that subjects were fully informed of the footwear and surface conditions so that they could anticipate the slipperiness of the conditions and modify their gait if required.

(A) Footwear Conditions

The three footwear conditions examined in the present study included two experimental footwear conditions and a control condition in which subjects walked barefoot. For the two experimental footwear conditions, subjects wore toe slippers (“Julie”, Grosby Footwear, Pacific Brands Holdings Pty Ltd, Victoria, Australia; 16 to 22 g) and closed back slippers (“Rhonda”, Grosby Footwear, Pacific Brands Holdings Pty Ltd, Victoria, Australia; 18 to 22 g). Both slipper types are depicted in Figure 5.2. Grosby slippers were chosen for the present study as they represented the leading brand of slippers for family value ranging from $15 to $25 at department and variety stores. “Grosby” was also commonly reported by survey respondents in Experimental Section A as the choice of slipper brand worn, particularly as respondents considered that Grosby slippers were made in Australia. Apart from the toe slipper only having an

upper with a toe box and the closed back slipper including an upper with both a toe box and a heel counter, both slippers were the same in design, colour, material make-up (synthetic fabric upper, sock lining, thermoplastic outsole and 1.3 cm heel height) and sole tread pattern. Furthermore, as normal shoe wear considerably affects the frictional properties of the shoe, each subject wore brand new slippers. The two slipper types were chosen for the present study as survey respondents in Experimental Section A of this thesis (see Section 3.3.6) and older people who present to hospital with a fractured neck of femur[23,378] commonly report wearing one of these slipper styles around the home.

Figure 5.2 The closed-back slipper (A) and toe slipper (B) worn by subjects in the present study.

(B) Surface Conditions

Tile and linoleum surfaces were considered by survey respondents in Experimental Section A to be slippery (see Section 3.3.8) and have been implicated in falls in older people (see Section 2.2.3). These surfaces become increasingly slippery when wet and are considered problematic for many older people. Therefore, a vinyl tile surface (Sommer, BO 0504 07, Sommer, New South Wales, Australia) was used as the experimental surface in the present

study under two conditions, wet and dry. For the wet condition, water was applied in a fine mist, using a standard garden sprayer, onto the entire walkway ensuring an even spread of water. The control surface condition was that of carpet (anti-static Martinique Mark II loop pile Berber carpet, Daybreak Grey 71, 100% Olefin B.C.F. Yarn; Impressions by Beaulieu, New South Wales, Australia), as it was not considered slippery by survey respondents in Experimental Section A (see Section 3.3.8). Carpet is also a preferred surface for falls and injury prevention[383,384]. Both surfaces were untreated, of similar colour and thickness, light, easy to move between trials and laid directly over the regupol surface of the laboratory floor. During the walking trials, each surface covered the entire walkway, with a separate embedded surface piece attached directly to the force platform, to ensure a consistent gait pattern (see Figure 5.3). Furthermore, the surface joins were camouflaged as well as possible to minimise targeting of the force platform by subjects.

5.3.2 Kinematic Data Collection

The three-dimensional motion of each subject's dominant lower limb was quantified during gait using an OPTO***TRAK***® 3020 motion analysis system (Northern Digital Inc., Ontario, Canada). Twelve encased infrared emitting diodes (IRED; 7 or 6 g; 16 mm or 8 mm diameter) were connected to two 6-channel strober units (57 mm x 77 mm x 24 mm; 94 g) using standard co-axial cables (1.2 m). Each strober unit was then connected to a receiver unit (60 mm x 120 mm x 34 mm; 16 g) via a strober cable, and both were powered by a battery pack (8.4 V; 28 g) enabling the IREDs to emit infrared light. The strobers, receiver units and battery packs were securely fastened to the harness worn by the subjects (see Section 5.4) to minimise any additional mass to be carried by the subjects while walking.

The 3020 System Control Unit (240 V) determined IRED activation by emitting infrared timing signals through a transmitter (127 mm diameter), connected to the system control unit via a 4-pin lemo head communication cable. These signals were detected through a black window on each receiver

Figure 5.3: The walkway surface (vinyl tile condition) with the force platform embedded in the middle of the surface.

unit. The three-dimensional IRED coordinates were detected by three one-dimensional charge coupled devices (resolution = 1:200,000; field of view = 34° x 34°) built into a 3020 Position Sensor (1127 mm x 216 mm x 315 mm; 36.4 kg; 240 V). The Position Sensor was factory calibrated to long focus deriving accurate (± 0.1 mm for *x*, *y* coordinates; ± 0.15 mm for *z* coordinate) and repeatable results (± 0.01 mm) between 2.2 m and 6 m away from the Position Sensor[718]. The Position Sensor was securely mounted and levelled 0.85 m above the floor, 3.4 m lateral to the long axis of the force platform. Therefore, the working area (1.5 m length x 1.9 m height) was large enough to capture the motion of each subject's dominant lower limb during initial foot-ground contact*, that is, from midswing to midstance. Coordinates from the Position Sensor were then relayed to the system control unit via a 10-pin lemo head communication cable.

Kinematic data were sampled at 100 Hz for 4 s using OPTO***TRAK***® COLLECT software (Version 2.003, Northern Digital Inc., Ontario, Canada) and stored for later analysis via an interface adaptor card housed within a Pentium III PC and connected to the system control unit via a 15-pin D-shell adapter cable. This sampling rate was chosen so that it would be possible to identify initial foot-ground contact more accurately than at lower frequencies[719].

(A) Attachment of Markers

To define the position of each subject's lower limb in three-dimensional space, 14 IREDs were attached over the skin of their dominant lower limb at landmarks that should have been easily detected by the Position Sensor[308,720,721]. Figure 5.4 displays and Table 5.2 describes the IRED locations. These IRED landmarks were positioned to define the thigh, leg and foot segments as well as the knee and ankle joint centres[722,723]. A further three IREDs were placed on three sensors of the force platform during the static trials to allow both the force platform and the subject to be placed in the global coordinate system. However,

* Initial foot-ground contact was defined as the frame of data when the vertical ground reaction forces exceeded baseline by 2% of the maximum force recorded.

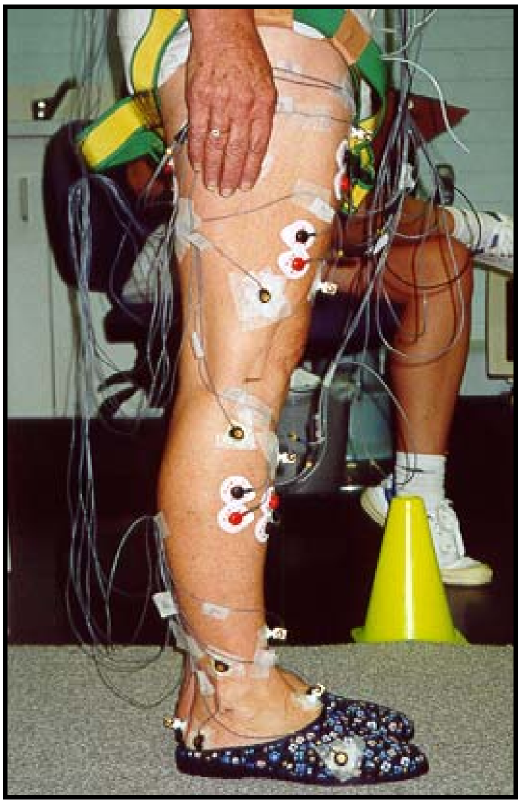

Figure 5.4: View of a subject's dominant lower limb with the IREDs attached.

Table 5.2: The anatomical landmarks upon which the IREDs were located.

IRED	Anatomical Location
1	Base of the 5th metatarsal when barefoot and the lateral aspect of the slipper upper corresponding to the base of the 5th metatarsal when shod.
2	Lateral inferior calcaneus when barefoot and when wearing the toe slipper but the lateral inferior heel counter of the closed-back slipper.
3	Posterior superior calcaneus when barefoot or when wearing the toe slipper but the posterior mid-heel counter of the closed-back slipper.
4	Superior apex of the navicular.
5	Apex of the lateral malleolus.
6	Lateral aspect of the leg, 3 cm above the lateral malleolus.
7	Anterior aspect of the leg, 5 cm above the lateral malleolus.
8	Anterior aspect of the leg at the base of the tibial tuberosity.
9	Lateral head of the fibula.
10	Apex of the lateral femoral condyle.
11	Lateral aspect of the thigh, 3 cm above the lateral femoral condyle.
12	Anterior aspect of the thigh, 5 cm above the superior pole of the patella.
13	Anterior aspect of the thigh, 20 cm above the superior pole of patella.
14	Lateral aspect of the thigh, 20 cm above the lateral femoral condyle.

as IREDs (5) and (10) and the force platform IREDs were hardwired to the system control unit, they were removed during the walking trials to minimise interference. The spatial coordinates of the IREDs were used to both describe the movements made at initial foot-ground contact (see Section 5.6.2) and as inputs for the inverse dynamics and power calculations (see Section 5.6.3).

(B) Reliability of Kinematic Measures and Marker Movement

A major limitation when quantifying the kinematics of human motion is the need to use external skin markers to simulate movements of the segments of the body. Errors in estimating skin marker location may occur due to the inertial properties and oscillations in the soft tissue and markers, as well as the soft tissue shifting over the joints[724,725]. Furthermore, movement of the skin across skeletal structures can differ between the proximal and distal aspects of a limb[726]. Studies by Amursky[727] and MacLeod & Morris[728] have reported that the magnitude of marker displacements can exceed 14 mm, depending on the movement type, marker location and subject morphology. Therefore, many researchers question whether external markers accurately portray motion of the underlying anatomical structures or segment as a whole[726,727,729,730]. Although efforts have been made to improve measurement techniques in an attempt to minimise skin movement artefacts[731], these artefacts cannot be eliminated unless markers are applied directly to the bones or through bone pins[732-734]. Therefore, using external skin markers placed at appropriate marker locations is currently the only feasible method for calculating kinematic variables when studying specific clinical populations[727]. Consequently, calculations of the kinematic variables and, in turn, the kinetic variables, via skin marker-based multi-link models should take account of skin movement artefact[733].

In the present study, IREDs were attached to the thigh, leg and foot/shoe segments. Markers attached to the foot/shoe and leg have been found to provide a good representation of lower limb skeletal motion during running[735]. Markers placed on the thigh, however, may not represent true femoral skeletal kinematics during running[735] because of muscle bulk and/or adipose tissue, which may undergo large relative movements[735,736]. The IREDs used in the present study

were small and light to reduce any potential inertial effects. In order to further minimise the effect of potential IRED movement in the present study, double-sided tape was used to affix the IREDs directly onto the subject's skin at locations away from joint centres and with minimal soft tissue[737]. Furthermore, both the IREDs and the leads extending from the IREDs were stabilised using surgical tape to reduce marker movement and to allow unencumbered motion. As subjects in the present study walked at slow velocities (see Section 2.3.1(F)), skin motion over the joint centres, adipose tissue oscillation and associated IRED movement were considered minimal.

To provide an estimate of marker movement in the present study, IRED movement was quantified by calculating the inter-marker distance for selected segments from the coordinates recorded during the walking trials and comparing these distances against the same inter-marker distances calculated when the subjects stood motionless. By using this method, total IRED movement in the present study was calculated to average 1.1 cm (range 0.3 to 4.1 cm; see Appendix B.5). These results are similar to past studies which have reported the largest skin marker movement to occur at the thigh (see Appendix B.5) recording movements of up to 3 cm[734] or 4 cm[736] in young subjects during gait. Furthermore, for a sample of lower limb-amputees, Zahedi and colleagues[738] reported that skin marker movement could lead to apparent changes in the length of the thigh segment of 3 cm to 4 cm during gait. Although an absolute change in segment length will not affect calculation of the kinematic variables (see 5.6.2(C)), marker movement in different planes will affect the accuracy of these data. However, marker movement is not uniform and therefore the accuracy of the calculated variables could only be estimated by further mathematical procedures. Therefore, the current IRED movement in the present study was considered acceptable for the sample and the movement performed.

5.3.3 Ground Reaction Force Data Collection

Ground reaction force signals generated by each subject during the walking trials were recorded using a calibrated Kistler Multichannel force platform (Type 9281B, Kistler Instrumente AG Winterthur, Switzerland; 600

mm x 400 mm). The force platform was placed on four steel mountings embedded into a concrete foundation according to manufacturer's specifications[739] and covered with the required surface (see Section 5.3.1(B)) so that it was level with the surface around it. The ground reaction force signals from four vertical (*y*), two anteroposterior (*x*) and two mediolateral (*z*) output channels of the force platform were then passed through a Kistler Multichannel Charge Amplifier (Type 9865A; Kistler Instrumente AG Winterthur, Switzerland), which converted the output signal of the platform into proportional electrical voltages (*y* = 10,000 pC; *x*/*z* = 5,000 pC). The amplifier was then connected to a junction box, which allowed the eight channels to be input to the OPTO***TRAK***® Data Acquisition Unit II (ODAU II; Northern Digital Inc., Ontario, Canada) via a ribbon cable. The ODAU II was connected to the system control unit via a 10-pin lemo head communication cable so that the ground reaction force signals could be synchronised with the capture of IRED coordinates.

The ground reaction force signals were collected for 4 s at 1000 Hz during both static and dynamic trials (see Section 5.5) using OPTO***TRAK***® COLLECT software (Version 2.003, Northern Digital Inc., Ontario, Canada). To ensure valid and reliable testing, the force platform was zeroed before each trial and calibrated following the manufacturer's specifications[739] before commencing data collection. The ground reaction force data were collected to enable later calculation of the centre of pressure location, the coefficient of friction and as input into the inverse dynamics and power calculations (see Section 5.6.3).

5.3.4 Recording Electromyographic Signals

Myoelectric signals were relayed from the bipolar electrodes placed on seven muscles of each subject's dominant lower limb to a Telemyo 8/16 Transmitter (Noraxon Oy, Cologne, Germany; 150 mm x 86 mm x 26 mm; 580 g; DC: +9 V; 580 g) via 1.2 m electrode leads[740]. The electrode leads were connected to the transmitter using a standard 8-channel input head with 10 piece 3-pole lemosa sockets. The input head was connected to the transmitter via a

25-pole D-SUB input socket, which interfaced with the signal sources of the amplifier card. The signals were then amplified ± 0.5 mV by a Telemyo 8/16 Amplifier card housed in the transmitter. The quadruple DIP switches of the amplifier card, housed within the transmitter, were set for an input level of ± 6.8 mV for each channel. The amplification setting was selected to allow for amplification of frequencies within the electromyography (EMG) signal with minimal signal distortion. Myoelectric signals were relayed from the transmitter to a Telemyo 8/16 receiver (AC: 220 V) via an antenna connected to the transmitter by an SMA connector (50 Ω). A magnet-mounted antenna attached to a steel surface[740], connected to the receiver using a BNC socket, detected the signal from the transmitter based on the used radio frequency. The receiver was then connected to a junction box, which allowed eight channels to be input to the ODAU II via a ribbon cable. Therefore, the collection of EMG signals was synchronized with collection of both the ground reaction force signals and the IRED coordinates. The EMG signals were sampled for 4 s at 1000 Hz (0 to 340 Hz bandwidth) using OPTO***TRAK***® COLLECT software (Version 2.003, Northern Digital, Ontario, Canada). The EMG signals were visually displayed and inspected before testing to check that no motion artefact was evident and clear meaningful signals were being received.

The transmitter was securely fastened around each subject's waist using a wide belt. Leads extending from the transmitter to the electrodes were secured using surgical tape to minimize electrode movement, movement artefact, and interference when the subject performed each walking trial[741]. The EMG signals were later used to calculate muscle activation patterns and to provide an indication of the muscular effort required when walking in each condition (see Section 5.6.4).

(A) Electrode Placement Sites and Preparation

Surface electrodes were placed on each subject’s dominant lower limb in a bipolar configuration over the muscle bellies of rectus femoris (RF), vastus lateralis (VL), semitendinosus (S), the long head of biceps femoris (BF), tibialis anterior (TA), peroneus longus (PL) and the medial head of gastrocnemius (G).

These muscles were selected due to their superficial location, their involvement in the gait cycle and their control of motion about the ankle and knee joints (see Section 2.5.4). Furthermore, the superficial location of these muscles assisted in reducing the possibility of cross talk[741-743] and increased the reliability of the EMG data collected during initial foot-ground contact[744]. The electrode placement sites for each muscle are detailed in Table 5.3. These electrode placement sites have been used previously in gait assessments[497] and were determined as the most appropriate sites for this sample of older women through extensive pilot testing. Before the electrodes were adhered, each site was confirmed by palpating the contracted muscle during movements representing the prime action of each individual muscle. A reference electrode was then placed over the medial femoral condyle, as few muscles are located in this area[741].

Table 5.3: The EMG electrode placement sites.

Muscle	Placement Site
RF	50% of the distance from the groin line to the proximal pole of the patella on a line extending from the anterior superior iliac spine to the middle of the patella.
VL	50% of the distance from the RF electrode placement site to the proximal pole of the patella and then 50% of the distance from this point to the lateral aspect of the thigh.
BF	50% of the distance along a line extending from the ischial tuberosity to the head of the fibula.
S	50% of the distance along a line extending from the ischial tuberosity to the medial epicondyle of the tibia.
G	The middle of the medial muscle belly found when subjects stood on the balls of their feet.
TA	25% of the distance along a line extending from the tibial tuberosity to the lateral malleolus.
PL	30% of the distance along a line extending from the head of the fibula to the lateral malleolus.

Each electrode placement site was prepared by shaving the site with a disposable razor, abrading with 3M® One-Step Skin Prep tape (3M®, New South Wales, Australia) and swabbing with diluted (50%) ethanol to remove both the dead cells on the surface of the skin and the skin's protective oils[745]. Miniature

silver/silver chloride surface electrodes (3M® Infant Pellet Electrodes, 3M®, New South Wales, Australia) were then arranged in a bipolar configuration (2 cm inter-electrode distance) directly over the prepared placement sites parallel to the line of action of the muscle fibres[746]. The impedance of the skin was then measured using a Cardiometric Artifact Eliminator® (Model CE01, Cardiometrics, Victoria, Australia), and was considered adequate if below 6 kΩ. Before testing, the source of the electric signal was checked using manual resistance tests[747], which would only elicit a response in one of the tested muscles and provide evidence of cross talk. Cross talk was minimised in the present study by using an appropriate electrode size with a small detection area[741,743,747] in a bipolar electrode configuration[741,747,748] and minimising the inter-electrode distance[743,747]. Furthermore, in subjects with excessive subcutaneous tissue, some muscles could not be monitored with accuracy and so were not included in data collection[743,748].

5.3.5 Subjective Estimations of Task Difficulty, Foot Pain, Shoe Comfort and Shoe/Surface Slipperiness

Following completion of each trial, subjects were asked to subjectively rate the task difficulty, foot pain intensity, shoe comfort and shoe/surface slipperiness. These subjective ratings were recorded using a Visual Analog Scale (VAS)[749]. The VAS is a 10 cm long white plastic card with a moveable indicator, the ends of which are marked with verbal anchors, on opposite ends of a continuum. The verbal anchors for measuring task difficulty were “no difficulty” and “so difficult, unable”; for foot pain intensity: “no pain” to “worst pain possible”; for shoe comfort: “uncomfortable” to “extremely comfortable”; and for shoe/surface slipperiness: “not slippery” to “extremely slippery”. Subjects were instructed to indicate their rating for each variable by moving the indicator to a position on the line between the two extremes. On the rear of the VAS, out of sight of the subject, was an ordinal scale from 0 to 10, to give each rating a quantitative value. To minimise bias based on previous responses, subjective measures were asked in a random order[750,751]. Baseline ratings were

taken before the trials while each subject was standing comfortably. Subsequent estimations were then normalised to this baseline rating.

The VAS was selected for use in the present study as it has been widely used to assess subjective states[752] in several different subject groups[544,752-755] and is considered simple, robust, sensitive and reproducible[749], as well as superior to other methods of determining subjective state[756,757]. Together with the assessment of subjective pain intensity[544,752,754,755], similar scales have been used to record subjective feelings of shoe comfort[460], slipperiness[85,422,426,461] and task difficulty[750,751,758].

5.4 Safety Monorail and Harness Design

During all walking tests, the subjects wore a commercially-available fully adjustable total body fall-arrest harness (Proflex, 011-1MP, Fallright International Pty Ltd, Queensland, Australia; 1.88 kg) to prevent floor contact if they fell during a trial (see Figure 5.5). The harness was suspended from a custom-designed overhead moveable trolley mounted on a monorail track via two attachments from a spreader bar (099-15, Fallright International Pty Ltd, Queensland, Australia; 74 g; see Figure 5.5). The Proflex harness was manufactured and in-house tested to meet the requirements of the Australian Standard 1891.1 - Safety belts and harnesses (see Appendix B.6). The overhead trolley and monorail system was designed by Forbes Rigby Pty Ltd (Wollongong, Australia), after significant input from the Chief Investigator as to the specifics of the design, and constructed by Department of Biomedical Science Technical Staff (University of Wollongong, Wollongong, Australia). Diagrammatic details and certification documents pertaining to the overhead trolley, monorail system and harness are presented in Appendix B.7 and Appendix B.8.

Each subject was fitted into the harness, which took approximately 5 minutes, and the harness was then attached to the spreader bar which, in turn, was attached to the trolley. Once appropriately fitted, all subjects practiced walking and falling in the harness system until they were confident that they would not sustain an injury if they fell during the trials, before the trials commenced. Similar harness and monorail designs have been used in past studies and have not been found to affect walking and

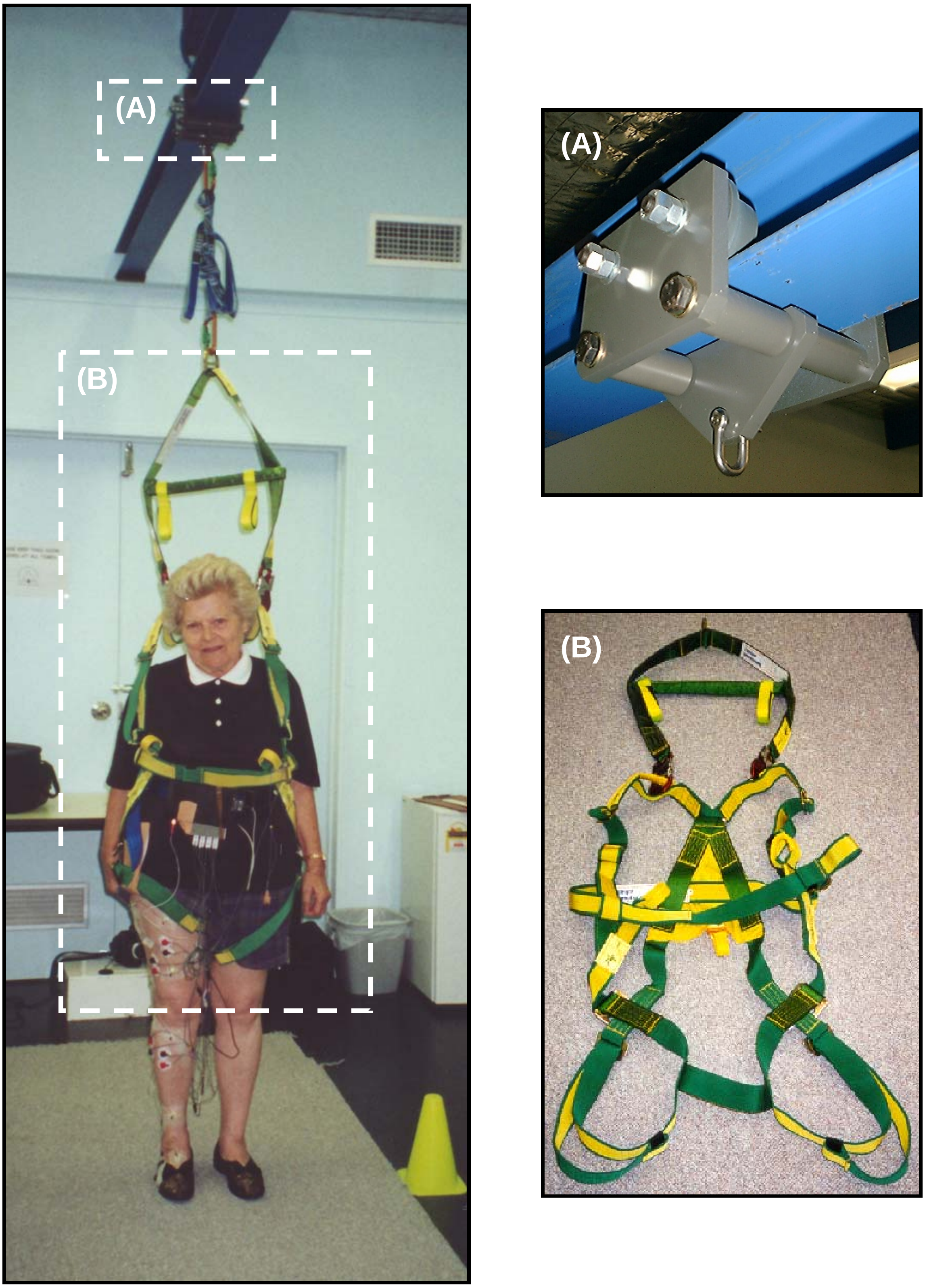

Figure 5.5 A subject ready to perform a walking trial under the closed back slipper and carpet condition wearing the Proflex harness (B) attached to the trolley (A), via two shoulder attachments on the spreader bar, on the monorail.

balance patterns in older and/or disabled individuals or postural reactions evoked by perturbation[158,759-761]. Furthermore, as the harness was worn by the subjects in all trials, any gait adaptations caused by the harness and monorail system would be consistent across all testing conditions and should not mask any differences due to the test conditions (see Section 5.3.1).

5.5 Data Collection Schedule

All testing was conducted in the Biomechanics Research Laboratory, University of Wollongong. Upon arrival at the laboratory, each subject received and read the Subject Information Package, signed the Informed Consent, and was assisted to complete the modified footwear questionnaire. Wearing minimal clothing, each subject's height and body mass were then taken, their dominant lower limb was established and their lower limb segment proportionality recorded. Following this, foot reaction time, lower limb strength, lower limb flexibility, plantar sensation and static and dynamic plantar pressures were assessed. The subject was then prepared for EMG and IREDs were attached to the skin overlying anatomical landmarks on their dominant lower limb, before being fitted into the total body harness. Each subject was then familiarised with walking in the harness system. Familiarisation also included practising falling in the harness system so that the subjects felt confident in their gait patterns regardless of the footwear or surface condition. Each subject then received initial instructions pertaining to the walking trials and was allowed to practice walking along the walkway at a self-selected speed to establish a starting position so that their dominant limb would naturally strike the force platform, minimising problems associated with targeting the force platform. Before the walking trials, and at the start of each condition, baseline EMG and body weight ground reaction force data were recorded while each subject stood motionless on the force platform (static trial). This static trial also established the spatial parameters of both the subject and the force platform to allow later transformation into the global coordinate system.

Following the static trial, kinematic, kinetic and EMG data were sampled for 4 s when each subject walked within the calibrated area along the walkway under the nine randomly allocated experimental conditions. The subject was allowed ample rest between trials to prevent fatigue, during which time they subjectively estimated task

difficulty, foot pain intensity, slipper comfort and shoe/surface slipperiness using the VAS. Sufficient trial repetitions were then recorded for each subject to ensure five successful trials per condition, that is, trials in which the dominant foot was placed wholly on the force platform with all IREDs visible by the Position Sensor[762,763]. A greater number of trials were not feasible due to the number of conditions, the age of the subjects and the time required to complete testing. All women with RA were tested at a time that reduced the effect of morning stiffness, a characteristic of the RA disease process. Total time required for testing was approximately 4 hours per subject. Despite the long and involved testing protocol there was only one fatigue-related withdrawal which involved an experimental subject who withdrew from the testing procedures after completing six out of the nine testing conditions (see Section 5.3.1).

5.6 Treatment and Analysis of the Walking Data

It was initially intended to conduct a three-dimensional kinematic analysis of the dominant lower limb of each subject during the walking trials and data were collected accordingly (see Section 5.3.2). However, the Position Sensor used to collect the kinematic data experienced a gradual degradation in the sensor detecting the location of IREDs within the *z* plane such that it eventually failed. The extent of this degradation was not obvious until after testing was completed and the Position Sensor was returned to the manufacturer for repair. After extensive communication with Northern Digital technical staff, it was deemed that two-dimensional coordinates could still be generated from the data for a within-subject analysis. As primary gait motion occurs in the sagittal plane[764,765] and, due to time restraints and difficulty in re-recruiting subjects, a two-dimensional analysis of the dominant lower limb was considered adequate and performed, particularly in view of the comprehensive EMG and kinetic data already obtained for the subjects. Therefore, only the procedures required to perform the two-dimensional data analysis on those variables directly related to slips that occur at initial foot-ground contact are presented in this chapter. It is acknowledged that additional information would be provided by an additional third spatial dimension, but this was not available in the present study due to these technical difficulties which were beyond the control of the Chief Investigator. Confirmation of the accuracy of the two-dimensional data is included in Section 6.6.4.

5.6.1 Estimating Segmental Mass and Inertia Parameters

Lower limb segmental mass and inertial parameters were estimated using the geometric model proposed by Zatsiorsky *et al.*[689]. The lower limb segmental lengths and circumferences recorded in Section 5.2.2(C) were input in the model using the following steps:

(1) The mass of the foot and leg segments were estimated by Equation 5.1:

$$m_i = K_i \cdot L_i \cdot C_i^2 \qquad \textbf{Equation 5.1}$$

where:
- m = mass of the segment (kg)
- K = segment mass coefficient
- L = biomechanical length of the segment (m)
- C = segmental circumference (cm)
- i = segment number

(2) The moments of inertia of the foot and leg segments were calculated relative to sagittal axis using Equation 5.2:

$$I_i = K_i \cdot m_i \cdot L_i^2 \qquad \textbf{Equation 5.2}$$

where:
- I = moment of inertia relative to the sagittal axis ($kg.m^{-2}$)
- K = moment of inertia coefficient relative to the sagittal axis
- m = mass of the segment (kg)
- L = biomechanical length of the segment (m)
- i = segment number

Body segment inertial parameters have been estimated using many methods as well as for many different populations[766-770]. The model proposed by Zatsiorsky *et al.*[689] was chosen for the present study as the equations to determine the segmental parameters were derived from living subjects as opposed to cadavers. Furthermore, use of the model reportedly gives more accurate results than the use of regression equations[689]. Although originally designed for younger subjects, the model can be used for other populations[689]. Estimating lower limb segment inertial parameters was necessary as input in determining kinetic data (see Section 5.6.3).

5.6.2 Kinematic Data Analysis

(A) Analysing the Positional Data

The two-dimensional temporal-spatial trajectory of each IRED was identified relative to the Position Sensor coordinate system[771]. Data were then transformed to meet the Cartesian coordinate system defined by the International Society of Biomechanics with positive *y* as upward and positive *x* as the direction of travel[772]. Positional data were filtered using a zero phase shift Butterworth filter (f_c = 12 Hz) to remove higher frequency noise[773]. The appropriate cut-off frequency was selected based on residual analyses (see Appendix B.9) comparing the difference between the unfiltered and filtered signals for each *x*-coordinate and *y*-coordinate for each IRED for cut-off frequencies ranging from 1 Hz to 30 Hz[774]. The mean cut-off frequency for the *x*-coordinate and *y*-coordinate of all IREDs throughout the trials was 15 Hz (range = 11 to 19 Hz). Although normal gait has been reported to have upper frequency limits between 4 and 6 Hz[774,775], a cut-off frequency of 12 Hz was considered appropriate to reject high frequency noise with the least attenuation of the positional data[716,776]. From these smoothed positional data, a two-dimensional three-segment rigid body model was constructed to characterise the sagittal plane kinematics of the dominant foot, leg and thigh segments of each subject during initial foot-ground contact.

(B) Calculating the Knee and Ankle Rotational Axes

Accurately estimating the rotational axes for each limb complex is important when constructing a rigid body model and subsequently calculating the kinematic and kinetic gait variables[777]. It is recognised that joint axes are not fixed points but rather axes, which move relative to the amount of angular displacement in other joints and/or other planes[778]. Therefore, total amplitude of sagittal plane joint motion may result from a number of joints and/or motions[779-781]. For example, Siegler & Chen[779] reported that the ankle contributes 80% of the motion to achieve maximum plantar flexion whereas the subtalar joint only contributes 20%. Consequently, the knee and ankle joints were determined by

using a least squares fitting of the thigh (IREDs 11 and 14), leg (IREDs 6 and 9) and foot (IREDs 1 and 2; see Table 5.2) segments. These IREDs were chosen as they were aligned with the long bone axis of each segment[774]. The centres of rotation of the knee and ankle were then predicted by the points at which the proximal leg and distal thigh and the proximal foot and distal leg bisected, respectively. Graphic representation of the calculated ankle and knee joint axis centres and, determination of the distances from the calculated joint centres and IREDs 9 and 10 in combination with anthropometric measurements confirmed the correct joint axis locations. The least squares approach allowed the joint rotational axes to be more precisely calculated; where external skin markers placed to represent the knee and ankle axes of rotation (IREDs 5 and 10) may not truly replicate internal skeletal movements. This is particularly true for an older and/or diseased subject sample where five out of eight RA subjects and three out of eight control subjects had undergone knee replacement surgery. The least squares approach has been used previously and the location of the joint Furthermore, the least squares approach for calculating lower limb joint rotational axes also assisted in improving the accuracy of calculated joint moments of force[782] (see Section 5.6.3).

(C) Calculating the Kinematics of Initial Foot-Ground Contact

To determine how each subject altered their gait at initial foot-ground contact, as a result of RA or in response to the different shoe and surface conditions, or for input into further analyses, the following kinematic variables were calculated from the positional data in the sagittal plane (see Figure 5.6):

(1) Horizontal displacement (cm) of the proximal end of the foot was derived by calculating the displacement of IRED 2 from initial foot-ground contact to 15 ms following initial foot-ground contact to determine whether there was any heel slide during loading response (see Section 2.5.2).

(2) Segmental angle (°) of the foot (θ_f) was calculated as the angle of the foot with respect to the horizontal. Foot segmental angle was calculated to

indicate the likelihood of a slip, in that, a decreased foot segmental angle at initial foot-ground contact has been reported as a strategy by which slips may be averted[399].

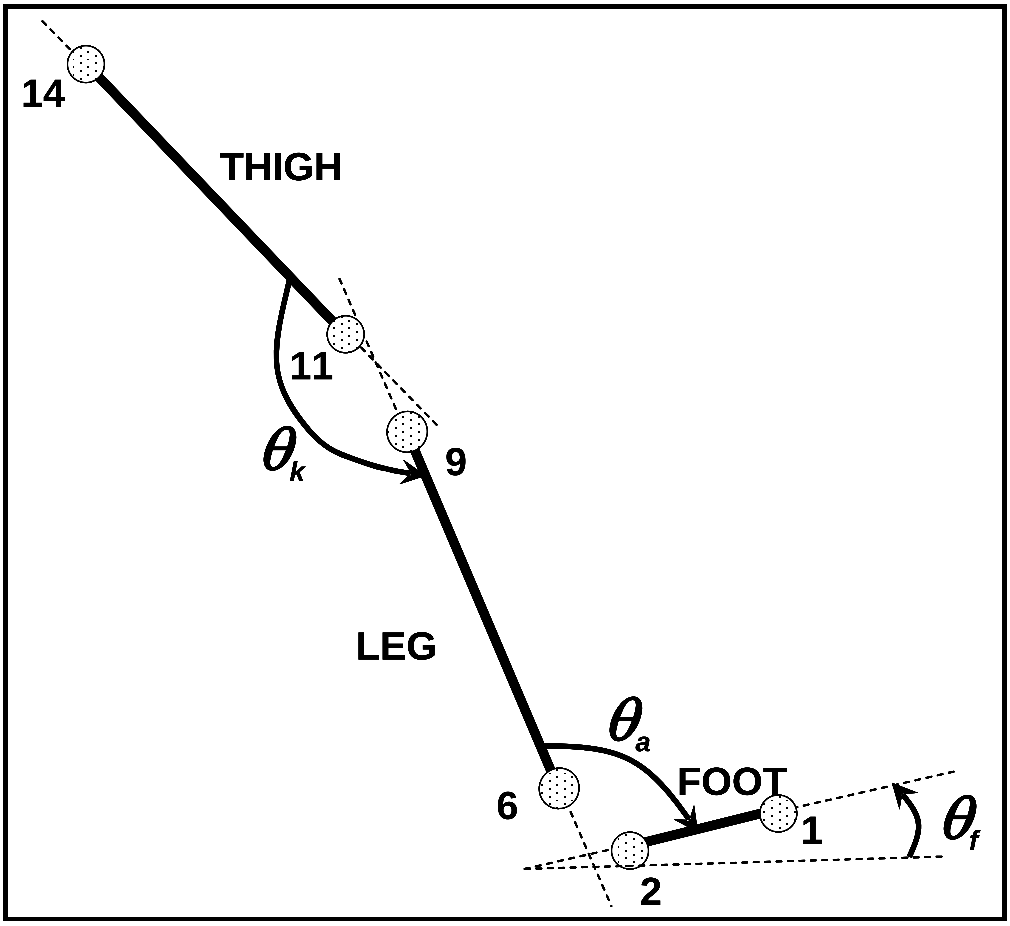

Figure 5.6 IREDs used to define the segments and the convention for calculating segmental and joint angles.

(3) Joint angle (°) of the ankle (θ_a) and knee (θ_k) were calculated to indicate the alignment of the lower limb at initial foot-ground contact.

(4) Horizontal velocity ($m.s^{-1}$) of the proximal end of the foot (IRED 2) was calculated during loading response to characterise the slip risk when walking under changing footwear and surface conditions whereby a higher horizontal velocity was considered to represent an increased slip risk[399].

(5) Angular velocity ($^\circ.s^{-1}$) of the foot segment was calculated during loading response to represent the rate at which the foot rotated towards the ground. This variable was used to characterise how quickly subjects achieved foot flat position (see Figure 2.4) such that transferring a greater percentage of body weight to the stance lower limb may assist in reducing slip risk (see Section 2.5.2).

Lower limb alignment in the static position was taken as the reference for all joint and segmental angles[783]. Velocities and accelerations were calculated by numerically differentiating the position data provided by each IRED using the method of Winter[774]. Angular velocities and accelerations were considered positive in the counter clockwise direction so that the velocity and acceleration data had the correct polarity for subsequent kinetic analyses[497] (see Section 5.6.3). All kinematic analyses were performed using NDI ToolBench Software (Version 1.1, Northern Digital Inc, Waterloo, Canada) and DAP software (Version 2.002, Northern Digital Inc, Waterloo, Canada).

5.6.3 Kinetic Data Analysis

(A) Calculating the Ground Reaction Force Data

The ground reaction force signals from the eight channels of the force platform were visually inspected to ensure all trials were clear representations of each subject's motion and that a single foot had contacted the force platform. The four vertical (y) and two anteroposterior (x) force channels[*] were then zero-offset, summed and scaled to obtain force-time curves in two orthogonal directions[772]. The vertical ground reaction force signals were used to determine stance time (s), such that initial foot-ground contact and terminal stance were deemed to have occurred when the vertical ground reaction forces deviated by 5 N from 0 N. Kinetic variables recorded and calculated from the force-time curves derived for loading response are described in Table 5.4 and depicted in Figure 5.7.

[*] A two-dimensional kinematic analysis was completed (see Section 5.6.2) and consequently only the vertical and anteroposterior ground reaction forces were analysed statistically.

Table 5.4: Kinetic variables recorded and calculated from the vertical and anteroposterior ground reaction force-time curves and the coefficient of friction curve.

Variable Description	Symbol	Units
Time from initial foot-ground contact to terminal stance, defined as 100% stance.	Stance	s
Peak braking vertical ground reaction force.[a]	F_{yB}	$N.kg^{-1}$
Time from initial foot-ground contact until F_{yB}.	IC-F_{yB}	%
Minimum vertical ground reaction force (midstance).	F_{yM}	$N.kg^{-1}$
Time from initial foot-ground contact until F_{yM}.	IC-F_{yM}	%
Peak braking anteroposterior ground reaction force.	F_{xB}	$N.kg^{-1}$
Time from initial foot-ground contact until F_{xB}.	IC-F_{xB}	%
Time from initial foot-ground contact until the anteroposterior ground reaction force polarity change.	IC-F_{xC}	%
Static coefficient of friction at initial foot-ground contact.	μ_S	--
Peak dynamic coefficient of friction after initial foot-ground contact.	μ_D	--
Time from initial foot-ground contact until μ_D.	IC-μ_D	%
Position of the centre of pressure under the foot throughout stance	x, y	--

[a] All magnitude data were normalised to body mass ($N.kg^{-1}$) and temporal data were normalised to stance time (%).

The force magnitudes (N) were normalised relative to each subject's body mass ($N.kg^{-1}$) and the temporal parameters (s) were normalised to stance time (%). Normalising the data minimised comparison errors induced by differences in subject mass and gait speed[784]. These kinetic variables were selected for analysis as they have previously been identified as important descriptors of the forces generated during different shoe-surface interactions and have been used as indicators for slip risk at initial foot-ground contact[399] (see Section 2.5.3).

The coefficient of friction (μ) was calculated by dividing the anteroposterior ground reaction force by the vertical ground reaction force[785] at initial foot-ground contact (μ_S) and at Peak 3 as classified by Perkins & Wilson (μ_D)[406]. Although four peaks were defined by these authors[406], Peak 1 and Peak 2, representative of the initial forwards foot-ground contact and change in force direction resulting from a backward force exerted on the heel, are often

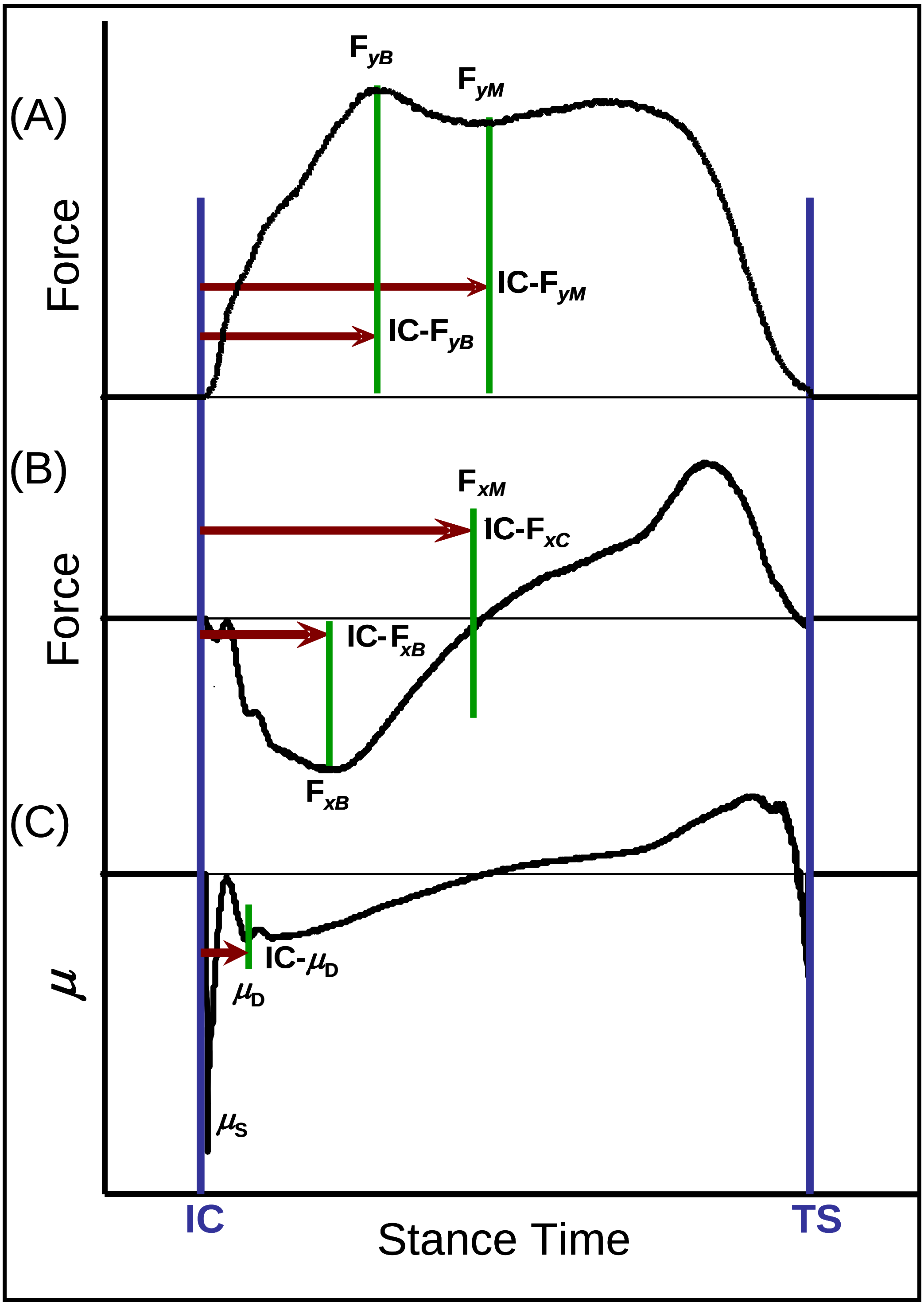

Figure 5.7 Kinetic variables calculated from the vertical (A) and anteroposterior (B) ground reaction force-time curves and the calculated coefficient of friction (C) from initial foot-ground contact (IC) to terminal stance (TS). Please refer to Table 5.4 for variable descriptions.

calculated when the vertical force is less than 50 N, resulting in erroneous results[91,785,786]. Furthermore, many subjects only display three peaks (four are displayed in Figure 5.7(C)), making the calculation of Peak 4 redundant. Consequently, Peak 3 was chosen for to represent μ_D[502] and encompasses the time that the most dangerous slips occur at the time of initial foot-ground[785].

The *x* and *y* coordinates of the centre of pressure were calculated using the algorithms presented by Bobbert & Schamhardt[787]. These algorithms improve the accuracy of the standard Kister[739] equations by correcting for symmetrically distributed errors found when comparing calculated and known points of force application[788]. The centre of pressure data were filtered using a fourth-order zero-phase shift Butterworth low pass filter[169,789]. The centre of pressure data were required to calculate the reaction forces, net moments of force and mechanical powers about the ankle and knee joints.

(B) Calculating the Net Joint Forces and Moments of Force

Computation of the net joint reaction forces and moments of force for the knee and ankle joints was completed using Newton-Euler free body inverse dynamic equilibrium equations[774,790,791]. The assumptions underlying the use of rigid body link segments models are listed in Section 4.5.

Joint reaction forces and net joint moments of force were calculated during loading response for each walking trial, moving progressively from the distal to the proximal end of the segment and from the foot to the leg segment. Segmental mass and moment of inertia were determined using the anthropometric algorithms developed by Zatsiorsky *et al.*[689] (see Section 5.6.1). Foot and leg centre of mass locations were determined as a percentage length using the data presented by Zatsiorsky *et al.*[689]. Linear accelerations of the segmental centres of mass were then derived by double differentiating the displacement data. The ground reaction force data obtained directly from the force platform were input as the forces acting on the distal end of the foot segment. The point of application of these forces was determined using the centre of pressure data obtained from the force platform. The calculated centre

of pressure data were transformed into global units in the laboratory coordinate system using the IREDs placed on the force platform during static assessment[792] and combined with the kinematic data to ensure accuracy of foot placement. Proximal joint reaction forces acting on the foot were derived as equal and opposite to the forces obtained from the force platform. Distal joint reaction forces acting on the leg were derived as equal and opposite to the proximal forces acting on the foot. Positional data obtained from the OPTO***TRAK***® 3020 motion analysis system were used to provide the required distances whereas segmental angular accelerations were calculated by double differentiating the angular displacement data.

Moments of force were calculated during loading response to represent the net effect of muscle activity occurring at the ankle and knee joints in the sagittal plane and to provide inputs to calculate mechanical power[774]. In particular, net joint moments were calculated at initial foot-ground contact as they have been found to be reduced when individuals encounter slippery surfaces or fall as a result of a slip[538]. Resultant joint moments were normalised to body mass (kg) and stance time (%) to enable inter-subject comparisons with positive joint moments of force indicating knee extension and ankle dorsiflexion.

(C) Calculating Mechanical Power

Mechanical powers were calculated for the dominant lower limb of each subject throughout the stance phase using Equation 5.3[345]:

$$P_i = M_i \cdot \omega_i \qquad \textbf{Equation 5.3}$$

where: P_i = mechanical power delivered to or taken from segment *i* (W)
M_i = joint moment vector acting on segment *i* (N.m)
ω_i = joint angular velocity of segment *i* (rad.s^{-1}; see Section 5.6.2)

Mechanical powers were normalised with respect to individual body mass and labeled according to Eng and Winter[793], such that when both the joint moments and corresponding angular velocities had the same polarity, mechanical power was labelled positive. Positive mechanical power is assumed to correlate with energy being generated during a concentric contraction[347]. When the polarities are different, mechanical power is negative, and it is

assumed that energy is being absorbed in an eccentric contraction[497]. The area under each power burst was then integrated to determine the mechanical work performed by the ankle and knee joints during each of the generating and absorbing phases of the stance phase[794]. Mechanical powers were calculated to provide a measure of the efficiency during loading response between the two subject groups under the different footwear-surface conditions. All kinetic analyses were completed using PROG software[795] (see Appendix B.10) and Microsoft Excel® 2002 for Windows (Microsoft Corporation, Washington, USA).

