The effects of safety flooring on sit-to-stand and quiet stance balance reactions in retirement home-dwellers

by

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Author's Declaration

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

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Abstract

Fall-related injuries in adults over the age of 65 pose an important public health issue especially with an increasing number of older adults living in retirement homes and nursing homes. Safety floors have been developed as an intervention to reduce the risk of these injuries. However, their effects on balance control reactions had never been tested during certain activities of daily living in retirement home dwellers. This research investigated how balance reactions are affected by the mechanical properties of safety flooring in older adults. The safety flooring showed minimal impact on the balance reactions while retaining force attenuation properties.

There were two studies as part of this thesis. The purpose of the first study was to determine whether the Nintendo Wii Balance Board (WBB) can be used as an appropriate substitution for a force plate when measuring balance reactions during common tests used to assess balance in older adults. Specifically, I characterized the technical specifications of the WBB and compared them to those of the force plate, showing that the two devices yielded similar responses during balance measures of quiet stance. The second study investigated the effect of two traditional floors and three safety flooring systems on balance control mechanisms (based on changes in underfoot centre of pressure) during sit-to-stand and quiet stance tasks in retirement home-dwellers. The results of this study provided evidence supporting the potential for safety floors to reduce fall-related injury risk without impairing balance and mobility of users. Additional research may want to assess WBB performance during dynamic tasks involving shear forces. The results from this study supports prospective clinical investigations of pilot installations of safety flooring in retirement and nursing home settings to evaluate their real life effects on fall related injuries.

Keywords: fall-related injuries; safety flooring; older adults; balance control; Nintendo Wii Balance Board; ageing;

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List of Acronyms

AP – Anterior-posterior

BOS – Base of support

COM – Centre of mass

COP – Centre of pressure

MDC – Minimal detectable change

ML – Medial-lateral

MMOS – Minimum margin of safety

QS – Quiet stance

STS – Sit-to-stand

TUG – Timed up and go test

VWP – The Village of Winston Park retirement home

WBB – Nintendo Wii Balance Board

Thesis Overview

Fall-related injuries in adults over the age of 65 pose an important public health issue especially with an increasing number of older adults living in retirement homes and nursing homes. Approximately 1 in 3 community dwelling adults over the age of 65 experiences a fall at least once a year and falls are responsible for 57% of injury-related deaths in female seniors. Safety floors, designed to attenuate forces upon impact but remain rigid under other circumstances, have been proposed as an intervention to reduce the risk of these injuries in environments of high risk such as retirement homes, nursing homes, and hospitals. However, their effects on balance control reactions had never been tested during certain activities of daily living in retirement home dwellers prior to this study. This research investigated how balance reactions are affected by the mechanical properties of safety flooring in older adults.

There were two studies as part of this thesis. The purpose of the first study was to determine whether the Nintendo Wii Balance Board (WBB) can be used as an appropriate substitution for a force plate when measuring balance reactions during common tests used to assess balance in older adults. Some basic specifications such as linearity and centre of pressure accuracy have previously been tested. I retested those technical specifications, as well as untested characteristics such as load accuracy, unloading characteristics, uniformity, and drift, of the WBB and compared them to those of the force plate. It was hypothesized that the specifications of the WBB would be comparable to those of a force plate, and that the centre of pressure measurements from both devices would be in clinical agreement.

The second study investigated the effect of two traditional floors and three safety flooring systems on balance control mechanisms (based on changes in underfoot centre of pressure) during sit-to-stand and quiet stance tasks in retirement home-dwellers. Previous studies have demonstrated the positive impact attenuation effects of safety floors in simulated head, hip and footfall impacts as well as its minimal impact on balance control effects in community dwelling older adults. However, limited testing on safety floors have been performed in retirement home dwelling populations. The purpose of the second study was to evaluate the effects on balance control during quiet stance and sit-to-stand tasks in retirement home-dwellers. It was hypothesized that balance control variables relating to displacement and rate of displacement of centre of pressure would not be significantly different between safety floors and a control condition of flooring representative of those used in retirement homes.

Chapter 1
Introduction and review of literature

1.1 Fall-related Injuries in Older Adults

1.1.1 Implications to society and scope of the problem

Fall-related injuries in adults over the age of 65 pose an important public health issue particularly with an increasing number of older adults living in retirement homes and nursing homes (Sattin, 1992; SMARTRISK, 2009). Approximately 1 in 3 community dwelling adults over the age of 65 experiences a fall at least once a year (Tinetti et al., 1988; Nevitt et al., 1989; Sattin, 1992; O'Loughlin et al., 1993), with half experiencing multiple falls (Tinetti et al., 1988; Rubenstein and Josephson, 2002). Fall incidence is even higher by 10% for those living in institutional settings (approximately 43%) (Tinetti et al., 1988; Rubenstein et al., 1996; Luukinen et al., 2010). Falls are the leading cause of injury in older adults (Grisso et al., 1991; Alexander et al., 1992) and are responsible for 57% of injury-related deaths in female seniors (Raina et al., 1997), and up to 85% of injury hospitalizations (CIHR, 2007). There are approximately 1.6 falls per person per year in long-term care facilities (Rubenstein and Josephson, 2002). They amount to an approximate cost of $2.8 billion to the Canadian economy annually (SMARTRISK, 2009). Aging can increase the risk, severity, and incidence of experiencing a fall related injury (Sterling et al., 2001; Rubenstein and Josephson, 2002).

The rate of injuries from falls in the elderly is between 20%-60%, depending on the population (Nevitt et al., 1991; Lord and Dayhew, 2001), and up to 25% of these falls result in some type of fracture (Tinetti et al., 1988; Nevitt et al., 1991; Thapa et al., 1996). Nearly half of fall related injuries result in the patients being discharged to nursing homes (Sattin et al., 1990; Tinetti and Williams, 1997). Around 30% of falls by older adults in institutional settings may lead to at least one injury (Jensen et al., 2002b). Hip fractures are especially common in older adults living in residential care facilities (Norton et al., 1999) and are among the most expensive incidences among serious fall related injuries (Chrischilles et al., 1994). One in five women has had a hip fracture by the age of 80, which increases to one in two by the age of 90 (Kannus et al., 1999). 25%-75% never recover to their pre-fracture level of function in activities of daily living (Magaziner et al., 1990) and 40% die within 6 months of sustaining the injury (Evans et al., 1979)

A history of falling, or even anticipation of falling can lead to a 'fear of falling', or higher subjective fall risk, that may develop from having fallen in the past, or even in the absence of recent falls (Maki and McIlroy, 2005), has the potential to limit the quality of daily living as well as increase fall risk in older adults (Tinetti et al., 1988; Campbell et al., 1989; Nevitt et al., 1989; Wolf et al., 1996). 30% - 70% of past fallers develop a fear of falling in the future, causing them to restrict activity amount, have losses of confidence, decrease dependence, develop depression and increase social isolation (Cumming et al., 2000; Rubenstein and Josephson, 2002). The incidence and potential risk factors of current fall related injuries in older adults is not one to be overlooked and must be further investigated to develop both preventative measures and impact reduction strategies.

1.1.2 Factors Associated with Falls and Fall-related Injuries

While not all falls result in injury, injuries resulting from falls can have life altering consequences especially with older adults. It is important to note that as the number of risk factors for falling and fall related injuries increases in an older adult, their chances of falling subsequently increases (Tinetti et al., 1988; Nevitt et al., 1989). For this reason risk factors for falling and resulting injuries must be identified in order to screen or develop interventions. These factors are discussed below, and are summarized in Table 1-1**.**

There are several age related changes that may be associated with an increased risk of falls including: gait or balance impairment, postprandial hypotension, decreased muscle mass (leading to gait disorders) and changes in visual, auditory or proprioceptive systems (Sekuler et al., 1980; Leibowitz and Shupert, 1985; Campbell et al., 1989; Gehlsen and Whaley, 1990; Jonsson et al., 1990; Lord et al., 1993). Whipple et al. (1987) found that fallers among older adults had significantly slower gait speed and shorter stride length. Along with changes in gait, older adults with impaired one leg balance are also at higher risk for suffering an injurious fall (Vellas et al., 1997). Older adults are more susceptible to certain illnesses such as arthritis, degenerative joint diseases, osteoporosis, Parkinson's disease and effects following a stroke, which are associated with fall risk (Campbell et al., 1989; Nevitt et al., 1989; Tinetti et al., 1995; Rubenstein and Josephson, 2002).

The use of certain types and the number of medications, and their association to fall risk has been studied. Older adults who take a greater number of medications (4+), especially psychotropic drugs, have been shown to be at greater fall risk (Tinetti et al., 1988; Campbell et al., 1989; Lord et al., 1993). This may be due to their effect on postural hypotension, postural reflexes, or reaction times. Frail and inactive older adults (Campbell et al., 1989; Rubenstein and Josephson, 2002), and patients with impaired mobility or those receiving community services (Lord et al., 1993; Studenski et al., 1994) also have a higher risk of falling.

Falls resulting in injury are caused by a number of identical factors as in falls that occur without a resulting injury, with some additional characteristics. Females, those with low body mass (both likely to be related to osteoporosis), and those participating in higher physical activity (resulting in an increase of environmental hazard exposure) are at greater risk for suffering an injury during a fall (Thapa et al., 1996; Rubenstein and Josephson, 2002). The risk of major injury during a fall is nearly six times greater for syncope related falls versus non-syncopal related falls (Nevitt et al., 1991). Those who have suffered from a stroke or respiratory disorder are approximately two times as likely to have a serious injury resulting from a fall (O'Loughlin et al., 1993).

The risk factors and the proportion of falls resulting in fall-related injuries are similar in community dwelling persons as in residential dwelling persons (Rubenstein and Josephson, 2002). However, as mentioned earlier, fall incidence is higher in residential care dwellers (43%), the number of fall-related injuries (1.6 per person year) is increased in this population. For this reason, intervention efforts should be developed and targeted towards this higher risk population. Associated environmental risk factors for both community dwelling and residential home dwelling individuals include wet floors, poor lighting and improper bed heights. In nursing homes particularly, most falls occur on a level surface by bedside or bathroom and are associated with low to moderate changes in position or posture; Such activities may include: arising from bed, going to and from the bathroom, and transferring to a bed, chair, or toilet (Kalchthaler et al., 1978; Berry et al., 1981; Tinetti, 1987; Tinetti et al., 1988). The knowledge of these fall risk factors demonstrates a need for more in depth research of the sit-to-stand movement and its related balance characteristics to reduce fall related injuries.

Table 1-1: Summary of risk factors associated with falling and fall-related injuries

Category	Characteristic	Associated changes
Living environment	Wet floors	Altered gait
	Poor lighting	Decreased quality of visual proprioception
	Improper bed height	Increased muscle strength challenges
Personal attributes	Age	Gait or balance impairment, postprandial hypotension, decreased muscle mass, changes in visual, auditory or proprioceptive systems
	Illnesses	Arthritis, degenerative joint diseases, osteoporosis, Parkinson's disease, post stroke effects
	Medications (Increase of number taken; use of psychotropic drugs)	Affecting postural hypotension, postural reflexes, reaction times
	Activity level	Frail and inactive older adults, patients with impaired mobility or those receiving community services
	History of falls	Fear of falling, restricted activity
Additional factors that increase risk for injury	Female	Linked to osteoporosis
	Low body mass	Linked to osteoporosis
	High physical activity	Increased environmental hazard exposure
	History of syncope, syncope or respiratory disorders	State of / loss of consciousness

1.1.2.1 Laboratory tests of balance that predict fall risk

It is important that the following three aspects of balance are evaluated when assessing fall risk because of their relation to real life situations: static balance maintenance, postural adjustments to voluntary movements, and postural responses to external perturbations (Chiu et al., 2003). The range of movement levels is necessary because static measures of balance have minimal association on dynamic responses to external movement (Maki et al., 1990; Owings et al., 2000; Pavol et al., 2002; Mackey and Robinovitch, 2005). This may be due to differences in neuromuscular control strategies or efforts between the two tasks (Morasso and Schieppati, 1999; Morasso and Sanguineti, 2002; Mackey and Robinovitch, 2005).

Balance measures from balance tests are often used as a predictor of fall risk (Berg, 1989; Maki and McIlroy, 1996; Stel et al., 2003). Centre of pressure (COP) and centre of mass (COM) trajectories are assessed as balance variables calculated from force plate and motion capture system data. COP is defined as the centroid of vertical ground reaction force (Winter, 2009), which governs the horizontal movement of whole body COM (estimated by the weighted average of individual body segment locations). As the COM moves towards the boundaries of the base of support (BOS) defined by the area between the feet, the COP rapidly exceeds the COM to ‘push’ the COM back to the neutral position, known as the ‘sheepdog’ effect (Winter, 2009). The minimum margin safety (MMOS) is defined as the distance between the furthest COP excursion and the BOS boundary.

One common task to measure static balance is postural sway during quiet stance because it reflects one’s ability to maintain balance during daily activities that require ‘feet-in-place’ strategies (Gatev et al., 1999). Higher measures of COP sway excursion has been associated with increased frequency of falls (Fernie et al., 1982; Lichtenstein et al., 1990) and increased age (Peterka and Black, 1990; Wolfson et al., 1992). Topper et al. (1993) found that sway measures during quiet stance tasks, and particularly in the medial-lateral (ML) direction, are good predictors of fall risk occurring from changes in the base of support or centre of mass. A higher frequency of falls is associated by increased ML sway (Maki et al., 1994; Stel et al., 2003). Greater ML COP sway is also correlated with an increased risk of injury during fall because falls to the side are more likely to result in injury compared to other directions (Nevitt and Cummings, 1993). Increased COP excursion and velocity, namely in the ML direction, have been found to be correlated with increased fall risk during quiet stance (Campbell et al., 1989; Maki et al., 1990; Prieto et al., 1996; Thapa et al., 1996).

There have also been a number of studies mimicking activities of daily living to assess postural adjustments to voluntary movements. In stroke patients, many falls occur during changes in position such as standing up, sitting down, and the initiation of walking (Nyberg and Gustafson, 1995). The sit-to-stand (STS) task becomes increasingly difficult with age and is a movement that is often studied as a predictor of fall risk and recurrent fallers (Tinetti et al., 1986). Researchers have studied sit-to-stand measures such as the time to stand up once, three, five and ten times, the number of sit-to-stand-to sit cycles completed in ten and 30 seconds, and measures of muscle performance while completing the STS motion (Bohannon, 1995). Older adults unable to arise from a seated position are more likely to fall (Topper et al., 1993). Additional factors found to predict fall risk includes: the risk of recurrent falling is 2.5 times larger in elderly adults unable to complete the STS motion (Nevitt et al., 1989), requiring two or more seconds to complete one STS (Nevitt et al., 1989; Najafi et al., 2002). One test that combines the sit-to-stand test with additional transitional postural changes, including transfers to and from sitting, gait initiation and turning, is the Timed Up and Go test (TUG) (Boulgarides et al., 2003; Laing and Robinovitch, 2009). Studies have shown associations between an increased time to complete the TUG and increased fall risk (Lundin-Olsson et al., 1998; Shumway-Cook et al., 2000; Chiu et al., 2003).

Reactions to external perturbations (e.g. being accidentally nudged or tripping over a raised edge) may be indicators of fall risk for older adults, who have been shown to sway more in response to a perturbation than young adults (Stelmach et al., 1989; Maki et al., 1990). Healthy older adults are also at greater risk than young adults of falling from a novel and unexpected perturbations, but both groups are able to learn to avoid falling during external perturbations during a STS task with repeated exposures (Pai, 1999; Pavol et al., 2002). The study of ‘stepping responses’ as a result of an external perturbations rather than feet-in-place strategies can provide insight into fall risk during real-life situations (Pai et al., 1998; Maki and McIlroy, 2005). Pavol et al. (2002) found that the majority of recurrent fallers (those who fell more than once in the study) did not use a step to recover from an external alteration of COM, implying a fall in clinical terms.

1.2 Current Strategies to decrease Fall-Related Injury Risk in Older Adults

1.2.1 Current methods for preventing falls

The process of identifying risk factors for falls in older adults has led to the development of several methods to prevent fall-related injuries. The most common interventions are having group exercise classes, removal or modification of environmental hazards, and fall prevention education (Province et al., 1995; Campbell et al., 1997; Cumming et al., 1999; Jensen et al., 2002a; Rapp et al., 2008). Exercise programs targeting balance, strength and gait are known to reduce overall fall incidence (Tinetti et al., 1994; Wagner et al., 1994). When performed as an exercise program, tai chi has also been associated with a decreased risk of one or more falls (Wolf et al., 1996; Kutner et al., 1997; Nowalk et al., 2001; Li et al., 2005). Jensen et al. (2002a) conducted a randomized control study in which the intervention group received environmental modification, repair or supply of walking aids as necessary, and removal of medications with side effects believed to increase fall risk. They found a

12% increase in the number of falls in the intervention group, and more recurrent fallers in the control group compared to the intervention group.

1.2.2 Strategies for preventing injuries in the event of a fall

There are three common protective measures used for decreasing the severity of potential injuries that may be sustained during a fall. They include the use of hip protectors, instruction of safe landing techniques, and installation of compliant flooring.

Jensen et al. (2002a) provided an intervention group with free hip protectors. They found that the intervention group experienced a decreased number of femoral fractures. Force attenuation effects have also been found with the use of hip protectors (Nevitt and Cummings, 1994; Robinovitch et al., 2000; Laing and Robinovitch, 2008b; Laing and Robinovitch, 2008a). However, in order for hip protectors to be protective against injuries, user compliance is required, which may be difficult due to its location (incorporated into undergarments) and embarrassment due to its bulky appearance. Furthermore, hip protectors only protect against hip fractures rather than all fall related injuries.

Safe landing techniques in the event of accidental falls have been investigated as a way to reduce the risk of injury and methods for safer fall methods have been evaluated as programs for older adults. The use of martial arts fall techniques by experienced practitioners as well as by young adults without prior experience have been found to decrease hip impact forces by up to 30% during falls (Groen et al., 2007; Weerdesteyn et al., 2008). In healthy older adults, martial arts fall training was found to decrease hip impact forces (Groen et al., 2010). However, the falls in the study were performed from kneeling, which may not be a common scenario with older adults.

The third potential preventative measure for reducing fall-related injuries is to implement safety flooring in long term care facilities. These floors have been designed with the goal of reducing the risk of fall-related injuries. Safety floors have been called novel compliant flooring or low stiffness flooring in the past, but I feel that these terms do not accurately portray the characteristic and purpose of these flooring systems. Designed to have a dual-stiffness response, safety flooring is rigid under low forces, such as walking, but collapses and absorbs forces during impact. This force attenuation effect may help to decrease the risk of sustaining an injury in the event of a fall. Although extensive testing on the novel compliant effects on force attenuation and balance is still limited, safety flooring is being increasingly implemented in residential care environments. Research on the effects on safety flooring has primarily been tested on community dwelling young and older adults but has yet to be extensively tested on older adults living in retirement homes.

1.2.2.1 Compliant flooring and force attenuation

Past epidemiologic studies have shown that falling onto soft surfaces reduces the risk of hip fractures (Nevitt and Cummings, 1993; Simpson et al., 2004; Healey 1994). Laboratory studies using mechanical test systems have found force attenuation of up to 7% for wooden floors, 15% for carpets, and 24% for carpets with under padding (Gardner et al., 1998, Maki and Fernie, 1990, Simpson et al., 2004).

Studies with human volunteers have also shown benefits of compliant flooring against the risk of hip fractures. Laing et al. (2006) found that compared to the rigid floor condition, peak hip impact forces were on average 8%-15% lower on compliant foam surfaces. Sran and Robinovitch (2008) found similar peak force attenuation applied to the buttocks during a backward fall from standing.

