# Novel Compliant Flooring Systems from Head to Toes: Influences on Early Compensatory Balance Reactions in Retirement-Home Dwelling Adults and on Impact Dynamics during Simulated Head Impacts

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. I understand that my thesis may be made electronically available to the public.

# ABSTRACT

The overall goal of my research was to advance our understanding of the potential for novel compliant flooring systems to reduce the risk for fall-related injuries in older adults, including fall-related traumatic brain injury (TBI). This entailed an assessment of how these floors affect the competing demands of fall-related TBI – impact severity attenuation in concert with minimal concomitant impairments to balance control and postural stability. Two studies are included as part of this thesis. The first study used a mechanical drop tower to assess the effects of four traditional flooring systems and six novel compliant flooring conditions on the impact dynamics of a surrogate headform during the impact phase of simulated 'worst-case' head impacts. The second study entailed an assessment of the effect of two traditional and three novel compliant floors on the initial phase of the compensatory balance reactions of older adult men and women living in a residential-care facility environment following an externally induced perturbation using a tether-release paradigm. Overall, this thesis demonstrates that novel compliant floors substantially attenuate the forces and accelerations applied to the head during simulated worst-case impacts when compared to traditional flooring surfaces such as vinyl and carpet with underpadding. These benefits are achieved without compromising indices of balance control, supported by the finding that parameters characterizing early compensatory balance reactions were unaffected by the novel compliant floors tested. This work supports the introduction of pilot installations of novel compliant flooring systems into environments with high incidences of falls to test their effectiveness at reducing fall-related injuries in clinical settings.

**Keywords:** impact biomechanics; injury prevention; fall-related injuries; concussions; traumatic brain injury; balance control; compliant floors; accidental falls; ageing;

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# CHAPTER 1 GENERAL INTRODUCTION & LITERATURE REVIEW

## 1.1 Fall-related Injuries in Older Adults

### 1.1.1 Scope of the Problem

Fall-related injuries in adults over the age of 65 are a major public health issue in Canada, representing the number one cause of injury-related hospitalizations and deaths for this age group (Health Canada, 2002; SMARTRISK, 2009). The direct annual costs associated with fall-related injuries in this population are in excess of $2 billion in Canada alone (SMARTRISK, 2009). It is estimated that 1 in 3 community-dwelling older adults will experience at least one fall per year, with 50% of this cohort suffering multiple falls (SMARTRISK, 2009). This rate is over three times higher for older adults residing in environments such as hospitals and nursing-care facilities, equating to approximately 1.4 falls per person per year (Cameron et al., 2010). The risk of fall-related injuries is also higher in these environments, with 22% of falls resulting in serious injuries in community-dwelling older adults (Speechley and Tinetti, 1991) versus one-third of all falls in residential-care settings (Nurmi and Luthje, 2002).

The risk for sustaining a fall-related injury increases dramatically with age. Health Canada reports that seniors over the age of 85 are 70% more likely to suffer an activity-limiting injury than persons aged 65-74 (Health Canada, 2002). The proportion of the Canadian population over the age of 65 is anticipated to climb to nearly 25% by the year 2041 (Health Canada, 2002), with the fastest growing demographic represented by adults over 80 years of age (Seidel et al., 2009) (Figure 1-1). Accordingly, the rates of fall-related injuries are expected to increase dramatically over the coming decades (Ferrell and Tanev, 2002). Serious injuries, such as hip fractures and traumatic brain injuries are associated with increased mortality and morbidity (Meyer et al., 2000), decreased mobility, physical activity, and functional independence (Wolinsky and Fitzgerald, 1994), as well as onset or progression of neuropsychologic disabilities (Adekoya et al., 2002). The most effective strategy to limit the negative physical, emotional, and economic consequences associated with these events is through primary prevention of falls and fall-related injuries in older adults.

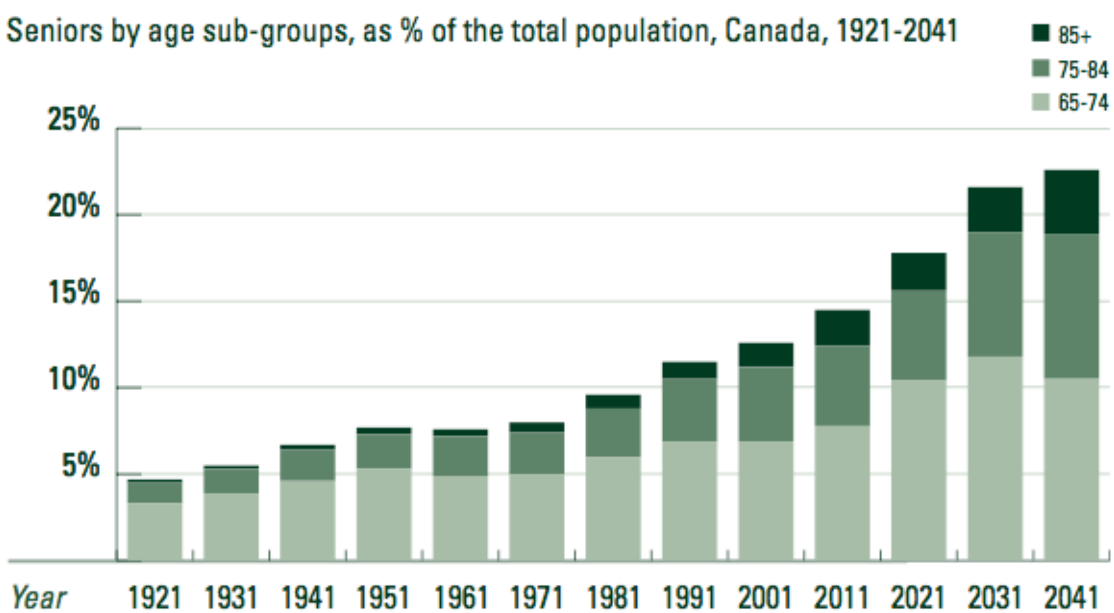


**Figure 1-1: Demographic projections for Canadian population (adapted from Health Canada (2002))**

### 1.1.2 Factors Associated with Fall and Fall-related Injury Risk

The risk for sustaining a fall-related injury is influenced by a multi-faceted interaction pathway (Figure 1-2). Predictably, this risk is directly dependent on the risk for sustaining a fall. It can therefore be inferred that any condition predisposing an individual to a fall will also increase his or her risk of sustaining a fall-related injury.

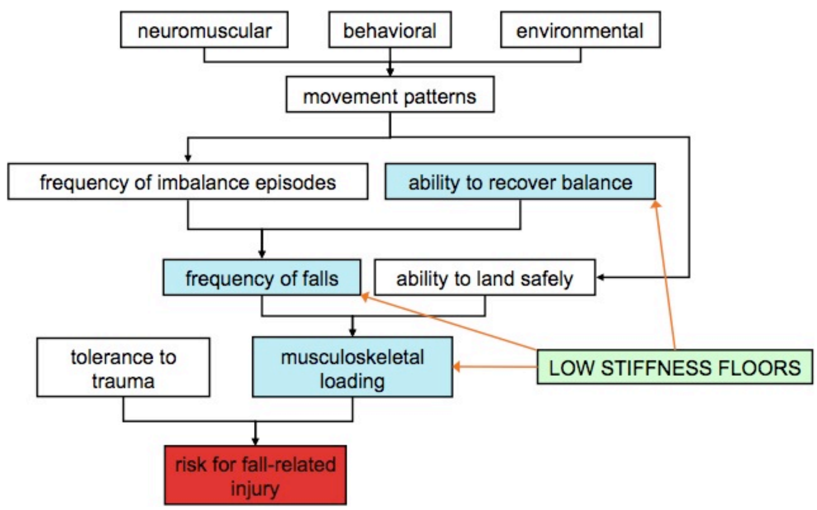


**Figure 1-2: Theoretical representation of the factors that influence risk for fall-related injury; Low-stiffness floors have the potential to substantially attenuate the loads applied to body tissues following impact, but it is essential that this is not accompanied by impairments to balance or mobility**

Many debilitating illnesses that affect older adults may increase their risk for falling. Degenerative joint diseases, arthritis, and other orthopaedic illnesses, Parkinson's disease, and residual impairments from stroke have all been associated with an increased risk for falling (Campbell et al., 1989; Tinetti et al., 1995; Ferrell and Tanev, 2002). Other age-related conditions that increase an older adult's chances of suffering a fall include orthostatic intolerance, foot disorders, and any impairments to the visual, auditory, vestibular, and/or proprioceptive systems (Duncan et al., 1992; Ferrell and Tanev, 2002; Tinetti, 2003).

Weakness of the upper and lower limbs has been found to increase risk for fall-related injury by two and five-fold, respectively (Nevitt and Cummings, 1993; Tinetti et al., 1995; Schwartz et al., 1998). The inability to rise from a chair without the use of one's arms increases the risk for hip fracture by a factor of 1.7 (Cummings et al., 1995). The use of certain medications has been linked to higher fall rates due to their influence on postural reflexes, reaction times, and orthostatic hypotension (Nevitt and Cummings, 1993; Ferrell and Tanev, 2002). The use of anticonvulsant drugs (versus not taking such drugs) has been associated with a doubled relative risk for hip fracture, while the use of long-acting benzodiazepines has been associated with a relative risk of 1.6 (Cummings et al., 1995). A history of falls and fractures (particularly after the age of 50) is also associated with an increased future risk of falling and injury (relative risk for hip fracture of 1.5 with previous fracture (Cummings et al., 1995)), and is consequently used as a basic screening question when assessing current fall risk (Kelsey and Samelson, 2009). History of maternal hip fracture has been associated with a doubled relative risk of hip fracture for women (Cummings et al., 1995). Sideways falls have been linked to a five-fold increase in risk for hip fracture, while falls that involve impact to the hip region increases the risk for hip fracture by nearly 22 times (Hayes et al., 1993). Even when injuries do not occur following a fall, the consequent "fear of falling" can dramatically influence an individual's behaviour, leading to diminished activity, mobility, and independence, which can accelerate age-related declines in muscle force and function (Alexander et al., 1992; Boulgarides et al., 2003). In fact, this "fear of falling" is reportedly developed in the *absence* of recent falls in 20-50% of older adults (Maki and McIlroy, 2005).

Traditionally, it has been emphasized that the majority of falls (up to 50%) occur during the act of walking (Nevitt and Cummings, 1993; Greenspan et al., 1994; Lee and Kim, 1997). Accordingly, many studies have reported that falls risk is closely associated with variability in temporal and spatial gait characteristics including stride time, stance time, step width, and step length (Maki, 1997; Brach et al., 2005). In fact, either too much *or* too little variability in step width has been associated with fall history in older adults walking at or near normal gait speed (Brach et al., 2005). This may be the result of age-related declines in function such as strength loss and/or visual impairments; while human locomotion is thought to be passively stable in the anterior-posterior direction, the substantial active control that is required for motion in the lateral direction (McGeer, 1990; Kuo, 1999; Bauby and Kuo, 2000; Donelan et al., 2004) may become challenging with age-related reductions in strength. Visual input plays a key role in moderating lateral variability as well. Bauby and Kuo (2000) report a 53% increase in lateral variability when participants were required to walk with eyes closed compared to an eyes open condition, as opposed to a 21% increase in sagittal variability. Thus, any age-related impairments to the visual system could also influence gait mechanics and affect fall risk.

### 1.1.3 Recent Evidence

All falls experienced by older adults do not lead to injury. Despite the finding that over 90% of hip fractures and TBI are due to falls (Grisso et al., 1991; Pickett et al., 2001), only 1-2% of falls result in serious injuries such as hip fractures (Tinetti et al., 1988; Nevitt et al., 1991). Thus, it is pertinent to consider the characteristics and mechanics of falls that are most often associated with injuries.

Traditionally, slips and trips have been implicated as the most common causes of falls leading to injury (Gabell et al., 1985; Cummings et al., 1988; Topper et al., 1993; Parkkari et al., 1999). Much of the data that has been gathered on this topic has been obtained through questionnaires and self-reports from patients following injury. However, many studies emphasize the potential of subject recall bias to be a significant source of error in their findings (Nevitt and Cummings, 1993; Greenspan et al., 1998; Parkkari, 1998; Schwartz et al., 1998; Wei et al., 2001; Keegan et al., 2004). Cummings et al. (1988) concluded that elderly participants often do not accurately recall having endured a fall over the preceding 3-12 months, let alone the exact characteristics of the fall. Other studies have found similar time-related errors in fall recollection for older patients (Ganz et al., 2005; Mackenzie et al., 2006). After imposing unexpected falls to young, healthy adults, Feldman and Robinovitch (2006) found that these participants were generally unable to accurately recall specific fall mechanics immediately afterwards. These limitations to our understanding of what types of falls lead to injuries have necessitated a more objective approach to answering this question.

A novel approach to studying fall mechanics was recently undertaken that challenges the traditional view of fall mechanics in elderly fallers. Robinovitch et al. (2009) analysed video footage from cameras that were installed in long-term care facilities and documented the causes and circumstances associated with falls. Of 81 falls that were captured on video, only 15% occurred during the act of forward walking, while 28% occurred during standing. Slips or trips caused only 15% of falls. The most common cause of falls was incorrect weight transfer; for example, rising from a seated to standing position, or transferring weight to put on a coat while standing. Also of interest is the finding that backwards falls occurred more than twice as often as forward or sideways falls, and head impact was involved in almost 30% of cases (Robinovitch et al., 2009). These findings have important implications in the design and testing of effective interventions to reduce fall-related injuries.

### 1.1.4 Balance Control and Compensatory Balance Reactions in Upright Stance

The control of postural stability arises from a complex interaction of various sensorimotor processes (Horak, 2006). From a biomechanical perspective, to maintain an upright posture during static or quasi-static activities such as quiet stance, the vertical projection of the whole-body centre-of-mass (COM) must fall within the limits of the base of support (BOS) defined by the anterior, lateral, and posterior borders of the feet (Winter, 2009). To accomplish this, models related to ankle stiffness (Winter et al., 1998; Winter et al., 2001) and reactive muscle strategies (Morasso and Schieppati, 1999; Morasso and Sanguineti, 2002) predict that adjustments in the location of the underfoot centre-of-pressure (COP) are used to guide or 'shepherd' the trajectory of the COM towards equilibrium, a region of space determined by the size of the BOS (also influenced by joint range of motion, muscle strength, and sensory information available to detect BOS boundaries) (Horak, 2006).

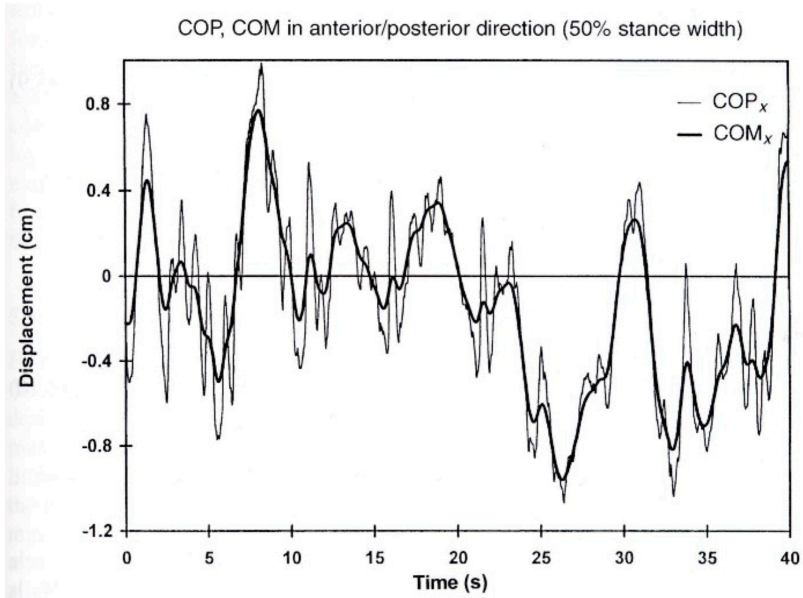


**Figure 1-3: Illustration of how underfoot centre-of-pressure is used to control trajectory of the whole-body centre-of-mass (from Winter, 2009)**

In the event of a balance perturbation that causes the COM to shift anteriorly (such as being nudged from behind), recovering balance without changing the BOS via stepping responses necessitates a rapid anterior shift of the COP to decelerate the COM. Any rate reduction or delay in COP displacement may preclude the COM trajectory from being altered quickly enough to prevent it from reaching the BOS boundaries, requiring the individual to increase the BOS by taking a step in order to prevent a forward fall (Maki et al., 2001; Winter, 2009). It has frequently been proposed that the earliest phase of compensatory responses to balance perturbation during upright stance occurs "automatically", involving activation of ankle muscles at short latencies of 80-140 ms (Nashner, 1976; Nashner and Cordo, 1981; Allum, 1983; McIlroy and Maki, 1993; Norrie et al., 2002; Mochizuki et al., 2010), which is then followed by a secondary stabilizing response. As a result, such early responses have historically been referred to as 'automatic' postural responses. The term 'automatic' is misleading however, as it does not acknowledge the role played by the autonomic nervous system in the control of body function, which acts largely below the conscious level. Consequently, these responses will be referred to as either early compensatory balance reactions or autonomic postural responses (APRs) in the remainder of the current document. The gain of such APRs has been shown to be modified by the magnitude of the triggering balance perturbation, with such responses being exhibited whether or not stepping is used to recover balance (McIlroy and Maki, 1993; Maki and McIlroy, 2007). Numerous reports have suggested that the evoked APRs require minimal cognitive drive (Brown et al., 1999; Rankin et al., 2000; Maki et al., 2001; Norrie et al., 2002), and are not affected if input from cutaneous mechaoreceptors on the plantar surface of the foot is reduced (Perry et al., 2000). Due to their prominence following multiple types of perturbations and responses, the initial APRs are widely reported when characterizing compensatory balance reactions to upright stance.

### 1.1.5 Biomechanical Paradigms for Assessing Balance and Falls Risk

Many different biomechanical paradigms exist for assessing balance and falls risk. Chiu et al. (2003) describe that a comprehensive balance test should evaluate three separate contexts: i) static balance maintenance; ii) postural adjustments to voluntary movements (dynamic voluntary); and iii) postural responses to external perturbations (dynamic external). It is important that all three contexts be assessed, as some previous studies have found minimal association between performances in two different contexts (Maki et al., 1990; Owings et al., 2000; Mackey and Robinovitch, 2005), which may be due to fundamental differences in the neuromuscular demands of each type of task (Mackey and Robinovitch, 2005). It is also essential that the tasks chosen for these balance tests most accurately represent conditions for which fall incidence is high.