5.6.4 Analysis of Muscle Activity

Raw EMG signals were initially inspected to discard any trials contaminated with noise or motion artefact. Then, following signal offset removal, the raw EMG signals were filtered using a zero-phase shift fourth-order Butterworth filter (high pass f_c = 15 Hz[*]; low pass f_c = 250 Hz[†])[741,774]. The filters were used to attenuate high frequency noise, minimise direct current drift and reduce movement artefact without signal distortion[762,774,796]. The filtered EMG signals were then full wave rectified and filtered using a zero-phase shift fourth-order Butterworth low pass filter[774] to obtain linear envelopes (mV).

A threshold detector screened the linear envelopes representing each muscle studied during the walking trials. Muscle burst onset and offset were deemed to have occurred when 14 consecutive samples of each linear envelope exceeded and passed back under, respectively, a threshold of 10% of the maximum amplitude of the linear envelope representing the muscle activity occurring at initial foot-ground contact[‡]. Computer generated threshold detection of muscle burst onset has been used previously[797,798] and found to be more reliable than simple manual detection of muscle burst onset[799]. However,

[*] A f_c of 15 Hz was deemed most effective in removing any possible movement artefact while minimising signal distortion and was selected after trialing a range of high pass f_c (5 Hz to 25 Hz) and visually inspecting the data after filtering.

[†] A f_c of 250 Hz was selected as the low pass f_c as most surface EMG signals occur between 20 and 200 Hz and 250 Hz represented 25% of the sampling frequency.

[‡] A 10% threshold was selected after trialing a range of thresholds (3 - 15%) and comparing the output against onsets and offsets manually observed from the filtered EMG data and the linear envelopes.

the validity of any computer generated detection method is gained by comparing the results to results gained via visual detection[799,800]. Therefore, visual confirmation using the EMG traces, which were filtered only to remove noise, ensured each value calculated from the linear envelope truly represented the temporal characteristics of each muscle burst during initial foot-ground contact[531]. Figure 5.8 represents an example of a filtered TA and a G trace and their linear envelopes displaying muscle burst onset, offset and the vertical ground reaction force during stance, which was used as a temporal reference.

As most slips that result in falls occur at initial foot-ground contact (see Section 2.2.3), temporal characteristics of the muscle bursts occurring immediately before initial foot-ground contact for each muscle, except G, were derived (ms) to understand how subjects recruited their lower limb muscles to prepare the stance limb for initial foot-ground contact. Small muscle bursts, thought to indicate balance maintenance, occurred at initial foot-ground contact for G. However, only the onset and offset for the major G muscle burst occurring before terminal stance were recorded. Therefore, muscle burst duration (ms) and muscle burst onset and offset relative to initial foot-ground contact (ms) were calculated for each muscle burst of the seven muscles per trial for each subject. Muscle burst onset and offset were then normalised to stance time (%).

The intensity of activity of the RF, VL, BF, S, TA, PL and G muscle bursts were determined by integrating the filtered and rectified data for the duration of each muscle burst. Each integrated EMG (IEMG, V·ms) result was then normalised with respect to the control condition (barefoot on carpet). Muscle intensity has been normalised using a variety of methods[*], although there is currently no consensus with regard to the most appropriate and accurate method that accounts for the portion of the muscle within the viewing area of the electrode, or the muscles' force-velocity and length-tension relationships[805]. The method of normalising muscle intensity used in the present study was considered

[*] Muscle intensity has been expressed as a function of maximal EMG of the same muscle during another exercise[801,802], as a function of the highest EMG of the muscle of interest during the exercise of interest[803], as a function of the force produced about the joint of interest[762], and as a ratio of another active muscle[802,804].

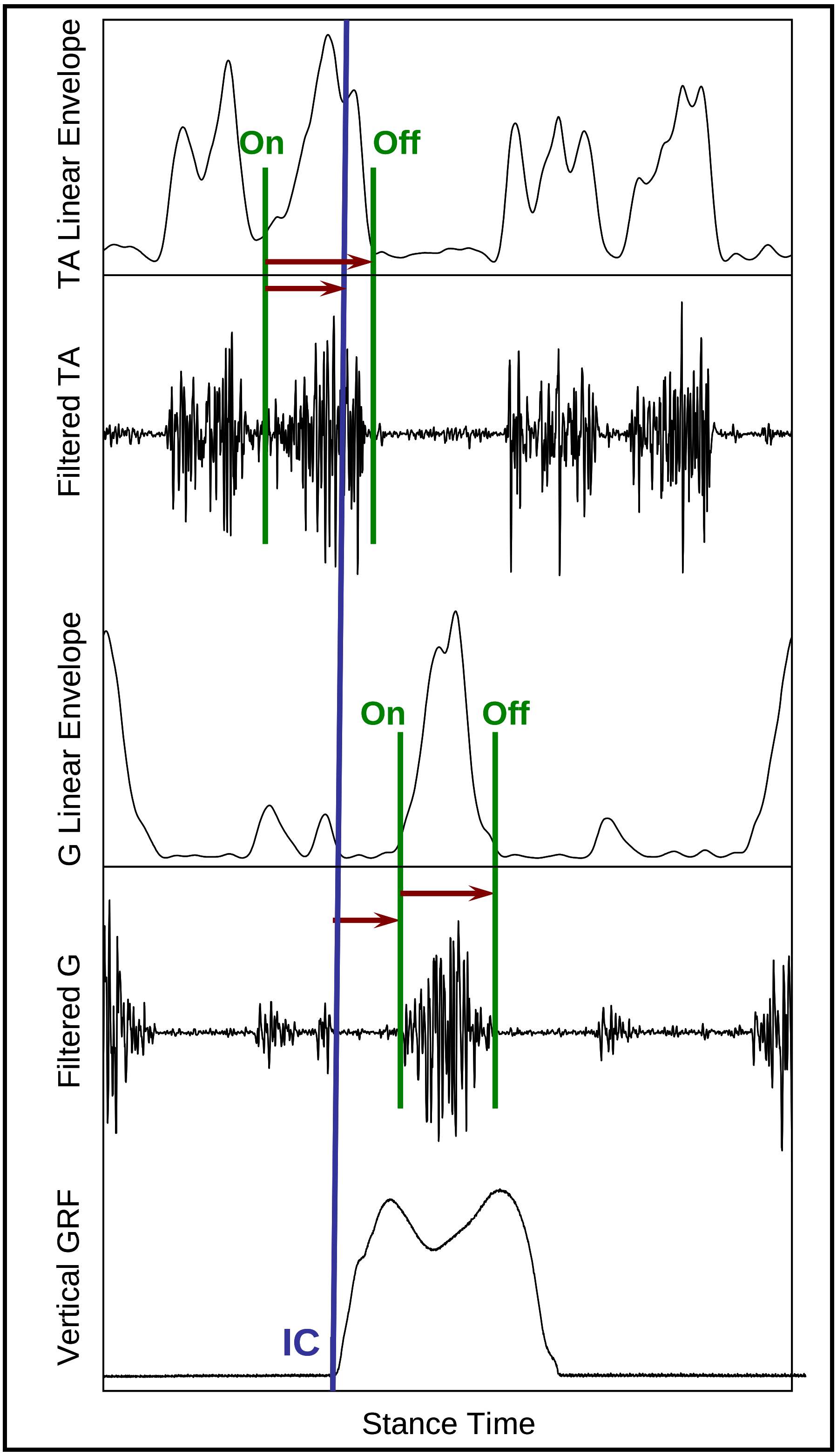

Figure 5.8 Example filtered EMG traces and linear envelopes for tibialis anterior (TA) and gastrocnemius (G), relative to initial foot-ground contact (IC; as indicated by the vertical ground reaction force (GRF) curve), recorded for a control subject's dominant lower limb during a walking trial.

appropriate, as the experimental task was a dynamic movement with changing ankle and knee joint angles[806] and normalising the data against a control condition enabled relative changes in muscle intensity to be compared both within and between subjects. The IEMG data were calculated to provide an indication of the amount of effort required by each muscle to control foot placement at initial foot-ground contact as previous literature has reported increased co-contraction between the muscles when individuals are placed in situations considered dangerous (see Section 2.5.4). PROG software[795] and Digital Signal Processing (DSP) software[807] were used to analyse all EMG waveforms (see Appendix B.10).

5.7 Statistical Analysis

5.7.1 Preliminary Assessment Variables

The two subject groups were initially described by calculating the group means and standard deviations for the variables of height, mass, AIMS2, FFI, lower limb segment proportionality, ankle plantar flexion and dorsiflexion strength and range of motion, knee flexion and extension strength and range of motion, foot reaction time, plantar sensation and both static and dynamic plantar pressure characteristics. Coefficients of variation were also calculated for the static and dynamic plantar pressure characteristics[497]. Paired *t*-tests found no significant within-subject limb-to-limb differences between these dependent variables. Therefore, independent *t*-tests were used to compare the dependent variables collected for the dominant limb of the two subject groups. The purpose of this design was to determine whether the RA subjects differed significantly ($p \leq 0.05$) to the control subjects for any of the preliminary assessment variables.

5.7.2 Walking Trial Variables

Subject group means, standard deviations and coefficients of variation[497] were calculated for the kinematic, kinetic and neuromuscular variables over the

five trials per subject to describe the two subject groups when walking in each condition. The absolute values were analysed using a mixed repeated measures three-way ANOVA design with one between factor (subject group: RA and control) and two within factors (surface: carpet, dry vinyl tile and wet vinyl tile; and shoe type: barefoot, closed back slipper, toe slipper). The main purpose of this design was to determine whether shoe or surface type significantly ($p \leq 0.05$) influenced the gait patterns of the RA or control subjects, particularly those parameters at initial foot-ground contact that may contribute to a slip. Due to the low subject numbers and the high number of conditions compared with the number of subjects in each group, multivariate analyses were not considered appropriate. Where a main effect of subject group, surface type or shoe type, or interactions between the three factors were demonstrated, step down procedures were conducted using one-way repeated measures ANOVAs and pairwise comparisons[808]. A maximum of three pairwise comparisons were performed for each significant main effect or interaction. However, no adjustment was made to the alpha level, according to the Bonferroni method, as a number of findings were anticipated and, although adjustment to the alpha level may reduce the chance of making a Type I error, it would inflate the likelihood of Type II errors[809]. This is particularly the case for small sample sizes[809].

The repeated measures three-way ANOVA is a parametric test that assumes normality, homogeneity of variances and sphericity[808]. Normality of the population was confirmed by assessing the distribution of all dependent variables using a Kolmogorov-Smirnov test (with Lilliefors' correction). The Levene Median test was used to determine that the groups came from populations, which had equal or nearly equal variances in the scores of the dependent variable. Sphericity was tested using Mauchly's test of sphericity to protect from making a Type I error. If this test was violated, a Greenhouse-Geisser epsilon correction was used to reduce the degrees of freedom making it more difficult to find significant F values[808]. Mauchly's tests of sphericity, completed for each dependent variable together with the results of the repeated measures three-way ANOVA and corrected degrees of freedom, are presented in Appendix B.11 and Appendix B.12.

Pearson product moment correlations were also computed to establish whether there were any relationships between the subjective estimation data and the walking trial kinetic or EMG variables for each subject group. All statistical analyses were conducted using SPSS for Windows Version 11.0 statistical package[810].

Chapter 6

Results and Discussion

6.1 Subject Characteristics

Descriptive and statistical information pertaining to the RA and control subjects who participated in Experimental Section B are summarised in Table 6.1. There were no significant differences between the two subject groups for age, height or mass (see Table 6.1) and seven of eight subjects in both groups reported being right foot dominant (see Section 5.2.2). Therefore, consistent with the subject inclusion criteria (see Section 5.1); the two subject groups were appropriately matched for age, height, mass and limb dominance. In addition, all RA subjects had been diagnosed with RA for longer than 5 years (mean ± standard deviation (SD); 19.3 ± 14 years) and had not had foot reconstructive surgery in the 5 years before participating in the study. Therefore, both subject groups adequately fulfilled the subject inclusion criteria for the present study. The AIMS2 and FFI data are discussed in Section 6.2.

Table 6.1: Descriptive characteristics of the RA (n = 8) and control (n = 8) subjects.

Variable	RA Subjects		Control Subjects		*t*-value	*p*-value
	Mean	(SD)	Mean	(SD)		
Age (years)	67.8	(7.3)	65.3	(3.1)	0.90	0.385
Height (cm)	159.2	(7.4)	164.5	(2.8)	-1.89	0.080
Mass (kg)	73.8	(17.7)	74.3	(17.8)	-0.06	0.953
AIMS2[a]	1.95	(2.28)	0.59	(1.20)	1.49	0.157
FFI	28.0	(20.5)	3.1	(4.4)	3.35	0.005*

[a] AIMS2 encompasses mobility, walking and bending, self-care and household sub-scales.
* Mann-Whitney Rank Sum Test indicated statistical significance at $p \leq 0.05$.

6.2 Questionnaire

The questionnaire described in Experimental Section A (see Section 3.2.3) was completed by subjects in the present study to ensure they were representative of the survey sample, as well as to describe each subject group. Although a statistical analysis could not be completed on the survey responses due to the low sample size,

subjectively, subjects in the present study reported results that were similar to the survey sample in all sections of the questionnaire (see Chapter 3). As anticipated, compared to control subjects, the RA subjects reported poorer health, a greater number of diagnosed medical conditions, a greater number of foot problems and a greater incidence of foot pain and/or discomfort, particularly during walking (see Figure 6.1). Five of the eight control subjects preferred not to wear any shoes around the home whereas seven of the eight RA subjects wore foot coverings around the home, with five wearing closed back slippers, two wearing lace-tied shoes and one wearing thick socks. Furthermore, none of the control subjects reported difficulty in donning their shoes compared to six of the eight RA subjects, who reported that they found it hard to bend down and get their shoes on to their feet. Therefore, the household shoe wearing patterns of the RA subjects, together with their reported shoe donning difficulty, appeared to be a direct result of their RA and the associated increased number and severity of foot problems, foot pain and/or discomfort (see Section 2.6.2). Interestingly, most subjects (80%), regardless of RA incidence, considered wet surfaces, particularly wet tile and linoleum surfaces, to be the most slippery when walking around the home.

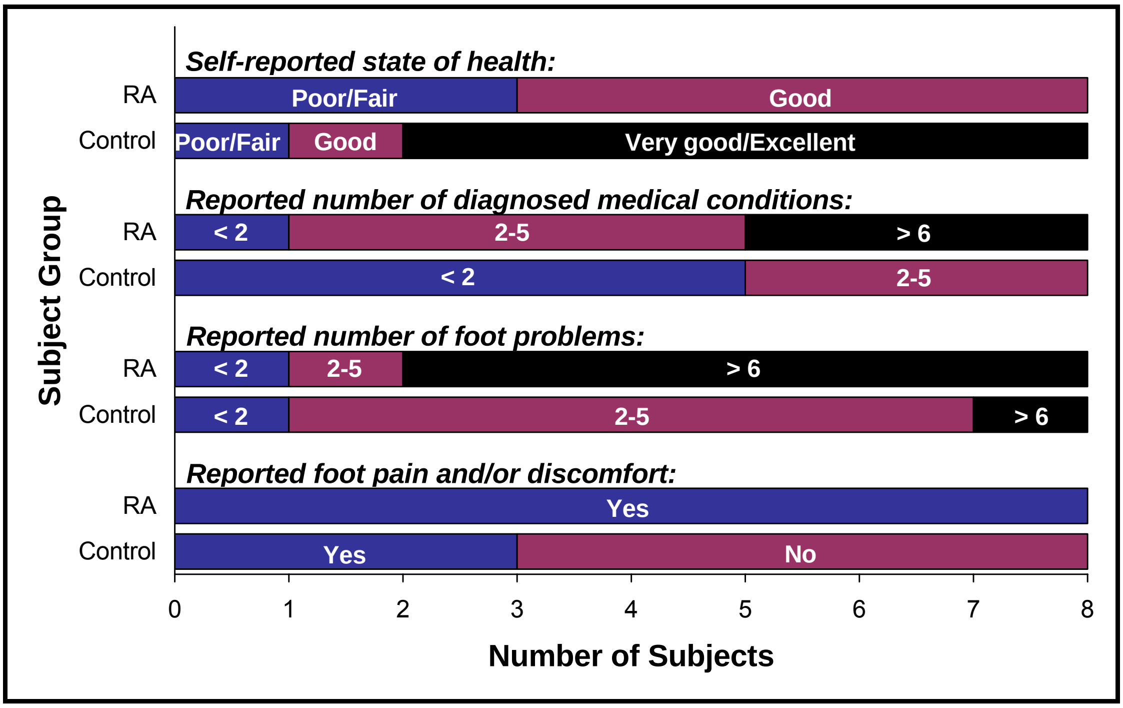

Figure 6.1: Questionnaire item responses pertaining to state of health, reported medical conditions, foot problems and foot pain and/or discomfort for the RA (n = 8) and control (n = 8) subjects.

Despite the subjective differences recorded between subject groups in the questionnaire items, independent *t*-tests revealed no significant differences between the RA and control subjects for their scores on the AIMS2 (see Table 6.1), which assessed the effect of arthritis on mobility and functional independence (see Section 5.2.1). However, the RA subjects reported a significantly higher average FFI score (see Section 5.2.1) compared to the control subjects (Table 6.1). Of the three sub-scales that compose the total FFI score, this difference was mainly due to the increased foot pain reported by the RA subjects (t = 3.19; p = 0.007) as well as greater difficulty when performing activities because of their feet (t = 3.43; p = 0.005), rather than greater activity limitation (t = 1.53; p = 0.148) when compared to the control subjects (see Figure 6.2). Therefore, although experiencing an increased incidence of foot pain and difficulty when performing tasks, perhaps due to their higher incidence of foot problems, the RA subjects in the present study were able to cope with the consequences of their arthritis in such a way that they did not limit their activity levels or impose functional restrictions on their lifestyle. Nevertheless, the older women with RA would appear to have a greater need for specialised footwear to cater for their increased foot problems, foot pain and/or discomfort as well as to minimise consequent altered joint loading during gait that can further contribute to the joint pain and deformities characteristic of RA (see Section 2.6).

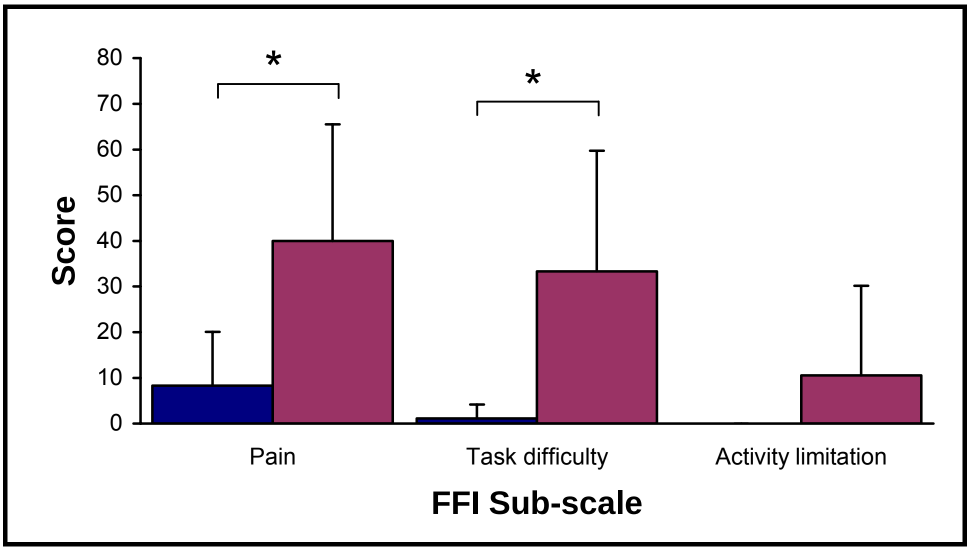

Figure 6.2: The mean (SD) foot function index (FFI) sub-scale scores for the control (n = 8; ■) and RA (n = 8; ■) subjects (* indicates statistical significance at $p \leq 0.05$).

6.3 Physical Assessment of Subjects

6.3.1 Lower Limb Segmental Proportionality

Descriptive statistics pertaining to the lower limb proportionality results for the RA and control subjects are presented in Table 6.2. Apart from both standing and lying foot length, where the RA subjects had significantly shorter feet compared to control subjects, both subject groups had similar lower limb proportionality (see Table 6.2). Therefore, any variations in gait parameters[811,812] (see Section 6.6) were unlikely to be due to between-subject group limb length discrepancies.

Table 6.2: Lower limb segmental proportionality measurements for the RA (n = 8) and control (n = 8) subjects.

Variable	RA Subjects		Control Subjects		*t*-value	*p*-value
	Mean	(SD)	Mean	(SD)		
Circumference (cm)						
Standing foot	22.2	(1.4)	21.8	(1.4)	0.62	0.549
Lying foot	21.1	(1.3)	20.9	(1.4)	0.17	0.870
Calf	36.1	(4.5)	36.4	(3.7)	-0.15	0.881
Upper thigh	57.0	(6.7)	58.2	(7.3)	-0.35	0.730
Length (mm)						
Standing foot	235.7	(10.2)	248.1	(8.6)	-2.55	0.024*
Lying foot	218.6	(10.2)	234.9	(8.0)	-3.56	0.003*
Calf	372.0	(30.6)	389.6	(12.8)	-1.51	0.155
Upper thigh	357.5	(25.4)	371.8	(17.3)	-1.31	0.211
Width (mm)						
Knee	124.8	(13.4)	119.6	(13.4)	0.76	0.458
Ankle	65.9	(5.0)	67.1	(2.6)	-0.63	0.242
Forefoot	86.4	(6.2)	88.4	(5.6)	-0.68	0.507

* Indicates statistical significance at $p \leq 0.05$.

6.3.2 Slipper Size

Although the RA subjects had significantly shorter standing and lying foot lengths compared to control subjects (see Table 6.2), there was no significant difference between the RA and control subjects for slipper size for either the toe or closed back slippers (t = 1.42; p = 0.179), with the median

slipper size worn by both subject groups being size 8. As the RA subjects did not appear to display the characteristic splaying of the forefoot (see Section 2.6.2) that would result in a broader forefoot (see Table 6.2), it was expected that the RA subjects would wear smaller slippers than the control subjects because of their shorter feet. However, contrary to this expectation, the RA subjects required larger slipper sizes to comfortably fit their feet. It is postulated that larger slippers were required because of the foot problems experienced by the RA subjects in the present study. For example, six of the eight RA subjects reported swollen feet compared to three of the eight control subjects. Therefore, based on foot proportionality, older women with RA have different footwear needs to older women without RA. Consequently, the Brannock® foot-measuring device (see Section 5.2.2) may not be the most accurate way to measure the size of an older woman's foot, particularly one with foot problems, to determine correct shoe size. Further research is warranted to determine which measurements of external foot shape and function may better portray the older person's, perhaps deformed, foot in terms of shoe size.

6.3.3 Lower Limb Muscle Strength

Knee and ankle muscle strength results recorded for the two subject groups using the MMT are presented in Table 6.3. Studies investigating the average muscle strength of older women and/or older women with RA report similar findings to the present study, although direct between-study comparison of the results is difficult due to the different methods used to assess lower limb muscle strength[108,109,126,127,813,814]. As expected in the present study, the RA subjects were significantly weaker in knee flexor and ankle dorsiflexor muscle strength compared to control subjects (see Table 6.3).

Deficits in muscle strength, particularly knee extension and ankle dorsiflexion strength have been associated with decreased normal walking speed, gait alterations, functional decline, mobility limitations and the risk of recurrent falls and fractures[15,183,815] (see Section 2.3.1(B)). Differences in gait between the two subject groups in the present study, possibly caused by strength differences, are discussed in later sections. Reduced muscle strength may also

lead to earlier muscle fatigue when performing daily living activities, leaving the older individual with a reduced capacity to avoid, or in fact recover from, a slip when confronted with hazardous or slippery conditions. Declines in muscle strength may be further exacerbated by the RA disease process (see Section 2.6). However, as the RA subjects in the present study did not record activity limitations (see Section 6.2) and were living independently in the community, it is postulated that these RA patients are at a greater risk of falls due to slips, particularly when combined with inadequate footwear. This was confirmed in the present study, as only one control subject had suffered a fall 12 months before testing compared to four RA subjects, who suffered five falls. All of these reported falls resulted from trips (2 falls) and slips (4 falls).

Table 6.3: Lower limb strength measurements for the RA (n = 8) and control (n = 8) subjects.

Variable[a]	RA Subjects		Control Subjects		*t*-value	*p*-value
	Mean	(SD)	Mean	(SD)		
Knee flexion	10.5	(3.3)	14.8	(3.0)	2.68	0.018*
Knee extension	15.9	(5.6)	20.7	(5.3)	1.75	0.102
Ankle dorsi[b]	12.1	(3.7)	17.7	(4.0)	2.92	0.011*
Ankle plantar[c]	17.1	(6.0)	22.3	(3.6)	1.91	0.077

[a] Strength measurements were measured in kilograms of force (see Section 5.2.2).
[b] Ankle dorsiflexion.
[c] Ankle plantar flexion.
* Indicates statistical significance at $p \leq 0.05$.

6.3.4 Lower Limb Joint Range of Motion

Knee and ankle joint range of motion results for the RA and control subjects are presented in Table 6.4. It was anticipated that the RA subjects would have reduced joint range of motion, due to the RA disease process, compared to their control counterparts (see Section 2.6). However, there were no statistically significant differences between the two subject groups for knee extension or ankle joint range of motion (see Table 6.4), with both subject groups displaying joint ranges of motion in the sagittal plane that were considered normal in an older population[339,486,781]. Furthermore, the control

subjects recorded significantly less knee flexion compared to the RA subjects (see Table 6.4). This result was unexpected, particularly as more RA subjects (5 of 8) had undergone knee replacement surgery compared to control subjects (3 of 8; see Section 5.6.2).

Table 6.4: Lower limb joint range of motion data recorded for the RA (n = 8) and control (n = 8) subjects.

Variable[a]	RA Subjects		Control Subjects		*t*-value	*p*-value
	Mean	(SD)	Mean	(SD)		
Knee flexion	59.3	(20.8)	42.3	(7.7)	-2.17	0.048*
Knee extension	175.0	(5.7)	179.5	(5.1)	1.67	0.118
Ankle dorsi[b]	113.1	(5.4)	106.6	(8.0)	-1.91	0.076
Ankle plantar[c]	164.0	(9.1)	169.3	(6.9)	1.30	0.216

[a] Range of motion measurements were measured in degrees (see Section 5.2.2).
[b] Ankle dorsiflexion.
[c] Ankle plantar flexion.
* Indicates statistical significance at $p \leq 0.05$.

6.4 Assessment of Foot Functionality

6.4.1 Foot Reaction Time

Simple and choice reaction time decline with age in many daily living activities, placing people with slower reaction times at a greater risk of falls and fractures[119,198] (see Section 2.2.1). The foot press test used in the present study assessed both simple and choice reaction time as it required the subjects to see, and then respond to, a LED stimulus, by depressing a switch with their feet[183] (see Section 5.2.3). Average foot reaction times recorded for older women using this apparatus have been found to range from 230 ms to 305 ms[149,183,190,191,816]. In the present study, despite the RA subjects recording average foot reaction times (310.8 ± 126.3 ms) which were slower than both the range reported by Lord *et al.*[149,190] as well as the control subjects (262.8 ± 21.9 ms), this between-subject group difference was not statistically significant ($T = 62.00$; $p = 0.574$). The high variability noted in the data for the RA subjects was due to one RA subject recording foot reaction times consistently above 500 ms. When this subject's data was omitted from analysis the average foot reaction time for the

RA subjects was 266.9 ± 25.3 ms although the statistical result remained the same. Therefore, both subject groups were considered to be at an equal risk of sustaining a fall due to impaired foot reaction time.

6.4.2 Plantar Sensation

The number of RA and control subjects who could sense each Semmes-Weinstein monofilament, with ratings of 4.17, 5.07 and 6.10, respectively (see Section 5.2.3), are displayed in Table 6.5. Previous research has advocated that the foot can be considered "at risk" of injury if any portion of the foot is insensitive to the 5.07 monofilament[704]. In the present study, there was no significant difference between the control and RA subjects in terms of plantar sensation at the four sites tested. However, only five of the eight control subjects were able to detect the 5.07 monofilament at the heel compared to seven of the eight RA subjects, suggesting an approximate 98% loss of their sensory ability[817]. As two of these control subjects did not wear shoes around the home, they are at risk of sustaining an injury due to loss of plantar sensitivity (see Section 2.3.1(C)). Simoneau *et al.*[185] attributed this inadequate protective sensation at the heel in older women to skin dryness or the formation of heel calluses, which are common problems seen in the older foot (see Section 2.3.1(E)).

Table 6.5: The number of RA (n = 8) and control (n = 8) subjects who could sense the 4.17, 5.07 and 6.10 Semmes-Weinstein monofilaments.

Site	RA Subjects			Control Subjects			χ^2-value	*p*-value
	4.17	**5.07**	**6.10**	**4.17**	**5.07**	**6.10**		
Hallux	4	7	7	7	8	8	0.347	0.841
1st MTP[a]	4	7	7	6	7	7	0.296	0.863
5th MTP	5	8	8	4	8	8	0.087	0.958
Heel	2	7	7	1	5	6	0.176	0.916

[a] MTP = metatarsophalangeal joint.

During walking the foot is the first part of the body to contact the external environment. Therefore, cutaneous receptors on the foot's plantar surface are required to provide the necessary sensory feedback to plan subsequent steps[270,460,818] as well as to respond to slippery surfaces[419,461] and/or unexpected obstacles or perturbations during walking[270]. Although subjects in the present study were assumed to have adequate plantar sensation, they are part of a population considered at risk of plantar insensitivity. Consequently, shoes are recommended to guard against plantar cutaneous injury. However, the effect of footwear on plantar sensitivity remains controversial. For example, the wearing of shoes may obstruct normal sensory input, leading to gait and posture disturbances and redistributing the pressures experienced at the foot's plantar surface[712,819,820]. Without adequate pressure or pain sensation to avoid them, repetitive, excessive pressures, whether on the dorsal or plantar aspect of the foot, combined with reduced or impaired sensation, can lead to skin breakdown, the results of which may be devastating[273,821,822].

6.4.3 Static and Dynamic Plantar Pressures

(A) Static Plantar Pressures

Descriptive data for static peak pressure, peak force and maximal active area are displayed in Figure 6.3. In the present study, both the RA and control subjects experienced the greatest peak forces at the lateral heel (M01) and the 2nd metatarsal head (M06) with these forces being generated over similar maximal active areas across the various masked areas of the foot. Furthermore, consistent with the findings of Duckworth *et al.*[823] and Minns & Crawford[597], RA subjects displayed the greatest peak plantar pressures across the metatarsal heads (M05, M06 and M07). Conversely, the control subjects displayed results similar to Cavanagh, Rodgers & Iiboshi[824], exhibiting the greatest peak plantar pressures at the heel (M01 and M02; see Figure 6.3). However, despite this apparent disparity between the RA and control subjects, only two significant between-group differences were found in the static plantar pressures obtained in the present study. On average, RA subjects exhibited a significantly (t = 2.12; p =

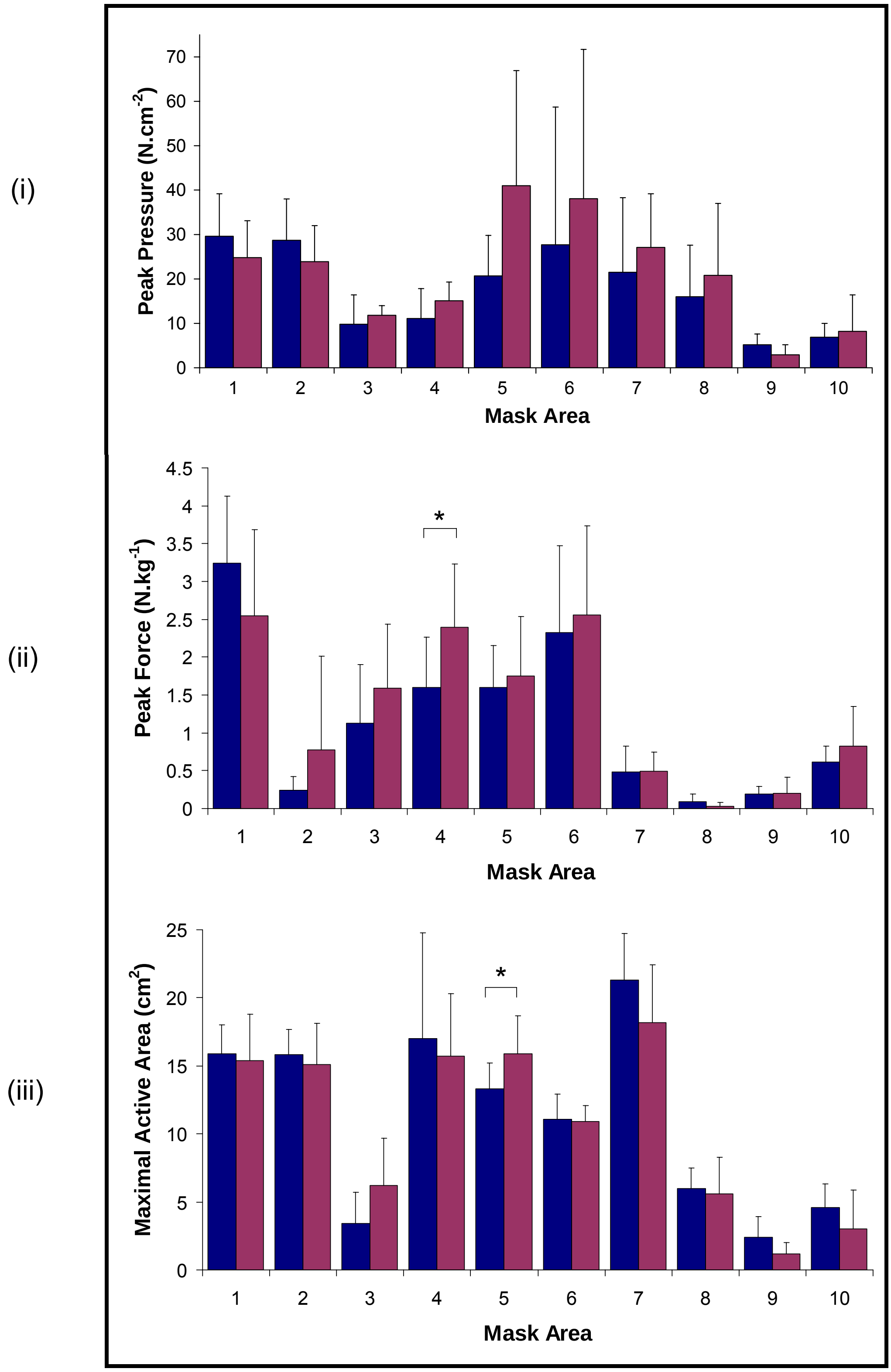

Figure 6.3: Peak pressure (i), peak force (ii) and maximal active area (iii) derived from the static maximum pressure picture for the control (n = 8; ■) and RA (n = 8; ■) subjects over the 10 masked areas of the foot (see Figure 5.1; * indicates statistical significance at $p \leq 0.05$).

0.052) higher peak force at the medial midfoot (M04) and a significantly (t = 2.12; p = 0.052) smaller maximal active area at the 1st metatarsal head (M05) compared to the control subjects (see Figure 6.3 (ii) and (iii)). The peak plantar pressures exhibited by each subject group at the 1st metatarsal head (M05) neared significance (t = 2.09; p = 0.056). This result, however, was achieved with a statistical power of only 39% and therefore requires further investigation.

Previous research reports associations among peak plantar pressure measurements and foot structure[825], foot type[826] and foot pathology[597,827]. As the RA subjects in the present study reported a greater number foot problems and foot pain relative to their non-RA counterparts (see Section 6.2), the differences in their static plantar pressure patterns may have resulted from these foot deformities, characteristic of the RA disease process (see Section 2.6.2). Closer inspection of the data presented in Figure 6.3 revealed large variation, evidenced by the high standard deviations. Furthermore, the calculated coefficients of variation were also high (peak pressure = 36% to 103%; peak force = 29% to 177%; and peak area = 13% to 75%), suggesting that between-group differences may have been masked by this variation. Therefore, further investigation is warranted to determine exactly how the feet of RA patients differ from non-RA individuals in order to correctly design comfortable shoes for this population that encompass the range of deformities characteristic of the RA disease process.

(B) Dynamic Plantar Pressures

Descriptive data for the dynamic peak pressure, peak force and maximal active area generated by the subjects walking across the pressure platform are displayed in Figure 6.4. In addition, the descriptive data for contact time, pressure-time integrals and force-time integrals are displayed in Table 6.6. Independent t-tests revealed that there were no significant differences between the RA and control subjects for contact time during walking or in any of the other temporal parameters measured using the pressure platform in the present study (see Table 6.6). Therefore, both subject groups appeared to have similar times of loading across the regions of their feet. Furthermore, as dynamic

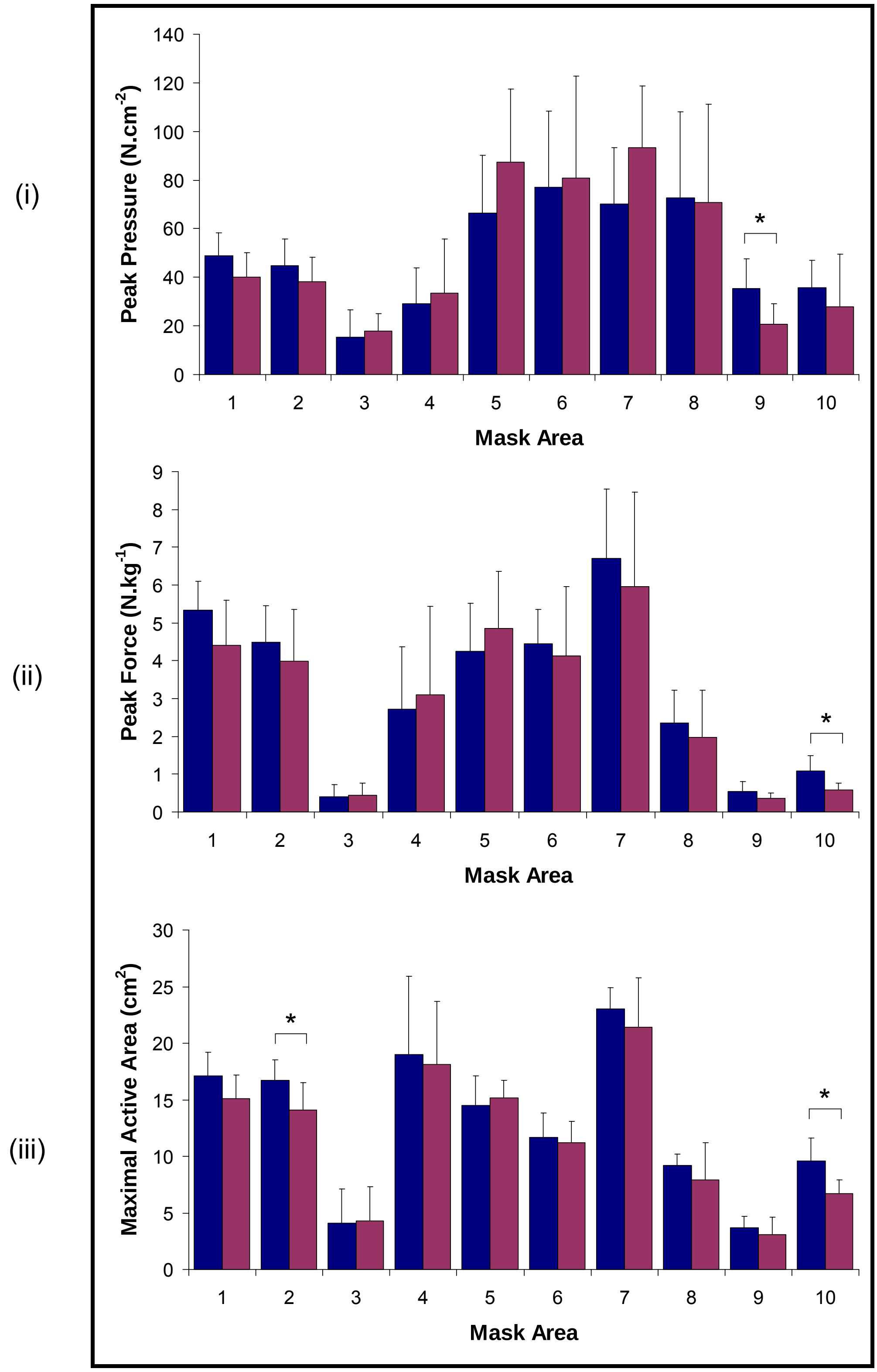

Figure 6.4: Peak pressure (i), peak force (ii) and maximal active area (iii) derived from the dynamic maximum pressure picture for the control (n = 8; ■) and RA (n = 8; ■) subjects over the 10 masked areas of the foot (see Figure 5.1; * indicates statistical significance at $p \leq 0.05$).

Table 6.6: Descriptive and statistical dynamic plantar pressure data for the RA (n = 8) and control (n = 8) subjects from the maximum pressure picture for the 10 masked areas of the foot (see Figure 5.1).

Variable	RA Subjects		Control Subjects		*t*-value	*p*-value
	Mean	(SD)	Mean	(SD)		
Time (ms)						
Total foot	965.0	(482.9)	775.0	(102.5)	1.09	0.295
Lateral heel	61.4	(17.0)	63.0	(5.8)	0.25	0.805
Medial heel	61.5	(16.4)	63.0	(5.8)	-0.25	0.803
Lateral midfoot	44.6	(20.4)	44.3	(20.7)	1.09	0.292
Medial midfoot	72.3	(11.5)	67.1	(6.9)	1.01	0.329
1st MTP[a]	85.1	(8.6)	84.6	(4.0)	0.16	0.875
2nd MTP	86.1	(6.4)	86.1	(3.8)	0.01	0.993
3rd-5th MTP	88.6	(6.2)	88.3	(3.4)	0.14	0.891
Hallux	74.8	(19.0)	69.6	(13.0)	0.65	0.526
2nd phalange	52.1	(19.0)	56.8	(10.8)	-0.61	0.550
3rd-5th phalange	75.1	(18.8)	72.2	(13.1)	0.36	0.725
Pressure-Time Integral ($N.cm^{-2}\cdot s$)						
Total foot	67.7	(38.5)	51.7	(15.3)	1.10	0.292
Lateral heel	14.4	(10.7)	14.2	(3.4)	0.14	0.892
Medial heel	14.2	(10.2)	13.8	(3.1)	0.12	0.912
Lateral midfoot	7.8	(8.7)	4.3	(3.5)	1.07	0.303
Medial midfoot	15.4	(13.3)	9.3	(4.4)	1.23	0.240
1st MTP	41.2	(32.3)	24.0	(11.0)	1.43	0.176
2nd MTP	31.7	(12.6)	28.0	(15.3)	0.53	0.603
3rd-5th MTP	41.2	(24.9)	25.1	(10.2)	1.69	0.113
Hallux	22.0	(14.9)	22.2	(15.2)	-0.04	0.973
2nd phalange	8.8	(10.3)	8.2	(4.1)	0.15	0.883
3rd-5th phalange	10.7	(8.9)	10.2	(4.1)	0.15	0.882
Force-Time Integral ($N.kg^{-1}\cdot s$)						
Total foot	867.7	(400.1)	761.0	(184.2)	0.69	0.504
Lateral heel	121.8	(113.2)	110.6	(29.9)	0.27	0.790
Medial heel	104.0	(79.5)	96.4	(28.6)	0.26	0.803
Lateral midfoot	15.4	(22.4)	8.1	(7.5)	0.87	0.397
Medial midfoot	97.0	(123.6)	66.7	(54.3)	0.63	0.536
1st MTP	149.5	(87.8)	112.1	(50.1)	1.05	0.313
2nd MTP	118.6	(35.4)	114.6	(33.6)	0.23	0.819
3rd-5th MTP	189.2	(75.4)	175.7	(33.8)	0.46	0.652
Hallux	46.1	(41.8)	44.5	(20.7)	0.10	0.926
2nd phalange	11.7	(13.6)	9.1	(5.5)	0.50	0.628
3rd-5th phalange	15.0	(10.5)	23.3	(13.8)	-1.35	0.199

[a] MTP = metatarsophalangeal joint.

plantar pressure distributions are dependent upon walking speed[828-832], any between-group differences in the magnitude of the dynamic plantar pressures were not attributed to differences in gait speed.

With respect to the magnitude data, and similar to the results of Evanski[260] and Edelstein[25], all subjects exhibited less force and pressure on the heel compared to the forefoot during walking, with both the RA and control subjects experiencing their largest peak force and maximal active area values at the 3rd-5th metatarsal heads (M07; see Figure 6.4). RA subjects also experienced peak plantar pressures at the 3rd-5th metatarsal heads (M07) whereas the control subjects experienced peak plantar pressures at the 2nd metatarsal head (M06). These results are in agreement with past studies that have found the maximum plantar pressures to occur under the 3rd metatarsal head[473,833,834]. However, other studies have reported these maximal plantar pressures to occur under the 1st metatarsal head[597,835,836], the 2nd metatarsal head[837,838] or the hallux[473,839,840], being distributed either evenly[833,841-843] or unevenly[835,840] within the metatarsal region when walking. This conflicting evidence in the literature with respect to metatarsal plantar pressure distribution during walking are most likely due to variations in the experimental protocols used (for example, barefoot walking over a pressure plate compared to in-shoe pressure measurements) as well as different analysis procedures[844]. Regardless, the forefoot region appears to be at highest risk for tissue damage, pain and consequent skin ulceration due to raised plantar pressures[704,705,821,822,845-847], particularly if coupled with high pressure-time integrals[833]. In the present study, although not significant, the RA subjects displayed consistently higher peak pressures as well as higher pressure-time and force-time integrals in the forefoot region (M06-M08; see Figure 6.4 and Table 6.6). Therefore, RA patients would appear to have an elevated risk of forefoot plantar tissue damage, pain (see Section 6.6) and consequent skin ulceration[592,827], due to their characteristic foot deformities (see Section 2.5.1). Consequently, specialised footwear advice is warranted for the RA patient in an attempt to reduce these risks[601,848].

Despite the above differences in the magnitude of the dynamic forefoot plantar pressure data, only four statistically significant between-group

differences were identified. That is, the RA subjects recorded significantly reduced peak plantar pressure at the 2nd phalange (M09; t = -2.79; p = 0.014), reduced peak force at the 3rd-5th phalanges (M10; t = -3.26; p = 0.006), a lesser maximal active area at the medial heel (M02; t = -2.50; p = 0.025) and a lesser maximal active area at the 3rd-5th phalanges (M10; t = -3.50; p = 0.004) compared to the control subjects (see Figure 6.4). Although higher in the control subjects, the absolute plantar pressure values were low relative to the forefoot regions and therefore not likely to prove problematic. However, the significantly reduced plantar pressures in the phalanges appear consistent with toe deformities characteristic of older women with RA (see Section 2.6.1). The typical shuffling pattern that has been used to describe the gait of RA patients (see Section 2.6.3) may explain the lesser heel area displayed by the RA subjects in the present study.

Similar to the static plantar pressure data (see Figure 6.3), between-group differences may have been masked by the wide variance in the dynamic plantar pressure data, as evidenced by the high standard deviations and coefficients of variation (peak pressure: 38% to 58%; peak force = 22% to 73%; peak area = 14% to 69%; time = 6% to 41%; pressure-time integral = 46% to 109%; and force-time integral = 21% to 150%), particularly in the metatarsal region of the foot (see Figure 6.4 and Table 6.6). High variability in maximum peak pressure data has been reported previously in the metatarsal region[840], the toe region[842,843] and the midfoot region[849] of the foot. Furthermore, considerable variation in both plantar pressure and force distributions[850,851] have been reported for RA populations, being attributed to either gross pathological changes in the forefoot (see Section 2.3.1(E)), to pain avoidance gait modifications[592,852], or to inconsistent walking speeds, which may result from both of these factors[828,831,832]. The RA subjects also displayed a lesser heel contact area indicating a more shuffling type gait, contributing to a reduced heel impact at initial foot-ground contact[466,853] (see Section 2.6.3). Consequently, it was anticipated that RA subjects would display kinematic, kinetic and neuromuscular differences at initial foot-ground contact compared to the control subjects. These differences are discussed in the ensuing sections within this chapter.

6.5 Summary of Physical and Foot Functionality Assessments

Both the RA and control subjects in the present study were well matched for age, height, mass, lower limb proportionality, lower limb dominance and functional independence. However, compared to the control subjects, the RA subjects reported increased foot problems and foot pain, greater difficulties performing activities because of their feet, required larger slippers relative to foot length, displayed lower limb muscle strength deficits and a trend for higher forefoot plantar pressures, although similar joint range of motion, plantar sensation and foot reaction time. As many of the characteristics displayed by the RA subjects in the present study have been associated with an increased risk of falling (see Section 2.3.1), it is imperative that strategies are implemented to decrease this risk. One such strategy is appropriate household footwear.

Any shoe designed for older women, particularly those with RA, needs to be safe and comfortable, reducing the risk of cutaneous injury, while still allowing adequate sensation of the environment to ensure appropriate gait modification can occur to reduce slips that lead to falls in the home. The results of the present study are consistent with previous recommendations for safe shoes (see Table 2.2 and Table 2.3) in terms of finding that older RA women in the present study purchase shoes that are lightweight with soft, easy to bend soles to reduce early onset muscle fatigue and allow for adequate plantar sensation, and extra depth and extra length shoes to ensure the shoes fit their feet adequately, particularly when toe deformities are present. Shoes with these characteristics should reduce foot pain and thereby these women will be more likely to perform their daily living activities, in turn allowing them to remain living independently for longer. However, whether different footwear constructions or surface characteristics influence the kinematics, kinetics and neuromuscular patterns required for walking in RA women compared to their non-RA counterparts are discussed in the following sections of this chapter.

