

**BIOMECHANICAL TESTING OF HIP PROTECTORS AND  
ENERGY-ABSORBING FLOORS FOR THE PREVENTION  
OF FALL-RELATED HIP FRACTURES**

by

Andrew Charles Thomas Laing  
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# APPROVAL

**Name:** Andrew Charles Thomas Laing  
**Degree:** Doctor of Philosophy  
**Title of Thesis:** Biomechanical testing of hip protectors and energy-absorbing floors for the prevention of fall-related hip fractures

**Examining Committee:**

**Chair:** **Dr. Scott Lear**  
Associate Professor, School of Kinesiology

---

**Dr. Stephen Robinovitch**  
Senior Supervisor  
Associate Professor, School of Kinesiology

---

**Dr. Max Donelan**  
Supervisor  
Assistant Professor, School of Kinesiology

---

**Dr. Tom Oxland**  
Supervisor  
Professor, Department of Orthopaedics, University of British Columbia

---

**Dr. Peter Cripton**  
Internal Examiner  
Assistant Professor, Department of Mechanical Engineering,  
University of British Columbia

---

**Dr. Stefan Duma**  
External Examiner  
Professor, Department of Mechanical Engineering, Virginia Polytechnic Institute and State University

**Date Defended:** June 18<sup>th</sup>, 2008

## **ABSTRACT**

The general objective of my thesis research was to characterize the stiffness and force distribution characteristics of the hip region during the impact phase of sideways falls, and to advance our understanding of the potential for external engineering interventions (e.g. hip protectors and compliant floors) to reduce hip fracture risk by reducing the force applied to the proximal femur during such falls. This thesis is comprised of five studies. In the first I characterized the degree of non-linearity in pelvic stiffness, and examined the influence of stiffness characterization methods on the accuracy of mathematical models in predicting impact dynamics during falls on the hip. In the second study I employed a pelvis release paradigm (a method of inducing low severity but clinically relevant falls) to examine whether soft shell hip protectors alter the distribution of force throughout the hip region during impact. The third study entailed a sensitivity analysis to determine the influence of mechanical test system properties on the force attenuation provided by hip protectors. In the fourth study I used pelvis release experiments to determine how the force applied to the pelvis is affected by body impact configuration and floor stiffness; I also examined the ability of a mass-spring model to predict these relationships. The final study used a mechanical fall simulator to assess the attenuation in femoral neck force provided by four low stiffness floors compared to a standard rigid floor, and assessed the influence of these floors on fall risk through a range of static and dynamic balance tests with fifteen elderly women. Overall, this thesis demonstrates that compliant floors and soft shell hip protectors substantially reduce the force applied to the proximal femur during the impact stage of sideways falls. Of equal importance, this work demonstrates the need for international standards for the biomechanical testing and market approval of these devices. These are essential steps for increasing the quality of hip protectors and compliant floors available in the marketplace, and consequently, for enhancing their ability to reduce hip fracture risk in vulnerable populations.

**Keywords:** injury biomechanics; injury prevention; hip fractures; hip protectors; compliant floors; accidental falls; ageing

**Subject Terms:** Biomechanics; Human mechanics; Musculoskeletal system -- Wounds and injuries; Musculoskeletal system -- Mechanical properties; Impact -- Physiological effect; Biomedical engineering

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# **CHAPTER 1 GENERAL INTRODUCTION AND LITERATURE REVIEW**

## **1.1 Hip fractures – scope of the problem**

Hip fractures are a major public health issue. The lifetime risk of hip fracture for white American women from 50 years of age onwards is approximately 17% (Melton, 2000). There are approximately 23,000 cases of hip fracture every year in Canada with associated treatment costs of about \$1 billion (Papadimitropoulos et al., 1997). The costs are even higher in the U.S. with treatment costs approximated at \$10 billion (Melton, 1993). Hip fractures cause increased mortality (Empana et al., 2004a, Meyer et al., 2000, Wolinsky et al., 1997), and declines in mobility, physical activity, and functional independence (Norton et al., 2000a, Wolinsky et al., 1997). Without improvements in prevention, hip fracture incidence is expected to increase 4-fold by the year 2041 due to demographic shifts towards a more aged population and the exponential increase in fracture risk with advancing age (Jaglal et al., 1996, Papadimitropoulos et al., 1997, Wiktorowicz et al., 2001). Recent evidence suggests that current screening and strategies and pharmaceutical interventions have tempered the expected rate of increase in hip fracture incidence in Ontario where all seniors have universal access to medical treatment (Jaglal et al., 2005). However, it is not clear whether this trend will continue as the demographics shift towards an increasingly aged population. In addition, this trend may not represent the reality for populations such as the U.S. in which health care provision

policies are markedly different. As such, there is a clear need for additional prevention efforts to reduce the burden associated with hip fractures.

## **1.2 Hip fractures – risk factors**

The risk of hip fracture is dependent upon two variables: force applied to the hip and the strength of the hip (Hayes et al., 1991). This is reflected in figure 1.1, in that all factors ultimately operate through these two parameters. The risk factors associated with bone strength are briefly summarized in section 1.2.2. The factors that influence applied force – the focus of my thesis - are presented in detail in section 1.2.3.

### **1.2.1 Mechanical testing of bone strength**

Mechanical test systems can be used to assess the biomechanical properties of bone related to strength. After appropriate fixation, loads are applied and the resulting tissue deformation is observed. For bone, there is an initial linear relationship between load and deflection termed structural stiffness (Burr and Turner, 2003). The upper end of this linear elastic region is termed the yield point, below which the material will return to its initial length upon load removal. Above the yield point the slope of force-deflection curve decreases, and permanent damage occurs in the structure (as indicated by residual deformation in the absence of applied load). The upper limit of this plastic region is termed ultimate failure, the point at which a catastrophic fracture occurs.

While a variety of features from fracture experiments could be used to characterize bone strength, the most commonly used outcome is the magnitude of applied force at the time of catastrophic failure (which in almost all cases is the peak force). This ‘fracture strength’ or ‘fracture threshold’ is intuitive and is easily measured in materials

testing systems. In contrast, impulse (the integral of the force vs. time trace) is rarely reported, as the effect of loading rate on proximal femur strength is relatively minor (influencing fracture threshold by up to 20%)(Courtney et al., 1994). Similarly, the energy absorbed before fracture (termed 'toughness') is not commonly cited, perhaps due to its large dependence on the degree of plastic deformation after initial yield. For these reasons, my primary measure of fracture risk throughout this thesis was peak force applied to the hip region. Using the peak applied force as the primary outcome parameter also allowed me to take a "factor of risk" approach, which is common in engineering design (Hayes et al., 1991), of comparing the applied force to the failure force in order to characterize the protective value of interventions including hip protector and compliant floors.

### **1.2.2 Risk factors operating through bone strength**

The fracture threshold of the proximal femur is dependent on age, and to a lesser extent, the rate and direction of applied load. Lotz and Hayes (1990) tested cadaveric femora from older donors simulating a sideways fall impact and reported fracture force to range from 778 to 4040 N (mean 2110, SD 1060). Courtney et al. (1994) positioned the femora to simulate sideways falls resulting in direct impact to the greater trochanter, and studied the influence of age and loading rate on fracture strength. They found the strength to be significantly higher for both young (mean age = 32 (SD 13) yrs) and elderly (mean age = 74 (SD 7) yrs) specimens at the higher loading rate (100 m/s vs. 2 mm/s). In addition, fracture strength was significantly lower for the elderly specimens. Specifically, fracture strength of the elderly specimens was 3440 N (SD 1330) and ~4100 N (SD ~1300) at loading rates of 2 and 100 mm/s, respectively. In contrast, fracture strength of

the younger specimens was 7200 N (SD 1090) and ~7800 N (SD ~1400) at loading rates of 2 and 100 mm/s, respectively (fracture thresholds not exact as they were approximated from figures). Using a loading rate of 2 mm/s, Pinilla et al. (1996) reported fracture threshold of femurs from elderly donors to be 4050 (SD 900), 3820 (SD 910), and 3060 (SD 890) for femur orientations of 0, 15, and 30°, respectively (simulating slightly rolled forward, sideways, and slightly rolled backwards impact configurations, respectively). However, there is substantial variability in the fracture threshold across, and within, studies (coefficients of variability of approximately 30 % are typical). Numerous lines of research have attempted to explain this variability (e.g. the influence of specimen age, rate and direction of loads, bone mineral density, bone quality, proximal femur geometry) in order to enhance predictive models of fracture risk (Bouxsein et al., 2007, Courtney et al., 1994, Courtney et al., 1995, Lochmuller et al., 2000, Manske et al., 2006, Pinilla et al., 1996).

Epidemiological studies demonstrate that hip fracture risk is associated with numerous bone-strength related factors including muscular strength, weight, and diet (see the review of Cumming et al. 1997 in table 1.1). One of the best such predictors is bone mineral density, which confers 2 to 3-fold increase in risk for each decrease of one standard deviation (Cumming et al., 1997, Cummings et al., 1995, Dargent-Molina et al., 1996, Nguyen et al., 1996). This explains why those with osteoporosis (a standardized bone mineral density score of more than 2.5 standard deviations below normal peak values for young adults) are at such a high risk of hip fracture.

A number of pharmacologic interventions have been shown to increase bone mineral density and/or decrease fracture risk. For example, bisphosphonates (bone

resorption inhibitors such as alendronate and risedronate) have been shown to increase bone mineral density (Cummings et al., 1998, Liberman et al., 1995, Orwoll et al., 2000, Watts and Becker, 1999) and to decrease the risk of hip fracture (Black et al., 1996, McClung et al., 2001). Similarly, parathyroid hormone has been shown to increase bone mineral density (Black et al., 2003, Finkelstein et al., 2003) and decrease the risk of vertebral and non-vertebral fragility fractures (Neer et al., 2001). Estrogen replacement therapy has been shown to decrease the risk of hip fracture, especially for those women who start treatment within 5 years of menopause (Cauley et al., 1995). Recently, Reginster et al. (2005) reported that a 3 year course of treatment with oral strontium ranelate decreased hip fracture risk.

However, there are limitations associated with pharmacologic interventions. For example, alendronate and strontium ranelate have both proven ineffective at reducing hip fracture risk for women without confirmed osteoporosis (McClung et al., 2001, Reginster et al., 2005). As a substantial portion of those who suffer hip fractures have normal bone strength (Dargent-Molina et al., 1996, Taylor et al., 2004), this subgroup will not benefit from therapies aimed at increasing bone strength. In addition, pharmacologic interventions (especially bisphosphonates) are unacceptable to some at-risk persons due to undesirable side effects such as gastro-intestinal tract irritation (Body et al., 2004). As such, it is questionable how effectively pharmacologic interventions alone can reduce the overall burden of hip fractures. Therefore, interventions acting via other mechanisms (e.g. reduction of applied force) should be considered.