The most commonly reported method to assess balance maintenance during static conditions uses measurements of postural sway during quiet stance (Ring et al., 1989; Lord et al., 1991; Teasdale et al., 1991; Lord and Menz, 2000; Gill et al., 2001). This generally entails examination of the trajectories of the COM and COP. The amplitude of COM and COP excursion during quiet standing postural sway, particularly in the medio-lateral direction, is thought to be one of the best tools to identify individuals with a high risk of falling, and has frequently been associated with increased fall risk (Campbell et al., 1989; Maki et al., 1994; Thapa et al., 1996). Furthermore, visual input has been reported to significantly associate with amplitude and velocity of postural sway; in the absence of visual feedback (as with the eyes closed), postural sway increases substantially (Redfern et al., 1997; Dickinson et al., 2001).

Postural responses to voluntary movements represent the second important component to comprehensive balance assessment. The traditional view of fall mechanics acknowledges that up to 21% of falls are associated with reaching tasks (Nevitt and Cummings, 1993; Lee and Kim, 1997). Consequently, the functional reach test exemplifies one commonly used and highly relevant test paradigm. Lower scores on this test have been associated with fall risk in elderly persons on multiple occasions (Duncan et al., 1992; Huang et al., 2006; Huang, 2006). An individual’s ability to transfer from quasi-static to dynamic situations has also been implicated in risk of falling (Topper et al., 1993). Reaching or turning at the time of a fall increases the risk of injury 3-fold (Nevitt et al., 1991). The Timed Up and Go test (TUG) combines numerous transitional activities, including transfers to and from sitting and standing positions, gait initiation, and turning (Mathias et al., 1986; Podsiadlo and Richardson, 1991). Several studies have reported associations between the time taken to complete the TUG and fall risk (Podsiadlo and Richardson, 1991; Lundin-Olsson et al., 1998; Shumway-Cook et al., 2000; Chiu et al., 2003).

Lastly, it is important to assess postural responses to externally invoked perturbations. A number of experimental paradigms are commonly used to simulate balance loss, including sudden floor/platform translations (McIlroy and Maki, 1996; Pavol et al., 2002; Laing and Robinovitch, 2009; Wright and Laing, 2011), introduction of a slippery surface during walking (Pavol et al., 1999; Pijnappels et al., 2004), and external pushing or pulling forces applied to the trunk of the body during upright stance (Luchies et al., 1994; Rogers et al., 2001). These tests effectively simulate many real-life balance perturbations, such as tripping, stepping onto a wet floor or being pushed by another person unintentionally. Another common biomechanical paradigm for inducing external perturbations involves the use of a tether-release system (Wojcik et al., 1999; Robinovitch et al., 2002; Grabiner et al., 2005; Grabiner and Troy, 2005). In this type of experiment, participants are inclined into a stationary leaning position (usually forward) and held in place by a horizontal tether. The participants are instructed to maintain balance following a sudden release of the tether. The ability to accomplish this task decreases substantially with age (Hsiao-Wecksler, 2008). Participants may be asked to recover balance while keeping a fixed BOS, or they may be allowed to take one or more steps; both simulate real-life situations. Stepping responses are prevalent following large magnitude and unexpected perturbations (Maki and McIlroy, 2005). While such change-in-support responses are unquestionably important in preventing falls, a variety of real-life situations (such as being nudged in a crowd, or standing on a bus that accelerates quickly) permit only fixed BOS responses to recover balance. Recent evidence indicating that most injurious falls occur in situations where the feet remain stationary (Section 1.1.3) further supports the importance of assessing fixed BOS responses to external perturbations. Whether or not the BOS is permitted to change, participants' compensatory balance reactions to externally-induced perturbations can be evaluated by characterizing autonomic postural responses (Section 1.1.4), in terms of latency to initial COP displacement, latency and magnitude of maximum COP displacement, and maximum rate of COP displacement (McIlroy and Maki, 1993; Perry et al., 2000; Maki et al., 2001; Wright and Laing, 2011).

Thus, there are three conceptual types of tests that comprise a comprehensive balance assessment. Any intervention to reduce fall-related injuries that may have an influence on balance and balance control responses should be evaluated using multiple test paradigms to evaluate these three contexts.

## 1.2 Traumatic Brain Injuries in Older Adults

The overall objective of this thesis was to assess the influence of novel compliant flooring systems on factors associated with risk for fall-related injuries, including traumatic brain injuries (TBI), in older adults. Consequently, the following sections will describe the scope of the TBI problem, and the biomechanics and pathophysiology associated with TBI.

### 1.2.1 Epidemiology

Traumatic brain injuries have a bimodal distribution among the human population, with the highest incidence occurring in young adults as a result of motor vehicle accidents, and a second peak occurring in the elderly population (Flanagan et al., 2005). TBI in older adults are 10-times more likely to be as a consequence of unintentional falls (up to 90%)

than the second leading cause of motor vehicle accidents (9%) (Pickett et al., 2001; Thompson et al., 2006). Fall-related TBI can account for a substantial portion of the costs due to fall-related injuries in older adults. Seniors are hospitalized twice as often as the general population for fall-related TBI, while over half of all fall-related deaths in older adults are due to TBI (Thomas et al., 2008). The rates of fall-induced TBI-related deaths have been on the rise, increasing by over 25% between 1989-1998 (Adekoya et al., 2002) and accounting for 52,000 annual deaths in the United States (Ferrell and Tanev, 2002). The risk for fall-related TBI increases substantially with age; persons over the age of 85 are hospitalized for fall-related TBI over twice as often as those aged 75-84, and over six times as often as those aged 65-74 (Coronado et al., 2005). The majority of elderly patients hospitalized for fall-related TBI are not discharged to the home, but instead to other facilities such as nursing homes and long-term care facilities (Thomas et al., 2008). The current epidemiological data clearly establish fall-related TBI as a major public health issue (Cameron et al., 2008).

### 1.2.2 Biomechanical Causes of Fall-related TBI

Although the exact pathway between mechanical insult and resultant cognitive deficit is yet to be fully understood (Feng et al., 2010), traumatic brain injuries are fundamentally due to straining of the brain tissue and blood vessels within the brain (Shorten and Himmelsbach, 2003; Cory et al., 2001). This generally occurs via one of three primary mechanisms. Firstly, external forces may be gradually applied to a stationary head. Secondly, rapid acceleration or deceleration of the head in the absence of contact with any external object can lead to impulsive forces applied to the brain, causing injury. Thirdly, and most often the case with fall-related TBI, the head may directly contact another surface

(Ferrell and Tanev, 2002; McHenry, 2004). Even without fracture of the skull, direct impact can cause linear and rotational accelerations of the brain within the brain cavity, creating local pressure changes and shear strains that may cause disruption and tearing of small blood vessels (King, 2000; Ferrell and Tanev, 2002; Shorten and Himmelsbach, 2003; Singh et al., 2006; Hardy et al., 2007; Ivancevic, 2009). Falls involving direct head impacts typically result in focal brain damage, including contusions, lacerations, and intracranial haemorrhaging. The damage may occur close to the point of impact, as in a *coup*-type injury, or opposite to the point of impact, classified as a *contrecoup* injury.

Regardless of the location of the focal damage, rupture of the subdural bridging veins can cause subdural hematoma (Cory et al., 2001; Ferrell and Tanev, 2002). This phenomenon occurs more easily and frequently in older adults, as age-related brain atrophy causes stretching of the bridging veins, making them more susceptible to tearing (Flanagan et al., 2005). Subdural hematomas typically expand slowly in older adults, leading to gradual accumulation of fluid over the weeks following the initial injury. The resultant increase in intracranial pressure may decrease perfusion to the brain, creating ischemic conditions and potentially furthering cerebral damage. However, subdural hematomas also have the potential to grow very rapidly and cause expeditious deterioration of neurologic function (Flanagan et al., 2005).

After brain cells have torn, leaking of potassium and calcium ions in the surrounding interstitial fluid result in a disruption of the ion concentration gradients that are essential for proper transmission of neural signals (Katayama et al., 1990; Park et al., 2008). Excessive levels of calcium cause initiating factors to be released from depolarized mitochondrial membranes, triggering programmed cell death. Furthermore, the remaining healthy brain cells metabolize higher levels of glucose, consuming more energy and leaving the brain more vulnerable to another injury (Shorten and Himmelsbach, 2003).

### 1.2.3 Age-related Differences in Long-term TBI Outcomes

Compared to younger adults, older adults have a much greater risk of substantial cognitive decline and death following TBI (Goleburn and Golden, 2001; Ferrell and Tanev, 2002; Flanagan et al., 2005). TBI accelerates both the age-related loss of neuronal networks and compromised integrity of the white matter matrix (Green et al., 2008). Up to 60% of older adults who experience TBI develop major depression in the year following injury (Hibbard et al., 1998), while other psychiatric disorders such as anxiety and behavioural dyscontrol may also arise (Flanagan et al., 2005). Although still a topic of debate, various studies have suggested a link between TBI and the development and progression of Alzheimer's disease and other forms of dementia (Nemetz et al., 1999; Fleminger et al., 2003). While many of the long-term effects of TBI in older adults have yet to be clarified, research to develop effective prevention of such injuries remains to be the optimal approach to minimize the burden of TBI (Adekoya et al., 2002).

## 1.3 Biomechanical Tools for Modelling Head Impact

The first study proposed as part of this thesis (Chapter 2) will use a mechanical test system to evaluate impact dynamics during simulated head impacts. The following section presents relevant background information relating to established injury criteria, test system designs and testing protocols.

### 1.3.1 Criteria for Predicting Head Injury

Since the 1960's, researchers have devised numerous methods to predict criterion values for the tolerance limit of the head to impact accelerations (McElhaney, 1976; Shorten and Himmelsbach, 2003; Cory et al., 2001). Such limits have been very useful for the development of safety standards for automobile crash tests and athletic equipment performance. Nevertheless, all of the current criteria have inherent shortcomings, leaving industry and researchers still in search of one optimal head injury predictor (McHenry, 2004; Cory et al., 2001; Hardy et al., 2007).

#### 1.3.1.1 Wayne State Tolerance Curve

The first model developed to assess the relationship between the tolerance of the human head to impact was the Wayne State Tolerance Curve (WSTC), initially introduced in 1960 (McElhaney, 1976). The WSTC was developed to ascertain the impact parameters required to fracture embalmed cadaver heads that were dropped onto flat, rigid surfaces. The roughly logarithmic WST curve provides an estimate of the required combinations of head acceleration and pulse duration during impact to elicit permanent damage, and established that the tolerable acceleration level decreases with increased pulse width (McElhaney, 1976; Shorten and Himmelsbach, 2003; McHenry, 2004).

The applicability of the WSTC is limited primarily by the fact that a mechanism linking to functional brain damage could not be established. Furthermore, this curve is incapable of accounting for rotational accelerations of the head that often accompany impacts. It is also heavily criticized for having been developed using only 6 data points that were recorded using questionable instrumentation methods (McHenry, 2004). Despite its limitations, the WSTC has formed the foundation for today's accepted indices of head injury tolerance.

#### 1.3.1.2 Gadd Severity Index

In 1966, Gadd presented an approach for predicting head injury severity based on an extension of the WSTC (Gadd, 1966). He determined that the logarithmic slope of the WSTC is roughly 2.5, a weighting factor he then used in the calculation of the Gadd Severity Index (GSI, also referred to as simply the Severity Index) (McElhaney, 1976; McHenry, 2004). The GSI is calculated through the following integration of the time varying acceleration history of the head, *a(t)*, following an impact of pulse duration τ:

$$GSI = \left[ \int_0^{\tau} a(t) \right]^{2.5} dt \tag{1.1}$$

A threshold GSI value of 1000 was proposed to represent the limit above which the probability of a life-threatening brain injury is greater than zero (McElhaney, 1976; Shorten and Himmelsbach, 2003). The main criticism of the GSI is the prediction of unrealistically high scores for long duration, low intensity head impacts (Shorten and Himmelsbach, 2003).

#### 1.3.1.3 Head Injury Criterion Score

The shortcoming of the GSI was addressed in 1972 when the National Highway Traffic Safety Administration (NHTSA) introduced the Head Injury Criterion (HIC) score. Although it is derived using a similar method to the GSI, calculation of the HIC requires that "portions of the acceleration-time pulse be analyzed to determine the starting and ending points that yield the highest score" (Shorten and Himmelsbach, 2003). This emphasizes high magnitude accelerations while de-emphasizing lower magnitude accelerations with long durations. The time interval ($t_1$, $t_2$) (where $t_1$ occurs at some time point after the initiation of the pulse and $t_2$ occurs before the cessation of the pulse) along the acceleration-time history of the impact that produces the largest GSI score defines the HIC score:

$$HIC = \max\left[(T_1 - T_0)\left[\frac{1}{T_1 - T_0}\int_{=T_0}^{T_1} a_r(t)dt\right]^{2.5}\right] \tag{1.2}$$

A diagram illustrating the differences between GSI and HIC is provided in Figure 1-4.

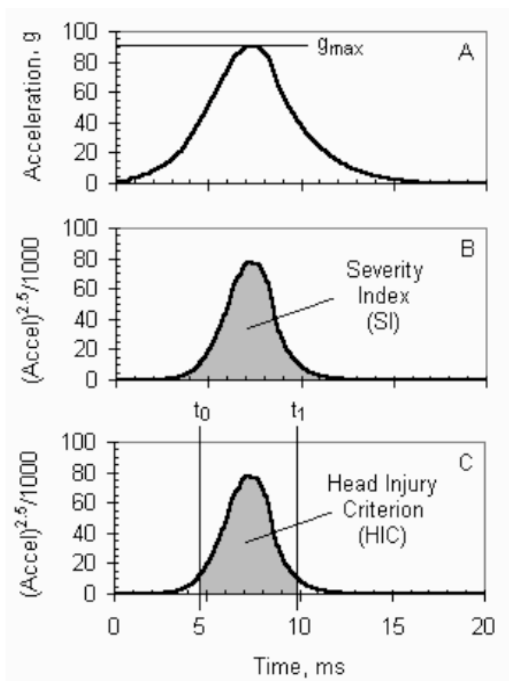


**Figure 1-4: Demonstration of the relationship between acceleration-time history of a head following impact, Gadd Severity Index, and Head Injury Criterion score intervals (from Shorten and Himmelsbach, 2003)**

Empirical measures describing the relationship between HIC scores and the probability of injury are provided by the Expanded Prasad-Mertz curves (Prasad and Mertz, 1985). These curves, illustrated in Figure 1-5, have been used widely in automotive and athletic industries to predict injury risk. Above a HIC score of 1000, the risk of sustaining no injury approaches zero while the risk of a fatal injury becomes greater than zero. At this HIC level, there is an 18% probability of severe head injury, 55% probability of a serious injury, and a 90% probability of moderate head injury for the average adult (Mackay, 2007).

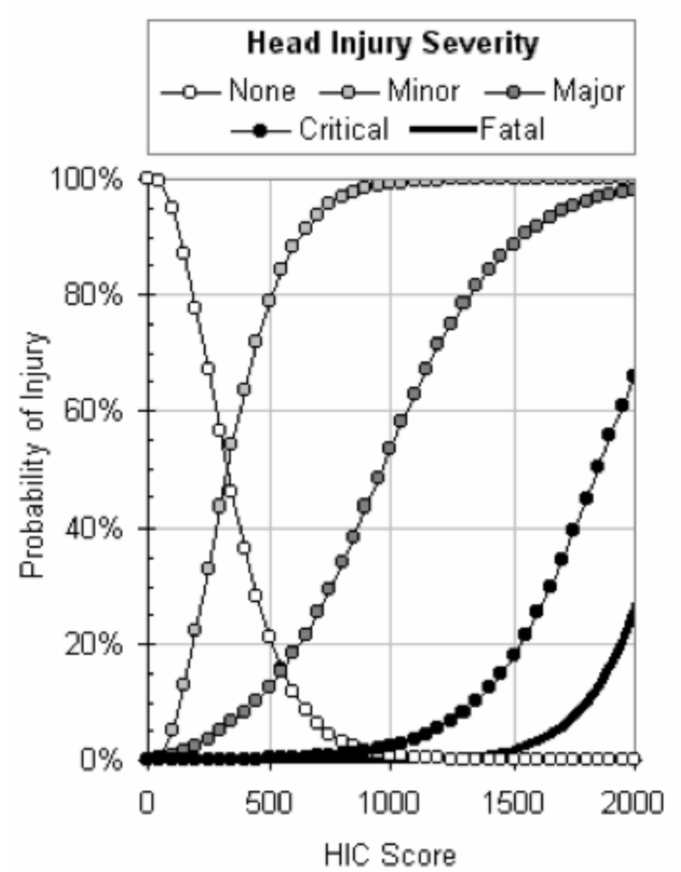


**Figure 1-5: Expanded Prasad-Mertz curves demonstrating the relationship between HIC score and probability of head injury (from Shorten and Himmelsbach, 2003); At a HIC score of 1000, the probability of suffering no injury is reduced to zero, while the probability of a fatal injury becomes non-zero**

Although the HIC addresses the shortcomings of the GSI, it is not without limitations of its own. The determination of this criterion is evidently reliant on the kinematics of the head, which are measured externally. It is not sensitive to direction of impact, nor is it able to model the intracranial behaviour of the brain during impact. Furthermore, the HIC is derived from measures of linear acceleration; while the relative importance of rotational versus linear accelerations is still a matter of debate, rotational accelerations are still thought to be a primary risk factor for TBI (Ommaya and Gennarelli, 1974; Guskiewicz and Mihalik, 2011). The complex etiology of head injuries has created a great degree of difficulty in developing accurate injury criteria and associated thresholds. Consequently, the HIC continues to be the best and most widely used risk prediction tool currently available (Cory et al., 2001; Guskiewicz and Mihalik, 2011).

### 1.3.2 Impact Testing

A very common method of testing the risk of head injury due to impact is to use surrogate biofidelic human headforms (Halstead, 2001). The headforms are generally dropped onto a surface in a guided free fall. Accelerometers mounted at the centre of gravity of the headforms provide quantitative measures of accelerations during these impacts, which may be used to calculate severity indices like those described above. It is often desirable to use triaxial accelerometers for this purpose to compensate for impacts that are not aligned directly with the centre of gravity (Halstead, 2001). Such tests have found widespread use in the development of safety standards for devices including helmets, airbags, seatbelts, and playground surfaces. For example, a GSI score of 1200 is used by the National Operating Committee on Standards for Athletic Equipment (NOCSAE) as the performance limit for certification of helmets for athletic use (NOCSAE, 2004; NOCSAE, 2009). The American Society for Testing and Materials (ASTM International Standard 1292-04) has defined a

HIC score of 1000 to be used for determining the 'critical drop height' (maximum allowable height of playground structures) for a given playground surface (ASTM, 2004).