6.6 Biomechanical Data Characterising Initial Foot-Ground Contact During the Walking Trials

Only the data directly relevant to the following discussion will be included in Section 6.6. Mauchly's tests of sphericity and the three-way repeated measures

ANOVA results for all of the subjective estimations and biomechanical data are included in Appendix B.11 and Appendix B.12, respectively.

6.6.1 Subjective Estimations of Task Difficulty, Foot Pain, Shoe Comfort and Shoe/Surface Slipperiness

As anticipated, the between-group subjective estimation data reported in the present study failed the assumptions of normality and equal variance. However, as there is no nonparametric counterpart, mixed three-way repeated measures ANOVAs, tested and corrected according to the assumption of sphericity, were completed on the data[698]. Consequently, although the results of the statistical analyses should be regarded with caution, the author is confident of true statistical significance when low alpha levels resulted, particularly given the low subject number and high variability within the data.

(A) Between-Group Main Effects

Descriptive and statistical information pertaining to between-group subjective estimations of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness are displayed in Table 6.7. There were no significant differences between the RA and control subjects for estimations of shoe comfort or shoe/surface slipperiness when the data were pooled across footwear and surface condition (see Table 6.7). However, as anticipated, there was a significant main effect of subject group on task difficulty and the foot pain recorded during the walking trials. That is, the RA subjects estimated the walking trials to be significantly more difficult and experienced significantly more pain during the walking trials than the control subjects (see Table 6.7).

Pain resulting from RA has been found to be a reliable predictor of medication use, depression and anxiety[676,854,855] and thereby, an increased risk of falls (see Section 2.3.1). However, musculoskeletal pain reported by older people in general has been suggested to be a major contributor to disability[856-858], mobility decrement[680,858-862], gait difficulty[680,858-862] and increased falls risk[253,856].

For example, the results of the Women's Health and Aging Study* revealed that increased pain was associated with difficulty in carrying out daily living activities[857,859], difficulty in walking[857] and a greater number of both single and recurrent falls in a 12 month period[863]. Furthermore, those women who reported widespread pain and displayed poorer physical performance, including slower gait, slower chair stand time and lower knee extension strength, were also found to have a higher prevalence of arthritic conditions affecting more than one musculoskeletal region than older women who reported less pain[863]. However, upon further analysis as to the site of the pain, only foot pain, as opposed to pain in the hands, wrists or back, was significantly associated with increased walking difficulty, slower gait, slower chair stand time and lower knee extension strength[864], as well as an increased risk of falls[863]. These results were confirmed by Menz & Lord[253] who demonstrated a link between foot pain and falls as well as foot pain and difficulty in carrying out daily living activities in community-dwelling older Australian individuals.

Table 6.7: Descriptive and statistical information for the subjective estimates of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness for the RA (n = 8) and control (n = 8) subjects.

Variable	RA Subjects		Control Subjects		$F_{(1,7)}$	*p*-value
	Mean	(SEM)[b]	Mean	(SEM)		
Task difficulty	17.8	(4.2)	3.0	(3.9)	6.65	0.023*
Foot pain	20.9	(5.7)	0.0	(5.3)	7.27	0.018*
Shoe comfort	82.1	(5.2)	90.8	(4.9)	1.49	0.243
Slipperiness[a]	15.5	(4.9)	5.5	(4.6)	2.19	0.163

[a] Slipperiness indicates shoe/surface slipperiness.
[b] SEM indicates the standard error of the mean.
* Indicates statistical significance at $p \leq 0.05$.

In the present study, and consistent with the subjective estimation data, compared to the control subjects, RA subjects reported significantly more foot pain and significantly greater difficulty in performing daily tasks because of their feet as evidenced by a tendency for high FFI scores (see Section 6.2). In

* The Women's Health and Aging Study[863] conducted home interviews and assessments of 1,002 older community-dwelling American women aged 65 years and above over 3 years.

addition, seven of the eight RA subjects reported difficulty in performing the walking trials compared to only four of the eight control subjects and, although six of the eight RA subjects reported foot pain when completing the walking trials, not one of the control subjects reported any foot pain (see Table 6.7). Foot pain and/or discomfort can have a considerable effect on gait and shoe wearing difficulties, together with an individual's mobility, independence and importantly, falls risk (see Section 2.3.1(E)). Therefore, foot pain may lead to expectations of task difficulty when walking and, if walking is perceived as more difficult, older people may curb their activity as a technique to reduce this pain, leading to a destructive cycle of activity decline with consequent strength and mobility loss[594,863], thereby placing them at a greater risk of sustaining a fall. Consequently, older women with RA need to have access to household shoes, which can reduce foot pain and ensure trouble-free gait in order to facilitate their mobility and maintain their ability to live independently.

(B) Within-Footwear Main Effects

Descriptive data pertaining to the subjective estimations of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness for the different footwear types are displayed in Table 6.8. The *F*-ratios and alpha levels for the within-subject sources of variation for subjective estimation data are documented in Table 6.9. When the data were pooled across subject group and surface condition, there was a significant main effect of footwear on shoe comfort and shoe/surface slipperiness (see Table 6.9). Furthermore, strong trends were evident with respect to the effects of footwear on task difficulty[*] and foot pain. That is, subjects recorded the greatest task difficulty and most foot pain when walking wearing the toe slippers with the least task difficulty recorded when subjects walked wearing the closed back slippers (see Table 6.8). However, the powers of these statistical tests were low at 51% and 55%, respectively, and, as the data were pooled across subject group, the pain

[*] A significant *p*-value was calculated for the effects of footwear on task difficulty. However, this was removed with application of the Greenhouse-Geisser correction factor.

recorded by the RA subjects may have contaminated the data. Therefore, further research is warranted into the effects of footwear on task difficulty and foot pain.

Table 6.8: Descriptive information, pooled across subject group and surface condition, for the subjective estimates of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness when subjects (n = 16) walked barefoot, in closed back slippers or in toe slippers.

Variable	Barefoot		Closed slipper		Toe slipper	
	Mean	(SEM)	Mean	(SEM)	Mean	(SEM)
Task difficulty	10.8	(2.9)	2.4	(0.7)	17.9	(6.8)
Foot pain	8.4	(3.4)	10.6	(4.3)	12.3	(4.2)
Shoe comfort	88.3	(2.9)	93.6	(2.4)	77.4	(7.0)
Slipperiness[a]	17.2	(3.4)	3.5	(1.4)	10.7	(6.4)

[a] Slipperiness indicates shoe/surface slipperiness.

Table 6.9: *F*-ratios and alpha levels for the subjective estimation data.

Variable	Shoe	Shoe x Group	Surface	Surface x Group	Shoe x Surface	Shoe x Surface x Group
	$F_{2,26}$ *p*-value	$F_{2,26}$ *p*-value	$F_{2,26}$ *p*-value	$F_{2,26}$ *p*-value	$F_{4,52}$ *p*-value	$F_{4,52}$ *p*-value
Task difficulty	3.94† 0.057	1.86† 0.193	9.26† 0.007*	1.26† 0.287	6.72† 0.004*	2.32† 0.118
Foot pain	3.11 0.061	3.11 0.061	0.87† 0.403	0.87† 0.403	2.03† 0.135	2.03† 0.135
Shoe comfort	5.43† 0.025*	0.47† 0.556	3.35† 0.076	0.11† 0.813	3.91† 0.025*	1.22† 0.315
Slipperiness[a]	4.49† 0.040*	0.90† 0.384	19.01† <0.001*	1.12† 0.315	21.83† <0.001*	0.57† 0.536

[a] Slipperiness indicates shoe/surface slipperiness.
* Indicates statistical significance at $p \leq 0.05$.
† Use of Greenhouse-Geisser correction factor (see Table B.11.1, Appendix B.11).

Pairwise comparisons completed on the shoe comfort and shoe/surface slipperiness data indicated that subjects in the present study perceived the toe slippers to be the most uncomfortable footwear condition compared to walking barefoot ($t = 2.43$; $p = 0.019$) or in closed back slippers ($t = 3.54$; $p = 0.001$; see Table 6.8). Furthermore, when subjects walked wearing closed back slippers

they found it significantly less slippery than when walking barefoot (t = 3.79; p < 0.001) or wearing toe slippers (t = -2.00; p = 0.051; see Table 6.8). These results appear to support the trends with respect to task difficulty and foot pain. That is, if the shoe was considered more comfortable, one would expect that foot pain would decrease and, in turn, if a shoe was considered less slippery when walking, the task may be considered easier to complete (see Table 6.8).

Shoe comfort has been defined as an individual subjective feeling[388]. Therefore, what is comfortable to one person may be uncomfortable to another based on their personal preferences and the footwear to which they have become accustomed. Although it is difficult to measure[865], and just as difficult to interpret, the perception of shoe comfort is one of the most important parameters for shoe manufacturers and shoe markets[460]. For comfort, the shoe should not produce local irritations to the foot and should generate minimal force and pressure between the plantar surface of the foot and the shoe insole[460,865]. Soames & Evans[495] stated that any shoe deemed comfortable was safe for the older individual (see Section 2.4.3) and past research has suggested relationships between perceived comfort and certain measurable parameters such as ground reaction forces, plantar pressure distributions and energy cost[270,388,460,712].

In the present study, toe slippers were considered to be the most uncomfortable footwear condition, possibly because subjects had trouble keeping the shoe on their foot. In fact, the feet of some subjects overhung the lateral aspects of the toe slippers, or their feet continually slipped out of the slippers altogether (see Figure 6.5), at times contacting the supporting surface when they walked. In addition, older people with toe deformities or restricted toe motion may have trouble grasping with their toes to keep the toe slipper on their foot. Alternatively, as five of the eight control subjects preferred to walk barefoot around the home and seven of the eight RA subjects wore closed back slippers or lace-tied shoes around the home (see Section 6.2), subjects may have simply been unaccustomed to toe slippers as they are not typically chosen for purchase due to their lack of comfort.

The reaction of people to shoe-surface interfaces perceived as slippery may partially determine the outcome of a movement, that is, whether the

individual slips, recovers from a slip, or sustains a fall resulting from a slip. Few studies were located that investigated how subjects perceived the slipperiness of different footwear types on different surface types and no studies were located investigating the slipperiness perceptions of older people for different household footwear on typical household surfaces. Of the studies that have been completed[85,90,400,411,416,419,422,426], strong associations have been shown between subjective ratings of slipperiness and friction measurements on dry and contaminated (wet, clay and oil) surfaces for walking (see Section 2.3.2(C)). However, many of these studies were completed when subjects either walked barefoot or used their hands to evaluate the degree of slipperiness. As the ability to adjust gait patterns during walking is influenced by subjective perceptions[461], age-related decline in function (see Section 2.3.1) may mean older individuals may not be able to accurately evaluate potential slip hazards. If a potential slip hazard is not perceived, one may not modify their movement patterns, thereby increasing their slip and injury risk (see Section 2.3.2(B) and Section 2.3.2(C)).

Figure 6.5: Examples of the feet of two subjects, captured from a rear video camera, slipping out of the toe slippers during quiet standing between the walking trials.

In the present study, both the closed back and toe slippers had the same outsole design, suggesting that both slipper types would interact in a similar fashion with any supporting surface. However, subjective comments pertaining to the toe slippers indicated that some subjects found them difficult to keep on their feet when walking, reporting them to be "slippery" because their feet kept slipping/moving within the slippers (see Figure 6.5). It is this definition of

slipperiness, that is, the foot slipping inside the shoe as compared to the shoe slipping on the surface that may be the reason as to why the toe slippers were considered to be more slippery than the closed back slippers. Unfortunately, no differentiation was made between these two definitions as subjects were simply asked “How slippery did you find the task?” (see Section 5.3.5). Therefore, further research is required to determine what effect: 1) the foot slipping inside the shoe, 2) the foot slipping out of the shoe and perhaps contacting the supporting surface and 3) the shoe slipping on the supporting surface, have on the perception of shoe/surface slipperiness as well as the variables of gait at initial foot-ground contact, particularly with respect to falls risk and footwear recommendations. For example, if foot slippage inside the shoe alters the gait of older women, then shoe types such as toe slippers and other household shoes that are sloppy and do not fit the foot well will place older women at a greater risk of falls and, in turn, should not be recommended for this population. Therefore, it is anticipated that subjects will display notable differences in the biomechanical indices that characterise initial foot-ground contact when walking barefoot and in toe slippers due to the main effects of footwear on shoe comfort and shoe/surface slipperiness.

(C) Within-Surface Main Effects

Descriptive data pertaining to the subjective estimations of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness when subjects walked across different surfaces types are displayed in Table 6.10. When the data were pooled across subject group and footwear condition, a significant main effect of surface type on task difficulty and shoe/surface slipperiness was evident (see Table 6.9). Pairwise comparisons revealed that subjects found the walking task both significantly more difficult and significantly more slippery when they walked across the wet vinyl tile surface compared to walking across dry vinyl tile (task difficulty: $t = -3.59$; $p = 0.001$; slipperiness: $t = -5.12$; $p < 0.001$) or carpet (task difficulty: $t = -3.19$; $p = 0.003$; slipperiness: $t = -5.26$; $p < 0.001$; see Table 6.10).

Table 6.10: Descriptive information, pooled across subject group and footwear condition, for the subjective estimates of task difficulty, foot pain, shoe comfort and shoe/surface slipperiness when subjects (n = 16) walked on carpet, dry vinyl tile and wet vinyl tile.

Variable	Carpet		Dry vinyl tile		Wet vinyl tile	
	Mean	(SEM)	Mean	(SEM)	Mean	(SEM)
Task difficulty	7.5	(2.6)	6.3	(2.5)	17.3	(4.4)
Foot pain	9.6	(3.2)	9.9	(4.1)	11.9	(4.6)
Shoe comfort	88.2	(3.4)	90.6	(3.7)	80.5	(5.5)
Slipperiness[a]	2.6	(2.1)	4.0	(3.4)	24.7	(5.9)

[a] Slipperiness indicates shoe/surface slipperiness.

As discussed in Section 2.3.2(C) and 6.6.1(B), the perception of shoe/surface slipperiness is important in terms of allowing the individual to modify their movement pattern if a potential slip hazard is perceived so that they remain upright and stable when encountering the slip hazard. Water adds to the risk of slips by providing a thin layer of fluid between the shoe and the surface[415] and lowering the coefficient of friction (see Section 2.3.2(C)). Interestingly, Gard & Lundberg[416] reported that more pedestrians chose to stay inside during wet weather compared to dry weather, possibly because of their perceptions of slippery surfaces and the increased likelihood of falls. Therefore, adapting one's gait is necessary when walking on a wet surface to avoid a slip and/or fall.

Differences have been previously reported in the number of reported falls, falls-related injuries and the walking patterns of older people when they walk across carpet and vinyl tile surfaces (see Section 2.3.2(B)). However, subjects in the present study did not appear to categorise the carpet and dry vinyl tile surfaces differently with respect to the subjective estimations of task difficulty and shoe/surface slipperiness. Therefore, as subjects in the present study perceived the wet vinyl tile surface to be slippery and difficult to walk across, it is postulated that unless a wet surface is recognised, older women will not change their walking patterns, and, in turn, will increase their falls risk. Consequently, it is anticipated that during the walking trials subjects will only display gait changes when walking on the wet vinyl tile surface with few gait

changes evident when walking on the dry vinyl tile surface (see Section 6.6.2(C), Section 6.6.3(C), Section 6.6.4(C) and Section 6.6.5(C)).

(D) Interaction Effects

There were no subject group x footwear, subject group x surface or subject group x footwear x surface interactions for the subjective estimation data in the present study. However, both the footwear and surface main effects were moderated by each other such that there were significant footwear x surface interactions for task difficulty, shoe comfort and shoe/surface slipperiness (see Table 6.9). Step down procedures pertaining to the subjective estimation data for task difficulty indicated that subjects considered walking in closed back slippers on the wet vinyl tile surface to be significantly easier than walking on the wet vinyl tile barefoot (t = 3.13; p = 0.007) or in toe slippers (t = -2.23; p = 0.043; see Figure 6.6). Therefore, although subjects generally considered walking in toe slippers to be the most difficult on all surface types, when they walked barefoot on the wet vinyl tile surface, subjects considered the task significantly more difficult than when walking shod or when walking barefoot on the other surface types.

Only one study was located which assessed ratings of task difficulty while subjects performed tasks wearing slippers[758]. Ten healthy, young women (mean age, 19.7 years) rated the task difficulty of climbing stairs with various tread/rise combinations while wearing different shoe types, one of which was slippers. Although no subject reported difficulty in stair climbing when wearing slippers compared to other low-heeled shoe types, the study did not provide details as to the type or design of the slipper and, although the stair dimensions were altered, the stair surface remained consistent. Therefore, it is postulated that surface type, surface condition and the presence of surface lubricant, may have a greater effect on task difficulty than merely shoe type in isolation.

Similar findings to those for task difficulty were found for the ratings of shoe/surface slipperiness. Step down procedures pertaining to the footwear x surface interaction for shoe/surface slipperiness revealed the subjects found

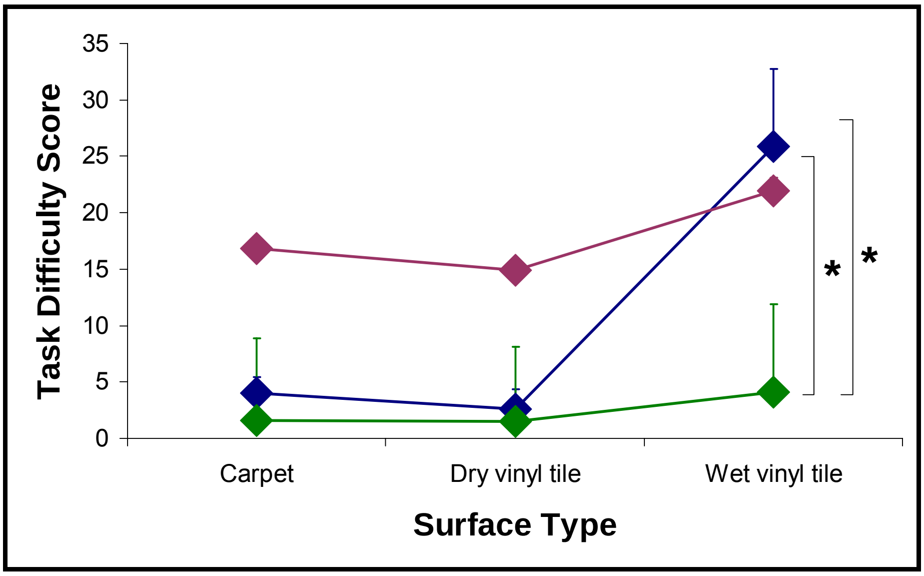

Figure 6.6: Footwear x surface interaction for the mean (SEM) subjective estimation of task difficulty when subjects (n = 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

walking barefoot on the wet vinyl tile surface to be significantly more slippery than walking shod on the same surface (closed back slippers: $t = 5.70$; $p < 0.001$; toe slippers: $t = 4.91$; $p < 0.001$; see Figure 6.7). Interestingly, subjects in the present study only considered barefoot walking on the wet vinyl tile surface to be significantly more slippery than walking shod on the same surface. That is, they did not consider walking barefoot on either the carpet or dry vinyl tile surfaces to be slippery. This may confirm the notion that increased shoe/surface slipperiness translates to increased task difficulty. However, if a an older person perceives a task, such as walking on a wet, slippery surface, to be difficult, they may direct a greater amount of attention to the walking process, assisting them to increase walking stability and reduce their falls risk (see Section 2.3.1(F)).

Step down procedures pertaining to the footwear x surface interaction for shoe comfort indicated that when subjects walked across the carpet surface, they reported the toe slippers to be the least comfortable footwear condition compared to walking barefoot ($t = 2.59$; $p = 0.021$) or in closed back slippers ($t = 2.57$; $p = 0.022$). Toe slippers were also considered to be significantly less

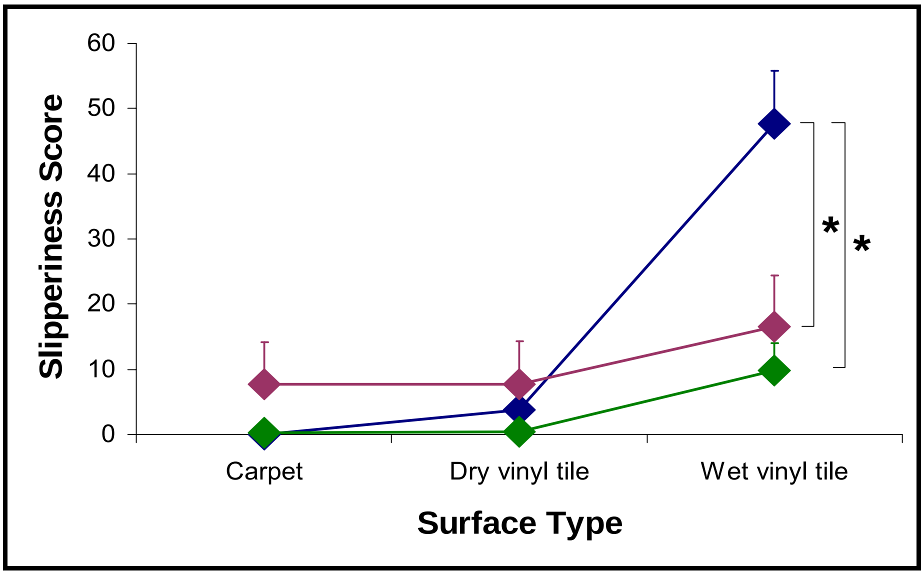

Figure 6.7: Footwear x surface interaction for the mean (SEM) subjective estimation of shoe/surface slipperiness when subjects (n = 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

comfortable than walking barefoot across dry vinyl tile ($t = 2.31$; $p = 0.036$; see Figure 6.8). However, unlike previous findings, when subjects walked across wet vinyl tile, they perceived walking barefoot to be significantly more uncomfortable than walking across wet vinyl tile in closed back slippers ($t = -2.31$; $p = 0.029$; see Figure 6.8). Therefore, similar to the task difficulty data (see Figure 6.6), although toe slippers were generally considered to be more uncomfortable than the other footwear types when walking across the carpet and dry vinyl tile surfaces (see Figure 6.8), this was not the case on the wet vinyl tile surface where subjects considered the barefoot condition to be the most uncomfortable footwear condition. Consequently, it would appear that for all three footwear x surface interactions, subjects had an acute negative reaction to walking barefoot on wet vinyl tile compared to the other conditions.

The extreme differences in subjective estimations recorded in the present study when subjects walked barefoot on the wet vinyl tile surface compared to walking barefoot on the other surface conditions warrants further consideration. When barefoot, the sensory receptors on the plantar surface of the foot are

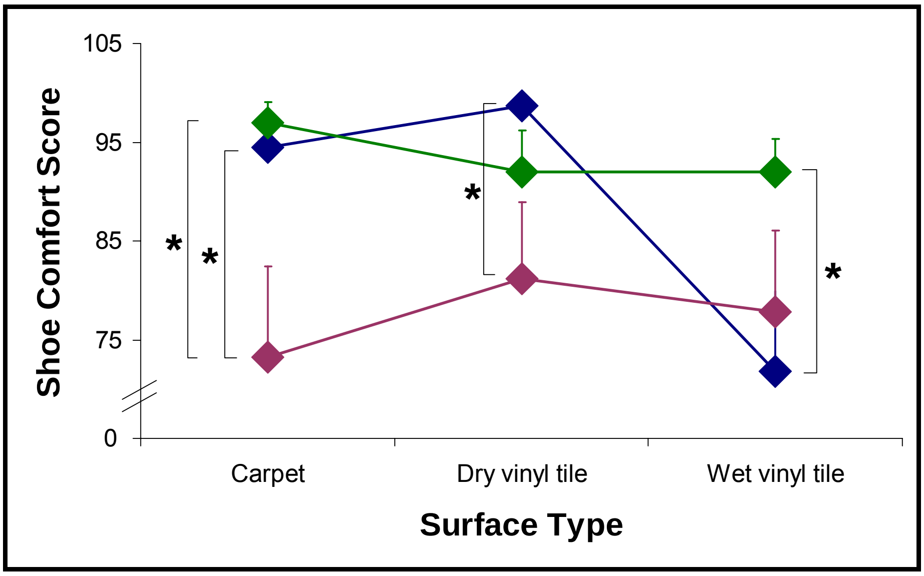

Figure 6.8: Footwear x surface interaction for the mean (SEM) subjective estimation of shoe comfort when subjects (n = 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

exposed to the supporting surface and can provide feedback to the individual about the condition of the supporting surface, allowing them to alter their movement patterns accordingly. Consequently, as barefoot walking appears to provide the greatest sensory feedback to the individual, one may recommend older people to walk barefoot around the home so that they can adjust their gait to changes in surface characteristics. However, walking barefoot increases the risk of sustaining cutaneous injury which, if combined with poor sensation and a reduced healing capacity (see Section 2.3.1(E)), could escalate into a major medical issue for the older individual.

When shod, little is known about the feedback provided to the individual from within the shoe or the effects of this feedback on movement patterns[270,866-868]. However, it appears that instead of relying on sensory input, older people become reliant upon past experiences of walking on selected surface types and memories of frictional values[400] (see Section 2.3.2(C)), thereby using their subjective perceptions of shoe/surface slipperiness to modify their gait. Unfortunately, if past experiences, memories or perceptions are incorrect for the

present footwear-surface interaction or inadequate sensory information is provided, the older person may again heighten their slip risk[91].

Based on the results of the subjective estimation data from the present study, it is anticipated that subjects will alter their gait pattern when walking in toe slippers on any surface and when walking barefoot, particularly on the wet vinyl tile surface. These gait changes will be discussed in the following sections. However, based on these data, further research is warranted to determine the true effect of subjective perceptions and sensory feedback when older people walk barefoot or shod, particularly in shoes with differing outsole thickness, hardness and roughness, on the variables characterising their gait and slip risk.

6.6.2 Lower Limb Muscle Activation Patterns in Preparation for Initial Foot-Ground Contact

(A) Between-Group Main Effects

An example of the lower limb phasic muscle burst activity used by the RA and control subjects in preparation for initial foot-ground contact is displayed in Figure 6.9. This sequence of muscle activation shows the synergistic actions of the muscles required to control the lower limb[869] in preparation for initial foot-ground contact and is similar to those reported in previous studies for both young and older subjects[352,497,499,500,512,531,870-873] (see Section 2.5.4). Therefore, subjects in the present study displayed normal phasic muscle activity during the terminal swing, initial foot-ground contact and loading response phases of gait (see Section 2.5.1 and Figure 2.4).

One notable feature in Figure 6.9 is the high variability shown within the mean phasic muscle activity as evidenced by the high standard deviations. This high variability is particularly evident at muscle burst offset for the RA subjects, with coefficients of variation ranging from 12% to 262% compared to the control subjects with coefficients of variation ranging from 8% to 100% for the same variables. High variability in muscle activation data during gait is often reported in the literature for young[347,874,875], healthy older[332,347,874,876] and

diseased populations[497,668,870,876] and various analysis techniques have attempted to normalise EMG data to reduce this variability[763,796,877,878]. However, it has been suggested that EMG variability is a normal phenomena within gait and is due to inconsistent motor unit recruitment patterns, possibly reflecting impaired postural control[876].

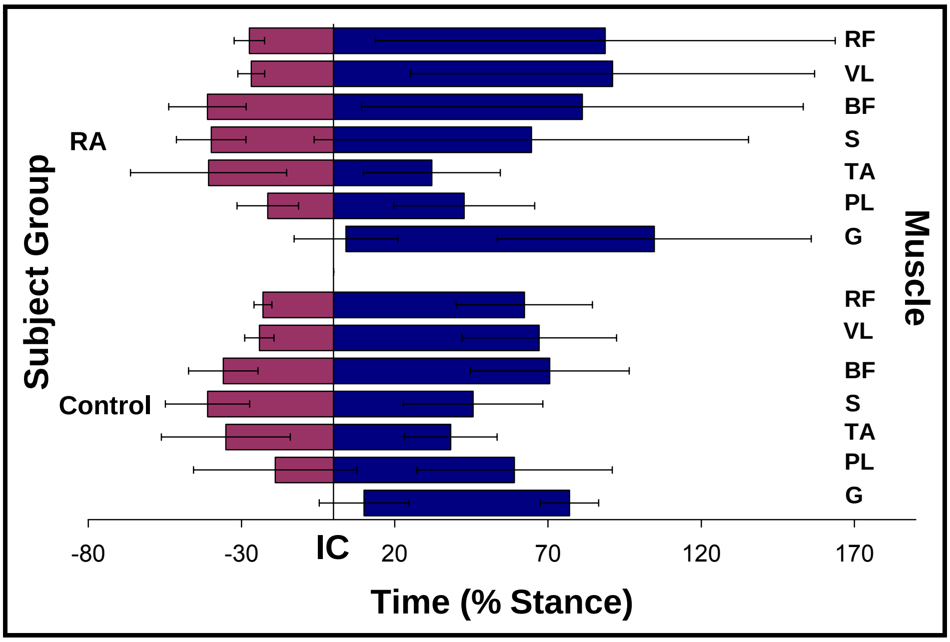

Figure 6.9: Mean (SD) muscle burst onsets and offsets at initial foot-ground contact (IC) for the condition of walking barefoot on carpet for the RA (n = 8) and control (n = 8) subjects.

For example, Miller *et al.*[876] recorded high variability in both the shape and timing of the muscle activation profiles for G, TA and VL when 18 healthy older subjects (mean age, 68 years) and 19 Parkinson's patients (mean age, 71 years) walked at a self-selected pace. The greatest shape variability was exhibited by TA, a muscle that has a very complex role during gait, functioning to control toe clearance, position the foot at initial foot-ground contact and allow a smooth lateral weight transfer (see Section 2.5.4). However, the Parkinson's patients displayed significantly reduced timing variability in gastrocnemius compared to the healthy older subjects, suggesting that increased gastrocnemius timing variability was associated with a more normal gait pattern[876]. As a result, variability in EMG data as well as other kinetic and kinematic variables (see

following sections) may reflect natural adaptations or compensations of muscle activity to allow for stable locomotion[874,876], rather than deficient gait or age-related pathology[876]. Therefore, despite possibly masking significant between-group differences, the high variability displayed by subjects in the present study, particularly for TA and PL burst intensity (see Table 6.11), may be a positive finding. That is, high variability suggests an ability of subjects to alter their movement strategies via altered central processing control patterns when walking under conditions of increased postural threat and/or heightened anxiety[879]. High variability was also found for the within-footwear (see Section 6.6.2(B)) and within-surface (see Section 6.6.2(C)) main effects and consequently, caution must be observed when drawing conclusions about the neuromuscular control of gait in the present study[332,875].

The between-group data pertaining to lower limb phasic muscle activity and muscle burst intensity data are presented in Table 6.11. In the present study, when the data were pooled across footwear and surface conditions, there was no significant main effect of subject group on any of the muscle activation variables (see Table 6.11). However, a strong trend was evident for the RA subjects to display a later G burst offset leading to a longer G burst duration compared to the control subjects (see Table 6.11). It is suggested that this trend, although recorded with low statistical power (45%), could reflect delayed heel rise in the RA subjects (see Section 2.6.3), resulting from increased foot pain (see Section 6.2 and Section 6.6.1(A)) or the use of G to control knee motion compensating for soleus or hamstring weakness[743] (see Section 6.3.3). However, despite this trend and, inconsistent with the subjective estimations of increased task difficulty (see Section 6.6.1(A)), the RA subjects used similar muscle activation strategies to the control subjects to prepare the lower limb for initial foot-ground contact, regardless of footwear type or surface condition.

(B) Within-Footwear Main Effects

Descriptive data pertaining to the lower limb phasic muscle activity and muscle burst intensity patterns displayed by both the RA and control subjects

Table 6.11: Descriptive and statistical information, pooled across footwear and surface condition, pertaining to the lower limb muscle activation patterns at initial foot-ground contact for the RA (n = 8) and control (n = 8) subjects.

Variable	RA Subjects		Control Subjects		$F_{(1,7)}$	*p*-value
	Mean	(SEM)	Mean	(SEM)		
Muscle Burst Duration (ms)						
RF	674.8	(86.3)	721.2	(86.3)	0.15	0.712
VL	734.0	(74.4)	753.5	(80.4)	0.03	0.862
BF	776.7	(73.8)	805.4	(79.7)	0.07	0.796
S	626.2	(78.3)	704.2	(84.5)	0.46	0.512
TA	586.1	(74.0)	548.2	(79.9)	0.12	0.734
PL	675.8	(58.7)	581.3	(69.5)	1.08	0.323
G	589.2	(55.3)	430.3	(59.7)	3.82	0.077
Initial Foot-Ground Contact to Muscle Burst Onset / Stance (%)[a]						
RF	-30.3	(2.5)	-26.2	(2.5)	1.37	0.269
VL	-29.5	(2.7)	-27.1	(2.9)	0.37	0.554
BF	-42.5	(4.4)	-38.5	(4.8)	0.37	0.553
S	-42.8	(6.2)	-44.9	(6.7)	0.05	0.823
TA	-43.7	(7.1)	-44.8	(7.6)	0.01	0.914
PL	-28.9	(5.3)	-21.9	(6.3)	0.73	0.414
G	7.7	(4.9)	17.7	(5.3)	1.94	0.191
Muscle Burst Offset to Initial Foot-Ground Contact / Stance (%)						
RF	63.7	(8.2)	73.7	(8.2)	0.75	0.407
VL	71.4	(8.0)	78.2	(8.6)	0.34	0.571
BF	66.3	(6.5)	74.4	(7.0)	0.72	0.413
S	42.4	(5.1)	55.1	(5.5)	2.88	0.118
TA	36.0	(4.6)	37.4	(5.0)	0.04	0.844
PL	67.6	(6.0)	61.8	(7.1)	0.39	0.546
G	87.9	(3.2)	78.6	(3.4)	4.02	0.070
Muscle Burst Intensity / Barefoot + Carpet Condition (%)[b]						
RF	105.1	(7.1)	107.2	(7.1)	0.05	0.836
VL	101.7	(3.4)	105.8	(3.7)	0.63	0.444
BF	124.3	(9.9)	100.8	(10.7)	2.62	0.134
S	106.7	(9.1)	117.3	(9.8)	0.62	0.448
TA	147.5	(25.4)	111.7	(27.4)	0.92	0.358
PL	191.6	(40.6)	111.3	(48.1)	1.63	0.231
G	94.8	(2.9)	96.0	(3.1)	0.09	0.771

[a] Negative value indicates muscle onset occurred before initial foot-ground contact.
[b] Muscle burst intensity data were normalised to the barefoot + carpet condition, denoted as 100% (see Section 5.6.4).

under different footwear conditions are displayed in Table 6.12 and Figure 6.10, respectively. The *F*-ratios and alpha levels for the within-subject sources of variation for the lower limb phasic muscle activity and muscle burst intensity data are presented in Table 6.13. When the data were pooled across subject group and surface condition, there was a significant main effect of footwear on BF and PL burst duration; RF, VL, TA and PL burst onset relative to initial foot-ground contact; G burst offset relative to initial foot-ground contact; and TA burst intensity (see Table 6.13).

Table 6.12: Descriptive information, pooled across subject group and surface condition, for lower limb muscle synchrony when subjects (n = 16) walked barefoot, in closed back slippers or in toe slippers.

Variable	Barefoot		Closed slipper		Toe slipper	
	Mean	(SEM)	Mean	(SEM)	Mean	(SEM)
Muscle Burst Duration (ms)						
RF	718.6	(54.9)	675.1	(61.0)	700.2	(75.0)
VL	762.0	(56.7)	727.2	(52.5)	742.1	(57.2)
BF	843.3	(52.0)	758.2	(54.0)	771.5	(66.4)
S	659.6	(48.4)	667.8	(65.8)	668.3	(70.4)
TA	521.3	(63.3)	547.6	(65.7)	632.7	(51.5)
PL	591.7	(47.9)	588.7	(61.8)	705.2	(49.4)
G	512.5	(40.4)	512.4	(44.2)	504.3	(41.8)
Initial Foot-Ground Contact to Muscle Burst Onset / Stance (%)[a]						
RF	-30.1	(2.3)	-26.3	(1.4)	-28.4	(2.0)
VL	-29.9	(2.3)	-26.6	(1.9)	-28.3	(2.0)
BF	-43.1	(3.5)	-38.0	(3.2)	-40.5	(4.2)
S	-44.9	(4.4)	-43.6	(5.0)	-43.0	(4.9)
TA	-41.4	(6.2)	-37.0	(6.1)	-54.2	(5.8)
PL	-23.0	(4.9)	-19.4	(3.0)	-33.9	(5.7)
G	9.7	(4.2)	14.9	(4.1)	13.5	(3.2)
Muscle Burst Offset to Initial Foot-Ground Contact / Stance (%)						
RF	72.1	(6.8)	66.2	(5.8)	67.9	(6.7)
VL	77.9	(6.4)	73.1	(5.7)	73.4	(6.0)
BF	76.3	(4.9)	66.9	(5.3)	67.9	(6.0)
S	48.8	(4.9)	49.0	(4.2)	48.4	(4.6)
TA	35.2	(4.2)	38.4	(3.6)	36.5	(3.1)
PL	63.8	(4.9)	63.2	(6.8)	67.1	(5.3)
G	81.3	(2.5)	85.6	(2.6)	82.8	(2.2)

[a] Negative value indicates muscle onset occurred before initial foot-ground contact.

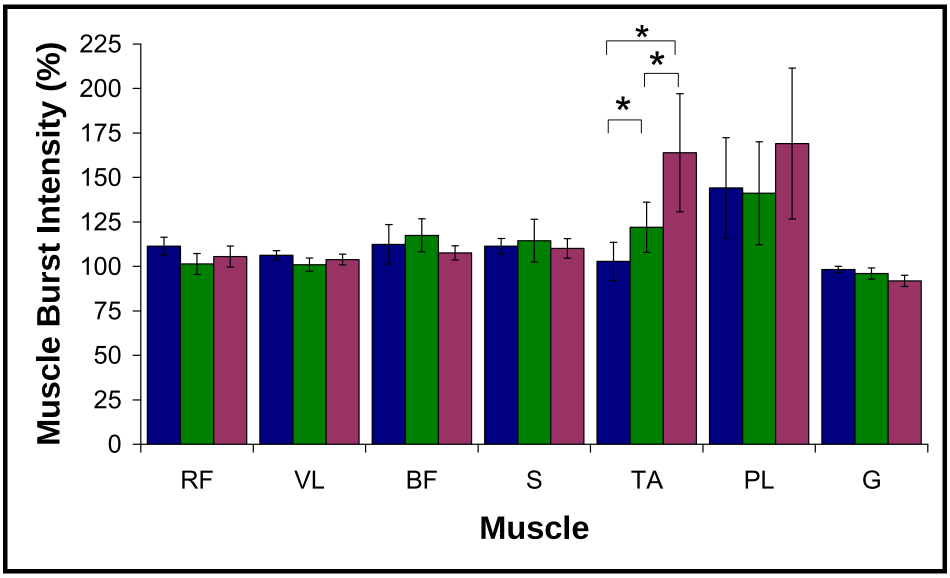

Figure 6.10: The mean (± SEM) muscle burst intensities, pooled across subject group and surface condition, displayed by subjects (n = 16) when they walked barefoot (■), in closed back slippers (■) and in toe slippers (■; * indicates statistical significance at $p \leq 0.05$).

Pairwise comparisons pertaining to quadriceps muscle activity revealed that when the subjects walked barefoot, RF and VL onset occurred significantly earlier compared to when walking in closed back slippers (RF: t = -3.45; p = 0.001; VL: t = -3.09; p = 0.004; see Table 6.12). Furthermore, VL was activated significantly earlier when subjects walked in toe slippers compared to walking in closed back slippers (t = 2.33; p = 0.025; see Table 6.12). When walking barefoot or in toe slippers, earlier quadriceps activation may have been required for greater lower limb control before initial foot-ground contact[497,870], assisting in knee extension and preparing for shock absorption while maintaining balance control[869,870,880,881] compared to walking in closed back slippers (see Section 2.5.4). Earlier RF onset has been suggested to assist the vastii muscles in providing lower limb stability and to ensure adequate toe clearance (see Section 2.3.1(F)). However, it has been suggested that the RF activity seen at initial foot-ground contact may be due to crosstalk from vastus intermedius, as at a normal walking speed, RF does not appear to have a functional role[882]. That is, immediately after initial foot-ground contact, the hip begins to extend and the knee flexes. If RF assists the vastii muscles in controlling knee flexion,

Table 6.13: Factorial ANOVA results for the lower limb muscle activation patterns at initial foot-ground contact.

Variable	Shoe	Shoe x Group	Surface	Surface x Group	Shoe x Surface	Shoe x Surface x Group
	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{4,44}$ *p*-value	$F_{4,44}$ *p*-value
Muscle Burst Duration (ms)						
RF	0.80 0.462	0.90 0.424	7.30 0.004*	0.50 0.614	5.83 0.001*	1.25 0.305
VL	2.57† 0.125	4.55† 0.042*	3.76 0.040*	0.21 0.810	2.03† 0.144	1.96† 0.153
BF	3.55 0.046*	0.09 0.913	0.18 0.833	0.82 0.455	0.62† 0.559	0.59† 0.576
S	0.03 0.972	0.91 0.419	0.76† 0.432	0.48† 0.551	0.91 0.467	1.32 0.277
TA	3.27 0.057	0.29 0.754	1.43 0.261	1.26 0.303	1.10 0.367	0.35 0.842
PL	3.74 0.042*	1.87 0.180	2.38 0.118	0.23 0.795	0.84 0.509	1.38 0.258
G	0.12 0.887	1.79 0.190	1.46 0.253	0.02 0.984	4.06 0.007*	1.36 0.264
Initial Foot-Ground Contact to Muscle Burst Onset / Stance (%)						
RF	3.98 0.035*	1.00 0.386	0.98 0.394	0.45 0.646	1.25† 0.309	0.57† 0.583
VL	5.69 0.010*	0.60 0.556	0.34† 0.618	0.73† 0.439	1.61† 0.226	1.15† 0.331
BF	1.56 0.234	1.24 0.309	1.35 0.281	0.06 0.944	1.47† 0.252	1.03† 0.374
S	0.35† 0.621	0.23 0.702	0.48† 0.551	2.25† 0.152	2.26 0.078	0.06 0.992
TA	5.79 0.010*	1.11 0.348	0.25 0.783	0.45 0.641	1.03 0.403	0.19 0.942
PL	7.55 0.004*	3.05 0.070	1.55† 0.242	0.19† 0.710	0.90 0.439	0.57 0.607
G	2.58 0.099	2.14 0.141	0.69† 0.464	0.45† 0.575	1.16† 0.329	2.54† 0.108
Muscle Burst Offset to Initial Foot-Ground Contact / Stance (%)						
RF	0.72 0.498	0.39 0.680	6.72 0.006*	1.89 0.177	1.92 0.126	2.36 0.069
VL	2.41† 0.141	1.76† 0.210	2.11 0.146	0.44 0.652	0.64 0.635	1.38 0.258
BF	2.69 0.090	0.86 0.439	0.14 0.874	1.19 0.323	0.79† 0.455	0.50† 0.597
S	0.01 0.990	1.09 0.354	1.75 0.197	0.02 0.984	0.63 0.642	1.04 0.397
TA	0.83 0.449	1.28 0.298	4.81 0.018*	4.41 0.024*	0.15 0.961	0.67 0.614

Continued on next page

Table 6.13: Factorial ANOVA results for the lower limb muscle activation patterns at initial foot-ground contact (continued).

Variable	Shoe	Shoe x Group	Surface	Surface x Group	Shoe x Surface	Shoe x Surface x Group
	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{2,22}$ *p*-value	$F_{4,44}$ *p*-value	$F_{4,44}$ *p*-value
PL	0.27 0.768	0.30 0.746	1.35 0.282	0.47 0.634	1.00 0.420	0.49 0.743
G	7.74† 0.009*	1.76 0.208	0.33 0.726	0.30 0.743	2.64 0.083	0.91 0.427
Muscle Burst Intensity / Barefoot + Carpet Condition (%)[a]						
RF	2.36 0.104	1.23 0.315	5.68 0.011*	0.81 0.461	7.77† 0.005*	0.71† 0.487
VL	1.41 0.265	2.23 0.131	2.70 0.089	0.89 0.423	3.39 0.017*	1.42 0.243
BF	0.74† 0.433	0.50† 0.531	0.43 0.657	2.34 0.120	1.02† 0.353	1.00† 0.359
S	0.17† 0.759	1.66† 0.223	6.11† 0.025*	0.23† 0.672	3.46† 0.039*	0.18† 0.874
TA	5.26† 0.035*	0.24† 0.668	0.33† 0.600	0.73† 0.424	1.63 0.185	0.92 0.463
PL	1.04 0.372	0.97 0.396	2.32† 0.149	0.75† 0.434	1.96 0.119	1.52 0.215
G	2.09 0.148	0.71 0.504	2.08† 0.168	0.79† 0.423	2.26 0.077	1.46 0.231

* Indicates statistical significant at $p \leq 0.05$.
† Use of Greenhouse-Geisser correction factor (see Table B.11.1 to Table B.11.8, Appendix B.11).
[a] Muscle burst intensity data were normalised to the barefoot + carpet condition, denoted as 100% (see Section 5.6.4).

additional hip extensor activity would be required to counteract the RF activity, whereas knee flexion can be solely controlled by the vastii muscles without impeding hip extension[882]. Consequently, further research is warranted to determine the different muscle activation patterns for RF and perhaps vastus intermedius when walking barefoot compared to walking shod.

Pairwise comparisons pertaining to hamstring muscle activity indicated that when subjects walked barefoot, BF displayed a significantly longer muscle burst duration compared to when they walked in closed back slippers ($t = 2.22$; $p = 0.033$; see Table 6.12). This increased muscle burst duration appeared to be

the product of a strong trend towards a later BF burst offset relative to initial foot-ground contact (see Table 6.11) and may have led to the slight decrease in BF burst intensity (see Figure 6.10) when barefoot compared to walking in closed back slippers. Eccentric contractions of the hamstring muscles act to decelerate the swinging limb in preparation for initial foot-ground contact (see Section 2.5.4). However, compared to the other hamstring muscles, BF may play a larger role in controlling hip extension during the loading response[348]. Therefore, the longer BF muscle burst duration combined with the earlier activation of RF may reflect the need for additional lower limb stability when walking barefoot, a footwear condition reported as slippery (see Section 6.6.1(B)), compared to walking in either slipper type. This altered quadriceps and hamstring co-contraction pattern may have also assisted to reduce joint contact forces and joint pain when additional impact attenuation was required when barefoot compared to shod walking[883] (see Section 2.5.4).

Pairwise comparisons pertaining to G activity determined that when subjects walked barefoot, G ceased its activity significantly earlier compared to walking either in toe slippers (t = -2.02; p = 0.049) or closed back slippers (t = -3.60; p = 0.001; see Table 6.12). Furthermore, G ceased its activity significantly earlier when subjects walked in toe slippers compared to wearing closed back slippers (t = 2.48; p = 0.018; see Table 6.12). Earlier cessation of G activity may have indicated that subjects had a decreased propulsive force at toe-off, lifting their foot from the supporting surface rather than using their forefoot for propulsion. Therefore, when walking barefoot or in toe slippers, subjects appeared to display compromised functioning of the muscle primarily responsible for generating 80% of the power for propulsion (see Section 2.5.4).

Subjects in the present study displayed significantly earlier quadriceps activation, significantly longer hamstring activation and significantly shorter gastrocnemius activation when walking barefoot compared to walking shod. These differences in muscle activation strategies suggest earlier preparation of lower limb alignment and increased shock absorption at initial foot-ground contact as well as greater lower limb, knee and trunk control during stance when walking barefoot. Altered ground reaction force (see Section 6.6.3(B)) and

kinematic (see Section 6.6.4(B)) profiles displayed when subjects walked barefoot compared to walking in either slipper type confirmed this notion.

Pairwise comparisons pertaining to the TA and PL muscles revealed that when subjects walked wearing the toe slippers, both TA and PL were activated significantly earlier compared to walking either barefoot (TA: $t = 4.22$; $p < 0.001$; PL: $t = 3.52$; $p = 0.001$) or in closed back slippers (TA: $t = 5.04$; $p < 0.001$; PL: $t = 4.88$; $p < 0.001$; see Table 6.12). Furthermore, PL exhibited a significantly earlier onset when subjects walked barefoot compared to walking in closed back slippers ($t = -2.00$; $p < 0.053$; see Table 6.12). Earlier TA and PL onset contributed to a strong trend towards a longer TA burst duration and a significantly longer PL burst duration when subjects walked wearing toe slippers compared to walking either barefoot ($t = -3.45$; $p = 0.001$) or in closed back slippers ($t = -2.35$; $p = 0.024$; see Table 6.12). In addition, TA displayed significantly greater muscle burst intensity when subjects walked in toe slippers compared to walking either barefoot ($t = -4.61$; $p < 0.001$) or in closed back slippers ($t = -3.16$; $p = 0.003$) and when subjects wore closed back slippers compared to when barefoot ($t = -2.86$; $p = 0.007$; see Figure 6.10). However, despite notable PL burst intensity differences between the footwear conditions (see Figure 6.10), high variability[*] was evident, possibly masking statistical differences in the data.