During development, a version of safety flooring using buckling columns (Penn State Safety Floor) was validated using a finite element model to show that they would be able to meet the maximum deflection of 2 mm during normal locomotion, while effectively decreasing impact forces during falls

(Casalena et al., 1998a; Casalena et al., 1998b). More recently, safety flooring was found to potentially decrease hip forces during falls by up to 50% in older women with minimal effects on balance and mobility during quiet standing (Laing and Robinovitch, 2009). Similarly, during a study of the effect of safety flooring on head impact forces during worst case scenarios using mechanical systems, forces and accelerations were significantly attenuated upon impact on safety flooring compared to traditional flooring (Wright, 2011). Results have demonstrated the protective measure of safety flooring against the severity of injuries resulting from fall-related impacts through force attenuation.

1.2.2.2 Compliant flooring and its effects on balance

Despite its potential for decreasing impact forces, compliant flooring has the potential to impair balance and mobility. This is important because decreased balance control can result in a decreased minimum margin of safety and increase fall risk. If the stiffness of the flooring surface is not high enough during activities of daily living (i.e. low force), the risk of falling may increase. Previous studies have shown this by demonstrating the increase of sway amplitude during quiet standing on compliant foam surfaces compared to the rigid control floors (Ring et al., 1989; Lord and Menz, 2000). This decreased balance may result from a decreased quality of information from golgi tendon ankle proprioceptors and pressure receptors on the plantar foot surface (Ring et al., 1989; Lord and Menz, 2000; Betker et al., 2005), and an increase in energy expenditure during walking (Redfern et al., 1997). During gait, the initial step is also affected by foam surfaces due to potential decreased trunk stability from a lowered centre of mass trajectory. However, balance can be restored during subsequent steps through maintained toe clearance and by changes to stride characteristics (Marigold and Patla, 2005; MacLellan and Patla, 2006).

Safety floors also have potentially negative effects on balance control during activities of daily living. Equations for a simple mass-spring model predict that if floor stiffness is decreased (i.e. more compliant), magnitude and time to peak torque generation would be decreased compared to traditional flooring (McMahon et al., 1987; Laing et al., 2006). However, recent results from research with older female participants who were evaluated during quiet stance, a get up and go test and backwards floor perturbations (the three balance aspects as outlined by Chiu et al. (2003), see Section1.1.2.1) show only a minimal influence on balance from safety floors compared to traditional floors (Laing and Robinovitch, 2009). Similarly, Wright (2011) assessed the balance responses in retirement home dwellers during an unexpected forward leaning tether release perturbation and found no difference between traditional flooring and safety floor conditions. A summary of the known effects of safety flooring on force attenuation and balance properties can be found in Table 1-2.

Although the number of studies focusing on the effect of compliant flooring on the balance of older adults is increasing, there are still a few key factors that need to be taken into account with future research. Limitations of past studies includes an insufficient range of flooring stiffness' studied (only two safety systems in Laing (2009)), the use of mechanical systems to simulate falls instead of human volunteers, and simulation of a single body configuration. One important issue with previous floor and balance studies is that centre of pressure during a sit-to-stand task has never been measured on compliant flooring to determine the effects on balance due to floor condition. This is important because the sit-to-stand motion is a functional task for all populations. My thesis will address this lack of an evidence base by testing balance measures on compliant flooring in retirement home dwellers during quiet stance. I will also test balance measures during a sit-to-stand task on compliant flooring. This study is described in Chapter 3.

Table 1-2: Summary of the effect of compliant flooring on force attenuation, balance and mobility

Compliant flooring effect	Associated consequences
Force attenuation	Decreased magnitude and time to peak force of hip and head impact forces during simulated falls Redistributed pressure distribution, decreasing peak pressure
Balance and mobility:	Degraded proprioception at ankle Diminished quality of mechanoreceptors on plantar foot surface Increased energy expenditure Decreased ankle stiffness Decreased trunk stability Response time Rate of torque generation from feet Clearance

1.3 Systems for measuring balance

1.3.1 Force plates

Force plates are commonly used equipment to assess measures of balance and postural control. A force plate typically quantifies forces and moments applied to its recording surface in terms of a predetermined coordinate system. They can be used to analyze motions such as gait, balance, and impact forces.

There are several types of transducers designed to sense changes in load which are implemented in force plates. Each type of transducer has a specific loading situation that it is better suited for. Piezo-electric sensors accurately register high frequency situations such as impacts, but have poorer direct-current response during low frequencies (Gautschi, 2006). Piezo-resistive sensors are designed similarly and have good frequency response, are extremely temperature sensitive but the measured responses will drift (Liu, 2006). As such, the sensors are better for dynamic situations rather than static loading and are commonly used in pressure sensors. The Kistler brand uses both piezo-electric and piezo-resistive sensors in their products to measure pressure, force, acceleration and torque, and are often used for biomechanical research. Strain gauges are another popular device to measure changes in stress of an object based on deformation of itself. They are used to measure several different types of resistances such as capacitance, inductance, impedances, forces, temperatures and pressures. Strain gauges are very commonly implemented in force plates that are used for biomechanical purposes and have high stiffness, high sensitivity and low crosstalk (Murray and Miller, 1992). Hall-effect sensors measure changes in magnetic field and can be used in used wireless applications, current sensors, fluid flow sensors, pressure sensors and in devices with potentiometers build a non-contact sensor. However, the magnetic flux may be affected if surrounded by other electric fields. They are a more economical force plate option if high sampling rate or sensitivity is not as important. Advanced Mechanical Technology Inc. (AMTI, Watertown, MA, USA) currently produces economic force plates using Hall Effect technology, as well as higher standard force plates using strain gauge technology.

Force plates are currently the gold standard for biomechanical research involving the collection of forces and moments because of their high accuracy and long-term durability. The characteristics of good systems include high sensitivity and resolution, excellent linearity and hysteresis measurements and low crosstalk. In respect to durability, the measurements are repeatable over time and are temperature resistant. In addition to being able to precisely measure forces and moments in three axes, sensors can be customized and mounted in equipment such as treadmills, stairs, or walking aids. Despite these benefits, researchers have been looking for alternatives to using force plates because they tend to be quite heavy (~10-45 kg), which reduces portability. The cost of purchasing a force plate is also quite high ($15,000-$20,000 CAD) relative to simple balance boards used for video games (<$100 CAD).

1.3.1.1 COP calculation using a force plate

To calculate true COP from tri-axial force plates, force and moments values from/about multiple axes are incorporated as seen in Eq. 2-1 (Winter, 2009):

$$COP_y = \frac{F_y \cdot C + M_x}{F_z} \qquad \textbf{Eq. 1-1}$$

where F_y is the force in the sagittal COP direction of interest, F_z is the force in the vertical direction, C is the height from the centre of the force plate to the feet, and M_x is the moment normal to the F_y and F_z plane.

When little to no shear forces or moments are present, the COP location is mostly a function of forces applied in the vertical direction. The inclusion of shear forces and moments into the equation allows for an accurate representation of COP in dynamic situations. As well, the ‘C’ height variable accounts for the height of any platforms that may be placed on top of the force plate surface.

1.3.2 Nintendo Wii Balance Board

The Nintendo Wii Balance Board (WBB) has recently had an increased presence in research, rehabilitation and exercise settings in place of force plates. It has a useable surface of 45 cm x 26.5 cm and contains one 16-bit pressure sensor located at each the four corners to register force in the vertical direction (Figure 1-1). Overall COP location and path can then be extrapolated using a series of equations (Please refer to Eqs. 2-2 and 2-2).

While the Nintendo Wii Balance Board was designed to complement the Nintendo Wii gaming console as a video game controller, it has also found its way into the rehabilitation world as well as being used in exercise programs for older adults. Past studies have included interviews discussing WBB use with rehabilitation specialists and direct care staff of patients or older adults in centres that had used the Nintendo Wii system (Fung et al., 2010; Higgins et al., 2010). General opinions were that using the Nintendo Wii was a good alternative treatment method that was easy and safe to operate, promoted wellbeing, and increased patient participation in rehabilitation. One game often used for the rehabilitation and exercise settings is the Wii Fit because the WBB encourages whole body movements and targets balance and fitness. However, to more specifically aid with rehabilitation for those who had sustained neurological injuries affecting balance and mobility, independent game based rehabilitation system using the WBB to increase trunk control, lower extremity stability, balance and controlled transfer of body weight have been developed and tested (Anderson et al., 2010; Gonzalez-Fernandez et al., 2010; Lange et al., 2010).

The WBB is also being used in an increasing number of research studies in lieu of an instrument grade force plate, due to its portability and affordability. The Wii Fit has been examined for use as a fall risk assessment in community-dwelling older women and it was found that the non-faller group had significantly better performance in the ‘Basic Step' module involving the WBB compared to the faller group (Yamada et al., 2011). As a fall prevention program, it was found that using the Wii Fit has the potential to safely improve balance in older adults (Williams et al., 2010). Another study placed the WBB on top of a force plate to analyze COP path differences from the force plate data between novice and experienced players for Wii Sports and Wii Fit, finding that children with previous experience with Wii Fit games showed greater movement quantity (Levac et al., 2010).

Characterization of WBB specifications is still in its early stages and there have only been two published scientific studies thus far specifically investigating the WBB’s performance with balance outcomes. Clark et al. (2010) measured a higher minimum detectable change in WBB force values than with the force plate. Linear correlations individually calculated for each sensor exceeded $R^2 = 0.999$, showing excellent linearity (Clark et al., 2010; Young et al., 2011). Clark et al. (2010) also found fairly high accuracy of COP coordinates on the WBB surface with percent errors of less than 3%. Pagnacco et al. (2011) determined COP resolution of the WBB to be approximately 0.5 mm.

The two studies also compared the performance of the WBB to force plates. Clark et al. (2010) compared test-retest reliability of COP path length measurements from the WBB and from an industrial grade force plate on two separate days for the following tasks: single limb sway with eyes open, single limb sway with eyes closed, double limb sway with eyes open and feet apart and double limb sway with eyes closed and feet together. They found that both devices showed excellent test-retest reliability but that the WBB had higher mean COP path length values. One limitation in the study by Clark et al. (2010) is the lack of direct comparison between a WBB and an industrial grade force plate. This could have been avoided by placing the WBB on the surface of the force plate and simultaneously collecting data from both devices. Pagnacco et al. (2011) studied device performance by placing the WBB on top of a commercially available posturography measuring device during quiet stance. The measured mean velocity of the COP path was higher with the WBB than the posturography device. However, Pagnacco et al. (2011) did not perform a statistical analysis and it is therefore unknown if the difference found was significant.

Specific WBB specifications such as drift, hysteresis, uniformity of response and COP accuracy have yet to be characterized. As well balance and gait researchers interested in using the WBB as a more feasible alternative to force plates in certain situation would benefit from an unbiased comparison of performance characteristic. However, no study to date has directly compared in-depth COP measures of the two devices.

1.3.2.1 Centre of pressure calculation using Nintendo Wii Balance Board

The calculation of true COP incorporates shear forces and moments, as well as an adjustment for platform heights (Section 1.3.1.1). However, WBBs only measure in the vertical direction, implying that shear forces and moments are not taken into account. The equations for COP in the x and y directions are shown below.

$$Xcop = \frac{FR+BR-FL-BL}{Fz} \times x_{dist} \qquad \textbf{Eq. 1-2}$$

$$Ycop = \frac{FL+FR-BL-BR}{Fz} \times y_{dist} \qquad \textbf{Eq. 1-3}$$

The x direction (ML) is along the long axis of the WBB, while the y direction (AP) is along the short axis of the WBB. FL, BL, FR and BR refer to the force values from the front left, back left, front right and back right sensors respectively (power button is along back side). Fz is the total sum of forces from all sensors. xdist and ydist represent the distance from the geometric centre of the WBB to the sensor location along the long and short axes respectively. See Figure 1-1 for sensor locations and variable names.

When two WBBs were used to calculate COP in Study 2, it was assumed that they were lined up along their longitudinal axes with a distance of 30 cm between the centres of the WBBs (Figure 3-2). The x axis then becomes the AP direction and the y axis is in the ML direction. The equations can be seen below in Eqs. 1-4 and 1-5.

$$COPx = (Xcop.left - 15cm) * \%left + (Xcop.right + 15cm) * \%right \qquad \textbf{Eq. 1-4}$$

$$COPy = Ycop.left * \%left + Ycop.right * \%right \qquad \textbf{Eq. 1-5}$$

Where Xcop is the ML COP location, Ycop is the AP COP location, and %left and % right refer to the percentage of mass distributed on the left and right WBB respectively.

1.3.2.2 Limitations of the Nintendo Wii Balance Board

The WBB COP calculation is more suited towards static loads because fewer forces and moments are generated in the non-vertical planes and therefore COP calculations would be more accurate without knowledge of other axial forces. More dynamic loads may present more forces in the non-vertical planes and require tri-axial information for more accurate calculations of COP. Without moment information, it is assumed that the forces are applied directly to the collection surface (i.e. no platform present). The sampling frequency of the WBB is approximately 100 Hz which is controlled by the device (Pagnacco et al., 2011). Although this is well above the frequency of daily static tasks (5-10 Hz) (Winter, 2009), tasks involving high frequencies such as jumping may not be collected accurately due to the low sampling frequency. However, this is an understandable limitation because the WBB was not designed to withstand impact, as Nintendo warns users to not jump while using the WBB.

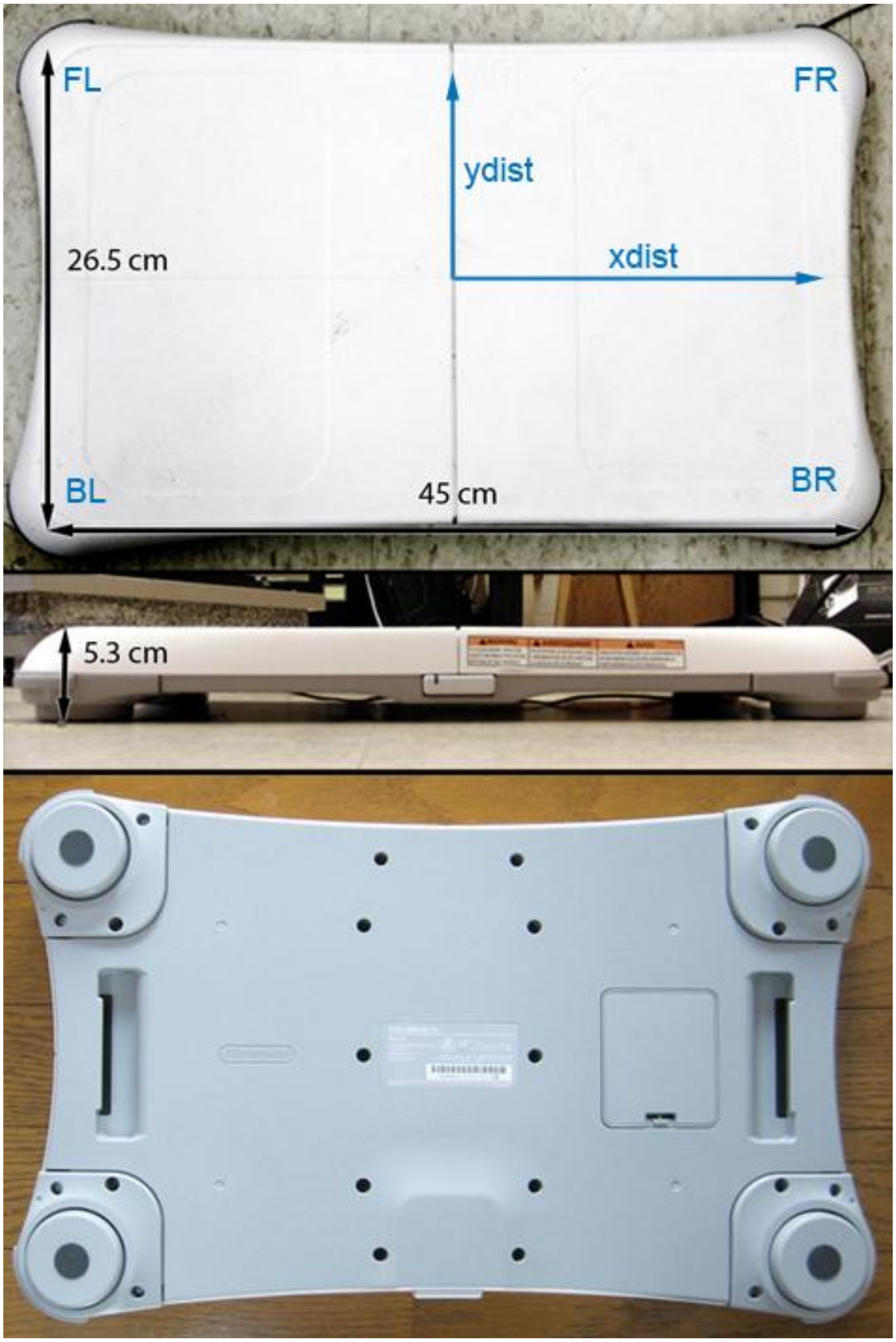


Figure 1-1: Dimensions of the Nintendo Wii Balance board (WBB); Top image - View of WBB from above; Middle - View of WBB from side; Bottom – Underside view of WBB

1.4 Thesis Objective and Summary of Studies

The primary objective of my thesis was to determine whether a range of safety floors, which have been shown to attenuate forces and accelerations during fall-related impacts, interfere with balance control during some activities of daily living in retirement home dwellers. There are two studies as part of this thesis. In the first study, I characterized Nintendo Wii Balance Board specifications and determine the appropriateness of its use in place of an instrument grade force plate in measuring balance responses. This was determined over a variety of laboratory specification tests, as well as a quiet stance task comparison between the balance board and force plate. The second study evaluated performances of both quiet stance and sit-to-stand tasks across traditional and safety floors to assess whether floor surfaces affect balance measures in older adults living in residential care facilities. It was hypothesized that i) values from centre of pressure measures from the WBB would be comparable to those from force plate outputs during quiet stance task based on suitability criteria determined from previous literature (Study #1) and ii) metrics of balance control derived from the underfoot centre of pressure on safety floors would not be significantly different compared to traditional flooring as measured from retirement home dwellers (Study #2).

Chapter 2
Nintendo Wii Balance Board Testing

2.1 Introduction

2.1.1 Existing types of force plates

As detailed in earlier in this thesis document, there are several existing types of force plates. Two commonly used brands presently used in biomechanics research are Advance Manufacturing Technologies Inc. (AMTI) and Kistler. There are sensors built into the force plates which register forces and moments, generally in three dimensional coordinates. The instrument grade force plates are extremely accurate and durable. The recordings are resistant to change in time or temperature. However, force plates have low portability due to mass (10 to 45 kg), and they are costly ($15,000-$20,000 for one). Further details can be found in Section 1.3.1.

2.1.2 Wii balance board

Nintendo Wii Balance Boards (WBB) (Nintendo, Redmond, WA, USA) have made their way into biomechanics research as a cheaper and more portable alternative to using force plates. There is a sensor in each of the four corners, measuring force in only the vertical direction. One issue with using a WBB to calculate COP is that shear forces and moments cannot be taken into account. As a consequence, these additional forces are neglected and problems may arise when trying to assess dynamic movements or static forces applied in the horizontal plane. It is also unknown how the WBB performs directly compared to the force plate during realistic balance measures. In this portion of my thesis, quiet stance (QS) data from a WBB was chosen to be evaluated against a force platform as it can be used to provide insight into balance sway measures and centre of pressure changes. Despite these unknowns, some researchers considering the use of the WBB in balance studies because it is very portable, weighing 3.5 kg, and affordable as it costs less than $100 CAD. Please refer to Section 1.3.2 for more information on the WBB.