#### 1.3.2.1 Headform Design

The mass, shape, and material properties of a headform will strongly influence the behaviour of the surface that it impacts, which will in turn dictate the magnitudes of the forces at accelerations it experiences. Some test standards suggest the use of solid spherical aluminium 'missiles' as surrogate headforms (ASTM, 2004). Due to the very low compliance of these surrogates, and consequently faster deceleration upon impact, a conservative over-estimation of the severity of injury is anticipated for any given impact (ASTM, 2004). Contrastingly, other surrogate headforms are designed to provide high biofidelity with respect to both anthrompometry and dynamic response (Higgins et al., 2007). NOCSAE has developed a set of three headforms of varying sizes, designed to match the anthropometric characteristics of 'average' human heads (Figure 1-6). These headforms are constructed with a high durometer urethane skull, covered with a lower durometer urethane that forms the skin and anatomical features (e.g. ears, nose, lips), and have also been manufactured with a glycerin-filled brain cavity to optimally simulate the behaviour of a human head in response to impact accelerations (Higgins et al., 2007; NOCSAE, 2009). Drop tests conducted using biofidelic headforms should therefore predict injury severity for a given impact with a higher degree of accuracy.

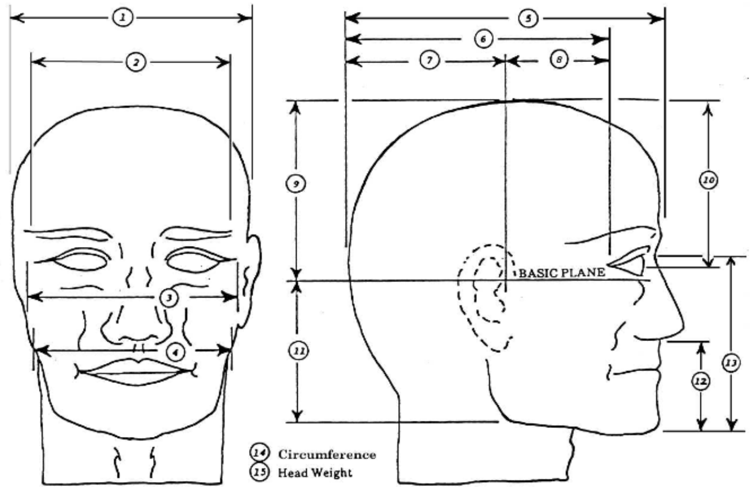


**Figure 1-6: Pictorial description of the anthropometric dimensioning used for NOCSAE headforms (adapted from NOCSAE, 2009)**

## 1.4 Current Strategies to Reduce Fall-related TBI

The body of literature pertaining to the development and testing of interventions designed to stem the rates of fall-related injury in older adults highlights two general conceptual approaches. The first approach is aimed towards decreasing the risk for falls through enhancing balance maintenance and recovery abilities. Interventions such as these involve resistance and agility training (Province et al., 1995; Liu-Ambrose et al., 2004) and exercise programs including Tai Chi (Lan et al., 1996; Kutner et al., 1997; Nowalk et al., 2001; Li et al., 2005). A thorough review of medications by a pharmacist has also been suggested to lower the rate of falls in nursing home residents (Zermansky et al., 2006; Cameron et al., 2010). The second type of approach aims to decrease fall frequency and the likelihood of injury in the result of a fall. This includes programs that teach elderly adults safer fall techniques (Hsiao and Robinovitch, 1998; Groen et al., 2006), making modifications to the environments where falls often occur (reviewed in Cameron et al., 2010), and pharmacologic interventions to increase tissue tolerance (such as bisphosphonates (Liberman et al., 1995; Cummings et al., 1998; Orwoll et al., 2000; McClung et al., 2001) and parathyroid hormones (Neer et al., 2001; Black et al., 2003; Finkelstein et al., 2003) to increase bone mineral density and decrease fracture risk). Additionally, protective devices such as wearable hip protectors have been developed to decrease the forces applied to the body following impact (Cummings and Nevitt, 1994; Kannus et al., 1996; Robinovitch et al., 2000; Keegan et al., 2004; Laing and Robinovitch, 2008a; Laing and Robinovitch, 2008b).

Questions exist as to the efficacy and cost-effectiveness of many current interventions (Cameron et al., 2010). One of the main limitations with many of the existing intervention strategies is the choice of the user to comply with the intervention. For example, the effectiveness of strength training programs depends largely on an individual's dedication to that program. Protective devices like specific models of wearable hip protectors (padding system incorporated into undergarments) have been shown to significantly reduce the risk for hip fracture in older adults (Parkkari et al., 1995; Kannus et al., 1999; Laing and Robinovitch, 2008a; Laing and Robinovitch, 2008b), but also require active user compliance in order to be clinically effective (Haines et al., 2006). Furthermore, this type of device protects only against one type of injury (i.e. hip fracture), and thus will have no influence on the risk for other severe fall-related injuries including TBI. The use of certified helmets would likely reduce TBI risk in the event of a fall, but this approach is impractical for the majority of older adults during normal activities of daily living. As no cure exists for TBI once is has occurred, the single best treatment remains prevention (Ferrell and Tanev, 2002).

One promising approach for reducing both the incidence and severity of fall-related injuries in older adults, including TBI, involves the installation of novel low-stiffness (compliant) flooring systems. This is particularly relevant for environments such as nursing homes, seniors' centres, and residential care facilities where a large number of falls and fall-related injuries occur (Section 1.1.1). This 'passive' intervention is always present thereby removing the issue of non-compliance or non-adherence with the intervention. Furthermore, they protect against multiple types of injury. Novel compliant floors (NCFs) are generally designed to provide a dual-stiffness response, which allows the floor to remain firm under standard loads associated with locomotion, but to deform and absorb energy once a critical buckling load has been exceeded. At least two basic design approaches have been developed to date. One design incorporates a continuous surface layer over an array of cylindrical columns (often rubber), as demonstrated by SmartCell (SATech, Chehalis, WA, USA) and SofTile (SofSurfaces, Petrolia, ON, Canada) floors (Figure 1-7). A second approach uses closed cell polyurethane foams beneath a continuous surface to provide the desired response, such as those floors developed by Kradal (Acma Industries Limited, Upper Hutt, Wellington, New Zealand).

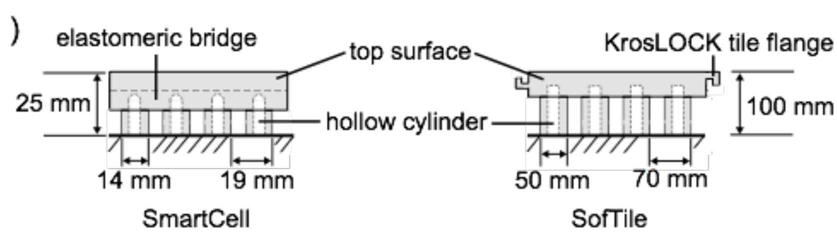


**Figure 1-7: Schematic representation of the buckling column designs of SmartCell (left) and SofTile (right) novel compliant floors**

### 1.4.1 Novel Compliant Floors and Impact Force Attenuation

Two preliminary lines of evidence exist that support the use of novel compliant floors to reduce serious fall-related injuries. Numerous reports have shown that falling on a soft surface (such as padded carpet or grass) reduces the risk for hip fracture compared to falling on a hard surface (such as concrete, linoleum, or vinyl) (Nevitt and Cummings, 1993; Healey, 1994; Simpson et al., 2004). Similarly, laboratory studies have demonstrated that decreased floor stiffness can attenuate the peak impact forces applied to the hip during simulated falls by up to 73% (Maki et al., 1990; Gardner et al., 1998; Minns et al., 2004). Regarding TBI, stiffness of the impact surface highly influences the type and severity of resultant intracranial injuries following direct impact (Gennarelli, 1984; McLean and Anderson, 1997; Cory et al., 2001). Unsuitable surfacing has been found to account for between 79-100% of severe head injuries in playground environments (Mack et al., 2000), while 75% of fall-related deaths near playground equipment involve catastrophic head injury (Tinsworth and McDonald, 2001). Furthermore, it has been demonstrated that the risk of a serious head injury following an impact is 1.7 times greater on grass than on sand (Laforest et al., 2000). Such epidemiologic evidence suggests the merit of further biomechanical studies to determine potential mechanisms underlying these injury trends.

The force attenuative properties of two NCFs have been characterized recently. Laing and Robinovitch (2009) report that certain NCFs are capable of reducing the impact force applied to the proximal femur by 25-50% during simulated sideways falls to the hip. Additionally, peak force attenuation increased with higher impact velocities. Thus, these floors appear to provide a significant protective capacity against hip fractures. This level of force attenuation is larger than those that have been reported for simulated hip impacts onto wooden floors (7%), carpets (15%), and carpets with underpadding (24%) (Maki et al., 1990; Gardner et al., 1998; Simpson et al., 2004). While a greater degree of force attenuation has been reported for simulated hip impacts onto vinyl and carpet floors with PVC underlay (56% and 73%, respectively), the protective capacity of such surfaces has likely been overestimated, as the effective compliance of the pelvic region was not incorporated into the test system (Nabhani and Bamford, 2002; Minns et al., 2004; Nabhani and Bamford, 2004). No independently conducted studies to date have assessed the influence of NCFs relative to traditional flooring surfaces on forces and accelerations applied to the head during simulated falls involving head impact. As such, the first study of this thesis will seek to assess the influence of NCFs compared to traditional flooring surfaces on impact dynamics during simulated head impacts with a mechanical test system.

### 1.4.2 Novel Compliant Floors and Balance

In order for NCFs to successfully reduce the risk for fall-related injuries including TBI, adequate force and acceleration attenuation must be provided without concomitant impairments in balance and mobility; the floor stiffness must not be decreased so substantially as to increase the risk for falling during daily activities. Earlier studies investigating the effect of floor/surface stiffness on balance and locomotion provide insights into the importance of this consideration. Postural sway amplitude during quiet stance on compliant foam surfaces has been shown to increase substantially compared to rigid floor conditions (Ring et al., 1989; Lord et al., 1991; Teasdale et al., 1991; Lord and Menz, 2000; Gill et al., 2001). Excessive reductions in floor stiffness have been associated with a degraded quality of proprioceptive and pressure information from the receptors on the plantar surface of the foot (Lord and Menz, 2000; Betker et al., 2005). Gait mechanics are altered following a single step on compliant foam surfaces, such that the trajectory of the COM is lowered through forward pitching of the trunk, suggesting the possibility of decreased trunk stability. Toe clearance is maintained on subsequent steps however, through modulated muscle activity (Marigold and Patla, 2005). Instability may be avoided during multiple steps on compliant foam surfaces through increases in step length, step width, and step width variability (MacLellan and Patla, 2006). Following transient support surface translations, compliant foam surfaces have been shown to affect COP and COM displacement rates, leading to reduced margins of safety, a possible consequence of reduced effective stiffness at the ankles while standing on the compliant surface (Wright and Laing, 2011). It has become evident that reductions in floor stiffness have the potential to increase the likelihood of postural instability and of suffering an imbalance episode.

There are multiple potential mechanisms by which compliant surfaces may impair an individual's ability to maintain balance using feet-in-place responses in the event of a perturbation. Let us consider the perturbation described in Section 1.1.4 (anterior pitch of the COM, as in a nudge from behind). Prior to the onset of perturbation, degraded quality of information from the mechanoreceptors on the plantar aspect of the feet may impair the ability to detect the anterior, lateral, and posterior borders of the base of support (i.e. the front, side, and back contact points between feet and the floor). Immediately following the onset of perturbation, underfoot surface deformations may reduce rotation at the ankles, causing a delay in the triggering of proprioceptive feedback from triceps surae muscle spindles. Furthermore, if we consider the body, or in this case the foot (of stiffness $k_b$), striking a compliant surface (of stiffness $k_f$), a simple mass-spring model illustrates that more compliant floors would reduce the effective stiffness ($k$) of the foot-floor system, requiring more time to generate peak force ($t_{max}$) under the foot and potentially leading to consequent postural instability (explained by the following equations (McMahon et al., 1987; Laing et al., 2006)):

$$t_{\max} = 1/\omega_n(\pi - \tan^{-1})(u\omega_n/g) \quad \textbf{(1.3)}$$

where $u$ is the impact velocity, the effective mass is $m$, and the natural frequency ($\omega_n$) of the system is:

$$\omega_n = \sqrt{k/m}\ , \quad \textbf{(1.4)}$$

$$k = (k_b k_f)/(k_b + k_f) \quad \textbf{(1.5)}$$

Stabilization of balance following the initial compensatory balance reaction could potentially be compromised by these same mechanisms (inability to detect base of support boundaries, reduced effective stiffness at the ankles leading to reduced magnitude and rates of ankle moments).

Promising results have been reported with respect to any influences on balance and mobility provided by NCFs. Laing and Robinovitch (2009) used a comprehensive balance assessment strategy (see Section 1.2.3) including static tasks, response to voluntary movements, and response to unexpected perturbations. They report that the average time taken to complete the TUG test for older community-dwelling women was not significantly different between a 'rigid' vinyl control condition and the two NCF conditions tested. However, these times were significantly greater for two other compliant surfaces of excessively low stiffness. Secondly, the proportion of successful balance recovery trials following backwards translation of the floor was not different between the rigid and NCF conditions. Thirdly, the root mean square amplitude and velocity of postural sway in the medio-lateral direction in both eyes open and eyes closed conditions were not significantly different between the rigid condition and one of the novel compliant floors (SmartCell). This was also true for postural sway in the anterior-posterior direction for the eyes closed condition. Wright and Laing (2011) extended from these findings, demonstrating that the minimum margins of safety for both the COP and COM were unaffected by SmartCell and SofTile floors, with COP displacement rates being unaffected by the SmartCell floor. These results are a promising indication that appropriately designed novel compliant floors may provide minimal impairments to mobility and postural stability.

Despite the encouraging results regarding the limited influence of NCFs on balance and mobility, there are major limitations to these studies (Laing and Robinovitch, 2009; Wright and Laing, 2011) that require consideration. Perhaps most importantly, the participants included in these studies were community-dwelling older women, as opposed to the residents of environments with higher rates of falls and fall-related injuries (e.g. residential care facilities) where the population of older adults will be increasingly frail, and NCFs are most likely to be installed (2009; Wright and Laing, 2011). Additional studies are required to determine whether their findings are generalizable to this higher-risk population. Secondly, both studies only recruited females under the rationale that women are significantly more likely to suffer hip fractures than men. However, exclusion of males is questionable as the incidence of fall-related TBI in men over the age of 65 is roughly twice that for women (Colantonio et al., 2009). Thirdly, it is pertinent to evaluate the relative influence of NCFs on balance control as compared to traditional compliant flooring systems, such as carpets with underpadding, which have also been shown to exhibit modest force attenuative properties and a consequent protective capacity against injuries. These limitations will be addressed in the second study of this thesis that investigates the influence of a range of flooring conditions on the earliest phase of compensatory balance responses of men and women from a retirement-home environment following external perturbations. Such information is necessary to inform design and implementation strategies for compliant floors intended for use in high-fall-rate areas such as residential care facilities and hospitals.

## 1.5 Thesis Objective and Summary of Studies

The overall objective of my research was to evaluate the effectiveness of a range of NCF designs to reduce the risk for fall-related injuries. In particular, I aimed to determine the influence of these floors on factors associated with fall-related TBI risk. To accomplish this objective, analysis of the competing demands of this intervention approach are required- namely, attenuation of impact force and acceleration without concomitant balance impairments.

Two studies are included as part of this thesis. The first entails the use a mechanical head impact simulator system to assess the influence of NCFs on indices of TBI risk during simulated 'worst-case' head impacts. Specifically, the forces and accelerations applied to a surrogate headform during impact are used to evaluate impact dynamics across a range of flooring and impact velocity conditions. The second study utilized a tether-release paradigm to evaluate the influence of NCFs, compared to traditional flooring surfaces, on autonomic postural responses during compensatory balance reactions in older men and women residing in a retirement home setting.

# CHAPTER 2 THE INFLUENCE OF HEADFORM ORIENTATION AND FLOORING SYSTEMS ON IMPACT DYNAMICS DURING SIMULATED FALL-RELATED HEAD IMPACTS

## 2.1 Background

Fall-related injuries in adults over the age of 65 are a major public health issue in Canada, and are associated with direct annual costs of over $2 billion (SMARTRISK, 2009). A substantial portion of this figure may be attributed to fall-related traumatic brain injuries (TBI), which are precipitated by falls in up to 90% of cases (Pickett et al., 2001). Seniors are hospitalized twice as often as the general population for fall-related TBI, while over half of all fall-related deaths in older adults are due to TBI (Thomas et al., 2008). The incidence of fall-induced TBI and associated deaths has been rising at alarming rates, increasing by over 25% between 1989-1998 (Adekoya et al., 2002). The risk for fall-related TBI increases substantially with age; persons over the age of 85 are hospitalized for fall-related TBI over twice as often as those aged 75-84, and over 6 times as often as those aged 65-74 (Coronado et al., 2005). As there is no cure for TBI once it has occurred, prevention remains the optimal approach for reducing associated injury and disability (Adekoya et al., 2002). Considering the ageing Canadian population (Health Canada, 2002), it is imperative that effective intervention strategies be designed and implemented to stem the social and economic impact of the anticipated rise in fall-related TBI incidence over the coming decades.

Development of effective intervention strategies necessitates an understanding of the cause of TBI. While the exact pathway between mechanical insult and cognitive deficit is not yet fully understood (Feng et al., 2010), it is generally recognized that the majority of fall-related TBI occur as a result of the head directly striking another surface (Ferrell and Tanev, 2002; McHenry, 2004). Even without fracture of the skull, direct impact can cause linear and rotational accelerations of the brain within the brain cavity, creating pressure fluctuations and shear strains that may lead to the tearing of small blood vessels and widespread disruption of axons (King, 2000; Ferrell and Tanev, 2002; Singh et al., 2006; Hardy et al., 2007; Ivancevic, 2009). The type and severity of intracranial injuries resulting from direct head impact is highly influenced by the stiffness of the impact surface (Gennarelli, 1984; McLean and Anderson, 1997; Cory et al., 2001). Indeed, previous research reports that unsuitable surfacing has been found to account for between 79-100% of severe head injuries in playground environments (Mack et al., 2000).

Towards the goal of reducing fall-related TBI in older adults, one promising approach entails the installation of novel low-stiffness, or compliant flooring systems. Novel compliant flooring systems (NCFs) are generally designed to provide a dual-stiffness response characterized by minimal deflection during locomotion, and a transition to increased compliance at the higher loads associated with fall-related impacts. Certain models of these commercially available products have been shown to attenuate the impact force applied to the proximal femur by up to 50% during simulated lateral falls compared to commercial-grade vinyl (Laing and Robinovitch, 2009), suggesting a significant protective capacity against hip fractures. This degree of force attenuation is far greater than levels that have been reported for wooden floors (7%), carpets (15%), and carpets with underpadding

(24%) (Maki et al., 1990; Gardner et al., 1998; Simpson et al., 2004). However, no independently obtained information is currently available with respect to the influence of novel versus traditional compliant flooring systems on impact dynamics during simulated head impacts.