During normal gait, TA, with assistance from PL[497,540], exercises its major activity at the end of the swing phase, contracting isometrically to maintain the foot in a slightly dorsiflexed position, allowing for safe foot placement at initial foot-ground contact[238,239,514,884] (see Section 2.5.4 and Section 6.6.4). Immediately after initial foot-ground contact, TA contracts eccentrically to lower the foot to the ground[335,535], control foot pronation[238] and prevent buckling of the stance limb[539,540] (see Section 2.5.4). Therefore, sufficient force output of TA appears critical to foot stability during initial foot-ground contact[15,183,885], without which, unsafe foot placement may result, placing older women at an increased risk of falls[885]. In fact, studies evaluating the force output of all major lower-extremity muscle groups revealed that ankle

[*] Coefficients of variation: barefoot = 114%; closed back slipper = 81% and toe slipper = 113%.

dorsiflexor weakness was the strongest predictor of falls among older people[815,885,886]. Furthermore, Wolfson *et al*.[532] concluded that ankle dorsiflexor muscle strength among older people with a history of repetitive falls was one tenth that of nonfallers.

As increased shock absorption provided from a shoe sole has been associated with reduced TA burst intensity[887] and the altered muscle control strategies discussed previously suggest an increased need for shock absorption, it was anticipated that TA burst intensity would decrease when subjects walked shod compared to walking barefoot. Instead, it is postulated that the increased TA and PL activation evident in the present study was required by the subjects when they walked in the toe slippers, as subjects reported that these slippers were difficult to keep on their feet (see Section 6.6.1(B)). That is, earlier TA and PL activation may have been required to maintain ankle dorsiflexion (see Section 6.6.4(B)) to ensure adequate shoe clearance during swing thereby reducing trip propensity (see Section 2.2.3). Earlier TA and PL activation may have also been required when wearing toe slippers to ensure correct ankle alignment at initial foot-ground contact (see Section 2.3.2(C)). In addition, and commensurate with the subjective estimations of shoe slipperiness (see Section 6.6.1(B)), Li[888] reported finding increased TA and PL activity when younger females (mean age, 23 years) walked wearing less slip-resistant shoes. Interestingly, the earlier offset of G displayed when subjects walked in toe slippers compared to wearing closed back slippers, suggested subjects may have reduced ankle plantar flexion at terminal stance (see Figure 2.4) so that their foot did not slide out of the toe slippers (see Figure 6.5). Finally, the increased TA intensity required when subjects walked in either slipper pair compared to walking barefoot may have been due to the slight posterior heel flare and increased heel height of the slippers, requiring increased muscular control at initial foot-ground contact to prevent forefoot slap[889].

When muscle weakness is present, as for the RA subjects (see Section 6.3.3), it is possible for muscle activation levels to be higher than expected as a greater proportion of the muscle fibres have to be activated to resist a given load. In the present study, both the RA and control subjects may have had to use a

greater percentage of their available muscle strength when walking in toe slippers compared to the other footwear conditions, using on average 70% more muscle intensity than that required when walking barefoot (see Figure 6.10). Consequently, these older women may be placed at risk of muscle fatigue and therefore fatigue-related falls[538] due to prolonged lower limb muscle activity (see Section 2.3.1(F)). Therefore, toe slippers would not be recommended for older women due to the altered TA, PL and G activation patterns required and the resulting gait changes, which may lead to an increased risk of falls in this population, particularly the RA subjects who had strength deficits in these muscles (see Section 6.3.3).

(C) Within-Surface Main Effects

Descriptive data pertaining to the lower limb phasic muscle activity and muscle burst intensities displayed at initial foot-ground contact when the subjects walked across the different surface types are displayed in Table 6.14 and Figure 6.11, respectively. When the data were pooled across subject group and footwear condition, significant main effects of surface type were calculated for RF and VL muscle burst duration; RF and TA muscle burst offset relative to initial foot-ground contact; and RF and S muscle burst intensity (see Table 6.13).

Pairwise comparisons pertaining to the quadriceps muscle activation patterns revealed that when subjects walked across the wet vinyl tile surface, RF ceased its activity significantly later compared to when walking across carpet ($t = -3.39$; $p = 0.002$) or dry vinyl tile ($t = -4.11$; $p < 0.001$; see Table 6.14). This delayed muscle burst offset led to a significantly longer RF burst duration when subjects walked on the wet vinyl tile compared to walking across carpet ($t = -4.28$; $p < 0.001$) or dry vinyl tile ($t = -4.28$; $p < 0.001$; see Table 6.14). As these changes in RF reflected the activity of VL, RF may be recruited to assist the vastii muscles in their role of controlling knee flexion during stance (see Table 6.14) or, as noted previously, could be the result of crosstalk from vastus intermedius (see Section 6.6.2(B)). However, when subjects walked across the wet vinyl tile surface, they required significantly greater RF intensity compared to walking across carpet ($t = -3.54$; $p = 0.001$) or dry vinyl tiles ($t = -3.26$; $p =$

0.002; see Figure 6.11). In comparison, when the subjects walked across the carpet surface, VL burst duration was significantly shorter compared to when they walked across the dry (t = -2.18; p = 0.035) or wet (t = -2.26; p = 0.030) vinyl tile surfaces (see Table 6.14), reflecting the activation patterns of RF (see Table 6.14) and perhaps increased walking stability on the carpet surface.

Table 6.14: Descriptive information, pooled across subject group and footwear condition, for the lower limb muscle activation patterns at initial foot-ground contact when subjects (n = 16) walked across carpet, dry vinyl tile or wet vinyl tile.

Variable	Carpet		Dry vinyl tile		Wet vinyl tile	
	Mean	**(SEM)**	**Mean**	**(SEM)**	**Mean**	**(SEM)**
Muscle Burst Duration (ms)						
RF	666.2	(58.8)	666.4	(63.9)	761.4	(66.8)
VL	709.1	(54.1)	757.1	(57.0)	765.1	(57.6)
BF	783.0	(55.6)	788.6	(61.7)	801.5	(54.1)
S	647.1	(66.1)	683.9	(59.2)	664.6	(54.4)
TA	601.9	(48.3)	567.8	(61.2)	531.9	(67.4)
PL	579.8	(38.8)	694.3	(61.2)	611.6	(62.3)
G	514.0	(41.8)	492.6	(44.6)	522.5	(39.4)
Initial Foot-Ground Contact to Muscle Burst Onset / Stance (%)[a]						
RF	-27.0	(1.4)	-29.0	(2.1)	-28.8	(2.3)
VL	-27.6	(1.9)	-28.8	(2.0)	-28.4	(2.5)
BF	-39.0	(3.3)	-40.2	(3.6)	-42.3	(3.5)
S	-44.0	(4.7)	-44.5	(4.6)	-43.0	(4.7)
TA	-45.4	(5.6)	-44.5	(4.8)	-42.8	(6.4)
PL	-24.3	(5.6)	-28.9	(4.9)	-23.1	(2.7)
G	11.8	(3.6)	14.3	(3.8)	12.1	(4.1)
Muscle Burst Offset to Initial Foot-Ground Contact / Stance (%)						
RF	66.7	(5.9)	64.9	(5.9)	74.6	(6.1)
VL	71.4	(5.8)	77.0	(6.5)	76.0	(6.0)
BF	70.3	(4.9)	71.2	(5.5)	69.6	(4.8)
S	46.1	(4.0)	51.7	(3.8)	48.4	(4.5)
TA	40.8	(3.5)	37.7	(4.2)	31.5	(3.8)
PL	60.8	(5.7)	70.6	(5.2)	62.6	(6.8)
G	83.0	(2.3)	83.7	(2.2)	83.0	(2.6)

[a] Negative value indicates muscle onset occurred before initial foot-ground contact.

When walking under different footwear conditions, RF onset was significantly altered, suggesting the need for greater lower limb stabilisation at

initial foot-ground contact when wearing different footwear (see Section 6.6.2(B)). However, when walking on different surface types, these significant differences were also found in RF offset, suggesting greater lower limb stabilisation was also required during the stance phase of gait when the surface characteristics are altered (see Section 2.3.1(F)). More specifically, these changes in RF activation appeared to provide a more stable gait pattern when subjects walked on the wet vinyl tile surface, a surface considered difficult, slippery and uncomfortable to walk upon (see Section 6.6.1(C)). Conversely, when walking on the carpet surface, the subjects displayed greater ease of controlling the lower limb, reflecting a more confident and stable gait pattern.

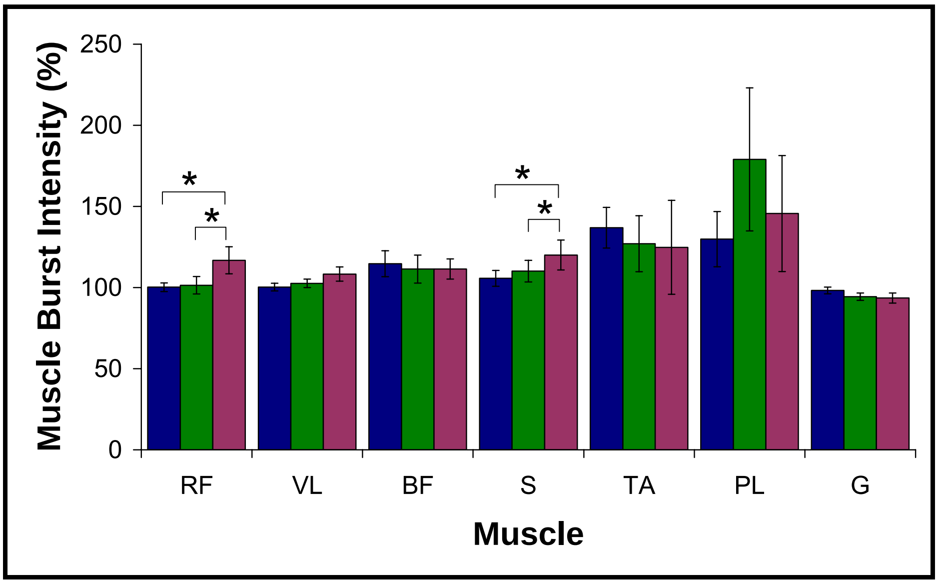

Figure 6.11: The mean (± SEM) muscle burst intensities, pooled across subject group and footwear condition, displayed by subjects (n = 16) when they walked across carpet (■), dry vinyl tile (■) and wet vinyl tile (■; * indicates statistical significance at $p \leq 0.05$).

Pairwise comparisons pertaining to hamstring muscle activation indicated that subjects required significantly greater S burst intensity when they walked across the wet vinyl tile surface compared to walking across the carpet (t = -2.79; p = 0.008) or dry vinyl tile (t = -2.51; p = 0.016) surfaces (see Figure 6.11). Increased S burst intensity may have assisted the other hamstring muscles in decelerating the lower limb in preparation for initial foot-ground contact.

However, this increase in S intensity may have also been required to compensate for increased RF activation, indicating greater co-contraction of the knee extensor and flexor muscles during loading to stabilise the knee joint when absorbing body weight[332] (see Section 2.5.3). Therefore, it is anticipated that due to this co-contraction, the knee moments will be reduced allowing the individual to adapt to the more "dangerous" situation (see Section 2.5.4). The knee moments generated during the walking trials are discussed further in Section 6.6.5(C).

Pairwise comparisons pertaining to TA activity determined that when subjects walked across the wet vinyl tile surface, TA ceased its activity significantly earlier compared to walking across either carpet (t = 2.32; p = 0.025) or dry vinyl tile (t = 2.40; p = 0.021; see Table 6.14). Contrary to the results indicated for different footwear types (see Section 6.6.2(B)), earlier TA offset may have indicated that subjects contacted the supporting surface with a less dorsiflexed foot and a lower foot/shoe to floor angle when walking across the wet surface (see Section 6.6.4(C)). A flatter foot at initial foot-ground contact would reduce the time until foot flat as well as increase the available surface area at contact, transferring a greater percentage of body weight to the supporting lower limb in an attempt to produce a more stable gait on this potentially hazardous wet surface (see Section 2.5.2).

(D) Interaction Effects

There were no significant subject group x footwear x surface interactions pertaining to muscle burst activation patterns in the present study (see Table 6.13). However, despite no between-group main effects (see Section 6.6.2(A)), subject group was moderated by both footwear and surface condition such that a significant subject group x footwear interaction was calculated for VL burst duration and a significant subject group x surface interaction was calculated for TA burst offset relative to initial foot-ground contact (see Table 6.13).

Step down procedures pertaining to the subject group x footwear interaction for VL burst duration indicated that, despite a significant interaction,

only the barefoot condition neared significance ($F_{1,44}$ = 3.32; p = 0.075; see Figure 6.12) when the data were pooled across surface type, although both the barefoot and toe slipper conditions displayed similar trends. That is, whereas when walking in closed back slippers both the RA and control subjects recorded similar VL burst durations, when walking barefoot or in toe slippers, the RA subjects displayed a longer VL burst duration than the control subjects (see Figure 6.12). Consequently, the longer VL burst duration may have been instrumental in controlling knee flexion during stance, as well as contributing to lower limb stability when walking under footwear conditions perceived as uncomfortable (see Section 6.6.1(B)). However, further research is warranted to investigate the ramifications of this finding.

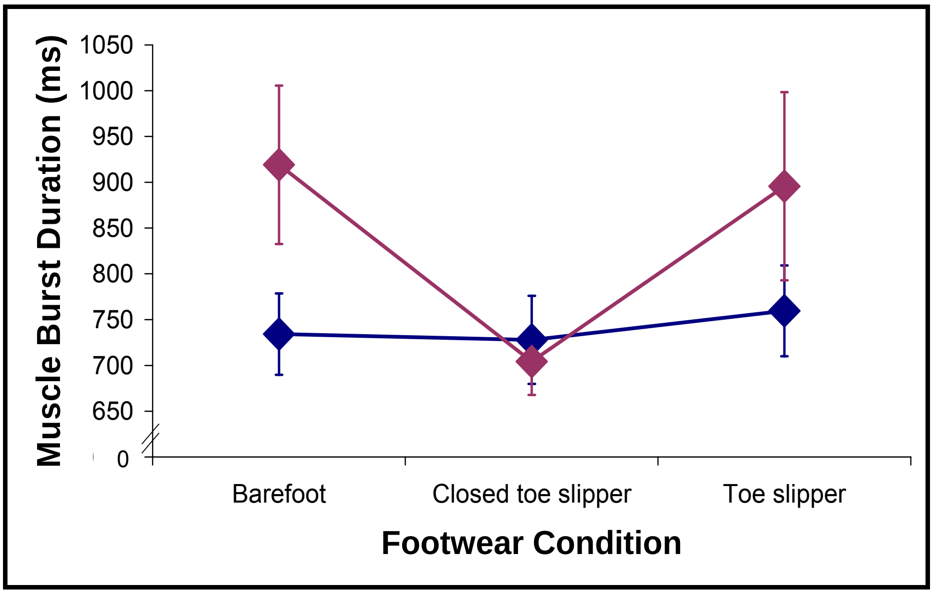

Figure 6.12: Subject group x footwear interaction for vastus lateralis burst duration for the control (n= 8; ▬) and RA (n= 8; ▬) subjects under different footwear conditions.

Step down procedures pertaining to the subject group x surface interaction for TA burst offset relative to initial foot-ground contact indicated that the control subjects displayed a significantly later TA offset relative to initial foot-ground contact compared to the RA subjects but only when walking across the carpet surface ($F_{1,44}$ = 2.28; p = 0.027; see Figure 6.13). That is, when

the control subjects walked across the dry vinyl tile surface, they ceased TA activity earlier than when walking on the carpet and when walking across the wet vinyl tile, TA offset occurred earlier than for the RA subjects. Prolonged TA activity has been reported for diabetic patients who demonstrate increased co-contraction of TA with G[300]. This co-contraction is required to stabilise the ankle upon initial foot-ground contact and improve foot stability during early stance[300] (see Section 2.5.4), perhaps even functioning as an adaptive strategy to compensate for diminished sensory information from the ankle and foot[890]. One explanation for this prolonged TA activity displayed by the control subjects could be that the control subjects in the present study recorded decreased plantar sensation compared to the RA subjects (see Section 6.4.2) whereby their diminished sensory information compromised TA activity. However, further research is warranted to determine why TA activity was prolonged when the control subjects walked on the carpet surface compared to why it was not prolonged when both subject groups walked on the slippery, wet vinyl tile surface (see Section 6.6.1(D)).

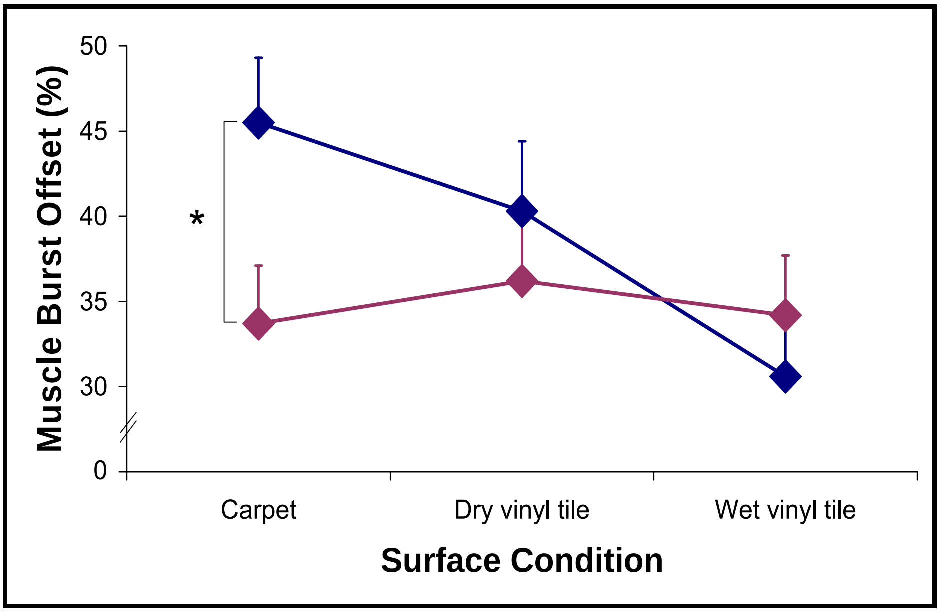

Figure 6.13: Subject group x surface interaction for tibialis anterior burst offset with respect to initial foot-ground contact for the control (n= 8; —) and RA (n= 8; —) subjects under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

Together with the significant two-way interactions involving subject group, significant footwear x surface interactions were also calculated for RF and G burst duration as well as RF, VL and S burst intensity (see Table 6.13). Step down procedures pertaining to the significant footwear x surface interactions indicated that when subjects walked barefoot on the wet vinyl tile surface, they recorded significantly longer RF and G burst durations compared to walking in closed back slippers (RF: t = 3.49; p = 0.005; see Figure 6.14; G: t = 2.33; p = 0.038; see Figure 6.15). Furthermore, although not significant for G, subjects in the present study displayed a significantly longer RF burst duration when they walked barefoot on the wet vinyl tile surface compared to walking in toe slippers (t = 3.21; p = 0.007; see Figure 6.14).

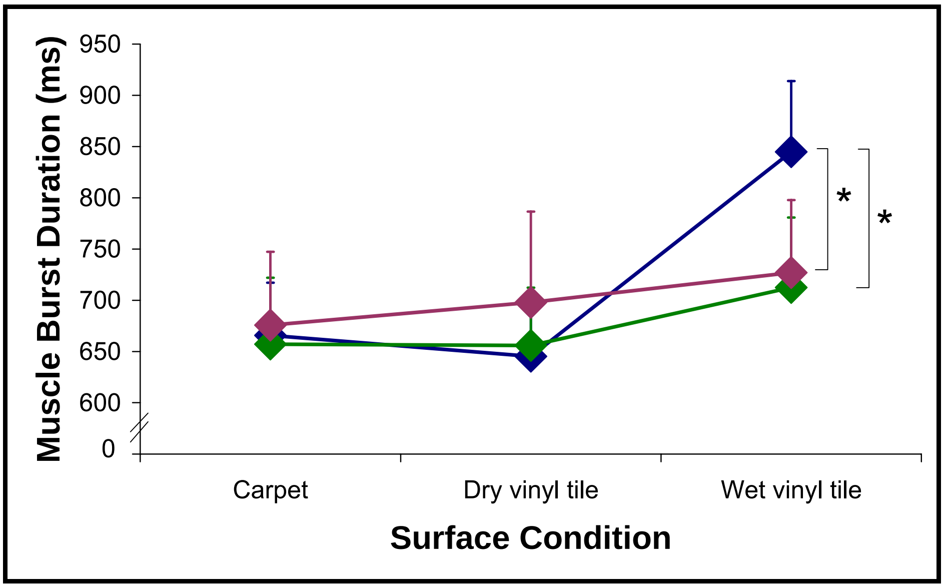

Figure 6.14: Footwear x surface interaction for rectus femoris burst duration when subjects (n= 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

Step down procedures pertaining to the footwear x surface interactions for muscle burst intensity indicated that when subjects walked on the wet vinyl tile surface, they required significantly greater RF and VL burst intensity when walking barefoot compared to walking in closed back slippers (RF: t = 3.05;

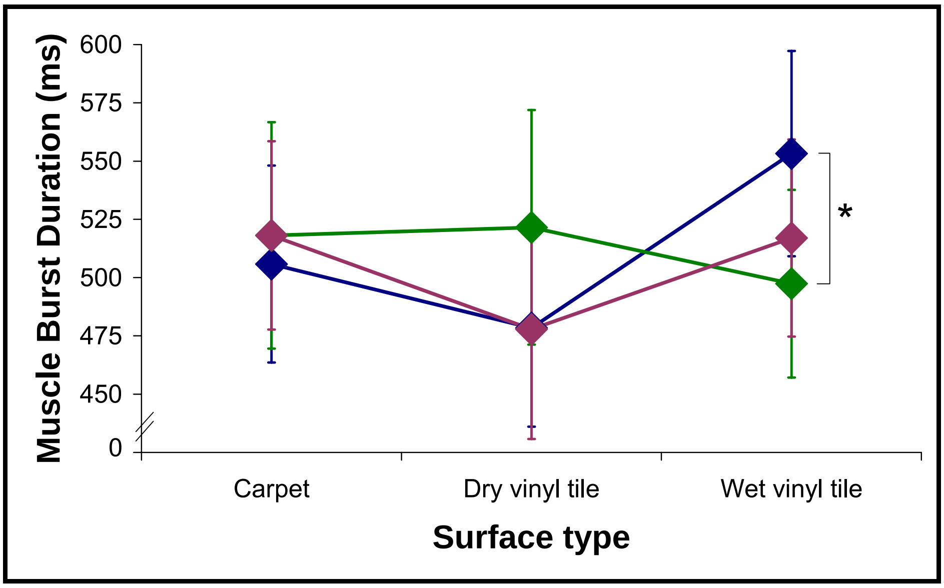

Figure 6.15: Footwear x surface interaction for gastrocnemius burst duration when subjects (n= 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

p = 0.011; see Figure 6.16; VL: t = 2.27; p = 0.042; see Figure 6.17). Furthermore, subjects in the present study displayed significantly greater RF and S burst intensity when they walked barefoot on the wet vinyl tile surface compared to walking in toe slippers (RF: $t = 2.76$; $p = 0.016$; see Figure 6.16; S: $t = 3.66$; $p = 0.003$; see Figure 6.18). A similar trend was evident for increased VL burst intensity when subjects walked barefoot on the wet vinyl tile surface compared to walking in toe slippers ($t = 1.91$; $p = 0.077$; see Figure 6.17).

Few studies have assessed the muscle activation patterns of older people, particularly those with arthritis and foot pain, when walking in different types of footwear and on different surfaces. Those studies located appear only to explain how the muscle activation patterns alter when people wear high heels[888,891] or prescribed shoes or orthotics designed for specific foot and/or lower limb problems[892,893]. However, different shoes, orthotics and supporting surfaces act as filters to the information provided to the sensory receptors in the foot[497,503,533,619]. This altered sensory input, in turn, may lead to changing muscle activation patterns[894]. Consequently, it was interesting to note that all

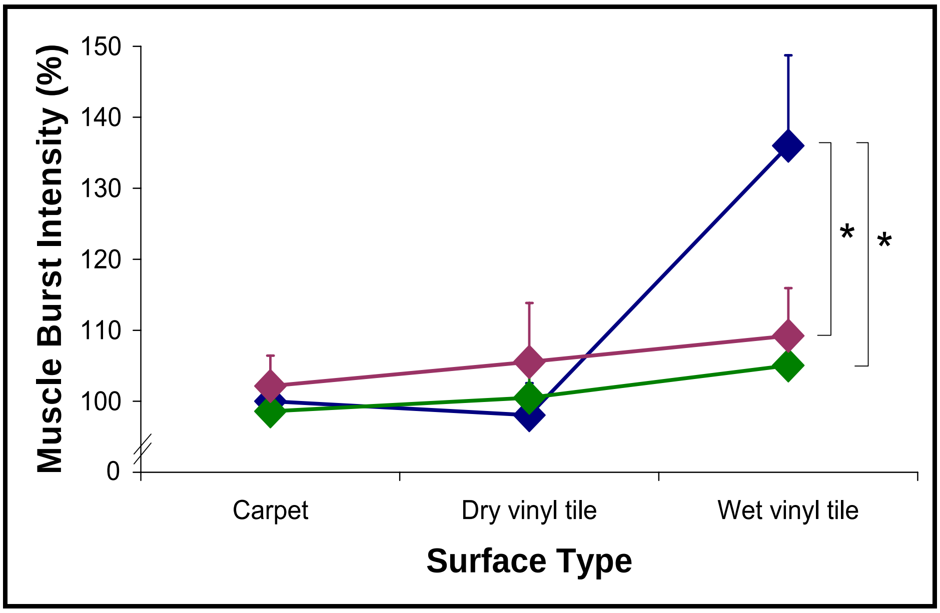

Figure 6.16: Footwear x surface interaction for rectus femoris burst intensity when subjects (n= 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

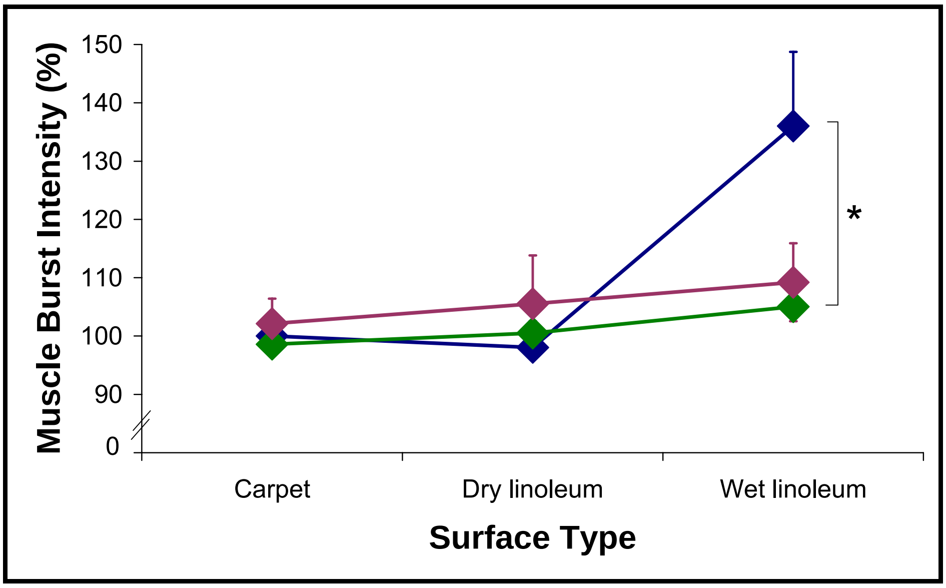

Figure 6.17: Footwear x surface interaction for vastus lateralis burst intensity when subjects (n= 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

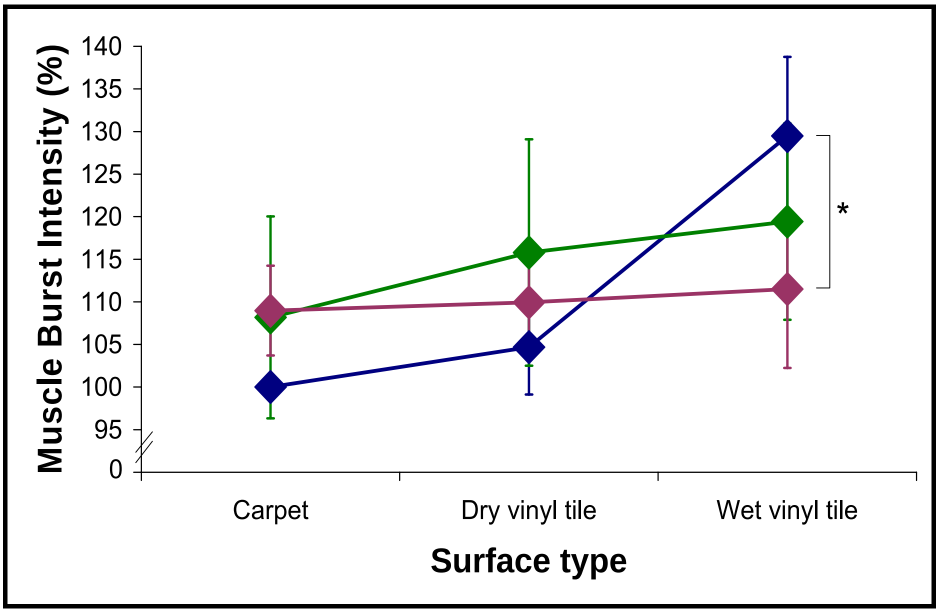

Figure 6.18: Footwear x surface interaction for semitendinosus burst intensity when subjects (n= 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

the significant footwear x surface interactions related to altered muscle control strategies when subjects walked barefoot on the wet vinyl tile surface (see Figures 6.14 to 6.18). Furthermore, despite the main effects of footwear on the muscles controlling ankle motion (see Section 6.6.2(B)), the muscles affected by barefoot walking on the wet vinyl tile surface were the quadriceps, hamstrings and gastrocnemius muscles. These muscles play a large role in stabilising both the lower limb and trunk during stance[895] (see Section 2.5.4) and may provide assistance to each other[534], as well as compensate for any muscle deficiencies at the ankle, knee and hip joints[896-898]. Therefore, these altered muscle control strategies may have been required by the subjects in the present study to ensure they remained upright when walking barefoot on the wet vinyl tile surface by ensuring greater lower limb stability and shock absorption during initial foot-ground contact and the during the stance phase of gait[270,867], relative to their usual muscle control strategies. However, it is important to realise that this greater control may necessitate a greater energy cost during gait[893], perhaps

increasing falls risk due to fatigue if the women were required to walk on a wet slippery surface for an extended period.

The altered muscle control strategies noted when subjects walked barefoot on the wet vinyl tile surface were consistent with the subjective estimation data whereby significant footwear x surface interactions were also noted for task difficulty, shoe/surface slipperiness and shoe comfort (see Section 6.6.1(D)). Consequently, barefoot walking, particularly on slippery surfaces, would not be recommended for older women around the home. Although five of the eight control subjects preferred to walk barefoot around their homes (see Section 6.2), it has been proposed that people become so accustomed to wearing shoes that footwear becomes a habitual extension of the foot[454,899] and that the barefoot state is unfamiliar[899]. As this difference was only evident when the subjects were placed in situations deemed more threatening to stability, namely when walking on the slippery floor surface, further investigation is warranted to determine the true effects of walking barefoot compared to shod on the muscle control strategies required by older women.

6.6.3 Ground Reaction Forces and the Coefficient of Friction at Initial Foot-Ground Contact

(A) Between-Group Main Effects

The descriptive data pertaining to the vertical and anteroposterior ground reaction forces and the coefficient of friction data calculated at initial foot-ground contact are presented in Table 6.15. The ground reaction force data recorded for the RA and control subjects were very similar both to each other (see Table 6.15) and to previous reports[385,833,900,901]. However, the RA subjects recorded longer time from initial foot-ground contact until the force peaks suggesting a slower rate of joint loading compared to the control subjects (see Table 6.15). In addition, although not significant, the RA subjects displayed longer stance times suggestive of slower gait (see Section 2.3.1(D)). Similarly, although the RA and control subjects displayed comparable static and dynamic coefficients of friction (see Table 6.15), the calculated values were lower than

the static (0.50) and dynamic (0.28) values reported in previous studies for safe walking[86,385,399,411,888,902] (see Section 2.3.2(C)). In fact, the dynamic coefficients of friction were at values suggested to evoke slips, albeit with a recoverable loss of balance, with the 0.15 value, the value recorded for the RA subjects (see Table 6.15), being the threshold below which it is suggested that slips generally result in unrecoverable losses of balance and consequent falls[398,399,411] (see Section 2.3.2(C)).

Table 6.15: Descriptive and statistical information, pooled across footwear and surface condition, for the ground reaction force and coefficient of friction variables for the RA (n = 8) and control (n = 8) subjects.

Variable[a]	RA Subjects		Control Subjects		$F_{(1,7)}$	*p*-value
	Mean	(SEM)	Mean	(SEM)		
Vertical Ground Reaction Forces						
Stance (ms)	751.5	(40.9)	700.4	(57.8)	0.52	0.494
F_{yB} (N.kg^{-1})	10.3	(0.1)	10.3	(0.2)	0.01	0.943
IC-F_{yB} (%)	29.3	(1.6)	28.0	(2.3)	0.19	0.673
F_{yM} (N.kg^{-1})	8.5	(0.3)	8.2	(0.5)	0.25	0.164
IC-F_{yM} (%)	49.8	(0.6)	47.2	(0.8)	6.68	0.036*
Anteroposterior Ground Reaction Forces						
F_{xB} (N.kg^{-1})	1.2	(0.2)	1.2	(0.2)	0.11	0.746
IC-F_{xB} (%)	19.0	(0.5)	19.0	(0.6)	0.01	0.918
IC-F_{xC} (%)	55.3	(4.6)	60.4	(5.0)	0.54	0.477
Coefficient of Friction						
μ_S	0.49	(0.06)	0.40	(0.06)	1.12	0.400
μ_D	0.15	(0.03)	0.17	(0.03)	0.40	0.593
IC-μ_D (%)	7.30	(0.24)	8.07	(0.24)	5.11	0.152

[a] See Table 5.4 for variable definitions.
* Indicates statistical significance at $p \leq 0.05$.

When the data were pooled across footwear and surface condition, there were no significant between-group main effects for any of the ground reaction force or coefficient of friction parameters except IC-F_{yM} (see Table 6.15). That is, the RA subjects displayed a significantly longer time from initial foot-ground contact to midstance, suggesting that the RA subjects may have delayed terminal stance, spending a greater amount of time in early stance (see Section 2.6.3). Spending increased time in early stance would, in turn, delay loading of the

metatarsal heads (see Section 2.6.3) and foot pain often associated with metatarsal head loading (see Section 6.6.1(A)). However, despite this one significant finding, subjects in the present study, independent of their RA status, exhibited similar loading and frictional patterns at initial foot-ground contact when performing the walking trials (see Table 6.15). These results would suggest that the RA subjects tested in the present study were not at any greater risk of slipping than the control subjects regardless of footwear or surface condition.

(B) Within-Footwear Main Effects

Descriptive data pertaining to the vertical and anteroposterior ground reaction forces and the coefficient of friction data calculated at initial foot-ground contact when subjects walked under different footwear conditions are displayed in Table 6.16. The *F*-ratios and alpha levels for the within-subject sources of variation for the vertical and anteroposterior ground reaction forces and the coefficient of friction data are presented in Table 6.17. When the data were pooled across subject group and surface condition, there was a significant main effect of footwear on peak F_{yB}, peak F_{xB} and μ_S (see Table 6.17).

Pairwise comparisons conducted on the significant findings revealed that the peak F_{yB} were significantly reduced when subjects walked barefoot compared to walking in toe slippers ($t = -2.04$; $p = 0.050$; see Table 6.16). Furthermore, subjects displayed significantly reduced peak F_{xB} when subjects walked barefoot compared to when they walked in closed back ($t = -4.11$; $p < 0.001$) or toe ($t = -3.74$; $p = 0.001$) slippers (see Table 6.16). Pairwise comparisons indicated that the μ_S was significantly reduced when subjects walked barefoot compared to when walking in closed back slippers ($t = -2.53$; $p = 0.017$; see Table 6.16).

The forces imposed on the body at initial foot-ground contact must be attenuated by the body's anatomical structures, together with coordinated motion of the lower limb, learned anticipatory muscular actions and shock absorption provided by a shoe sole (see Section 2.5.3). However, if the efficiency of the

Table 6.16: Descriptive information, pooled across subject group and surface condition, for the ground reaction force and coefficient of friction variables when subjects (n = 16) walked barefoot, in closed back slippers or in toe slippers.

Variable[a]	Barefoot		Closed slipper		Toe slipper	
	Mean	(SEM)	Mean	(SEM)	Mean	(SEM)
Vertical Ground Reaction Forces						
Stance (ms)	716.4	(38.7)	735.0	(33.8)	726.4	(34.4)
F_{yB} ($N.kg^{-1}$)	10.1	(0.1)	10.3	(0.1)	10.4	(0.1)
IC-F_{yB} (%)	27.7	(0.9)	27.6	(0.8)	30.7	(2.9)
F_{yM} ($N.kg^{-1}$)	8.4	(0.3)	8.3	(0.3)	8.3	(0.3)
IC-F_{yM} (%)	48.6	(1.4)	48.6	(0.9)	48.3	(0.6)
Anteroposterior Ground Reaction Forces						
F_{xB} ($N.kg^{-1}$)	1.1	(0.1)	1.3	(0.1)	1.2	(0.1)
IC-F_{xB} (%)	19.8	(1.1)	18.9	(0.4)	18.2	(0.8)
IC-F_{xC} (%)	56.9	(2.4)	64.0	(8.8)	52.7	(1.7)
Coefficient of Friction						
μ_S	0.25	(0.05)	0.31	(0.05)	0.30	(0.05)
μ_D	0.16	(0.01)	0.17	(0.04)	0.15	(0.02)
IC-μ_D (%)	8.06	(0.31)	7.27	(0.07)	7.72	(0.22)

[a] See Table 5.4 for variable definitions.

body's natural shock absorption mechanisms are reduced because of musculoskeletal disease and/or pain, an individual may adapt their gait to elicit pain-avoidance gait strategies, which may in some instances increase their falls risk (see Section 2.5.3 and Section 2.3.1(F)). Therefore, the reduced forces displayed by the subjects at initial foot-ground contact when walking barefoot compared to when walking shod may reflect an attempt to reduce joint loading and consequent joint pain[903,904], compensating for the lack of cushioning normally provided by a shoe sole[522,905,906] (see Section 2.5.3). Reduced braking forces have also been reported in slip-avoidance gait patterns (see Section 2.3.1(F), Section 2.5 and Section 2.6.3) and, as subjects perceived barefoot walking to be more slippery than walking shod (see Section 6.6.1(B)), these altered force profiles, in turn, increased the slip-resistance of barefoot walking (see Table 6.16). Therefore, it appears that the compensatory muscle control strategies (see Section 6.6.2(B)) used to alter the ground reaction force and coefficient of friction profiles, in part determined by the subjective estimations

of the footwear conditions (see Section 6.6.1(B)), allowed subjects to successfully maintain their required coefficient of friction for each footwear condition (see Table 6.16).

Table 6.17: *F*-values and alpha levels for the three-way repeated measures ANOVA results for the ground reaction force and coefficient of friction variables.

Variable[a]	Shoe	Shoe x Group	Surface	Surface x Group	Shoe x Surface	Shoe x Surface x Group
	$F_{2,14}$ *p*-value	$F_{2,14}$ *p*-value	$F_{2,14}$ *p*-value	$F_{2,14}$ *p*-value	$F_{4,28}$ *p*-value	$F_{4,28}$ *p*-value
Stance (ms)	2.51 0.117	0.39 0.687	1.26 0.310	0.40 0.566	5.78† 0.022*	0.87† 0.425
F_{yB} ($N.kg^{-1}$)	4.05 0.041*	0.38 0.690	3.53 0.057	0.40 0.675	0.35† 0.620	0.62† 0.488
IC-F_{yB} (%)	1.53† 0.258	2.38† 0.161	4.01† 0.084	2.53† 0.155	1.97† 0.202	1.77† 0.224
F_{yM} ($N.kg^{-1}$)	0.24 0.792	3.02 0.081	1.43 0.272	1.35 0.291	0.67† 0.526	0.86† 0.443
IC-F_{yM} (%)	0.03† 0.883	0.20† 0.684	0.55 0.587	0.15 0.865	0.14† 0.801	0.47† 0.573
Anteroposterior Ground Reaction Forces						
F_{xB} ($N.kg^{-1}$)	12.12 <0.001*	2.77 0.085	12.96† 0.002*	0.85† 0.401	7.61 <0.001*	0.62 0.649
IC-F_{xB} (%)	0.84 0.398	0.29 0.639	1.56 0.232	1.08 0.357	0.86† 0.440	1.32† 0.287
IC-F_{xC} (%)	1.29† 0.284	1.18† 0.304	1.34† 0.272	0.91† 0.364	1.06† 0.329	1.46† 0.254
Coefficient of Friction						
μ_S	14.19 0.015*	5.06 0.080	9.23 0.032*	0.76 0.929	0.32 0.856	0.44 0.777
μ_D	0.20 0.824	0.41 0.688	0.29 0.764	0.29 0.761	1.30 0.348	0.88 0.516
IC-μ_D (%)	5.18 0.078	1.67 0.297	3.98 0.112	0.74 0.535	0.38† 0.605	2.85† 0.231

[a] See Table 5.4 for variable definitions.
* Indicates statistical significance at $p \leq 0.05$.
† Use of Greenhouse-Geisser correction factor (see Table B.11.9 to Table B.11.11, Appendix B.11).

(C) Within-Surface Main Effects

The ground reaction forces generated at initial foot-ground contact together with the coefficient of friction data calculated when subjects walked across the different surface types are displayed in Table 6.18. When the data were pooled across subject group and footwear condition, although there was a strong trend towards subjects increasing their peak F_{yB} when they walked across carpet compared to walking across vinyl tile (see Table 6.18), there was no significant main effect of surface type on any of the vertical ground reaction force variables (see Table 6.17). However, there was a significant main effect of surface type on the peak F_{xB} and μ_S. That is, subjects displayed significantly reduced peak F_{xB} and μ_S when they walked across the wet vinyl tile surface compared to walking across carpet (peak F_{xB}: $t = 4.71$; $p < 0.001$; μ_S: $t = 4.75$; $p < 0.0$) or dry vinyl tile (peak F_{xB}: $t = 6.06$; $p = 0.001$; μ_S: $t = 4.63$; $p < 0.001$; see Table 6.18).

At initial foot-ground contact, to avoid a slip it is imperative that the shear forces generated by older individuals do not exceed the frictional capabilities, that is, the available coefficient of friction of the shoe/surface interface (see Section 2.5.3). If the available coefficient of friction is exceeded, a slip or forward foot-slide would be the likely consequence (see Section 2.3.2(C)). When combined with the reduced capacity of older people to recover from a slip or foot-slide (see Section 2.3.1), falls and severe falls-related injuries, such as hip fractures, often result. Many floors, particularly smooth floor types such as vinyl tile, become dangerously slippery when they are wet due to a fluid film being trapped under the shoe, such that the shoe is partially supported by the fluid film, allowing the shoe to easily slide across the floor[453,907]. Therefore, in the present study, when subjects walked on the wet vinyl tile surface, they significantly reduced their peak F_{xB} as a strategy to increase the frictional force available to them during early stance (see Table 6.18). This finding is analogous with the subjective estimation results pertaining to shoe/surface slipperiness (see Section 6.6.1(C)) and altered muscle activation patterns (see Section 6.6.2(C)). It is postulated that these strategies were used to decrease their slip potential, particularly as the coefficient of friction calculated at initial foot-ground contact

depicts a value representing how easily the foot/shoe could slide on the wet surface.

Table 6.18: Descriptive information, pooled across subject group and footwear condition, for the ground reaction force and coefficient of friction variables when subjects (n = 16) walked across carpet, dry vinyl tile or wet vinyl tile.

Variable[a]	Carpet		Dry vinyl tile		Wet vinyl tile	
	Mean	**(SEM)**	**Mean**	**(SEM)**	**Mean**	**(SEM)**
Vertical Ground Reaction Forces						
Stance (ms)	720.4	(35.1)	717.8	(32.0)	739.6	(41.5)
F_{yB} ($N.kg^{-1}$)	10.5	(0.2)	10.2	(0.1)	10.2	(0.1)
IC-F_{yB} (%)	26.9	(1.0)	27.3	(1.0)	31.7	(2.7)
F_{yM} ($N.kg^{-1}$)	8.4	(0.3)	8.2	(0.3)	8.4	(0.3)
IC-F_{yM} (%)	47.7	(1.1)	48.9	(0.7)	48.9	(0.8)
Anteroposterior Ground Reaction Forces						
F_{xB} ($N.kg^{-1}$)	1.3	(0.1)	1.3	(0.1)	1.0	(0.1)
IC-F_{xB} (%)	17.7	(0.7)	19.3	(1.2)	20.0	(0.6)
IC-F_{xC} (%)	52.7	(1.6)	56.6	(2.1)	64.2	(8.9)
Coefficient of Friction						
μ_S	0.38	(0.06)	0.34	(0.04)	0.17	(0.04)
μ_D	0.16	(0.02)	0.17	(0.01)	0.15	(0.03)
IC-μ_D (%)	7.88	(0.02)	7.34	(0.27)	7.84	(0.25)

[a] See Table 5.4 for variable definitions.

(D) Interaction Effects

There were no significant subject group x footwear, subject group x surface or subject group x footwear x surface interactions pertaining to the vertical or anteroposterior ground reaction forces or the coefficient of friction data in the present study (see Table 6.17). However, the footwear and surface main effects were moderated by each other such that there were significant footwear x surface interactions for stance time as well as for peak F_{xB} (see Table 6.17).

Step down procedures pertaining to stance time indicated that when subjects walked across the carpet surface, they displayed a significantly shorter stance time when barefoot compared to when walking in closed back slippers

(t = -2.95; p = 0.012; see Figure 6.19). In addition, when walking across the dry vinyl tile surface, subjects recorded significantly shorter stance times when barefoot compared to when walking in closed back (t = -3.94; p = 0.002) or toe (t = -2.17; p = 0.048) slippers (see Figure 6.19). However, when walking across the wet vinyl tile surface, no significant within-footwear differences were evident with respect to stance time. That is, although walking barefoot on the carpet and dry vinyl tile surfaces revealed significantly shorter stance times compared to when shod, walking barefoot on the wet vinyl tile surface led to increased stance times, a common modification displayed by older people to ensure a more stable gait pattern (see Section 2.3.1(F)) and a reduced falls risk.

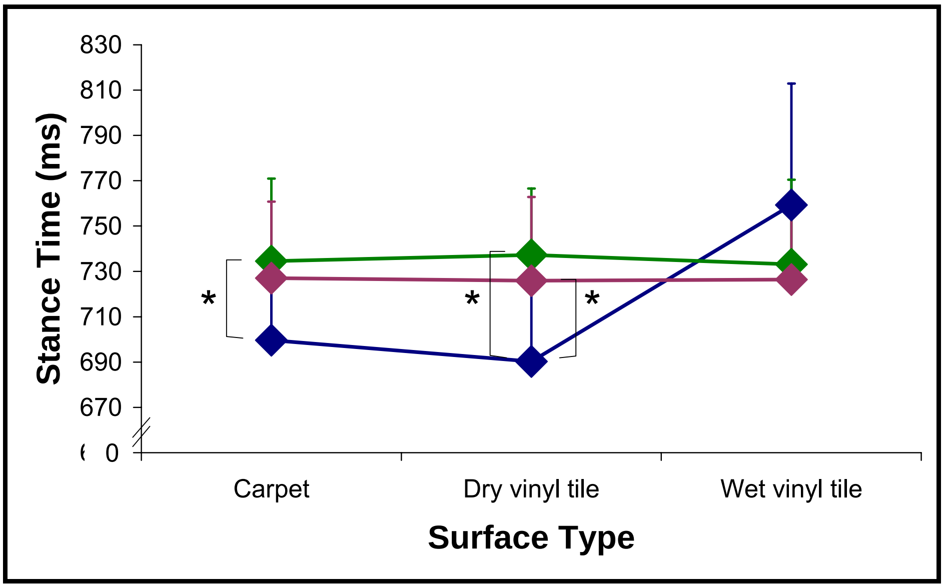

Figure 6.19: Footwear x surface interaction for stance time (ms) when subjects (n = 16) walked barefoot (—), in closed back slippers (—) and in toe slippers (—) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

When subjects walked barefoot on dry vinyl tile they recorded a significantly reduced peak F_{xB} at initial foot-ground contact compared to walking in closed back slippers (t = -2.47; p = 0.029; see Figure 6.20). Similarly, subjects also recorded significantly reduced F_{xB} when they walked barefoot on wet vinyl tile compared to walking in closed back (t = -5.27; p < 0.001) or toe (t = -5.81; p < 0.001) slippers (see Figure 6.20). Interestingly, the peak F_{xB} was

reduced under all footwear conditions when subjects walked across the wet vinyl tile surface; it was just increased when subjects walked barefoot. Therefore, it appeared that subjects were able to avoid a slip by altering their stance time as well as reducing the shear forces displayed at initial foot-ground contact to increase the available friction and reduce their slip potential, particularly when walking barefoot on wet vinyl tile.