2.1.3 Study rationale

Some data is currently known about the ability of WBBs to assess balance (refer to Section 1.3.2). However, there remain some important unknown performance characteristics. For example, load accuracy, unloading characteristics, drift, and COP accuracy have never been tested. High load accuracy and COP accuracy refer to the precision of known outputs compared against measured outputs and are desired so that studies using the WBB as a measurement tool can be considered reliable. Incremental loading and unloading characteristics are important because they show that a repeated change in mass will be measured as a consistent increase or decrease in force regardless of the existing static force loaded on the device. Drift calculations demonstrate whether the measured force output of a static load changes over time. This may occur if the device requires a 'warm up' time. It is also currently unknown if battery life affects the performance of the WBB. These unknowns described above are the main focus of this chapter of my thesis and are testing of these factors are outlined below in Section 2.2.

2.1.4 Purpose and hypotheses

The purpose of Study #1 was to test the hypotheses that: i) the specifications of the WBB are comparable to those of the gold standard AMTI OR6-6 laboratory grade force plate; ii) COP measurements taken on a WBB are in clinical agreement to those from a force plate; and iii) performance of the WBB is not affected by battery life.

2.2 Methods

2.2.1 Data collection

Vertical force data from a single Nintendo Wii Balance Board was collected at 64 Hz. The WBB was modified by a lab technician to be used with an external AC source, allowing the voltage to be varied, and therefore ‘battery life’ (i.e. voltage changes) to be assessed. The battery lives chosen were: 4.50 V (low), 5.25 V (medium), and 6.00 V (high). The WBB does not perform under 4.5 V of battery life. A custom software program written in the C# language (WBB Program V 1.5.1, Simon Jones, University of Toronto) was used to extract data from the WBB using a Bluetooth connector. The output from this program that was used in this thesis was the time-varying mass (kg) measured by each of the four vertical sensors. The origin was defined as (0, 0) at the geometric centre of the WBB. Where appropriate, measures from the WBB were simultaneously collected and compared to those from our laboratory AMTI force plate (AMTI, Watertown, MA, USA) and the WBB was placed directly on top of the force plate. In the event that a force plate was used, the device was allowed to ‘warm up’ for a period of 60 minutes, and set at a sampling rate of 256 Hz. The WBB was not given a warm up time. The study was performed in a biomechanics laboratory at the University of Waterloo.

2.2.2 Experimental Procedures

Two separate experiments were completed as part of this study. The first involved a detailed characterization of the WBB technical specifications. The second involved a comparison of the WBB output to that measured from a laboratory grade force plate (considered a gold standard) during quiet stance trials performed by volunteers. Each experiment is separately described below.

2.2.3 Experiment #1: Characterizing WBB specifications

Based on technical specifications often provided by the manufacturers for force plates, six WBB characteristics were tested**:** drift, linearity, hysteresis, mass accuracy, uniformity of response, and COP accuracy. All six tests were performed separately at low, medium, and high battery lives. A maximum of 150 kg of gym weight plates (six 20.45 kg (45 lb) plates, two 11.36 kg (25 lb) plates, and one 4.54 kg (10 lb) plate) was used to for testing, which matches the maximum recommended mass to apply to a WBB as determined by its manufacturer (Nintendo, Redmond, WA, USA). The mass accuracy test was collected simultaneously by a WBB placed on top of a force plate so that the applied masses could be precisely measured. Details on each test are provided below.

Static loading

Drift: The WBB was loaded with 150 kg for a duration of 120 minutes.

Mass loading and unloading test

The following test was used to extract linearity, hysteresis and mass accuracy data. A force reading was collected after every addition and/or removal of mass.

Linearity: To test linearity, mass were added to the centre of the WBB every 30 seconds through ten increments, up to a total of 150 kg.

Hysteresis: Following the linearity test, the ten loads were removed individually in 30 second intervals.

Mass Accuracy: The values from the loading and unloading responses of the linearity and hysteresis tests were used to calculate mass accuracy.

Spatial loading response

COP accuracy: A point mass of 20 kg was manually applied using a Chatillon force transducer (AMETEK, Largo, FL, USA) custom fitted with a screw end to fifteen (5 wide by 3 tall) points spread across the WBB. The set up and grid layout can be seen in Figure 2-1 and Figure 2-7.

Uniformity of response: 150 kg of mass was separately loaded over a 3 cm diameter spacer at each of the four corners, and at the geometric centre of the WBB.

2.2.3.1 Summary of Wii Balance Board characterization conditions

Effects from battery life of WBB: (1) WBBs x (3) battery levels x (6) tests at each battery level = 18 total tests from static load, loading/unloading, and spatial loading response procedures. The tests and outcome variables are summarized below in Table 2-1.

2.2.3.2 Data Analysis

All data analysis was completed using a custom software program written in Matlab (Mathworks, Natick, MA, USA). A 4^{th} order dual pass Butterworth filter with a 6 Hz cut-off frequency was applied to the data from both devices, with the cut-off frequency determined by performing a residual analysis on quiet stance force plate data. The 6 Hz cut-off was also found to be most appropriate for the WBB after the data was filtered at various cut-off frequencies (2 to 20 Hz) and compared the filtered force plate data.

This section outlines the calculations required for the six specification tests. Drift was defined as the absolute percent difference (Eq. 3-1) between the WBB at 0 min compared to at 30, 60 and 120 min. Linearity was expressed as a coefficient of determination R^2 value calculated using the force output from each load increment. Hysteresis was determined as a percentage difference between the areas (using the trapezoidal rule of calculating area Eq. 3-2) of individually loaded then unloaded gym weight plates (six 20.45 kg plates, followed by two 11.36 kg plates, followed by one 4.54 kg plate) from 0 to 150 kg. Mass accuracy was defined as the average percent difference between the mass measured by the WBB and the known mass measured from the force plate. Uniformity of response across the WBB was determined by calculating the individual percent differences between the outputs of a 150 kg load applied to the four sensors separately and of a 150 kg load at the centre of the board. COP Accuracy was determined by calculating the absolute difference between each calculated position using the WBB data, and its respective known COP location at each of the 15 grid locations.

$$\%\, Error = \left|\frac{x_2 - x_1}{x1}\right| x100\%, \qquad \textbf{Eq. 2-1}$$

where x_1 and x_2 represent the expected (FP) outcome and measured (WBB) outcome respectively.

$$f(x)dx \approx (b-a)\frac{f(a)+f(b)}{2}, \qquad \textbf{Eq. 2-2}$$

where a and b are two successive known loads, and f(a) and f(b) are the measured outputs.

Table 2-1: Wii balance board characterization tests to determine technical specifications

Measurement	Test	Outcome Variables
Drift	Place 150 kg load on top of WBB and measure output at 0, 30, 60 and 120 min	% error in output load at 30, 60 and 120 min compared to 0 min
Linearity	Individually load ten gym weight plates totaling 150 kg to the centre of the WBB	Linear regression correlation
Hysteresis	Individually load then unload gym weight plates totaling 150 kg from the centre of the WBB	% error between loading and unloading areas
Accuracy	Same test as hysteresis	% error between known and measured mass at each increment
COP accuracy	20 kg point mass applied in to fifteen points (5 wide by 3 tall)	% error between known and measured COP location at each grid unit
Uniformity of response	150 kg mass separately applied to 4 corners and centre	Compare % error of COP location output from 4 sensors to output from centre of WBB

Figure 2-1: Equipment setup for COP accuracy testing. The Chatillon force transducer was custom fitted with a screw end and loaded with 20 kg of force at each of the 15 grid points.

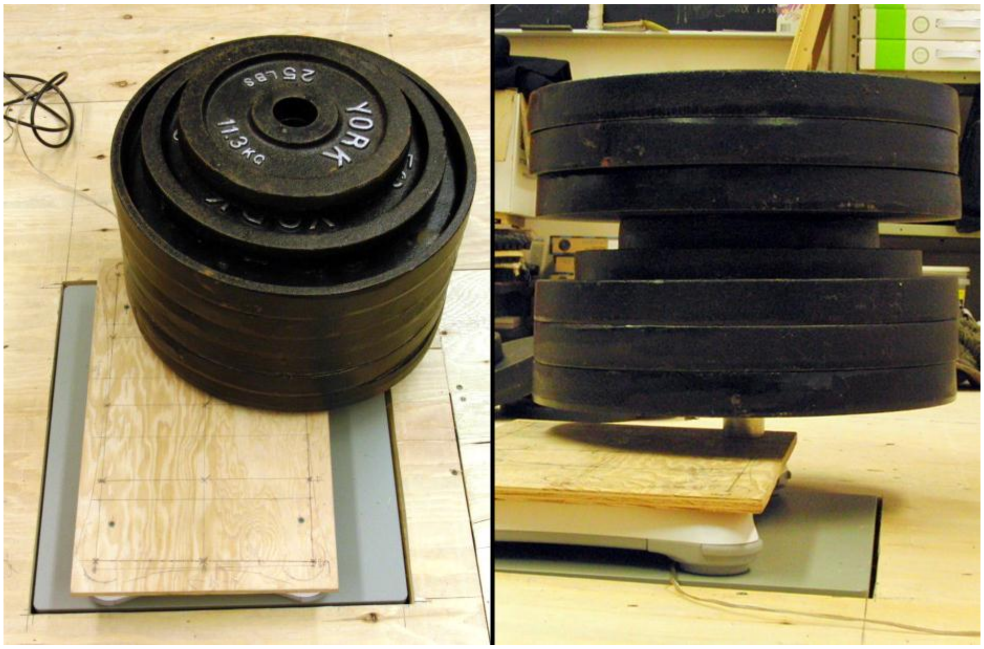


Figure 2-2: Equipment setup for uniformity of response testing. 150 kg was loaded on a spacer measuring 3 cm in diameter at each of the 4 corners and at the centre of the WBB

2.2.3.3 WBB suitability criteria

There are several criteria that should be taken into consideration when deciding whether a WBB can be used in place of a force plate to measure indices of balance control including similarity in output measures, and the intended use of the resulting data (e.g. whether clinical decisions are going to be driven by the outputs). My thesis focuses on the former category. I used the following criteria as a means to evaluate the appropriateness of the WBB for balance assessment purposes:

a. Drift, hysteresis, and uniformity of response should have less than 1% error.

b. Linearity should have a R^2 correlation value of greater than 0.99 (Clark et al., 2010).

c. Mass accuracy should have less than 3% error (Clark et al., 2010).

d. COP accuracy should have less than 10 mm error.

2.2.4 Experiment #2: WBB – force plate comparison during quiet stance

2.2.4.1 Participants

A total of six young individuals (5 female, 1 male) participated in the study. Participant characteristics were as follows: mean (SD) age = 22.5 (2.0) years, height = 170.3 (10.3) cm, mass = 68.6 (14.4) kg. Inclusion criteria included no lower limb musculoskeletal disorders in the past six months, and the ability to stand without taking a step or altering the base of support for 60 seconds.

2.2.4.2 Experimental procedure

During a single data collection session participants performed completed quiet stance (QS) trials with eyes open and eyes closed. For all trials the participant stood on a WBB positioned on top of a force plate (Figure 2-3 and Figure 2-4). The WBB was powered by an external source that allowed the experimenter to simulate three 'battery life' conditions: low, medium and high as described in Section 2.1.1.

For all trials participants were instructed to stand with their hands at their sides, with their heels shoulder width apart, determined by the distance between the acromion. Participants self-selected the stance angle during the first practice trial, which was standardized for the rest of the session. For the eyes open condition, participants were instructed to look straight ahead at an imaginary target on the wall (Baloh et al., 1994; Stel et al., 2003; Mackey and Robinovitch, 2005). During eyes closed trials, participants maintained their head position as if holding a straight gaze. Three trials were performed per condition. Conditions included: eyes open at each battery life, and eyes closed at full battery life only. Each trial was 30 seconds in duration. The order of each task was randomized across participants. To synchronize the data from the WBB and force plate, a 9.1 kg mass was removed from the WBB at the start of each collection. Details on the experimental conditions are summarized in Table 2-2.

Figure 2-3: Set up of testing environment for quiet stance tasks. The WBB is placed directly on top of the force plate.

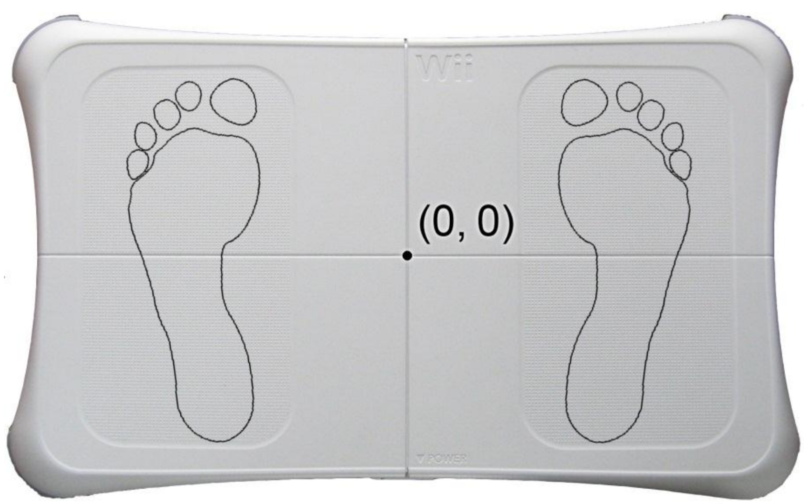


Figure 2-4: Orientation of feet during the quiet stance trials.

Table 2-2: Subject task conditions to compare Wii balance board and force plate responses

Condition	Protocol Details
Tasks (2)	Quiet stance - eyes open (QSEO) Quiet stance - eyes closed (QSEC)
Battery (3)	QSEO only - low, medium, and high
Trials (3)	Three trials per task and battery combination
Participants (6)	5 female, 1 male; 22.5 (2) years; 170.3 (10.3) cm; 68.6 (14.4) kg

2.2.4.3 Data Analysis

A trial was considered successful if the participant did not need to take a step during the quiet stance task. Data was analyzed and filtered as in Experiment #1 using the same custom software program with a 6 Hz cut-off frequency as that from the characterization experiment (Section 2.2.3). Time-carrying COP trajectories were calculated in the anterior-posterior (AP) and medial-lateral (ML) directions over the 30 second trials (Eq. 2-1 for force plate data, and Eqs. 1-2 & 1-3 for WBB data).

The two devices were synchronized at the time when a 20 kg mass was removed from the WBB surface. This point was picked in the custom Matlab program by finding the time at which the total force on each device decreased by 1 kg. Four common COP summary variables were calculated in the AP and ML direction during the quiet stance (Prieto et al., 1996; Thapa et al., 1996; Boulgarides et al., 2003; Mackey and Robinovitch, 2005; Laing and Robinovitch, 2009) - equations involved are presented at the end of this paragraph. Based on differences in origins across devices, maximum range of COP displacement (QS_{range}) was calculated from the force plate data using Eq. 2-3, and calculated using Eq. 2-4 for the WBB data. Root mean square (QS_{RMS}) and was defined as per Eq. 2-5, providing a distinct COP location over the trial. Mean velocity of COP (QS_{vel}) was defined as the total COP distance travelled over the trial, divided by the duration of the trial (i.e. 30 s) as in Eq. 2-6 (Bohannon, 1995; Laing and Robinovitch, 2009). Mean sway frequency (QS_{freq}) was calculated using Eq. 2-7, which represents the number of revolutions (i.e. cycles) completed in one second of a distance equivalent the ratio of the mean velocity to mean distance travelled. (Maki et al., 1994; Prieto et al., 1996; Mackey and Robinovitch, 2005).

$$QS_{range\_FP} = highest\ COP\ value - lowest\ COP\ value \qquad \textbf{Eq. 2-3}$$

$$QS_{range\_WBB} = highest\ COP\ value + |lowest\ COP\ value| \qquad \textbf{Eq. 2-4}$$

$$QS_{rms} = \sqrt{\frac{{x_1}^2+{x_2}^2+\cdots+{x_n}^2}{n}} \qquad \textbf{Eq. 2-5}$$

where $x_1 \ldots x_n$ represent the COP locations over the trial; n is the total number of frames.

$$QS_{vel} = \frac{total\ COP\ distance}{t} \qquad \textbf{Eq. 2-6}$$

where t is the trial length.

$$QS_{freq} = \frac{mean\ velo}{2\pi(average\ distance - mean\ COP)} \qquad \textbf{Eq. 2-7}$$

2.2.4.4 Statistical Analysis

Due to the nature of the WBB characterization tests, no explicit statistical approaches were employed for Experiment #1.

For Experiment #2, one way repeated analysis of variance (ANOVA) was performed to assess potential effects of battery life on the WBB output during the eyes open quiet stance trials. Comparisons between the WBB and force plate results were conducted only on the eyes closed data at full battery life because we expected that the eyes closed data would be more dynamic and therefore be the condition which had the largest effects on WBB accuracy. Specifically, a one factor, repeated ANOVA was conducted to analyze the effect of device on each balance variable. In addition, average percent differences were computed across devices for each variable. Finally, Bland-Altman plot analyses were conducted for each variable as they are the preferred method when comparing the difference in results from a new method (WBB) to a gold standard (force platform) (Giangregorio and Cook, 2008). In addition to plotting the 95% confidence intervals, I interpreted the Bland-Altman data with respect to differences reported in the literature between older adult fallers and non-fallers (a means of assessing the clinical significance of the WBB-force plate differences). The clinical values for AP and ML range were drawn from Melzer et al. (2010), who tested community dwelling fallers and non-fallers in a narrow quiet stance eyes closed test. The remaining RMS, velocity and frequency values in the AP and ML directions were taken from Maki et al. (1994), who conducted a prospective study of community dwelling older adults over a one-year period. If the differences between the devices lie between the differences between faller statuses, then I considered the WBB-force platform discrepancies to have limited clinical significance. All statistical analyses were conducted with a significance level of 0.05 using statistical analysis software (SPSS Version 18.0, SPSS Inc., Chicago, IL, USA).

2.3 Results

2.3.1 Experiment #1: WBB Characterization

The results of the characterization tests for drift, linearity, hysteresis, mass accuracy and uniformity are shown in Table 2-3. The WBB system produced a coefficient of determination R^2 value of 1.0 across all battery lives when testing for linearity (Figure 2-5). Hysteresis was less than 0.12% during incremental loading and unloading (Figure 2-5). The WBB displayed a mass accuracy that ranged from 99.39% to 99.80% and a uniformity of response of 99.39% to 99.79% across different loading regions and battery lives (Table 2-3). The drift of the WBB at 30, 60 and 120 min was less than 0.12% when compared to the start of the trial (Figure 2-6). Results from each specification test at every battery level met the cut-off criteria for WBB usage in Study 2 (Section 2.2.3.3). The suitability criteria for the characterization tests set in Section 2.2.3.3 have mostly been met in this part of the study. All of drift, hysteresis and uniformity of response tests produced less than 1% difference between WBB and force plate readings. The linearity had an R^2 correlation value of 1.0, which was greater than the 0.99 criterion. Mass accuracy was greater than 99%, which meets the < 3% error criterion.

Variations in COP accuracy across locations on the WBB surface can be seen Table 2-4 and Figure 2-7. The centre of pressure measurements were most accurate at the centre and decreased in the accuracy as the load distance from the centre increased. Specifically, the difference between the known and measured centre of pressure locations across the WBB ranged from 0.1 mm to 16.0 mm, varying by a maximum of 2.4 mm at the centre of the WBB to a maximum of 16.0 mm at the corners. While the COP accuracy was higher near the centre of the WBB (Columns B,C,D in Figure 2-7), displaying less than a 10 mm difference of measurement and meeting the criteria, the difference was greater than 10 mm at the outer edges (columns A and E in Figure 2-7), not meeting the criterion.