Evaluation of head impact dynamics is commonly accomplished using mechanical impact simulators. Such tests have found widespread use in the development of safety standards for devices including helmets, airbags, and playground surfaces. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) has developed biofidelic surrogate human headforms that match the anthropometric characteristics of ‘average’ human heads, and include a glycerin-filled ‘brain cavity’ to optimally simulate the behaviour of the human head in response to impact (Higgins et al., 2007; NOCSAE, 2009). Decades of head impact research have produced risk curves and associated injury thresholds for skull fracture and TBI following impact based on force and acceleration profiles, as well as derived injury criteria such as the Head Injury Criterion (*HIC*) (Gurdjian et al., 1966; Prasad and Mertz, 1985; ASTM, 2004; Mackay, 2007; Funk et al., 2011). Simulated head impacts have been widely used to evaluate head injury risk, including during falls on taekwondo mats (Hrysomallis and McLaughlin, 1999), falls onto playground surfaces (ASTM, 2004), and impacts during athletic competition (Pellman et al., 2003). Despite the widespread use of simulated head impacts using surrogate headforms, the effect of headform orientation, and consequent impact location, has rarely been reported.

## 2.2 Purpose and Hypotheses

Accordingly, the objectives of the current study were to determine: (a) the ‘worst-case’ orientation for simulated head impacts using a biofidelic surrogate human headform based on measures associated with risk for skull fracture and TBI, including peak resultant acceleration ($g_{max}$), Head Injury Criterion score (*HIC*), and peak force ($F_{max}$); and (b) the influence of 10 flooring surfaces on these outcome variables during ‘worst-case’ impacts, relative to a traditional compliant flooring surface (commercial-grade carpet with underpadding). It was hypothesized that the added compliance associated with the headform’s ear (during side impacts) and nose (during front impacts) would lead to reductions in the magnitudes of all outcome variables compared to impacts of the back of the headform. Furthermore, it was hypothesized that during impacts in the ‘worst case’ head orientation, impacts onto novel compliant flooring systems would result in lower applied forces and accelerations (e.g. $g_{max}$, *HIC*, and $F_{max}$) compared to impacts onto a commercial-grade carpet. Finally, it was also hypothesized that the commercial carpet would provide significant force and acceleration attenuation relative to a commercial-grade vinyl.

## 2.3 Methods

The experimental protocol used for this investigation involved a custom protocol that integrated elements from ASTM Standard F 1292 – 04 (Standard Specification for Impact Attenuation of Surfacing Materials within the Use Zone of Playground Equipment) (ASTM, 2004) and NOCSAE Document 001 – 08m08b (Standard Test Method and Equipment used in Evaluating the Performance Characteristics of Protective Headgear/Equipment) (NOCSAE, 2009).

### 2.3.1 Test System

A mechanical drop tower (Dixon and Brodie, 1993) was used to impact a medium-sized surrogate human headform developed by the National Operating Committee for Standards on Athletic Equipment (NOCSAE) onto various flooring surfaces (Figure 2.1).

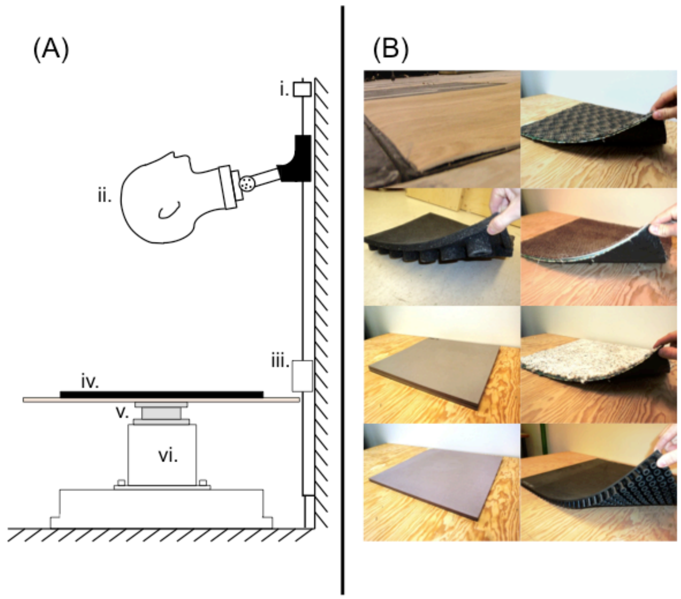


**Figure 2-1:** (A) **Schematic of the mechanical head impact simulator with the following elements highlighted; i. mechanical release; ii. surrogate headform with accelerometer mounted at centre of mass; iii. light gate velocimeter; iv. flooring sample; v. load cell; vi. concrete base.** (B) **Pictures of the floor conditions tested (clockwise from top left): Vinyl (*V*), Commercial Carpet (*CC*), Residential Carpet (*RC*), Berber Carpet (*BC*), SmartCell (*SC*), 12 mm Kradal ($KR_{12}$), 24 mm Kradal ($KR_{24}$), and SofTile (*ST*); not shown in this figure are SmartCell with vinyl overlay (*SC-V*) and SofTile with vinyl overlay (*ST-V*).**

Detailed headform specifications have been reported (Higgins et al., 2007; NOCSAE, 2009), but in brief, the headform was comprised on a glycerin-filled acrylonitrile butadiene styrene (ABS plastic) brain cavity, surrounded by separate urethane skull and facial features. An adjustable mechanical release enabled impacts at velocities of 1.5, 2.5, and 3.5 m/s, which were verified for each trial using an infrared light gate velocimeter (Model VS300, GHI Systems, Aurora, ON, Canada). These impact velocities were decided upon using energy conservation principles based on falls from standing height, pilot testing, and previous research using mechanical impact simulators to assess impact dynamics of the hip (Laing and Robinovitch, 2009). A triaxial accelerometer (Model 2707A, frequency range: 0 – 2000 Hz; Endevco Corporation, San Juan Capistrano, CA, USA) mounted at the centre of mass of the headform recorded impact accelerations, while a load cell (Model 925M113, Kistler Instrument Corporation, Amherst, NY, USA) mounted beneath the impact surface measured impact forces. Force and acceleration data were sampled at 20,000 Hz. In all cases, 3 sequential trials were completed for each impact condition. The flooring samples were moved between trials to prevent repeated impacts onto the same location.

### 2.3.2 Determining the 'Worst-Case' Impact Orientation

A level loop 'Commercial Carpet' (*CC*) (The Carpet Store, Waterloo, ON), pile height = 6 mm, face weight = 882 g/m$^2$) with 6 mm underpad (a traditional compliant flooring system often found in commercial housing settings) was used as a control condition in this study. In order to determine the 'worst-case' impact orientation, trials were conducted onto *CC* flooring only, using three headform orientations (front (*F*), side (*S*), and back (*B*)) at three impact velocities (1.5, 2.5, and 3.5 m/s), with order of condition combination randomly determined.

### 2.3.3 Novel Compliant Floors versus Traditional Flooring Systems

Nine additional flooring conditions were tested in this study under the 'worst-case' impact orientation at 1.5, 2.5, and 3.5 m/s impact velocities. These included a commercial-grade Vinyl, two additional traditional compliant floors (Residential Carpet, Berber Carpet) and six NCF conditions (Figure 1b). The 'Vinyl' (*V*) condition entailed a 2 mm thick layer of rubber appropriate for installation over concrete or wooden subfloors in institutional settings (Noraplan Classic, Nora Systems Inc, Lawrence, MA, USA). The 'Residential Carpet' (*RC*) condition entailed a polypropylene pile-loop carpet (pile height = 9 mm, face weight = 1085 g/m$^2$) with 6 mm foam-rubber underpadding designed for residential settings. The 'Berber Carpet' (*BC*) condition was the thickest of the carpets, consisting of a synthetic weave looped polypropylene (pile height = 10 mm, face weight = 1221 g/m$^2$) used primarily is residential settings (The Carpet Store, Waterloo, Canada). Six NCF conditions were also tested. 'SmartCell' (*SC*) (SATech, Chehalis, WA, USA) was a 25 mm tall synthetic, 50-durometer rubber flooring system comprising a continuous surface layer overlying a series of cylindrical rubber columns 14 mm in diameter and 19 mm apart. The 'SofTile' (*ST*) floor (SofSurfaces, Petrolia, ON, Canada) used a similar design, with 50 mm diameter columns spaced at 70 mm intervals; the 50 mm thick model was tested. The SofTile and SmartCell floors were also tested with a vinyl overlay (*ST-V* and *SC-V*, respectively), representing a design scenario likely necessary for clinical settings. Two designs from Kradal (Acma Industries Ltd., Upper Hutt, Wellington, New Zealand) comprised the final NCF conditions, including 12 mm ($KR_{12}$) and 24 mm ($KR_{24}$) thick tiles with a relatively stiff top surface over a closed cell polyurethane base layer. During testing, the order of floor-velocity combination was randomly determined.

### 2.3.4 Data Analysis

Accelerometer data was processed according to ASTM Standard F1292-04 for testing impact attenuation of surfacing materials during simulated head impacts (ASTM, 2004). In brief, a fourth-order, dual-pass, low-pass digital Butterworth filter (1000 Hz cutoff) was used before calculating the resultant acceleration from the accelerations in each of the three orthogonal axes as follows:

$$a_r(t) = \sqrt{a_x(t)^2 + a_y(t)^2 + a_z(t)^2} \tag{2.1}$$

$g_{max}$ was recorded as the single largest value from the resultant acceleration-time history for each impact. The *HIC* score was also calculated for each impact, according to the following equation (ASTM, 2004):

$$HIC = \max\left[(T_1 - T_0)\left[\frac{1}{T_1 - T_0}\int_{t=T_0}^{T_1} a_r(t)dt\right]^{2.5}\right] \tag{2.2}$$

where $a_r$ is the resultant acceleration profile and $T_0$ and $T_1$ define the time interval that maximizes the *HIC* score. $F_{max}$ (Figure 2.2) was determined from the force-time profile after filtering using a dual-pass, low-pass 4$^{th}$ order digital Butterworth (500 Hz cutoff, determined from residual analysis).

### 2.3.5 Statistics

#### 2.3.5.1 Determination of the 'Worst-Case' Headform Orientation

A two-way ANOVA was used to assess the influence of impact orientation and impact velocity on $g_{max}$, *HIC*, and $F_{max}$. When significant interactions were found, a one-way ANOVA was used to determine the influence of impact orientation at each impact velocity, with Tukey's post-hoc to compare across the three orientations.

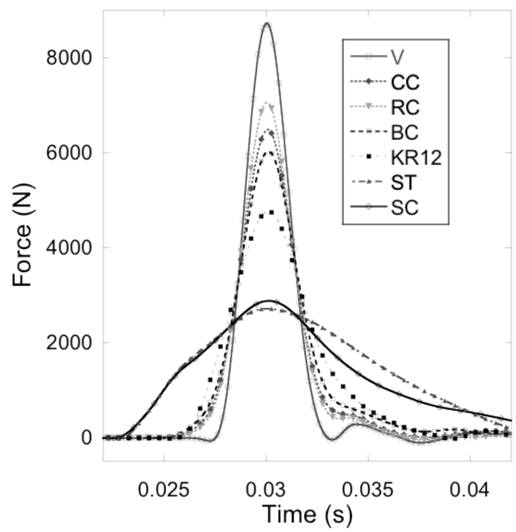


**Figure 2-2: Representative force versus time profiles from 2.5 m/s impacts onto a subset of the floor conditions tested.**

#### 2.3.5.2 Floor Testing

A two-way ANOVA was used to assess the influence of floor condition and impact velocity on each of the outcome parameters. If a significant interaction was found, a one-way ANOVA was used to determine the influence of floor condition at each impact velocity. Dunnett's post-hoc was used to compare each floor relative to the control condition, *CC*.

All statistical analyses were conducted with an experiment-wide significance level of 0.05 using SPSS statistical software package (Version 19.0, SPSS Inc., Chicago, IL, USA).

## 2.4 Results

### 2.4.1 'Worst-case' Orientation

Results from the two-way ANOVAs indicated a significant interaction between impact orientation and impact velocity for all outcome parameters (p always <0.001). Subsequent ANOVA results indicated a significant effect of orientation for all variables at each impact velocity (p always <0.001). Tukey's post-hoc indicated that *B* and *S* impacts consistently yielded higher $g_{max}$, $F_{max}$, and *HIC* values when compared to *F* impacts at all impact velocities (Figure 2.3 and Table 2.1). During impacts at velocities of 1.5 and 2.5 m/s, no differences in any of the outcome parameters were found between *B* and *S* orientations. At 3.5 m/s, $g_{max}$ values were not different, however *HIC* and $F_{max}$ values were significantly greater for *B* impacts.

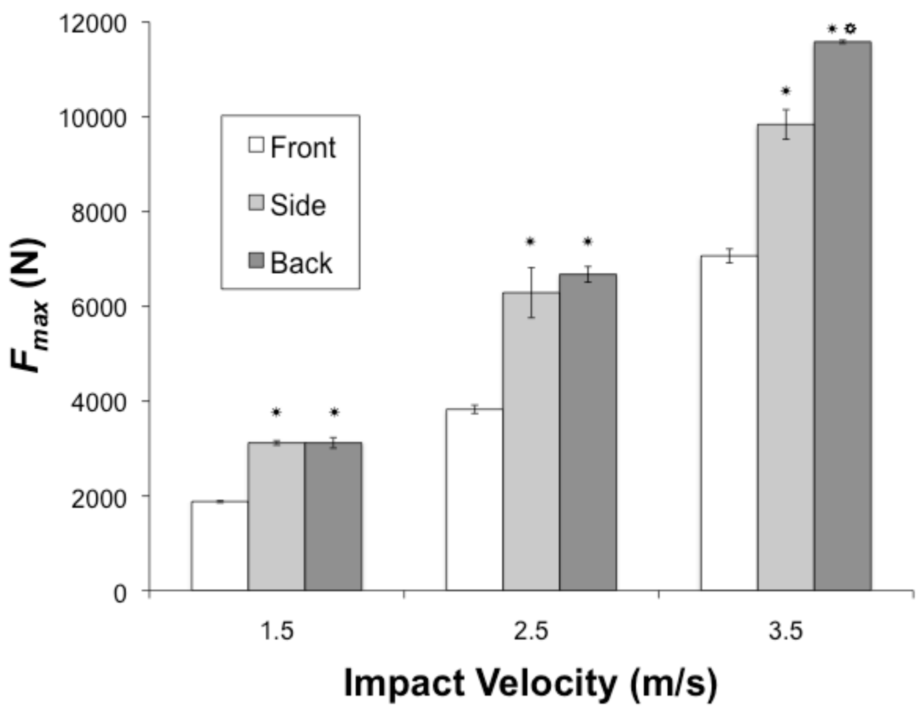


**Figure 2-3: Mean (SD) of peak force ($F_{max}$) for impacts onto the front (*F*), side (*S*), and back (*B*) of the surrogate headform at each impact velocity (testing completed on the control Commercial Carpet condition). * indicates significant (p<0.05) increase relative to *F*, ✲ indicates significant (p<0.05) increase relative to *S*.**

**Table 2-1: Mean (SD) of peak resultant acceleration ($g_{max}$) and Head Injury Criterion (*HIC*) for impacts onto the front (*F*), side (*S*), and back (*B*) of the surrogate headform at each impact velocity on the control Commercial Carpet condition. These results informed the decision to use impacts on the back of the headform to compare across floor conditions.**

| Variable | Orientation | Impact Velocity (m/s$^2$) | | |
|---|---|---|---|---|
| | | 1.5 | 2.5 | 3.5 |
| $g_{max}$ (*g*) | *Front* | 30.7 (0.4) | 62.6 (2.0) | 94.1 (5.7) |
| | *Side* | 62.8 (7.0) * | 123.3 (5.3) * | 263.0 (9.9) * |
| | *Back* | 54.7 (3.4) * | 122.7 (3.8) * | 262.1 (11.1) * |
| *HIC* | *Front* | 27.0 (1.3) | 107.9 (9.1) | 250.4 (15.8) |
| | *Side* | 48.8 (9.3) * | 282.8 (58.4) * | 827.9 (29.0) * |
| | *Back* | 39.1 (3.9) | 258.0 (23.7) * | 1068.0 (40.6) *,# |

* significantly greater than Front orientation (p < 0.05)

# significantly greater than Side orientation (p < 0.05)

### 2.4.2 NCFs versus Traditional Flooring Surfaces

Based on the results in Section 3.1, all additional testing was completed using impacts to the back (*B*) of the surrogate headform. The data is summarized below and in Figures 2-4 – 2-6. It should be noted that impacts at 3.5 m/s were not conducted for the Vinyl (*V*) floor in order to protect the mechanical integrity of the testing system.

#### 2.4.2.1 Peak Acceleration ($g_{max}$)

Peak accelerations ranged from 54 - 262 *g* for impacts onto the carpet conditions, from 90 – 170 *g* onto the Vinyl floor, and from 27 – 157 *g* on the NCFs (Figure 2-4). ANOVA indicated a significant interaction between floor condition and impact velocity ($F_{17,58}$ = 137.6, p<0.001). At 1.5 m/s, there was a significant effect of floor ($F_{9,20}$ = 92.6, p<0.001). Dunnett's post-hoc demonstrated that, compared to *CC*, peak accelerations were lower for all of the NCF conditions (p always ≤ 0.002). Similar trends were found for impacts at 2.5 m/s ($F_{9,20} = 431.4$, $p<0.001$) and 3.5 m/s ($F_{8,18} = 558.7$, $p<0.001$), whereby peak accelerations were consistently lower for the NCF conditions compared to *CC* (p always < 0.001). Across all three impact velocities, $g_{max}$ was attenuated by at least 25% and up to 70% for impacts onto NCFs compared to *CC*.

Contrastingly, during impacts at both 1.5 m/s and 2.5 m/s, $g_{max}$ was significantly larger for impacts onto *V* (64% and 39% larger, respectively (p always <0.001)) relative to *CC*. This trend was also observed for impacts onto *RC* at 1.5 m/s and 2.5 m/s (20% larger, p=0.01; and 12% larger, p=0.001, respectively) relative to *CC*. However, at 3.5 m/s, $g_{max}$ was 17% lower for *RC* relative to *CC* (p<0.001). *BC* impacts were not different from *CC* at 1.5 m/s (p=0.975), but yielded reduced $g_{max}$ values at 2.5 m/s (7%, p=0.047) and 3.5 m/s (18%, p<0.001).