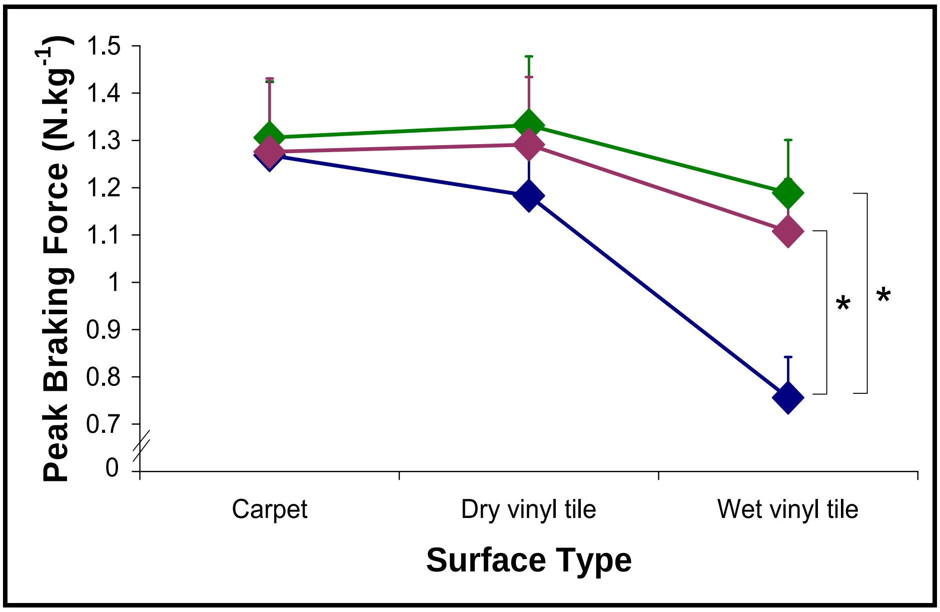

Figure 6.20: Footwear x surface interaction for the peak anteroposterior braking ground reaction force (F_{xB}; $N.kg^{-1}$) when subjects (n = 16) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions (* indicates statistical significance at $p \leq 0.05$).

To prevent a slip at initial foot-ground contact, the frictional demand must be less than the friction available between the supporting surface and the foot/shoe sole[91,502,505]. Section 2.3.2(C) details the coefficient of friction values for safe walking conditions and, although friction coefficients could be determined for each of the shoe-surface conditions assessed in the present study, using one of many described protocols and apparatus[88,411,785,908-916] (see Section 2.3.2(C)), they may also be inferred from previously published data[385,411,891,917-919]. However, provided they were tested under the same conditions (see Section 2.3.2(C)), the closed back slipper and the toe slipper should record the same

coefficient of friction values on each surface type, as the soles of the two slipper types were identical in material, structure and tread pattern (see Section 5.3.1).

While a good frictional interaction exists when walking barefoot on dry surfaces[420,920], this interaction diminishes when walking on smooth or wet floor surfaces, perhaps due to the reduced points of contact on the plantar surface (toes, ball and heel of the foot) when compared with the contact areas of a normal shoe[785,920]. Furthermore, in contrast to the tread pattern on the slipper sole, which allows water to escape to improve the frictional interaction[453,907], the plantar surface of the foot is usually smooth and untextured[453,907] with sensory thresholds capable of influencing skin friction and contributing to subjective sensations of skin smoothness, greasiness, and moistness[921]. Consequently, it is postulated that the lowest friction coefficient would have been recorded for the barefoot on wet vinyl tile condition compared to the other conditions assessed[785]. Therefore, although no slips occurred during the walking trials, the gait adaptations noted in the present study when subjects walked barefoot on the wet vinyl tile surface may have been required to maintain the coefficients of friction when walking under this condition. Consequently, these gait adaptations allowed subjects in the present study to successfully complete each trial without slipping.

6.6.4 Lower Limb Kinematics at Initial Foot-Ground Contact

It was originally intended to quantify the three-dimensional kinematic characteristics of each subject's dominant lower limb during initial foot-ground contact. However, preliminary analysis of the data revealed that the Position Sensor used to collect the kinematic data experienced a gradual degradation in the sensor detecting IRED location within the *z* plane such that it eventually failed (see Section 5.6). As the Position Sensor was positioned perpendicular to the direction of travel, was level in the horizontal plane and degradation occurred in the sensor detecting depth, it was postulated that, if three-dimensional coordinates were unable to be calculated, two-dimensional coordinates could be generated.

Each of the three charge couple devices located within the Position Sensor scans the *x*, *y* and *z* planes, locating each IRED within each plane. This planar data is then transformed, using commercially confidential algorithms developed by Northern Digital Inc. (Ontario, Canada), to calculate the three-dimensional coordinates of each IRED. Although it was originally postulated, and suggested by Northern Digital Inc. technical staff, that two-dimensional analyses could be performed using the full raw data format, or the planar data, from the operational sensors, visual inspection and preliminary analysis of the planar data indicated that the calculated IRED positions did not adequately describe lower limb motion during initial foot-ground contact. Consequently, the full raw data and camera files generated during testing were sent to Northern Digital Inc. technical staff so that alternate transformation algorithms could be written to analyse the data in two dimensions, namely to generate the *x* and *y* coordinates required for analysis without input from the degraded sensor.

Despite Northern Digital Inc. technical staff and programming mathematicians producing what was deemed to be an appropriate transformation algorithm, visual inspection of graphical representation of the output indicated the IRED positional data still did not realistically represent the position of the lower limb during initial foot-ground contact. For example, during the static trials when each subject stood stationary on the force platform, which was horizontal to the camera (see Section 5.6.2), the transformed data representing the force platform indicated the platform was lying at an angle to the floor, with this angle sometimes approximating 30°. Furthermore, calculation of IRED movement revealed that during these “stationary” trials, some markers were purportedly moving in excess of 40 mm, with some IREDs appearing to move below the force platform. There was also a large amount of missing data following the transformation. Unfortunately, licensing laws prevented Northern Digital Inc. from disclosing the algorithms used to transform the planar data from the Position Sensor into coordinate data. Therefore, solutions to the data problems were reliant upon the mathematicians at Northern Digital Inc., that is, mathematical advice from elsewhere could not be sought without the original transformation algorithms.

The error inherent in the data calculated for both RA and control subjects 3 to 8 meant that if these data were to be used in the kinematic analysis, different IRED combinations would have had to be used to calculate segmental and joint angles, velocities and accelerations, without any guarantee of accuracy. Consequently, the *x* and *y* coordinates derived from the three-dimensional data that had been calculated using the degraded sensor information, could not be used for analysis. Unfortunately, as there was a 15 month time delay between the original testing sessions and the recognition that no solution to the kinematic data problem could be found, retesting the same subjects was not feasible. Furthermore, due to the long testing sessions and the difficulty in finding volunteers willing and able to participate in the study, recruiting a further 12 subjects was not possible. Therefore, a full two-dimensional kinematic and kinetic analysis were restricted to two RA subjects and two control subjects, whose data were collected before the Position Sensor degraded. Descriptions of these data are presented in the relevant sections below and subjective descriptions of the involved subjects are included in Table 6.19.

Table 6.19: Descriptive characteristics of the RA (n = 2) and control (n = 2) subjects whose kinematic data were analysed.

Variable	RA Subjects		Control Subjects	
	Subject 1	Subject 2	Subject 1	Subject 2
Age (years)	77	60	63	67
Height (cm)	154.2	173.7	168.0	164.0
Mass (kg)	55.8	81.7	60.3	95.6
AIMS2[a]	3.33	0.33	0.00	0.25
FFI total[a]	31.8	26.3	1.9	2.6

[a] AIMS2 encompasses mobility, walking and bending, self-care and household sub-scales. FFI total encompasses pain, task difficulty and activity limitation sub-scales.

The two-dimensional data from these four subjects were considered appropriate for further analysis as the determination of foot, shank and thigh segment lengths and force platform dimensions were within ± 1 cm when the positional data derived from the static trials were compared to measurements made between IRED locations. This was not the case for the remaining 12 subjects. However, although visual inspection of the analysed two-dimensional

data appear similar to past studies[87,399,504], due to the error inherent in three-dimensional techniques and the unknown error in the calculation of the three-dimensional coordinates from the Position Sensor, three dimensional analyses were not completed to ensure results with the smallest amount of error could be produced.

(A) Between-Group Patterns

Descriptive variables pertaining to the kinematics of the lower limb at initial foot-ground contact for the two RA and two control subjects are displayed in Table 6.20. In the present study, the RA subjects displayed similar muscle control strategies (see Section 6.6.2(A)), ground reaction forces and coefficient of friction results (see Section 6.6.3(A)) to the control subjects. Therefore, as anticipated, there were only a few between-group kinematic differences (see Table 6.20) and all subjects recorded kinematic characteristics similar to those reported in the literature[296,318,323,324,338,497,507-509,514,922] (see Section 2.5.2 and Section 2.3.1(F)).

Table 6.20: Descriptive information, pooled across footwear and surface condition, for the lower limb kinematic variables displayed by the RA (n = 2) and control (n = 2) subjects at initial foot-ground contact.

Variable	RA Subjects		Control Subjects	
	Mean	(SD)	Mean	(SD)
Gait speed (m.s^{-1})[a]	0.96	(0.17)	1.12	(0.10)
Time of foot/shoe flat (%)	15.0	(3.1)	11.8	(2.0)
Lower Limb Alignment Variables				
Ankle angle (θ_a; °)[b]	-8.0	(1.9)	-3.9	(2.6)
Knee angle (θ_k; °)	176.1	(5.6)	173.1	(2.3)
Slip Potential Variables				
Horizontal heel slide (cm)	1.47	(0.52)	2.64	(0.75)
Foot/shoe angle (θ_f; °)	20.2	(2.7)	20.0	(2.8)
Heel velocity (m.s^{-1})	0.21	(0.06)	0.31	(0.09)
θ_f angular velocity (°.s^{-1})	144.4	(38.6)	194.8	(37.3)

[a] Horizontal velocity of the greater trochanter was used to represent gait speed.
[b] A negative number indicates ankle dorsiflexion.

With respect to the between-group differences, the control subjects recorded a faster average gait speed and consequently, a shorter time to foot/shoe flat than the RA subjects. In contrast, the RA subjects tended to contact the supporting surface with both greater knee extension and ankle dorsiflexion compared to the control subjects (see Table 6.20). Reduced gait speed has been associated with arthritis activity, muscle strength deficits and pain (see Section 2.3.1(F)). The RA subjects in the present study recorded reduced knee flexion and ankle dorsiflexion muscle strength (see Section 6.3.3), increased foot pain (see Section 6.2 and Section 6.6.1(A)) and recorded greater activity limitation due to foot pain and difficulty in performing daily living tasks (see Table 6.19) compared to the control subjects. However, the RA subjects also displayed greater knee flexion and ankle dorsiflexion at initial foot-ground contact combined with no between-group neuromuscular differences (see Section 6.6.2(A)) and consequently, they may be at a greater risk of fatigue-related falls compared to the control subjects in the present study.

The kinematics of the foot/shoe at initial foot-ground contact play an important role in the potential for slips and falls[410] (see Section 2.5.2). At terminal swing (see Figure 2.4), muscular action rapidly decelerates the lower limb before initial foot-ground contact (see Section 2.5.4 and Section 6.6.2(A)). Then, at impact, the foot/shoe slides along the supporting surface[411,505], at times following a variable pattern (see Section 2.5.2). After the foot/shoe ceases sliding, it rapidly rotates from a contacting foot/shoe angle near 20° down to the supporting surface, reaching a foot/shoe flat position at about 15% of stance time[85,411,923] (see Section 2.5.2). Slips during gait have been reported to result from high horizontal heel velocities and slow foot/shoe angular velocities at initial foot-ground contact together with large heel slide magnitudes after initial foot-ground contact[324,399,502,534] (see Section 2.5.2).

In the present study and consistent with the literature (see Section 2.5.2), the RA and control subjects displayed a foot/shoe angle of 20° at initial foot-ground contact with positive (forward) horizontal heel velocities and horizontal heel slide magnitudes (see Table 6.20). However, in comparison to the control subjects, RA subjects recorded 44% slower horizontal heel velocities, 32% less

horizontal heel slide and 26% slower foot/shoe angular velocities (see Table 6.20). Therefore, the lack of significant findings in the coefficient of friction data (see Section 6.6.3(A)) may have been influenced by the altered gait kinematics displayed by the RA subjects in an attempt to reduce their slip potential. Instead, when comparing the magnitudes of the slip potential variables displayed by both subjects groups in the present study to previous literature reporting the variables noted when subjects slipped at initial foot-ground contact, the control subjects may be at a greater risk of incurring a slip (see Table 6.20). However, as discrepancies exist in the data, for example, the RA subjects displayed reduced horizontal heel slide and horizontal heel velocity although a slow foot/shoe angular velocity (see Table 6.20); further research is warranted to determine which kinematic variable has a greater influence on slip potential in older women.

(B) Within-Footwear Patterns

Descriptive variables pertaining to the kinematics of the lower limb at initial foot-ground contact when subjects walked under different footwear conditions are displayed in Table 6.21. In the present study, there were no within-footwear outcomes with respect to gait speed although subjects achieved foot/shoe flat earliest when walking barefoot and latest when walking in the closed back slippers (see Table 6.21). An earlier time of foot/shoe flat may explain the altered loading forces sustained when barefoot compared to when shod (see Section 6.6.3(B)). Furthermore, although the subjects displayed similar ankle and knee joint angles when they walked barefoot or in closed back slippers, when subjects walked wearing the toe slippers, they displayed greater ankle dorsiflexion and increased knee flexion at initial foot-ground contact compared to the other footwear conditions (see Table 6.21). Interestingly, this altered lower limb alignment coincided with earlier activation of VL and TA together with increased TA muscle activity (see Section 6.6.2(B)). Therefore, the increased ankle dorsiflexion combined with TA activation may have been required to assist in keeping the shoe on the foot (see Section 6.6.1(B)). Consequently, the increased knee flexion and altered VL muscle activation

evident when subjects walked wearing the toe slippers may have been required to compensate for the increased ankle dorsiflexion to maintain the same foot/shoe angle at initial foot-ground contact[538] (see Table 6.21).

With respect to the slip potential variables, subjects in the present study recorded similar foot/shoe angles and horizontal heel velocities at initial foot-ground contact regardless of footwear condition (see Table 6.21). In contrast, subjects recorded the greatest horizontal heel slide when they walked in the toe slippers and the greatest foot/shoe angular velocity when they walked wearing closed back slippers compared to the other footwear conditions (see Table 6.21). The increased horizontal heel slide may reflect the foot sliding inside the toe slipper (see Section 6.6.1(B)), rather than the shoe sliding on the supporting surface and, as no IRED was placed on the sole of the toe slipper, further research is required to support or deny this claim. However, it was interesting to note that when subjects walked barefoot or in toe slippers, two footwear conditions considered to be slippery (see Section 6.6.1(B)), they displayed angular foot/shoe angular velocities which have been reported previously in walking trials that result in slips[324,399,502,534] (see Section 2.5.2).

Table 6.21: Descriptive information, pooled across subject group and surface condition, for the lower limb kinematic variables at initial foot-ground contact when subjects (n = 4) walked barefoot, in closed back slippers or in toe slippers.

Variable	Barefoot		Closed slipper		Toe slipper	
	Mean	**(SD)**	**Mean**	**(SD)**	**Mean**	**(SD)**
Gait speed ($m.s^{-1}$)[a]	1.00	(0.18)	1.10	(0.17)	1.01	(0.12)
Time of foot/shoe flat (%)	12.2	(3.3)	14.5	(1.0)	13.5	(3.7)
Lower Limb Alignment Variables						
Ankle angle (θ_a; °)[b]	-5.2	(3.1)	-5.8	(3.3)	-7.1	(2.7)
Knee angle (θ_k; °)	175.4	(5.1)	175.7	(4.4)	172.7	(3.3)
Slip Potential Variables						
Horizontal heel slide (cm)	1.73	(0.84)	1.98	(0.46)	2.47	(1.09)
Foot/shoe angle (θ_f; °)	19.7	(3.0)	20.5	(3.1)	20.0	(2.0)
Heel velocity ($m.s^{-1}$)	0.25	(0.09)	0.24	(0.06)	0.30	(0.11)
θ_f angular velocity ($°.s^{-1}$)	164.2	(59.1)	186.1	(39.7)	158.2	(31.2)

[a] Horizontal velocity of the greater trochanter was used to represent gait speed.
[b] A negative number indicates ankle dorsiflexion.

Aside from the average magnitude data, high between-subject variability, as evidenced by the high standard deviations, was evident in the data for the four subjects, particularly with respect to the calculated foot/shoe angular velocities when subjects walked barefoot (see Table 6.21). High variability in kinematic data obtained during gait studies has been reported previously[399,503,898,906,924]. For example, research conducted by Kurz & Stergiou[906,924] revealed that when eight healthy males (mean age, 28 years) ran barefoot, they recorded greater ankle joint variability during the running stance period compared to when they ran in shoes. Furthermore, this variability was only found to occur at the ankle, suggesting that ankle motion is required to compensate for changes in footwear. The authors suggested that this variability may be related to the ability of the heel and forefoot mechanoreceptors to sense the magnitude of impact during stance[868,924]. That is, a varied joint pattern may assist in dispersing the joint forces across various tissues, reducing repetitive impact to the same structures and preventing over-use injuries[531,618,925-928]. As footwear design affects sensory information (see Table 2.2), footwear may compromise the capacity of the foot to convey information about foot position[929], leading to both reduced kinematic variability[868,896] and increased potential for injury. Interestingly, Waddington & Adams[930] found that when conventional smooth insoles inside football boots were replaced with textured insoles, movement discrimination was restored to barefoot levels in elite female soccer players, suggesting shoes can be redesigned to allow adequate sensory feedback for accurate foot/shoe positioning.

Footwear, however, has also been reported to enhance performance in tasks performed by older people involving locomotion and dynamic foot-ground interaction[931]. For example, research conducted by Waddington & Adams[899] found that when they assessed 20 community-dwelling subjects aged 65 to 85 years, active movement at the ankle was significantly better discriminated when they were wearing shoes compared to when they were tested barefoot. This confirmed an earlier study by Robbins *et al.*[454] who found that the ability of subjects to balance when barefoot declined in advancing years, such that older subjects had significantly fewer imbalances during beam walking when wearing footwear than when barefoot. Whether age-related changes in the feet, the

incidence of arthritis or other age-related decline (see Section 2.3.1) make standing more comfortable in shoes or because older people are accustomed to wearing shoes to the extent that the barefoot state is unfamiliar is unknown. Therefore, further research is warranted to investigate how footwear design and its impact upon sensory feedback affect the kinematics of the lower limb with respect to slip potential and falls reduction in older people.

(C) Within-Surface Patterns

Descriptive variables pertaining to the kinematics of the lower limb at initial foot-ground contact when subjects walked across different surface conditions are displayed in Table 6.22. Consistent with increased slipperiness and task difficulty perceptions (see Section 6.6.1(C)), altered muscle activation patterns (see Section 6.6.2(C)) and reduced peak F_{xB} (see Section 6.6.3(C)), subjects in the present study displayed the slowest gait speed when walking on the wet linoleum surface and the fastest gait speed when walking across the carpet surface (see Table 6.22). Willmott[384] found that hospital patients displayed a significantly slower gait speed when walking on a reflective vinyl tile surface compared to walking on a carpet surface, concluding that subjects displayed a more efficient and confident gait pattern when they walked on the carpet surface (see Section 3.2.3(B)). In contrast, Dickinson[386] reported older adults to walk significantly slower across carpet compared to vinyl tile, suggesting the subjects were more hesitant when they encountered a more compliant surface. In the present study, the carpet surface did not have an underlay and was a low pile carpet (pile height = 10 mm; see Section 5.3.1(B)). Therefore, subjects may not have sensed much difference in the compliance of the carpet or vinyl tile surfaces. Furthermore, no within-surface differences were noted for time to foot/shoe flat, ankle dorsiflexion or knee flexion at initial foot-ground contact (see Table 6.22). Therefore, the changes in gait speed in the present study appear indicative of confident and efficient gait when walking across a surface not deemed to be slippery (see Section 6.6.1(C)).

There were no within-surface differences in the slip potential variables of horizontal heel slide or horizontal heel velocity (see Table 6.22). However,

when subjects walked across the wet vinyl tile surface, they recorded a decreased foot/shoe angle and reduced foot/shoe angular velocity compared to walking on the carpet or dry vinyl tile surfaces (see Table 6.22). When comparing kinematic characteristics of steps prior to and onto a known slippery area, Andres and colleagues[542] reported similar gait adaptations to those reported in the present study, such as reduced foot angle and foot angular velocity at initial foot-ground contact. Furthermore, Cham and Redfern[505] reported that subjects who recovered from slip events when walking across an oily surface, walked with smaller foot contact angles and angular velocities than the values recorded on dry conditions. Interestingly, these data were in general similar to those observed during falling trials[505]. Therefore, although purported to be mechanisms leading to slips, reductions in foot/shoe angle and angular velocity at initial foot-ground contact may also be kinematic adaptations made to reduce the shear forces at foot-ground impact and, consequently, the likelihood of slipping[399,406,503].

Table 6.22: Descriptive information, pooled across subject group and footwear condition, for the lower limb kinematic variables at initial foot-ground contact when subjects (n = 4) walked across carpet, dry vinyl tile or wet vinyl tile.

Variable	Carpet		Dry vinyl tile		Wet vinyl tile	
	Mean	**(SD)**	**Mean**	**(SD)**	**Mean**	**(SD)**
Gait speed ($m.s^{-1}$)[a]	1.11	(0.15)	1.03	(0.14)	0.97	(0.17)
Time of foot/shoe flat (%)	13.8	(2.5)	13.2	(3.3)	13.3	(3.4)
Lower Limb Alignment Variables						
Ankle angle (θ_a; °)[b]	-5.6	(3.1)	-6.1	(3.0)	-6.3	(3.2)
Knee angle (θ_k; °)	175.5	(4.0)	175.2	(4.3)	173.0	(4.9)
Slip Potential Variables						
Horizontal heel slide (cm)	1.93	(0.77)	2.20	(1.02)	2.06	(0.86)
Foot/shoe angle (θ_f; °)	21.4	(2.4)	19.9	(2.6)	18.9	(2.8)
Heel velocity ($m.s^{-1}$)	0.24	(0.10)	0.27	(0.09)	0.27	(0.09)
θ_f angular velocity ($°.s^{-1}$)	183.7	(35.8)	183.6	(37.7)	141.1	(49.9)

[a] Horizontal velocity of the greater trochanter was used to represent gait speed.
[b] A negative number indicates ankle dorsiflexion.

(D) Interaction Patterns

No statistical analyses were completed on the kinematic data to reveal statistical interactions due to the low subject number in each group. However, when the data were pooled across surface type, the subject group x footwear data for knee joint angle and horizontal heel slide displayed similar patterns to the earlier results reported for VL burst duration (see Section 6.6.2(D)). That is, whereas the RA subjects displayed consistently greater knee extension at initial foot-ground contact when walking barefoot or in closed back slippers compared to the control subjects, when walking in toe slippers the RA subjects displayed a comparable knee joint angle to the control subjects (see Figure 6.21). Furthermore, although the RA and control subjects displayed similar magnitudes of horizontal heel slide when walking in the closed back slippers, when walking barefoot or in toe slippers the control subjects recorded notable increases in horizontal heel slide compared to the RA subjects who recorded reduced horizontal heel slide values (see Figure 6.22).

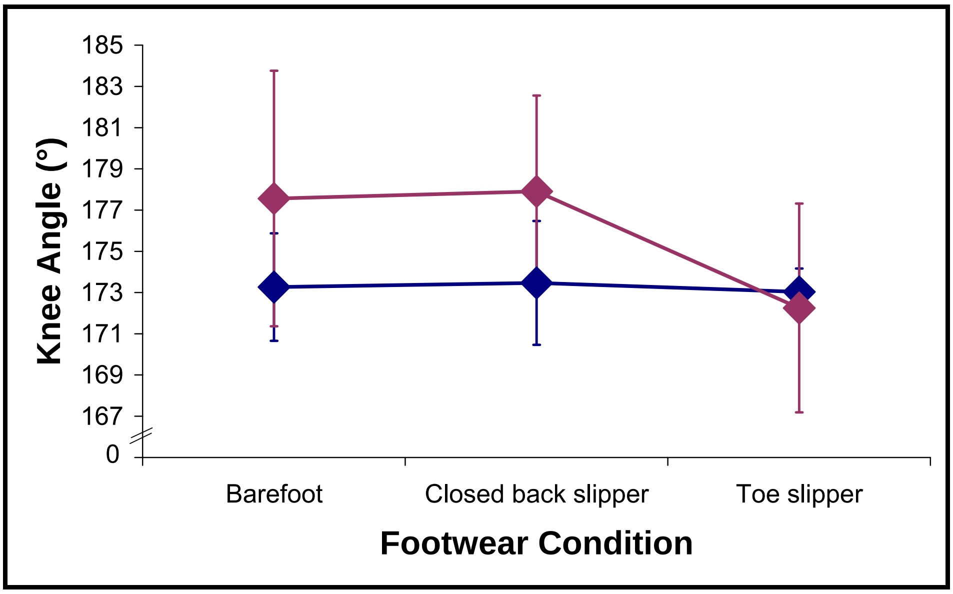

Figure 6.21: Subject group x footwear interaction for knee joint angle at initial foot-ground contact for the control (n= 2; ▬) and RA (n= 2; ▬) subjects under different footwear conditions.

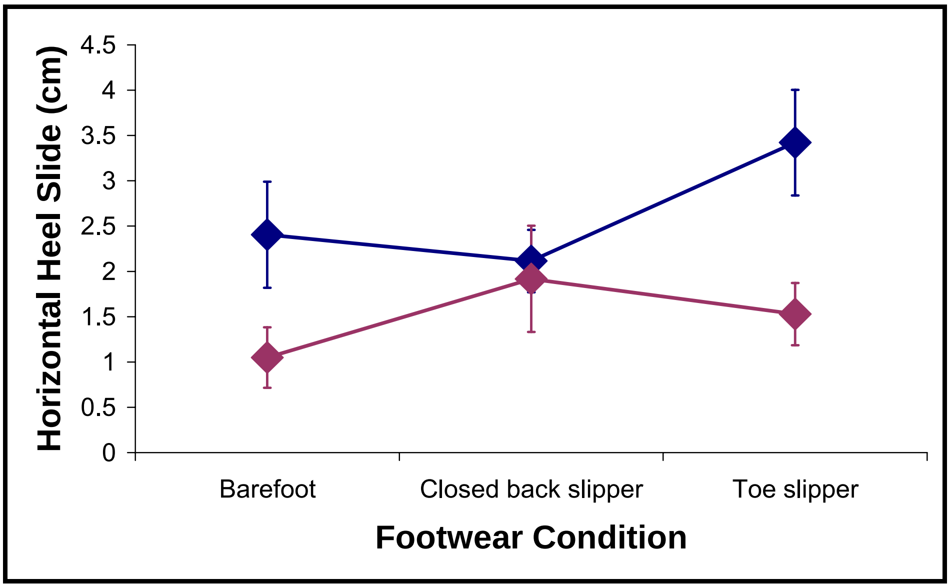

Figure 6.22: Subject group x footwear interaction for the horizontal heel slide that occurs at initial foot-ground contact for the control (n= 2; ▬) and RA (n= 2; ▬) subjects under different footwear conditions.

Decreased knee flexion during stance has been previously reported for patients with knee osteoarthritis as they attempt to avoid large eccentric quadriceps contractions to reduce the compressive knee joint forces and consequent knee pain[306,479]. Consistent with the results of the present study, similar gait mechanisms have also been reported for RA patients who suffer lower limb pain (see Table 6.19 and Section 2.6.3). However, when walking in toe slippers, the greater knee flexion required by the RA subjects required an altered quadriceps contraction, as evidenced by the extended VL burst duration (see Figure 6.12 and Section 6.6.2(D)) and may have contributed to greater knee pain. Although knee pain was not evaluated in the present study, a strong subject group x footwear type trend was evident for perceived foot pain (see Section 6.6.1(D)) whereby the RA subjects recorded the greatest foot pain when walking in toe slippers. Therefore, further research is warranted to determine whether this increased need for knee flexion when walking in the toe slippers would place the RA subjects at a greater risk of sustaining either a fatigue- or pain-related fall, due to their decreased quadriceps muscle strength (see Section 6.3.3) or altered movement patterns that may affect lower limb stability.

Data pertaining to horizontal heel slide values indicated that the control subjects had much greater foot movement inside the toe slipper (see Section 6.6.1(B) and Section 6.6.4(B)) or much greater heel slide when wearing the toe slipper compared to the RA subjects. Interestingly, when walking under footwear conditions considered to be slippery or difficult (see Section 6.6.1(B)), whereas the control subjects displayed increased horizontal heel slide, the RA subjects decreased their horizontal heel slide. Therefore, the RA subjects appeared to alter their movement pattern to decrease their slip risk. In fact, when walking wearing the toe slippers, control subjects recorded a mean horizontal heel slide value above the 3 cm threshold reported by Leamon & Li[506], which may place control subjects at a greater risk of slipping (see Figure 6.22 and Section 2.5.2). As previously reported, further research pertaining to foot movement inside the toe slipper is required to determine how these footwear types affect slip risk.

When footwear and surface condition were pooled across subject group, an interesting footwear x surface pattern was noted for foot/shoe angular velocity. That is, when subjects walked on the carpet or dry vinyl tile surfaces, they displayed similar mean foot/shoe angular velocities regardless of footwear condition (see Figure 6.23). However, when subjects walked across the wet vinyl tile surface, they displayed slower mean foot/shoe angular velocities in all footwear conditions, although this decrease was particularly evident when subjects walked barefoot (see Figure 6.23).

This change in movement pattern specifically under the barefoot + wet vinyl tile condition is consistent with increased perceptions of task difficulty and shoe/surface slipperiness (see Section 6.6.1(D)), altered muscle activation patterns (see Section 6.6.2(D)) and altered ground reaction force variables (see Section 6.6.3(D)). Therefore, the plantar cutaneous sensory information appears to play an important role in movement regulation to prevent slips. Consequently, further research is warranted to determine whether this sensory feedback reduces the slip risk as well as how this sensory feedback response is affected and/or attenuated by habitual shoe wear[932].

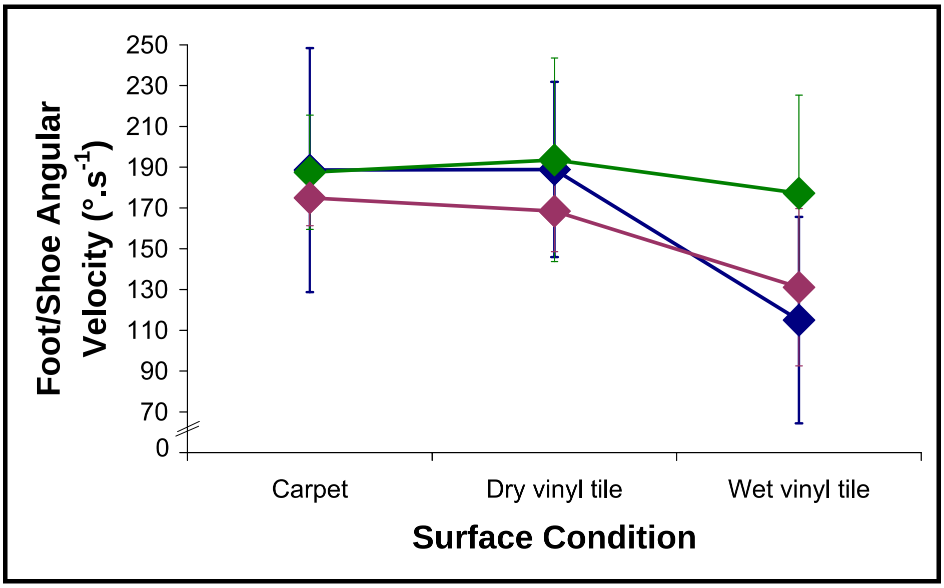

Figure 6.23: Footwear x surface interaction for the foot/shoe angular velocity recorded at initial foot-ground contact when subjects (n = 4) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions.

6.6.5 Lower Limb Joint Moments and Powers at Initial Foot-Ground Contact

(A) Between-Group Patterns

Descriptive data pertaining to the joint moments and powers evident at initial foot-ground contact for the two RA and two control subjects are presented in Table 6.23. Consistent with the muscle activation (see Section 6.6.2(A)), ground reaction force (see Section 6.6.3(A)) and kinematic (see Section 6.6.4(A)) data, there were no between-group differences in the ankle or knee joint moments at initial foot-ground contact. However, the RA subjects displayed larger mean ankle and knee joint powers compared to the control subjects (see Table 6.23).

During "normal" gait, at initial foot-ground contact, the muscles cocontract to control placement of the lower limb to the floor and absorb the energy of impact[327,497,502,933]. If all muscles contract somewhat equally, then, as was seen in the present study, both the ankle and knee joint moments approach zero[509] (see Table 6.23). However, if this co-contraction occurs with large

activation of the surrounding muscle groups, then, although the strain and shear forces at the joint may be reduced, the compressive forces and thereby joint loading will increase[934,935], perhaps contributing to greater lower limb joint pain. Furthermore, increased intensity of muscle co-contraction could accelerate muscle fatigue, leading to fatigue-related falls. However, the muscle burst intensities recorded in the present study did not display increased activity compared to previous reports (see Section 6.6.2).

Table 6.23: Descriptive information, pooled across footwear and surface condition, for the lower limb joint moment and power variables displayed by the RA (n = 2) and control (n = 2) subjects at initial foot-ground contact.

Variable	RA Subjects		Control Subjects	
	Mean	(SD)	Mean	(SD)
Lower Limb Joint Moments[a]				
Ankle ($N.m.kg^{-1}$)	0.009	(0.026)	0.001	(0.004)
Knee ($N.m.kg^{-1}$)	-0.002	(0.010)	-0.002	(0.003)
Lower Limb Joint Powers[b]				
Ankle ($W.kg^{-1}$)	0.041	(0.180)	0.007	(0.065)
Knee ($W.kg^{-1}$)	0.047	(0.407)	0.029	(0.116)

[a] Positive net joint moment = ankle dorsiflexion and knee extension; Negative net joint moment = ankle plantar flexion and knee flexion.
[b] Positive power = energy generation, Negative power = energy absorption.

Immediately following initial foot-ground contact, there is typically a small ankle dorsiflexion moment as the ankle dorsiflexor muscles eccentrically control the lowering of the foot (ankle plantar flexion) to the floor[497,871,936,937] (see Section 2.5.4). However, although the power curve at this time shows energy absorption, the power associated with the lowering of the foot is small and is thereby considered insignificant in the ankle power curve[345]. Interestingly, both subject groups displayed positive ankle joint power, indicative of concentric ankle dorsiflexor foot control at initial foot-ground contact, with the RA subjects recording increased ankle joint power compared to the control subjects (see Table 6.23). The slight difference in ankle joint power between the RA and control subjects may reflect an apropulsive gait pattern, typically reported in RA subjects, whereby the loads on the lower limbs are

redistributed so that weight-bearing is avoided on painful joints, such as the metatarsal heads (see Section 2.6.3).

At the knee, a small knee flexor moment may occur at initial foot-ground contact while the hamstring muscles are activated to absorb most of the energy from the swinging leg and foot and to decelerate the lower limb for safe foot placement (see Section 2.5.4). However, this knee flexor moment is rapidly (within approximately 4% of the gait cycle) converted to a knee extensor moment, whereby the quadriceps eccentrically contribute to balance control and weight acceptance as well as prevent excessive vertical translation of the body centre of gravity by controlling the amount of knee flexion[345,348,497,502,933] (see Section 2.5.4). During this time there is an important major energy absorption phase in the knee power curve, which has been deemed important for stability during stance[348] (see Section 2.5.4). In the present study, at initial foot-ground contact, the knee moment was zero, although very slightly negative, indicating a knee flexor moment. This knee flexor moment was combined with positive knee joint power, indicating a concentric contraction of the knee flexors to slow the forward horizontal velocity of the lower limb to ensure safe lower limb placement at initial foot-ground contact. Furthermore, similar to the ankle power magnitude, the knee power in the present study was higher in the RA subjects compared to the control subjects (see Table 6.23), perhaps indicative that the RA subjects required greater lower limb control at initial foot-ground contact.

Few differences were seen in the results of the ankle and knee joint moment and power magnitudes in the sagittal plane in the present study. More differences may have occurred in the sagittal plane ankle and knee joint moments and powers late in stance as has been discussed previously for RA patients (see Section 2.6.3). However, it is not perceived that any sagittal plane ankle and knee joint moment and power differences evident in late stance would affect slip risk at initial foot-ground contact and, therefore, further discussion is not within the scope of the present thesis. Furthermore, data calculated in the frontal and transverse planes may have revealed differences between the two subjects groups due to the typical lower limb problems seen in RA patients (see

Section 2.6.1 and Section 2.6.3). However, although only calculated for two RA and two control subjects, large variation was seen in the ankle and knee joint power magnitudes, particularly the knee joint power for the RA subjects, as evidenced by the high standard deviations (see Table 6.23). This variability, also reported in other data in the present study (see Section 6.4.3 and Section 6.6.2), may be related to an adaptation made by the subjects, particularly the RA subjects towards safer and more stable locomotion[318,876].

(B) Within-Footwear Patterns

Descriptive variables pertaining to the joint moments and powers at initial foot-ground contact when subjects walked barefoot or in either slipper type are displayed in Table 6.24. In the present study, although all approached zero, the largest mean ankle joint moment and power magnitudes were calculated when subjects walked in the toe slippers compared to walking barefoot or in the closed back slippers (see Table 6.24). Therefore, concentric ankle dorsiflexor activation again suggested that increased muscular activation was required to help keep the toe slipper on the foot for safe foot-ground contact (see Section 6.6.2(B)). Interestingly, the ankle power magnitude recorded at initial foot-ground contact when subjects walked barefoot was negative (see Table 6.24), indicating the ankle dorsiflexor muscles were eccentrically contracting to absorb impact. It is postulated that the absorption of energy occurs earlier when barefoot as the forces imposed on the body must be attenuated by the body's anatomical structures compared to when shod whereby some shock absorption is provided by the shoe sole (see Section 2.5.3).

With respect to the knee, there was no effect of footwear on the net joint moments at initial foot-ground contact, which were zero, suggesting co-contraction between the knee flexor and extensor muscle groups (see Table 6.24). However, compared to the other footwear conditions, the mean knee joint power magnitudes were largest when subjects walked barefoot (see Table 6.24), indicating concentric contraction of the knee flexors to ensure safe lower limb placement at initial foot-ground contact. Furthermore, when subjects walked in the closed back slippers, they displayed a mean negative knee joint power

magnitude (see Table 6.24). It is postulated that subjects felt confident walking in the closed back slippers (see Section 6.6.1(B)) and, consequently, did not require as much control during initial foot-ground contact. That is, when subjects walked barefoot or in the toe slippers, they still appeared to be preparing their lower limb for contact whereas when they walked in the closed back slippers they started to absorb energy at impact.

Table 6.24: Descriptive information, pooled across subject group and surface condition, for the lower limb joint moment and power variables at initial foot-ground contact when subjects (n = 4) walked barefoot, in closed back slippers or in toe slippers.

Variable	Barefoot		Closed slipper		Toe slipper	
	Mean	**(SD)**	**Mean**	**(SD)**	**Mean**	**(SD)**
Lower Limb Joint Moments[a]						
Ankle ($N.m.kg^{-1}$)	0.001	(0.004)	0.002	(0.007)	0.015	(0.036)
Knee ($N.m.kg^{-1}$)	-0.006	(0.012)	0.000	(0.005)	0.000	(0.004)
Lower Limb Joint Powers[b]						
Ankle ($W.kg^{-1}$)	-0.020	(0.056)	0.038	(0.109)	0.062	(0.229)
Knee ($W.kg^{-1}$)	0.144	(0.475)	-0.044	(0.295)	0.047	(0.056)

[a] Positive net joint moment = ankle dorsiflexion and knee extension; Negative net joint moment = ankle plantar flexion and knee flexion.
[b] Positive power = energy generation, Negative power = energy absorption.

Few studies have reported the effects of footwear on the sagittal ankle and knee joint moments and powers at initial foot-ground contact. Instead, studies report on the major phases of both power curves and how they vary following orthotic or footwear interventions during walking or running gait as well as other movement types in various subject populations[938-947]. However, one study by Kerrigan *et al*.[944,945] reported that 20 healthy women displayed significantly altered peak ankle and knee moments in both the sagittal and coronal planes when they walked wearing high heels compared to walking barefoot. In addition, although different in the coronal plane, at initial foot-ground contact, the sagittal plane ankle and knee joint moments were very similar between conditions. Therefore, the moments and powers that occur in the frontal and transverse planes are of importance when assessing these interventions[938,939], particularly in individuals with lower limb pathology[938-947]

and it was unfortunate that they could not be calculated in the present study (see Section 6.6.4). Similarly, Johnson *et al*.[947] revealed no major differences in the sagittal plane ankle moments at initial foot-ground contact in 164 subjects (aged 5 to 85 years) regardless of orthotic intervention. However, the knee joint moment was dependent upon orthotic intervention, such that one of the peaks occurred at initial foot-ground contact[947]. Interestingly, and in accordance with the findings of the present study, Johnson *et al*.[947] reported high variation in the knee joint moments (see Table 6.24). This variability may have concealed any within-footwear differences in the ankle and knee joint moments and powers in the present study. Therefore, as the data from only two RA and two control subjects was analysed, further investigation is warranted to determine the effects of footwear on these moments and powers at initial foot-ground contact.

(C) Within-Surface Patterns

Descriptive variables pertaining to the joint moments and powers at initial foot-ground contact when subjects walked across each surface condition are displayed in Table 6.25. In the present study there was no effect of surface condition on the mean ankle or knee joint moments (see Table 6.25). Therefore, thoughts that hamstring and quadriceps co-contraction would reduce the knee moments more so when subjects walked on the wet vinyl tile surface compared to the other surface types, allowing the individual to adapt to the more "dangerous" situation (see Section 2.5.4), was not evident at initial foot-ground contact. However, similar to the between-group and within-footwear data, there was some variability in the mean ankle and knee joint powers within the three surface conditions as evidenced by the high standard deviations (see Table 6.25). For example, when subjects walked on the dry vinyl they displayed opposite power profiles at initial foot-ground contact compared to walking across carpet or wet vinyl tile (see Table 6.25). That is, when subjects walked across the dry vinyl tile surface that they perceived as the easiest and most comfortable to walk across (see Section 6.6.1(C)), they displayed positive ankle joint power and negative knee joint power. However, further research is required to determine whether this is a consistent response.

Very little research has been conducted on the effects of surface type on the moments and powers of the knee and ankle joints, particularly at initial foot-ground contact. Cham & Redfern[503] reported a significant surface effect on joint moment variables, with significantly higher moments displayed by subjects walking on rough floors compared to vinyl and smooth floor types. In addition, both ankle and knee moments have been reported to decrease when individuals encounter and/or anticipate slippery surfaces, with additional reductions seen in the incidence of an actual fall event[238,503]. Therefore, as the lower limb joint moments are a reflection of overall muscle reactions[503], these findings suggest that the knee and hip appear to be used more than the ankle to control slip potential[503]. Similar results were noted in the present study, as greater changes were made with respect to quadriceps and hamstring muscle activation patterns when subjects altered the surface that they walked upon compared to the muscles that surround and control ankle movement (see Section 6.6.2(C)).

Table 6.25: Descriptive information, pooled across subject group and footwear condition, for the lower limb joint moment and power variables at initial foot-ground contact when subjects (n = 4) walked across carpet, dry vinyl tile or wet vinyl tile.

Variable	**Carpet**		**Dry vinyl tile**		**Wet vinyl tile**	
	Mean	**(SD)**	**Mean**	**(SD)**	**Mean**	**(SD)**
Lower Limb Joint Moments[a]						
Ankle ($N.m.kg^{-1}$)	-0.001	(0.003)	0.011	(0.032)	0.007	(0.015)
Knee ($N.m.kg^{-1}$)	0.001	(0.004)	-0.004	(0.013)	-0.002	(0.003)
Lower Limb Joint Powers[b]						
Ankle ($W.kg^{-1}$)	-0.006	(0.086)	0.111	(0.195)	-0.027	(0.101)
Knee ($W.kg^{-1}$)	0.040	(0.189)	-0.014	(0.206)	0.094	(0.495)

[a] Positive net joint moment = ankle dorsiflexion and knee extension; Negative net joint moment = ankle plantar flexion and knee flexion.
[b] Positive power = energy generation, Negative power = energy absorption.

(D) Interaction Patterns

No statistical analyses were completed on the net joint moment or power data to reveal statistical interactions due to the small subject number. However, when the data were pooled across surface type, footwear type or subject group,

there were very few consistent group x footwear, group x surface or footwear x surface patterns, respectively, noted in the interaction graphs. It is postulated that this was due to the high variation recorded among the data combined with the low subject number.

When the data were pooled across surface condition, the subject group x footwear data for ankle and knee joint moment and power magnitude (see Figure 6.24), indicated that the RA subjects required greater ankle joint moment and power compared to the control subjects. As the RA subjects recorded lower ankle dorsiflexion muscle strength relative to their control counterparts (see Section 6.3.3), these muscles may have had reduced force producing capability, requiring increased intensity of contraction (see Section 6.6.2(B)) when the RA subjects walked wearing the toe slippers. However, the toe slipper condition also recorded the highest variation within the ankle data, suggesting a variety of strategies were required by the subjects to keep the toe slippers on their feet as well as to control foot placement.

When the data were pooled across footwear condition, the subject group x surface data revealed similar patterns for the knee joint moment and power magnitudes (see Figure 6.25). However, when walking on carpet, the RA subjects displayed an ankle plantar flexion moment at initial foot-ground contact compared to the control subjects who displayed a zero ankle moment. This ankle plantar flexion moment may reflect that the two RA subjects prematurely activated G, perhaps together with delaying TA onset, compared to the two control subjects[300]. The RA subjects also appeared to require greater ankle joint power when they walked on the dry vinyl tile surface compared to the control subjects. However, the greatest variation around the mean was also evident on this surface for the RA subjects, indicating inconsistent walking strategies at initial foot-ground contact, possibly due to footwear.

When footwear and surface condition were pooled across subject group, there were interesting footwear x surface patterns noted for the ankle joint moment and power magnitudes (see Figure 6.26). That is, when subjects walked in the toe slippers on the dry vinyl tile surface they required a greater ankle dorsiflexion moment combined with greater energy generation, again perhaps

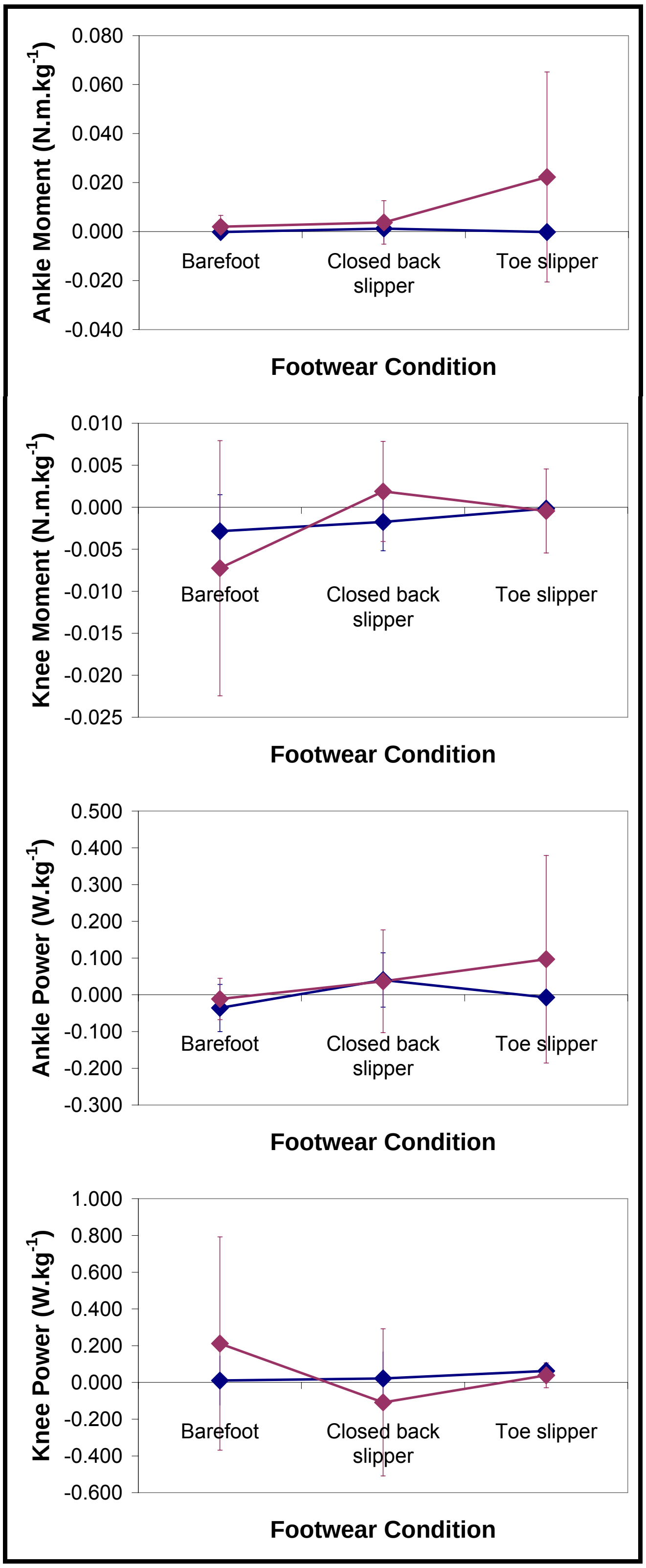

Figure 6.24: Subject group x footwear interactions for the ankle and knee joint moments ($N.m.kg^{-1}$) and powers ($W.kg^{-1}$) calculated at initial foot-ground for the control (n= 2; ▬) and RA (n= 2; ▬) subjects under different footwear conditions.