### **1.2.3 Force applied to the hip during a fall**

Falls onto the hip, more specifically the greater trochanter, are the event most directly linked to hip fractures in the elderly (Nevitt and Cummings, 1993, Zuckerman, 1996). Over 90% of hip fractures are due to falls (Grisso et al., 1991). As such, falls provide the necessary circumstances during which the force applied to the proximal femur exceeds the fracture threshold. A requisite for the prediction of hip fracture risk is knowledge of the magnitude of these applied forces (Hayes et al., 1991). Average peak force applied to the hip during self-initiated sideways, postero-lateral, and posterior falls onto a 13 cm foam mattress overlying a force plate have been reported as 2252 N (SD 442), 2498 N (SD 457) and 3247 N (SD 587), respectively (Nankaku et al., 2005). However, quantifying these forces for falls onto rigid surfaces is a conundrum for researchers – as the epidemiologic evidence suggests such falls result in applied forces sufficient to cause fracture in a significant portion of the elderly population, it is obviously not ethical to measure the impact forces associated with falls onto rigid floors in-vivo.

An alternative to experimental measures is the use of an appropriately tuned mathematical model to predict the peak force applied to the hip during falls from standing height. Robinovitch et al. (1991) used a Voigt support model (a mass above parallel spring and damper elements) to predict average peak forces of 5600 and 8600 N for falls in muscle-relaxed and -active states, respectively. Using improved parameter estimates, they later predicted peak forces ranging from 1150 to 5288 N (Robinovitch et al., 1997a). In addition, they demonstrated that ignoring effective damping (i.e. using a mass-spring model) eliminated the non-physiologic instantaneous force for non-zero impact velocities

predicted by a Voigt model without appreciably reducing the accuracy of peak force predictions (Robinovitch et al., 1997b). van den Kroonenberg et al. (1995) used such a mass-spring model to predict peak forces of 2900, 3580, and 4260 N for the 5, 50, and 95 percentile females during sideways falls from standing height. They concluded that these forces are well below the average femoral fracture threshold of young adults, but are sufficient to result in hip fracture in about 50% of elderly persons.

The quality of model outputs is usually dependent on the accuracy of model inputs and / or complex interactions between model inputs. For example, an infinite combination of effective mass, stiffness, and impact velocity values in a mass-spring model of impact result in an identical prediction of peak force (figure 1.2). However, I will later show that the choice of input parameters has important consequences for the accuracy of model predictions of the force attenuation provided by protective equipment. As such, it is important to critically evaluate the methods by which the various research groups calculated the parameters in their predictive models. Based on free-fall dynamics and the principle of energy conservation, Robinovitch et al. (1991) estimated an impact velocity of 3.7 m/s for a fall from standing height. Support for this value was provided by van den Kroonenberg et al. (1996) who measured the impact velocity of the hip during self initiated sideways falls onto a foam mat to range from 2.14 to 4.79 m/s with a mean of 3.17 m/s, and by Feldman and Robinovitch (2007) who observed a mean hip impact velocity of 3.0 (SD 1.0) m/s in young persons during unexpected sideways falls. The remaining parameters for mass-spring or Voigt models were estimated by Robinovitch et al. (1991, 1997a) using pelvis release experiments with healthy young subjects. Effective mass depends on the portion of total body mass that has a non-zero velocity during

impact. Effective stiffness and damping depend on the force-length and force-velocity properties of the soft tissues and underlying skeletal structures. During the pelvis release trials, the participant lay on the ground with her pelvis in a sling. The pelvis was raised and suddenly released, introducing a step change in force applied to the lateral aspect of the hip. Effective mass was originally calculated as the resting force applied to the hip (Robinovitch et al., 1991), and was reported to be dependent on the degree of activation of the trunk musculature. However, more elegant parameter identification techniques were later employed which showed effective mass to be independent of muscle activation and equal to approximately 50% of total body mass (Robinovitch et al., 1997a). Effective stiffness and damping were calculated based on the decaying oscillation in ground reaction force (termed the free vibration response approach). Robinovitch et al. (1991) demonstrated that both parameters initially increased with increasing applied force, but plateaued at applied forces greater than 300 N. This observation was used to justify the use of linear (constant) effective stiffness and damping in models employed by van den Kroonenberg et al. (1995) and Robinovitch et al. (1997a).

However, the validity of using a linear estimate of effective stiffness is questionable for a number of reasons. First, the peak forces in the experimental validation trials were approximately one order of magnitude lower than the model predictions of peak force for standing height falls. Second, Mills (1996, 2007) reports that the force-deflection behaviour of soft tissues in the hip region is non-linear, with stiffness increasing as applied force increases. Collectively, these issues cast some uncertainty on the validity of using constant values of effective stiffness in models that predict the force applied to the hip during sideways falls. An alternative approach would be to estimate

effective stiffness based on the force-deflection characteristics of the pelvis during lateral impact. This technique has never been employed with living subjects, and as such, it is unknown how well the calculated stiffness would match that determined from the vibration response method, and whether one would result in more accurate model predictions of impact force.

These issues were examined in the first study of this thesis, described in Chapter 2 of this document. Healthy young subjects engaged in pelvis release trials with impact velocities up to 1 m/s. The force applied to the hip was measured with a force plate, while a motion capture system measured the subject's pelvic compression during impact. Effective stiffness was estimated via the standard frequency response approach (Robinovitch et al., 1991, Robinovitch et al., 1997a) and by fitting linear and non-linear curves to the measured force-deflection data. The accuracy and sensitivity of model predictions of peak force to these different estimates of effective stiffness were examined by comparing model output to experimentally measured forces. This study provides insight into the validity of previous peak force predictions, and will inform decisions on the best methods to use when expanding existing mathematical models to investigate other aspects of fall dynamics e.g. floor stiffness.

All of the experimental and modelling efforts to date have measured or predicted the total force applied to the entire contact area of the hip region during impact. However, these models fail to indicate what portion of this total force is applied to the proximal femur, which is the parameter of primary interest for predicting hip fracture risk (Lauritzen, 1997). Forces directed into the soft tissues anterior and posterior to the greater trochanter may not influence hip fracture risk. In addition, the area of contact between the

hip region and the floor during the impact phase of a fall has been suggested as an important factor in the etiology of hip fractures. While total contact area between the hip region skin surface and ground has been estimated as 110 cm<sup>2</sup> (Askegaard and Lauritzen, 1995), the contact area of the greater trochanter has been estimated at 2 - 6 cm<sup>2</sup> based on the size of hematomas in hip fracture patients (Lauritzen, 1997).

The dual importance of force magnitude and contact area points towards the importance of considering the pressure applied to the hip region during the impact phase of a fall. Using a surrogate pelvis matched to the average surface geometry of 20 elderly women, Minns (2004a) reported a pressure of ~180 mm Hg acting on the skin overlying the greater trochanter in a simulated sideways sleeping position. Unfortunately, no information is available on the distribution of pressure throughout the hip region during the impact phase of a sideways fall on the hip. Such data would enhance our ability to predict the actual force applied to the proximal femur during fall impacts, allowing more precise estimate of hip fracture risk during falls.

This gap in the literature was examined in the second study of this thesis in which I used pelvis release experiments with human subjects to determine how applied force, and subsequently pressure, is distributed throughout the hip region during the impact phase of a sideways fall. These data afforded us insights into the portion of total impact force that is directed into the proximal femur during a fall impact. In addition, they provided us with a baseline on which to compare the ability of external interventions (e.g. hip protectors) to reduce the force applied to the regions directly overlying and surrounding the proximal femur.

### **1.3 Strategies to reduce the force applied to the hip during falls**

Over 90% of hip fractures are due to falls (Grisso et al., 1991). However, only 1 - 2% result in hip fractures (Nevitt et al., 1991, Tinetti et al., 1988). As such, it is pertinent to consider the mechanics of falls, and to identify those characteristics that are associated with higher risk of fracture. Case control studies examining fall mechanics demonstrate that landing on or near the hip increases hip fracture risk by 30-fold (Nevitt and Cummings, 1993, Schwartz et al., 1998, Tinetti et al., 1988). Falling sideways increases the probability of landing on the hip, and increases hip fracture risk by 5-6 fold (Greenspan et al., 1994, Nevitt and Cummings, 1993, Schwartz et al., 1998). Lower and upper limb weakness increases risk of fall-related injury by 5 and 2-fold, respectively (Nevitt and Cummings, 1993, Schwartz et al., 1998, Tinetti et al., 1995). In contrast, a decrease of 1 standard deviation in bone mineral density only is associated with only a 2 to 3-fold increase in hip fracture risk (Cummings et al., 1995, Dargent-Molina et al., 1996, Nguyen et al., 1996). As such, the direction of a fall, the impact configuration, and a person's ability to engage in protective landing strategies are at least as important in determining hip fracture risk as factors related to bone strength. In recognition of this, there is an emerging body of literature studying the effectiveness of interventions aimed at improving balance maintenance and recovery (Liu-Ambrose et al., 2004, Province et al., 1995, Robertson et al., 2002, Wolf et al., 1996) or safer fall techniques (Groen et al., 2006, Hsiao and Robinovitch, 1998, Leavitt, 2003, Robinovitch et al., 2003, Sabick et al., 1999). However, there are questions as to the efficacy of these interventions for specific groups of elderly population, and their cost-effectiveness. As such, it is pertinent to

consider other avenues by which forces applied to the proximal femur might be decreased.

An alternative approach is to alter the environment in such a manner as to reduce the applied force in the event of a fall. Low stiffness floors and hip protectors have been proposed as interventions which have the potential to achieve this end (Cummings and Nevitt, 1994, Kannus et al., 1996, Kannus et al., 2005, Keegan et al., 2004, Robinovitch et al., 2000b) and will be studied in depth in Chapters 3 to 6 of this thesis.

### **1.3.1 Hip protectors**

Hip protectors are external padding systems worn over the hip region which aim to reduce the force applied to the proximal femur during the impact phase of a fall. Hard shell protectors are typically comprised of a rigid dome or horseshoe that does not directly contact the skin overlying the greater trochanter (figure 1.3A). During impact, the rigid elements direct a portion of the energy away from the proximal femur into the surrounding soft tissues, and as such, are often described as energy shunting protectors. In contrast, soft shell protectors are usually comprised of compliant foam or rubber materials positioned directly on the skin overlying the greater trochanter and proximal femur. During impact, the compliant materials are presumed to absorb a portion of the total impact energy (figure 1.3B). Consequently, soft shell protectors are often described as energy absorbing hip protectors even though very few studies have actually measured the mechanisms by which they act.

### **1.3.1.1 Clinical effectiveness of hip protectors**

Currently there is substantial debate about the clinical effectiveness of hip protectors at reducing hip fractures. Initial prospective studies in which the intervention and control groups were randomly assigned by ‘clusters’ (e.g. nursing homes, or individual wards within institutions) indicated that hip protectors reduced hip fracture risk for elderly persons living in institutional settings (Ekman et al., 1997, Harada et al., 2001, Kannus et al., 2000, Lauritzen et al., 1993). However, more recent studies in which group status was randomized at the individual level have not confirmed these results (Cameron et al., 2001, Chan et al., 2000, Hubacher and Wettstein, 2001, Kiel et al., 2007, van Schoor et al., 2003b). Furthermore, hip protectors have not been found to reduce hip fracture risk for elderly individuals living in the community (Birks et al., 2003, Birks et al., 2004, Cameron et al., 2003). Two recent reviews suggest that there is little support for clinical effectiveness for community dwellers, and that their effectiveness is uncertain for those dwelling in institutional settings (Parker et al., 2006, Sawka et al., 2005).