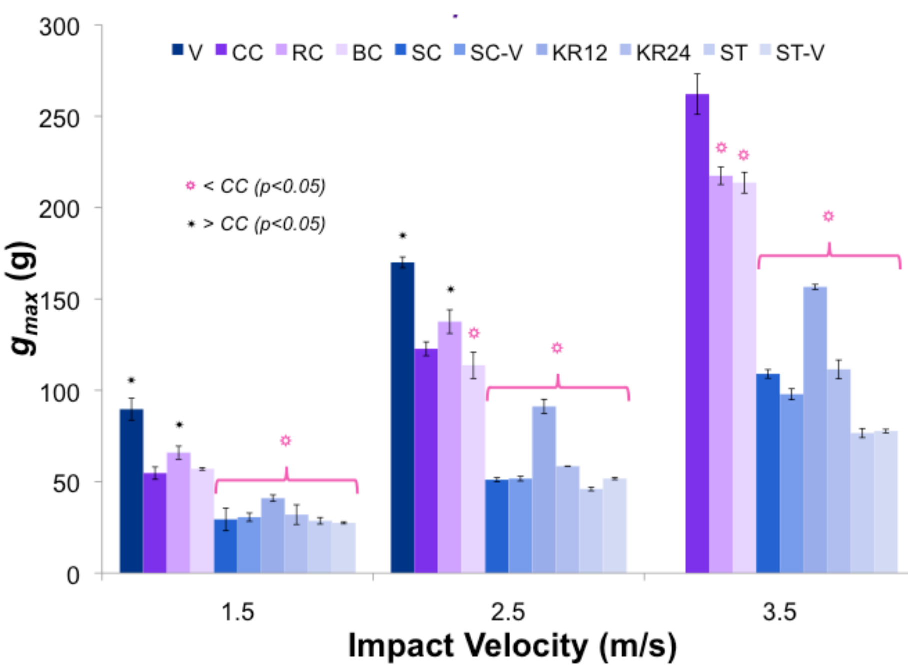


**Figure 2-4: Mean (SD) peak accelerations for impacts to the back of the headform at each impact velocity across all flooring conditions**

#### 2.4.2.2 Head Injury Criterion (HIC)

*HIC* scores ranged from 39 - 1068 for carpeted conditions across all tested impact velocities, between 101 and 496 onto the Vinyl floor (not tested at 3.5 m/s), and from 14 – 482 onto NCF conditions (Figure 2-5). A significant interaction was found between floor condition and impact velocity whereby the attenuation in *HIC* scores provided by NCFs increased as impact velocity increased ($F_{17,58} = 268.3$, $p<0.001$). Subsequent one-way ANOVAs indicated that floor condition was associated with *HIC* at each impact velocity ($F_{9,20} = 236.5$, 640.2 at 1.5 and 2.5 m/s, respectively, and $F_{8,18} = 356.5$ at 3.5 m/s, p always <0.001). Dunnett's post-hoc revealed that *HIC* scores were consistently lower for impacts onto NCFs relative to *CC*. NCFs reduced *HIC* scores by 33-63% at 1.5 m/s, by 41-76% at 2.5 m/s, and by 55-85% for impacts at 3.5 m/s (p always <0.001).

Similarly to the results for $g_{max}$, the *HIC* scores for impacts onto *V* were 159% larger than those onto *CC* at 1.5 m/s (p<0.001), and 92% larger at 2.5 m/s (p<0.001). *HIC* was significantly larger for impacts onto *RC* compared to *CC* at 1.5 m/s (p=0.001) and 2.5 m/s (p<0.001), but were reduced at 3.5 m/s (p=0.008). Compared to *CC*, *HIC* was not different for impacts onto *BC* at 1.5 m/s (p=0.92), but was significantly lower at both 2.5 m/s and 3.5 m/s (p<0.001).

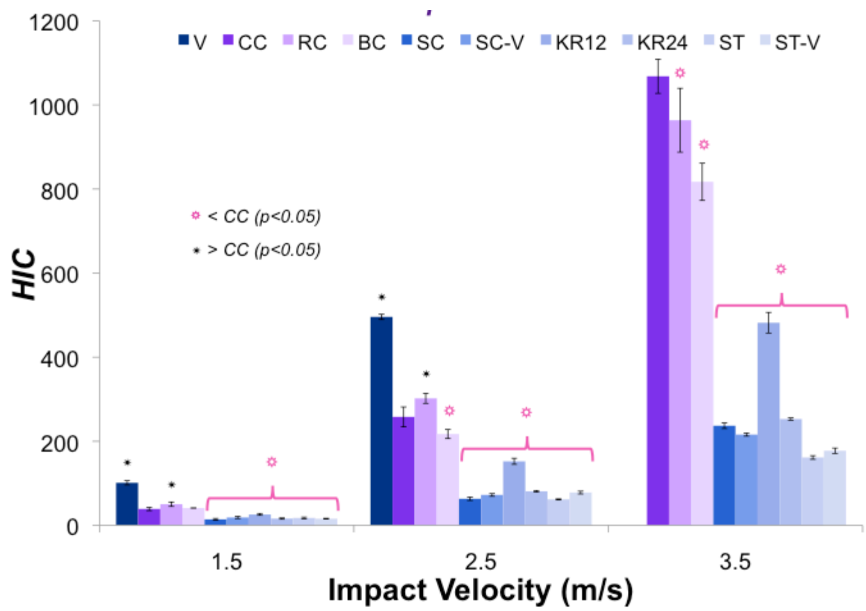


**Figure 2-5: Mean (SD) Head Injury Criterion (HIC) scores for impacts onto the back of the headform at each impact velocity across all flooring conditions**

#### 2.4.2.3 Peak Force ($F_{max}$)

Peak impact force across impact velocities ranged from 3045 - 11583 N for impacts onto the carpet conditions, from 4676 – 8721 N for the Vinyl condition, and between 1487 – 8552 N onto the NCFs (Figure 2-6). A significant interaction between floor condition and impact velocity ($F_{17,58}$ = 395.7, p<0.001) was observed. During impacts at 1.5, 2.5, and 3.5 m/s, floor was significantly associated with $F_{max}$ ($F_{9,20}$ = 1085, 1252, and $F_{8,18}$ = 1522 respectively, p always <0.001). Post-hoc analysis provided that, compared to *CC*, $F_{max}$ was always significantly lower for impacts onto NCFs (p always <0.001). At 1.5 m/s, peak force attenuation provided by the NCFs ranged from 27-52%, similar to that at 2.5 m/s (29-59%) and 3.5 m/s (26-64%).

*CC* provided some force attenuation relative to *V*; $F_{max}$ values were 50% larger for impacts onto *V* at 1.5 m/s, and 31% larger for impacts at 2.5 m/s. $F_{max}$ was 8% larger for impacts onto *RC* at 1.5 m/s (p<0.001), 4% larger at 2.5 m/s (p=0.023), but not significantly different at 3.5 m/s (p=0.251). Compared to *CC*, impacts onto *BC* produced peak forces that were not significantly different at 1.5 m/s (p=0.493); $F_{max}$ was, however, 12% lower at 2.5 m/s and 7% lower at 3.5 m/s (p<0.001).

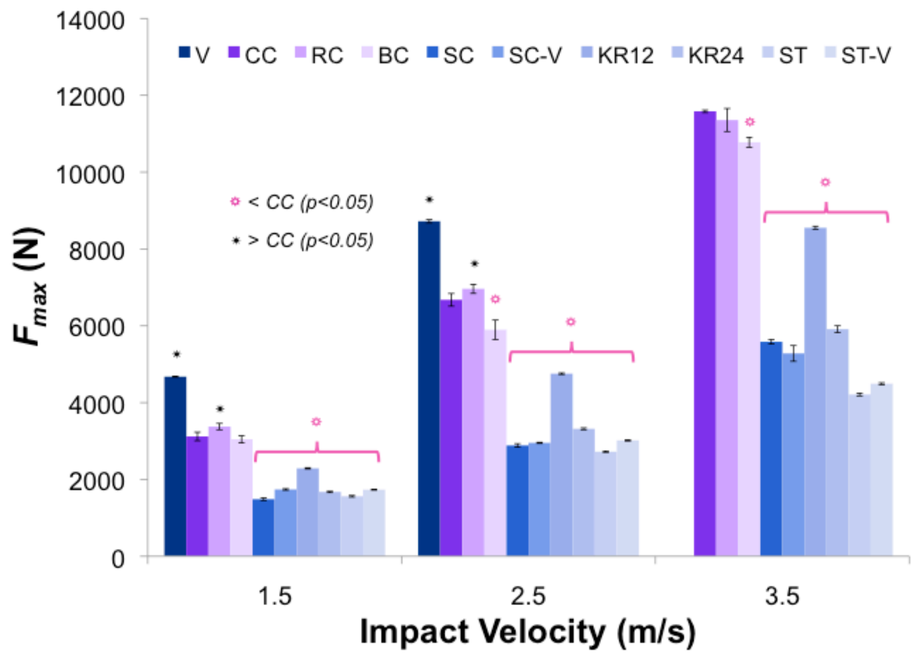


**Figure 2-6: Mean (SD) peak forces applied following impacts to the back of the headform at each impact velocity across all flooring conditions**

## 2.5 Discussion

In the current study, the influence of surrogate headform orientation on indices of skull fracture and TBI risk was examined and it was found that impacts onto the back of the headform represented the 'worst-case' orientation based on resultant acceleration and force profiles. The influence of flooring type on head impact dynamics during these 'worst-case' impact scenarios was then assessed. The hypothesis that the headform would experience lower forces and accelerations during impacts onto novel compliant floors (NCFs) than onto the Commercial Carpet was supported in 54 of 54 possible comparisons (6 floors * 3 impact velocities * 3 variables ($F_{max}$, $g_{max}$, $HIC$)). Regarding the second hypothesis, impacts onto Commercial Carpet yielded significantly lower values for all outcome variables compared to Vinyl in six of six possible comparisons (2 impact velocities * 3 variables). Although not compared statistically, it can be inferred that the outcomes for the NCFs would also be significantly reduced compared to Vinyl based on their relationship to the Commercial Carpet. Interestingly, an interaction effect between floor condition and impact velocity was observed for all three outcome parameters. This interaction was generally characterized by increased attenuation in outcomes in the NCF conditions as impact velocity increased, suggesting that the protective capacity of these floors may be greater as impact severity increases. Overall, these results indicate that the NCFs tested in this study are capable of substantially reducing indices of skull fracture and TBI risk compared to traditional flooring materials during simulated falls involving head impacts.

The most likely explanation for the observation of the backwards headform orientation presenting as a worst-case impact scenario may relate to the construction of the NOCSAE headform itself. The headform is comprised of a high durometer urethane skull covered with a lower durometer urethane that forms the skin and anatomical features of the head (such as the nose, ears and lips). A mass-spring model of the headform-floor system suggests that impact orientations with the lowest effective stiffness (which are likely for headform aspects with the thickest low durometer urethane elements) will result in lower peak forces and accelerations during impact events. For a NOCSAE headform, the thickness of the urethane ear is relatively low in comparison to the nose, while there is only a thin layer of urethane covering the occipital region. These characteristics correspond to the observation of increasing impact severity across front, side, and backwards headform orientations for this test system, respectively.

The definition of the back of the headform as a 'worst-case' impact orientation is specific to the test system used in the current study, and is not intended to contribute to the discussion regarding the effect of impact location / direction on head injury risk during real-world falls involving head impact. Early studies suggested that real-world impacts to the lateral aspect of the human head are most likely to lead to concussion (Hodgson et al., 1983), which corresponds to finite-element models demonstrating a lower tolerance for lateral impacts compared to anterior-posterior or axial impacts (Zhang et al., 2001; Kleiven, 2003; Zhang et al., 2004). In an analysis of head impacts experienced by collegiate football players, 46% of concussive impacts occurred to the top of the head, whereas frontal, lateral, and occipital impacts were responsible for 31%, 15%, and 8% of observed concussions, respectively. However, no significant correlation was found between impact location and clinical outcome severity (Guskiewicz et al., 2007). The finding in the current study that impact severity was substantially affected by headform orientation during impact suggests that this factor should be considered and reported in future research involving simulated impacts with surrogate headforms. Furthermore, although the headform used in this study was chosen based on its high level of biofidelity including a glycerin-filled cavity to simulate brain movement (Higgins et al., 2007), further development of surrogate headforms that aim to mimic the orientation-sensitive response of the human head might be warranted.

It is worthwhile to consider the observed *HIC* scores in context with proposed injury thresholds. Based on animal and cadaveric data, the expanded Prasad-Mertz curves suggest that a *HIC* score of 1000 is associated with a non-zero risk of a fatal head injury, an 18%

probability of severe head injury, a 55% probability of a serious injury, and a 90% probability of moderate head injury for the average adult (Prasad and Mertz, 1985; Mackay, 2007). For impacts at 3.5 m/s onto Commercial Carpet, *HIC* scores exceeded this threshold (mean (SD) = 1068.0 (40.6)). Furthermore, the *HIC* for the Vinyl floor condition was 495.9 (6.2) at an impact velocity of only 2.5 m/s. In contrast, the largest mean *HIC* value from 3.5 m/s impacts onto any of the NCFs was less than 500 ($KR_{12}$ : 482.0 (24.5)), with all other NCF surfaces yielding average *HIC* scores of less than 300. In clinical terms, these results suggest that the risk of moderate head injury for an average adult is 5-25% for a fall involving 3.5 m/s head impact onto the novel compliant floors tested here, compared to an 80-90% risk onto the traditional compliant floors. For the NCFs, outcomes corresponded with floor thickness. For example, at 3.5 m/s the *HIC* was 161.5 (4.3) for the 50 mm *ST* product (likely the least appropriate for indoor implementation), compared to 237.0 (6.6) for the 25 mm *SC*, and 482.0 (24.5) for the 12 mm $KR_{12}$ floor. Additional studies should be considered which investigate the design features that are most predictive of biomechanical effectiveness during head impact, in addition to practical factors including usability, durability, and ease of implementation.

The current results are in accordance with previous reports of the force attenuative properties of specific novel and traditional compliant flooring systems. Maki and Fernie (1990) used a mechanical fall simulator to determine peak deceleration and peak force during simulated hip impacts onto traditional flooring surfaces (although they did not specify the impact velocity achieved). They report that, in comparison to impacts onto a vinyl floor similar to that used in the current study, padded carpets provided the greatest level of impact attenuation (up to 23%). Others have reported force attenuative values as high as 56% and

73% when incorporating PVC underlay beneath vinyl and carpet floors, respectively (Nabhani and Bamford, 2002; Minns et al., 2004; Nabhani and Bamford, 2004); however, these values may overestimate the protective capacity of these flooring conditions as the effective compliance of the pelvic region was not incorporated into their testing system. Most recently, Laing and Robinovitch (2009) reported that the same SmartCell floor tested in this study attenuated peak femoral impact force by 17.3% compared to a commercial-grade vinyl for impacts at 2 m/s, and 22.5% at 3 m/s, while a 100 mm thick SofTile product provided peak force attenuation of 44.9% and 45.5% at 2 and 3 m/s, respectively. In the current study, Commercial Carpet lowered peak forces by 30% compared to the Vinyl floor for impacts at 2.5 m/s. Impacts onto the SmartCell floor produced peak forces that were at least 51% lower than the Commercial Carpet (thus, 80% lower than Vinyl). SofTile reduced peak forces by at least 59% compared to Commercial Carpet (~90% lower than peak forces onto Vinyl). These values are higher than those reported for simulated falls on the hip as the surrogate headform is likely much stiffer than the pelvis' effective stiffness of approximately 40 kN/m (Laing and Robinovitch, 2010; Robinovitch et al., 2009). These data, in conjunction with additional studies that have assessed the influence of floor stiffness during falls on the upper limb (Robinovitch and Chiu, 1998) and buttocks (Sran and Robinovitch, 2008), suggest that compliant floors have the potential to decrease the risk of a wide range of fall-related injuries.

For novel compliant floors to be an effective intervention strategy in reducing fall-related injuries, they must have the capacity to decrease impact loads and accelerations while having minimal concomitant influences on the balance and mobility of the target users. Numerous reports have established that some compliant surfaces may decrease postural stability and consequently increase the likelihood of falling. Compared to rigid surfaces, compliant foam surfaces have been associated with increased postural sway during quiet stance (Lord et al., 1991; Redfern et al., 1997; Lord and Menz, 2000; Gill et al., 2001), as well as a lowered trajectory of the whole-body centre-of-mass (COM), reduced toe clearance, and increased step length, step width, and step width variability during gait (Marigold and Patla, 2005; MacLellan and Patla, 2006). Regarding traditional compliant flooring systems tested in this study, thick carpet has been shown to increase anterior-posterior sway for older adults when visual fields are altered (Redfern et al., 1997), although these effects are not observed under normal vision conditions (Dickinson et al., 2001; Dickinson et al., 2002). Regarding novel compliant flooring systems, Laing and Robinovitch (2009) found that medial-lateral postural sway on a SmartCell floor was not different than on a rigid surface for community-dwelling elderly women, and that scores on the Timed Up and Go test (a predictor of fall risk (Podsiadlo and Richardson, 1991; Lundin-Olsson et al., 1998; Chiu et al., 2003)) were not different for SmartCell, SofTile and a rigid floor condition. Furthermore, Wright and Laing (2011) found that the displacement profiles of both the centre of mass (a balance indicator) and the underfoot centre-of-pressure (a balance control variable) were not affected by SmartCell and SofTile floors in community-dwelling elderly women during backwards perturbations. Despite these encouraging findings, further research is needed to ascertain if and how balance control is affected on these floors during activities of daily living for older adults residing in settings with high rates of falls and associated injuries (retirement homes, nursing homes, hospitals) where NCFs are most likely to be installed.

There were several limitations associated with this study, the majority of which are specific to the test apparatus. First, while little conclusive information is available with respect to the characteristics of 'typical' falls and subsequent head impacts experienced by older adults (Klenk et al., 2011), it is unlikely that all injurious real-world falls involving head impact are characterized by the purely vertical cranial trajectory that this test system simulated. Although the relative importance of linear versus rotational accelerations in TBI pathology is still a matter of debate, rotational accelerations are nonetheless hypothesized to be a primary risk factor (Ommaya and Gennarelli, 1974; Guskiewicz and Mihalik, 2011), and such rotational effects were minimized in the current simulated impacts. However, the test method used was similar to those used for national standards on assessing the protective capacity of playground surfaces (ASTM, 2004) and sports helmets (NOCSAE, 2009), allowing for comparisons of the protective capacity offered by these differing intervention strategies. Second, the Head Injury Criterion outcome that is reported is reliant on measurements of external linear kinematics of the head, and is not specific to direction of impact, nor is it able to reflect the response of the brain within the cranial cavity (Hardy et al., 2007; Marjoux et al., 2008). However, the complex etiology of head injuries makes it immensely difficult to establish accurate injury criteria and associated injury thresholds, so despite its limitations, the *HIC* represents the best and most widely-used risk prediction tool currently available (Cory et al., 2001; Guskiewicz and Mihalik, 2011). Lastly, the impact velocities used in this study may not reflect those experienced during 'typical' head impacts, which may be greater than 3.5 m/s. Pilot testing at higher impact velocities caused damage to the mechanical test system, including a rupture of one headform's glycerin-filled brain cavity. A compromise between headform biofidelity and external validity could be achieved through the use of a more durable headform (e.g. Hybrid III) to assess the protective capacity offered by flooring surfaces at higher impact velocities. However, future research is needed to characterize the inputs that should be incorporated into a biofidelic test method for simulating fall-related head impacts in older adults (e.g. distributions of head orientation, impact velocities, and load trajectories), in addition to the risk of injury across these loading scenarios, similar to research being conducted for sports-related head impacts (Zhang et al., 2004; Guskiewicz et al., 2007; Greenwald et al., 2008; Marjoux et al., 2008; Duma and Rowson, 2011; Guskiewicz and Mihalik, 2011).