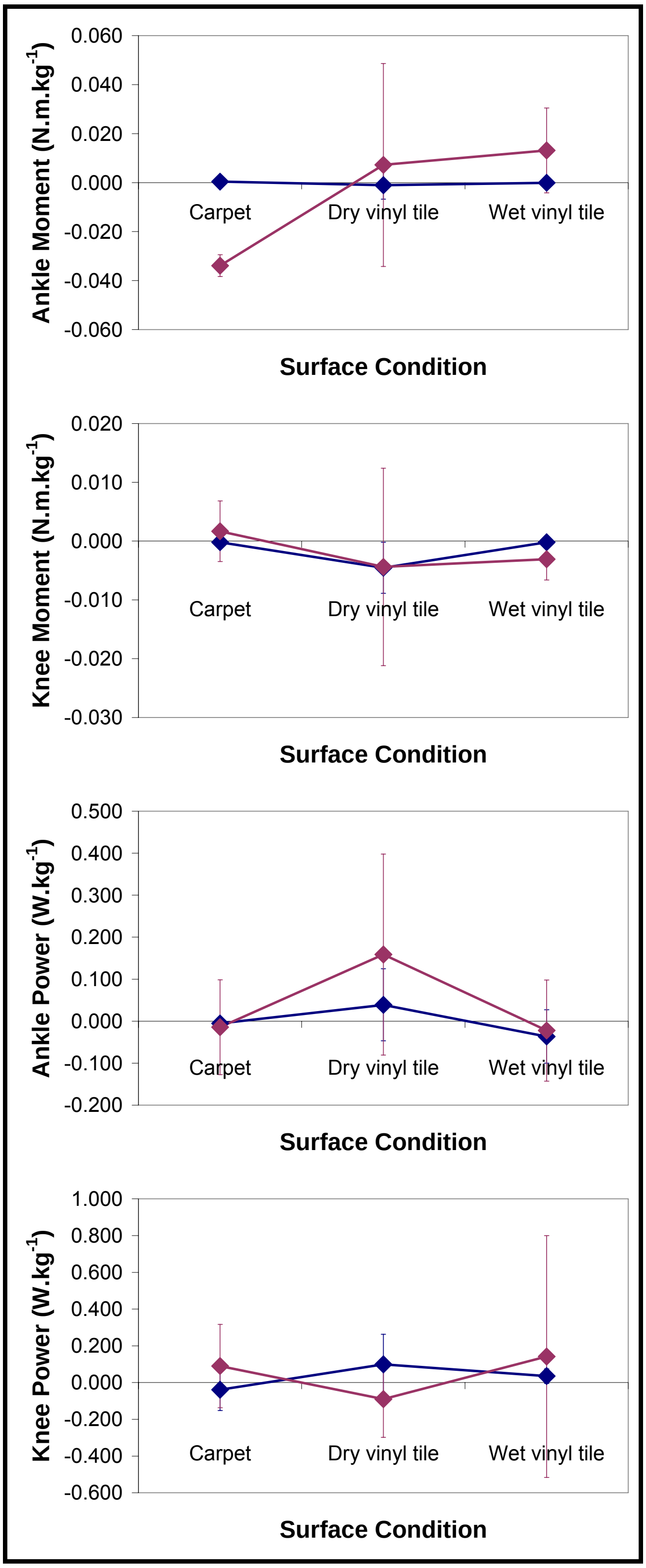

Figure 6.25: Subject group x surface interactions for the ankle and knee joint moments ($N.m.kg^{-1}$) and powers ($W.kg^{-1}$) calculated at initial foot-ground contact for the control (n= 2; ▬) and RA (n= 2; ▬) subjects under different surface conditions.

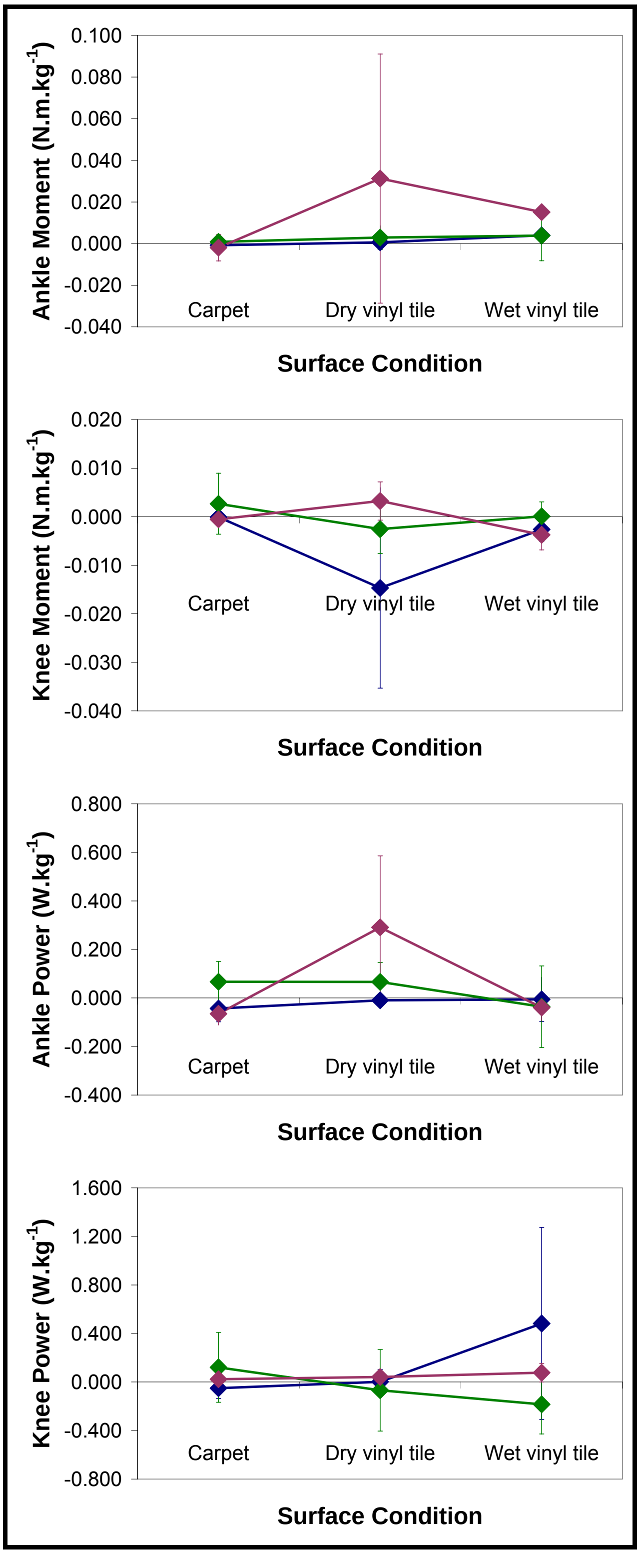

Figure 6.26: Footwear x surface interactions for the ankle and knee joint moments ($N.m.kg^{-1}$) and powers ($W.kg^{-1}$) calculated at initial foot-ground contact when subjects (n = 4) walked barefoot (▬), in closed back slippers (▬) and in toe slippers (▬) under different surface conditions.

reflecting the difficulty of keeping the toe slippers on the feet (see Section 6.6.1(B)). However, there were no other comparable interactions found in the present study (see Section 6.6.2(D), Section 6.6.3(D) and Section 6.6.5(D)) and therefore this result warrants further investigation. In fact, the only footwear x surface pattern that was similar to earlier interactions was noted for knee joint power, whereby subjects required greater energy generation when they walked barefoot on the wet vinyl tile surface compared to the other footwear-surface interactions (see Figure 6.26). Therefore, it would appear that when stability was compromised, the knee contributed a larger role compared to the ankle as has been previously reported (see Section 6.6.5(C)).

The potential sources of error when calculating joint moments and, in turn powers are numerous, particularly due to the error inherent in the centre of pressure values calculated during initial foot-ground contact[503,948]. Consequently, although it was interesting to note that descriptions of the net ankle and knee joint moments and powers at initial foot-ground contact did not adequately infer what was occurring within the musculature surrounding the joints, these values may have been inaccurate. Furthermore, as the moments and powers in the present study were only calculated in the sagittal plane, the neuromuscular control of movement, as explained by the muscle activation patterns (see Section 6.6.2) would appear more reliable than the calculated ankle and knee joint moments and powers, particularly as it applied to multiplanar movement. Therefore, although between-group, within-footwear and within-surface differences with respect to ankle and knee joint moments were anticipated, these may have been more prolific in the frontal and transverse planes or, within the major phases of the moment and power curves compared to the sagittal plane moments at initial foot-ground contact. However, further research with larger subject numbers would be required to confirm or deny this notion.

Chapter 7

Summary & Conclusions

7.1 Summary of Results

Home falls and their resultant injuries are documented to be the leading cause of unintentional injury, disability, hospitalisation and death in the world for older people. Consequently, falls in the home place an ever-increasing strain on the community's financial resources and health care system. The most common mechanism of home falls in older people is that of slips, most of which occur at initial foot-ground contact. Of the many risk factors implicated in home slips, the major factors cited include inappropriate footwear, such as slippers, slippery surfaces and gait modifications caused by musculoskeletal problems such as arthritis and foot pathologies. A link between home falls and slipper wear has emerged in the literature and, despite many health professionals and researchers advocating otherwise, based on the results of Experimental Section A, older people still wear slippers around the home. Therefore, it is postulated that slippers may provide some fundamental design characteristics that should be incorporated into safe household shoes designed for older individuals, particularly those suffering arthritis and associated foot pathologies.

Few researchers have focused on the complex interplay between footwear-surface interactions, walking and balance when developing recommendations for selecting safe shoes, instead tending to discuss these items in isolation. However, it is information pertaining to how the three-component system, namely that of the person, the shoe and the supporting surface, interact that is vital in order to design safe household shoes for older people. Therefore, it was the purpose of Experimental Section B to identify how different household shoe-surface interactions, derived from Experimental Section A, altered the biomechanical indices characterising initial foot-ground contact in older women with RA, a population shown in Experimental Section A to be at risk of falls, have foot pathologies and require specialised shoes. The results of the experiments conducted on eight older women with RA and eight matched controls are summarised below, with particular attention given to those household shoe design characteristics that would appear to protect older people against home falls.

7.1.1 Preliminary Assessment Tasks

In partial agreement with Hypothesis (1), RA subjects reported a significantly higher incidence of foot problems, greater foot pain and increased difficulty in performing activities because of their feet, together with significantly reduced knee flexion and ankle dorsiflexion muscle strength in the preliminary assessment tasks compared to the control subjects. In addition, the RA subjects also required larger slipper sizes relative to foot length than the control subjects. However, in contrast with Hypothesis (1), the RA subjects did not display significantly increased foot reaction time, significantly decreased knee and ankle joint range of motion, consistently reduced plantar sensation or increased static and dynamic plantar pressures, particularly in the forefoot region, when compared to the control subjects. Therefore, although the RA subjects may be at greater risk of falls due to impaired lower limb muscle strength and have a greater need for specialised footwear due to their foot problems and foot pain, they were similar to the control subjects in terms of functionality and independence.

7.1.2 Between-Group Effects

In partial agreement with Hypothesis (2(i)), compared to the control subjects, the RA subjects displayed trends towards decreased knee flexion, decreased horizontal heel velocity and decreased foot/shoe angular velocity at initial foot-ground contact. However, in disagreement with Hypothesis (2(i)), the RA subjects did not display a trend towards decreased foot/shoe angle or significantly increased stance time compared to the control subjects. In fact, contrary to Hypothesis (2(i)), the RA subjects displayed a trend towards greater ankle dorsiflexion at initial foot-ground contact compared to the control subjects. Furthermore, in disagreement with Hypothesis (2(ii)) and Hypothesis (2(iii)), the RA subjects did not display any significant differences compared to the control subjects in the ground reaction forces, coefficient of friction values, joint moments or the neuromuscular variables calculated at initial foot-ground contact, although a trend towards increased ankle and knee joint powers at initial

foot-ground contact was noted. Therefore, although agreement was reached with Hypothesis (2(iv)) in that the RA subjects perceived significantly greater task difficulty and foot pain when completing the walking trials compared to the control subjects, these subjective perceptions did not transfer to altered lower limb control at initial foot-ground contact. Instead, as suggested following the preliminary assessment tasks (see Section 7.1.1), the RA and control subjects displayed similar strategies to control their lower limb at initial foot-ground contact regardless of footwear or surface condition.

7.1.3 Within-Footwear Effects

Subjects performed the walking trials under three footwear conditions, namely barefoot, closed back slippers and toe slippers. In partial agreement with Hypothesis (3(i)), when subjects performed the walking trials wearing the toe slippers, they displayed trends towards increased ankle dorsiflexion and decreased foot/shoe angular velocity; although increased knee flexion and horizontal heel velocity and no change in foot/shoe angle or stance time compared to walking barefoot or in closed back slippers. Furthermore, consistent with Hypothesis (3(ii)), subjects recorded significantly decreased vertical braking forces when they walked barefoot compared to walking in toe slippers, as well as significantly decreased anteroposterior braking forces and significantly increased static coefficient of friction when they walked barefoot compared to walking shod. However, in disagreement with Hypothesis (3(ii)), there were no differences in braking forces or coefficient of friction variables between the two slipper conditions and no within-footwear main effects on the time from initial foot-ground contact to the braking force peaks, coefficient of friction variables or knee joint moments. Interestingly, weak trends displaying increased ankle joint moment and power magnitudes when subjects walked wearing the toe slippers compared to the other footwear conditions and increased knee joint power magnitude when subjects walked barefoot compared to walking shod were noted in direct opposition to Hypothesis (3(ii)).

When subjects performed the walking trials wearing toe slippers, they recorded significantly earlier VL, TA and PL activation, significantly longer PL

burst duration, significantly increased TA burst intensity and significantly earlier G offset compared to walking in closed back slippers. In addition, subjects recorded significantly earlier TA and PL activation, significantly longer PL burst duration, significantly greater TA burst intensity and significantly delayed G offset when they walked wearing toe slippers compared to walking barefoot. When subjects walked barefoot, they recorded significantly earlier RF, VL and PL onset, significantly longer BF burst duration, significantly reduced TA burst intensity and significantly earlier G offset compared to walking in closed back slippers. However, there were no within-footwear main effects pertaining to earlier, or increased hamstring or G activation, reduced quadriceps burst intensity or task difficulty. Furthermore, subjects perceived the closed back slippers to be the most comfortable and least slippery footwear condition to walk in, followed by barefoot walking and then walking in toe slippers. Therefore, these results were in partial agreement with Hypothesis (3(iii)) and Hypothesis (3(iv)).

When subjects walked wearing either slipper type they appeared better able to reduce their ground reaction forces and did not require as many neuromuscular adaptations as when walking barefoot to achieve safe initial foot-ground contact. However, when compared to closed back slippers, the toe slippers were deemed the most uncomfortable and slipperiest footwear type. Toe slippers also required the greatest number of kinematic and neuromuscular modifications, in particular ankle dorsiflexor muscle activity, for the achievement of safe initial foot-ground contact. Therefore, the toe slippers were not only the most uncomfortable footwear type, but were also the most difficult to keep on the foot. Furthermore, older people with toe deformities or restricted toe motion, such as the typical RA patient, may be unable to grasp with their toes to help keep the toe slipper on their foot. Consequently, toe slippers are not recommended for this population of older women, particularly those with foot problems, as they would contribute to foot pain and place the individual at an increased risk of sustaining a slip and fall. Instead, closed back slippers could be worn by older women with foot problems around the home in place of walking barefoot.

7.1.4 Within-Surface Effects

Subjects performed the walking trials under three surface conditions, namely carpet, dry vinyl tile and wet vinyl tile. In disagreement with Hypothesis (4(i)) and Hypothesis (4(ii)), there were no effects of surface type on ankle angle, knee angle, horizontal heel velocity, stance time, vertical braking force, dynamic coefficient of friction, timing of the ground reaction force variables or the knee and ankle joint moments or powers. However, in partial agreement with Hypothesis (4(i)), subjects displayed trends towards decreased foot/shoe angle and angular velocity when walking on the wet vinyl tile surface compared to the dry vinyl tile surface, as well as decreased foot/shoe angle and increased horizontal heel slide when they walked on the vinyl tile surfaces compared to the carpet surface. Furthermore, consistent with Hypothesis (4(ii)), subjects displayed significantly decreased anteroposterior braking forces and static coefficient of friction variables when they walked on the wet vinyl tile surface compared to the other surface conditions.

When subjects walked across the wet vinyl tile surface, they recorded a significantly delayed RF offset which contributed to a significantly longer RF burst duration, significantly earlier TA offset as well as significantly greater RF and S burst intensity compared to when walking across carpet or dry vinyl tile. In comparison, when the subjects walked across the carpet surface, VL burst duration was significantly shorter compared to when they walked across the dry vinyl tile or wet vinyl tile surfaces. Therefore, these results were mainly contrary to Hypothesis (4(iv)), although there were no effects of surface type on hamstring timing and, differing to the within-footwear effects (see Section 7.1.3), there were no effects of surface type on PL or G burst activity.

As proposed in Hypothesis (4(iv)), subjects perceived the carpet surface to be the least slippery and the wet vinyl tile to be the most slippery surface when walking. However, in partial agreement with Hypothesis (4(iv)), although the wet vinyl tile surface was considered significantly more difficult to walk across, carpet was not considered the easiest surface to traverse. Instead, subjects perceived the dry vinyl tile as the easiest surface to walk across. Gait adaptation is a necessary adjustment when walking on a wet surface to avoid a

slip and/or fall. Therefore, as subjects in the present study perceived the wet vinyl tile surface to be slippery and difficult to walk across, they altered their walking patterns so that they could safely walk without slipping. Interestingly, there were very few differences noted when subjects walked across the carpet and dry vinyl tile surfaces, perhaps indicating the low pile carpet and vinyl tile are perceived similarly by older women.

7.1.5 Interaction Effects

There was only one significant subject group x footwear interaction in the present study whereby the RA subjects displayed significantly longer VL burst duration when they walked in the toe slippers compared to the control subjects. In addition, although not analysed statistically, subject group x footwear interaction patterns similar to that of VL burst duration were derived for knee flexion angle and horizontal heel slide as well as the ankle moment and power at initial foot-ground contact. These patterns indicated that different strategies were required by the RA subjects when they walked in the toe slippers compared to the control subjects. Therefore, although anticipated due to the high incidence of forefoot problems in this population, only two RA and two control subjects were assessed and further research is warranted to confirm or deny this notion. Further two-way interactions with subject group would be anticipated with less functional RA subjects and, due to their increased incidence of foot problems and foot pain as well as reduced knee flexion and ankle dorsiflexion muscle strength, it would be anticipated that RA patients would have significantly different footwear needs compared to their non-arthritic counterparts. However, in the present study and in disagreement with Hypothesis (5), the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking barefoot or wearing a specific slipper were not dependent upon subject group.

Consistent with Hypothesis (6), there was only one significant subject group x surface interaction such that the control subjects displayed a significantly later TA offset relative to initial foot-ground contact compared to the RA subjects but only when walking across the carpet surface. Instead, both

subject groups prepared for initial foot-ground contact in a comparable fashion when walking on different surface types. Therefore, the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking on a specific surface were not dependent upon subject group.

In agreement with Hypothesis (7), there were significant footwear x surface interactions pertaining to the subjective estimations of task difficulty, shoe/surface slipperiness and shoe comfort; RF and G burst duration; RF, VL and S burst intensity; stance time and peak horizontal braking forces in the present study. Furthermore, although not conducted statistically, footwear x surface interaction patterns were revealed for foot/shoe angular velocity as well as the ankle joint moment and power calculated at initial foot-ground contact.

When subjects walked barefoot on the wet vinyl tile surface, they displayed slower mean foot/shoe angular velocities; perceived the interaction as significantly more uncomfortable, slippery and difficult to walk across and recorded significantly reduced horizontal braking forces compared to walking in closed back or toe slippers on the same surface. Furthermore, walking barefoot on the wet vinyl tile surface, lead to significantly longer RF and G burst durations as well as significantly greater RF and VL burst intensities compared to walking in closed back slippers and a significantly longer RF burst duration and significantly greater RF and S burst intensities compared to walking in toe slippers on the same surface. Interestingly, when subjects walked across the carpet surface, they reported the toe slippers to be the least comfortable footwear condition compared to walking barefoot or in closed back slippers and, displayed a significantly shorter stance time when walking barefoot on carpet compared to walking in closed back slippers. Toe slippers were considered less comfortable and required trends towards increased ankle moments and powers compared to walking barefoot across dry vinyl tile. In addition, when walking across the dry vinyl tile surface, subjects recorded significantly shorter stance times when barefoot compared to when walking in closed back or toe slippers and significantly reduced horizontal braking forces at initial foot-ground contact compared to walking in closed back slippers.

The majority of footwear x surface interactions revealed that when subjects walked barefoot on the wet vinyl tile surface, they used significantly altered strategies to control their lower limb at initial foot-ground contact compared to the other conditions. The wet vinyl tile surface itself, was perceived as slippery and thereby commanded altered movement strategies to reduce slip risk (see Section 7.1.4). However, these altered movement strategies appeared to be compounded when subjects walked across the wet vinyl tile surface barefoot. When barefoot, subjects receive sensory feedback from the supporting surface and, in this case, the wet vinyl tile surface felt slippery to the subjects, resulting in cautious gait changes. In comparison, when shod, this sensory feedback was occluded by the shoe sole and, consequently, subjects may have been unable to distinguish between a slippery and non-slippery surface, thereby recording no change in their gait pattern and, perhaps increasing their slip risk. Alternatively, subjects may alter their gait initially based on their subjective perception of surface slipperiness, although return to their normal gait pattern if their sensations do not confirm these perceptions. Given that the frictional properties of the plantar surface of the foot and the rubber sole of the slippers differed, gait changes would be anticipated when subjects walked barefoot on the wet, slippery surface. However, not altering gait due to the habitual use of footwear and a dependence on the role provided by shoes, particularly shoes which are marketed as possessing “non-slip” soles, could also place the older individual at a heightened risk of incurring a slip. Therefore, the biomechanical parameters characterising initial foot-ground contact and the subjective perceptions when walking barefoot or wearing a specific slipper were dependent upon the surface walked across. As such, the design and prescription of appropriate footwear must take into consideration the surfaces within an older person’s home, if home falls in older women are to be reduced.

In disagreement with Hypothesis (8), there were no significant subject group x footwear type x surface type interactions for any of the biomechanical parameters characterising initial foot-ground contact or the subjective perceptions in the present study. It is postulated that the relatively functional RA sample contributed to this finding by minimising between-group differences

(see Section 7.1.2). Consequently, further research examining a less functional RA sample is warranted.

7.2 Conclusions

Based on the results of the present thesis it was concluded that toe slippers were unsafe for older women. This is because toe slippers were perceived as being more difficult to wear and required greater muscular effort to keep the shoe on the foot compared to the other footwear conditions. Furthermore, although walking barefoot allowed the older women to adjust their gait to the more slippery condition to avoid slipping, subjects did not like walking barefoot under all surface conditions. In addition, together with possible skin traumas associated with habitual barefoot gait, unprotected barefoot walking is not considered safe for this population. Therefore, it is recommended that older women, particularly those with RA, not wear toe slippers or walk barefoot around the home. However, the results of the present thesis revealed that well-fitted closed back slippers could be worn safely by older women when walking on typical household surfaces, including older women with RA. Interestingly, the closed back slippers also encompassed many of the published design recommendations for safe footwear. Therefore, it is recommended that older women, particularly those with RA, can safely wear well-fitted closed back slippers around the home when walking on carpet and vinyl tile surfaces. However, further research is required to determine whether the wearing of well-fitted closed back slippers by older women could reduce falls in the home.

7.3 Recommendations for Further Research

Based on the findings of the present thesis, the following recommendations were suggested to further the research in the area of examining the effects footwear-surface interactions on the gait of older people:

(1) Despite being diagnosed with adult onset RA for over 5 years, the RA subjects who volunteered to participate in Experimental Section B appeared to have their RA controlled by medication such that they had no major functional limitations.

Therefore, further research is warranted to determine the effects of footwear and surface type on a less functional RA subject group.

(2) Research in the present study was conducted on two groups of older women and consequently, further research should be conducted to determine whether older men display similar changes when walking in household footwear on typical household surfaces.

(3) Although the RA and control subjects in the present study had different foot lengths, they required the same slipper sizes. Therefore, further investigation is recommended to better characterise anthropometrical parameters of the feet of older people, to ensure that the shape of shoe last, upon which shoes for older women are based, adequately encompass the older foot.

(4) Due to equipment failure, the present study analysed the biomechanical indices describing initial foot-ground contact in the sagittal plane. It is suggested that, further investigation determines how household footwear, surface type and RA incidence affect the biomechanical indices describing initial foot-ground contact in all three planes of motion.

(5) Two common household shoe designs and two common household surfaces were assessed in the present study, providing valuable information with respect to household shoe design for older women. Further research examining other household and outdoor shoe designs combined with typical household and outdoor surfaces should be completed to contribute to developing recommendations for safe shoe design for older people to reduce falls.

(6) When subjects walked wearing toe slippers in the present study, they reported that their feet slipped both in and out of the shoes. Therefore, further research is warranted to determine how foot slippage both inside and outside the slipper affects the perception of shoe/surface slipperiness as well as the biomechanical variables characterising gait at initial foot-ground contact, particularly with respect to falls risk and footwear recommendations.

(7) Often an older individual will fall as a result of slipping on an unseen puddle of water. In the present study, the wet vinyl tile surface was fully wetted. Therefore, further research is warranted to examine the gait of older people

walking across a surface with an uneven surface coating of water or pools of water.

(8) Slips and trips, the leading cause of falls in older people, can occur at any stage of the gait cycle. Whereas the present thesis examined the biomechanical indices describing gait at initial foot-ground contact, further research is warranted to examine the biomechanics of gait throughout the entire gait cycle to determine how gait events, such as terminal stance and swing, are influenced by household footwear-surface interactions.

(9) Older people require increased attention for gait control such that many falls occur when an older individual's locus of attention is diverted from the required task[879,949,950]. In the present study, subjects were made fully aware of the footwear and surface changes and were able to give their full attention to successfully completing the walking task. Therefore, further investigation is warranted to determine how older people alter their walking patterns when wearing different footwear types and traversing different surfaces when provided with different visual, vestibular and proprioceptive inputs to determine how sensory conflict affects the postural control system.

(10) In the present study, the foot was modelled as a rigid body being controlled by the muscles surrounding the ankle. However, structures of the foot can move relative to each other under the control of the foot's intrinsic musculature. In fact, atrophy of the intrinsic muscles of the foot may severely compromise the muscle strength and motor function of the foot as well as lead to foot deformities and altered foot pressures during gait[951-953]. Therefore, further knowledge as to how footwear and surface type affect the kinematics, kinetics and muscular control of the foot has implications for the design and construction of functional footwear for older people, warranting further research in this area.

(11) Many research studies have implicated household footwear as contributors to home falls in older people. As the shoe design parameters thought to be beneficial for older women in the present study are encompassed in the design of household shoes, further investigation is warranted to determine whether household shoes lead to falls in older people or, whether older people fall regardless of the footwear worn.

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Appendix A

Experimental Section A Documentation

HREC Approval: Survey

UNIVERSITY OF WOLLONGONG
Office of Research

APPROVED
In reply please quote: DC:KM HE96/59
Further Information: Karen McRae (Ext 4457)

17 April 1996

Ms Bridget Munro
Department of Biomedical Science
University of Wollongong

Dear Ms Munro,

I am pleased to advise that the following Human Research Ethics application has been approved:

Ethics Number:	HE 96/59
Project Title:	Footwear purchasing and wearing habits of persons aged 65 years and above
Name of Researcher:	Bridget Munro Julie Steele
Approval Date:	11 April 1996
Duration of Approval:	10 April 1997

This certificate relates to the research protocol submitted in your application of 1 April 1996. It will be necessary to inform the Committee of any changes to the research protocol and seek clearance in such an event.

Please note that experiments of long duration must be reviewed annually by the Committee and it will be necessary for you to apply for renewal of this application if experimentation is to continue beyond one year.

Professor G.D. Calvert
Chairperson
Human Research Ethics Committee
cc. Julie Steele, Supervisor

Address: Northfields Avenue, Wollongong 2522 Australia
Telephone: (042) 213 386 (direct); (042) 213 555 (switchboard) Facsimile: (042) 214 338

FOOTWEAR SURVEY:

WHAT DO YOU WEAR ON YOUR FEET AT HOME ?

Survey Number

Subject Number

---

1. Are you male or female?

Male	1
Female	2

2. What is your date of birth? / /

3. What is the postcode of where you are currently living?

4. Are you an Australian citizen?

Yes...	1
No	2

5. In which country were you born? ______________________

If Australia go to Q 6. If not, go to Q 7.

6. Are you of Aboriginal or Torres Strait Islander origin?

No	1
Yes, Aboriginal...	2
Yes, Torres Strait Islander	3

7. How would you rate your overall state of health?

Excellent	1
Very good..	2
Good	3
Fair...	4
Poor	5

8. Have you ever been diagnosed with any of the following medical conditions? **YES NO**

Meniere's Disease/Vertigo.............................	1
Epilepsy	2
Cataracts/Glaucoma/Poor vision	3
Bad circulation in the legs...............................	4
Vascular disease/Leg ulcers...........................	5
Diabetes...	6
Stroke/Tias...	7
Heart Attack	8
Angina/Palpitations/Heart failure.....................	9
High blood pressure.......................................	10
High cholesterol	11
Osteoporosis..	12
Arthritis..	13
Lung problems (Asthma/Emphysema/etc)	14
Foot problems	15
Other (Please specify): ___________________	16

9. On an average day, how long can you walk for before you need a rest?

Less than 5 minutes...	1
5 - 10 minutes	2
10 - 15 minutes	3
15 - 30 minutes	4
30 minutes to 1 hour	5
More than 1 hour...	6

10. Do you need any help to perform the following activities?

Cooking..	1
Cleaning..	2
Bathing/toileting	3

Shopping 4

11a) Have you been to visit your doctor in the last 2 weeks?

Yes 1

No 2

b) **If yes**, what problems/conditions did you visit your doctor for?

____

____

____

____

12a) Have you started taking any new medications in the last two weeks?

Yes 1

No 2

b) **If yes**, why have you started to take these medications?

____

____

____

____

13a) Were you in hospital in the last year?

Yes 1

No 2

b) **If yes**, how many times were you in hospital in the last year?__

c) **If yes**, for what problems/conditions were you in hospital for?

____

____

____

____

14. Do you currently have any of the following foot problems?

Tick if you do have the problems

Corns	1
Bunions	2
Callouses	3
Swollen feet	4
Flat feet	5
Gout	6
Amputated toes	7
Rash	8
Blisters	9
Hammer toes	10
Osteoarthritis	11
Rheumatoid Arthritis	12
Dry skin	13
Foot drop	14
Ingrown toenails	15
Hard thick nails	16
Warts	17
Other (please specify: ________________________	18

15a) Do you ever experience pain and/or discomfort on your feet?

Yes	1
No	2

b) **If yes**, where do you experience pain and discomfort on your feet (eg heels, toes)?

____

____

c) **If yes**, when do your feet experience pain and discomfort (eg morning, hot weather)?

____

____

d) **If yes**, what activities cause your feet pain and discomfort (eg walking)?

____

16a) Have you ever visited health/medical personnel about your feet?

Yes.. 1

No 2

b) **If yes**, who did you visit? (eg podiatrist, GP): _______________

c) **If yes**, for what foot problems did you visit health/medical personnel for?

____

17a) Do you currently wear a special insole/appliance in your shoe?

Yes.. 1

No 2

b) **If yes**, what is the appliance? ____________________________

c) **If yes**, who provided it? (eg podiatrist, GP) ________________

18a) Have you suffered any falls in the past year?

By fall we mean an accidental loss of balance/trip/slip where you find yourself on the ground

Yes...................... **GO TO Q 19** 1

No .... **GO TO Q 20** 2

b) **If yes**, how many times have you fallen in the past year? _____

19a) Were you wearing shoes at the time of your fall?

Yes...................... **GO TO Q 19c)** 1

No .... **GO TO Q 19b)** 2

b) **If no**, describe what you were you wearing when you fell?

Barefeet 1

Thick socks 2

Thin socks.. 3

Nylons (stockings/hose)... 4

Other (please specify: ______________________ 5

c) **If yes**, describe the type of shoe you were wearing at the time of your fall?

Thongs, slip-ons, sandals without straps at the back 1

Sandals with straps at the front & back (open toes) .. 2

Sandals with straps at the front & back (closed toes) 3

Slippers (only cover front foot)................................. 4

Slippers (cover whole foot) 5

Closed toed slip-ons (excluding above).................... 6

Open toed slip-ons (excluding above) 7

Tie or buckle shoes (not athletic).......................... 8

Tied Athletic shoe 9

Boots.. 10

Galoshes/gumboots.. 11

Orthopaedic moulded shoes 12

Other (please specify): ______________________ 13

IF TIE/BUCKLE/FASTEN SHOE ANSWER Q 19 (d).

d) Were your shoes fastened when you fell?

Yes... 1

No 2

Unsure 3

20a) Do you ever find it hard to put your shoes on your feet?

Yes... 1

No 2

b) **If yes**, in what way(s) do you find it hard to put your shoes on your feet?

Reaching down to feet 1
Doing up shoe fastenings (ie laces/buckles)............. 2
Getting shoe on foot.. 3
Other (please specify):______________________ 4

21a) Do you wear shoes in and around the house?

Yes..................... **GO TO Q 21c)** 1
No .... **GO TO Q 21b)** 2

b) **If no**, can you describe what you wear around the house?

Barefeet 1
Thick socks 2
Thin socks... 3
Nylons (stockings/hose)... 4
Other (please specify: ______________________ 5

If you answered **NO TO Q 21 (a)**, thank you very much for being involved in the survey. Please remember to place the survey in the stamped self addressed envelope and place in the mail box.

If you answered **YES TO Q 21 (a)** please go and get your household shoes so you can answer the following questions easily.

c) **If yes**, can you describe the type of shoes that you wear in and around the house ?

Thongs, slip-ons, sandals without straps at the back 1
Sandals with straps at the front & back (open toes) .. 2
Sandals with straps at the front & back (closed toes) 3
Slippers (only cover front foot)................................. 4
Slippers (cover whole foot) 5
Closed toed slip-ons (excluding above).................... 6
Open toed slip-ons (excluding above) 7
Tie or buckle shoes (not athletic).......................... 8
Tied Athletic shoe.. 9

Boots 10
Galoshes/gumboots 11
Orthopaedic moulded shoes 12
Other (please specify): ________________________ 13

IF TIE/BUCKLE/FASTEN SHOE ANSWER Q 21 (d).

d) What type of fastening have your household shoes/slippers?

Zipper 1
Buckle 2
Laces 3
Velcro 4
Other (please specify): _________________________ 5

22a) Are these shoes always fastened when you wear them?

Yes 1
No 2

b) **If no**, why are they not always fastened when you wear them?

____

23. What is the colour of the shoe/slipper you currently wear around the house? ___________________________________

24. What type of material is the top of your household shoe/slipper made of?

Fabric 1
Hard Leather 2
Soft leather 3
Vinyl 4
Rubber 5
Plastic 6
Cloth (ie canvas) 7
Sheepskin 8
Mixture of materials 9

Other (please specify):________________________ 10

25. What is the sole of your household shoe/slipper like?
    - Smooth, flat.. 1
    - Rough 2
    - Patterned 3
    - Other (please specify):________________________ 4

26. What is the inside material/lining of your household shoe/slipper made of?
    - Fabric... 1
    - Foam sponge.. 2
    - Leather.. 3
    - Vinyl 4
    - Sheepskin 5
    - Plastic 6
    - Rubber 7
    - No lining.. 8
    - Other (please specify):________________________ 9

27. What sort of material is the sole of your household shoe/slipper?
    - Leather.. 1
    - Moleskin.. 2
    - Hard synthetic (eg rubber/plastic).............................. 3
    - Soft synthetic (eg rubber/plastic) 4
    - Unsure 5
    - Other (please specify):________________________ 6

28. Is the sole easy to bend with your hands?
    - Yes.. 1
    - No 2

29. How high are the heels on your current household shoes/slippers?

Flat/no heel (<1”/2.5 cm)...	1
Medium (1”/2.5 cm)..	2
High (higher than 1”/2.5 cm)	3
Other (please specify):_______________________	4

30. Do you ever wear these shoes outside the house (eg in the garden, down the street, shopping, to the doctor)?

Yes...	1
No	2

31. On which surface does your household shoe/slipper feel the most slippery?

Carpet, low pile	1
Carpet, high pile..	2
Smooth tile	3
Rough tile/Outdoor paving	4
Linoleum	5
Concrete/asphalt/blacktop	6
Grass	7
Floor boards..	8
Unsealed surface - gravel..	9
Other (please specify):_______________________	10

32. How long have you had your current household shoes/slippers for?

Less than 1 year	1
1-2 years..	2
2-5 years..	3
More than 5 years...	4

33. What are the signs your shoes show from ageing (eg worn sole, holes)?

____

34. When do you usually wear your household shoes/slippers?

Only at night..	1

Only in the morning.. 2
Morning and night 3
All day 4
When feet are cold.. 5
When feet are sore 6
Only in the garden... 7
At bathing times 8
Varies.. 9
Other (please specify):_______________________ 10

35. How long do you wear your household shoes/slippers each day?

Less than 1 hour 1
1-2 hours.. 2
2-5 hours.. 3
More than 5 hours... 4

36a) Do your household shoes/slippers fit your feet well?

Yes.. 1
No 2

b) **If no**, why don't they fit your feet well?

A little loose... 1
Very loose 2
Very tight.. 3
A little tight 4
Other (please specify):_______________________ 5

37. Why did you buy the shoe/slipper that you currently wear around the house?

Old shoes were worn out/falling apart........................ 1
Fashion style... 2
Saw them and like them.. 3
Didn't buy - Gift 4

Medical advice 5
Foot problem... 6
Weather 7
Comfortable 8
Ease of putting them on 9
Other (please specify):________________________ 10

38a) Do you find your household shoes/slippers comfortable?

Yes...................... **GO TO Q 38b)** 1
No .... **GO TO Q 38c)** 2

b) **If yes**, why do you find your shoe comfortable/uncomfortable?

____

____

39a) Do you do anything to your household shoes/slippers that makes them more comfortable for your feet?

Yes... 1
No 2

b) **If yes**, what do you do to make them more comfortable for your feet?

____

____

40. Do you wear your special insole or appliance in your household shoes/slippers?

Yes... 1
No 2

41a) Do you wear stockings/socks with your household shoes/slippers?

Yes... 1
No 2

b) What type of stockings/socks do you wear with your household shoes/slippers?

Thin socks..	1
Thick socks	2
Full length stockings	3
Knee high stockings..	4
Other (please specify):________________________	5

42. What is important to you when buying household shoes/slippers?

____

____

43. How often do you buy household shoes/slippers (mark all that apply)?

Under 6 months	1
6 months - 1 year	2
1 - 2 years	3
Only when I need to	4
At yearly sales...	5
When others tell me I need to	6
When I see something I like......................................	7
I don't buy shoes...	8
Other (please specify):________________________	9

44. When was the last time you had your feet measured at a shoe store?

Less than 6 months ago..	1
6 months - 1 year ago	2
1 - 2 years ago	3
2 - 5 years ago	4
More than 5 years ago	5
Other (please specify):________________________	6

45. What colour shoe/slipper do you prefer when buying household shoes/slippers? ____________________________

46. Where do you usually go when you are trying to buy household shoes/slippers?

Variety store without assistance (eg. Kmart, Coles) ..	1
Department store with assistance (eg. Grace Bros) ..	2
Shoe store (eg. Williams, Mathers)............................	3
Custom made footwear store....................................	4
Chemist...	5
Second hand store..	6
Other (please specify):_______________________	7

47. Do you prefer to purchase household shoes/slippers with a brand name you know?

Yes..	1
No	2

48. Do you prefer to purchase household shoes/slippers that are made in Australia?

Yes..	1
No	2

49. How much money would you be prepared to pay for a new pair of household shoes/slippers?

Less than $20	1
$20 - $30..	2
$30 - $50..	3
$50 - $100..	4
More than $100...	5

50. How much money would you be prepared to pay for a new pair of shoes to go out in?

Less than $20	1
$20 - $30..	2
$30 - $50..	3

$50 - $100 4
More than $100 5

51. What is your approximate yearly income?
Less than $8000 1
$8,001 - $12,000 2
$12,001 - $20,000 3
$20,001 - $30,000 4
$30,001 - $40,000 5
$40,001 and over 6

Thank you very much for completing the questionnaire

Appendix B

Experimental Section B Documentation

Appendix **Page**

HREC Approval: Laboratory Investigation

University of Wollongong

Office of Research

University of Wollongong
NSW 2522 Australia

Tel (042) 21 3386
Fax (042) 21 4338
International +61+42

CONDITIONAL APPROVAL
In reply please quote: SD:KM HE97/048
Further Information: Karen McRae (Ext 4457)

24 April 1997

Ms Bridget Munro
Biomedical Science
University of Wollongong

Dear Ms Munro,

I am pleased to advise that the following Human Research Ethics application has been **conditionally approved**:

Ethics Number:	HE97/048
Project Title:	Effects of footwear-surface interactions on locomotor tasks performed by elderly women
Name of Researchers:	Bridget Munro Julie Steele Mark Brown
Clearance Date:	22 April 1997
Duration of Clearance:	21 April 1998

This approval is granted subject to the following conditions:

(i) the use of the term "older" instead of "elderly" and a plain English alternative to the word "locomotor" on the Subject Information Package/Consent Form

Please provide written confirmation to the Secretary of the Committee before continuing your research, or approval will be withdrawn.

This certificate relates to the research protocol submitted in your original application 7 April 1997. It will be necessary to inform the Committee of any changes to the research protocol and seek clearance in such an event.

Please note that experiments of long duration must be reviewed annually by the Committee and it will be necessary for you to apply for renewal of this application if experimentation is to continue beyond one year.

Dr S. Dodds
Chairperson
Human Research Ethics Committee

cc. Julie Steele, Supervisor
Head, Biomedical Science

Modified Arthritis Impact Measurement Scales 2 (AIMS)

A score on the AIMS2[671] was achieved by summing the item responses to produce scale scores and then standardising these scores between 0 and 10. Each question was asked starting with "**During the past month…**"

Arthritis Scale
(Answer using: All days (1), Most days (2), Some days (3), Few days (4), No days (5))
- How often did you have severe pain from your arthritis?
- How often did you have pain in two or more joints at the same time?
- How often did your morning stiffness last > 1 hr from the time you woke up?
- How often did your pain make it difficult for you to sleep?
- How often have you had to take medication for your arthritis?

(Answer using: Severe, Moderate, Mild, Very Mild, None)
- During the past month, how would you describe your usual arthritis pain?

Mobility & Physical Activity Scale
(Answer using: All days (1), Most days (2), Some days (3), Few days (4), No days (5))
- How often were you physically able to drive a car/use public transportation?
- How often were you out of the house for at least part of the day?
- How often were you able to do errands in the neighbourhood?
- How often did someone have to assist you around outside your home?
- How often were you in a bed or chair for most or all of the day?
- Did you have trouble doing vigorous activities such as running, lifting heavy objects, or participating in strenuous sports?
- Did you have trouble walking several blocks/climbing a few flights of stairs?
- Did you have trouble bending, lifting or stooping?
- Did you have trouble either walking one block or climbing one flight of stairs?
- Were you unable to walk unless assisted by another person or by a cane, crutches, or walker?

Self-Care & Household Tasks Scale
(Answer using: Always (1), Very often (2), Sometimes (3), Almost never (4), Never (5))
- Did you need help to take a bath or shower?
- Did you need help to get dressed?
- Did you need help to use the toilet?
- Did you need help to get in or out of bed?
- If you had the necessary transportation, could you go shopping for groceries without help?
- If you had kitchen facilities, could you prepare you own meals without help?
- If you had household tools and appliances, could you do your own housework without help?

- If you had laundry facilities, could you do your own laundry without help?

Foot Function Index (FFI)

The total item scores for each sub-scale of the FFI[670] are divided by the maximum score possible for each sub-scale, multiplied by 100, and averaged to obtain a score that represents total foot function.

SUBSCALE	SCORE
FOOT PAIN **How severe is your foot pain?** (0 = No pain to 10 = Worst pain imaginable) • At worst • In the morning • When standing barefoot • When you walk barefoot • When standing with shoes • When walking with shoes • When standing with orthotics • When walking with orthotics • At end of day **TOTAL** **TOTAL/90 *100** **DISABILITY** **How much difficulty do you have because of your feet?** (0 = No difficulty; 10 = So difficult, unable) • Walking in the house • Walking outside • Walking 4 blocks • Climbing stairs • Descending stairs • Standing tip toe • Getting up from chair • Climbing curbs • Walking fast **TOTAL** **TOTAL/90 *100** **ACTIVITY LIMITATION** **How much of the time do you?** (0 = None of the time to 100 = All of the time) • Stay inside all day because of your feet • Stay in bed all day because of your feet • Limit activities because of your feet • Use assistive devices indoors • Use assistive devices outdoors **TOTAL** **TOTAL/50 *100**	
TOTAL SCORE - AVERAGE OF THREE SUB-SCALE SCORES	

Physical Assessment Reliability Procedure

Interrater reliability of the Chief Investigator for each of the physical assessments detailed in Section 5.2.2 was established by taking three measurements of the same physical dimensions for three older women (mean age, 62 years) on three consecutive days. Intraclass correlation coefficients (ICC) type (3,1) were then calculated on the data using the method described by Vincent[698]. Mean data for each day and calculated ICCs are presented for height, body mass, lower limb segmental proportionality and slipper size in Table B.4.1 and for lower limb strength and flexibility measurements in Table B.4.2. As the ICCs exceeded 0.93 for each of the physical assessment tests, the results obtained by the Chief Investigator were considered highly reproducible and, therefore, reliable.

Table B.4.1: Data pertaining to the reliability testing for anthropometry measurements for three subjects over three days.

Variable	Day 1			Day 2			Day 3			ICC
Subject	1	2	3	1	2	3	1	2	3	
Height (m)	1.50	1.56	1.64	1.48	1.56	1.64	1.49	1.56	1.65	0.997
Mass (kg)	61	68	92.4	61	68	92.3	62	67	92.3	0.999
Circumferences (cm)										
Lying Foot	20.7	20.6	21.7	20.6	20.4	21.8	20.6	20.5	21.8	0.996
Standing Foot	22.0	21.4	22.1	21.9	21.3	22.1	21.9	21.3	22.1	0.995
Calf	34.7	33.2	35.8	34.8	33.4	35.8	34.6	33.3	35.9	0.998
Upper Thigh	52.6	56.3	63.8	52.8	56.3	63.8	53.2	56.1	64.0	0.999
Lengths (mm)										
Lying Foot	224	215	227	225	214	227	224	214	226	0.997
Standing Foot	239	227	247	239	227	247	239	227	246	0.999
Calf	368	360	401	367	358	402	367	358	400	0.999
Upper Thigh	347	326	340	346	333	340	345	329	341	0.979
Widths (mm)										
MTP heads [a]	111	85	128	112	85	128	112	85	129	0.999
Ankle	64	62	66	65	61	66	65	61	67	0.983
Knee	85	120	92	85	120	91	85	119	92	0.999

[a] MTP = metatarsophalangeal

Table B.4.2: Data pertaining to the reliability testing for the lower limb strength and flexibility measurements for three subjects over three days.

Variable	Day 1			Day 2			Day 3			ICC
Subject	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	
Ankle Strength (kg)										
Dorsiflexion	19.5	13.2	23.6	19.0	14.1	24.6	20.1	12.8	25.0	0.995
Plantar flexion	28.8	19.7	17.8	30.1	17.8	17.1	29.1	18.5	17.8	0.996
Knee Strength (kg)										
Flexion	14.4	8.1	10.1	14.8	8.5	13.1	14.7	8.2	10.6	0.971
Extension	19.7	17.0	20.8	21.7	18.1	20.4	22.7	17.5	20.8	0.928
Ankle Flexibility (°)										
Dorsiflexion	95	100	111	95	104	115	90	98	114	0.976
Plantar flexion	165	163	175	168	165	169	170	163	168	0.927
Knee Flexibility (°)										
Flexion	41	44	46	39	47	45	42	45	45	0.923
Extension	182	177	170	184	174	170	182	176	172	0.985

IRED Marker Movement Estimations

Data from motion analysis systems are used to make objective suggestions with respect to human gait and recommendations of interventions may be prescribed from these data. Therefore, it is critical that good repeatability of the data is established to allow detection of actual changes in gait patterns between successive trials and/or interventions. One factor affecting the repeatability of the biomechanical data between sessions is marker movement as these data are used for successive calculations such as for joint moment and joint power calculations. Therefore, marker movement was kept to a minimum to decrease the error inherent in these successive calculations (see Section 5.3.2).

Movement of the IREDs during the walking trials in the present study was calculated using the three-dimensional positional data collected using the OPTO***TRAK*** 3020 motion analysis system (see Section 5.3.2). The x and y coordinates of each IRED that defined the foot (IREDs 1 and 2), leg (IREDs 6 and 9) and thigh (IREDs 11 and 14; see Table 5.2 and Figure 5.6) segments were used to calculate IRED movement by determining the inter-IRED distance for each segment during the walking trials with respect to the static trial data. The IRED movement for each trial for the two control and two RA subjects whose kinematic data were deemed acceptable for further analysis are reported in Table B.5.1, Table B.5.2, Table B.5.3 and Table B.5.4. Using this method, mean marker movement in the present study was 9.7 ± 3.8 mm for the foot, 6.8 ± 2.3 mm for the leg and for the 16.6 ± 9.8 mm for the thigh.

Table B.5.1: Descriptive statistics for the mean IRED marker movement (mm) over each trial for each condition for Control Subject 1.