Table 2-3: Summary of drift, linearity, hysteresis, mass accuracy and uniformity results from specification testing.

	Drift (% change)	**Linearity (R^2)**	**Hysteresis (% diff)**	**Mass Accuracy (% diff)**	**Uniformity (% diff)**
High battery	< 0.04	1.00	0.12	0.20 (0.23)	- 0.61 (0.29)
Medium battery	< 0.20	1.00	0.01	0.61 (0.29)	- 0.21 (0.51)
Low battery	< 0.05	1.00	0.12	0.25 (0.26)	- 0.46 (0.33)

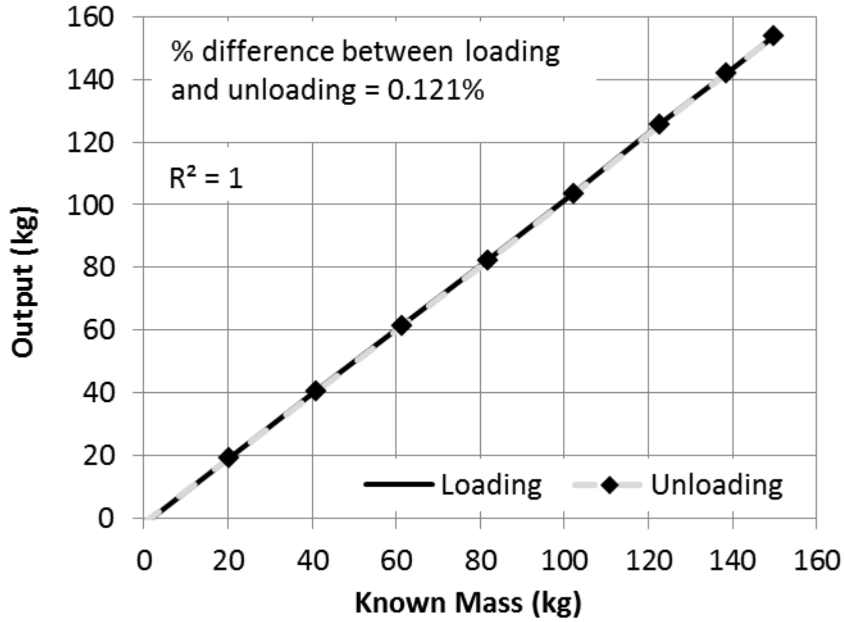


Figure 2-5: WBB measurements from the loading and unloading of 0 to 150 kg of mass on the WBB surface. Linearity is represented by the R_2 correlation of determination value, while hysteresis is represented by the % difference.

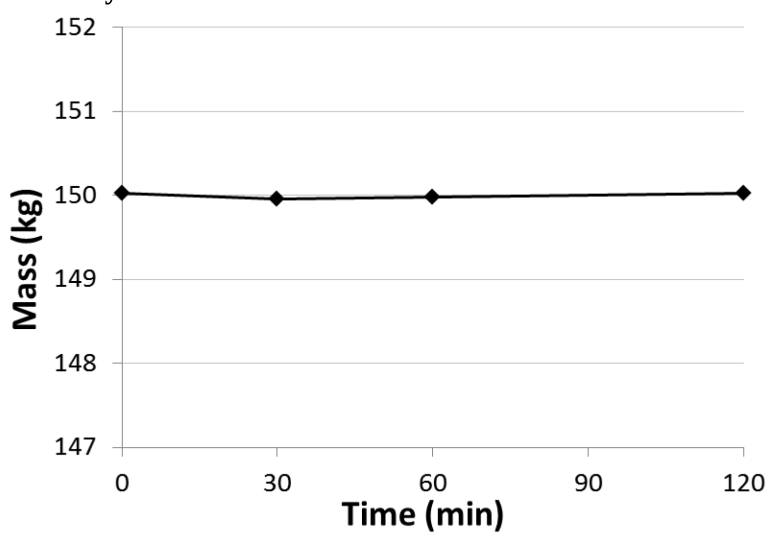


Figure 2-6: Mass measurements from a WBB loaded with a 150 kg mass over a two hour period.

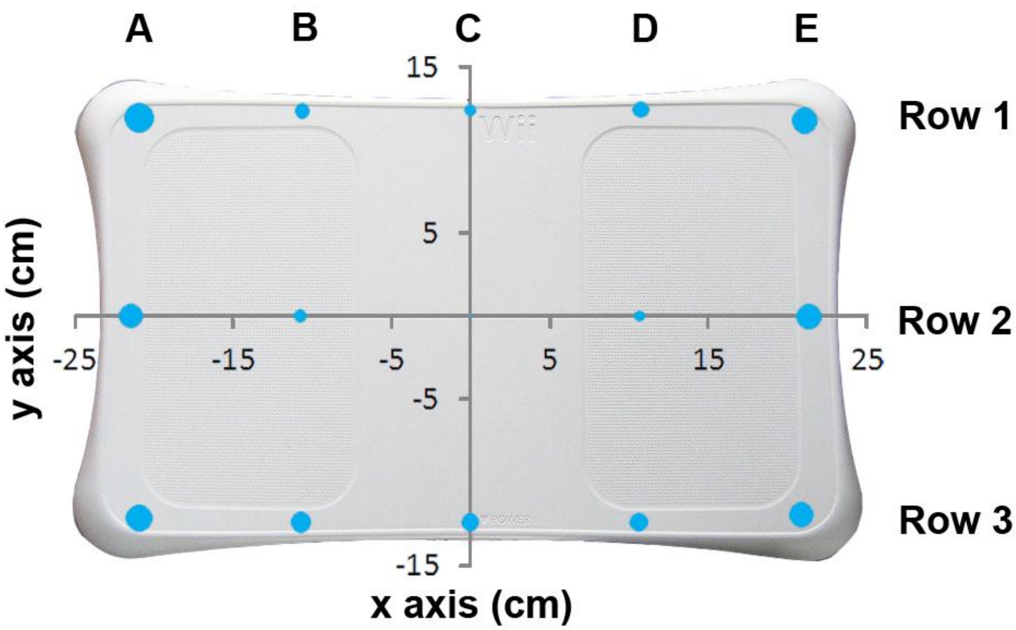


Figure 2-7: Results from COP accuracy testing. The radius of each circle represents the difference between the known COP location and the COP measurements recorded each grid point. The difference is calculated as the resultant distance of the horizontal and vertical difference vectors.

Table 2-4: Results from COP accuracy testing. The difference between the known and measured COP location at each grid point is shown.

	Difference between known and measured COP location (mm)					
	Full battery		Half battery		Low battery	
COP location	COPx	COPy	COPx	COPy	COPx	COPy
Row 1.A	16.0	5.6	14.3	7.8	10.9	8.2
Row 1.B	7.4	5.4	9.3	6.6	6.9	6.9
Row 1.C	0.8	7.4	0.5	6.6	3.1	9.8
Row 1.D	7.2	7.1	4.0	7.4	1.6	9.5
Row 1.E	14.1	5.5	10.2	7.1	10.3	7.9
Row 2.A	13.8	0.8	11.7	2.9	11.7	1.1
Row 2.B	8.2	0.5	7.7	1.1	7.1	0.2
Row 2.C	1.1	2.4	0.3	0.5	1.1	0.4
Row 2.D	6.9	0.1	2.9	0.4	5.2	0.5
Row 2.E	14.2	2.7	13.1	2.3	10.1	0.8
Row 3.A	12.1	10.5	2.9	0.4	11.3	6.2
Row 3.B	8.4	8.4	13.1	2.3	7.8	4.9
Row 3.C	0.8	10.3	9.9	8.1	0.7	7.7
Row 3.D	6.3	9.1	9.6	6.3	5.2	8.2
Row 3.E	11.1	8.9	6.5	8.0	9.3	7.4

2.3.2 Experiment #2: WBB vs. FP during quiet stance

A one way repeated ANOVA was used for the QS data to test for the effect of device (repeated: 2 factors) on any differences between the force plate outcome variables and WBB outcome variables. Significant differences were shown in the following measures: AP QS_{range} ($F_{1,10}$ = 74.858, p = 0.000), AP QS_{RMS} ($F_{1,10}$ = 57.285, p = 0.001), ML QS_{range} ($F_{1,10}$ = 32.390, p = 0.002), ML QS_{RMS} ($F_{1,10}$ = 33.084, p = 0.002), ML QS_{vel} ($F_{1,10}$ = 12.936, p = 0.016). The remaining factors showed no significant between devices: AP QS_{vel} ($F_{1,10}$ = 0.567, p = 0.485), AP QS_{freq} ($F_{1,10}$ = 5.405, p = 0.068), ML QS_{freq} ($F_{1,10}$ = 0.321, p = 0.596). To visualize the trends, the WBB data has been overlaid on the force plate data during one quiet stance eyes open trial in Figure 2-8.

Since statistical differences were found, a more detailed analysis was performed to see if the differences were clinically significant. A comparison of devices revealed that the devices had a discrepancy that ranged from 1% (velocity) to 4% (RMS) and 3% (frequency) to 13% (range) for AP and ML directions respectively (Figure 2-8 and Figure 2-9). The high discrepancy in the significantly different ML variables amounts to an actual difference of 1.3 cm in range, 0.2 cm in QS_{RMS}, and 0.5 cm/s in QS_{vel} measurements. Significantly different range and QS_{RMS} variables in the AP and ML direction amounted to a difference of 1.5 mm and 0.4 mm respectively (Table 2-7). Additional Bland-Altman plots were created for each balance sway measure to provide a more visual comparison between the devices. The plots revealed that most points lay within the 95% confidence intervals in both AP and ML directions, showing low variation in the difference between force plate and WBB values. There are between zero and one outliers outside of the confidence intervals in each balance variable. As well, the differences in the WBB calculations do not tend to increase or decrease as the magnitude of the force plate data increases. The differences between the devices were less than the clinically significant margins between fallers and non-fallers with the exception of QS_{freq} in the ML direction. The percentage of device differences compared to faller and non-faller differences can be found in Table 2-7. The device differences were less than 40% of the clinically significant ranges. The Bland Altman plots from the quiet stance eyes closed data can be seen in Figure 2-11. Data points from the WBB were also plotted against the force plate data over a y = x line to gain a sense of correlation. The WBB calculations underestimated those from a force plate in three of eight variables: ML QS_{range}, ML QS_{RMS} and ML QS_{vel} (Figure 2-12).

Regarding the influence of battery life, ANOVA showed that battery life (3 levels) did not significantly affect the WBB output across AP QS_{range} ($F_{2,10}$ = 0.037, p = 0.964), AP QS_{RMS} ($F_{2,10}$ = 0.550, p = 0.593), AP QS_{vel} ($F_{2,10}$ = 0.944, p = 0.421), AP QS_{freq} ($F_{2,10}$= 0.017, p = 0.983), ML QS_{range}

($F_{2,10}$ = 0.783, p = 0.483), ML QS_{RMS} ($F_{2,10}$ = 3.106, p = 0.089), ML QS_{vel} ($F_{2,10}$ = 0.586, p = 0.575), and ML QS_{freq} ($F_{2,10}$ = 0.074, p = 0.929). Subsequent statistical analyses presented in this section were performed on data from the full battery life, eyes closed condition only as there was no effect of battery life.

Table 2-5: ANOVA table comparing effects of battery life on balance variables during quiet stance with eyes open

	ANOVA	
	Battery Life	
Anterior-posterior	F	p
Range (cm)	0.037	0.964
RMS (cm)	0.550	0.593
Mvelo (cm/s)	0.944	0.421
Mfreq (Hz)	0.017	0.983
Medial-lateral	F	p
Range (cm)	0.783	0.483
RMS (cm)	3.106	0.089*
Mvelo (cm/s)	0.586	0.575
Mfreq (Hz)	0.074	0.929

* indicates p < 0.05 for effect of battery life

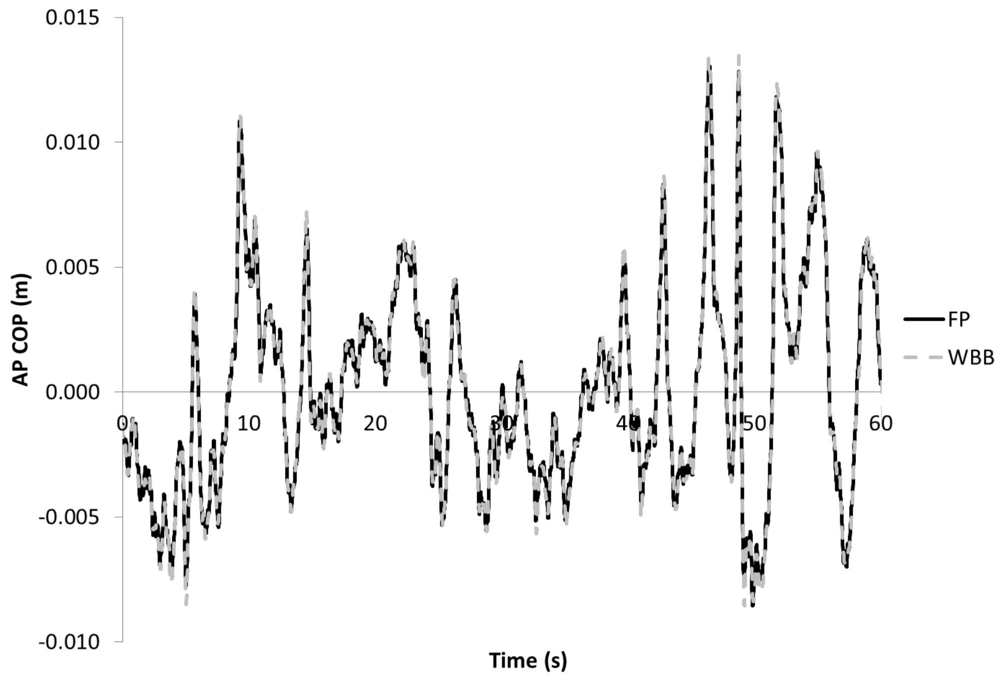


Figure 2-8: Sample centre of pressure data in the anterior-posterior direction over a 60 second quiet stance eyes open trial for one participant

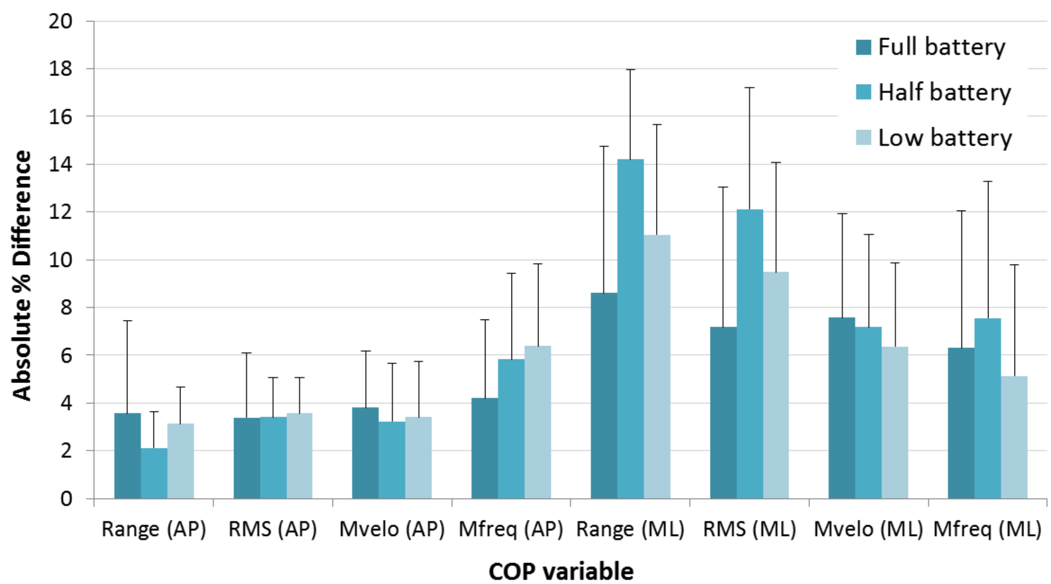


Figure 2-9: Quiet stance eyes open task. Mean percent difference between force plate and Nintendo Wii Balance Board balance variable calculations across low, medium and high battery conditions.

Table 2-6: Range, RMS, velocity and frequency results from quite stance eyes open test. The four balance variables were calculated using WBB data at low, medium and high battery lives and compared to synchronized COP measurements from the force plate.

	Quiet stance eyes open	Range (cm)	RMS (cm)	Velocity (cm/s)	Frequency (Hz)
AP	Force plate	1.79 (0.35)	0.40 (0.10)	0.47 (0.10)	0.27 (0.06)
	100% WBB	1.71 (0.31)	0.38 (0.09)	0.46 (0.10)	0.28 (0.08)
	Force plate	1.84 (0.69)	0.41 (0.13)	0.50 (0.13)	0.27 (0.06)
	75% WBB	1.89 (0.74)	0.41 (0.15)	0.52 (0.14)	0.28 (0.07)
	Force plate	1.98 (0.50)	0.44 (0.12)	0.53 (0.11)	0.28 (0.09)
	50% WBB	1.94 (0.49)	0.42 (0.11)	0.53 (0.11)	0.29 (0.09)
ML	FP Eyes Open	0.73 (0.30)	0.14 (0.06)	0.44 (0.25)	0.80 (0.41)
	WBB Eyes Open	0.65 (0.26)	0.12 (0.06)	0.41 (0.22)	0.84 (0.41)
	Force plate	0.89 (0.50)	0.15 (0.07)	0.44 (0.16)	0.76 (0.31)
	75% WBB	0.77 (0.44)	0.13 (0.07)	0.40 (0.13)	0.80 (0.31)
	Force plate	0.95 (0.30)	0.17 (0.06)	0.47 (0.14)	0.72 (0.36)
	50% WBB	0.84 (0.27)	0.15 (0.05)	0.43 (0.12)	0.73 (0.35)

* indicates $p < 0.05$ for battery life effects

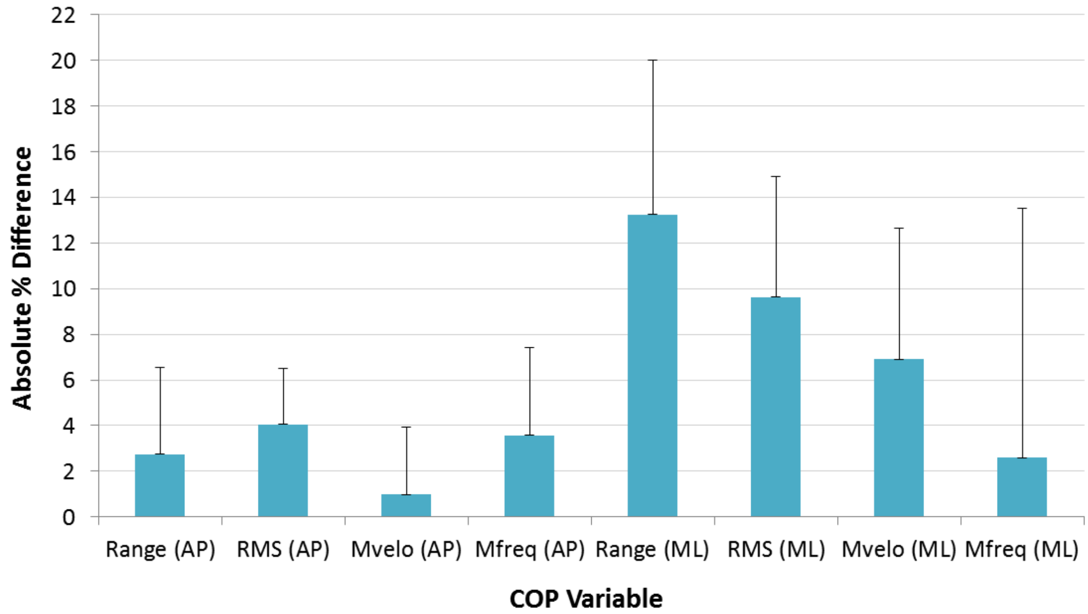


Figure 2-10: Quiet stance eyes closed task. Mean percent difference between force plate and Nintendo Wii Balance Board balance variables.