There are additional biomechanical issues that need to be studied to fully characterize the potential protective capacity of novel compliant floors during head impacts. For example, additional studies should investigate the potential influence of surface compliance on the rotational accelerations experienced within the brain cavity during oblique head impacts. Furthermore, the deformation of compliant floors around the skull during obliquely oriented head impacts might increase the system coefficient of friction leading to a slowing of the head's horizontal velocity and a concomitant increase in neck bending loads due to the inertia of the body. Future studies should characterize such factors towards the goal of developing flooring systems that optimize potentially competing demands for protection across a range of injury types. Nonetheless, the results of the current study are encouraging, indicating that the novel compliant floors tested can substantially reduce the magnitudes of widely used indices of skull fracture and TBI risk compared to traditional flooring products.

In order to limit the expected increase in the incidence of fall-related TBI (and other fall-related injuries) in seniors over the coming decades, it is imperative that effective intervention strategies be designed and implemented. Novel compliant flooring systems appear to be a promising approach, capable of providing substantial protective capacity against head injury and other fall-related injuries without introducing impairments to balance and mobility (Laing and Robinovitch, 2009; Wright and Laing, 2011). The added benefit of being a passive intervention approach precludes the need for active user compliance and adherence to ensure effectiveness, unlike intervention strategies such as exercise, pharmacological agents, and wearable hip protectors. The results of this study further support the the introduction of pilot installations to aid in the development of clinical trials to test the effectiveness of NCFs in environments with high rates of falls and injuries such as hospitals, seniors' centres, and residential-care facilities.

# CHAPTER 3 EFFECT OF NOVEL COMPLIANT FLOORS ON EARLY COMPENSATORY BALANCE REACTIONS FOLLOWING EXTERNAL PERTURBATION IN RETIREMENT HOME-DWELLING OLDER ADULTS

## 3.1 Background

Falls are a major cause of injuries in adults over the age of 65, and are responsible for up to 90% of both hip fractures and traumatic brain injuries (Grisso et al., 1991; Pickett et al., 2001). One in three community-dwelling older adults will experience at least one fall per year, while 50% of this cohort will suffer multiple falls (SMARTRISK, 2009). This rate is even higher for older adults residing in environments such as nursing homes, hospitals, and residential-care facilities (Cameron et al., 2010). Furthermore, one-third of all falls in residential-care settings result in injury (Nurmi and Luthje, 2002). Fall-related injury risk increases exponentially with age, clearly illustrated by the fact that seniors over the age of 85 are 70% more likely to suffer an activity-limiting injury than persons aged 65-74 (Health Canada, 2002). As the proportion of Canadian seniors is expected to near 25% by the year 2041, effective intervention strategies are desperately needed to minimize the anticipated social and economic difficulties associated with the ageing Canadian population.

Novel compliant flooring systems (NCFs) have demonstrated a substantial protective capacity against hip fractures, by providing peak reductions in the impact forces applied to the proximal femur of 25-50% during simulated sideways falls compared to a vinyl surface (Laing and Robinovitch, 2009). These findings are complemented by the results from Study 1 of this thesis (Chapter 2), which indicated that NCFs are capable of reducing the impact forces and accelerations applied to the back surrogate headform by up to 80% compared to a commercial-grade carpet with underpadding, and provide even greater reductions relative to vinyl surfacing. This intervention is particularly relevant for environments such as nursing homes, hospitals, and residential care facilities where large numbers of falls and associated injuries occur. However, it is essential that these low-stiffness floors are capable of providing sufficient force attenuation without effectively increasing the risk for falling by impairing the balance and mobility of target users.

The control of postural stability arises from a complex interaction of various sensorimotor processes (Horak, 2006). In order to maintain an upright posture during quiet upright stance, from a biomechanical perspective, the vertical projection of the COM must fall within the limits of the BOS (Winter, 2009). Adjustments in the location of the underfoot COP are used to guide or 'shepherd' the trajectory of the COM towards equilibrium (Winter et al., 1998; Morasso and Schieppati, 1999; Winter et al., 2001; Morasso and Sanguineti, 2002; Horak, 2006). In the event of a balance perturbation that causes the COM to shift anteriorly (e.g. being nudged from behind), balance recovery necessitates a rapid anterior shift of the COP to decelerate the COM before it crosses the anterior BOS boundary. If there is an initial delay or slowing of COP displacement, a foot-in-place approach may be insufficient to decelerate the COM, requiring the individual to take a step in order to increase the BOS to prevent a forward fall (Maki et al., 2001; Winter, 2009).

It has frequently been proposed that the initial compensatory reaction to balance perturbation during upright stance occurs "automatically" (Allum, 1983; Nashner, 1976; Nashner and Cordo, 1981; McIlroy and Maki, 1993; Norrie et al., 2002; Mochizuki et al., 2010), which is then followed by a secondary stabilizing response. Such autonomic postural responses (APRs) have been shown to be modified by the magnitude of the balance perturbation, and are exhibited whether or not a stepping response is used to recover balance (McIlroy and Maki, 1993; Maki and McIlroy, 2007). The evoked APR is thought to require minimal attentional effort (Brown et al., 1999; Rankin et al., 2000; Maki et al., 2001; Norrie et al., 2002), and is not affected if inputs from cutaneous mechaoreceptors on the plantar surface of the foot are reduced (Perry et al., 2000).

Low stiffness surfaces have the potential to affect balance and balance control responses through degraded proprioceptive and pressure sensitivity from receptors on the plantar surface of the foot (Lord and Menz, 2000; Betker et al., 2005). For example, postural sway during quiet stance on compliant foam surfaces has been shown to increase substantially on compliant foam surfaces compared to rigid surfaces (Ring et al., 1989; Lord et al., 1991; Teasdale et al., 1991; Lord and Menz, 2000; Gill et al., 2001). It has been demonstrated that walking on extremely compliant foam surfaces affects gait mechanics, by which the COM trajectory is lowered through increased step length, step width, and forward pitching of the trunk, which might suggest decreased trunk stability (Marigold and Patla, 2005; MacLellan and Patla, 2006). Following rearward translation of the support surface, compliant foam surfaces have been shown to affect measurements associated with APRs, namely the magnitudes and rates of displacement of the COM and COP, possibly as a result of decreased effective stiffness at the ankles while standing on the compliant foam (Wright and Laing, 2011). Thus, it is apparent that compliant support surfaces have the potential to impair aspects of balance control and postural stability.

Nevertheless, NCFs appear to provide minimal affect to balance maintenance and stability for older community-dwelling women (Laing and Robinovitch, 2009; Wright and Laing, 2011). This is supported by the results of postural sway during quiet stance and the Timed Up and Go (TUG) test, both of which have been shown to associate with fall risk (Campbell et al., 1989; Maki et al., 1994; Lundin-Olsson et al., 1998; Chiu et al., 2003). Specifically, compared to a rigid control surface, the root mean square amplitude and velocity of postural sway were not significantly different in the medio-lateral direction for one NCF, and the times required to complete the TUG were not different for two NCFs (Laing and Robinovitch, 2009). Furthermore, two NCFs were shown to have minimal effect on aspects of APRs, as the margin of safety and initial rates of displacement of the COM and COP were not changed compared to a control vinyl surface (Wright and Laing, 2011).

However, as outlined in detail in Section 1.4.2, there are limitations associated with the studies that have assessed the effects of NCFs on balance control that must be addressed (Laing and Robinovitch, 2009; Wright and Laing, 2011). Specifically, the participants from these studies were recruited from a community-dwelling population of elderly women, despite the fact that NCFs are most likely to be installed in environments with higher rates of falls and associated injuries (e.g. residential care facilities) where the population of older adults are increasingly frail. Additionally, no studies to date have investigated the effect of NCFs on APRs compared to traditional compliant flooring surfaces including carpets.

## 3.2 Purpose and Hypotheses

The purpose of this study was to determine the influence of four flooring surfaces (one traditional, three novel compliant flooring systems) on variables characterizing autonomic postural responses relative to a control surface (commercial-grade carpet with underpadding). It was hypothesized that no differences would be found across floors for: i) time to onset of an autonomic postural response, defined by initial movement of the COP ($APR_{onset}$); ii) the minimum margins of safety of the COP (*MMOS*); iii) the time to *MMOS* ($t_{MMOS}$); iv) the peak velocity of COP movement ($v_{max}$); and v) time to peak COP velocity ($t_{vmax}$). Based on these expected results, it was also hypothesized that no difference would be found in: vi) the minimum margin of safety for the whole-body COM ($MMOS_{COM}$).

## 3.3 Methods

### 3.3.1 Participants

15 healthy older adults (13 female, 2 male) recruited from a local retirement home facility participated in this study, with a mean (SD) age of 83.9 (3.1) years (range: 79 – 89), mean body mass of 70.7 (8.1) kg (range: 57.5 – 85 kg), mean height of 159.7 (5.2) cm (range: 150.5 – 168.5 cm), and mean body mass index of 27.8 (3.3) kg/m$^2$ (range: 20.4 – 33.6 kg/m$^2$). Exclusion criteria included: a) a history of falls within the past 6 months; b) a demonstrated willingness and ability to successfully stand for 60 seconds without any external aid; c) stand from a seated position without using chair armrests; and d) successful balance maintenance on at least two out of a possible three trials using a spinal nudge test. This study was approved by the Office of Research Ethics at the University of Waterloo. All participants provided written informed consent prior to participation.

### 3.3.2 Flooring Conditions

Five separate flooring conditions were tested during this study, all of which were tested in the second chapter of this thesis (Figure 3-1). Floor stiffnesses were estimated from the slope of force-deflection tests under a simulated 816 N footfall using a rigid foot-shaped indenter mounted within a servohydraulic materials testing system (i.e. stiffness = (816 N – 0 N)/observed deflection at 816 N). Similarly to Chapter 2, the control condition entailed a level loop 'Commercial Carpet' (*CC*) (The Carpet Store (Waterloo, ON), pile height = 6 mm, face weight = 882 g/m$^2$, stiffness = 220 kN/m) with 6 mm underpad (a traditional compliant flooring system often found in commercial housing settings). A second traditional surface was also tested, using a commercial-grade 'Vinyl' (*V*) comprised of a 2 mm thick layer of rubber appropriate for installation over concrete or wooden subfloors in institutional settings (Noraplan Classic, Nora Systems Inc, Lawrence, MA, USA). Three NCF conditions were also tested. The 'SmartCell' condition (SATech, Chehalis, WA, USA, stiffness = 583 kN/m) was a 25 mm tall synthetic, 50-durometer rubber flooring system comprising a continuous surface layer overlying a series of cylindrical rubber columns 14 mm in diameter and 19 mm apart (*SC-V*). The 'SofTile' (*ST-V*) (SofSurfaces, Petrolia, ON, Canada, stiffness = 429 kN/m) condition used a similar design, with 50 mm diameter columns spaced at 70 mm intervals; the 50 mm thick model was tested. Both *SC-V* and *ST-V* conditions included a vinyl overlay in order to represent a design scenario likely necessary for clinical settings. The final floor condition entailed a 12 mm thick tile from Kradal (Acma Industries Ltd., Upper Hutt, Wellington, New Zealand, stiffness = 680 kN/m) with a relatively stiff top surface over a closed cell polyurethane base layer. All floor conditions had substantially higher stiffnesses than those floors that have previously been reported to have significant influences on early compensatory balance responses (~11 kN/m; Wright and Laing, 2011). During testing, the order of floor condition was randomly determined and subsequently tested as a block.

### 3.3.3 Experimental Protocol

18 infrared-emitting markers were placed over the left and right acromion, lateral epicondyle of the humerus, styloid process of the radius, greater trochanter, lateral condyle of the femur, calcaneus, lateral malleolus, head of the fifth metatarsal, and distal phalange of hallux. These markers were tracked using a 12-sensor motion capture system (collected at 201 Hz; Optotrak Certus, Northern Digital Inc., Waterloo, ON).

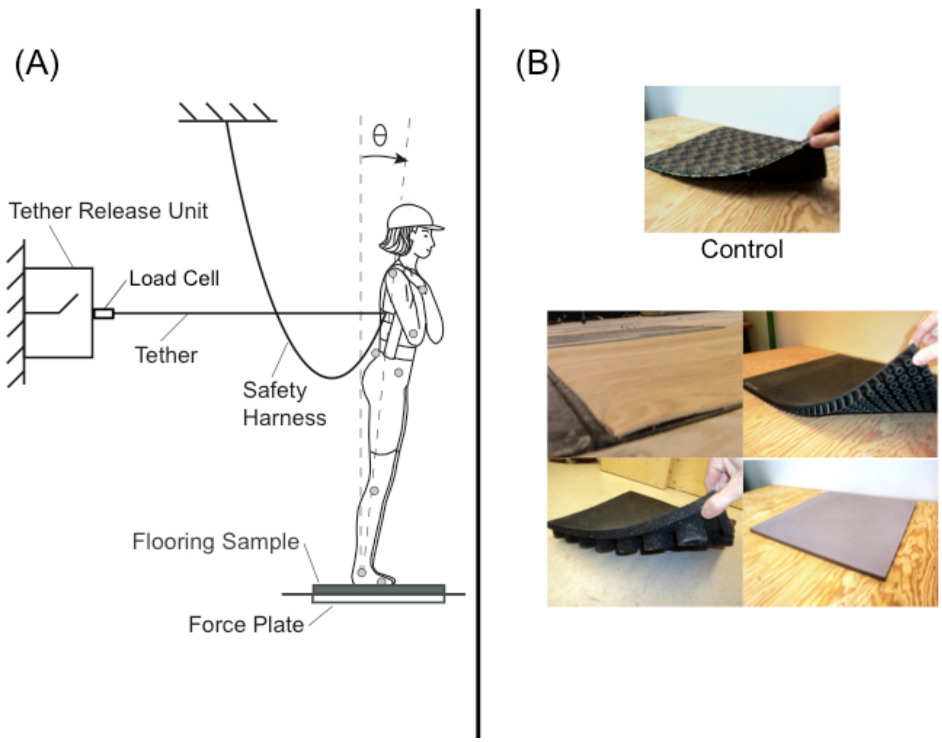


**Figure 3-1: A) Schematic of the experimental setup to used for the tether-release experiment; B) Pictures of the floor conditions tested: Top – Commercial Carpet (*CC* - control); additional floors (clockwise from top left): Vinyl (*V*), SmartCell (*SC-V*), 12 mm Kradal (*KR*), and SofTile (*ST-V*); note that SmartCell and Vinyl were tested with vinyl overlay.**

The participants initially stood barefoot, arms on the chest, on a flooring sample mounted over a force plate (collected at 2010 Hz; Model OR6-7, Advanced Mechanical Technology Incorporated, Watertown, MA, USA), with a foot width (second toe-to-second toe) equal to the breadth of the anterior superior iliac spines. The participants were instructed to lean forward at the ankle, inclining them to a stationary forward leaning position where they were held by horizontal tether attached to a chest harness worn by the participants at one end (at approximately the level of the 9$^{th}$ thoracic vertebrae) and to a wall mount at the other (Figure 3-1). A load cell (collected at 2010 Hz; MLP-100, Transducer Techniques, Temecula, California, USA) placed in series with the horizontal tether allowed for the quantification of the force exerted on the tether due to the participants' lean angle, and provided the indication for perturbation onset time. For the purposes of safety, the chest harness was also attached to an overhead support with another tether; the second tether did not impair participants' movement. The participants were provided with real-time visual feedback of their pre-perturbation COP location using a custom program created in the Lab-VIEW environment (v8.5; National Instruments, Austin, TX, USA). They were instructed to maintain their initial COP location within 4-6 cm anterior to the ankles (chosen to match average COP location relative to the ankles during quiet stance for a similar population). In order to control for changes in pre-perturbation effective ankle stiffness across floor conditions, auditory biofeedback (Myotrac, Thought Technology Ltd., Montreal, Canada) was provided to ensure that the participants maintained an inclined position without co-contraction of the muscles spanning the ankle joint. Electrodes were placed unilaterally on the right medial gastrocnemius (over the most prominent bulge of the muscle belly) and right tibialis anterior (1/3$^{rd}$ of the distance from the head of the fibula to the medial malleolus). Thresholds for auditory feedback were set to match the peak level of activity exhibited during a 15 s quiet stance trial.

After providing a “ready” cue, the horizontal tether was released following a random time delay of between 1 and 5 seconds. The participant was required to maintain his or her balance using feet-in-place responses only (i.e. without taking a step), simulating the type of balance disruption that may be presented if nudged from behind, or standing on a bus that decelerates quickly, by which the COM is pitched anteriorly towards the toes. Hip flexion and arm movements were allowed. Initial lean angle was monitored in real-time, defined as the angle between the vertical and a vector connecting the lateral malleolus to the acromion, and was set to 3 degrees of incline relative to upright stance. This was chosen based on previous tether-release studies that have found community-dwelling older women were able to recover from an average maximum initial lean angle of 4.6 degrees (Mackey and Robinovitch, 2005). Thus, the paradigm was designed to present a challenging, yet sub-maximal perturbation. Due to the sensitivity of lean angle readings to subtle movements by the participant (especially at the shoulders), lean angle was also monitored based on the force exerted by the participant on the tether recorded by the tether load cell (Mochizuki et al., 2010). After completing three practice trials in the absence of a flooring sample, each participant performed five successive trials on each of the five flooring conditions. There were approximately 30 s between each perturbation, while two-minute breaks were provided every five trials, or at the request of the participant, so as to minimize any potential influence of fatigue.

### 3.3.4 Data Analysis

Successful trials were defined as a recovery of balance without taking a step or making contact with the investigator who was spotting the participant. All analysis of data was completed in the Matlab environment (version R2007b, Mathworks, Natick, MA, USA).