However, the clinical effectiveness of hip protectors is directly related to user compliance, as emphasized in Ekman et al.’s (1997) statement that “The effect of hip protectors is obviously linked to where the protector is, whether it is on the hip or in the drawer.” A systematic review by van Schoor et al. (2002) reported that primary acceptance with hip protectors ranged from 37% to 72% (median 68%), while compliance varied between 20% and 92% (median 56%). Perhaps surprisingly, there is little evidence which suggests that hip protector characteristics play a prominent role in determining user compliance (O’Halloran et al., 2005, Suzuki et al., 1999). In contrast, user characteristics (e.g. dementia, dependency on staff, fear of falling), residence

characteristics (e.g. size of nursing home), training and education, and staff characteristics have been reported as the most important factors influencing compliance (Burl et al., 2003, Cryer et al., 2006, Forsen et al., 2004, O'Halloran P et al., 2006, O'Halloran et al., 2005, Thompson et al., 2005, van Schoor et al., 2003a, Warnke et al., 2004). Despite these inconsistencies, there is universal agreement that user compliance is a critical component influencing the clinical effectiveness of hip protectors, and that efforts to improve compliance should be considered (e.g. through design features and education programs).

#### **1.3.1.2 Biomechanical effectiveness of hip protectors**

The biomechanical effectiveness of a hip protector is typically defined as the percentage decrease in an outcome of interest (e.g. peak force, stress, pressure applied to the proximal femur) it provides compared to an unpadding or unprotected condition. As it is not safe to study standing height falls onto rigid surfaces in-vivo, mechanical fall simulators are often used to evaluate the biomechanical effectiveness of hip protectors. Typically, a test system is composed of a surrogate pelvis that strikes an impact surface, simulating the impact phase of a fall. Sensors measure the force associated with these impacts with and without the hip protector in place. The data generated from these test systems are crucial for the design and development of hip protectors, and for informing consumer purchase decisions.

##### ***1.3.1.2.1 Hip protector test results***

Despite the uncertainty about the clinical effectiveness of hip protectors, laboratory-based studies have overwhelmingly demonstrated the biomechanical effectiveness of hip protectors. Although all hip protectors reported in the literature have

been shown to reduce the force applied to the proximal femur, there is a wide range in the reports of biomechanical effectiveness (tables 1.2 and 1.3).

In part, the variability in biomechanical effectiveness can be explained by the variation in design features across hip protectors (e.g. geometry, materials) (see the *within* column comparisons in tables 1.2 and 1.3). For example, Robinovitch et al. (1995a) reported that a horseshoe-shaped protector composed of a dilatant material (a concentrated suspension of microparticles in a liquid medium) decreased peak trochanteric force by 64.7% compared to an unpadding impact, more than double the amount of any of the other eight pads they tested. Furthermore, they observed that the attenuation in force was linearly related to the thickness of the protector. Similarly, Mills (1996) reported that the biomechanical effectiveness of the most common hip protector on the market was enhanced when the thickness of the soft foam back was increased. A Finnish group (Kannus et al., 1999, Parkkari et al., 1995) demonstrated that the biomechanical effectiveness of their rigid, dome-shaped hard shell protector was independent of impact energy, in contrast to two common competitor products. Okizumi et al. (1998) demonstrated that adding a resin cover to a gel based hip protector increased its biomechanical effectiveness from 30.2 to 46.1%. Nabhani et al. (2002) demonstrated that a hard shell hip protector designed to be glued directly to the surface of the hip provided almost as much force attenuation (77.4%) as common garment based protectors, in addition to a surprising finding that a crude polystyrene prototype with “little shunting and very poor energy absorption” attenuated peak force by 58.1 %. Derler et al. (2005) tested the biomechanical effectiveness of 10 hip protectors (3 hard shell, 7 soft shell) and found pad characteristics to significantly affect biomechanical effectiveness (range of

26.0 to 63.1% for the 49.1 J impact energy). Although they do not report the specific hip protectors tested, they observed that energy shunting hip protectors tend to transmit higher loads to the femoral neck than energy absorbing hip protectors at higher impact energies. van Schoor et al. (2006) also found pad characteristics to significantly affect biomechanical effectiveness (range of 18.7 to 86.5% across 10 hip protectors). However, in contrast to Derler et al. (2005) they reported that hard shell protectors were superior to soft shell protectors. Cumulatively, these studies clearly demonstrate that the biomechanical effectiveness of hip protectors is dependent on their design characteristics.

However, the range in biomechanical effectiveness values reported in the literature is a function of more than just hip protector design features as different laboratories have reported widely varying values of biomechanical effectiveness *for the same hip protector* (see the *between* column comparisons in tables 1.2 and 1.3). For example, the biomechanical effectiveness reports range from 18.2% to 83.7 for the Sahva protector (Robinovitch et al. 1995 and Nabhani and Bamford 2002, respectively), 18.7 to 78.9% for Safetypants FI (van Schoor et al. 2006 and Kannus et al. 1999, respectively) and 49.5 to 80.9% for Safehip (Kannus et al. 1995 and Nabhani and Bamford 2002, respectively). In addition, the relative ranking of hip protectors do not always correspond across studies. For example, Kannus et al. (1999) reported reasonably close results for biomechanical effectiveness of Safehip and Safetypants FI. In contrast, van Schoor et al. (2006) reported the biomechanical effectiveness of Safetypants FI to be approximately 50% lower than the value for Safehip. Collectively, these results point towards the importance of considering the influence of test system characteristics on the force attenuation provided by hip protectors.

#### ***1.3.1.2.2 Mechanical systems for testing hip protectors***

Unfortunately, there are no standard guidelines outlining the characteristics that should be incorporated into test systems (Minns, 2006). Therefore it is pertinent to discuss the test system characteristics that have the potential to influence the reports of hip protector biomechanical effectiveness. As demonstrated in figure 1.2, for a mass-spring model the total force during an impact is dependent on the effective mass, stiffness and velocity of the body at impact and realistic peak forces for falls from standing can be produced by an infinite combination of these variables. However, a mass-spring model of the hip in-series with a compliant material (e.g. interventions such as soft shell hip protectors or low stiffness floors) predicts significantly different values of force attenuation provided by the intervention across different combinations of input parameters (figure 1.4). Specifically, the attenuation in peak force provided by a compliant hip protector increases with increasing effective body stiffness and impact velocity and decreases with increasing effective body mass. Theoretically then, test systems need to incorporate physiologic values of effective mass, stiffness, and impact velocity if they are to provide meaningful estimates of the biomechanical effectiveness of hip protectors.

The force applied to the proximal femur during an impact also depends on the characteristics of the soft tissues (adipose, muscle, etc.) overlying the hip region. Women who suffer hip fractures have a significantly decreased thickness of soft tissues overlying the greater trochanter compared to healthy controls, even for those with the same BMI (Lauritzen and Askegaard, 1992). During an impact, energy absorption and peak force are positively, and negatively, correlated with soft tissue thickness, respectively (Etheridge et al., 2005, Lauritzen and Askegaard, 1992, Robinovitch et al., 1995b). As

soft tissue thickness is positively correlated with body mass index (Lauritzen and Askegaard, 1992, Maitland et al., 1993) the energy absorption capacity of soft tissues is one explanation for the decreased hip fracture risk for those with higher BMIs (Greenspan et al., 1994, Hayes et al., 1993, Keegan et al., 2004, Wolinsky and Fitzgerald, 1994). In addition, Robinovitch et al. (1995a) report the area directly overlying the greater trochanter to be more than four-fold stiffer than the area 6 cm posterior (35.0 (SD 14.0) kN/m vs. 8.5 (SD 2.9) kN/m, respectively). Although it is not clear how this variability influences trochanteric force, it is clear that soft tissue stiffness and thickness have direct implications for hip protector performance, particularly rigid energy shunting protectors which attempt to form a bridge over the greater trochanter. If the edges of the protector intrude into the soft tissues sufficiently, the hip protector will come in direct contact with the tissues overlying the greater trochanter resulting in compromised energy shunting capacity. As such, it is likely important to incorporate physiologic simulants of soft tissue stiffness and thickness into test systems that evaluate the biomechanical effectiveness of hip protectors.

Table 1.4 presents the characteristics of the test systems used to evaluate the biomechanical effectiveness of hip protectors. Disconcertingly, there is considerable variability in test system characteristics, especially in the values of effective mass, stiffness, and impact velocities. The effects of these differences in test system characteristics can readily be observed by comparing the peak impact force for an impact with no hip protector in place (i.e. unpadding, unprotected) at comparable impact energies (table 1.2). In cases in which the same hip protector has been tested in different test systems, we can also compare whether the biomechanical effectiveness of said hip

protector appears to correspond to test system properties (table 1.3). Overall, the Robinovitch et al. (1995a) test system appears to be the most biofidelic as it incorporates realistic estimates of effective stiffness, mass, impact velocity and soft tissue stiffness and thickness. This results in three possible outcomes for the other test systems: i) peak forces that are much higher than those predicted (Kannus et al., 1999, Parkkari et al., 1995, Wiener et al., 2002); ii) a reduction of effective mass to unrealistic levels in order to produce realistic peak impact forces (Derler et al., 2005, Mills, 1996, Okuizumi et al., 1998); or iii) a reduction in impact velocity to unrealistic levels in order to produce realistic peak forces (van Schoor et al., 2006). As a consequence (and as predicted by a mass-spring model of the hip in series with a compliant hip pad) each of these systems tends to estimate substantially higher biomechanical effectiveness values compared to those reported by Robinovitch et al. (1995a).

Although the above review and simple modelling examples indicate the importance of test system characteristics in evaluating the biomechanical effectiveness of hip protectors, they are not a sufficient data set from which to draw conclusions. In contrast, they serve as a basis upon which research questions and hypotheses can be generated. For example, the above-mentioned model predictions are based on a simplistic energy absorbing hip protector with linear stiffness characteristics. However, it is unclear how isolated changes in variables such as impact velocity will affect the biomechanical effectiveness of more complex hip protectors systems (e.g. those which incorporate rigid energy shunting components, those which do not directly overlie the greater trochanter) or soft, energy absorbing protectors with non-linear stiffness properties. In addition, it is unclear how soft tissue stiffness and surface geometry affect the force attenuation

provided by hip protectors. Unlike effective mass, stiffness, and impact velocity, the effects of these variables are not amenable to investigation with simple mathematical models. These gaps in the literature were examined in the third study of this thesis through a sensitivity analysis of the extent to which impact velocity, soft tissue stiffness, and surface geometry affected the biomechanical effectiveness of hip protectors (described in Chapter 4). This study assists in explaining the large variability in biomechanical effectiveness reports in the literature, and will aid in the development of standards and guidelines on the characteristics that should be incorporated into test systems that evaluate the biomechanical effectiveness of hip protectors.

#### ***1.3.1.2.3 Pressure distribution – another potential measure of biomechanical effectiveness***

Despite efforts to improve the accuracy and reliability of systems that evaluate the biomechanical effectiveness of hip protectors, there are several limitations with current approaches. For example, despite our best efforts, it may not be possible to adequately simulate realistic human characteristics. In addition, test systems usually attempt to emulate the average characteristics of a certain subset of the population. However, there may be marked differences in the hip protector biomechanical effectiveness for individuals who don't exhibit these average characteristics. Consequently, it is important to augment test system based studies of hip protector effectiveness with those performed with living human participants (Parkkari et al., 1997).