Table 2-7: Range, RMS, velocity and frequency results from quite stance eyes closed test. The four balance variables were calculated using WBB data full battery and compared to synchronized COP measurements from the force plate. The difference between the device values were divided by the difference between fallers and non-fallers to obtain a percentage (Maki et al., 1994; Melzer et al., 2010).

		Range (cm)	**RMS (cm)**	**Velocity (cm/s)**	**Frequency (Hz)**
AP	FP Eyes Closed	2.40 (0.53)	0.49 (0.12)	0.71 (0.14)	0.33 (0.08)
	WBB Eyes Closed	2.25 (0.54)*	0.45 (0.13)*	0.68 (0.17)	0.34 (0.09)
	% Ratio of device difference to faller/non-faller difference	26.3%	40.0%	5.0%	16.7%
ML	FP Eyes Closed	0.94 (0.30)	0.16 (0.05)	0.49 (0.18)	0.75 (0.30)
	WBB Eyes Closed	0.81 (0.28)*	0.14 (0.05)*	0.44 (0.16)*	0.76 (0.30)
	% Ratio of device difference to faller/non-faller difference	17.8%	16.7%	23.5%	20%

* indicates $p < 0.05$ for effect of device – see Table 2-8 for details

Table 2-8: ANOVA table comparing effects of device (WBB or force plate) on balance variables during quiet stance with eyes closed

	ANOVA	
Anterior-posterior	F-ratio	p-value
Range (cm)	74.858	0.000*
RMS (cm)	57.285	0.001*
Mvelo (cm/s)	0.567	0.485
Mfreq (Hz)	5.405	0.068
Medial-lateral	F-ratio	p-value
Range (cm)	33.390	0.002*
RMS (cm)	33.084	0.002*
Mvelo (cm/s)	12.936	0.016*
Mfreq (Hz)	0.321	0.596

* indicates $p < 0.05$ for effect of device (WBB and force plate)

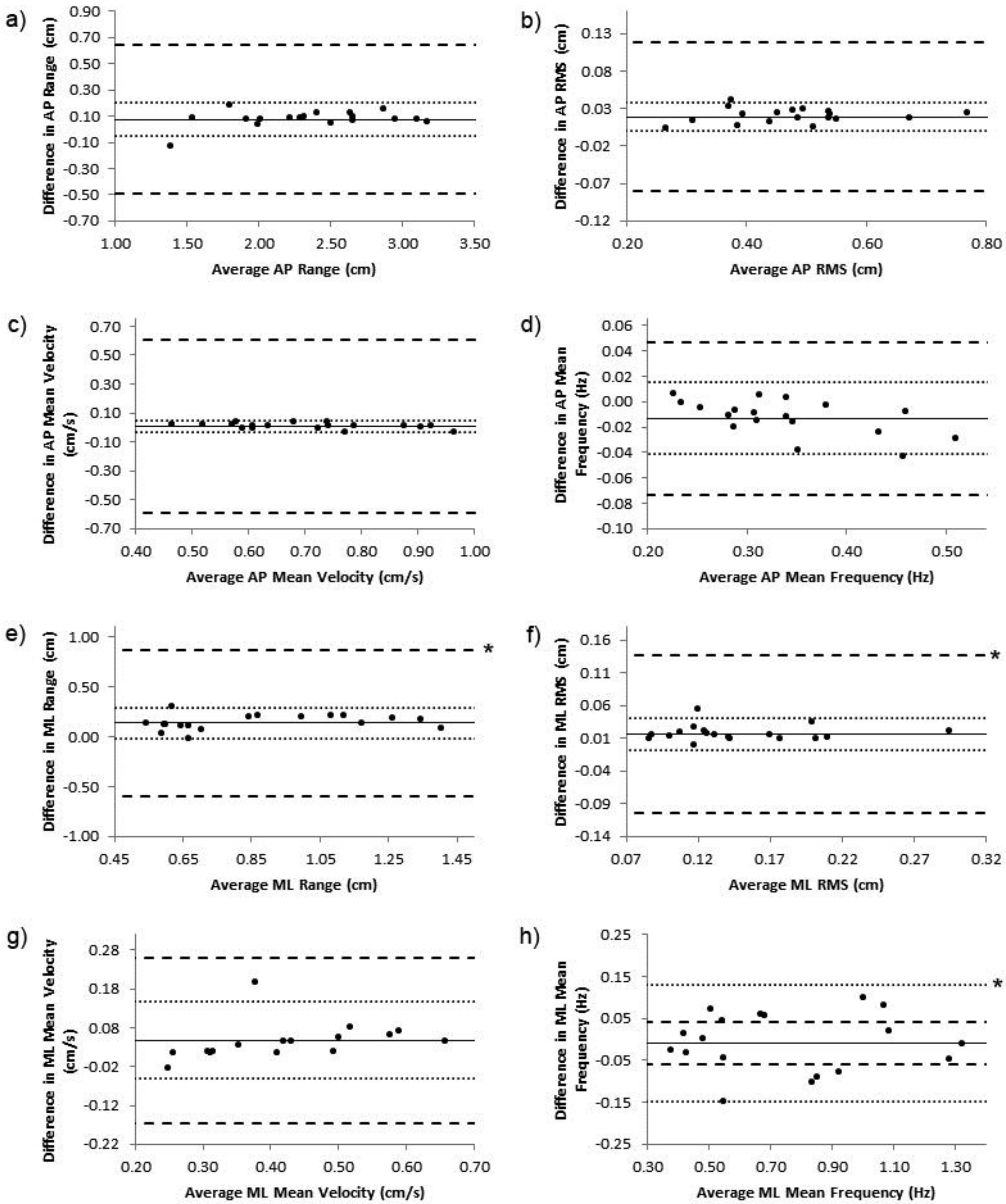


Figure 2-11: Bland-Altman plots from quiet stance eyes closed data. The dotted lines represent 95% confidence intervals. The dashed lines represent the difference between older adult fallers and non-fallers. A star (*) beside a dashed line indicates that the difference was significantly different in the research source. a) and e) reference lines from Melzer et al. (2010). b), c), d), f), g), h) reference lines from Maki et al. (1994).

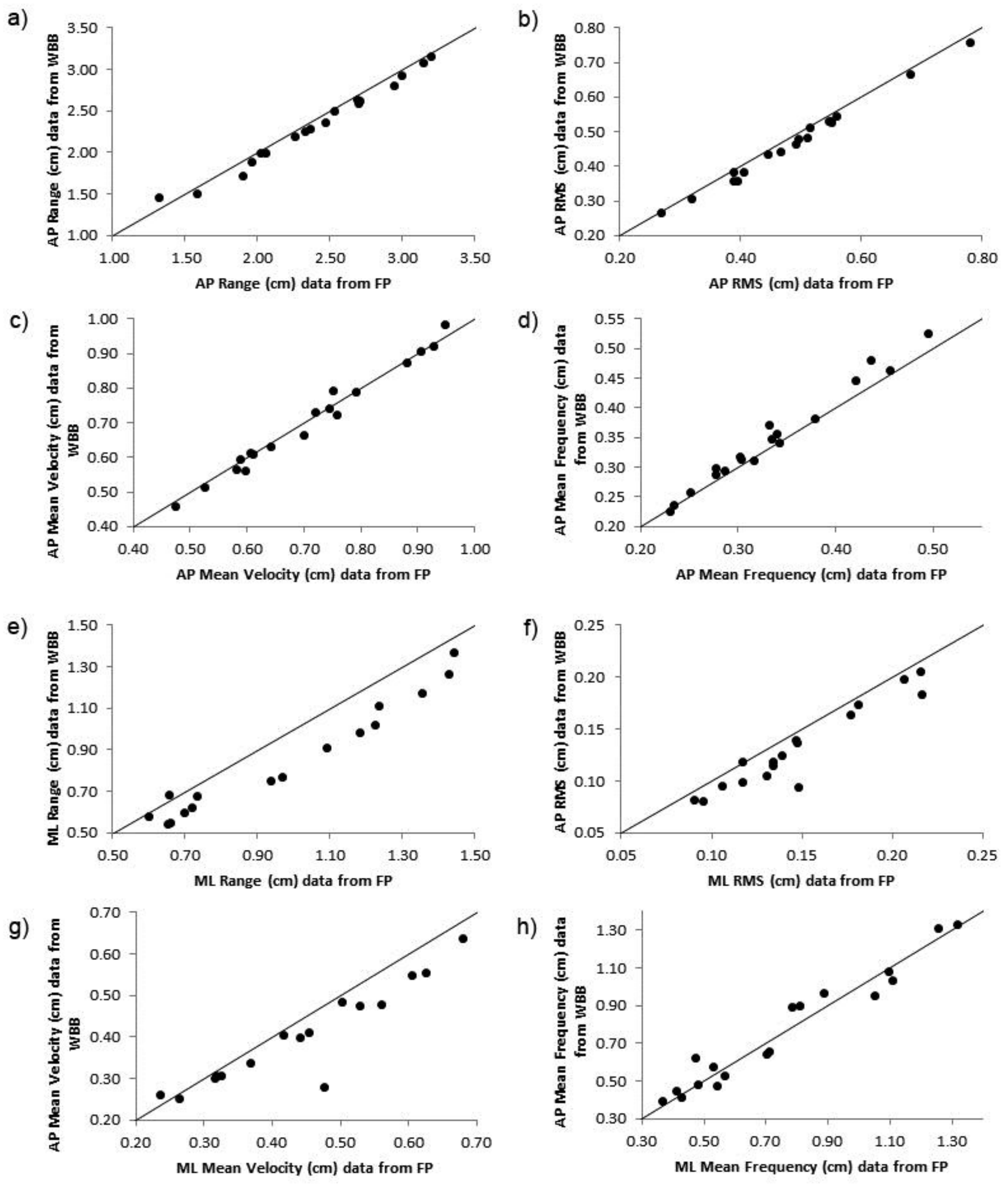


Figure 2-12: Linear relationship portion of Bland-Altman tests. The COP calculations from the gold standard force plate are plotted on the x-axis. The COP calculations from the Nintendo Wii Balance Board are plotted on the y-axis.

2.3.3 Discussion

The WBB has been used in balance research in place of force plates without having been properly tested as a suitable alternative. In this Chapter, I tested the appropriateness of using the WBB in biomechanics research through specification testing, and through direct comparisons to balance measurements using the force plate. The study was supportive of the hypothesis that the specifications (six of six tests) of the WBB would be comparable to those of a laboratory grade force plate. Three of four criteria were fully met and the remaining criterion was partially met. It was supportive of the hypothesis that COP measurements on a WBB were in clinical agreement to those taken on a force plate. Lastly, the study was supportive of the hypothesis that performance of the WBB was not affected by battery life. More details are provided in the following paragraphs.

In agreement with my first hypothesis, I have shown in Experiment #1 that when compared to a force plate, the WBB meets the suitability criteria in terms of loading response and accuracy. The suitability criteria set in Section 2.2.3.3 include: i) less than 1% error for drift, hysteresis and uniformity of response; ii) a R^2 correlation value of greater than 0.99 for linearity; iii) less than 3% error for mass accuracy and COP accuracy. The WBB system showed excellent linearity across incremental input ranges, producing a highest possible R^2 correlation value of 1.0 across all battery lives. The test for linearity met the suitability criteria. There was very minimal (<1% error) hysteresis during incremental loading and unloading, showing that varying loads over time do not affect the reliability of WBB measurements. The results of the linearity and hysteresis tests are of interest to researchers who wish to measure changes in load over time. In addition, the WBB had high total mass accuracy, showing less than 1% difference between ten measured mass values compared to their known mass values. The WBB measurements were consistent across the board surface, as there was less than 1% difference between measurements of a 150 kg load at each of the four corners compared to at the centre of the surface. Researchers can be confident that the loads output by the WBB are representative of their true values. The amount of drift over a two hour period was under 1%, meeting the acceptable criteria and showing good consistency in WBB measurements. The accuracy of centre of pressure values recorded by the WBB changed depending on the location of applied load. At the corners, the difference between the measured and known values averaged to approximately 15 mm, which is over the 10 mm suitability criteria. However, the clinical relevance of this test location is questionable, as it is likely that a substantial portion of the foot would need to hang over the edge of the WBB in order to position the COP in these corner locations. In contrast, the difference decreased as the COP location approached the centre of the WBB, at which the difference was approximately 1 mm (lower than the criteria set in Section 2.2.3.3). Related implications are that tandem stance tasks should not be evaluated using the WBB because portions of the feet may approach or extend beyond the edges of the WBB (where COP accuracy is poorest). For similar reasons, a wide stance during quiet standing should not be evaluated unless using one WBB per foot. However, the WBB is an appropriate alternative for force plates when analyzing balance control during a narrow quiet stance task with eyes open, a Romberg's test, or regular quiet stance with eyes open or closed. The overall message is that researchers should be aware that output accuracy is very good when the COP location is near the centre of the WBB, but that some caution is warranted when performing tests near the edge of the board.

As hypothesized, when comparing COP range, RMS, velocity and frequency in both the anterior-posterior and medial-lateral directions for the quiet stance eyes open task, my tests showed that battery life does not statistically affect the performance of the WBB. Following this test, only the quiet stance eyes closed (QSEC) task at full battery life was chosen to be subsequently analyzed because of its relevancy to Study 2. While most centre of pressure measurements from the QSEC WBB data in the

AP direction were relatively close to those output from a force plate (~4% difference), the gap between the devices increased to as much as 13% in the ML direction. However, it should be noted that the percent differences between devices amounted to an average difference of less than 1.5 mm in significantly different AP variables (range and RMS), less than 1.3 mm in the ML direction for the range and RMS variables, and less than 0.5 mm/s for ML velocity. Despite these statistical differences, I believe these between-device differences are acceptable because they are within the clinical significance boundaries between fallers and non-fallers as supported by the Bland-Altman plots. One variable that did not follow the trends was frequency in the ML direction. While the ~2% difference between the devices for ML frequency was not significant ($p > 0.05$), the difference was greater than the clinically significant difference between fallers and non-fallers according to Maki (1994). However, the differences may not be comparable because while ML frequency values of older adults during quiet stance in the literature range between 0.41 – 0.50 Hz (Maki et al., 1994; Freitas et al., 2005; Mackey and Robinovitch, 2005), my values for young adults were 0.75 Hz and 0.76 Hz for the force plate and WBB respectively.

As my WBB characterization study is novel, there are few detail in the literature that are relevant for comparison purposes. Two studies that include basic system characteristics are Clark et al. (2010) and Pagnacco et al. (2011). My linearity values ($R^2 = 1.00$) are in agreement with those from Clark et al. (2010) who measured excellent linearity ($R^2 = 0.99$). They also found COP accuracy values of less than 3% (difference between known and output locations) in an 8.4 cm by 8.4 cm square around the centre of the WBB. My greater detailed analysis approach demonstrated a similar degree of accuracy when loads were applied close to the centre of the board; however, my approach found that this accuracy decreased to potentially unacceptable levels when the COP approached the edges of the board. When comparing quiet stance force plate measurements to a WBB, Clark et al. (2010) found that the WBB overestimated COP path length values. Similarly, Pagnacco et al. (2011) compared a posturography measuring device to WBB measurements and found that the WBB had higher velocity values. In contrast to this, my findings from comparing balance variables showed that the WBB slightly underestimated the force plate values (see scatterplots in Figure 2-12) as shown by the linear comparison plots. However, the differences observed by Clark et al. (2010) may be erroneous as they did not directly compare device data from the same trials – separate trials were collected on a WBB and a laboratory grade force plate. The difference that Pagnacco et al. (2011) found may be due to the fact that they used a CAPS Lite posturographic force platform (can only measure vertical ground reaction force like a WBB), whereas I tested a laboratory grade force plate. Regardless of the slight differences in the output comparisons, my study is in general agreement with the two previous studies in providing support for WBBs being used as instruments to assess balance control in clinical research settings.

There are a number of limitations associated with this study. While I provided a comprehensive battery of tests of the technical specifications, there are still some characteristics that were not feasible to test. First, I did not assess the effect of the WBB duration of use (e.g. relatively new vs. 1 year of regular use) on output characteristics. A study assessing how the WBB performs after an extended period of time compared to an 'off the shelf' WBB would be beneficial for researchers conducting long term, repeated testing projects. Second, the 10 mm suitability criteria set for COP accuracy may introduce error in certain scenarios. For example, while Study 1 has shown that the WBB is suitable for measures of COP during quiet stance, a 10 mm COP error may introduce unacceptable errors for applications such as calculation of joint moments using inverse dynamics. Third, as there were a limited number of studies in the literature comparing older adult fallers to non-fallers for the specific balance variables that I tested, it was difficult to be completely confident that the quiet stance differences I found between the two devices were not clinically significant. Fourth, the study participants were university aged adults. It may be worthwhile to repeat this study on older adults if researchers want to specifically validate WBB use as an instrument for measuring balance in older adult populations. Finally, the study only assessed WBB accuracy for the relatively static activity of quiet stance. Although the second study of this thesis analyzes a sit-to-stand task, it was not assessed during the in vivo portion of the WBB validation. However, sit-to-stand data has been collected by another member of the laboratory for WBB validation purposes, and is intended to be analyzed in follow up studies. Further testing is suggested to determine WBB COP accuracy during the performance of more dynamic activities.

In conclusion, my novel characterization study tested technical specifications of the WBB that were previously unreported. Because of the increasing number of researchers using the WBB for balance testing, the establishment of the suitability (or unsuitability) of the WBB use as an alternative for the industrial grade force plate is very important. Due to its high portability and low cost, my support for the WBB as a research tool will be especially beneficial for clinicians and researchers who are unable to test metrics of balance control in a laboratory setting. The WBB showed minimal drift, low hysteresis, and high linearity, uniformity, mass accuracy and centre of pressure accuracy. The WBB measures were not affected by battery life, allowing us to be more confident with the reliability of the WBB. Similarly, the linearity, hysteresis and mass accuracy tests show that the WBB will produce accurate mass measurements unaffected by the load placed on the WBB. Through direct comparison of balance variables from a force plate to those from a WBB during a quiet stance task, I have shown that two devices produce similar output during this relatively static task. Although the WBB underestimate three of eight measures, analyses demonstrated a limited clinical significance of the differences.

Accordingly, this study supports the use of the WBB as a surrogate for a force plate when measuring balance variables during relatively static tasks such as quiet stance.

Chapter 3
Clinical measures of balance variables of older adults

3.1 Background

In older adults, falls are the leading cause of injury, being responsible for 57% of injury-related deaths in female seniors (Raina et al., 1997) and up to 85% of injury hospitalizations for persons over 65 years of age (CIHR, 2007). They amount to an approximate cost of $2.8 billion to the Canadian economy annually (SMARTRISK, 2009). It is clear that falls are an important public health concern that needs to be addressed. Please refer to Section 1.1.1 for more information on fall-related injury incidences.

One potential preventative measure for fall-related injuries is to implement compliant flooring systems in settings at high risk for falls (e.g. residential care facilities, hospitals). Safety flooring systems are a type of compliant flooring that has been designed to remain rigid under low forces, such as walking, but to collapse and absorb forces during impact. For safety floors to be an effective intervention, they must attenuate impact loads during falls, while having minimal effects on balance, mobility, and fall risk.