Trials were excluded if any of the kinematic markers were not tracked for more than 200 ms (Howarth and Callaghan, 2010). Data points where the marker was obstructed were interpolated using a cubic spline routine. Based on the results of residual analyses, dual-pass, $4^{th}$ order digital Butterworth filters were used to filter kinematic data (3 Hz cutoff frequency), force plate data (5 Hz cutoff frequency), and tether load cell data (3 Hz cutoff frequency). It was not anticipated that filtering kinematic and kinetic data at different cutoff frequencies would induce significant artifacts; while a previous report has suggested that using different cutoff frequencies can induce artifacts in the relation between calculated peak forces and moments at the knee during the impact phase of a jump (Bisseling and Hof, 2006), the movements in the current study were of considerably lower frequency. For each successful trial, kinematic data was used to construct a transverse planar 11-segment rigid link model (Winter, 2009), allowing for the calculation of the location of whole-body centre of mass (COM) over the course of each trial (Figures 3-2, 3-3). Centre of pressure (COP) trajectories were determined from the force plate data (Figures 3-2, 3-3). Perturbation onset was determined from the sharp drop in tether load (Figure 3-2). Initial lean angle was verified, defined as the angle at perturbation onset between the vertical and a vector connecting the midpoint of the lateral malleoli to the midpoint of the acromia, relative to the same angle during upright stance. The boundaries of the base of support (BOS) were defined by markers on the toe, heel, and head of the fifth metatarsal (Figure 3-3). The time between perturbation onset and the initiation of an autonomic postural response ($APR_{onset}$)

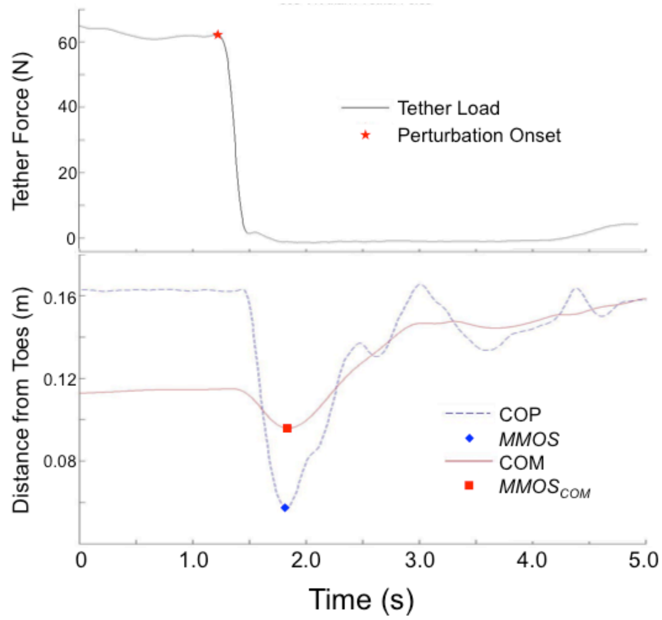


**Figure 3-2: Representative force profile recorded from the tether load cell over the entire course of one trial on the commercial carpet, indicating the moment of perturbation onset (upper); Representative COP and COM displacement profiles over the entire course of one trial, plotted as distance from the toes (lower)**

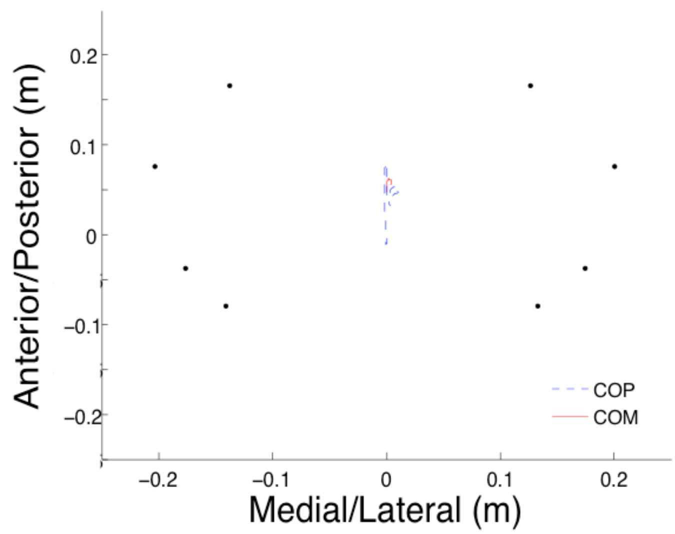


**Figure 3-3: Representative anterior/posterior versus medial/lateral displacement profiles for the COP and COM following perturbation onset for one trial on the commercial carpet (with respect to the centre of the force plate); black dots indicate location of (from top to bottom) markers on the hallux, fifth metatarsal, lateral malleolus, and heel of the left and right feet**

was determined from the COP profile at the point where a sharp, sustained decrease in slope occurred (Figure 3-4). The minimum margin of safety (*MMOS*) of the COP was defined at the point of maximum anterior excursion of the COP, when the distance from the BOS boundary was at a minimum (i.e. the smallest distance between the COP and the toes during the balance recovery response; Figure 3-4). The time elapsed between perturbation onset and the point at which *MMOS* was achieved was defined ($t_{MMOS}$). COP velocities were determined from the displacement profiles using three-point central difference differentiation. Peak velocity of the COP ($v_{max}$) during the autonomic postural response and time to peak velocity ($t_{vmax}$) were determined (Figure 3-4). Finally, the MMOS was also calculated for the whole-body COM ($MMOS_{COM}$) (Figure 3-4). The outcome parameters from each trial were analyzed, and three trials in total were removed from subsequent analysis as they represented obvious outliers that did not follow the consistent trends in the data (for example, one trial was excluded as the calculated $APR_{onset}$ time was roughly three times longer than typical calculated values). For each subject, the values of each outcome parameter from all included trials were averaged for each floor and used for statistical analysis.

### 3.3.5 Statistics

A one-way repeated-measures ANOVA was used to test for the effect of floor condition on each of the six outcome variables. Where the assumption of sphericity was violated, a Hyunh-Feldt correction was applied. If appropriate, post-hoc analyses were conducted using a paired t-test using Bonferroni correction, comparing each of the floors to the control condition represented by the commercial carpet. All statistical analyses were carried out with an experiment-wide significance level of 0.05 using statistical analysis software (SPSS Version 19.0, SPSS Inc., Chicago, IL, USA).

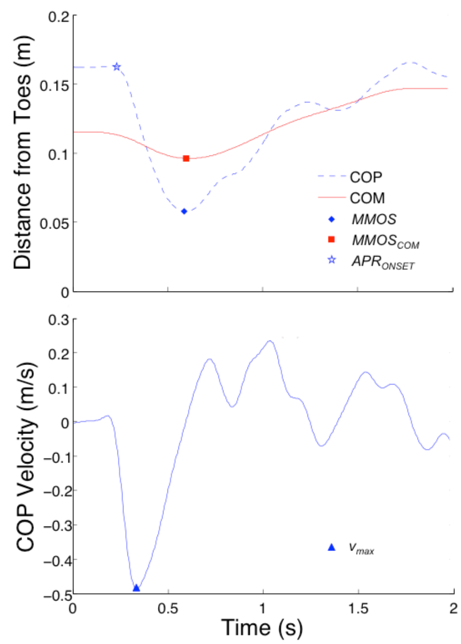


**Figure 3-4: Representative COP and COM displacement profiles for 2 seconds following perturbation onset for one trial on the commercial carpet floor, plotted as distance from the toes (upper); corresponding COP velocity profile (lower)**

## 3.4 Results

The pre-perturbation lean angles across all participants were not significantly different across floor conditions ($F_{4,56} = 0.207$, $p = 0.933$), with mean (SD) values of 3.8° (1.3°), 3.5° (1.9°), 3.6° (1.7°), 3.7° (1.7°), and 3.6° (1.4°) for the *CC*, *V*, *SC-V*, *KR*, and *ST-V* floor conditions respectively. Expressed as a percentage of body weight, the mean (SD) loads recorded in the tether immediately prior to perturbation onset were 5.6 (1.2)%, 5.6 (1.9)%, 5.3 (1.4)%, 5.3 (1.0)%, and 5.5 (1.5)% for the *CC*, *VN*, *SC-V*, *KR*, and *ST-V* conditions, respectively. On average, participants were able to successfully maintain balance in 82.7%, 80.4%, 85.8%, 80.4%, and 87.1% of trials in the *CC*, *VN*, *SC-V*, *KR*, and *ST* conditions. Table 3-1 summarizes the data across outcome variables and floor conditions. The average COP displacement and COP velocity profiles across all subjects for each floor are shown in Figure 3-5. For the control floor condition, the mean (SD) $APR_{onset}$ time was 227 (19) ms, with a *MMOS* of 5.1 (1.8) cm, and $t_{MMOS}$ of 597 (105) ms. $v_{max}$ was, on average, 0.70 (0.25) m/s, while $t_{vmax}$ was 343 (29) ms. Finally, the mean $MMOS_{COG}$ was 7.6 (2.1) cm.

**Table 3-1: Mean (SD) values for outcome parameter across the commercial carpet (*CC*), vinyl (*VN*), SmartCell (*SC-V*), Kradal (*KR*), and SofTile (*ST-V*) floor conditions.**

| Parameter | Floor | | | | |
|---|---|---|---|---|---|
| | *CC* | *VN* | *SC-V* | *KR* | *ST-V* |
| *MMOS* (cm) | 5.1 (1.8) | 5.2 (2.0) | 5.7 (1.8) | 5.2 (1.5) | 5.5 (1.6) |
| $t_{MMOS}$ (ms) | 597 (105) | 634 (141) | 572 (116) | 575 (111) | 566 (112) |
| $v_{max}$ (m/s) | 0.70 (0.25) | 0.70 (0.20) | 0.69 (0.19) | 0.70 (0.19) | 0.76 (0.23) |
| $t_{vmax}$ (ms) | 343 (29) | 350 (34) | 346 (40) | 347 (27) | 337 (24) |
| $APR_{onset}$ (ms) | 227 (19) | 224 (17) | 227 (17) | 229 (24) | 227 (14) |
| $MMOS_{COM}$ (cm) | 7.6 (2.1) | 8.6 (2.0) | 8.6 (2.1) | 8.9 (2.4) | 8.5 (1.6) |

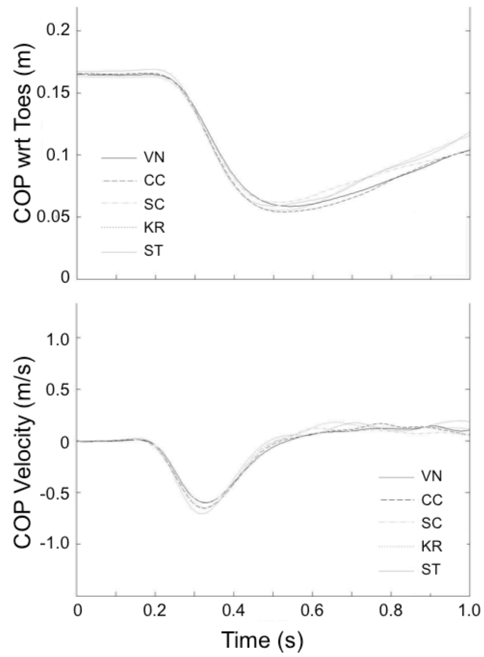


**Figure 3-5: Averaged across all subjects for 1.0 second following perturbation onset, this figure illustrates the average distance of the COP with respect to the toes (upper), and average COP velocity profile (lower) for each floor condition**

The average within subject coefficient of variation (CV) in $APR_{onset}$ for the control condition (commercial carpet) was 0.07 (range: 0.03 – 0.14). This was similar to the average coefficient of variation in $APR_{onset}$ for the SmartCell condition (mean: 0.08; range: 0.03 – 0.18). Furthermore, there was no apparent trend in CV between these two flooring conditions across subjects (i.e. CV was not consistently higher or lower for the SmartCell condition compared to the commercial carpet condition).

ANOVA results indicated that floor condition was not significantly associated with $APR_{onset}$ ($F_{4,56}$ = 0.238, p = 0.842), $MMOS$ ($F_{4,56}$ = 1.169, p = 0.334), $t_{MMOS}$ ($F_{4,56}$ = 1.837, p = 0.161), $v_{max}$ ($F_{4,56}$ = 1.030, p = 0.400), $t_{vmax}$ ($F_{4,56}$ = 0.699, p = 0.596), or $MMOS_{COM}$ ($F_{4,56}$ = 0.644, p = 0.476).

## 3.5 Discussion

In the current study, the influence of flooring surface (5 conditions) on indices of early compensatory balance reactions were examined following a lean and release balance perturbation in elderly adults who reside in a retirement home setting. The results strongly indicate that the novel compliant flooring systems tested in this study cause minimal effects on early balance control characteristics for this type of perturbation. It was hypothesized that the early compensatory balance reactions would not be influenced by flooring condition, as described by five variables: i) the time between the onset of perturbation (release of the tether) and the initiation of an autonomic postural response ($APR_{onset}$; indicated by the initial movement of the COP); ii) the peak rate ($v_{max}$), and; iii) time to peak rate ($t_{vmax}$) of COP displacement; iv) COP minimum margin of safety (*MMOS*), and; v) time to *MMOS* ($t_{MMOS}$). The results support each of these hypothesis elements, with no significant differences being detected in any of the outcome parameters across floor conditions. The second hypothesis was that the minimum margin of safety of the whole-body centre of mass ($MMOS_{COM}$) would be unaffected by flooring surface, which the results also supported.

An immediate question that may arise from the results of the current study is whether or not the measurements used to assess the early compensatory balance reactions are, in fact, sensitive to changes in surface/floor stiffness. In theory, compliant surfaces have the potential to degrade the quality of information from plantar surface mechanoreceptors, limiting the ability to detect the anterior, lateral, and posterior borders of the base of support. Furthermore, local underfoot deformations due to increased pressure of the forefoot following perturbation could serve to reduce rotation of the ankle and consequent feedback from spindle proprioceptors. Underfoot deformations could also decrease the effective stiffness at the ankle and potentially slow the displacement of the COP, requiring greater excursion before it is able to 'catch' the COM. Accordingly, Wright and Laing (2011) demonstrated that compliant foam surfaces affected the rates of COP displacement, ultimately reducing the minimum margins of safety of the COP (from 2.6 to 1.7 cm, a 35% decrease) and COM (from 5.8 to 2.7 cm, a 53% decrease) compared to a vinyl control floor. These results support that the measures used in the current study are indeed influenced by surface compliance.

While Wright and Laing (2011) similarly found that SmartCell and SofTile floors had minimal effects on the magnitudes and rates of COP displacement during the initial balance recovery response, an important difference between the previous study and the current investigation lies in the perturbation paradigm used. The previous study used a platform/support surface translation paradigm, whereas this study used a tether-release strategy. Although the two paradigms are similar in terms of the evoked compensatory balance reaction, by which feet-in-place balance recovery requires a similar pattern of COP displacement, there are important differences. While platform translation paradigms allow the COM and COP to start in approximately the same positions in the sagittal plane, the initial pre-perturbation location of the COM in the current study was already shifted anteriorly relative to the COP. It follows that, upon perturbation onset, the tether-release paradigm likely places a greater emphasis on rapid COP displacement. Indeed, the COP displacement rates found in the current study (~0.70 m/s) were greater than the largest such rates reported by Wright and Laing (2011) following backward platform translation (~0.56 m/s). This might explain the finding that the minimum margins of safety were larger in the current study than in the previous study for both the COP (~0.053 m versus ~0.026 m) and

COM (~0.084 m versus ~0.057 m). The displacement rates reported by Wright and Laing, however, were for specific portions of the COP profile as opposed to peak displacement rates. It is noteworthy that although both the platform translation and tether-release paradigms are commonly used in the assessment of balance control, they induce large, abrupt perturbations. While the exact characteristics of 'typical' falls is not definitely known, evidence exists to suggest that a substantial proportion of falls occur following slow drifts of body weight towards the limits of the base of support (Robinovitch et al., 2009). Future research is thus needed to develop perturbation paradigms that mimic this behaviour. Nevertheless, the current results indicate that the NCFs tested had no effect on indices associated with early compensatory postural reactions for a tether-release paradigm, agreeing with previous reports of minimal such effects for older community-dwelling women (Wright and Laing, 2011), and complementing previous findings of their minimal effect on voluntary postural stability and mobility (Laing and Robinovitch, 2009).

Interestingly, the average time to onset of APR was ~220 ms; generally, reported APR onset times are much shorter than this, with latencies of 80-140 ms (McIlroy and Maki, 1993; Maki and McIlroy, 2007). It should be noted that onset latencies associated with early autonomic postural responses reported in the literature are often determined from electromyographic (EMG) signals (McIlroy and Maki, 1993; Maki et al., 2001). For example, from gastrocnemius EMG activity, Maki et al. (2001) found an average APR onset latency for community-dwelling elderly participants of 128 ms following posterior support surface translations. Anterior COP excursion can provide an indication of the biomechanical effects of changes in EMG signals, as it is roughly proportional to changes in flexor/extensor ankle torque (McIlroy and Maki, 1993; Robinovitch et al., 2002). The electromechanical delay between muscle activation and force generation in the muscles at the ankle (which are primarily responsible for COP changes) could, at least in part, account for the observed difference in response onset latency in the current study. However, previous studies have used COP-based parameters (e.g. generation of ankle torque) to characterize onset latency for feet-in-place postural responses using a tether-release paradigm, and have reported mean reaction times of 99 ms for young adults (Robinovitch et al., 2002) and 125 ms for elderly community-dwelling women (Mackey and Robinovitch, 2005). As the participants in the current study were both older (mean age of 84 versus 78) and of a different resident population (community-dwelling versus retirement home-dwelling), age-related declines in muscle strength and reaction time may help explain the longer latency to $APR_{onset}$ compared to those reported by Mackey and Robinovitch (2005). Another possible explanation for the observed difference in the current study is that the imposed perturbation was of relatively low magnitude. The average lean angle reported by Robinovitch et al. (2002) was 5.9 degrees, while the lean angle reported by Mackey and Robinovitch (2005) was 4.77 degrees; the release angle in the current study was ~3.6 degrees. It was not infrequent that participants would have to engage the tibialis anterior in order to successfully maintain this small lean angle. Following perturbation, successful maintenance of balance would therefore require a deactivation of the ankle dorsiflexors prior to onset of the plantar flexors, inducing an additional time delay before COP movement would be detected. Compared to larger lean angles, release from a smaller lean angle would likely cause a relatively lower rate of ankle rotation, perhaps decreasing the influence of ankle proprioceptors on the resultant postural response. Previous studies have demonstrated that minimizing ankle rotation by combining a rearward support surface translation with an anterior platform tilt caused a delay in the onset of gastrocnemius activity and ankle torque generation until 200-300 ms following onset (Nashner, 1976). This notion is supported by other reports that have suggested that the average latency to initiation of postural response may be affected by the magnitude of perturbation (McIlroy and Maki, 1993). The current investigation required participants to monitor the initial location of the underfoot centre-of-pressure, as well as the initial activity of muscles spanning to ankle. The dual-task nature of this paradigm may have also caused delays in the onset compensatory balance responses. Towards the goal of this study however, the delayed onset to postural response is evidently not a function of flooring surface, given the consistent $APR_{onset}$ timing across conditions, but is a function of the perturbation itself.