In addition, current mechanical test systems have focussed on how hip protectors affect the peak forces and stresses applied to the proximal femur. However, little is known about how the total impact force is distributed throughout the hip region, and whether hip protectors alter this distribution. Minns (2006) observed that most hip

protectors tend to increase the pressure applied to the greater trochanter region during simulation of a static, sideways sleeping position. However, no information is available on pressure distribution throughout the hip during the impact phase of a sideways fall, likely due to the high cost of appropriate pressure arrays as previously mentioned. Presumably, biomechanically effective hip protectors reduce the pressure applied to the proximal femur during impact. However, this has never been proven experimentally. Similarly, it is unknown whether hip protectors alter pressures in other areas of the hip, and whether pressure distributions patterns are dependent on the characteristics of hip protectors. Such information would provide insights into the mechanisms by which hip protectors absorb and/or shunt energy away from the greater trochanter during impacts.

These gaps in the literature were examined in the second study of this thesis described in Chapter 3. I used pelvis release experiments with human subjects to determine how pressure is distributed throughout the hip region during the impact phase of a sideways fall, and whether soft shell hip protectors with different geometric properties alter this distribution. This study provides insights into what portions of total impact force are directed directly into the proximal femur during a fall impact, and the mechanisms by which soft hip protectors attenuate force applied to this region.

### **1.3.2 Compliant floors and impact force attenuation**

One of the main limitations with hip protectors is that their ability to reduce hip fracture risk relies on active user compliance. Strictly from the perspective of intervention intensity removing a user's choice of non-compliance should increase the clinical effectiveness of an intervention. One strategy for achieving this that is particularly relevant to high-risk environments (such as nursing homes, hospitals, or

senior centres) is to reduce the stiffness of the floor. This ‘passive’ intervention requires no decision on the part of the user, and as such, compliance is 100%. Consequently, the clinical effectiveness of low stiffness floors is based solely on their mechanical properties.

There are two lines of evidence to support the potential of low stiffness floors as a means to decrease hip fracture risk. First, epidemiological studies have reported that, when compared to falling on a hard (concrete or linoleum) surface, falling on a soft (padded carpet, grass, or loose dirt) surface creates reduced risk for hip fracture (Nevitt and Cummings, 1993, Simpson et al., 2004). Similarly, Healey (1994) reported that in a hospital for aged persons, falls onto carpets resulted in a significantly lower rate of injury compared to falls onto vinyl floors. Second, laboratory studies have found that low stiffness floors can reduce the peak impact force during the impact phase of simulated sideways fall by up to 73% (Casalena et al., 1998a, Gardner et al., 1998, Maki and Fernie, 1990, Minns et al., 2004c, Nabhani and Bamford, 2002). Specifically, Minns et al. (2004c) used a mechanical drop mass system (see description of the system in table 1.3 under Nabhani and Bamford, 2002) to study the effect of floor surfaces (carpet and vinyl) and various under-padding materials on peak impact force during simulated falls on the hip. They found that, compared to an impact on a rigid surface, peak impact force was decreased by approximately 12.5% for a surface covered with carpet (fabric backed luxury pile with open tufted polyester fibres). Force attenuation increased to 73% when 12.5 mm PVC foam under-padding was added to the carpet. However, the validity of these results is unclear as their fall simulation system did not incorporate the effective stiffness of the body during an impact. Using the same drop mass test system, Nabhani

and Bamford (2002) observed that carpet underlay materials decreased peak impact force by an average of approximately 55% (SD 4%) and that force attenuation was positively correlated with underlay thickness. Casalena et al. (1998a, 1998b) used an impact pendulum to assist in the development of a floor that deflected < 2mm at forces associated with walking, but underwent greater deflection and absorbed more energy via 'collapsible column supports' during the higher forces associated with fall impacts. Their best prototype met their deflection criteria and decreased peak impact force by 15.3%. However, it proved too expensive to be widely implemented. Gardner et al. (1998) developed an impact transducer and mathematical model to study the effect of floor surface on impact forces. They measured the acceleration of the transducer mass during impact, which was used to estimate impact force. They report that a floor surface comprised of pile carpet plus under-padding reduces peak force by 6.7 % compared to a concrete floor. However, they admit that their system may not accurately simulate human values of mass, stiffness, damping and muscle activation, and as such, they state that their system is not intended to predict the absolute values of impact forces. Despite this, they feel their approach adequately determines the relative difference in impact force associated with different floor surface conditions. Finally, Maki and Fernie (1990) developed an impact pendulum to study the effect of various floor surfaces on peak impact force. They measured system deceleration during impact, which was used to calculate the impact forces. Compared to the most rigid floor condition (linoleum on a terrazzo/concrete slab), thick pile carpet plus under-padding reduced peak force by 24.0 %. However, the validity of these results is unclear as the effective stiffness of their

system (540 kN/m) was approximately 10-fold higher than the values later determined from human subjects (Robinovitch et al., 1991, Robinovitch et al., 1997a).

Collectively, these studies were limited in their ability to quantify the relationship between floor stiffness and peak impact force for several reasons. First, they didn't examine a sufficient range of conditions to evaluate the true force attenuation capacity of low stiffness floors. Second, they used mechanical systems to simulate falls, which may exhibit impact dynamics quite different from human subjects. Finally, most (Gardner et al., 1998, Maki and Fernie, 1990, Nabhani and Bamford, 2002) did not specify the stiffness of the floor materials that were studied, which limits the ability of others to duplicate their experimental conditions.

These limitations were addressed in the fourth and fifth studies of this thesis (described in Chapters 5 and 6) in which I used pelvis release experiments with human subjects to determine how impact force is affected by floor stiffness and impact velocity. In addition, I used a mechanical hip impact simulator to examine the effects of floor stiffness during falls from standing height. These studies provide us with insights into the force attenuation we can expect to attain through the use of low stiffness floors, which will aid engineers and architects in the design of low stiffness floors for 'safe movement environments' in high risk areas such as hospitals, nursing homes, residential housing, and rehabilitation gymnasiums.

#### **1.4 Compliant floors and fall risk**

For low stiffness floors to result in decreased hip fracture their properties must not substantially increase the risk of falls during a variety of activities of daily living. It is

possible that low stiffness floors might impair stability by decreasing the quality of information obtained through somatosensory feedback systems (Betker et al., 2005, Lord and Menz, 2000, Ring et al., 1989). For example, foam rubber floor materials could degrade the quality of proprioceptive information sensed by spindle organs in the muscles spanning the ankle, due to a reduction in ankle rotation accompanying sway (caused by the foot depressing into the foam). In addition, foam rubber floors might reduce the variation in pressure and thereby alter signals from the mechanoreceptors in the sole of the foot. Low stiffness floors might also impair balance through mechanical means. During stance, foam rubber floors may compress in a non-uniform manner corresponding to the varied distribution in pressure under foot. These varied deflections may decrease stability and increase the chance of ankle sprains due to excessive inversion or eversion. Similarly, trip risk may be increased if toe clearance is reduced during the swing phase due to compression of the floor materials. Despite these potential drawbacks, common low stiffness floors such as carpet have never been implicated as a factor that increases fall risk. However, it seems evident that, at some point, lowering floor stiffness will impair balance maintenance and / or recovery abilities and increase the risk of falls (imagine trying to maintain balance while performing activities of daily living in the inflatable jumping rooms often found in amusement parks). What is unknown is the specific floor stiffness threshold at which balance becomes substantially impaired during a range of static and dynamic activities.

Two important aspects of a study aiming to determine these floor stiffness thresholds are the conditions, and tasks, under which the effect of floor stiffness on balance is examined. Chiu et al. (2003) state that a comprehensive test should evaluate

balance maintenance, postural adjustment to voluntary movements, and ability to respond to external perturbations. Furthermore, for an investigation to provide maximum insight into fall risk, the tasks chosen should represent those for which fall incidence is relatively high and / or those which are significantly associated with fall risk. An assessment strategy that satisfies both these criteria will provide insight into both the potential clinical applications of low stiffness floors, and the physiologic mechanisms by which elderly persons maintain balance in the face of changes in floor stiffness.

Up to 50% of falls occur during the act of walking (Greenspan et al., 1994, Lee and Kim, 1997, Nevitt and Cummings, 1993). As such, it is relevant to evaluate the effect of floor stiffness on balance during locomotion (Redfern et al., 1997). Decreased floor stiffness was initially reported to increase step length during running (McMahon and Greene, 1979). However, recent studies suggest little association between these variables as runners maintain global mechanics (e.g. stride length and frequency, peak vertical ground reaction force) in response to expected changes in floor stiffness by immediately altering their leg stiffness (Ferris et al., 1998, Kerdok et al., 2002). One step onto an unexpected low stiffness surface results in increased vertical centre of mass displacement and trunk pitch, suggesting decreased trunk stability (Marigold and Patla, 2005). However, muscle activity is modulated during recovery to ensure that toe clearance during the subsequent swing phase is not diminished. When a change in floor stiffness is expected, stepping onto and walking on a compliant foam surface results in increased step width and length, and increased step width variability (MacLellan and Patla, 2006). These studies indicate that both floor stiffness and prior knowledge of impending changes in floor stiffness are factors that might moderate balance control during locomotion.

Only about 2-4% of non-syncopal falls occur during the task of standing still (Lee and Kim, 1997, Nevitt and Cummings, 1993). However, increased amplitude of postural sway during quiet stance is associated with increased fall risk (Campbell et al., 1989, Maki et al., 1994, Thapa et al., 1996), and probably associates with sway during other activities. Therefore, quiet stance is a relevant activity for which to evaluate the effect of floor stiffness on balance. Compared to a rigid floor condition, postural sway during quiet stance has been demonstrated to increase when standing on an extremely compliant foam surface (Gill et al., 2001, Lord et al., 1991, Lord and Menz, 2000, Ring et al., 1989, Teasdale et al., 1991). These observations do not apply to common low stiffness floors such as carpets, which under normal sensory conditions, have never been associated with increased sway (Dickinson et al., 2002, Dickinson et al., 2001, Redfern et al., 1997). However, when vision is altered and / or the floor surface is translated, carpeted floors have been associated with increased sway in elderly persons (Dickinson et al., 2001, Redfern et al., 1997). Redfern et al. (1997) are the only group to have evaluated sway over a number of carpet and under-pad conditions. This approach needs to be expanded to consider a larger range of floor stiffness conditions to identify the specific threshold below which sway is significantly increased during quiet stance.

Finally, risk of falls is associated with a person's ability to transfer from quasi-static to dynamic situations. For example, Topper et al. (1993) observed that elderly persons with low scores in activity-based clinical tests of transferring, turning, and reaching were more likely to suffer falls during these tasks in every day life compared to elderly persons with normal scores. The "Timed Up & Go Test" combines several transitional activities (transferring from sitting to standing, gait initiation, turning, and

transfer from standing to sitting) into a single assessment tool (Mathias et al., 1986, Podsiadlo and Richardson, 1991). The time taken to complete the test associates with fall risk (Chiu et al., 2003, Lundin-Olsson et al., 1998, Okumiya et al., 1998, Podsiadlo and Richardson, 1991, Shumway-Cook et al., 2000) and identifies persons who require assistance with functional mobility tasks (chair and toilet transfers, tub or shower transfers, walking aid, stair climbing) (Podsiadlo and Richardson, 1991). However, the relationship between floor stiffness and balance and stability during transitional tasks has never been quantified.