Past epidemiologic studies have shown that falling onto soft surfaces reduces the risk of hip fractures (Nevitt and Cummings, 1993; Healey, 1994; Simpson et al., 2004). Laboratory simulation studies have found force attenuation of up to 7% for wooden floors, 15% for carpets, and 24% for carpets with under padding (Maki and Fernie, 1990; Gardner et al., 1998; Simpson et al., 2004). Laing and Robinovitch (2009) found that safety flooring may decrease hip forces during falls by up to 50% in older women. A similar decrease in attenuation of impact forces (> 50%) occurs during simulated head impacts on safety flooring compared to commercial carpet (Wright, 2011). Safety floors have demonstrated biomechanical effectiveness during falls from an impact attenuation perspective.

To be clinically effective, safety floors should have minimal influence on balance, mobility and fall risk. It is generally recognized that extremely compliant surfaces may impair balance and mobility through modification of sensory inputs as well as by affecting the mechanical effectiveness. Modified sensory input on compliant surfaces is caused by a decreased quality of information from ankle proprioceptors and pressure receptors on the plantar foot surface, leading to a delayed or altered balance control response (Ring et al., 1989; Lord and Menz, 2000). One mechanical alteration is a reduction in toe clearance during walking due to vertical height absorption, which is also associated with an increase in energy expenditure during walking (Betker et al., 2005). Interestingly though, both carpets, and safety floors appear to have minimal influence on balance and mobility (Redfern et al., 1997; Dickinson et al., 2002). Safety floors minimally affect balance measures in healthy older adults during quiet stance, get up and go tests and external perturbations (Laing and Robinovitch, 2009; Wright, 2011). They have also associated with 3-10 times the energy absorption to deflection ratios compared to commercial carpet during simulated footfall testing (Glinka et al., 2012).

The balance assessment tests chosen for this thesis are based on the recommendations of Chiu et al. (2003). To cover the scope of fall risk in real life situations, they determined the need for the testing of balance maintenance, postural adjustment to voluntary movements, and the ability to respond to external perturbations. This novel study examined the static sway aspect and adjustment to voluntary movements because they better target the realistic functional capabilities inclusive of the entire retirement home population whereas previous flooring balance studies with older adults have only included higher functionality groups from the community or retirement home who have been able to maintain their balance in response to an external perturbation. Similarly, these past balance and floor studies did not include an evaluation of the sit-to-stand task, which is a functional task for all populations that has never been studied on safety flooring.

3.2 Hypotheses

The purpose of this study was to investigate responses during quiet stance and sit-to-stand tasks in retirement home dwellers. Both tasks are common activities of daily living, and certain performance characteristics in these tasks are associated with fall risk (Tinetti et al., 1988). The specific goal of this study was to test the hypotheses that:

1. Variables that relate to *displacement* of centre of pressure (COP) will not be significantly different across flooring conditions for both the quiet stance and sit-to-stand tasks.
2. Variables that relate to *rate* of COP displacement will not be significantly different across flooring conditions for both the quiet stance and sit-to-stand tasks.

3.3 Methods

3.3.1 Participants

Twelve healthy older adults (6 males, 6 female) residing at The Village of Winston Park (VWP) care facility participated in this study, with a mean age of 86.9 (SD = 4.6; range: 78 – 95) years, mean height of 167.6 (SD = 12.1; range: 151 – 189) cm, and mean body mass of 71.5 (SD = 9.7); range: 58 – 90) kg. Additional details on medical and fall history, and participant characteristics can be seen in Table 3-1 and Table 3-2. These characteristics provide some general normative data to allow comparison with other study populations. Future research may incorporate these data into interpretations of the main outcome variables measured in this study. These participants were recruited utilizing a research paradigm which was inclusive of many interested retirement home dwellers.

Specifically, I excluded interested participants only if they were unable to communicate in English, unable to stand continuously for 60 seconds without taking a step, or if they could not perform three consecutive sit-to-stands without using armrests.

Recruitment was coordinated in conjunction with staff from VWP as well as through the Schlegel-UW Research Institute for Aging (RIA). Following free information seminars that were held for the residents about fall risks and my study logistics, information letters were distributed to residents by RIA staff. I followed up with each interested participant to schedule a screening session to assess whether they met the inclusion criteria of the study. This study was approved by the Office of Research Ethics at the University of Waterloo. All participants provided written informed consent at the start of the first screening session.

Table 3-1: Self-reported participant characteristics related to indices of mobility

		Count	% of respondents
Ambulatory Comfort Setting*	community	0	0
	within retirement home	12	100
Ambulatory Aids	none	5	41.7
	cane (only)	1	8.3
	rollator (only)	1	8.3
	cane or rollator	5	41.7
	walker	0	0
	wheelchair	0	0
Falls Within Past Year	0	7	58.3
	1	3	25.0
	2	2	16.7
	3	0	0
	>3	0	0
Exercise (overlap between groups)	exercise class in	9	75.0
	independent exercise	1	8.3
	walking	9	75.0
	elliptical	0	0
	tai chi	7	58.3
Activities Balance and Confidence Scale score	>= 75%	8	66.7
	< 75%	4	33.3

* setting in which participant was comfortable moving independently.

Table 3-2: Self and staff reported participant characteristics related to indices of health

		Count	% of respondents
Smoker	no	9	75.0
	yes	0	0
	ex	3	25.0
Alcohol	none	2	16.7
	some	10	83.3
	heavy	0	0
Uncorrected Vision		6	50.0
Diabetes		1	8.3
Major Stroke Past 2 Years		1	8.3
Complaints of Arthritic Symptoms		1	8.3
Medications	Hypnotics / Anxiolytics	2	16.7
	Anti-depressants	2	16.7
	Anti-psychotics	0	0
	Anti-hypertensives	4	33.3
	Sedatives	0	0
	Anti Arrythmics	1	8.3
	Anti-inflammatories	10	83.3
Blood Pressure while supine (mmHg)	High >=140/90	1	8.3
	Normal = (100-140)/60-90	11	91.7
Blood Pressure standing 0 min after lying supine (mmHg)	High >=140/90	1	8.3
	Normal = (100-140)/60-90	11	91.7
Blood Pressure standing 3 min after lying supine (mmHg)	High >=140/90	0	0
	Normal = (100-140)/60-90	12	100.0
Heart Rate while supine (beats per min)	50-59	2	16.7
	60-69	3	25.0
	70-79	5	41.7
	80-89	2	16.7
MOCA test score	Normal >= 26/30	5	41.7
	Below normal < 26/30	7	58.3

3.3.2 Flooring Conditions

Five flooring conditions were tested as part of this study (Figure 3-1). Two of the five floors are commonly used in retirement homes and were labelled as ‘traditional’. The first was dense resilient rolled sheeting (RRS) (2 mm thick), representing a baseline (control) condition. The second traditional floor was an institutional grade carpet (CAR) (5 mm thick). Three commercially available safety flooring systems were also tested. SmartCell (SATech, WA, USA) is a synthetic rubber floor system that is comprised of vertical cylindrical columns 14 mm in diameter and 19 mm apart overlaid with a flat continuous surface. Twenty-five mm thick SmartCell systems was tested in two separate scenarios: SmartCell covered with resilient rolled sheeting (SCR) and SmartCell covered with carpet (SCC). The fifth condition was Kradal (Acma Industries Limited, Upper Hutt, New Zealand) floor tiles, a 12 mm thick, firm foam flooring. It was tested without any overlaying surface as recommended by the manufacturer.

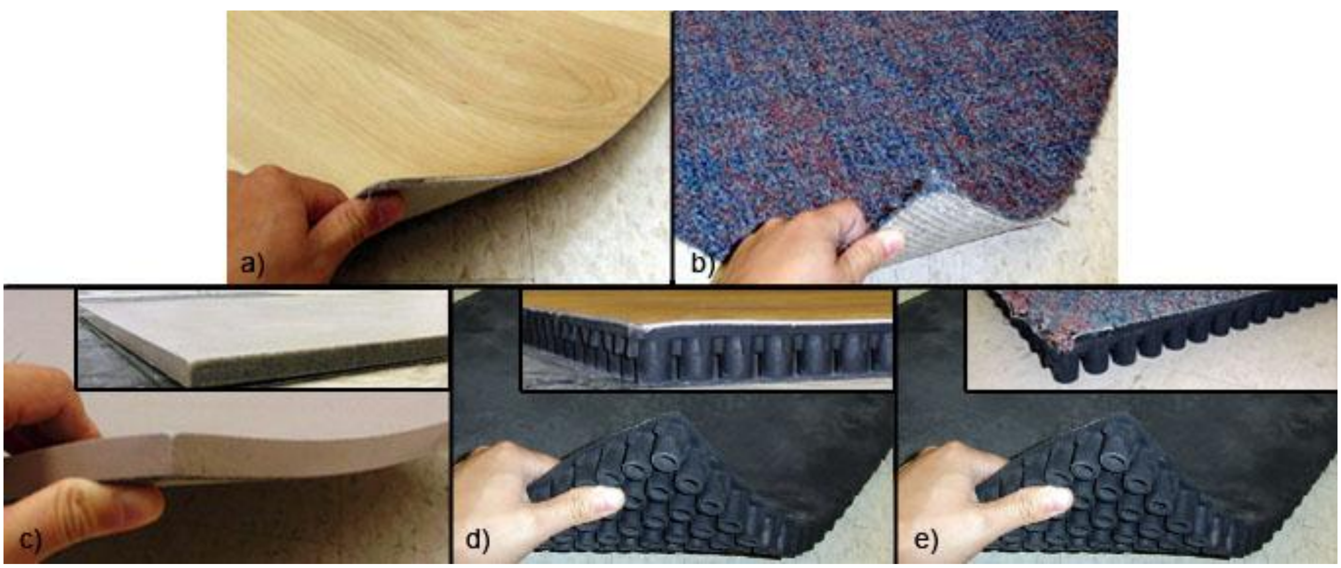


Figure 3-1: Illustration of flooring conditions. Samples included: a) resilient rolled sheeting (RRS) (a rigid control); b) carpet (CAR); c) Kradal (KRD); d) SmartCell covered by RRS; e) SmartCell covered by CAR.

3.3.3 Experimental Protocol

3.3.3.1 Setting

All data was collected at The Village of Winston Park residential care facility. I made use of a pre-existing data collection facility that was constructed as part of the Functional Fitness Assessment program. This allowed for the inclusion of participants who were unable to travel to a traditional biomechanics laboratory at the University of Waterloo.

3.3.3.2 General protocol details

The collection process consisted of three meetings with each participant. An initial meeting was used as a pre-screening session to determine if the participant was able to perform both quiet stance and sit-to-stand tasks, as well as to collect fall and medical histories, cognitive state screening (using Montreal Cognitive Assessment), and orthostatic vitals. Sessions two and three involved assessing the influence of flooring conditions on balance during quiet stance and sit-to-stand tasks, respectively. The order of flooring conditions was randomly assigned for the quiet stance task and repeated for the sit-to-stand session.

During the balance assessments, flooring conditions were placed on top of two separate plywood platforms mounted to two adjacent Nintendo Wii Balance Boards (WBB) (Nintendo, Redmond, WA, USA) (Figure 3-3). The foot and board orientation on the WBBs can be seen in Figure 3-2. Each WBB collected data from a separate foot. During the sit-to-stand task, plywood platforms were used to ensure that the top surface of the experimental floor conditions and the surrounding floor (which the chair was positioned on) were flush. Foot placement and location was standardized for both quiet stance and sit-to-stand tasks. The heels were set 17 cm apart at an angle of 16.6º between lines drawn from the heel to middle of the big toe. A stance width of 17 cm was chosen as it has been a proposed standard for quiet stance trials, while 16.6º was the average preferred stance angle in a study measuring quiet stance in older adults (McIlroy, 1997). During the sit-to-stand trials, the heels were set 10 cm in front of the seat pan edge. The boundaries of the BOS relative to the WBB coordinate system were collected for each subject (Figure 3-4). Specifically the COP locations of the anterior tip of the big toe, lateral edge of the 5th metatarsal, and the medial edge of the 1st metatarsal were recorded. The protocol setup is shown in Figure 3-3.

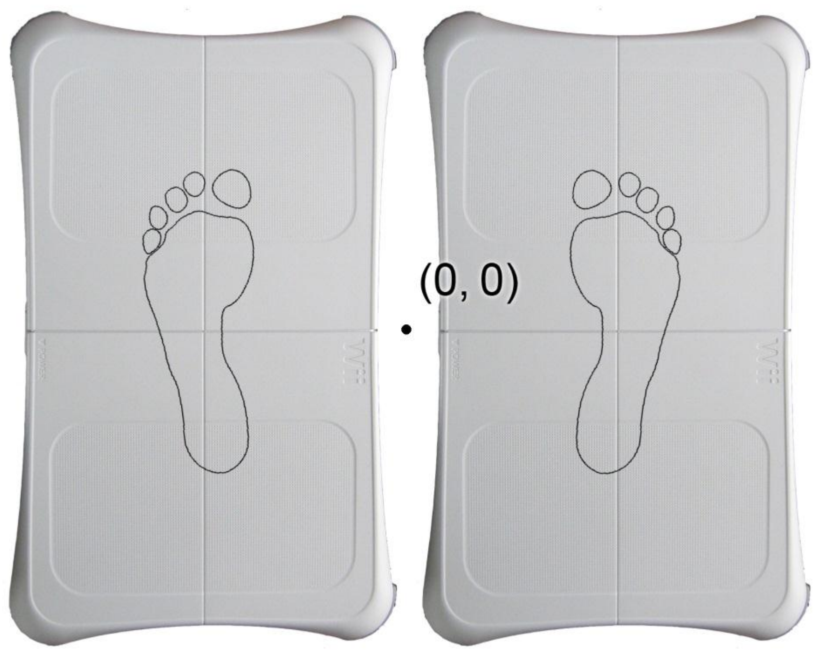


Figure 3-2: Orientation of feet on two Nintendo Wii Balance Boards during the quiet stance and sit-to-stand trials.

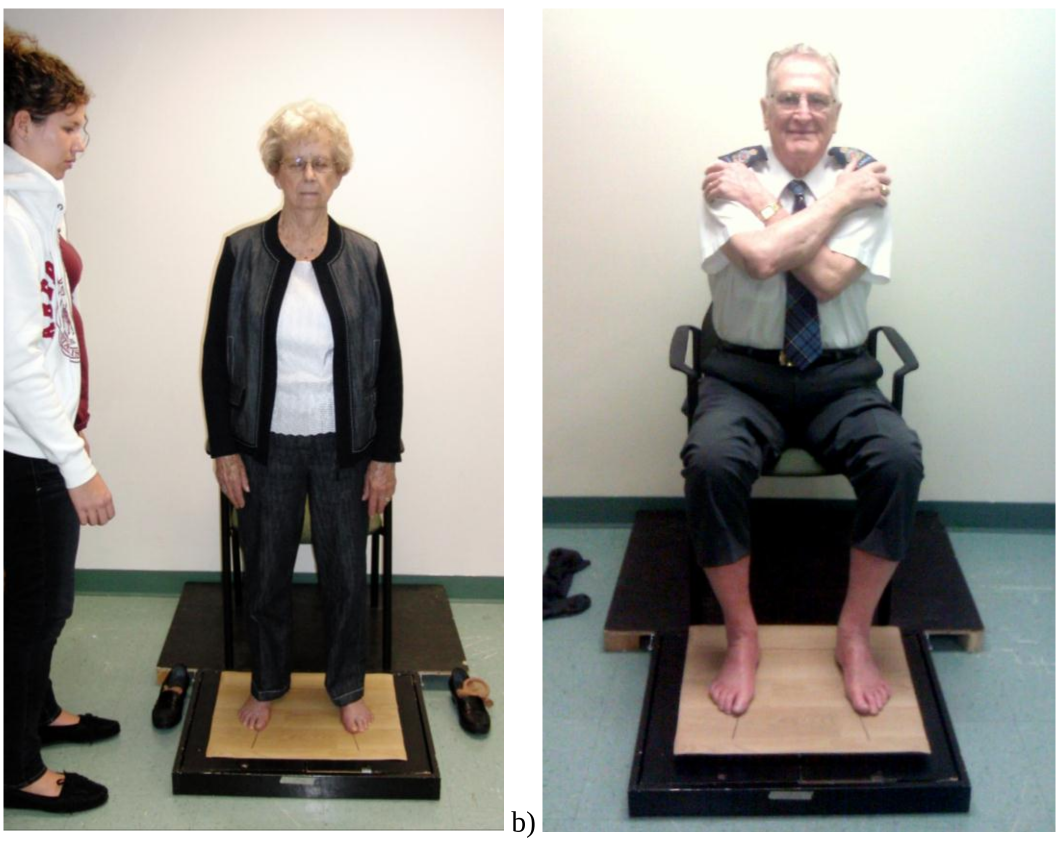
3.3.3.3 Quiet Standing

The quiet standing (QS) task was measured in the first of two testing sessions. Each trial was 30 seconds long where the participant stood barefoot on a flooring sample. For all trials, participants were instructed to initially gaze straight ahead and then close their eyes for the entire trial with their arms at their side without speaking or changing their base of support unless necessary to maintain their balance

(Baloh et al., 1994; Stel et al., 2003; Mackey and Robinovitch, 2005). There was an investigator standing beside the participant for the duration of each trial in case the participant needed assistance regaining their balance. The participant was given one practice trial with their eyes open and one 15 second practice trial with their eyes closed. Three successive trials were performed per randomized flooring condition for a total of 15 trials. A minimum 1-minute sitting rest period was provided to the participant between floor conditions, or at the request of the participant, to minimize the influence of fatigue.

3.3.3.4 Sit-to-stand

Participants performed a sit-to-stand (STS) task scheduled on a separate day after the quiet standing session. Separate QS and STS sessions were used to avoid possible interacting fatigue effects from the two tasks. Participants sat on a slightly padded chair (Bohannon et al., 1994; Thapa et al., 1994) with a seat pan height of 44 cm (Bohannon et al., 1994; Pai and Lee, 1994; Yamada and Demura, 2004; Abe et al., 2010). The front of the seat pan was lined up with a third of the length of the participants' femur from the proximal end across trials (Papa and Cappozzo, 1999; Papa and Cappozzo, 2000). The bottom of the chair legs were kept flush with the flooring condition through the use of wooden platforms (Figure 3-3). In the trial start position, participants folded their arms across their chest (Guralnik et al., 1994; Pai and Lee, 1994; Schenkman et al., 1996; Tully et al., 2005) with their vision directed straight ahead (Mackey and Robinovitch, 2005). Foot placement was set in the standard location described in Section 3.3.3.2 with ankles are dorsiflexed 10 degrees (Cheng et al., 1998; Pavol et al., 2002). It should be noted that foot position has not been found to affect total COP excursion or stabilization time (Akram and McIlroy, 2011) and was standardized so that COP starting positions would be consistent across trials. The participants were instructed to stand at their own pace without moving their feet and without the use of external aids or arm rests once given the "ready" cue. After successfully standing, participants returned their arms and hands to the side and maintained their balance using feet-in place responses only (i.e. without taking a step) (Nevitt et al., 1989; Pai and Lee, 1994) for the remainder of the 10 second trial. During initial pilot tests, participants were instructed to hold their arms crossed through the entire trial so that the COP measurements would not be affected by arm swing but this proved to be too difficult for a number of participants. The randomized flooring order from the quiet stance session was used in the sit-to-stand session. One sit-to-stand was recorded per floor condition before recording the second trial per floor, for a total of 10 trials. A minimum of 2-minutes of rest was provided after each trial, or at the request of the participant.



a) b)

Figure 3-3: Set up of testing sessions for resilient rolled sheet condition during a) quiet stance and b) sit-to-stand tasks. The two Nintendo Wii Balance Boards are covered by a wooden platform, upon which the experimental floor samples are mounted (foreground).