It is worthwhile to consider the current results alongside the results of Chapter 2 of this thesis, as well as other reports of the force attenuative properties of various flooring surfaces. Chapter 2 illustrates that traditional compliant flooring systems, such as the commercial-grade carpet with underpadding used in the current study, may provide significant force and acceleration attenuation during simulated impacts to the back of a biofidelic surrogate headform compared to vinyl floors. Impacts onto a vinyl surface resulted in peak forces that were up to 50% larger, peak accelerations that were up to 64% higher, and Head Injury Criterion (HIC) scores that were nearly 160% greater than those measured during impacts onto the commercial carpet with underpadding. However, the protective capacity of the traditional compliant flooring systems appears to be modest when compared to NCFs; Chapter 2 also demonstrated that, compared to the same commercial carpet with underpadding, the NCFs tested in the current study were able to reduce the peak force applied to the back of a biofidelic surrogate human headform by up to 56%, 29%, and

61% for the *SC-V*, *KR*, and *ST-V* conditions, respectively. This was accompanied by relative attenuations in peak resultant acceleration of up to 63%, 40%, and 70%, and reductions in HIC scores by 80%, 55%, and 83% for the *SC-V*, *KR*, and *ST-V* conditions, respectively. Paired with reports that SmartCell and SofTile floors provide attenuation of peak force applied to the proximal femur of up to 25-50% during simulated sideways falls to the hip, compared to vinyl floors (Laing and Robinovitch, 2009), it appears that NCFs have the potential to provide a significant protective capacity against a range of fall-related injuries often suffered by older adults, including hip fractures and traumatic brain injury. Furthermore, this protection appears to be provided with limited influence on indices associated with autonomic postural responses during compensatory balance reactions.

While the primary outcome parameters reported in this study were biomechanically grounded, it is relevant to note that subjective ratings of each floor condition were also recorded. After each block of five trials on each of the floors, participants were asked to answer the following question: "On a scale of 1-10, how difficult was it to maintain your balance on this floor, where 1 = incredibly difficult, and 10 = incredibly easy." No obvious trends were apparent with respect to which floor condition scored highest by the participants. While the mean (SD) subjective ratings across floors were similar (8.2 (1.2), 7.3 (2.0), 7.4 (2.8), 7.5 (2.7), 7.7 (2.1) for *CC*, *V*, *SC-V*, *KR*, and *ST-V* respectively), the participants' comments provide further insight into how agreeable the floors were for them. Most commonly, the participants reported that they enjoyed and preferred the 'feel' and slight 'give' of the rubber NCF conditions underfoot. Another common phrase was that they enjoyed the 'traction' provided by the carpet condition. It is a worthy reminder that two of the three NCF conditions were tested with a vinyl overlay. However, a carpeted overlay could alternatively be used during installation, which might achieve the enhanced ‘traction’ provided by the carpet in concert with the ‘feel’ of the NCFs. Anecdotal support for these floors is likely of great interest to manufacturers, as well as management teams from facilities where pilot NCF installations might be considered.

There are several limitations associated with this investigation. Firstly, the sensory system responsible for detection of the perturbation using a tether-release paradigm, and thus the driver of the compensatory balance response, likely does not replicate detection of real-world imbalance episodes. Using this paradigm, the leaning participant is supported by the chest harness; thus, it is likely that the onset of perturbation is initially detected by mechanoreceptors at the trunk following the change in pressure exerted by the harness. Secondly, the participants were required to maintain their balance using feet-in-place responses following a perturbation restricted to the sagittal plane. This was based on a recent, objective report on video-recorded fall mechanics in elderly fallers (Section 1.1.3), demonstrating that 32% of falls occurred during feet-in-place activities, as a result of inappropriate weight transfer skills (e.g. rising from a chair, involving an anterior-posterior weight shift with stationary foot placement) (Robinovitch et al., 2009). Change-in-support balance recovery strategies, which involve taking a step, are also prevalent reactions following external perturbations of small and large magnitude (Maki and McIlroy, 2005). In the current study, participants were successfully able to maintain balance using feet-in-place responses for only 80.4 – 87.1 % of trials across all floors; important information may lie within the additional 13 – 20% of ‘failed’ trials that involved a change-in-support strategy, warranting additional studies to assess the influence of NCFs on such responses. Furthermore, numerous studies have implicated medial-lateral stability to be of great importance in preventing falls (Kuo, 1999; Bauby and Kuo, 2000; Donelan et al., 2004), providing another important avenue for future research. Thirdly, this study focussed primarily on centre-of-pressure-based outcomes, while kinematic data was used only to determine the whole-body centre-of-mass. This was done in an effort to assess the impact of flooring on balance control at the foot-floor interface, as centre-of-pressure movement is understood to control the location of the centre-of-mass. Future studies might consider a more in-depth analysis of the kinematic-based responses to perturbation to ascertain if flooring condition affects joint angle kinematics and restabilization strategies. Fourthly, the participants were barefoot while completing this study, in an attempt to isolate the influence of flooring condition (and not footwear) on indices of APRs. Consequently, the findings reported here might not be directly applicable to conditions where footwear is worn. Lastly, the population chosen for this study represented a highly functional group of retirement home-dwelling seniors with good mobility. While these participants were residents of the type of setting where falls and fall-related injuries occur at higher rates, further research is underway to address the effects of NCFs on the postural stability of residents who are at the *highest* risk of falling, including individuals with low-level mobility, during quiet standing and sit-to-stand tasks.

The anticipated rise in the incidence of fall-related injuries in older adults needs to be addressed with effective and robust intervention strategies. Novel compliant flooring systems appear to be one promising approach, as they have demonstrated a significant protective capacity against multiple types of fall-related injuries including hip fractures (Laing and Robinovitch, 2009) and traumatic brain injuries (Chapter 2) in a simulated setting. This protection appears to be provided with minimal influences on voluntary balance control (Laing and Robinovitch, 2009) and compensatory balance responses (Wright and Laing, 2011) in community-dwelling older women. Additional support for this intervention strategy is provided by the current results, which indicate that autonomic postural responses are not affected by novel compliant flooring systems including SmartCell, Kradal, and SofTile following a lean and release balance perturbation. These results provide support for pilot installations to inform the development of clinical trials that test the effectiveness of novel compliant floors at reducing fall-related injuries in older adults.

# CHAPTER 4 THESIS SYNTHESIS AND CONCLUSION

The demographic shift towards a more aged population within Canada is anticipated to lead to a greater incidence of fall-related injuries in older adults, posing significant emotional and financial burdens on individuals, families, health-care providers, and the Canadian health care system. Consequently, much attention has been devoted towards the development of effective intervention strategies to minimize such injuries and resultant costs over the coming decades. The central aim of this thesis was to evaluate one such strategy, involving the installation of energy-absorbing novel compliant flooring systems (NCFs). The challenge for this strategy is the design of floors that can provide a substantial protective capacity against a variety of injury types in the event of a fall without impairing balance control and postural stability, thereby increasing the chances of falling.

Previous research had demonstrated that certain models of these floors are capable of substantially reducing impact forces applied to the hip during simulated sideways falls compared to a vinyl surface, providing a protective capacity against hip fracture in the event of a fall (Laing and Robinovitch, 2009). However, other research has established that common traditional flooring systems including carpet with underpadding are also capable of reducing peak hip impact forces during simulated falls, potentially raising the question, "are novel compliant flooring systems more capable of preventing injury than traditional floors?" Chapter 2 of this thesis provides the first direct comparison of novel compliant floors to traditional compliant flooring systems, and suggests that, while floors like carpet with underpadding do provide a degree of protective capacity against head injury compared to vinyl surfaces during simulated head impacts, the attenuation of impact forces and accelerations is incredibly modest when compared to the NCFs tested. Thus, certain novel compliant flooring systems seem capable of providing a protective capacity against at least two of the most devastating types of fall-related injuries suffered by older adults (hip fracture and brain injury), to a level above and beyond that which is provided by traditional flooring systems.

Certain models of NCFs had been suggested to minimally affect fall risk relative to vinyl floors for older community-dwelling women, as supported by results of postural sway during quiet stance, times to perform the Timed Up and Go test, and on indices of autonomic postural responses following rearward support surface translations (Laing and Robinovitch, 2009; Wright and Laing, 2011). However, novel compliant flooring systems are most likely to be installed into settings where falls and fall-related injuries occur most frequently, including residential care facilities, hospitals, and retirement homes, where the residents are increasingly frail compared to community-dwelling seniors. As such, it is critical that the potential impact of novel compliant floors on the balance control characteristics of the target user population be assessed, a task that has begun to be accomplished in this thesis. Chapter 3 demonstrates that, compared to both vinyl and underpadded carpet, at least three commercially-available NCFs do not have a significant effect on indices of autonomic postural responses for feet-in-place reactions following release from a 3.5 degree forward lean for 15 healthy, high-mobility retirement home-dwelling seniors. Hence, the NCFs tested in Chapter 3 do not seem to impact the initial phase of compensatory balance reactions, suggesting that they may not provide an increased risk for anterior-posterior falls for this population.

While this thesis provides further support that certain NCFs appear to be a very promising intervention strategy towards accomplishing the goal of reducing fall-related injuries, there are many additional lines of research that still warrant attention. Directly extending from the work presented in this thesis, there are additional balance assessments that need to be conducted to confirm that novel compliant floors will not lead to increased fall risk for residents of settings with high incidences of falls and fall-related injuries. Prior to recruitment, the participants in Chapter 3 were identified as being of relatively high mobility, as the balance task they were asked to complete was fairly challenging. Consequently, the population tested is likely not representative of the full spectrum of mobility levels exhibited by residents of settings with high fall incidences, and likely did not include individuals at the highest risk of falling. Further research should be conducted to investigate the potential effect of NCFs on the balance control characteristics of the lower mobility groups. While these floors are suggested as a strategy to prevent injuries to older adults in the event of a fall, the needs of the care staff must also be taken into consideration. It is essential that the altered compliance of the floors does not substantially increase work demands for these individuals, such as pushing wheelchairs, equipment carts, or using lift assists. Furthermore, the creep behaviour of the floor under prolonged loading needs to be assessed (such as the weight of a bed over time). With regards to retrofitting existing facilities with NCFs, issues that have successfully been addressed include installation of ramps and transition markers between traditional and compliant flooring zones, ensuring sufficient clearance for doors, and maintaining standard heights for infrastructure including toilets and sinks.

Despite the encouraging evidence regarding the potential of novel compliant flooring systems, the goal of this thesis is not to suggest that NCFs represent the single best intervention approach towards reducing fall-related injuries. This approach has many strengths, but also has limitations. Regarding strengths, compared to many other intervention approaches (including protective devices like wearable hip protectors, balance/strength training programs, and the use of pharmacological agents) NCFs are not dependent on active user compliance for clinical effectiveness. Furthermore, this approach protects against multiple types of injuries, including hip fracture and head injury, which might be considered as two of the most devastating injuries suffered by older adults. However, NCFs will provide no protective capacity in the event that, during a fall, impact is made with something other than the floor, such as the wall or a piece of furniture. Consequently, they may represent only one part of an optimal combination of intervention strategies that may hopefully prevent the majority of fall-related injuries.

In order to stem the expected increase in fall-related injuries, effective intervention strategies are urgently needed. Laboratory-based studies indicate that novel compliant flooring systems appear to be one very effective strategy, as they are able to substantially reduce impact forces and accelerations applied to various body parts while inducing only minimal concomitant impairments to indices of balance control and postural stability in older adults. This thesis has provided further support for this approach, and supports the development of pilot installations to test their effectiveness in clinical settings.

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# APPENDIX A – VERIFICATION OF IMPACT VELOCITIES DETERMINED FROM LIGHT GATE VELOCIMETER

In Chapter 2 of this thesis, simulated head impacts were completed at multiple impact velocities to determine the effect of flooring surface on head impact dynamics. Impact velocities were recorded using a light gate velocimeter. Prior to collection of the data presented in Chapter 2, pilot tests were conducted in order to verify that the velocity readings obtained from the light gate velocimeter were accurate. This was accomplished through the attachment of an infrared emitting marker on the flag that passes through the light gate immediately prior to impact. Two simulated head impacts were completed at each of two release heights (arbitrarily chosen). The displacement of this marker was tracked using three Optotrak sensors. Using three-point central difference differentiation, the velocity of the marker at impact was calculated and compared to the impact velocity determined from the velocimeter. Table A.1, shown below, summarizes the results of this pilot test, and indicates that the average error in light gate velocimeter readings (relative to the velocity determined from the kinematic approach) was 0.04%. Based on this finding, it was assumed that impact velocities recorded from the light gate velocimeter in subsequent testing sessions were accurate.

**Table A-1: Summary of impact velocities determined from kinematic data compared to those determined using the light gate velocimeter.**

| Trial | Release Height | Impact Velocity (m/s) | |
|---|---|---|---|
| | | Kinematics | Velocimeter |
| 1 | Low | -4.28 | -4.31 |
| 2 | Low | -4.20 | -4.23 |
| 1 | High | -5.03 | -4.99 |
| 2 | High | -5.09 | -5.05 |

# APPENDIX B – EFFECT OF PLYWOOD MOUNTING SURFACE ON HEAD IMPACT DYNAMICS

In Chapter 2 of this thesis, head impact dynamics were assessed following simulated impacts onto various flooring surfaces. In order to mount these flooring surfaces in the drop tower apparatus, they were supported by a plywood sheet (1 cm thickness) that was mounted onto to the top of the load cell. In order to assess the influence of this plywood sheet on the outcome parameters reported in Chapter 2, pilot impacts were conducted directly onto the plywood, and also directly onto the load cell, using a release height of 85 cm. The results of this pilot test are summarized below in Table B.1, and indicate that the plywood support surface had the effect of reducing peak resultant accelerations ($g_{max}$) by 9.6%, Head Injury Criterion (*HIC*) scores by 5.9%, and peak impact forces ($F_{max}$) by 11.2%. As such, it is anticipated that the values reported in Chapter 2 for each of these parameters are slightly lower than values that would have been obtained if the flooring surfaces could have been mounted directly onto the load cell (i.e. without the plywood support surface).

**Table B-1: Comparison of head impact outcome parameters following impacts onto the plywood support surface versus impacts directly onto the load cell**

| Impact Surface | Trial | $g_{max}$ (g) | *HIC* | $F_{max}$ (N) |
|---|---|---|---|---|
| Plywood | 1 | 271.7 | 2143.0 | 5296788 |
| Plywood | 2 | 270.8 | 2059.2 | 5265897 |
| Plywood | 3 | 270.3 | 2122.0 | 5292270 |
| *Mean$_{Plywood}$ (SD)* | | *270.9 (0.7)* | *2108.1 (43.6)* | *5284985 (16684)* |
| Load Cell | 1 | 297.6 | 2180.0 | 5925601 |
| Load Cell | 2 | 300.1 | 2252.4 | 5937645 |
| Load Cell | 3 | 301.8 | 2288.8 | 5999728 |
| *Mean$_{Load Cell}$ (SD)* | | *299.9 (2.1)* | *2240.4 (55.4)* | *5954325 (39779)* |

# APPENDIX C – SPECIFIC VALUES FOR OUTCOME PARAMETERS PRESENTED IN CHAPTER 2

**Table C-1: Means (SD) for peak force ($F_{max}$), peak acceleration ($g_{max}$), and Head Injury Criterion (*HIC*) during simulated head impacts at 1.5, 2.5, and 3.5 m/s across the ten flooring conditions tested in Chapter 2 of this thesis; * indicates significant ($p<0.05$) difference compared to *CC*.**

| Variable | Floor | Impact Velocity (m/s$^2$) | | |
|---|---|---|---|---|
| | | 1.5 | 2.5 | 3.5 |
| $F_{max}$ (N) | *CC* | 3119 (111) | 6676 (165) | 11583 (34) |
| | *V* | 4676 (11) * | 8721 (43) * | - |
| | *RC* | 3376 (83) * | 6961 (118) * | 11356 (300) |
| | *BC* | 3045 (93) | 5896 (253) * | 10776 (131) * |
| | *SC* | 1487 (34) * | 2886 (45) * | 5584 (58) * |
| | *SC-V* | 1740 (20) * | 2953 (13) * | 5281 (205) * |
| | $KR_{12}$ | 2289 (15) * | 4753 (21) * | 8552 (40) * |
| | $KR_{24}$ | 1676 (17) * | 3318 (25) * | 5913 (88) * |
| | *ST* | 1562 (22) * | 2722 (17) * | 4211 (36) * |
| | *ST-V* | 1733 (11) * | 3015 (15) * | 4491 (26) * |
| $g_{max}$ (*g*) | *CC* | 54.7 (3.4) | 122.7 (3.8) | 262.1 (11.1) |
| | *V* | 89.7 (6.0) * | 170.0 (3.0) * | - |
| | *RC* | 65.9 (3.6) * | 137.6 (6.5) * | 217.4 (4.8) * |
| | *BC* | 57.0 (0.7) | 113.7 (7.2) * | 213.6 (5.7) * |
| | *SC* | 29.4 (6.2) * | 51.1 (1.1) * | 109.0 (2.4) * |
| | *SC-V* | 30.6 (2.3) * | 51.7 (1.3) * | 97.9 (3.0) * |
| | $KR_{12}$ | 41.0 (1.8) * | 91.2 (3.9) * | 156.6 (1.4) * |
| | $KR_{24}$ | 32.0 (5.4) * | 58.5 (0.1) * | 111.5 (5.1) * |
| | *ST* | 28.6 (1.8) * | 46.0 (1.0) * | 76.6 (2.5) * |
| | *ST-V* | 27.5 (0.6) * | 51.7 (0.6) * | 77.7 (1.0) * |
| *HIC* | *CC* | 39.1 (3.9) | 258.0 (23.7) | 1068.0 (40.6) |
| | *V* | 101.4 (5.1) * | 495.9 (6.2) * | - |
| | *RC* | 50.4 (5.2) * | 302.2 (12.0) * | 963.5 (75.9) * |
| | *BC* | 41.4 (0.5) | 217.4 (10.7) * | 817.1 (44.2) * |
| | *SC* | 14.4 (2.1) * | 63.3 (3.9) * | 237.0 (6.6) * |
| | *SC-V* | 18.9 (2.6) * | 72.5 (3.2) * | 216.0 (3.6) * |
| | $KR_{12}$ | 26.1 (2.1) * | 152.4 (7.2) * | 482.0 (24.5) * |
| | $KR_{24}$ | 16.3 (1.5) * | 81.1 (1.7) * | 252.9 (2.7) * |
| | *ST* | 17.6 (2.1) * | 61.9 (1.2) * | 161.5 (4.3) * |
| | *ST-V* | 16.0 (0.8) * | 78.1 (3.3) * | 177.5 (6.4) * |