These gaps in the literature were examined in the fifth study of my thesis, described in Chapter 6. Elderly women engaged in common static and dynamic activities (transferring, walking, quiet stance, balance recovery during a translating floor perturbation) while floor stiffness was varied. Markers of balance were measured to identify the lower boundary of floor stiffness beyond which stability was substantially impaired. This study provided general insights into how floor stiffness affects balance maintenance and stability in elderly women at highest risk of falls. In addition, the range of activities examined allowed me to determine whether floor stiffness had a differential influence on static and dynamic measures of stability (Mackey and Robinovitch, 2005, Maki et al., 1990, Owings et al., 2000). In combination with study four (the effect of floor stiffness on impact force), these results provide crucial information which can assist engineers and architects in optimizing compliant flooring's potentially competing demands of force attenuation and balance maintenance and recovery in efforts to decrease hip fracture risk.

## **1.5 Thesis objective and summary of studies**

The general objective of my thesis was to characterize the stiffness and force distribution characteristics of the hip region during unpadded sideways falls, and to advance our understanding of the potential for external engineering interventions (e.g. hip protectors and compliant flooring) to reduce hip fracture risk by reducing the magnitude of force applied to the hip during the impact phase of a fall.

To achieve this objective I employed a combination of experimental and mathematical modelling approaches. Specifically, for facets of the research program investigating impact dynamics, healthy women engaged in ‘pelvis release experiments’ to simulate the impact stage of a sideways fall on the hip. For fall heights that were unsafe for study in human participants, I used mathematical models and/or a mechanical fall simulator constructed to match the characteristics of elderly females at high risk of hip fracture. In all studies I considered a worst case scenario in which protective responses were insufficient to avoid direct impact between the floor and the lateral aspect of the hip (Nevitt and Cummings, 1993). For aspects of the research program studying balance I used measures of natural sway, task success, and participant ratings to study the effects of floor stiffness on fall risk.

This thesis uses a manuscript-based format to describe five studies I completed as part of my doctoral research. In the first I characterized the degree of non-linearity in pelvic stiffness, and examined the influence of stiffness characterization methods on the accuracy of mathematical models in predicting impact dynamics during falls on the hip. In the second study I employed a pelvis release paradigm to further our understanding of the potential for hip protectors to alter the distribution of force throughout the hip region

during the impact phase of sideways falls. The third study entailed a sensitivity analysis to determine the extent to which the characteristics of mechanical test systems (specifically impact velocity, pelvis size, and soft tissue stiffness) affect estimates of the biomechanical effectiveness of hip protectors. In the fourth study I used pelvis release experiments to determine how attenuation in peak force is affected by impact configuration and floor stiffness as modified by changing the thickness of foam-rubber floor mats, and also examined the ability of a mass-spring model to predict these relationships. The final study used a mechanical fall simulator to assess the attenuation in femoral neck force provided by four low stiffness floors compared to a standard rigid floor, and assessed the influence of these floors on fall risk through a range of static and dynamic balance tests with fifteen elderly women. Each of these studies is presented in detail in the following chapters.

**Table 1-1: Review of clinical risk factors for hip fracture in women (Cumming et al. 1997). \* represents factors that might operate through bone strength.**

Risk factor	Comparison
Established risk factors for hip fracture in women	
<b>Relative risk <math>\geq 2.0</math></b>	
BMD of the femur *	Per 1 SD decrease
Fall on hip	Fall on hip vs. fall on other part of the body
Neuromuscular impairment *	Unable vs. able to rise from chair without using arms
Race	Whites vs. Asians (in Asia) or whites vs. blacks
Ultrasound attenuation *	Per 1 SD decrease
<b>Relative risk 1.0 to 1.9</b>	
Age	Per 5 year increase
BMD at non-femoral sites *	Per 1 SD decrease
Falls in past year	$\geq 2$ falls vs. 0 falls
Hormone replacement therapy *	Current users vs. users
Neuromuscular impairment	
Grip strength *	Per 5 kg decrease
Gait speed	Per 0.2 m/s decrease
Weight *	Per 10 kg decrease
Factors found to be risk factors in several studies, but some uncertainty still exists	
<b>Relative risk <math>\geq 2.0</math></b>	
Health status	Poor/fair vs. good/excellent
Parkinson's disease	Present vs. absent
Race	Whites vs. Hispanics
Weight change *	20% weight loss
<b>Relative risk 1.0 to 1.9</b>	
Coffee	$> 2$ cups/day vs. no coffee
Calcium intake *	$< 400$ mg/day vs. $> 1000$ mg/day
Fractures since age 50 *	Fractures vs. no fractures
Height	Per 10 cm increase
Hip axis length	Per 1 SD increase
Psychotropic drugs	Users vs. non-users
Smoking	Smokers vs. non-smokers
Thiazides	Non-users vs. users
Vision	Acuity worse than 20/40
Physical activity *	Not walking vs. walking for exercise

**Table 1-2: Peak force (N) in unpadded and hip protector conditions as measured by mechanical test systems that evaluate the biomechanical effectiveness of hip protectors.**

	Robinovitch et al. 1995	Parkkari et al. 1996	Mills 1996	Okizumi et al. 1998	Kannus et al. 1999	Nabhani & Bamford 2002	Wiener et al. 2002	van Schoor et al. 2006	Derfer et al. 2005					
impact energy (J)	120	78	132	24.1	20.6	41	74	110	120	52.9	82	19.6	19.6	49.1
condition												1" tissue	0.5" tissue	
unpadded	5810	5590	10400	3910	3117	3740	6130	9190	4300	13372	17102	3998	6378	5654
implant pad (Proshok)	4530													
HipGuard	4200													
Strock pad	4800													
Wortberg pad	4500													
Safety Pants (FI)	3900					790	2760	5770				2330	5186	
FICSIT pad	4530													
Sahva	4750		880						700			1298	2061	
dilatant horseshoe	2050													
Hornsby Healthy Hip														
KPH1		1040	1810			590	780	1360						
KPH2						510	760	1170				804	900	
Safehip						1080	2240	4640	820			911	1817	
Impactwear Garment												971	2105	
Gerihip												1957	4948	
Hipsaver												1689	3472	
Lyds												1984	4423	
Safety Pants (NL)												1520	3415	

	Robinovitch et al. 1995	Parkkari et al. 1996	Mills 1996	Okizumi et al. 1998	Kannus et al. 1999	Nabhani & Bamford 2002	Wiener et al. 2002	van Schoor et al. 2006	Derler et al. 2005						
impact energy (J)	120	78	132	24.1	20.6	41	74	110	120	52.9	82	19.6	19.6	49.1	
condition												1"	0.5"	tissue tissue	
N & B self adhesive															
N & B crude															
OHK gel + cover					1681										
OHK gel					2176										
WANC firm														1197	2160
WANC soft														1234	2603

**Table 1-3: Summary of the literature reports of force attenuation provided by hip protectors (% reduction in peak force compared to the unpadded condition). Within column comparisons indicate the effect of hip protector design characteristics. Across column comparisons indicate the effect of test system characteristics.**

	Robinovitch et al. 1995	Parkkari et al. 1996	Mills 1996	Okizumi et al. 1998	Kannus et al. 1999	Nabhani & Barnford 2002	Wiener et al. 2002	van Schooret al. 2006					
impact energy (J)	120	78	132	24.1	20.6	41	74	110	120	52.9	82	19.6	19.6
condition												1"	0.5"
implant pad (Prosthek)	22.0											tissue	tissue
HipGuard	27.7												
Strock pad	17.4												
Wortberg pad	22.5												
Safety Pants (FI)	32.9				78.9	55.0	37.2					41.7	18.7
FICSIT pad	22.0								83.7			67.5	67.7
Sahva	18.2												
dilatant horseshoe	64.7												
Hornsby Healthy Hip												78.6	86.5
KPH1		81.4	82.6		84.2	87.3	85.2					79.9	85.9
KPH2					86.4	87.6	87.3					77.2	71.5
Safehip					71.1	63.5	49.5		80.9			75.7	67.0
Impactwear Garment												51.1	22.4
Gerhip												57.8	45.6
Hipsaver												50.4	30.7
Lyds												62.0	46.5
Safety Pants (NL)													



**Table 1-4: Characteristics of test systems that measure the biomechanical effectiveness of hip protectors. The Robinovitch et al. (1995) test system incorporates physiologic estimates of effective stiffness, mass, impact velocity and soft tissue stiffness and thickness.**

	Robinovitch et al. 1995	Parkkari et al. 1996	Mills 1996	Okizumi et al. 1998	Kannus et al. 1999	Nabhani & Bamford 2002	Wiener et al. 2002	Derler et al. 2005	van Schoor et al. 2006
impact velocity (m/s)	2.6	2.0, 2.6, 3.7	1.1 - 3.3	1.0 - 3.1 in 0.4 increments	1.4, 1.9, 2.3	3.12	3.43, 4.27	3.1	1.25
effective mass (kg)	35	50.7*	12.3	8.4	52.0**	24	9.0	5, 10, 15	25
impact energy (J)	120	78, 132, 272	8 - 66	4.1 - 41.2 in 4.1 increments	41, 74, 110	120.0	52.9, 82.0	24.5, 49.1, 74.6	19.6
effective stiffness (kN/m)	24 to 78 (low vs. high force)	75	NI	NI	75	NI	NI	NI	NI
soft tissue stiffness (kN/m)	8.5 to 35	NR	70	NI	NR	NR	NR	NR	NR
soft tissue thickness (mm)		25	10 or 20	NI	20	5	5	20	12.7 or 24.5