3.3.4 Data Collection

Vertical force data from two Nintendo Wii Balance Boards were collected at 100 Hz. A custom software program (Simon Jones, University of Toronto) was used to extract data from the boards using a Bluetooth connector. The output from this program was given as the mass (kg) from each of the vertical sensors (two boards, eight sensors total). An overall COP location was calculated relative to a designated origin (0, 0) at the geometric centre of the two boards under the feet.

3.3.5 Data Analysis

A quiet stance trial was considered successful if the participant did not take a step, grab the spotter for support, or speak throughout the trial (i.e. true feet-in-place support). Successful sit to stand trials were defined as a completion of the task upon the first try without moving their feet or making contact with the spotter. Data analysis of COP variables were performed using custom software written in Matlab (Version R2010b, Mathworks, Natick, MA, USA). Force data were filtered using a 4^{th} order dual pass Butterworth filter with a 6 Hz cut-off frequency, as determined through prior residual analyses of data from Chapter 2. COP trajectories were calculated in the anterior-posterior (AP) direction as well as the medial-lateral (ML) direction using the vertical force data from the WBB using Eqs. 1-2 & 1-3. The boundaries of the base of support (BOS) were measured and defined during testing at the following locations: head of first distal phalange, head of first metatarsal, head of fifth metatarsal, and calcaneus. The four outcome variables in the quiet stance condition calculated in the second study of Chapter 2 were calculated for Chapter 3. Three COP variables were collected in the sit-to-stand condition and are outlined in Section 3.3.5.2.

3.3.5.1 Quiet stance

Four common COP variables in each of the AP and ML planes were assessed for the quiet stance eyes closed task. Range of COP displacement (QS_{range}) (Eq. 2-4), root mean square (QS_{RMS}) (Eq. 2-5), mean velocity (QS_{vel}), and mean sway frequency in Hz (QS_{freq}) was calculated over the entire trial. The outcome variables are summarized in Table 3-3. Please refer to Section 2.2.4.3 for more detailed descriptions on the outcome variable calculations.

3.3.5.2 Sit-to-stand

There are three phases associated with the sit-to-stand movement (Figure 3-5). The first is the preparation phase, which begins at the start of a sharp rise in COP in the AP direction and ends when the AP COP starts to decline (initiation of movement until chair is unloaded). The movement phase then begins, and ends after the hip joint reaches full extension (Akram and McIlroy, 2011). The stabilization phase starts from the time of full hip extension to the time when COP sway falls within 95% CI of quiet standing sway. Because kinematic measures were not recorded in this study, the exact timing of the movement and stabilization phases could not be calculated. However, outcome variables spanning the two phases were calculated. I standardized the location of the participants' toes across trials and used this as the frame of reference for my COP minimum margin of safety (MMOS) variables (Wright, 2011).

Four COP variables were assessed in the AP direction for the sit-to-stand task (Figure 3-4 and Figure 3-5). The range of COP displacement during the preparation phase (STS_{range}) was calculated and defined as the COP at the start of preparation phase to the furthest point of COP displacement in the AP direction during the preparation phase (Figure 3-5). The average velocity (STS_{vel}) during the preparation phase was calculated by dividing the total displacement during the preparation phase by the time it took to complete the phase: $STS_{range} / \Delta t$. The minimum margin of safety (STS_{MMOS}) was calculated as the distance from the big toe relative to the BOS. This was defined as the difference between the maximum anterior excursion during each trial and the furthest anterior BOS location (Figure 3-5). The total distance (STS_{path}) travelled in the first 3 seconds immediately after the preparation phase following seat-off (Figure 3-5). Traditionally, the total distance travelled would be calculated throughout the exact movement phase. We could not determine the movement phase time without the use of kinematics and an approximate phase time of 3 seconds was therefore chosen. An additional trial success rate on each floor was recorded. Successful sit-to-stand trials are defined in Section 3.3.5. The outcome variables are summarized in Table 3-3.

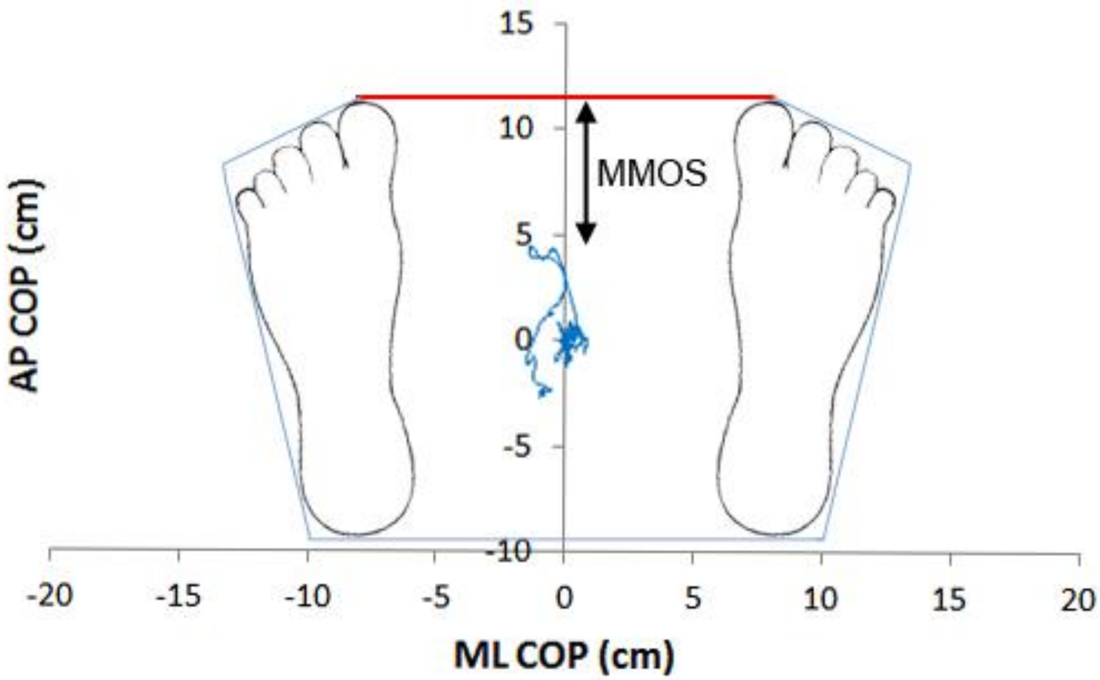


Figure 3-4: COP tracing (in dark blue) during a sit-to-stand task in anterior-posterior and medial-lateral directions; the solid red line indicates the most anterior portion of the BOS bounded by the toes; the solid light blue line outlines the base of support in conjunction with the red line; MMOS indicates the minimal margin of safety.

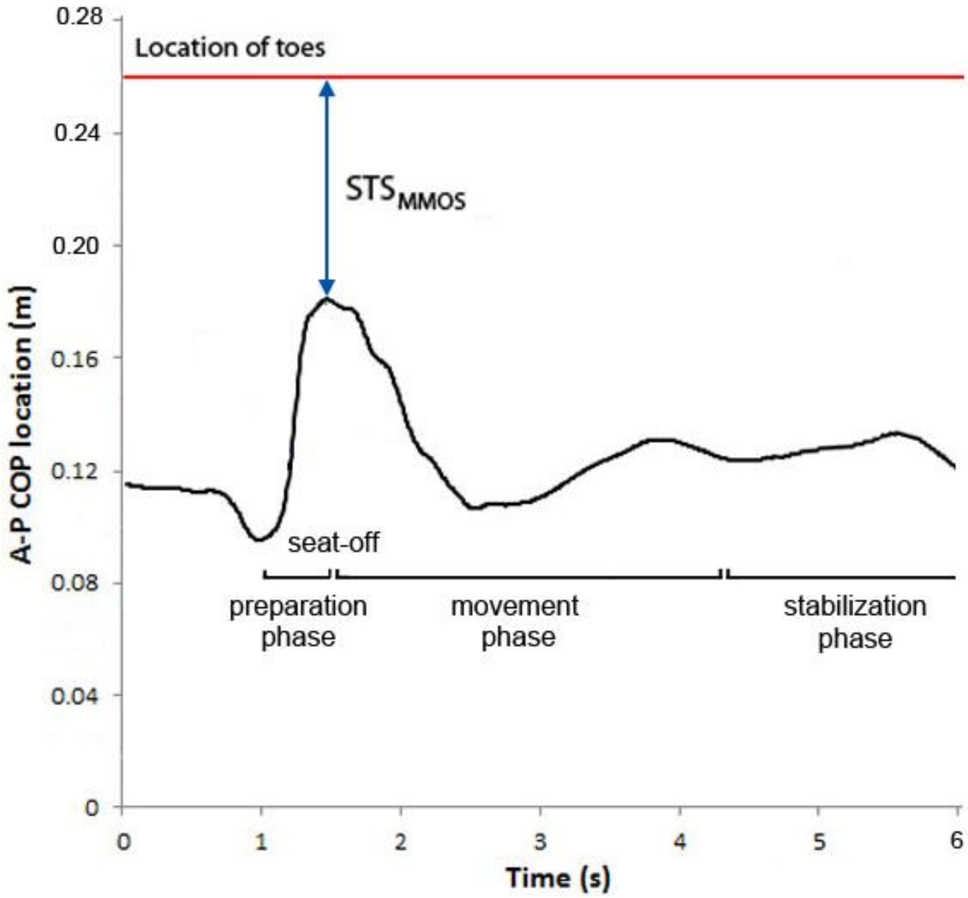


Figure 3-5: Sample STS tracing on rigid flooring showing the three phases. The preparation phase begins at the initiation of movement and ends once the buttocks leave the seat. The movement phase begins at seat off and ends once the stand has been completed (full hip extension). The stabilization phase is the time it takes to fall within the 95% CI of quiet stance sway. The full stabilization phase could not be shown within the 10 second trial.

Table 3-3: Outcome variables from WBB and force plate during static and dynamic tasks

Task	Outcome Variables
Quiet stance (4) **AP and ML**	QS_{range}: Maximum range of displacement QS_{RMS}: Root mean square over 30 second trial QS_{vel}: Mean velocity over the entire trial QS_{freq}: Mean sway frequency over the entire trial
Sit-to-stand (5) **AP only**	STS_{range}: COP displacement from the start to end of preparation phase (sit to stand transition) STS_{vel}: Mean velocity during preparation phase STS_{MMOS}: Minimum margin of safety between greatest AP displacement and the location of toes STS_{path}: Distance travelled by the COP in first 3 seconds of the movement phase following seat off Success rate: Percentage of successful trials on each floor. I.e. Participant was able to stand on first try and/or recover balance

3.3.6 Statistics

For the QS and STS tasks, separate one-way repeated measures analysis of variance (ANOVA) were used to test for the effect of floor condition (repeated: 5 levels) on each dependent variable listed in Table 3-3. Where the assumption of sphericity was violated, a Hyunh-Feldt correction was applied. When necessary, post-hoc analyses were conducted using pairwise comparisons, comparing each floor to the resilient rolled sheeting control condition. A chi-squared test was performed on the success rate variable in the STS task. All statistical analyses were conducted with a significance level of 0.05 using statistical analysis software (SPSS Version 18.0, SPSS Inc., Chicago, IL, USA).

3.4 Results

3.4.1 Quiet stance eyes closed

Table 3-4 summarizes the results of balance variable calculations across floor conditions for the quiet stance task. A sample trial on the control flooring is shown in Figure 3-6. Participants were able to maintain their balance during the quiet stance task for 100% of trials in all floor conditions. The four balance variables were not significantly different across floor conditions in either the AP or ML direction ($p > 0.05$; Table 3-4). For the control floor condition (RRS), the mean (SD) AP QS_{range} was 2.94 (1.28) cm, the AP QS_{RMS} location was 0.55 (0.28) cm, the AP QS_{vel} was 1.63 (1.07) cm/s and AP QS_{freq} was 0.66 (0.23) Hz. The mean (SD) ML QS_{range} was 1.77 (0.77) cm, the ML QS_{RMS} location was 0.36 (0.16) cm, the ML QS_{vel} was 0.73 (0.41) cm/s, and the AP QS_{freq} was 0.49 (0.27) Hz.

ANOVA results showed that floor condition was not significantly associated with WBB output across AP range, $F(4,44) = 0.749$, $p=0.564$, AP RMS, $F(4,44) = 0.311$, $p=0.869$, AP velocity, $F(4,44) = 0.709$, $p=0.590$, AP frequency, $F(4,44) = 0.1.483$, $p=0.224$, ML range, $F(4,44) = 0.524$, $p=0.718$, ML RMS, $F(4,44) = 1.835$, $p=0.139$, ML velocity, $F(4,44) = 0.993$, $p=0.420$, and ML frequency, $F(4,44) = 0.477$, $p=0.752$. The ANOVA results are summarized in Table 3-4.

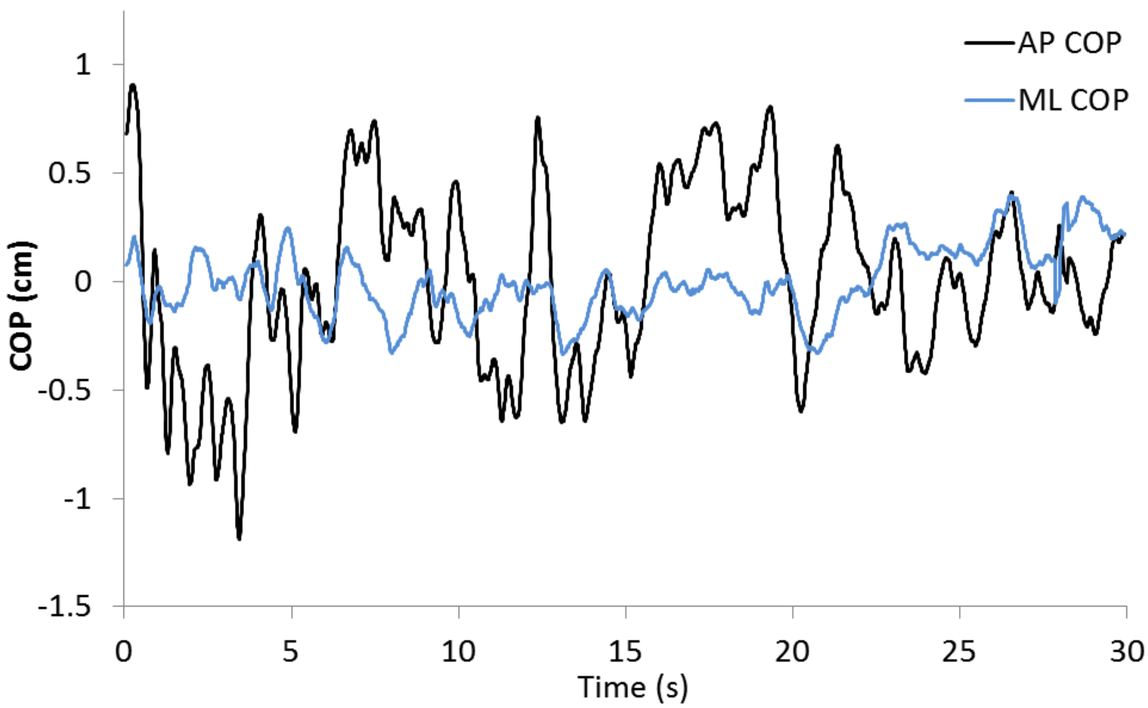


Figure 3-6: Sample 30 sec QSEC centre of pressure tracing on RRS (control) flooring

Table 3-4: Mean (SD) values and ANOVA results for balance variables during quiet stance eyes closed task across resilient rolled sheeting (RRS), carpet (CAR), Kradal (KRD), SmartCell with RRS (SCR), and SmartCell with carpet (SCC) floor conditions

	Parameter	Floor					ANOVA	
		RRS	**CAR**	**KRD**	**SCR**	**SCC**	**F-ratio**	**p-value**
AP	QS_{range} (cm)	2.94 (1.28)	2.95 (1.28)	3.15 (1.25)	3.04 (1.48)	2.80 (1.53)	0.749	0.564
	QS_{RMS} (cm)	0.55 (0.28)	0.55 (0.23)	0.57 (0.21)	0.56 (0.23)	0.52 (0.28)	0.311	0.869
	QS_{vel} (cm/s)	1.63 (1.07)	1.66 (0.90)	1.75 (0.89)	1.53 (0.84)	1.52 (0.94)	0.709	0.590
	QS_{freq} (Hz)	0.66 (0.23)	0.67 (0.20)	0.67 (0.23)	0.61 (0.21)	0.65 (0.20)	1.483	0.224
ML	QS_{range} (cm)	1.77 (0.77)	1.74 (0.68)	1.66 (0.63)	1.64 (0.69)	1.62 (0.71)	0.524	0.718
	QS_{RMS} (cm)	0.36 (0.16)	0.33 (0.11)	0.32 (0.12)	0.31 (0.11)	0.31 (0.11)	1.835	0.139
	QS_{vel} (cm/s)	0.73 (0.41)	0.74 (0.33)	0.73 (0.33)	0.68 (0.39)	0.66 (0.30)	0.996	0.420
	QS_{freq} (Hz)	0.49 (0.27)	0.53 (0.21)	0.54 (0.22)	0.51 (0.23)	0.50 (0.16)	0.477	0.752

* indicates p < 0.05 for effect of floor

3.4.2 Sit to stand

Sample sit-to-stand trials across flooring conditions from one participant can be seen in Figure 3-7. Participants were able to successfully stand up on the first attempt 99.2%, 99.2%, 100%, 97.5% and 99.2% of trials on RRS, CAR, KRD, SCV, and SCC. Table 3-5 summarizes the results of balance variable calculations across floor conditions. A chi-squared test revealed that the differences in success rate on floors was not significant, x^2(df = 4, N=120) = 0.10, p > 0.05.

For the control floor condition (RRS), the mean (SD) preparation phase range (STS_{range}) was 6.09 (1.92) cm, the mean preparation phase velocity (STS_{vel}) was 24.52 (12.67) cm/s, the mean preparation phase minimum margin of safety (STS_{MMOS}) was 3.95 (1.51) cm, and the mean movement phase path length (STS_{path}) was 18.51 (5.23). The ANOVA results are summarized in Table 3-6. Flooring condition had an overall significant effect on STS_{MMOS} ($F_{4,44}$ = 3.038, p = 0.027). Further post-hoc analysis indicated that the SCR condition resulted in significantly different STS_{MMOS} results ($p = 0.001$) compared to the control floor.

ANOVA results indicate that floor condition was not significantly associated with STS_{range} ($F_{4,44} = 1.543$, $p = 0.206$). The overall effects of flooring on STS_{vel} were not significant ($F_{4,44} = 2.572$, $p = 0.051$), or movement phase path length ($F_{4,44}=2.085$, $p = 0.099$). Post-hoc tests were also performed for STS_{vel} as it closely approached my alpha value. The SCR condition was found to be significantly different from the control floor in STS_{vel} (23.9% difference, $p = 0.023$), as well as the SCC condition (25.6% difference, $p = 0.042$).

Table 3-5: Mean (SD) values for balance variables during sit-to-stand task across resilient rolled sheeting (RRS), carpet (CAR), Kradal (KRD), SmartCell with RRS (SCR), and SmartCell with carpet (SCC) floor conditions.