Segment	Condition									
	Static	Carpet			Dry Vinyl Tile			Wet Vinyl Tile		
		Mean	Range	Movement	Mean	Range	Movement	Mean	Range	Movement
Barefoot										
Foot	135.9	135.3	(129.3-137.4)	8.1	135.1	(128.1-138.0)	9.9	133.8	(129.9-136.3)	6.4
Leg/Shank	316.1	316.7	(315.0-318.2)	3.2	316.4	(313.5-317.8)	4.4	315.6	(312.9-317.9)	5.0
Thigh	178.2	179.7	(172.1-190.3)	18.3	178.8	(175.0-188.1)	13.2	178.9	(175.1-191.4)	16.3
Closed Toe Slippers										
Foot	135.7	134.5	(128.8-136.8)	8.0	134.6	(128.4-138.0)	9.6	134.9	(128.2-137.8)	9.6
Leg/Shank	316.1	316.3	(314.7-317.8)	3.1	316.1	(313.0-318.2)	5.1	315.5	(311.4-319.2)	7.8
Thigh	177.9	179.3	(171.9-192.7)	20.8	179.9	(171.7-196.2)	24.6	179.1	(170.1-196.7)	26.6
Toe Slippers										
Foot	136.7	139.2	(132.6-142.1)	9.5	137.7	(132.2-142.5)	10.3	138.7	(134.2-142.3)	8.1
Leg/Shank	315.9	316.1	(313.4-318.1)	4.7	316.0	(313.1-318.0)	4.9	316.2	(314.7-318.1)	3.4
Thigh	178.4	179.1	(170.1-195.6)	25.5	180.4	(168.4-196.4)	28.0	178.9	(165.2-194.1)	28.8

Appendix B.5

Table B.5.2: Descriptive statistics for the mean IRED marker movement (mm) over each trial for each condition for Control Subject 2.

Segment	Condition									
	Static	Carpet			Dry Vinyl Tile			Wet Vinyl Tile		
		Mean	Range	Movement	Mean	Range	Movement	Mean	Range	Movement
Barefoot										
Foot	145.1	143.9	(139.2-145.9)	6.6	144.2	(138.4-146.1)	7.7	143.2	(138.6-146.0)	7.3
Leg/Shank	254.3	251.6	(246.6-253.3)	6.7	251.8	(249.0-253.8)	4.8	251.9	(249.2-253.5)	4.3
Thigh	113.5	114.0	(111.7-116.1)	4.4	114.1	(111.9-116.0)	4.1	114.7	(113.1-117.3)	4.2
Closed Toe Slippers										
Foot	145.0	143.4	(137.9-145.9)	8.0	144.3	(139.4-147.0)	7.6	144.1	(138.3-147.5)	9.2
Leg/Shank	252.9	251.2	(247.1-253.4)	6.3	251.6	(247.0-253.7)	6.7	251.3	(247.3-253.5)	6.3
Thigh	113.4	113.4	(107.2-117.1)	10.0	114.1	(111.2-121.2)	9.9	113.5	(111.0-121.1)	10.2
Toe Slippers										
Foot	134.1	136.9	(131.5-140.9)	9.4	137.2	(132.9-142.4)	9.4	139.3	(135.8-143.0)	7.2
Leg/Shank	253.3	251.3	(247.8-253.4)	5.7	252.1	(249.9-254.1)	4.1	251.4	(246.5-253.4)	6.9
Thigh	113.1	113.8	(108.6-122.6)	13.9	114.5	(111.9-122.0)	10.1	113.9	(111.6-124.0)	12.4

Table B.5.3: Descriptive statistics for the mean IRED marker movement over each trial for each condition for RA Subject 1.

Segment	Condition									
	Static	Carpet			Dry Vinyl Tile			Wet Vinyl Tile		
		Mean	Range	Movement	Mean	Range	Movement	Mean	Range	Movement
Barefoot										
Foot	101.7	104.9	(102.5-107.1)	4.6	102.0	(98.8-103.5)	4.7	96.4	(92.2-98.6)	6.5
Leg/Shank	287.3	285.4	(280.9-287.2)	6.3	286.1	(281.2-288.9)	7.7	285.3	(278.2-288.5)	10.3
Thigh	173.6	174.6	(170.8-180.9)	10.1	175.4	(171.1-180.1)	9.0	175.3	(172.2-180.8)	8.6
Closed Toe Slippers										
Foot	118.6	118.3	(115.0-120.9)	5.8	120.5	(114.8-127.3)	12.5	123.6	(117.3-129.8)	12.5
Leg/Shank	286.9	285.5	(277.9-289.0)	11.2	285.9	(280.2-289.4)	9.2	286.1	(280.5-289.5)	8.9
Thigh	174.0	175.3	(169.6-180.6)	11.0	176.7	(170.6-186.9)	16.3	175.8	(170.3-184.6)	14.3
Toe Slippers										
Foot	102.1	104.9	(101.1-121.6)	20.5	96.7	(85.2-109.7)	24.5	98.0	(92.8-102.4)	9.5
Leg/Shank	287.2	285.0	(281.4-287.8)	6.4	285.0	(278.2-288.0)	9.8	284.2	(278.2-287.2)	9.0
Thigh	173.8	175.1	(171.1-180.4)	9.3	175.7	(170.7-180.4)	9.8	175.4	(171.0-179.3)	8.3

Table B.5.4: Descriptive statistics for the mean IRED marker movement over each trial for each condition for RA Subject 2.

Segment	Condition									
	Static	Carpet			Dry Vinyl Tile			Wet Vinyl Tile		
		Mean	Range	Movement	Mean	Range	Movement	Mean	Range	Movement
Barefoot										
Foot	111.3	112.6	(106.8-116.7)	9.9	112.9	(105.5-117.2)	11.7	115.3	(110.2-119.0)	8.8
Leg/Shank	323.3	320.4	(313.4-324.2)	10.8	321.1	(316.9-323.6)	6.7	322.1	(318.9-324.6)	5.7
Thigh	133.5	137.7	(118.0-154.3)	36.3	132.7	(113.1-153.4)	40.2	131.8	(116.0-157.2)	41.2
Closed Toe Slippers										
Foot	127.5	123.2	(116.2-127.8)	11.6	124.3	(116.6-128.7)	12.1	125.4	(118.1-129.7)	11.5
Leg/Shank	324.0	320.6	(314.0-325.0)	11.0	321.1	(314.7-324.5)	9.9	320.9	(316.6-324.5)	7.9
Thigh	135.0	135.4	(130.8-143.5)	12.7	134.6	(123.1-147.4)	24.3	135.5	(121.4-148.9)	27.5
Toe Slippers										
Foot	125.0	118.4	(112.2-124.6)	12.4	120.7	(114.8-124.2)	9.4	125.8	(122.0-129.0)	7.0
Leg/Shank	324.4	319.8	(314.9-324.3)	9.4	321.7	(318.1-324.5)	6.3	321.1	(317.9-323.5)	5.6
Thigh	133.8	134.7	(130.6-144.3)	13.7	135.2	(130.7-141.9)	11.2	134.9	(128.2-141.6)	13.4

Proflex Harness Specifications and Approval

Fallright International Pty Ltd
an all Australian company
ACN 009 434 137

Quality Endorsed Company
ISO 9002 QEC6790
Standards Australia

Designers, manufacturers, trainers and exporters of specialist high quality height safety, confined space and rescue equipment.

NATIONALLY RECOGNISED TRAINING

Training Accreditation Council
WESTERN AUSTRALIA

PROFLEX HARNESS

The Proflex Harness (Code No: 011-IMP) has been manufactured and in-house tested to meet the requirements of AS1891.1 – Safety belts and harnesses.

At present the 012 Proflex harness is with Australian Standards being tested for accreditation to AS1891.1.

The only difference between the Proflex harness and our Australian Standards licenced products is the introduction of elastic into the webbing. This only affects the harness when it is not under load, making it far more comfortable to wear. When the webbing is loaded and the elasticity is utilised, the webbing performs to exactly the same specifications as the Australian Standards licenced product.

KEITH HARWOOD
QUALITY MANAGER

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Head Office
42 Belmont Avenue, Belmont, Western Australia 6104
Tel:+61 (08) 9461 2000 Fax:+61 (08) 9461 2001
Email: enquire@fallright.com.au

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If it's Fallright – it's Allright

IQNet

Australian Standard
AS 1891.1 for
Belts & Harnesses
Licence No 1270

Monorail and Trolley Technical Drawings

FORBES RIGBY PTY LTD
engineers • planners • scientists
278 Keira Street, Wollongong NSW 2500
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email: secretary@forbesrigby.com.au
Forbes Rigby Pty Ltd ACN 003 936 981

MEMORANDUM

Total Pages (incl this one): 4
Dispatch by:
Hand: ☐ Mail: ☐ Courier: ☐
Fax: x Fax No: *05 B&G at Uni

28.February 1999

FAXED

PROJECT: **Building 15 Biomechanics Lab**
TO: **Bridget Munro**
FROM: **Neil McKinlay** **FR REF:** **98043-1gen/memo 002**
SUBJECT: **Detail of Connection System of a Mono Rail I beam track to Underside of existing Roof Trusses to Support a Harness Trolley**

Bridget,

Please accept my apologies for the delay in getting my thoughts to you has to how you can connect the "I section" beam to the underside of the existing trusses. I am disappointed with my ability to draw the details in a fashion that your department technical officer will be able to understand.. I think that it would be best to meet with your technical officer to explain these sketches. Alternatively Kon Kudzielko may be able to arrange for a steel fabricator/boilermaker to meet with us in the lab so that the scope of your requirements can be explained and a quotation obtained. I will not be in the office until 3-3-99

The essence of the detail is

- Continuous "I section" beam approx 10.8 m long supported on the underside of the 2 trusses and supported at each end wall on a 75x75x5 angle. The end wall support angles would be coach bolted into each timber stud in the wall framing. The beam would weigh 162 kg so getting it into position, both in terms of length and weight, will need some thinking about. Is it possible to get a beam this long into position?
- I have tried to sketch a hinged plate fixed to the top flange of the I beam. This plate will held in position by a fabricated hinge and two bolts. When the bolts of are tensioned the top flange of I beam will be clamped to the underside of the bottom chord of the trusses. I'm not sure whether a clamping connection is needed at the end wall supports. If it is required a small clamp plate could be fitted after the beam is in position.
- The method to moving the beam to the non use position would be to loosen the clamping bolts at each truss and then slide the beam sideways. To make it easier to slide the beam sideways I suggest that a nylon bearing strips (say 20mm thick) be fixed to the top side of the bottom chord of the trusses.
- At the location where the beam is to be used the nylon bearing strip would be replaced by a 20mm thick steel plate that would make the clamping connection more ridgid.
- A 4 wheeled trolley (steel wheels?, nylon wheels?) needs to be fabricated to run on the beam, to suit the harness and the experiments that are to be carried out.

The attached sketches nominate a 100UC15 universal column section, 100mm high x 100mm wide. It would also be possible to use a 150UB14 which is 150 mm high x 75 mm wide. The

weight of the beam would be similar in each case. I have nominated the 100UC15 as it will be slightly stiffer when the beam is being moved sideways.

We trust that the above is sufficient for your present needs. Please do not hesitate to contact me if you would like me to meet your Technical Officer or a fabricator in the Lab..

Regards
For and on behalf of
FORBES RIGBY PTY LTD

Neil McKinlay

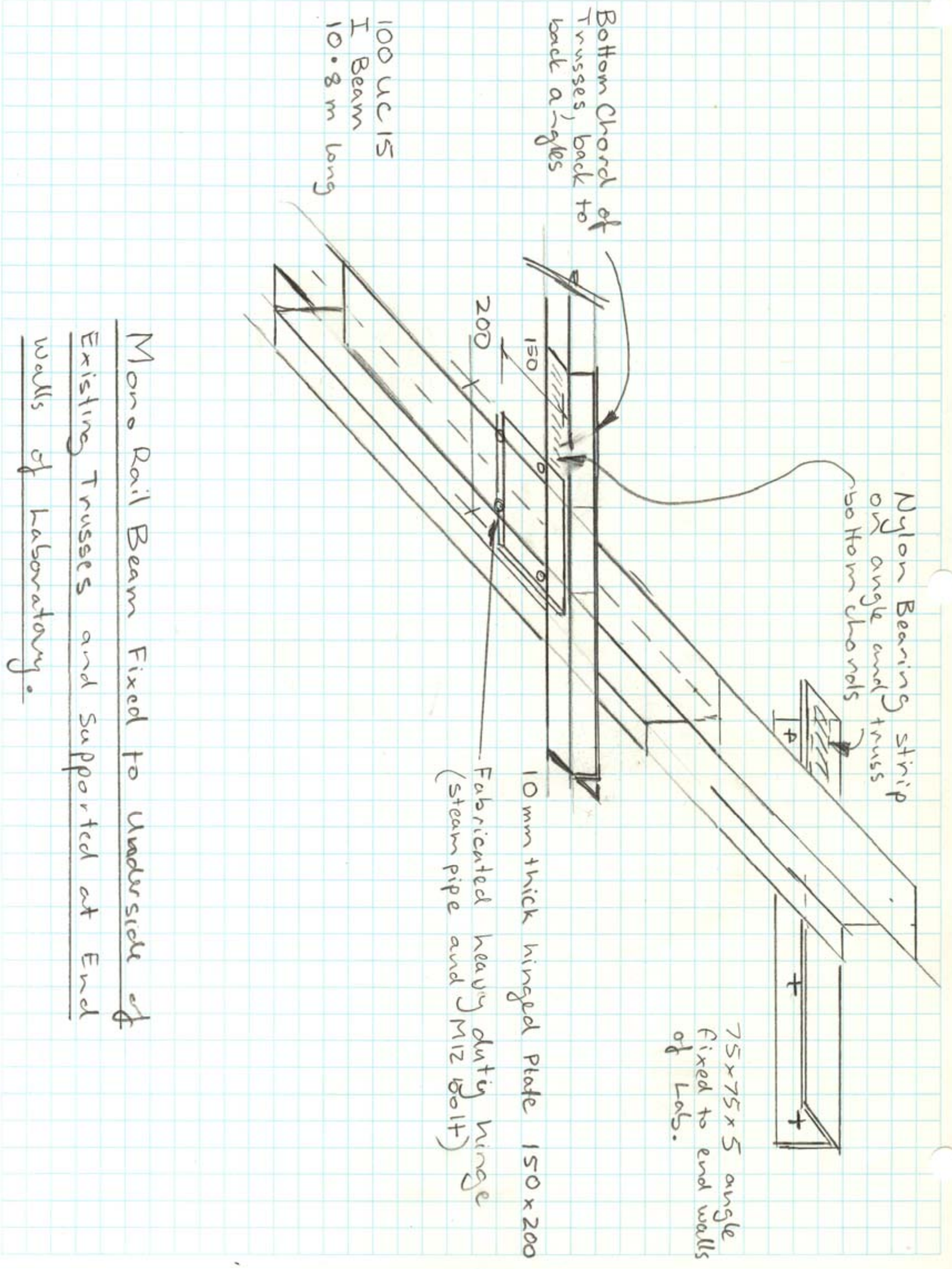
N W McKinlay (Director)

Mr K Kudzielko Uni B&G.

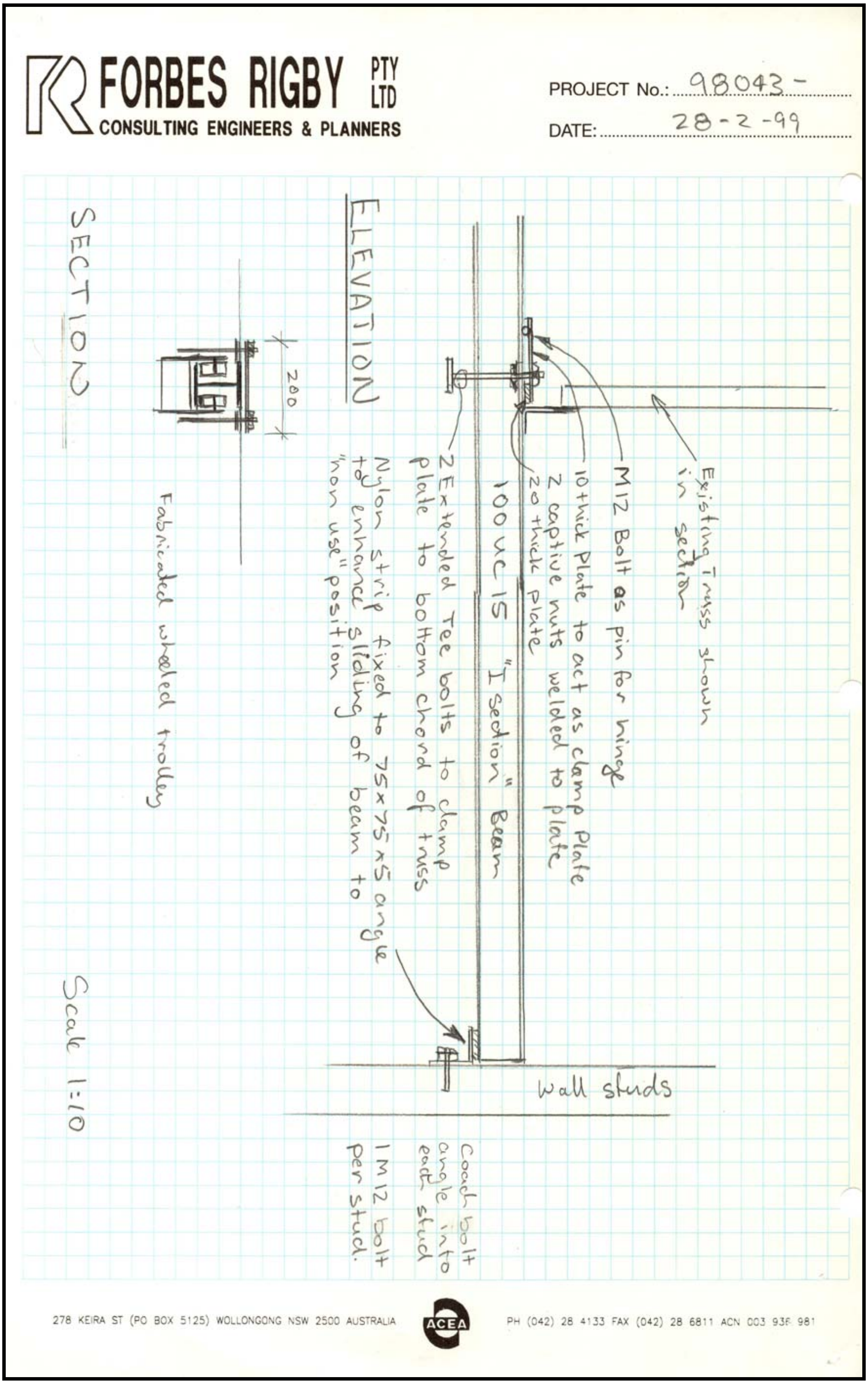
FORBES RIGBY PTY LTD
CONSULTING ENGINEERS & PLANNERS

PROJECT No.: 98043-1

DATE: 28-2-99



278 KEIRA ST (PO BOX 5125) WOLLONGONG NSW 2500 AUSTRALIA ACEA PH (042) 28 4133 FAX (042) 28 6811 ACN 003 936 981


FORBES RIGBY PTY LTD
CONSULTING ENGINEERS & PLANNERS
PROJECT No.: 98043-
DATE: 28-2-99
ELEVATION
SECTION
Existing Truss shown in section
M12 Bolt as pin for hinge
10 thick Plate to act as clamp Plate
2 captive nuts welded to plate
20 thick plate
100 UC 15 "I section" Beam
2 Extended Tee bolts to clamp plate to bottom chord of truss
Nylon strip fixed to 75x75x5 angle to enhance sliding of beam to "non use" position
200
Fabricated wheeled trolley
wall studs
Coach bolt angle into each stud
1 M12 bolt per stud.
Scale 1:10
278 KEIRA ST (PO BOX 5125) WOLLONGONG NSW 2500 AUSTRALIA
ACEA
PH (042) 28 4133 FAX (042) 28 6811 ACN 003 936 981

Monorail Certification Letter (Forbes Rigby Pty Ltd)

FORBES RIGBY PTY LTD
engineers • planners • scientists

OUR REF: 98043-1/gen/cert 001

YOUR REF:

10 November 1999

Ms B Munro
Biomedical Sciences
UNIVERSITY OF WOLLONGONG NSW 2522

Attention: Ms. B Munro

Dear Madam,

INSTALLATION OF MONORAIL BEAM
In BIOMEDICAL SCIENCES LAB
BUILDING 15

CERTIFICATE RE : APPROVAL OF INSTALLATION OF 100UC15 SECTION MONORAIL BEAM TO UNDERSIDE OF BOTTOM CHORD OF EXISTING ROOF TRUSSES
DOCUMENTS REF : FORBES RIGBY MEMO 98043- 002

We hereby certify that we have inspected the above installation. Based on our review of the works we are satisfied that the monorail beam is both safe and serviceable. To maintain maximum flexibility of Laboratory use, the monorail has been fixed to the bottom chord of the existing trusses via a clamping plate system which allows the monorail beam to be moved laterally when not in use.

We recommend that a Monorail capacity notice be posted on the wall of the Laboratory.

"Maximum Monorail Capacity 200 kg
Ensure Beam Clamping System at
each Truss is Tensioned Before use"

Yours faithfully,
for and on behalf of
FORBES RIGBY PTY LTD

Neil McKinlay

N W McKinlay (Director)

Inspection NM 9-11-99 Installation and clamping system

278 KEIRA STREET, WOLLONGONG NSW 2500
PH: (02) 4228 4133 FAX: (02) 4228 6811
email: secretary@forbesrigby.com.au
FORBES RIGBY PTY LTD ACN 003 936 981

directors: E.H. RIGBY BE, MEngSc, FIEAust, MASCE, MACS, CPEng - V.M. WATTS BE, ME(Hons), MIEAust, AIArbA, MEIA, CPEng
N.W. McKINLAY BE,FIEAust,CPEng - P.S. NICHOLS BE(Hons), MEngSc, MIEAust, CPEng - K.N. WILYMAN BSc(Hons), GradCE, MIEAust, CPEng
associates: D.F. LAING BNatRes, Grad Dip URP, MEIA, MRAPI, MNELA - M.C. WELLS BE(Hons), MIEAust, CPEng

PROG Subroutine – Residual Analysis

The following subroutines were used by PROG software[811] to perform residual analysis for cut-off frequencies ranging from 1 Hz to 30 Hz on the x-coordinate and y-coordinate for each IRED[779].

```
[PROGRAMSTART]     // tells PROG to start program at this point

[READ]             // read in appropriate file for analysis
// *** INPUT THE KINEMATIC FILE NAME AND CORRECT DIRECTORIES HERE
    filename="c:\analysis\RA\E8\optotrak\raw#028.txt" // filename where data resides
    filetype="TXT"                        // file type = ASCII TEXT
    columnLabels="T","1A","2A","3A","1B","2B","3B","MTP_Y","MTP_X","MTP_Z",
              "LC_Y","LC_X","LC_Z","PC_Y","PC_X","PC_Z","N_Y","N_X","N_Z",
              "DLL_Y","DLL_X","DLL_Z","DAL_Y","DAL_X","DAL_Z","TT_Y",
              "TT_X","TT_Z","PLL_Y","PLL_X","PLL_Z","DLT_Y","DLT_X","DLT_Z",
              "DAT_Y","DAT_X","DAT_Z","PAT_Y","PAT_X","PAT_Z","PLT_Y","PLT
              _X","PLT_Z","4A","4B","4C","5A","5B","5C"     // label data columns

[DELETEALL]        // deletes everything from memory except for these columns
    exclude="T","MTP_Y","MTP_X","LC_Y","LC_X","PC_Y","PC_X","DLL_Y","DLL_X",
          "PLL_Y","PLL_X","DLT_Y","DLT_X","PLT_Y","PLT_X"

[RESIDUAL]         // calculates the residual analysis protocol of Winter[779]
    columns="T","MTP_Y","MTP_X","LC_Y","LC_X","PC_Y","PC_X","DLL_Y","DLL_X",
          "PLL_Y","PLL_X","DLT_Y","DLT_X","PLT_Y","PLT_X"
                   // tells which columns to use for subroutine
    FREQUENCY=100  // data collection frequency (Hz)
    FREQLOWER=1    // Low pass frequency from 1
    FREQUPPER=30   // Low pass frequency to 30
    FREQGAP=1      // Low pass frequency every 1 from lower to upper

[DELETEALL]        // deletes everything from memory except for these columns
    exclude="MTP_XRA","MTP_YRA","LC_XRA","LC_YRA","PC_XRA","PC_YRA",
          "DLL_XRA","DLL_YRA","PLL_XRA","PLL_YRA","DLT_XRA","DLT_YRA",
          "PLT_XRA","PLT_YRA"

[WRITE]
// *** CHANGE THE FILE NAME THAT YOU WANT TO SAVE IT TO
    filename="c:\analysis\RA\E8\optotrak\raw#028r.txt"
    filetype="TXT"
    delimiterFormat=","
    dataFormat="8.4"

[PROGRAMEND]     // terminates program execution at this point
```

PROG Subroutine – Ground Reaction Force & EMG Analysis

The following subroutines were used by PROG software[811] to sum the ground reaction forces (N), calculate the centre of pressure coordinates, generate linear envelopes to determine muscle burst onset/offset and integrate EMG data. The [INTEGRATE] subroutine could only be used when muscle burst onset/offset as well as initial foot-ground contact and terminal stance had been determined.

```
[PROGRAMSTART]     // tells PROG to start program at this point

[READ]             // read in appropriate file for analysis
// *** INPUT THE FORCE FILE NAME AND CORRECT DIRECTORIES HERE
    filename="c:\analysis\RA\E8\odau\tc5.txt"    // filename where data resides
    filetype="TXT"                               // file type = ASCII TEXT
    columnLabels="T","Y1","Y2","Y3","Y4","X1","X2","Z1","Z2","RF","VL","TA","X","PL",
                    "BF","ST","G"                // labels to title columns of data

[FILTERHIGH]       // high pass filter
    columns="RF","VL","TA","PL","BF","ST","G"
    frequency=1000     // data collection frequency; hertz default = 1
    cutoff=15          // cut off frequency (hertz)
    deleteOriginal=1

[FILTER]           // low pass filter
    columns="RF","VL","TA","PL","BF","ST","G"
    frequency=1000
    cutoff=250
    deleteOriginal=1

[RECTIFY]          // full wave rectification
    columns="RF","VL","TA","PL","BF","ST","G"
    deleteOriginal=0
    tag=1              // default = 0 posttag analysed data; 1 = posttag used data
    posttag="_FIL"     // posttag name

[FILTER]           // generate linear envelopes
    columns="RF","VL","TA","PL","BF","ST","G"
    frequency=1000
    cutoff=10
    deleteOriginal=1

[RECTIFY]          // ensure no negative values - affects threshold detector in DSP
    columns="RF","VL","TA","PL","BF","ST","G"
    deleteOriginal=1

[TRANSLATESCALE]  // zero offset force data
    columns="Y1","Y2","Y3","Y4","X1","X2","Z1","Z2"
    transmode=1
```

```
    row=100
    order=0
    sign=-1
    deleteOriginal=1

[TRANSLATESCALE]  // scale vertical analog force signals from V to N
    columns="Y1","Y2","Y3","Y4"
    transmode=3         // 3 = translate using a constant value defined below
    transconst=0        // constant translation value
    order = 0
    scale=261.0966      // convert voltage data into newtons using the equation below:
// System sensitivity (V/N)   =   Force plate sensitivity (pC) / Amplifier range (pC/10V)
//                            =   3.83 / (10000/10)
//                            =   0.0038
// Force (N)   =   Voltage (V) / System Sensitivity (V/N)
//             =   V / 0.00383
// Scaling factor will only allow multiplication therefore 1 / 0.00383 = 261.0966
    deleteOriginal=1

[TRANSLATESCALE]  // scale horizontal analog force signals from V to N
    columns="X1","X2","Z1","Z2"
    transmode=3
    transconst=0
    order = 0
    scale=62.5          // convert voltage data into newtons using the equation below:
// System Sensitivity (V/N)   =   Force plate sensitivity (pC) / Amplifier range (pC/10V)
//                            =   8 / (5000/10)
//                            =   0.0016
// Force (N)   =   Voltage (V) / System Sensitivity (V/N)
//             =   V / 0.0016
// Scaling factor will only allow multiplication therefore 1 / 0.0016 = 62.5
    deleteOriginal=1

[RECTIFY]               // ensure no negative values - affects threshold detector in DSP
    columns="Y1","Y2","Y3","Y4"
    deleteOriginal=1

[WRITE]                 // save data in column order for further analysis
    columns="RF_FIL","VL_FIL","TA_FIL","PL_FIL","BF_FIL","ST_FIL","G_FIL","RF","V
              L", "TA","PL","BF","ST","G","Y1","Y2","Y3","Y4","X1","X2","Z1","Z2"
    filename="c:\analysis\columns.txt"          // write to filename
    filetype="TXT"                              // output file type = ASCII TEXT
    delimiterFormat=","                         // comma separated file type
    dataFormat="8.4"                            // allows 8 numbers + 4 decimal places

[DELETEALL]             // deletes everything in memory

[READ]
    filename="c:\analysis\columns.txt"
    filetype="TXT"
    LabelsInRow=1       // uses column labels already defined

[KISTLER]               // sums GRF and calculates COP coordinates[779]
// ISB: Y = vertical, X = anteroposterior (long axis), Z = mediolateral (short axis)
```

```
    columns="Y1","Y2","Y3","Y4","X1","X2","Z1","Z2"
    label="KF"            // prefixlabel to summed Z Y X and <label>copX <label>copY
    CentreOfPressure=1 // 0 = default is no; 1 = yes
    Zwidth=120            // half distance between sensors
    Xwidth=200            // half distance between sensors
    Ydepth=-81            // depth = 54 mm + carpet (27 mm) or linoleum (23 mm) height
    Resultant=5           // 5 = XYZ + all the above

[DELETEALL]               // deletes everything from memory except for these columns
    exclude="RF_FIL","VL_FIL","TA_FIL","PL_FIL","BF_FIL","ST_FIL","G_FIL","RF",
          "VL","TA","PL","BF","ST","G","KFY","KFX","KFZ","KFcopZ","KFcopX"

[WRITE]
// *** CHANGE THE FILE NAME THAT YOU WANT TO SAVE IT TO
    filename="c:\analysis\RA\E8\odau\tc5p.txt"
    filetype="TXT"
    delimiterFormat=","
    dataFormat="8.4"

[DELETEALL]

[READ]                    // read in appropriate file for analysis
// *** INPUT THE FORCE FILE NAME AND CORRECT DIRECTORIES HERE
    filename="c:\analysis\RA\E8\odau\tc5p.txt"
    filetype="TXT"                          // file type = ASCII TEXT
    LabelsInRow=1

[INTEGRATE]             // integrate EMG using simpson method[779]
    columns="RF"
    stepwidth=0.001     // e.g. 0.001 for 1 ms or 1 for 1M
    frequency=1         // method=simpson
    row=1428,3421       // rows indicate muscle burst onset, muscle burst offset
    posttag="_SRULE"

[INTEGRATE]
    columns="VL"
    stepwidth=0.001
    frequency=1
    row=1318,2541
    posttag="_SRULE"

[INTEGRATE]
    columns="TA"
    stepwidth=0.001
    frequency=1
    row=953,1653
    posttag="_SRULE"

[INTEGRATE]
    columns="PL"
    stepwidth=0.001
    frequency=1
    row=1429,1965
    posttag="_SRULE"
```

```
[INTEGRATE]
    columns="BF"
    stepwidth=0.001
    frequency=1
    row=1388,2118
    posttag="_SRULE"

[INTEGRATE]
    columns="ST"
    stepwidth=0.001
    frequency=1
    row=1135,2112
    posttag="_SRULE"

[INTEGRATE]
    columns="G"
    stepwidth=0.001
    frequency=1
    row=1429,2277
    posttag="_SRULE"

[INTEGRATE]         // integrate vertical GRF using simpson method[779]
    columns="KFY"
    stepwidth=0.001 // e.g. 0.001 for 1 ms or 1 for 1M
    frequency=1     // method=simpson
    row=1428,3421   // rows indicate initial contact, terminal stance
    posttag="_SRULE"

[DELETEALL]          // deletes everything from memory except for these columns
    exclude="RF_SRULE","VL_SRULE","TA_SRULE","PL_SRULE","BF_SRULE",
            "ST_SRULE","G_SRULE",”KFY_SRULE”

[WRITE]
// *** CHANGE THE FILE NAME THAT YOU WANT TO SAVE IT TO
    filename="c:\analysis\RA\E8\odau\tc5i.txt"
    filetype="TXT"
    delimiterFormat=","
    dataFormat="8.4"

[PROGRAMEND]         // terminates program execution at this point
```

Mauchly's Tests of Sphericity for Subjective Perception, EMG and Kinetic Data

Table B.11.1: Mauchly's sphericity tests for the subjective perception data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Task difficulty	0.379	11.656	2	0.003*	0.617
Foot pain	0.759	3.310	2	0.191	0.806
Shoe comfort	0.487	8.645	2	0.013*	0.661
Slipperiness [a]	0.484	8.699	2	0.013*	0.660
Surface					
Task difficulty	0.224	17.952	2	<0.001*	0.563
Foot pain	0.597	6.197	2	0.045*	0.713
Shoe comfort	0.461	9.296	2	0.010*	0.650
Slipperiness	0.175	20.930	2	<0.001*	0.548
Footwear x Surface					
Task difficulty	0.048	34.767	9	<0.001*	0.504
Foot pain	0.193	18.779	9	0.029*	0.659
Shoe comfort	0.083	28.392	9	0.001*	0.595
Slipperiness	0.001	79.277	9	<0.001*	0.398

[a] Slipperiness = shoe/surface slipperiness.
* Indicates statistical significance at $p \leq 0.05$.

Table B.11.2: Mauchly's sphericity tests for the rectus femoris (RF) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.731	2.816	2	0.245	0.788
Muscle onset (ms)	0.772	2.327	2	0.312	0.814
Muscle onset (%)	0.989	0.100	2	0.951	0.989
Muscle offset (ms)	0.575	4.973	2	0.083	0.702
Muscle offset (%)	0.557	5.273	2	0.072	0.683
IEMG (mV)	0.625	4.228	2	0.121	0.727
IEMG (%)	0.768	2.378	2	0.305	0.812
Surface					
Duration (ms)	0.625	4.224	2	0.121	0.728
Muscle onset (ms)	0.665	3.671	2	0.160	0.749
Muscle onset (%)	0.721	2.944	2	0.230	0.782
Muscle offset (ms)	0.586	4.903	2	0.091	0.707
Muscle offset (%)	0.864	1.313	2	0.519	0.880
IEMG (mV)	0.887	1.084	2	0.582	0.898
IEMG (%)	0.557	5.271	2	0.072	0.693
Footwear x Surface					
Duration (ms)	0.258	11.408	9	0.257	0.635
Muscle onset (ms)	0.022	32.253	9	<0.001*	0.509
Muscle onset (%)	0.024	31.429	9	<0.001*	0.522
Muscle offset (ms)	0.590	4.441	9	0.883	0.808
Muscle offset (%)	0.518	5.535	9	0.790	0.751
IEMG (mV)	0.047	25.807	9	0.003*	0.450
IEMG (%)	0.046	25.955	9	0.002*	0.429

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.3: Mauchly's sphericity tests for the vastus lateralis (VL) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.467	7.608	2	0.022*	0.652
Muscle onset (ms)	0.709	3.445	2	0.179	0.774
Muscle onset (%)	0.574	5.553	2	0.062	0.701
Muscle offset (ms)	0.544	6.082	2	0.048*	0.687
Muscle offset (%)	0.319	11.438	2	0.003*	0.595
IEMG (mV)	0.845	1.683	2	0.431	0.866
IEMG (%)	0.819	1.994	2	0.369	0.847
Surface					
Duration (ms)	0.877	1.309	2	0.520	0.891
Muscle onset (ms)	0.540	6.169	2	0.046*	0.685
Muscle onset (%)	0.397	9.241	2	0.010*	0.624
Muscle offset (ms)	0.745	2.943	2	0.230	0.797
Muscle offset (%)	0.808	2.126	2	0.345	0.839
IEMG (mV)	0.768	2.636	2	0.268	0.812
IEMG (%)	0.804	2.176	2	0.337	0.836
Footwear x Surface					
Duration (ms)	0.165	16.958	9	0.053*	0.611
Muscle onset (ms)	0.020	36.772	9	<0.001*	0.494
Muscle onset (%)	0.018	37.757	9	<0.001*	0.453
Muscle offset (ms)	0.223	14.111	9	0.124	0.649
Muscle offset (%)	0.311	10.992	9	0.283	0.750
IEMG (mV)	0.302	11.290	9	0.263	0.743
IEMG (%)	0.279	12.019	9	0.219	0.734

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.4: Mauchly's sphericity tests for the biceps femoris (BF) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.957	0.437	2	0.804	0.959
Muscle onset (ms)	0.770	2.618	2	0.270	0.813
Muscle onset (%)	0.710	3.429	2	0.180	0.775
Muscle offset (ms)	1.000	0.002	2	0.999	1.000
Muscle offset (%)	0.979	0.211	2	0.900	0.980
IEMG (mV)	0.663	4.110	2	0.128	0.748
IEMG (%)	0.379	9.703	2	0.008*	0.617
Surface					
Duration (ms)	0.846	1.674	2	0.433	0.866
Muscle onset (ms)	0.886	1.214	2	0.545	0.897
Muscle onset (%)	0.882	1.260	2	0.533	0.894
Muscle offset (ms)	0.673	3.962	2	0.138	0.753
Muscle offset (%)	0.893	1.134	2	0.567	0.903
IEMG (mV)	0.808	2.138	2	0.343	0.839
IEMG (%)	0.879	1.295	2	0.523	0.892
Footwear x Surface					
Duration (ms)	0.111	20.743	9	0.015*	0.534
Muscle onset (ms)	0.142	18.378	9	0.033*	0.524
Muscle onset (%)	0.121	19.874	9	0.020*	0.502
Muscle offset (ms)	0.113	20.564	9	0.016*	0.521
Muscle offset (%)	0.031	32.724	9	<0.001*	0.448
IEMG (mV)	0.066	25.575	9	0.003*	0.455
IEMG (%)	0.005	50.283	9	<0.001*	0.327

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.5: Mauchly's sphericity tests for the semitendinosus (S) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.847	1.664	2	0.435	0.867
Muscle onset (ms)	0.572	5.588	2	0.061	0.700
Muscle onset (%)	0.461	7.739	2	0.021*	0.650
Muscle offset (ms)	0.893	1.136	2	0.567	0.903
Muscle offset (%)	0.862	1.489	2	0.475	0.878
IEMG (mV)	0.306	11.852	2	0.003*	0.590
IEMG (%)	0.513	6.678	2	0.035*	0.672
Surface					
Duration (ms)	0.485	7.230	2	0.027*	0.660
Muscle onset (ms)	0.548	6.017	2	0.049*	0.689
Muscle onset (%)	0.466	7.643	2	0.022*	0.652
Muscle offset (ms)	0.633	4.572	2	0.102	0.732
Muscle offset (%)	0.787	2.399	2	0.301	0.824
IEMG (mV)	0.315	11.567	2	0.003*	0.593
IEMG (%)	0.274	12.964	2	0.002*	0.579
Footwear x Surface					
Duration (ms)	0.477	6.978	9	0.645	0.795
Muscle onset (ms)	0.087	23.013	9	0.007*	0.435
Muscle onset (%)	0.195	15.398	9	0.085	0.538
Muscle offset (ms)	0.445	7.624	9	0.579	0.799
Muscle offset (%)	0.406	8.490	9	0.492	0.802
IEMG (mV)	0.117	20.185	9	0.018*	0.588
IEMG (%)	0.108	20.997	9	0.014*	0.600

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.6: Mauchly's sphericity tests for the tibialis anterior (TA) data.

Measure	Mauchly's W	Approx Chi-Square	*Df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.988	0.125	2	0.939	0.988
Muscle onset (ms)	0.857	1.538	2	0.464	0.875
Muscle onset (%)	0.903	1.015	2	0.602	0.912
Muscle offset (ms)	0.987	0.128	2	0.938	0.987
Muscle offset (%)	0.982	0.186	2	0.911	0.982
IEMG (mV)	0.785	2.420	2	0.298	0.823
IEMG (%)	0.293	12.279	2	0.002*	0.586
Surface					
Duration (ms)	0.773	2.580	2	0.275	0.815
Muscle onset (ms)	0.884	1.238	2	0.538	0.896
Muscle onset (%)	0.908	0.968	2	0.616	0.916
Muscle offset (ms)	0.606	5.014	2	0.081	0.717
Muscle offset (%)	0.747	2.913	2	0.233	0.798
IEMG (mV)	0.424	8.572	2	0.014*	0.635
IEMG (%)	0.214	15.405	2	<0.001*	0.560
Footwear x Surface					
Duration (ms)	0.240	13.421	9	0.150	0.621
Muscle onset (ms)	0.157	17.423	9	0.045*	0.551
Muscle onset (%)	0.215	14.495	9	0.111	0.620
Muscle offset (ms)	0.371	9.342	9	0.413	0.713
Muscle offset (%)	0.381	9.076	9	0.437	0.709
IEMG (mV)	0.351	9.850	9	0.370	0.649
IEMG (%)	0.446	7.607	9	0.580	0.767

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.7: Mauchly's sphericity tests for the peroneus longus (PL) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.903	0.915	2	0.633	0.912
Muscle onset (ms)	0.589	4.771	2	0.092	0.708
Muscle onset (%)	0.689	3.353	2	0.187	0.763
Muscle offset (ms)	0.958	0.388	2	0.824	0.959
Muscle offset (%)	0.918	0.771	2	0.680	0.924
IEMG (mV)	0.812	1.877	2	0.391	0.842
IEMG (%)	0.973	0.243	2	0.886	0.974
Surface					
Duration (ms)	0.842	1.551	2	0.460	0.863
Muscle onset (ms)	0.338	9.752	2	0.008*	0.602
Muscle onset (%)	0.294	11.031	2	0.004*	0.586
Muscle offset (ms)	0.699	3.221	2	0.200	0.769
Muscle offset (%)	0.717	2.994	2	0.224	0.779
IEMG (mV)	0.945	0.509	2	0.775	0.948
IEMG (%)	0.434	7.521	2	0.023*	0.638
Footwear x Surface					
Duration (ms)	0.419	7.323	9	0.611	0.713
Muscle onset (ms)	0.060	23.666	9	0.006	0.551
Muscle onset (%)	0.092	20.080	9	0.019*	0.609
Muscle offset (ms)	0.378	8.197	9	0.523	0.705
Muscle offset (%)	0.506	5.727	9	0.772	0.761
IEMG (mV)	0.319	9.620	9	0.391	0.604
IEMG (%)	0.345	8.947	9	0.451	0.635

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.8: Mauchly's sphericity tests for the gastrocnemius (G) data.

Measure	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Duration (ms)	0.728	3.174	2	0.205	0.786
Muscle onset (ms)	0.844	1.693	2	0.429	0.865
Muscle onset (%)	0.799	2.248	2	0.325	0.832
Muscle offset (ms)	0.904	1.014	2	0.602	0.912
Muscle offset (%)	0.530	6.346	2	0.042*	0.680
IEMG (mV)	0.982	0.182	2	0.913	0.982
IEMG (%)	0.977	0.237	2	0.888	0.977
Surface					
Duration (ms)	0.652	4.272	2	0.118	0.742
Muscle onset (ms)	0.493	7.062	2	0.029*	0.664
Muscle onset (%)	0.548	6.019	2	0.049*	0.689
Muscle offset (ms)	0.593	5.222	2	0.073	0.711
Muscle offset (%)	0.995	0.053	2	0.974	0.995
IEMG (mV)	0.559	5.819	2	0.055	0.694
IEMG (%)	0.511	6.720	2	0.035*	0.671
Footwear x Surface					
Duration (ms)	0.571	5.278	9	0.813	0.808
Muscle onset (ms)	0.061	26.341	9	0.002*	0.453
Muscle onset (%)	0.041	29.983	9	0.001*	0.456
Muscle offset (ms)	0.183	16.003	9	0.071	0.555
Muscle offset (%)	0.053	27.632	9	0.001*	0.582
IEMG (mV)	0.300	11.348	9	0.260	0.750
IEMG (%)	0.274	12.178	9	0.210	0.734

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.9: Mauchly's sphericity tests for the vertical ground reaction force data.

Measure [a]	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
Stance (ms)	0.878	0.779	2	0.677	0.891
IMP_y (ms)	0.757	1.671	2	0.434	0.804
IMP_y (N.ms.kg^{-1})	0.869	0.844	2	0.656	0.884
F_{yB} (N)	0.387	5.691	2	0.058	0.620
F_{yB} (N.kg^{-1})	0.457	4.700	2	0.095	0.648
IC-F_{yB} (ms)	0.249	8.334	2	0.016*	0.571
IC-F_{yB} (%)	0.280	7.629	2	0.022*	0.582
F_{yM} (N)	0.976	0.146	2	0.930	0.977
F_{yM} (N.kg^{-1})	0.837	1.070	2	0.586	0.860
IC-F_{yM} (ms)	0.202	9.593	2	0.008*	0.556
IC-F_{yM} (%)	0.107	13.404	2	0.001*	0.528
Surface					
Stance (ms)	0.213	9.276	2	0.010	0.560
IMP_y (ms)	0.670	2.404	2	0.301	0.752
IMP_y (N.ms.kg^{-1})	0.901	0.627	2	0.731	0.910
F_{yB} (N)	0.632	2.753	2	0.252	0.731
F_{yB} (N.kg^{-1})	0.613	2.940	2	0.230	0.721
IC-F_{yB} (ms)	0.021	23.127	2	<0.001*	0.505
IC-F_{yB} (%)	0.034	20.206	2	<0.001*	0.509
F_{yM} (N)	0.786	1.442	2	0.486	0.824
F_{yM} (N.kg^{-1})	0.855	0.939	2	0.625	0.873
IC-F_{yM} (ms)	0.938	0.382	2	0.826	0.942
IC-F_{yM} (%)	0.755	1.689	2	0.430	0.803
Footwear x Surface					
Stance (ms)	0.005	28.919	9	0.001*	0.412
IMP_y (ms)	0.145	10.457	9	0.336	0.616
IMP_y (N.ms.kg^{-1})	0.166	9.721	9	0.395	0.625
F_{yB} (N)	0.005	28.343	9	0.001*	0.314
F_{yB} (N.kg^{-1})	0.006	27.338	9	0.002*	0.315
IC-F_{yB} (ms)	<0.001	47.941	9	<0.001*	0.269
IC-F_{yB} (%)	<0.001	52.882	9	<0.001*	0.267
F_{yM} (N)	0.036	18.008	9	0.043*	0.495
F_{yM} (N.kg^{-1})	0.036	17.999	9	0.043*	0.489
IC-F_{yM} (ms)	0.011	24.484	9	0.005*	0.338
IC-F_{yM} (%)	0.014	23.081	9	0.008*	0.356

[a] See Table 5.4 for description of variables.

* Indicates statistical significance at $p \leq 0.05$.

Table B.11.10: Mauchly's sphericity tests for the anteroposterior ground reaction force data.

Measure [a]	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
Footwear					
F_{xB} (N)	0.795	2.297	2	0.317	0.830
F_{xB} (N.kg^{-1})	0.727	3.193	2	0.203	0.785
IC-F_{xB} (ms)	0.417	8.738	2	0.013*	0.632
IC-F_{xB} (% stance)	0.322	11.338	2	0.003*	0.596
IC-F_{xC} (ms)	0.208	15.720	2	<0.001*	0.558
IC-F_{xC} (% stance)	0.169	17.759	2	<0.001*	0.546
Surface					
F_{xB} (N)	0.480	7.330	2	0.026*	0.658
F_{xB} (N.kg^{-1})	0.472	7.504	2	0.023*	0.655
IC-F_{xB} (ms)	0.417	8.738	2	0.013*	0.632
IC-F_{xB} (% stance)	0.322	11.338	2	0.003*	0.596
IC-F_{xC} (ms)	0.087	24.473	2	<0.001*	0.582
IC-F_{xC} (% stance)	0.063	27.720	2	<0.001*	0.516
Footwear x Surface					
F_{xB} (N)	0.379	9.142	9	0.431	0.649
F_{xB} (N.kg^{-1})	0.337	10.241	9	0.339	0.611
IC-F_{xB} (ms)	0.417	8.738	9	0.013*	0.632
IC-F_{xB} (% stance)	0.322	11.338	9	0.003*	0.596
IC-F_{xC} (ms)	0.000	88.184	9	<0.001*	0.273
IC-F_{xC} (% stance)	0.000	97.033	9	<0.001*	0.266

[a] See Table 5.4 for description of variables.
* Indicates statistical significance at $p \leq 0.05$.

Table B.11.11: Mauchly's sphericity tests for the coefficient of friction data.

Measure [a]	Mauchly's W	Approx Chi-Square	*df*	*p*-value	Greenhouse-Geisser Epsilon
μ_S	0.003	5.804	2	0.055	0.501
μ_D	0.442	0.817	2	0.665	0.642
IC-μ_D (ms)	0.251	1.384	2	0.501	0.572
IC-μ_D (%)	0.972	0.029	2	0.986	0.972
Surface					
μ_S	0.385	0.956	2	0.620	0.619
μ_D	0.022	3.797	2	0.150	0.506
IC-μ_D (ms)	0.401	0.913	2	0.633	0.626
IC-μ_D (%)	0.432	0.839	2	0.658	0.638
Footwear x Surface					
μ_S	0.288	6.740	9	0.680	0.698
μ_D	0.005	12.916	9	0.258	0.320
IC-μ_D (ms)	<0.001	23.177	9	0.017*	0.288
IC-μ_D (%)	<0.001	19.433	9	0.050*	0.352

[a] See Table 5.4 for description of variables.
* Indicates statistical significance at $p \leq 0.05$.