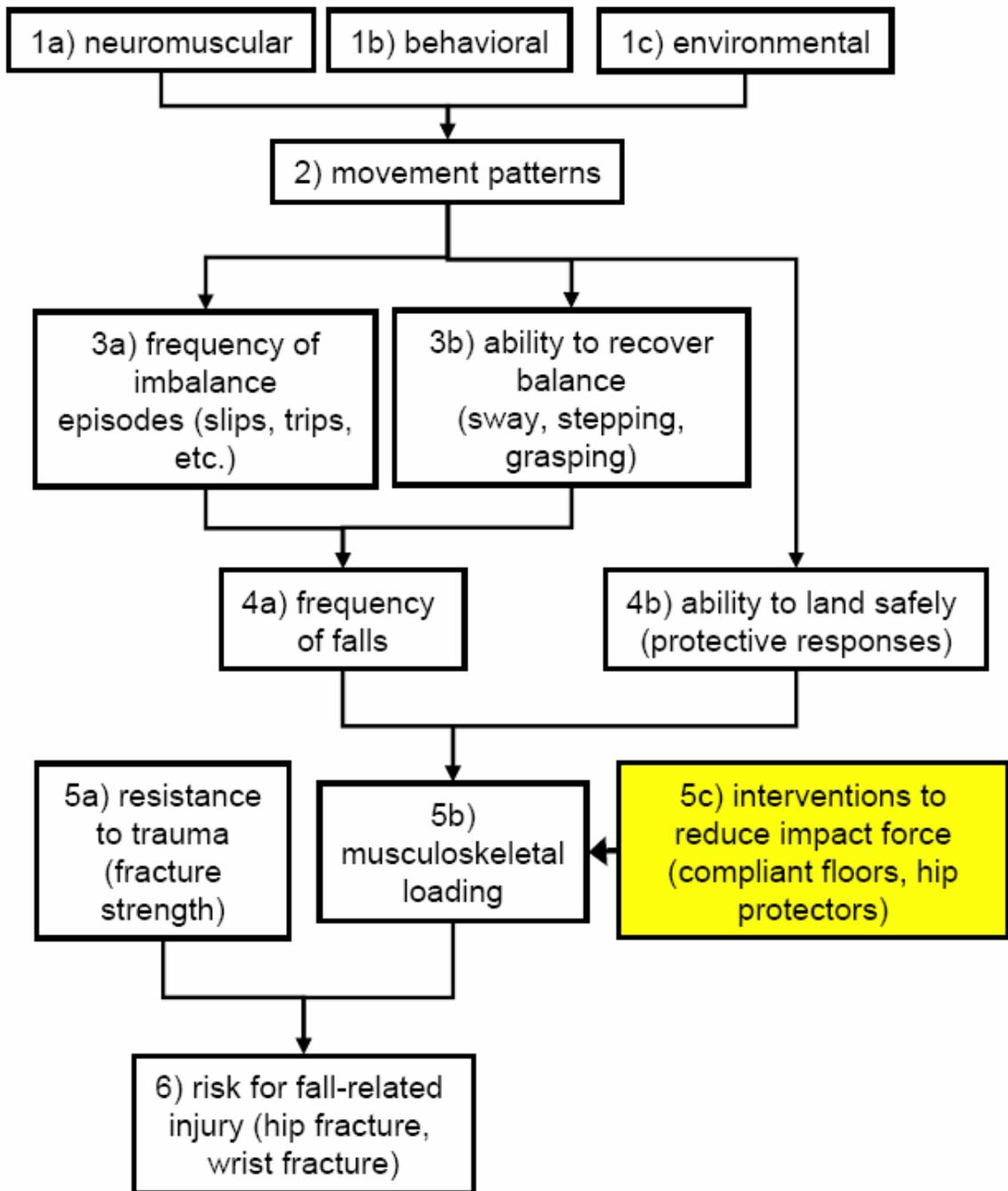
\* total effective mass comprised of 39kg impact pendulum and an 11.7 kg surrogate pelvis.

\*\* total effective mass comprised of 40.3 kg impact pendulum and an 11.7 kg surrogate pelvis.

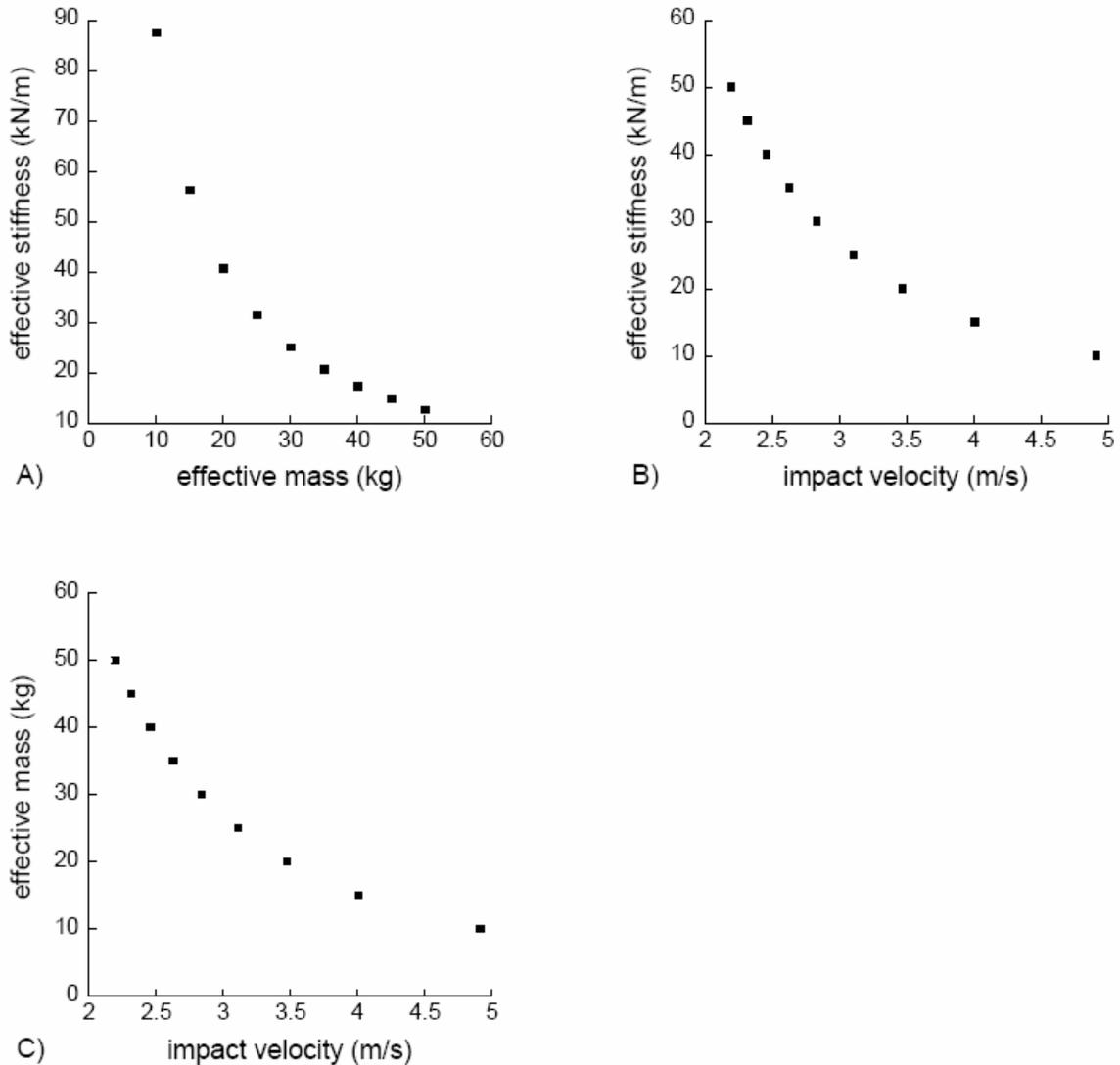
NI indicates parameter not incorporated into test system and thus value not reported.

NR indicates value not reported.

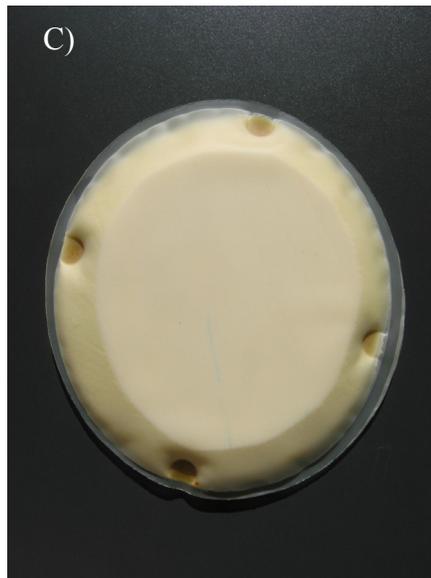
Figure 1-1: Conceptual model of the factors that influence risk for hip fracture.



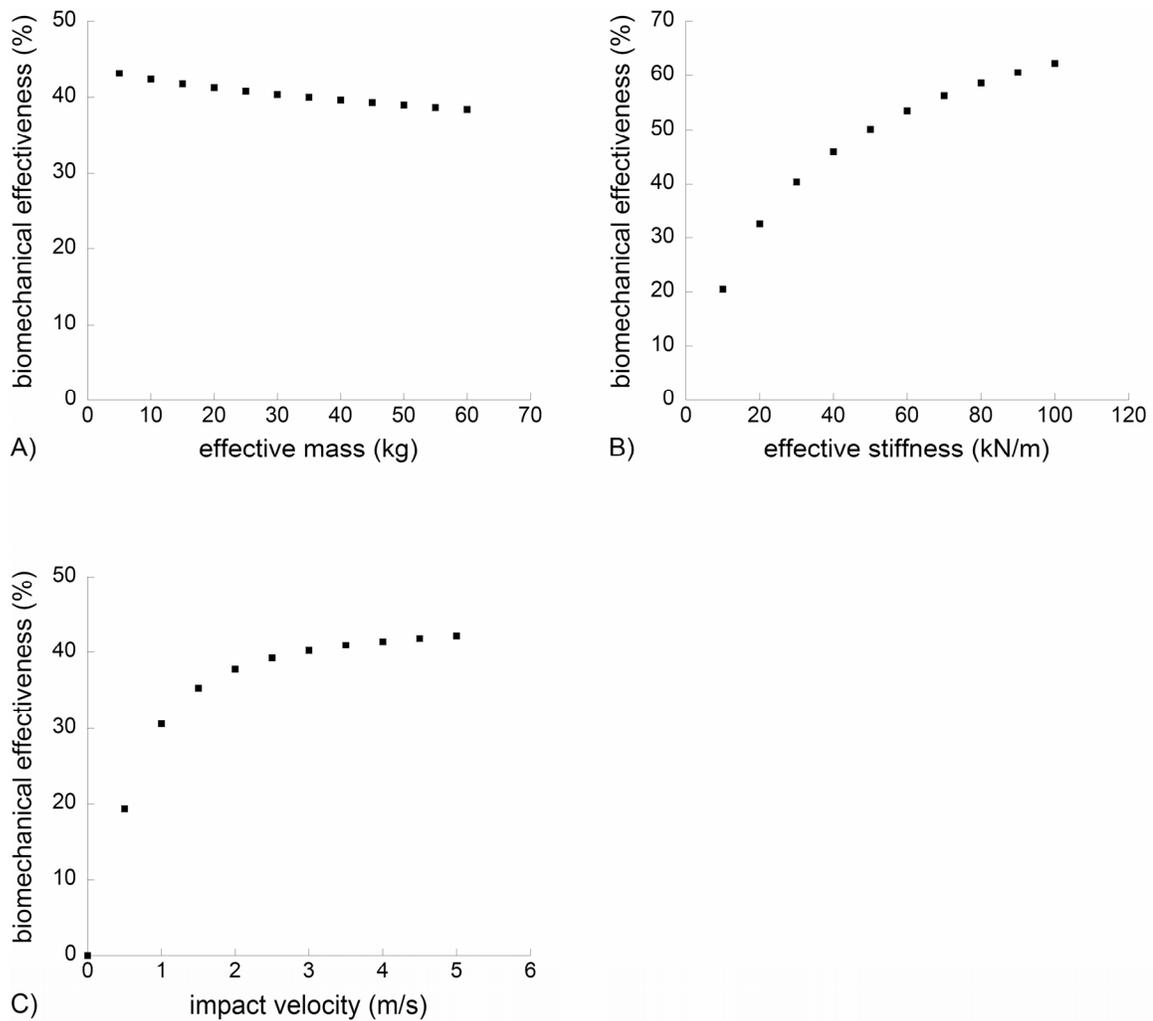
**Figure 1-2: Demonstration of the complex manner in which mass-spring model parameters can interact. McMahon et al. (1987) showed that the equations describing the time to peak force ( $t_{max}$ ) and magnitude of peak force ( $F_{max}$ ) are given by:  $t_{max} = 1/\omega_n (\pi - \tan^{-1}(u\omega_n/g))$  and  $F_{max} = mg((u\omega_n/g)\sin(\omega_n t_{max}) + 1 - \cos(\omega_n t_{max}))$  where  $u$  is the impact velocity (in m/s),  $m$  is the effective mass of the body (in kg),  $\omega_n$  is the natural frequency (in rad/s), and  $k$  is the effective stiffness of the body (in N/m). There exist an infinite combination of  $u$ ,  $m$ , and  $k$  that result in an identical peak force of 3000 N. A)  $k$  vs.  $m$  with  $u = 3.1$  m/s; B)  $k$  vs.  $u$  with  $m = 30$  kg; C)  $m$  vs.  $u$  with  $k = 30$  kN/m.**