Parameter	**Floor**				
	RRS (control)	**CAR**	**KRD**	**SCR**	**SCC**
STS_{range} (cm)	6.09 (1.92)	6.84 (1.82)	6.17 (1.01)	6.01 (0.88)	5.68 (1.22)
STS_{vel} (cm/s)	24.52 (12.67)	21.93 (11.94)	21.63 (8.85)	18.65 (8.69)*	18.25 (9.64)*
STS_{MMOS} (cm)	3.95 (1.51)	4.26 (1.71)	4.51 (1.46)	5.12 (1.41)*	4.48 (1.30)
STS_{path} (cm)	18.51 (5.23)	20.34 (5.83)	18.22 (4.31)	18.65 (8.69)	17.63 (3.79)
Success (%)	99.2	99.2	100.0	97.5	99.2

* indicates floor is significantly different from control floor ($p < 0.05$)

Table 3-6: ANOVA table comparing effects of floor condition on balance variables during sit-to-stand task. Pairwise comparisons were conducted for variables with a significant F value.

	ANOVA		**Pairwise Comparisons**			
	Floor		RRS/CAR	RRS/KRD	RRS/SCR	RRS/SCC
Anterior-posterior	F-ratio	p-value	p-value			
STS_{range} (cm)	1.543	0.206	--	--	--	--
STS_{vel} (cm/s)	2.572	0.051	0.378	0.294	0.023*	0.042*
STS_{MMOS} (cm)	3.038	0.027	0.504	0.238	0.001*	0.110
STS_{path} (cm)	2.085	0.099	--	--	--	--

* indicates effect of floor is significantly different from control floor ($p < 0.05$)

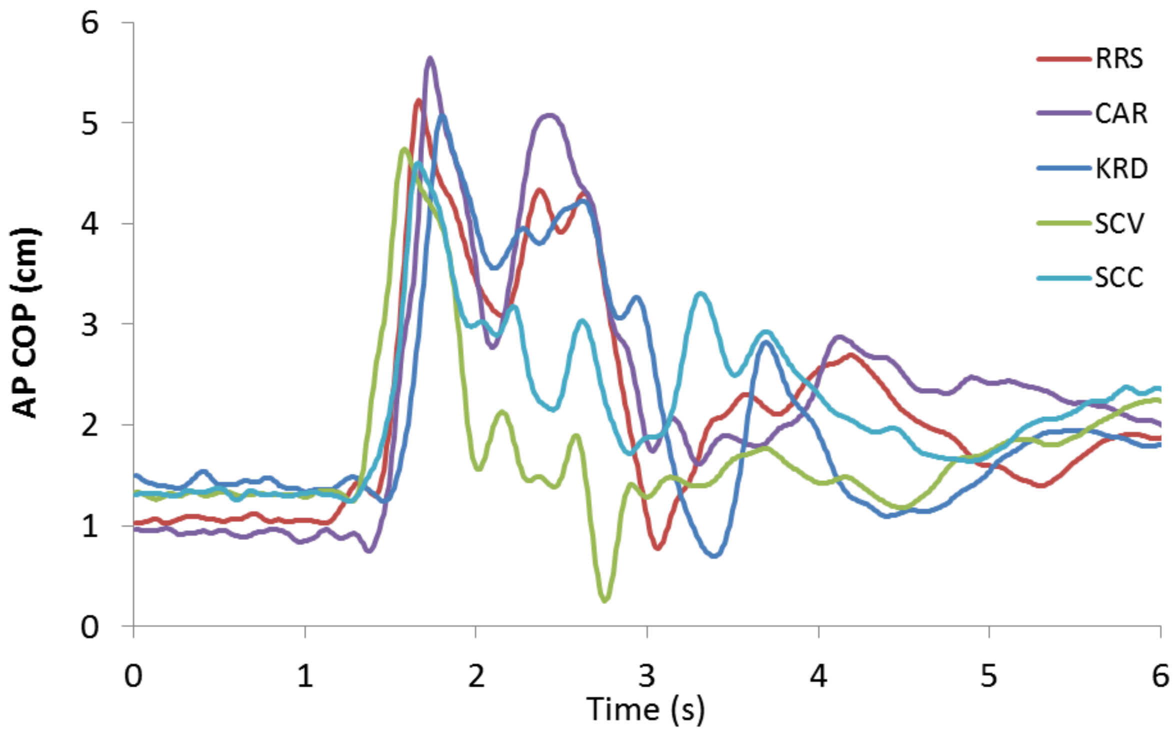


Figure 3-7: Sample sit-to-stand trials on each flooring condition by one participant

3.5 Discussion

In this thesis, the influence of flooring surface (5 conditions) on balance control of older retirement home dwelling adults was evaluated during quiet stance with eyes closed (QSEC) and sit-to-stand (STS) tasks. The results from the QSEC task indicate that the safety floors (3 conditions) have minimal effects on balance control during static feet-in-place tasks. The results from the STS task indicate that safety floors do not affect balance control characteristics during voluntary movements. Within the balance control variables calculated during these tasks, the results support the hypothesis that centre of pressure (COP) displacement variables are not affected by flooring condition. The results support the hypothesis that rate of COP displacement values are not affected by flooring condition.

As no statistically significant differences were seen between floors in the quiet stance group, it is worth considering whether the balance variables that we chose to evaluate are sufficiently sensitive to flooring compliance. First, we know that highly compliant surfaces can impair balance and mobility through factors such as decreased quality of information from both the ankle proprioceptors and pressure receptors on plantar foot surface. Past studies have found that during a floor shifting paradigm, certain balance variables are negatively affected on highly compliant floors compared to the control resilient rolled sheeting floor used in this study (Wright and Laing, 2011). If the safety floors do indeed mimic the responses of compliant floors, they certainly have the potential to affect balance control. Despite the fact that Wright and Laing tested (2011) community dwelling older women in an external perturbation task, the results justify our use of similar balance measures in our residential home dwelling group, who may not have be able to react to the external perturbation. Even a more basic quiet stance task had the potential to have had similar effects as the external perturbation in community dwelling older adults.

While the effect of safety flooring on balance control variables during a QSEC task performed by residential-dwelling older adults had never been tested prior to this study, Laing and Robinovitch (2009) evaluated the task in community dwelling older women. They obtained RMS and mean velocity values that were lower and slower (32.7% and 29.4% respectively in the AP direction, and 61.1% and 39.7% in the ML direction) than our values, which is expected in a more independent population (Table 3-7). There have been a range of reported values for quiet stance eyes open in community dwelling older adults (Table 3-7), which our values fall within or close to with the exception of AP and ML frequency. Our frequency values averaged 0.65 Hz in the AP direction, and 0.51 Hz in the ML direction across floors, higher than the literature reported range (0.43 – 0.56 in AP, 0.41 – 0.46 in ML). Interestingly, the ML frequency values from our young adults were even higher, with a range of 0.72-0.84 Hz. Mackey and Robinovitch (2005) reported an AP frequency value of 0.62 Hz for the QSEC task on a compliant surface, suggesting our participants may have even higher AP mean frequencies on compliant surfaces and therefore less balance control than the community dwelling older adults while standing on them. While the ML frequency value by Mackey and Robinovitch (2005) on compliant flooring, 0.42 (0.12) Hz, is higher than our value, both fall within the literature reported values of quiet stance on rigid floors. The rest of their reported values for QSEC are: AP mean (SD) range: 6.64 (1.94) cm, AP RMS: 1.50 (0.48) cm, AP velocity: 3.94 (1.45) cm/s, ML range: 5.48 (2.69) cm, ML RMS: 1.21 (0.55) cm, and ML velocity: 2.30 (1.28) cm/s. which, are all higher than the values I obtained. Our measures were also greater than those obtained with the younger adult participants in Chapter 2, indicating that the older adults generally had larger sway and higher velocity as expected. Maki (1994) reported values that were significantly associated with older adult fallers include having ML RMS 0.34 (0.23) cm and ML frequency of 0.41 (0.22) Hz. However, both these numbers are within the values reported for quiet standing on a rigid floor based on Table 3-7, and as a result it is difficult to say whether our group is representative of community dwelling older adult fallers based on the values alone.

Table 3-7: Literature reported balance control variables on rigid and safety flooring from a 10-second quiet stance task in community dwelling older adults (Baloh et al., 1994; Maki et al., 1994; Brauer et al., 2000; Stel et al., 2003; Freitas et al., 2005; Laing and Robinovitch, 2009; Melzer et al., 2010).

		Eyes Closed		Eyes Open
		Rigid (Control)	**SmartCell**	**Rigid**
AP	Range (cm)	--	--	1.7 – 4.0
	RMS (cm)	0.37 (0.12)	0.42 (0.13)	0.36 – 0.57
	Velocity (cm/s)	1.15 (0.36)	1.21 (0.40)	0.63 – 1.79
	Frequency (Hz)	--	--	0.43 – 0.56
ML	Range (cm)	--	--	0.88 – 4.40
	RMS (cm)	0.14 (0.05)	0.15 (0.05)	0.22 – 0.34
	Velocity (cm/s)	0.44 (0.16)	0.47 (0.13)	0.13 – 0.97
	Frequency (Hz)	--	--	0.41 – 0.46

Until recently, the sit-to-stand task had not often been well characterized from a biomechanical perspective; its performance had often been quantified based on completion quantity, time for completion, and/or part of a timed up and go task as well. As such, there are not many available resources from past studies investigating centre of pressure variables during the STS. Total sway has been shown to increase on highly compliant floors, suggesting that the path travelled during the movement phase may indicate whether the floor acts similarly to compliant floors in terms of effects on balance. In this study, the distance travelled in the first three seconds in the movement phase following seat off was not found to be significantly different across floor types, which would have been expected if the floors had enough compliance to impair balance maintenance responses. We found the velocity during the preparation phase to be significantly different on the two SmartCell conditions compared to the control condition. Since compliant floors have dampening effects, a higher horizontal velocity is often required to gain the momentum required when performing a STS. Both safety floor velocities were lower than the control by approximately 6 cm/s, which may indicate that the safety floors demonstrate some compliant properties. However, it must also be noted that if velocity is too fast prior to seat off, it is expected to be more difficult for individuals to slow their centre of mass once out of the chair. Therefore, the significantly slower velocity seen on the safety floors during the preparation phase may not have clinical significance. The minimal margin of safety was significantly larger on SmartCell with the RRS overlay compared to the control margin. This is desirable as it indicates that participants are further from their BOS. Once an individual's COP crosses the BOS, they will no longer be able to control their centre of mass using feet in place strategies which will either result in a stepping response to alter the base of support, or a fall.

While this study primarily investigated variables associated with balance control, subjective ratings on each floor were recorded after the completion of both sessions. I asked each participant what their favourite floor was by asking them the following question: "Can you please rate your favourite floor in terms of your 'balance' on each floor, and which floor had the nicest 'feel'?". Six of twelve participants had no preference in either balance or feel. An additional two of the twelve had no preference on feel but felt that their balance was most compromised on the RRS flooring, with no specification on whether it was bare RRS, or if it had the SmartCell underlay. In contrast to this, one participant reported that the RRS felt most 'secure'. Two more participants liked the carpet condition, with one participant also describing the carpet as feeling 'secure', and providing more 'traction'. Overall, there does not appear to be a difference in the subjective ratings of the flooring samples I examined in this study.

There are several limitations associated with this study. First, we chose to investigate the quiet stance task while participants kept their eyes closed because it was expected to be a more challenging task than with eyes open. However, it has relevance to worst-case scenarios in which individuals may be required to maintain balance at night without appropriate lighting. However, older adults are most likely to be walking in the dark at night time to go to the washroom after being in bed. My protocol did not involve testing my participants after they had being lying down, which may have offered additional information. Second, participants were limited to using feet-in-place strategies to perform the STS, which may contribute to large variation in COP patterns seen in participants since it may not have been natural to stand with a constrained foot position. Similarly, because the participants' feet were constrained to a position that may not have been their normal STS starting point, this may have contributed to the limited number of failed attempts at standing. One of the reasons for choosing to study feet-in-place strategies is that the majority of recurrent fallers have been found to not use a step to recover from an external alteration of COM, implying fall in clinical terms (Pavol et al., 2002). This 'faller' group is of greater interest as a target population for safety floors. As well, having feet-in-place strategies simplified the collection process and subsequent data analyses by minimizing within subject variance and increasing precision in QSEC results. Another reason for choosing feet-in-place strategies is that the STS motion is constrained more to the AP direction because participants were not permitted to step out to the side. The ML direction was not analyzed in this study because of this constraint, but should be done in follow-up work. Despite standardization of the physical starting location of STS trials, the starting underfoot COP still ranged between trials and floors within participants. The ability to sit up and maintain a 'straight' posture varied greatly between participants and trunk position was therefore self-selected, possibly leading to the difference in starting COP between trials. Participants were also allowed to drop their arms from the initial crossed arms position. Although previous STS studies have required participants to maintain crossed arms, I found that some of my older adult population was not able to comply with this requirement. As a result, my findings during the movement phase may not be directly comparable to other sit-to-stand data sets. However, the preparation phase dependent variables should not be affected by instances of arm drops. Third, the lack of kinematics collection and the duration of my STS trials (10 sec) were not sufficient to calculate the exact movement phase length and time to stabilization in the older adult population. Fourth, COP patterns tended to vary more during the movement phase than the preparation phase (Figure 3-7), and it is unclear whether this variance is a physiological phenomenon or an artifact associated with the increased shear forces which occur during the movement phase of the sit-to-stand task (and which are not measured by the WBB). Future validation of WBB COP accuracy is recommended for dynamic task activities such as the sit-to-stand task. Regardless, my overall conclusions are likely unaffected, as only one of four dependent variables were calculated from movement phase data. Finally, a longer trial length (e.g. 20 seconds) would have allowed us to calculate time to stabilization, which is a fall risk factor and has been found to be affected by balance factors such as stance width (Akram and McIlroy, 2011).

This study was novel because within the retirement home dwelling population, the effects of safety flooring on balance control have only been thoroughly investigated using a tether-release paradigm, simulating external perturbations. Floor shifting protocols have also been used to test a limited number of safety floors. Centre of pressure balance variables of quiet stance and STS tasks have previously been tested in community dwelling older adults only. My study provides more depth by increase the population range of these related studies. It is important to study the retirement home dwelling population because some of the older adults may have decreased functional capabilities resulting in reliance on assisted care. Retirement homes are also a realistic, feasible, environment to install safety floors in and are already being used as the pilot sites for prospective studies investigating the fall rates on safety floors. I studied static sway and adjustment to voluntary movements, which have been associated with fall risk and are very basic functional requirement that may be more challenging for retirement home dwelling individuals. In nursing homes, most falls occur on level surface and are associated with low to moderate changes in position or posture such as getting out of bed, transferring to bed, chair or toilet. This demonstrates the importance of studying the STS as a related factor in falls. Previous to this study the effect of safety flooring on QS and STS was unknown. However, safety floors must not be the only fall related injury intervention. Group exercise classes targeting balance, strength and gait, tai chi classes, modification of environment hazards, and fall prevention education have also been shown to reduce the rate of falls. Similarly, the supply of walking aids, hip protectors, and removal of medications associated with light-headedness and therefore relating to falls has been shown to potential decrease fall rate. However, there are issues with compliance of users of the aforementioned strategies.

This study has provided additional findings to past studies that have demonstrated the effectiveness of safety floors in reducing fall related injuries including hip fractures (Laing and Robinovitch, 2009). We have shown that the balance control on safety floors is not significantly affected compared to our rigid control condition for a quiet stance eyes closed task. The hypotheses of this study have been strongly supported, indicating that COP displacement, as well as COP rate of displacement balance control variables are not affected by safety floors compared to traditional floors. Although the safety floors showed some effect on the rate dependent COP variables, the values we obtained were still within the reported values of standing balance of older adults on rigid floors. The results provide further support for the use of safety floors, as well as for the continuation of pilot testing of the flooring in retirement home environments.

Chapter 4
Thesis Conclusion

As future population estimates show a shift towards the older adult population, an associated increase in fall related injuries is expected. This poses an increased financial burden on the Canadian health care system as well as health care providers. Retirement homes, hospitals and residential care facilities will be target areas for fall related environmental modifications as a high percentage of older adults will be living in those settings. The main focus of this study was to evaluate the potential for safety floor installation in care facilities with the intention of reducing fall related injuries. While these safety floors have successfully incorporated compliant properties to protect against injuries resulting from fall impacts, one challenge has been to include a design that allows users to maintain balance control during daily activities. A secondary goal of this study was to evaluate the potential use of a Nintendo Wii Balance Board (WBB) in place of a laboratory grade force plate in balance research. If suitable, the WBB provides a cheaper, more portable alternative for testing in non-laboratory settings such as retirement homes and clinics.

Safety floors have been shown to effectively reduce impact forces during simulated hip, head and footfall falls by at least 50% compared to control resilient rolled sheeting or commercial carpet conditions (Laing and Robinovitch, 2009; Wright and Laing, 2011; Glinka et al., 2012). Because of these results, there has been increasing interest for the use of these floors in settings of frequent falls and fall-related injuries, such as residential care facilities. Certain safety floors have been shown to minimally affect fall risk compared to traditional floors for community-dwelling older adults in quiet stance sway, Timed Up and Go tests, and backwards floor shifting tests (Laing and Robinovitch, 2009; Wright and Laing, 2011). Similarly, safety floors have not been shown to affect balance control significantly more than the control traditional floors in retirement home-dwelling older adults during a tether release external perturbation task (Wright and Laing, 2011). Chapter 3 of this thesis provides a further comparison of these floors during quiet stance and sit-to-stand tasks by retirement home-dwelling adults that strongly indicates that safety floors do not negatively affect balance control responses to a greater degree than traditional floors. Safety floors appear to be a promising intervention for reducing fall related injuries, without affecting aspects of balance control.

This thesis also provides insights into the validity of using Nintendo Wii Balance Boards as an alternative to laboratory grade force plates for balance assessment purposes. Laboratory grade force plates have traditionally been used for measuring balance control variables in biomechanics research. Recently, the WBB has been used as an alternative to the force plate when assessing balance due to its high portability (3.5 kg) and low cost (< 1% of a force plate). Previous studies comparing WBB to force plate technical specifications have been basic and only included linearity and COP accuracy. In these studies, the WBB showed excellent linearity (coefficient of determination = 0.99) and high COP accuracy (< 3% error) (Clark et al., 2010; Pagnacco et al., 2011). Chapter 2 of this thesis validates the previous findings, and also showed that the WBB has low drift, low hysteresis, high mass accuracy, and high uniformity. Past studies have also evaluated the performance of the WBB during quiet stance sway. However, one study used a non-laboratory grade force plate for comparison (Pagnacco et al., 2011), while the other compared separate trials from a WBB and a force plate (Clark et al., 2010). This thesis compared simultaneous quiet stance sway trials from a WBB and a force plate to obtain a more accurate comparison. The results showed that the differences between the WBB and force plate were not clinically significant and that the WBB would be an appropriate substitution for the force plate during quiet stance balance control measures. However, the COP measures were quite variable during the movement phase of the sit-to-stand as seen in Chapter 3 and may not be accurately recorded by the

WBB during this dynamic phase. More work should be done to validate WBB use to measure balance control during a sit-to-stand task.

In conclusion, this thesis provides support that Nintendo Wii Balance boards appear to be a reasonable alternative to force plates during quiet stance tasks. This supports the use of WBBs from the perspective of clinical balance training, in additional to research purposes. This thesis also demonstrates that safety floors do not affect balance control responses in some activities of daily living, complimenting the previously studies impact attenuating properties of the floors. Additional research may want to assess WBB performance during dynamic tasks involving shear forces. The results from this study supports prospective clinical investigations of safety flooring in retirement and nursing home settings to evaluate their real life effects on fall-related injuries.

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