Three-way Repeated Measures ANOVA Results for Subjective Perception, EMG and Kinetic Data

Table B.12.1: Three-way repeated measures ANOVA results for the subjective perception data.

Source	Sum Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Task difficulty	5379.5	1.233†	4361.2	3.94	0.057
Foot pain	334.7	2	167.3	3.11	0.061
Shoe comfort	6141.6	1.321†	4647.5	5.43	0.025*
Slipperiness	4201.9	1.320†	3184.3	4.49	0.040*
Footwear x Group					
Task difficulty	2530.1	1.233†	2051.2	1.86	0.193
Foot pain	334.7	2	167.3	3.11	0.061
Shoe comfort	529.4	1.321†	400.6	0.47	0.556
Slipperiness	843.9	1.320†	639.6	0.90	0.384
Error (Footwear)					
Task difficulty	17729.9	16.035†	1105.7		
Foot pain	1397.9	26	53.8		
Shoe comfort	14707.9	17.179†	856.1		
Slipperiness	12158.7	17.154†	708.8		
Surface					
Task difficulty	3241.1	1.126†	2878.1	9.26	0.007*
Foot pain	140.4	1.425†	98.6	0.87	0.403
Shoe comfort	2481.8	1.299†	1910.0	3.35	0.076
Slipperiness	13718.1	1.096†	12519.2	19.01	<0.001*
Surface x Group					
Task difficulty	440.5	1.126†	391.2	1.26	0.287
Foot pain	140.5	1.425†	98.6	0.87	0.403
Shoe comfort	78.5	1.299†	60.4	0.11	0.813
Slipperiness	805.0	1.096†	734.6	1.12	0.315
Error (Surface)					
Task difficulty	4548.4	14.640†	310.7		
Foot pain	2110.8	18.527†	113.9		
Shoe comfort	9634.1	16.893†	570.3		
Slipperiness	9382.8	14.245†	658.7		
Footwear x Surface					
Task difficulty	2289.6	2.018†	1134.6	6.72	0.004*
Foot pain	140.1	2.635†	53.2	2.03	0.135
Shoe comfort	4493.5	2.382†	1886.6	3.91	0.025*
Slipperiness	8951.7	1.593†	5620.5	21.83	<0.001*
Footwear x Surface x Group					
Task difficulty	790.5	2.018†	391.7	2.32	0.118
Foot pain	140.1	2.635†	53.2	2.03	0.135
Shoe comfort	1396.6	2.382†	586.4	1.22	0.315
Slipperiness	233.6	1.593†	146.7	0.57	0.536
Error (Footwear x Surface)					
Task difficulty	4430.2	26.234†	168.9		
Foot pain	897.2	34.255†	26.2		
Shoe comfort	14936.3	30.963†	482.4		
Slipperiness	5331.6	20.705†	257.5		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.1, Appendix B.11).

* Indicates statistical significance at $p \leq 0.05$.

Table B.12.2: Three-way repeated measures ANOVA results for the rectus femoris (RF) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	34250.2	2	17125.1	0.80	0.462
Muscle onset (ms)	7664.6	2	3832.3	2.72	0.090
Muscle onset (%)	<0.1	2	<0.1	3.98	0.035*
Muscle offset (ms)	7920.6	2	3960.3	0.19	0.826
Muscle offset (%)	<0.1	2	<0.1	0.72	0.498
IEMG (mV)	37.2	2	18.6	2.62	0.097
IEMG (%)	1803.9	2	901.9	2.54	0.104
Footwear x Group					
Duration (ms)	38238.1	2	19119.1	0.90	0.424
Muscle onset (ms)	2779.9	2	1389.9	0.99	0.390
Muscle onset (%)	<0.1	2	<0.1	1.00	0.386
Muscle offset (ms)	53852.8	2	26926.4	1.31	0.292
Muscle offset (%)	<0.1	2	<0.1	0.39	0.680
IEMG (mV)	20.7	2	10.4	1.46	0.256
IEMG (%)	871.6	2	435.8	1.23	0.315
Error (Footwear)					
Duration (ms)	426401.2	20	21320.1		
Muscle onset (ms)	28139.0	20	1406.9		
Muscle onset (%)	<0.1	20	<0.1		
Muscle offset (ms)	411646.5	20	20582.3		
Muscle offset (%)	0.9	20	<0.1		
IEMG (mV)	141.9	20	7.1		
IEMG (%)	7116.4	20	355.8		
Surface					
Duration (ms)	217361.3	2	108680.7	7.30	0.004*
Muscle onset (ms)	5872.9	2	2936.5	1.15	0.337
Muscle onset (%)	<0.1	2	<0.1	0.98	0.394
Muscle offset (ms)	152645.1	2	76322.6	6.21	0.008*
Muscle offset (%)	0.2	2	<0.1	6.72	0.006*
IEMG (mV)	59.4	2	29.7	4.78	0.020*
IEMG (%)	6132.8	2	3066.4	5.68	0.011*
Surface x Group					
Duration (ms)	14884.4	2	7442.2	0.50	0.614
Muscle onset (ms)	1965.9	2	982.9	0.39	0.685
Muscle onset (%)	<0.1	2	<0.1	0.45	0.646
Muscle offset (ms)	30350.3	2	15175.2	1.24	0.312
Muscle offset (%)	<0.1	2	<0.1	1.89	0.177
IEMG (mV)	8.4	2	4.2	0.68	0.520
IEMG (%)	869.2	2	434.6	0.81	0.461
Error (Surface)					
Duration (ms)	297732.5	20	14886.6		
Muscle onset (ms)	51084.2	20	2554.2		
Muscle onset (%)	<0.1	20	<0.1		
Muscle offset (ms)	245670.7	20	12283.5		
Muscle offset (%)	0.3	20	<0.1		
IEMG (mV)	124.3	20	6.2		
IEMG (%)	10793.2	20	539.7		
Footwear x Surface					
Duration (ms)	113019.1	4	28254.8	5.83	0.001*
Muscle onset (ms)	10151.3	2.034†	4990.7	1.82	0.186
Muscle onset (%)	<0.1	2.086†	<0.1	1.25	0.309
Muscle offset (ms)	57901.3	4	14477.6	4.07	0.007*

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Table B.12.2: Three-way repeated measures ANOVA results for the rectus femoris (RF) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Muscle offset (%)	<0.1	4	<0.1	1.92	0.126
IEMG (mV)	61.2	1.801†	34.0	5.08	0.020*
IEMG (%)	5395.4	1.715†	3146.1	7.77	0.005*
Footwear x Surface x Group					
Duration (ms)	24266.4	4	6066.6	1.25	0.305
Muscle onset (ms)	2892.7	2.034†	1422.1	0.52	0.605
Muscle onset (%)	<0.1	2.086†	<0.1	0.57	0.583
Muscle offset (ms)	25138.9	4	6284.7	1.78	0.154
Muscle offset (%)	<0.1	4	<0.1	2.36	0.069
IEMG (mV)	10.9	1.801†	6.1	0.91	0.412
IEMG (%)	490.2	1.715†	285.9	0.71	0.487
Error (Footwear x Surface)					
Duration (ms)	193941.2	40	4848.5		
Muscle onset (ms)	55644.2	20.341†	2735.6		
Muscle onset (%)	0.1	20.864†	<0.1		
Muscle offset (ms)	142236.2	40	3555.9		
Muscle offset (%)	0.3	40	<0.1		
IEMG (mV)	120.6	18.009†	6.7		
IEMG (%)	6943.0	17.140†	404.9		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.5, Appendix B.11).

* Indicates statistical significance at $p \leq 0.05$.

Table B.12.3: Three-way repeated measures ANOVA results for the vastus lateralis (VL) data.

Source	Sum of Squares	*Df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	23674.9	1.305†	18143.4	2.57	0.125
Muscle onset (ms)	7163.2	2	3581.6	5.12	0.015*
Muscle onset (%)	<0.1	2	<0.1	5.69	0.010*
Muscle offset (ms)	10818.4	1.374†	7874.0	0.92	0.368
Muscle offset (%)	<0.1	1.189†	<0.1	2.42	0.141
IEMG (mV)	26.3	2	13.2	1.37	0.275
IEMG (%)	558.9	2	279.5	1.41	0.265
Footwear x Group					
Duration (ms)	41962.7	1.305†	32158.4	4.55	0.042*
Muscle onset (ms)	81.6	2	40.8	0.06	0.944
Muscle onset (%)	<0.1	2	<0.1	0.60	0.556
Muscle offset (ms)	40874.6	1.374†	29750.0	3.46	0.072
Muscle offset (%)	<0.1	1.189†	<0.1	1.76	0.210
IEMG (mV)	67.3	2	33.6	3.50	0.048*
IEMG (%)	882.8	2	441.4	2.23	0.131
Error (Footwear)					
Duration (ms)	101484.5	14.354†	7070.3		
Muscle onset (ms)	15396.9	22	699.9		
Muscle onset (%)	<0.1	22	<0.1		
Muscle offset (ms)	130057.9	15.113†	8605.5		
Muscle offset (%)	0.3	13.084†	<0.1		
IEMG (mV)	211.4	22	9.6		
IEMG (%)	4358.1	22	198.1		

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Table B.12.3: Three-way repeated measures ANOVA results for the vastus lateralis (VL) data.

Source	Sum of Squares	*Df*	Mean Square	*F*-value	*p*-value
Surface					
Duration (ms)	71131.6	2	35565.8	3.76	0.040*
Muscle onset (ms)	1592.4	1.369†	1162.8	0.50	0.547
Muscle onset (%)	<0.1	1.248†	<0.1	0.34	0.618
Muscle offset (ms)	35542.8	2	17771.9	2.15	0.141
Muscle offset (%)	<0.1	2	<0.1	2.11	0.146
IEMG (mV)	39.7	2	19.8	3.02	0.070
IEMG (%)	1323.5	2	661.8	2.70	0.089
Surface x Group					
Duration (ms)	4040.4	2	2020.2	0.21	0.810
Muscle onset (ms)	1981.7	1.369†	1447.0	0.62	0.491
Muscle onset (%)	<0.1	1.248†	<0.1	0.73	0.439
Muscle offset (ms)	4608.8	2	2304.4	0.28	0.760
Muscle offset (%)	<0.1	2	<0.1	0.44	0.652
IEMG (mV)	11.9	2	6.0	0.91	0.418
IEMG (%)	438.1	2	219.1	0.89	0.423
Error (Surface)					
Duration (ms)	208308.7	22	9468.6		
Muscle onset (ms)	35015.6	15.064†	2324.4		
Muscle onset (%)	<0.1	13.723†	<0.1		
Muscle offset (ms)	182045.3	22	8274.8		
Muscle offset (%)	0.4	22	<0.1		
IEMG (mV)	144.8	22	6.6		
IEMG (%)	5388.1	22	244.9		
Footwear x Surface					
Duration (ms)	54414.0	2.444†	22260.8	2.03	0.144
Muscle onset (ms)	6604.6	1.974†	3345.2	2.09	0.149
Muscle onset (%)	<0.1	1.811†	<0.1	1.61	0.226
Muscle offset (ms)	18152.6	4	4538.1	0.77	0.549
Muscle offset (%)	<0.1	4	<0.1	0.64	0.635
IEMG (mV)	46.4	4	11.6	2.92	0.032*
IEMG (%)	1425.8	4	356.4	3.39	0.017*
Footwear x Surface x Group					
Duration (ms)	52741.3	2.444†	21576.5	1.96	0.153
Muscle onset (ms)	3964.8	1.974†	2008.1	1.25	0.305
Muscle onset (%)	<0.1	1.811†	<0.1	1.15	0.331
Muscle offset (ms)	29482.6	4	7370.6	1.25	0.302
Muscle offset (%)	<0.1	4	<0.1	1.38	0.258
IEMG (mV)	32.8	4	8.2	2.07	0.101
IEMG (%)	598.0	4	149.5	1.42	0.243
Error (Footwear x Surface)					
Duration (ms)	295547.3	26.888†	10991.7		
Muscle onset (ms)	34805.5	21.718†	1602.6		
Muscle onset (%)	<0.1	19.916†	<0.1		
Muscle offset (ms)	258589.6	44	5877.0		
Muscle offset (%)	0.5	44	<0.1		
IEMG (mV)	174.8	44	4.0		
IEMG (%)	4631.4	44	105.3		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.6, Appendix B.11).

* Indicates statistical significance at $p \leq 0.05$.

Table B.12.4: Three-way repeated measures ANOVA results for the biceps femoris (BF) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	162595.8	2	81297.9	3.55	0.046*
Muscle onset (ms)	14667.0	2	7333.5	1.05	0.368
Muscle onset (%)	<0.1	2	<0.1	1.56	0.234
Muscle offset (ms)	62949.3	2	31474.7	1.84	0.182
Muscle offset (%)	0.2	2	0.1	2.69	0.090
IEMG (mV)	387.6	2	193.8	0.66	0.525
IEMG (%)	1904.2	1.234†	1543.4	0.74	0.433
Footwear x Group					
Duration (ms)	4204.8	2	2102.4	0.09	0.913
Muscle onset (ms)	24544.2	2	12272.1	1.75	0.197
Muscle onset (%)	<0.1	2	<0.1	1.24	0.309
Muscle offset (ms)	24895.6	2	12447.8	0.73	0.494
Muscle offset (%)	<0.1	2	<0.1	0.86	0.439
IEMG (mV)	22.4	2	11.1	0.04	0.963
IEMG (%)	1282.6	1.234†	1039.6	0.50	0.531
Error (Footwear)					
Duration (ms)	503422.6	22	22882.8		
Muscle onset (ms)	154188.9	22	7008.6		
Muscle onset (%)	0.4	22	<0.1		
Muscle offset (ms)	376104.2	22	17095.6		
Muscle offset (%)	0.8	22	<0.1		
IEMG (mV)	6430.0	22	292.3		
IEMG (%)	28328.6	13.571†	2087.4		
Surface					
Duration (ms)	7032.4	2	3516.2	0.18	0.833
Muscle onset (ms)	16102.0	2	8051.0	2.30	0.124
Muscle onset (%)	<0.1	2	<0.1	1.35	0.281
Muscle offset (ms)	966.4	2	483.2	0.04	0.961
Muscle offset (%)	<0.1	2	<0.1	0.14	0.874
IEMG (mV)	305.8	2	152.9	1.12	0.344
IEMG (%)	290.2	2	145.1	0.43	0.657
Surface x Group					
Duration (ms)	31165.0	2	15582.5	0.82	0.455
Muscle onset (ms)	467.6	2	233.8	0.07	0.936
Muscle onset (%)	<0.1	2	<0.1	0.06	0.944
Muscle offset (ms)	20370.5	2	10185.2	0.84	0.447
Muscle offset (%)	<0.1	2	<0.1	1.19	0.323
IEMG (mV)	789.5	2	394.7	2.90	0.077
IEMG (%)	1584.1	2	792.0	2.34	0.120
Error (Surface)					
Duration (ms)	420538.4	22	19115.4		
Muscle onset (ms)	77163.7	22	3507.4		
Muscle onset (%)	0.2	22	<0.1		
Muscle offset (ms)	268232.8	22	12192.4		
Muscle offset (%)	0.4	22	<0.1		
IEMG (mV)	2999.7	22	136.3		
IEMG (%)	7448.7	22	338.6		
Footwear x Surface					
Duration (ms)	58621.4	2.137†	27425.6	0.62	0.559
Muscle onset (ms)	18747.1	2.098†	8937.3	1.07	0.364
Muscle onset (%)	<0.1	2.008†	<0.1	1.47	0.252
Muscle offset (ms)	61204.8	2.084†	29362.3	1.04	0.374

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Table B.12.4: Three-way repeated measures ANOVA results for the biceps femoris (BF) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Muscle offset (%)	0.1	1.793†	<0.1	0.79	0.455
IEMG (mV)	822.4	1.818†	452.3	0.61	0.541
IEMG (%)	6601.9	1.309†	5042.0	1.02	0.353
Footwear x Surface x Group					
Duration (ms)	55726.0	2.137†	26071.0	0.59	0.576
Muscle onset (ms)	23934.2	2.098†	11410.1	1.36	0.277
Muscle onset (%)	<0.1	2.008†	<0.1	1.03	0.374
Muscle offset (ms)	38140.1	2.084†	18297.3	0.65	0.540
Muscle offset (%)	<0.1	1.793†	<0.1	0.50	0.597
IEMG (mV)	1263.6	1.818†	695.0	0.93	0.403
IEMG (%)	6443.4	1.309†	4920.9	1.00	0.359
Error (Footwear x Surface)					
Duration (ms)	1047127.8	23.512†	44535.5		
Muscle onset (ms)	193623.7	23.074†	8391.5		
Muscle onset (%)	0.4	22.083†	<0.1		
Muscle offset (ms)	649866.1	22.929†	28342.4		
Muscle offset (%)	1.5	19.723†	<0.1		
IEMG (mV)	14959.2	19.999†	748.0		
IEMG (%)	71249.7	14.403†	4946.8		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.7, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

Table B.12.5: Three-way repeated measures ANOVA results for the semitendinosus (S) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	1825.3	2	912.6	0.03	0.972
Muscle onset (ms)	859.6	2	429.8	0.07	0.936
Muscle onset (%)	<0.1	1.300†	<0.1	0.35	0.621
Muscle offset (ms)	7596.3	2	3798.1	0.19	0.829
Muscle offset (%)	<0.1	2	<0.1	0.01	0.990
IEMG (mV)	28.8	1.180†	24.4	0.10	0.800
IEMG (%)	385.4	1.345†	286.6	0.17	0.759
Footwear x Group					
Duration (ms)	58668.2	2	29334.1	0.91	0.419
Muscle onset (ms)	3490.4	2	1745.2	0.27	0.766
Muscle onset (%)	<0.1	1.300†	<0.1	0.23	0.702
Muscle offset (ms)	56031.6	2	28015.8	1.39	0.270
Muscle offset (%)	<0.1	2	<0.1	1.09	0.354
IEMG (mV)	456.0	1.180†	386.3	1.55	0.240
IEMG (%)	3765.8	1.345†	2800.2	1.66	0.223
Error (Footwear)					
Duration (ms)	713301.5	22	32422.8		
Muscle onset (ms)	142614.6	22	6482.5		
Muscle onset (%)	0.2	14.297†	<0.1		
Muscle offset (ms)	442891.7	22	20131.4		
Muscle offset (%)	0.9	22	<0.1		
IEMG (mV)	3234.7	12.985†	249.1		
IEMG (%)	24979.0	14.793†	1688.5		

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Table B.12.5: Three-way repeated measures ANOVA results for the semitendinosus (S) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Surface					
Duration (ms)	26324.0	1.320†	19936.7	0.76	0.432
Muscle onset (ms)	269.1	1.377†	195.4	0.04	0.904
Muscle onset (%)	<0.1	1.303†	<0.1	0.48	0.551
Muscle offset (ms)	20640.1	2	10320.0	0.90	0.422
Muscle offset (%)	<0.1	2	<0.1	1.75	0.197
IEMG (mV)	266.1	1.187†	224.2	5.57	0.030*
IEMG (%)	4226.3	1.158†	3648.3	6.11	0.025*
Surface x Group					
Duration (ms)	16530.0	1.320†	12519.2	0.48	0.551
Muscle onset (ms)	11724.3	1.377†2	8512.4	1.90	0.189
Muscle onset (%)	<0.1	1.303†	<0.1	2.25	0.152
Muscle offset (ms)	1339.0	2	669.5	0.06	0.944
Muscle offset (%)	<0.1	2	<0.1	0.02	0.984
IEMG (mV)	15.6	1.187†	13.2	0.33	0.615
IEMG (%)	161.6	1.158†	139.5	0.23	0.672
Error (Surface)					
Duration (ms)	378811.9	14.524†	26081.5		
Muscle onset (ms)	68054.9	15.150†	4491.9		
Muscle onset (%)	0.1	14.338†	<0.1		
Muscle offset (ms)	253198.7	22	11509.0		
Muscle offset (%)	0.4	22	<0.1		
IEMG (mV)	525.4	13.053†	40.2		
IEMG (%)	7613.5	12.743†	597.5		
Footwear x Surface					
Duration (ms)	57241.2	4	14310.3	0.91	0.467
Muscle onset (ms)	15229.8	1.740†	8752.2	1.28	0.297
Muscle onset (%)	<0.1	4	<0.1	2.26	0.078
Muscle offset (ms)	22777.7	4	5694.4	0.47	0.755
Muscle offset (%)	<0.1	4	<0.1	0.63	0.642
IEMG (mV)	211.0	2.351†	89.7	2.92	0.065
IEMG (%)	3143.7	2.400†	1310.0	3.46	0.039*
Footwear x Surface x Group					
Duration (ms)	83139.4	4	20784.9	1.32	0.277
Muscle onset (ms)	787.5	1.740†	452.6	0.07	0.916
Muscle onset (%)	<0.1	4	<0.1	0.06	0.992
Muscle offset (ms)	59644.0	4	14911.0	1.24	0.308
Muscle offset (%)	<0.1	4	<0.1	1.04	0.397
IEMG (mV)	11.6	2.351†	4.9	0.16	0.884
IEMG (%)	161.3	2.400†	67.2	0.18	0.874
Error (Footwear x Surface)					
Duration (ms)	692786.7	44	15745.2		
Muscle onset (ms)	131221.8	19.141†	6855.4		
Muscle onset (%)	0.2	44	<0.1		
Muscle offset (ms)	529572.6	44	12035.7		
Muscle offset (%)	1.0	44	<0.1		
IEMG (mV)	796.1	25.864†	30.8		
IEMG (%)	10003.5	26.397†	379.0		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.8, Appendix B.11).

* Indicates statistical significance at $p \leq 0.05$.

Table B.12.6: Three-way repeated measures ANOVA results for the tibialis anterior (TA) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	262729.8	2	131364.9	3.27	0.057
Muscle onset (ms)	314183.4	2	157091.7	5.85	0.009*
Muscle onset (%)	0.6	2	0.3	5.79	0.010*
Muscle offset (ms)	23227.0	2	11613.5	1.71	0.204
Muscle offset (%)	<0.1	2	<0.1	0.83	0.449
IEMG (mV)	10577.2	2	5288.6	6.12	0.008*
IEMG (%)	75772.2	1.172†	64674.5	5.26	0.035*
Footwear x Group					
Duration (ms)	23060.0	2	11530.0	0.29	0.754
Muscle onset (ms)	41720.9	2	20860.5	0.78	0.472
Muscle onset (%)	0.1	2	<0.1	1.11	0.348
Muscle offset (ms)	27794.7	2	13897.3	2.04	0.153
Muscle offset (%)	<0.1	2	<0.1	1.28	0.298
IEMG (mV)	534.9	2	267.5	0.31	0.737
IEMG (%)	3506.6	1.172†	2993.0	0.24	0.668
Error (Footwear)					
Duration (ms)	885203.7	22	40236.5		
Muscle onset (ms)	590939.5	22	26860.9		
Muscle onset (%)	1.2	22	<0.1		
Muscle offset (ms)	149606.0	22	6800.3		
Muscle offset (%)	0.3	22	<0.1		
IEMG (mV)	19009.4	22	864.1		
IEMG (%)	158471.6	12.888†	12296.5		
Surface					
Duration (ms)	95231.6	2	47615.8	1.43	0.261
Muscle onset (ms)	4383.9	2	2191.9	0.16	0.851
Muscle onset (%)	<0.1	2	<0.1	0.25	0.783
Muscle offset (ms)	77106.4	2	38552.2	2.94	0.074
Muscle offset (%)	0.2	2	<0.1	4.81	0.018*
IEMG (mV)	1620.3	1.269†	1276.5	1.33	0.279
IEMG (%)	3203.3	1.120†	2860.1	0.33	0.600
Surface x Group					
Duration (ms)	84100.6	2	42050.3	1.26	0.303
Muscle onset (ms)	18800.8	2	9400.4	0.70	0.509
Muscle onset (%)	<0.1	2	<0.1	0.45	0.641
Muscle offset (ms)	86330.5	2	43165.2	3.29	0.056
Muscle offset (%)	0.2	2	<0.1	4.41	0.024*
IEMG (mV)	1733.7	1.269†	1365.9	1.43	0.262
IEMG (%)	7068.1	1.120†	6310.8	0.73	0.424
Error (Surface)					
Duration (ms)	732518.3	22	33296.3		
Muscle onset (ms)	296849.9	22	13493.2		
Muscle onset (%)	0.6	22	<0.1		
Muscle offset (ms)	288632.6	22	13119.7		
Muscle offset (%)	0.4	22	<0.1		
IEMG (mV)	13386.5	13.962†	958.7		
IEMG (%)	106445.2	12.320†	8640.1		
Footwear x Surface					
Duration (ms)	89762.9	4	22440.7	1.10	0.367
Muscle onset (ms)	34041.5	2.204†	15443.7	0.95	0.407
Muscle onset (%)	<0.1	4	<0.1	1.03	0.403
Muscle offset (ms)	10722.1	4	2680.5	0.31	0.872

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Table B.12.6: Three-way repeated measures ANOVA results for the tibialis anterior (TA) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear x Surface (continued)					
Muscle offset (%)	<0.1	4	<0.1	0.15	0.961
IEMG (mV)	1095.4	4	273.8	1.58	0.196
IEMG (%)	6921.0	4	1730.3	1.63	0.185
Footwear x Surface x Group					
Duration (ms)	28557.2	4	7139.3	0.35	0.842
Muscle onset (ms)	7525.9	2.204†	3413.8	0.21	0.831
Muscle onset (%)	<0.1	4	<0.1	0.19	0.942
Muscle offset (ms)	25469.6	4	6367.4	0.73	0.577
Muscle offset (%)	<0.1	4	<0.1	0.67	0.614
IEMG (mV)	246.7	4	61.7	0.36	0.838
IEMG (%)	3903.9	4	976.0	0.92	0.463
Error (Footwear x Surface)					
Duration (ms)	894377.5	44	20326.8		
Muscle onset (ms)	392846.8	24.247†	16202.2		
Muscle onset (%)	0.7	44	<0.1		
Muscle offset (ms)	384377.7	44	8735.9		
Muscle offset (%)	0.7	44	<0.1		
IEMG (mV)	7610.4	44	173.0		
IEMG (%)	46855.4	44	1064.9		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.9, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

Table B.12.7: Three-way repeated measures ANOVA results for the peroneus longus (PL) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	308737.2	2	154368.6	3.74	0.042*
Muscle onset (ms)	204743.9	2	102371.9	8.59	0.002*
Muscle onset (%)	0.4	2	0.2	7.55	0.004*
Muscle offset (ms)	18648.4	2	9324.2	0.28	0.756
Muscle offset (%)	<0.1	2	<0.1	0.27	0.768
IEMG (mV)	388.9	2	194.5	0.62	0.549
IEMG (%)	16558.1	2	8279.0	1.04	0.372
Footwear x Group					
Duration (ms)	154701.2	2	77350.8	1.87	0.180
Muscle onset (ms)	68518.2	2	34259.1	2.88	0.080
Muscle onset (%)	0.2	2	<0.1	3.05	0.070
Muscle offset (ms)	11735.2	2	5867.6	0.18	0.838
Muscle offset (%)	<0.1	2	<0.1	0.30	0.746
IEMG (mV)	430.7	2	215.4	0.68	0.516
IEMG (%)	15457.9	2	7729.0	0.97	0.396
Error (Footwear)					
Duration (ms)	825828.1	20	41291.4		
Muscle onset (ms)	238292.1	20	11914.6		
Muscle onset (%)	0.5	20	<0.1		
Muscle offset (ms)	657257.2	20	32862.9		
Muscle offset (%)	1.1	20	<0.1		
IEMG (mV)	6293.3	20	314.7		
IEMG (%)	159482.8	20	7974.1		

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Table B.12.7: Three-way repeated measures ANOVA results for the peroneus longus (PL) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Surface					
Duration (ms)	244706.4	2	122353.2	2.38	0.118
Muscle onset (ms)	25984.8	1.204†	21588.4	1.33	0.280
Muscle onset (%)	<0.1	1.172†	<0.1	1.55	0.242
Muscle offset (ms)	88301.5	2	44150.8	1.11	0.350
Muscle offset (%)	0.2	2	<0.1	1.35	0.282
IEMG (mV)	1386.3	2	693.2	4.11	0.032*
IEMG (%)	44170.3	1.277†	34594.1	2.32	0.149
Surface x Group					
Duration (ms)	23858.6	2	11929.3	0.23	0.795
Muscle onset (ms)	4447.3	1.204†	3694.8	0.23	0.686
Muscle onset (%)	<0.1	1.172†	<0.1	0.19	0.710
Muscle offset (ms)	20340.7	2	10170.3	0.26	0.777
Muscle offset (%)	<0.1	2	<0.1	0.47	0.634
IEMG (mV)	37.0	2	18.5	0.11	0.897
IEMG (%)	14307.0	1.277†	11205.2	0.75	0.434
Error (Surface)					
Duration (ms)	1026563.6	20	51328.2		
Muscle onset (ms)	194963.4	12.036†	16197.7		
Muscle onset (%)	0.4	11.720†	<0.1		
Muscle offset (ms)	797239.9	20	39862.0		
Muscle offset (%)	1.4	20	<0.1		
IEMG (mV)	3372.5	20	168.6		
IEMG (%)	190249.9	12.768†	14900.3		
Footwear x Surface					
Duration (ms)	176118.8	4	44029.7	0.84	0.509
Muscle onset (ms)	22767.3	2.206†	10322.5	0.75	0.495
Muscle onset (%)	<0.1	2.436†	<0.1	0.90	0.439
Muscle offset (ms)	202401.2	4	50600.3	1.10	0.370
Muscle offset (%)	0.3	4	<0.1	1.00	0.420
IEMG (mV)	506.4	4	126.6	0.57	0.687
IEMG (%)	32567.8	4	8142.0	1.96	0.119
Footwear x Surface x Group					
Duration (ms)	290193.5	4	72548.4	1.38	0.258
Muscle onset (ms)	12302.8	2.206†	5578.0	0.41	0.690
Muscle onset (%)	<0.1	2.436†	<0.1	0.57	0.607
Muscle offset (ms)	87711.5	4	21927.9	0.48	0.753
Muscle offset (%)	0.1	4	<0.1	0.49	0.743
IEMG (mV)	1057.8	4	264.5	1.19	0.331
IEMG (%)	25195.8	4	6298.9	1.52	0.215
Error (Footwear x Surface)					
Duration (ms)	2101019.6	40	52525.5		
Muscle onset (ms)	302500.7	22.056†	13715.2		
Muscle onset (%)	0.6	24.361†	<0.1		
Muscle offset (ms)	1841619.9	40	46040.5		
Muscle offset (%)	3.0	40	<0.1		
IEMG (mV)	8907.6	40	222.7		
IEMG (%)	165933.1	40	4148.3		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.10, Appendix B.11).

* Indicates statistical significance at $p \leq 0.05$.

Table B.12.8: Three-way repeated measures ANOVA results for the gastrocnemius (G) data.

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Duration (ms)	1702.7	2	851.4	0.12	0.887
Muscle onset (ms)	31485.5	2	15742.8	3.04	0.068
Muscle onset (%)	<0.1	2	<0.1	2.58	0.099
Muscle offset (ms)	41051.8	2	20525.9	8.72	0.002*
Muscle offset (%)	<0.1	1.361†	<0.1	7.74	0.009*
IEMG (mV)	77.4	2	38.7	1.77	0.194
IEMG (%)	799.6	2	399.8	2.09	0.148
Footwear x Group					
Duration (ms)	25292.4	2	12646.2	1.79	0.190
Muscle onset (ms)	25981.1	2	12990.6	2.51	0.105
Muscle onset (%)	<0.1	2	<0.1	2.14	0.141
Muscle offset (ms)	14955.6	2	7477.8	3.18	0.061
Muscle offset (%)	<0.1	1.361†	<0.1	1.76	0.208
IEMG (mV)	42.2	2	21.1	0.96	0.397
IEMG (%)	271.3	2	135.6	0.71	0.504
Error (Footwear)					
Duration (ms)	155440.0	22	7065.5		
Muscle onset (ms)	114030.1	22	5183.2		
Muscle onset (%)	0.2	22	<0.1		
Muscle offset (ms)	51790.9	22	2354.1		
Muscle offset (%)	<0.1	14.967†	<0.1		
IEMG (mV)	481.8	22	21.9		
IEMG (%)	4216.2	22	191.6		
Surface					
Duration (ms)	18444.5	2	9222.2	1.46	0.253
Muscle onset (ms)	5505.6	1.328†	4147.1	0.48	0.553
Muscle onset (%)	<0.1	1.377†	<0.1	0.69	0.464
Muscle offset (ms)	4163.9	2	2081.9	1.16	0.331
Muscle offset (%)	<0.1	2	<0.1	0.33	0.726
IEMG (mV)	61.5	2	30.8	2.41	0.113
IEMG (%)	466.7	1.343†	347.5	2.08	0.168
Surface x Group					
Duration (ms)	203.3	2	101.6	0.02	0.984
Muscle onset (ms)	4995.0	1.328†	3762.5	0.43	0.576
Muscle onset (%)	<0.1	1.377†	<0.1	0.45	0.575
Muscle offset (ms)	2344.8	2	1172.4	0.66	0.529
Muscle offset (%)	<0.1	2	<0.1	0.30	0.743
IEMG (mV)	19.6	2	9.8	0.77	0.477
IEMG (%)	178.0	1.343†	132.5	0.79	0.423
Error (Surface)					
Duration (ms)	138747.1	22	6306.7		
Muscle onset (ms)	126576.2	14.603†	8667.6		
Muscle onset (%)	0.2	15.149†	<0.1		
Muscle offset (ms)	39335.3	22	1788.0		
Muscle offset (%)	<0.1	22	<0.1		
IEMG (mV)	281.1	22	12.8		
IEMG (%)	2467.1	14.772†	167.0		
Footwear x Surface					
Duration (ms)	36584.1	4	9146.0	4.06	0.007*
Muscle onset (ms)	4897.3	1.813†	2701.6	0.59	0.547
Muscle onset (%)	<0.1	1.825†	<0.1	1.16	0.329
Muscle offset (ms)	16193.4	4	4048.3	5.28	0.001*

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Table B.12.8: Three-way repeated measures ANOVA results for the gastrocnemius (G) data (continued).

Source	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear x Surface (continued)					
Muscle offset (%)	<0.1	2.327†	<0.1	2.64	0.083
IEMG (mV)	72.1	4	18.0	2.19	0.086
IEMG (%)	601.7	4	150.4	2.26	0.077
Footwear x Surface x Group					
Duration (ms)	12243.6	4	3060.9	1.36	0.264
Muscle onset (ms)	16872.9	1.813†	9308.1	2.04	0.159
Muscle onset (%)	<0.1	1.825†	<0.1	2.54	0.108
Muscle offset (ms)	3077.7	4	769.4	1.00	0.416
Muscle offset (%)	<0.1	2.327†	<0.1	0.91	0.427
IEMG (mV)	61.0	4	15.3	1.85	0.136
IEMG (%)	387.6	4	96.9	1.46	0.231
Error (Footwear x Surface)					
Duration (ms)	99192.7	44	2254.4		
Muscle onset (ms)	90995.8	19.940†	4563.5		
Muscle onset (%)	0.2	20.080†	<0.1		
Muscle offset (ms)	33710.0	44	766.1		
Muscle offset (%)	<0.1	25.598†	<0.1		
IEMG (mV)	362.0	44	8.2		
IEMG (%)	2923.5	44	66.4		

† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.11, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

Table B.12.9: Three-way repeated measures ANOVA results for the vertical ground reaction force data.

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
Stance (ms)	4162.3	2	2081.1	2.51	0.117
F_{yB} (N)	4792.8	2	2396.4	4.12	0.039*
F_{yB} (N.kg^{-1})	1.1	2	0.5	4.05	0.041*
IC-F_{yB} (ms)	11561.1	1.142†	10119.8	1.77	0.224
IC-F_{yB} (%)	<0.1	1.163†	<0.1	1.53	0.258
F_{yM} (N)	606.3	2	303.1	0.56	0.585
F_{yM} (N.kg^{-1})	<0.1	2	<0.1	0.24	0.792
IC-F_{yM} (ms)	1463.0	1.112†	1315.2	0.40	0.567
IC-F_{yM} (%)	<0.1	1.057†	<0.1	0.03	0.883
Footwear x Group					
Stance (ms)	637.9	2	318.9	0.39	0.687
F_{yB} (N)	268.0	2	134.0	0.23	0.797
F_{yB} (N.kg^{-1})	<0.1	2	<0.1	0.38	0.690
IC-F_{yB} (ms)	17409.9	1.142†	15239.5	2.66	0.140
IC-F_{yB} (%)	<0.1	1.163†	<0.1	2.38	0.161
F_{yM} (N)	3200.9	2	1600.5	2.95	0.086
F_{yM} (N.kg^{-1})	0.7	2	0.4	3.02	0.081
IC-F_{yM} (ms)	537.5	1.112†	483.2	0.15	0.738
IC-F_{yM} (%)	<0.1	1.057†	<0.1	0.20	0.684
Error (Footwear)					
Stance (ms)	11594.1	14	<0.1		
F_{yB} (N)	8146.9	14	581.9		

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Table B.12.9: Three-way repeated measures ANOVA results for the vertical ground reaction force data (continued).

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
F_{yB} (N.kg^{-1})	1.8	14	0.1		
IC-F_{yB} (ms)	45839.3	7.997†	5732.1		
IC-F_{yB} (%)	<0.1	8.142†	<0.1		
F_{yM} (N)	7609.0	14	543.5		
F_{yM} (N.kg^{-1})	1.6	14	0.1		
IC-F_{yM} (ms)	25633.5	7.787†	3291.8		
IC-F_{yM} (%)	<0.1	7.396†	<0.1		
Surface					
Stance (ms)	6802.9	2	3401.4	1.26	0.310
F_{yB} (N)	6234.7	2	3117.4	3.29	0.068
F_{yB} (N.kg^{-1})	1.5	2	0.7	3.53	0.057
IC-F_{yB} (ms)	29379.6	1.011†	29068.3	3.25	0.114
IC-F_{yB} (%)	<0.1	1.108†	<0.1	4.01	0.084
F_{yM} (N)	2534.2	2	1267.1	1.41	0.276
F_{yM} (N.kg^{-1})	0.4	2	0.2	1.43	0.272
IC-F_{yM} (ms)	3005.7	2	1502.8	1.14	0.347
IC-F_{yM} (%)	<0.1	2	<0.1	0.55	0.587
Surface x Group					
Stance (ms)	2151.2	2	1075.6	0.403	0.566
F_{yB} (N)	511.7	2	255.8	0.270	0.767
F_{yB} (N.kg^{-1})	0.2	2	<0.1	0.404	0.675
IC-F_{yB} (ms)	17695.6	1.011†	17508.1	1.956	0.204
IC-F_{yB} (%)	<0.1	1.108†	<0.1	2.530	0.155
F_{yM} (N)	2157.9	2	1079.0	1.20	0.329
F_{yM} (N.kg^{-1})	0.4	2	0.2	1.35	0.291
IC-F_{yM} (ms)	937.5	2	468.8	0.36	0.706
IC-F_{yM} (%)	<0.1	2	<0.1	0.15	0.865
Error (Surface)					
Stance (ms)	37356.4	14	2668.3		
F_{yB} (N)	13273.0	14	948.1		
F_{yB} (N.kg^{-1})	2.9	14	0.2		
IC-F_{yB} (ms)	63326.1	7.075†	8950.8		
IC-F_{yB} (%)	<0.1	7.123†	<0.1		
F_{yM} (N)	12549.3	14	896.4		
F_{yM} (N.kg^{-1})	2.2	14	0.2		
IC-F_{yM} (ms)	18395.0	14	1313.9		
IC-F_{yM} (%)	<0.1	14	<0.1		
Footwear x Surface					
Stance (ms)	15713.7	1.648†	9535.4	5.78	0.022*
F_{yB} (N)	868.0	1.254†	692.0	0.35	0.617
F_{yB} (N.kg^{-1})	0.2	1.258†	0.2	0.35	0.620
IC-F_{yB} (ms)	28618.3	1.076†	26589.0	1.97	0.202
IC-F_{yB} (%)	<0.1	1.070†	<0.1	1.97	0.202
F_{yM} (N)	1505.0	1.980†	760.0	0.54	0.596
F_{yM} (N.kg^{-1})	0.4	1.956†	0.2	0.67	0.526
IC-F_{yM} (ms)	3543.0	1.353†	2619.1	0.63	0.495
IC-F_{yM} (%)	<0.1	1.423†	<0.1	0.14	0.801
Footwear x Surface x Group					
Stance (ms)	2358.0	1.648†	1430.9	0.87	0.425
F_{yB} (N)	1277.7	1.254†	1018.7	0.52	0.532
F_{yB} (N.kg^{-1})	0.4	1.258†	0.3	0.62	0.488
IC-F_{yB} (ms)	27393.8	1.076†	25451.3	1.88	0.211
IC-F_{yB} (%)	<0.1	1.070†	<0.1	1.77	0.224

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Table B.12.9: Three-way repeated measures ANOVA results for the vertical ground reaction force data (continued).

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
F_{yM} (N)	2477.1	1.980†	1250.9	0.88	0.436
F_{yM} (N.kg^{-1})	0.5	1.956†	0.2	0.86	0.443
IC-F_{yM} (ms)	1973.1	1.353†	1458.6	0.35	0.633
IC-F_{yM} (%)	<0.1	1.423†	<0.1	0.47	0.573
Error (Footwear x Surface)					
Stance (ms)	19018.6	11.536†	1648.7		
F_{yB} (N)	17333.4	8.780†	1974.2		
F_{yB} (N.kg^{-1})	4.1	8.807†	0.5		
IC-F_{yB} (ms)	101832.6	7.534†	13516.0		
IC-F_{yB} (%)	0.2	7.489†	<0.1		
F_{yM} (N)	19693.0	13.862†	1420.7		
F_{yM} (N.kg^{-1})	3.7	13.692†	0.3		
IC-F_{yM} (ms)	39553.5	9.469†	4177.0		
IC-F_{yM} (%)	<0.1	9.961†	<0.1		

[a] See Table 5.4 for description of variables.
† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.2, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

Table B.12.10: Three-way repeated measures ANOVA results for the anteroposterior ground reaction force data.

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
F_{xB} (N)	4843.9	2	2422.0	14.19	<0.001*
F_{xB} (N.kg^{-1})	0.9	2	0.4	12.12	<0.001*
IC-F_{xB} (ms)	1517.3	1.264†	1200.7	0.53	0.520
IC-F_{xB} (% stance)	<0.1	1.192†	<0.1	0.84	0.398
IC-FxC (ms)	129717.7	1.116	116251.5	1.67	0.223
IC-F_{xC} (% stance)	0.3	1.092	0.2	1.29	0.284
Footwear x Group					
F_{xB} (N)	739.6	2	369.8	2.17	0.138
F_{xB} (N.kg^{-1})	0.2	2	0.1	2.77	0.085
IC-F_{xB} (ms)	266.3	1.264†	210.7	0.09	0.822
IC-F_{xB} (% stance)	<0.1	1.192†	<0.1	0.29	0.639
IC-F_{xC} (ms)	102853.7	1.116	92176.3	1.32	0.278
IC-F_{xC} (% stance)	0.2	1.092	0.2	1.18	0.304
Error (Footwear)					
F_{xB} (N)	3754.4	22	170.7		
F_{xB} (N.kg^{-1})	0.8	22	<0.1		
IC-F_{xB} (ms)	31537.7	13.901†	2268.7		
IC-F_{xB} (% stance)	<0.1	13.109†	<0.1		
IC-F_{xC} (ms)	856223.1	12.274	69757.9		
IC-F_{xC} (% stance)	2.2	12.017	0.2		
Surface					
F_{xB} (N)	8900.7	1.316†	6762.4	14.92	0.001*
F_{xB} (N.kg^{-1})	1.7	1.309†	1.3	12.96	0.002*
IC-F_{xB} (ms)	8263.0	2	4131.5	2.37	0.117
IC-F_{xB} (% stance)	<0.1	2	<0.1	1.56	0.232
IC-F_{xC} (ms)	143286.1	1.045	137087.2	1.74	0.213
IC-F_{xC} (% stance)	0.3	1.032	0.3	1.34	0.272

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Table B.12.10: Three-way repeated measures ANOVA results for the anteroposterior ground reaction force data (continued).

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Surface x Group					
F_{xB} (N)	329.6	1.316†	250.4	0.55	0.516
F_{xB} ($N.kg^{-1}$)	0.1	1.309†	<0.1	0.85	0.401
IC-F_{xB} (ms)	3099.7	2	1549.8	0.89	0.425
IC-F_{xB} (% stance)	<0.1	2	<0.1	1.08	0.357
IC-F_{xC} (ms)	74537.2	1.045	71312.5	0.91	0.365
IC-F_{xC} (% stance)					
Error (Surface)					
F_{xB} (N)	6560.5	14.378†	453.1		
F_{xB} ($N.kg^{-1}$)	1.5	14.399†	0.1		
IC-F_{xB} (ms)	38315.1	22	1741.6		
IC-F_{xB} (% stance)	<0.1	22	<0.1		
IC-F_{xC} (ms)	903689.7	11.497	78599.4		
IC-F_{xC} (% stance)	2.2	11.355	0.2		
Footwear x Surface					
F_{xB} (N)	2862.6	4	715.7	8.21	<0.001*
F_{xB} ($N.kg^{-1}$)	0.6	4	0.2	7.61	<0.001*
IC-F_{xB} (ms)	5341.5	2.252†	2371.5	0.95	0.411
IC-F_{xB} (% stance)	<0.1	2.060†	<0.1	0.86	0.440
IC-F_{xC} (ms)	127703.8	1.091	117099.4	0.80	0.401
IC-F_{xC} (% stance)	0.4	1.064	0.4	1.06	0.329
Footwear x Surface x Group					
F_{xB} (N)	188.3	4	47.1	0.54	0.707
F_{xB} ($N.kg^{-1}$)	<0.1	4	<0.1	0.62	0.649
IC-F_{xB} (ms)	6139.8	2.252†	2725.8	1.09	0.358
IC-F_{xB} (% stance)	<0.1	2.060†	<0.1	1.32	0.287
IC-F_{xC} (ms)	235148.7	1.091	215622.1	1.47	0.253
IC-F_{xC} (% stance)	0.6	1.064	0.5	1.46	0.254
Error (Footwear x Surface)					
F_{xB} (N)	3837.7	44	87.2		
F_{xB} ($N.kg^{-1}$)	0.9	44	<0.1		
IC-F_{xB} (ms)	62011.7	24.777†	2502.8		
IC-F_{xB} (% stance)	0.1	22.661†	<0.1		
IC-F_{xC} (ms)	1764586.0	11.996	147096.0		
IC-F_{xC} (% stance)	4.3	11.701	0.4		

[a] See Table 5.4 for description of variables.
† Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.3, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

Table B.12.11: Three-way repeated measures ANOVA results for the coefficient of friction data.

Source [a]	Sum of Squares	*df*	Mean Square	*F*-value	*p*-value
Footwear					
μ_S	0.3	2	0.2	14.19	0.015*
μ_D	<0.1	2	<0.1	0.20	0.824
IC-μ_D (ms)	65.8	2	32.9	2.76	0.176
IC-μ_D (%)	3.7	2	1.9	5.18	0.078
Footwear x Group					
μ_S	0.1	2	<0.1	5.06	0.080
μ_D	<0.1	2	<0.1	0.41	0.688
IC-μ_D (ms)	21.0	2	10.5	0.88	0.482
IC-μ_D (%)	1.2	2	0.6	1.67	0.297
Error (Footwear)					
μ_S	<0.1	4	<0.1		
μ_D	<0.1	4	<0.1		
IC-μ_D (ms)	47.7	4	11.9		
IC-μ_D (%)	1.4	4	0.4		
Surface					
μ_S	0.4	2	0.2	9.23	0.032
μ_D	<0.1	2	<0.1	0.29	0.764
IC-μ_D (ms)	97.3	2	48.6	8.47	0.036*
IC-μ_D (%)	2.2	2	1.1	3.98	0.112
Surface x Group					
μ_S	<0.1	2	<0.1	0.76	0.929
μ_D	<0.1	2	<0.1	0.29	0.761
IC-μ_D (ms)	22.8	2	11.4	1.99	0.251
IC-μ_D (%)	0.4	2	0.2	0.74	0.535
Error (Surface)					
μ_S	<0.1	4	<0.1		
μ_D	<0.1	4	<0.1		
IC-μ_D (ms)	23.0	4	5.7		
IC-μ_D (%)	1.1	4	0.3		
Footwear x Surface					
μ_S	<0.1	4	<0.1	0.32	0.856
μ_D	<0.1	4	<0.1	1.30	0.348
IC-μ_D (ms)	28.3	1.111	25.5	1.37	0.362
IC-μ_D (%)	0.2	1.033	0.2	0.38	0.605
Footwear x Surface x Group					
μ_S	<0.1	4	<0.1	0.44	0.777
μ_D	<0.1	4	<0.1	0.88	0.516
IC-μ_D (ms)	60.6	1.111	54.5	2.92	0.221
IC-μ_D (%)	1.3	1.033	1.3	2.85	0.231
Error (Footwear x Surface)					
μ_S	0.1	8	<0.1		
μ_D	<0.1	8	<0.1		
IC-μ_D (ms)	41.5	2.229	18.7		
IC-μ_D (%)	0.9	2.067	0.4		

[a] See Table 5.4 for description of variables.
[†] Greenhouse-Geisser adjustment due to significant Mauchly's W (see Table B.11.4, Appendix B.11).
* Indicates statistical significance at $p \leq 0.05$.

“You have brains in your head
You have feet in your shoes
You can steer yourself
any direction you choose
You’re on your own
And you know what to do
And YOU are the one
who’ll decide where to go…..”

Dr Seuss (1904-1991)
Oh! The Places You’ll Go