**Figure 1-3: Examples of hip protectors. A) Hard shell. B and C) Soft shell.**



**Figure 1-4: Demonstration of the influence of mass-spring model parameters on the peak force attenuation (biomechanical effectiveness) provided by a compliant hip protector.  $m$  and  $k_b$  are the effective mass and stiffness of the body, respectively, and  $u$  is the impact velocity. Total effective stiffness  $k$  is calculated as the series combination of the hip protector spring ( $k_f = 13$  kN/m) and the effective spring stiffness of the body ( $k_b$ ) as  $k = \frac{k_b k_f}{k_b + k_f}$ . Biomechanical effectiveness vs. A)  $m$  with  $k_b = 30$  kN/m and  $u = 3.1$  m/s; B)  $k_b$  with  $m = 30$  kg and  $u = 3.1$  m/s; C)  $u$  with  $k_b = 30$  kN/m and  $m = 30$  kg. The biomechanical effectiveness of a low stiffness hip protector decreases with increasing effective mass and increases with increasing effective stiffness of body and impact velocity.**



## **CHAPTER 2 CHARACTERIZING THE STIFFNESS OF THE PELVIS AND PREDICTING IMPACT DYNAMICS DURING SIDEWAYS FALLS ON THE HIP**

### **2.1 Introduction**

Falls onto the hip, more specifically the greater trochanter, are the event most directly linked to hip fractures in the elderly (Grisso et al., 1991, Nevitt and Cummings, 1993, Zuckerman, 1996). As such, falls provide the necessary circumstances during which the force applied to the proximal femur exceeds the fracture threshold. A requisite for the prediction of hip fracture risk during falls is knowledge of the magnitude of these applied forces (Hayes et al., 1991). As safety concerns preclude the in-vivo measurement of these forces during standing height falls, researchers must consider other means of addressing this research question.

An alternative to experimental measures is the use of appropriately tuned mathematical models to predict impact dynamics during sideways falls on the hip. Robinovitch et al. (1997a) used a Voigt model (a mass supported by parallel spring and damper elements) to predict peak forces ranging from 1150 to 5288 N for a sample of healthy young participants. Using a mass-spring model, van den Kroonenberg et al. (1995) predicted peak forces of 2900, 3580, and 4260 N for 5, 50, and 95 percentile females during sideways falls from standing height. They concluded that these forces are well below the average femoral fracture threshold of young adults, but are sufficient to result in hip fracture in about 50% of elderly persons.

An important consideration when selecting the most appropriate model of impact is the influence of damping on the force applied to the hip during a sideways fall. During low energy falls in healthy participants, Robinovitch et al. (1991; 1997a) observed the body to be underdamped resulting in decaying oscillations in force over time. The damping was non-linear and force-dependent, best described by an exponential function in which damping increased with applied force before plateauing above a threshold of approximately 300 N. However, Robinovitch et al. (1997b) later demonstrated that inclusion of damping in mathematical models did not improve the accuracy of the predictions of peak force during standing-height falls simulated with a mechanical test system. Specifically, a Voigt model underpredicted peak force by an average (SD) of 10.2 (13.3) % compared to average mass-spring model accuracy of 2.2 (14.0) %. Furthermore, a Voigt model predicts an instantaneous force for non-zero impact velocities due the model's velocity-dependent damping element. Thus, while damping is required to adequately simulate the decaying oscillation in applied force in the latter stages of impact (greater than ~100 ms), it does not improve the quality of model predictions in the first half period of oscillation where applied force (and thus risk of fracture) is highest. These results support the earlier work of van den Kroonenberg et al. (1995) and form the basis of my decision in the current study to model impact force during sideways falls with a mass-spring model.

An inevitability of mathematical models is that the accuracy of their outputs is highly dependent on the accuracy of their inputs and / or input interactions. Consequently, it is important to critically evaluate the methods by which previous research groups calculated their model parameters. Based on free-fall dynamics and the

principle of energy conservation, Robinovitch et al. (1991) originally estimated a hip impact velocity of 3.7 m/s for a fall from standing height. This was later supported by van den Kroonenberg et al. (1996) who measured values during self-initiated sideways falls onto a foam mat to average 3.2 m/s (range of 2.1 to 4.8 m/s), and by Feldman and Robinovitch (2007) who observed a mean hip impact velocity of 3.0 (SD 1.0) m/s during sideways falls initiated by an unexpected floor surface translation. The remaining model parameters were estimated by Robinovitch et al. (1991, 1997a) using data from pelvis release experiments. Effective mass refers to the portion of total body mass that has a non-zero velocity during impact, and was reported to be approximately 50% of total body mass (Robinovitch et al., 1997a). Effective stiffness was calculated based on the free vibration response of ground reaction force during impact. It initially increased with increasing applied force, but plateaued at forces exceeding approximately 300 N (Robinovitch et al., 1991). This observation was used to justify the use of linear (constant) effective stiffness and/or damping in models employed by van den Kroonenberg et al. (1995) and Robinovitch et al. (1997a).

However, the validity of using a linear estimate of effective stiffness can be questioned from several perspectives. First, the peak forces in the experimental validation trials were approximately one order of magnitude lower than the model predictions of peak force for standing height falls (Robinovitch et al., 1991, 1997a). Second, Mills (1996, 2007) reports that the force-deflection behaviour of soft tissues in the hip region is non-linear, with stiffness increasing as applied force increases (which corresponds to my observations in Chapter 4). Third, despite cadaveric studies which report pelvic compressions of up to 10 cm during high-energy impacts without fracture (Cavanaugh et

al., 1990, Etheridge et al., 2005, Viano et al., 1989), some researchers suggest that the linear effective stiffness values used in previous modelling efforts would result in unreasonably large pelvic deformation during falls from standing height (Derler et al., 2005, van Schoor et al., 2006). Collectively, these issues cast some uncertainty on the accuracy of previous estimates of pelvic stiffness. An alternative approach to the vibration response method of parameter identification would be to directly measure the force-deflection characteristics of the pelvis during impact. Unfortunately, this technique has never been employed in living human subjects.

My objectives for this study were two-fold. First, I aimed to determine whether the in-vivo force-deflection properties of the hip during the impact phase of safe, sideways falls have significant non-linear components. My second goal was to examine whether the accuracies of mathematical model predictions of impact dynamics were influenced by the method used to estimate effective pelvic stiffness. These results will provide important insights into best methods for optimizing parameter identification methods to improve the accuracy of model predictions of impact dynamics during falls on the hip. In addition, this data will provide essential information for the design of mechanical test systems that assess the force attenuation provided by protective devices including hip protectors and low stiffness floors (Derler et al., 2005, Gardner et al., 1998, Kannus et al., 1999, Mills, 1996, Minns et al., 2004a, Nabhani and Bamford, 2002, 2004, Robinovitch et al., 1995a, van Schoor et al., 2006).

## **2.2 Methods**

### **2.2.1 Participants**

Study participants consisted of fifteen women ranging in age from 20 to 27 years (mean =  $22.9 \pm 2.4$  (SD) yrs), in body mass from 49.1 to 67.3 kg (mean =  $56.7 \pm 5.0$  kg), and in height from 1.55 to 1.71 m (mean =  $1.65 \pm 0.05$  m). I recruited subjects through flyers posted at the University. Respondents were interviewed by phone or in person to ensure they were in good general health, and free of musculoskeletal conditions (such as recent sprains, strains, or fractures) that would affect their ability to perform the experiment. All participants provided written informed consent and the study was approved by the Committee on Research Ethics at Simon Fraser University.

### **2.2.2 Experimental protocol**

The experimental protocol (previously termed “pelvis release experiments” (Robinovitch et al., 1991, 1997a) involved using a sling and electromagnet to raise and suddenly release the subject’s pelvis, simulating the impact stage of a sideways fall on the hip (figure 2.1). To conduct a trial, I first positioned the subject in a ‘lying’ sideways fall impact configuration on her left side, with her left shoulder flexed overhead, hips flexed at 45 degrees, and knees flexed at 75 deg. Surgical positioning mats (Olympic Vac Pac models 30 and 31, Olympic Medical Corp., Seattle, WA, USA) were used to ensure consistent postures between trials. The subject wore tight-fitting cycling shorts and a tank top during the trials.

The subject’s pelvis was cradled in a nylon sling with the inferior and superior borders contacting the upper thigh and iliac crest, respectively. A steel cable, the length of which could be adjusted via a turnbuckle, was connected at one end to the sling and at

the other end to an electromagnet (model DCA 600-110I; Automatic Equipment Corp., Cincinnati, OH, USA). I used the turnbuckle to raise the pelvis so a gap of 0, 1.25, or 5 cm existed between the ground and the skin overlying the greater trochanter. In the 0 cm or ‘impending impact’ condition, the pelvis just barely touched the floor as evidenced by minimal resistance to the movement of a piece of paper between the floor and pelvis. In the 1.25 and 5 cm conditions drop height was measured with wooden blocks. I then instructed the subject to completely relax her muscles, and to stay relaxed during impact. Once the participant confirmed that she was “ready,” I released the electromagnet (90% decay time = 15 ms), causing the subject to fall onto the landing surface. Throughout the trial, the lateral aspect of the shin and shoulder remained resting on the surgical positioning mats adjacent to the force plate; only the pelvis was lifted and released.

During each trial the time-varying force applied to the hip region during impact was measured with a force plate (model 6090-15; Bertec Corp., Columbus, OH, USA) sampled at 960 Hz. I also measured body segment movements with a 240 Hz, eight-camera motion measurement system (Motion Analysis Corporation, Santa Rosa, CA) that recorded the position of nine markers attached to various anatomical landmarks (front head, top head, back head, right shoulder, sternum, right anterior superior iliac spine, right greater trochanter, right knee, right ankle) and 5 markers attached to top of the impact surface. Markers were secured with standard double-sided tape. With proper calibration the system has a spatial accuracy of within 1 mm.

Each subject participated in four trials at the 0 (impending impact), 1.25, and 5 cm drop heights resulting in a total of 12 trials. While the four trials for a given condition

were acquired consecutively, I randomized the order of presentation of the different drop heights.

### 2.2.3 Data analysis

In all trials, the measured variation in impact force was dominated by a single natural frequency (figure 2.2). This allowed for easy detection of the time and magnitude (through a customized MATLAB routine) of the following four points on the force versus time trace: point of impact ( $T_{imp}, F_{imp}$ ), peak force ( $T_{max}, F_{max}$ ), first minimum force ( $T_{min}, F_{min}$ ), and final resting force ( $F_m$ ).

#### 2.2.3.1 Model parameter identification

Using the above data points from each trial in the 0 cm drop height (impending impact) condition, I used free vibration response methods to estimate parameter values for a mass-spring model of oscillation. Effective mass ( $m$ ) and period of oscillation ( $T$ ) were calculated as:

$$m = F_m / g \quad (2.1)$$

and

$$T = 2 * (T_{min} - T_{max}). \quad (2.2)$$

I calculated effective compressive stiffness ( $k_{vibe}$ ) as

$$k_{vibe} = \omega_n^2 m \quad (2.3)$$

where the natural frequency ( $\omega_n$ ) for equation (2.3) was calculated as:

$$\omega_n = 2\pi / T. \quad (2.4)$$

### 2.2.3.2 Force-deflection characteristics of the hip

I used the position of the right greater trochanter marker as a measure of pelvic deflection during impact. The start of impact was defined as the frame immediately preceding the instant when applied force exceeded 2 N. The initial compressive phase was defined as start of impact to peak force. Pelvic width was defined as the distance between the right greater trochanter marker and the floor surface marker throughout the impact. Pelvic compression was calculated as the change in pelvic width compared to the baseline value recorded at the start of impact. The data from all four trials at a given drop height condition were combined to increase the number of points in each data set. For each drop height, I created scatterplots of applied force ( $F$ ) versus pelvis deflection ( $x$ ) and generated best-fit curves consisting of first and second order polynomial functions using a least squares regression approach (figure 2.3 A & B). Both functions were forced through an intercept of zero, while the second order fit was constrained to positive coefficients to ensure that negative stiffness values did not exist at small deflections. The functions were differentiated to determine effective stiffness ( $k_{1st}$  and  $k_{2nd}$ ) as follows:

$$1^{st} \text{ order: } F = A * x \quad (2.5)$$

$$k_{1st} = A \quad (2.6)$$

$$2^{nd} \text{ order: } F = B * x^2 + A * x \quad (2.7)$$

$$k_{2nd} = 2 * B * x + A \quad (2.8)$$

### 2.2.4 Model predictions of impact dynamics

I performed model simulations to predict the compressive forces generated at the hip for all three drop heights. A preliminary analysis provided evidence of the limited

influence of damping on peak force during sideways falls, as peak force and peak deflection tended to occur at similar times (figure 2.4). The trend was observed at all three drop height conditions. These findings concur with Robinovitch et al.'s (1997b) conclusion that peak force during sideways falls is dominated by elastic (rather than velocity-dependent) mechanisms, and further supported my decision to use a mass-spring model to simulate impact dynamics.

I used subject specific parameter values of effective mass and stiffness calculated from the impending impact condition as model inputs. Initial condition for mass position was defined as  $x(0) = 0$  m. The greater trochanter position data was differentiated to provide velocity, and impact velocity was defined as  $\dot{x}(0) = \dot{x}(T_{imp})$  m/s. For my mass-spring model, I used MATLAB to solve the equation of motion

$$m\ddot{x} + kx = mg \quad (2.9)$$

for  $x(t)$ , and compressive force was given by

$$F = kx . \quad (2.10)$$

I performed model simulations using each of the three estimates of effective body stiffness ( $k_{vibe}$ ,  $k_{1st}$  and  $k_{2nd}$ ). Based on Robinovitch et al.'s (1991) observation that pelvic stiffness plateaued at forces greater than 300N, I also ran model simulations using a combination approach ( $k_{combo}$ ) where effective stiffness was equal to  $k_{2nd}$  for the low force range (0 to 300 N) and to the instantaneous value at 300 N for all points above this transition force (figure 2.3 C).

For all models I calculated the average % difference between the experimental values and model predictions of peak force ( $F_{dif}$ ) and time to peak force ( $T_{dif}$ ) at the 0, 1.25, and 5 cm drop heights.

### **2.2.5 Statistics**

Due to a technical problem I was unable to collect kinematic data from one participant. Consequently, this participant was excluded from the analyses, resulting in a final sample size of fourteen. I used paired t-tests ( $\alpha = 0.05$ ) to test for significant differences between linear effective stiffness values estimated via the free vibration response ( $k_{vibe}$ ) and force-deflection ( $k_{1st}$ ) methods. The quality of the matches between the force-deflection data and the 1<sup>st</sup> and 2<sup>nd</sup> order curve fits were evaluated via the coefficient of determination ( $r^2$ ). I *a priori* chose an  $r^2$  increase of 0.1 (translating to a 10% increase in the proportion of the variance in force explained by deflection) to signify a substantially improved fit. I used two-factor repeated measures analysis of variance (ANOVA) to test the effect of stiffness calculation method ( $k_{vibe}$ ,  $k_{1st}$ ,  $k_{2nd}$ ,  $k_{combo}$ ) and drop height (0, 1.25, and 5 cm drop heights) on  $F_{dif}$  and  $T_{dif}$ . When ANOVA results indicated a significant effect, I used paired t-tests to test for differences between individual conditions. All statistical analyses were performed with a software package using an  $\alpha$  of 0.05 (SPSS Version 16.0, SPSS Inc., Chicago, IL, USA).

## **2.3 Results**

### **2.3.1 Effective stiffness data**

I observed good correspondence between the linear effective stiffnesses estimated via the free vibration response and 1<sup>st</sup> order polynomial fit to the force-deflection data

(table 2.1). Specifically, there was no significant difference between  $k_{vibe}$  (mean (SD) = 23696 (7165) N/m) and  $k_{1st}$  (20922 (3482) N/m) ( $t(13) = -1.217, p = 0.245$ ). In contrast, the linear portion of  $k_{combo}$  at forces above 300 N (31784 (10266) N/m) was significantly higher than  $k_{vibe}$  ( $t(13) = 2.219, p = 0.045$ ) and  $k_{1st}$  ( $t(13) = 4.577, p = 0.001$ ).

The force-deflection properties of the pelvis demonstrated slight non-linearity characterized by increasing stiffness as deflection increased (figure 2.3). Averaging function coefficients across all subjects, the best-fit 2<sup>nd</sup> order polynomial for the impending impact condition could be described as  $F = 800478 * x^2 + 9306 * x$ . The associated function for effective stiffness was  $k_{2nd} = 1600956 * x + 9306$ , which produced  $k_{2nd}$  values of 10907, 25316, 41325, and 89355 at deflections of 0.001, 0.01, 0.02 and 0.05 m, respectively. Averaged across participants,  $k_{2nd}$  was equal to  $k_{1st}$  at a deflection of 8.2 (SD 1.7) mm, corresponding to an average applied force of 110.5 (SD 52.0) N. The average coefficient of determination ( $r^2$ ) value increased by 0.036 (from 0.899 to 0.935) for the 2<sup>nd</sup> order polynomials compared to the 1<sup>st</sup> order curve fits. The  $r^2$  increased by 0.091 (from 0.830 to 0.922) for the 1.25 cm drop height, and by 0.083 (from 0.783 to 0.866) for the 5 cm drop height condition.

### 2.3.2 Model accuracy

ANOVA demonstrated that stiffness calculation approach had a significant influence on the accuracy of  $F_{max}$  predictions ( $F_{3,13} = 43.6, p < 0.001$ ; table 2.2, figure 2.4).  $F_{dif}$  averaged across drop heights was best for the mass-spring model using  $k_{1st}$  (mean (SD) = 4.7 (13.9) %) and  $k_{vibe}$  (6.7 (13.7) %) followed by  $k_{combo}$  (13.2 (15.8) %) and  $k_{2nd}$  (49.2 (27.7) %).

Model predictions of  $F_{max}$  were also influenced by drop height ( $F_{2,13} = 11.3$ ,  $p = 0.001$ ; table 2.2, figure 2.4). Overall, the model accuracy was significantly better at the 5 cm drop height (mean (SD)  $F_{dif} = 8.1$  (28.6) %) compared to the impending impact ( $F_{dif} = 22.9$  (21.5) %) and 1.25 cm ( $F_{dif} = 24.3$  (24.3) %) drop heights.

I also observed a significant interaction effect indicating that the effect of stiffness calculation method on  $F_{max}$  differed across drop heights ( $F_{3,39} = 19.4$ ,  $p < 0.001$ ; table 2.2, figure 2.4).  $F_{dif}$  for the mass-spring model using  $k_{2nd}$  (which always over-predicted  $F_{max}$ ) remained relatively constant for the 5 cm compared to 0 cm drop height. In contrast,  $F_{max}$  predictions using  $k_{vibe}$ ,  $k_{1st}$ , and  $k_{combo}$  tended to improve with increasing drop height. At my highest drop height (5 cm), the mass-spring model using  $k_{combo}$  provided the most accurate match with  $F_{dif}$  averaging 0.3 (SD 11.6) %.

ANOVA demonstrated that  $T_{dif}$  was significantly associated with stiffness calculation approach ( $F_{3,13} = 15.5$ ,  $p < 0.001$ ; table 2.2, figure 2.5). Averaged across drop heights,  $T_{max}$  was overestimated by the mass-spring model with all effective stiffness estimates. Overall model accuracy was best using  $k_{2nd}$  (mean (SD) of  $T_{dif} = 27.3$  (39.8) %), then  $k_{combo}$  ( $T_{dif} = 54.9$  (54.1) %), and finally  $k_{vibe}$  ( $T_{dif} = 69.0$  (64.4) %) and  $k_{1st}$  ( $T_{dif} = 75.5$  (62.5) %).

Drop height had a significant influence on  $T_{dif}$  ( $F_{2,13} = 139.6$ ,  $p < 0.001$ ; table 2.2, figure 2.5). The models provided reasonable predictions of  $T_{max}$  for impending impact trials ( $T_{dif} = 3.2$  (16.9)), but substantially over-predicted  $T_{max}$  as drop height increased ( $T_{dif} = 118.4$  (50.1) % at the 5 cm condition).

I also observed a significant interaction whereby stiffness calculation method effects on  $T_{max}$  were mediated by drop height ( $F_{4,52} = 21.5$ ,  $p < 0.001$ ; table 2.2, figure

2.5). Although  $T_{dif}$  increased with increasing drop height for all stiffness estimates, the effect was less pronounced for  $k_{2nd}$  compared to  $k_{combo}$ ,  $k_{vibe}$  and  $k_{1st}$ .

## 2.4 Discussion

In the current study, I used pelvis release trials with healthy young women to examine the linearity of pelvic compressive stiffness during low-energy sideways falls. I observed slight non-linearities in the pelvis' force-deflection properties, but the mean  $r^2$  from a 2<sup>nd</sup> order polynomial regression increased by only 0.070 compared to linear curve fits, which was less than the 0.10 *a priori* selected to indicate a substantially improved fit. I also used mathematical models to predict the peak force ( $F_{max}$ ) and time to peak force ( $T_{max}$ ) measured from the experimental trials. A mass-spring model using a non-linear estimate of pelvic stiffness ( $k_{2nd}$ ) consistently overestimated  $F_{max}$  (by 49%), while linear estimates of pelvic stiffness ( $k_{vibe}$ ,  $k_{1st}$ ) provided the most accurate fits at the lowest drop heights ( $F_{dif}$  was approximately 12.0%). A combination non-linear/linear approach to estimating pelvic stiffness ( $k_{combo}$ ) provided the best fit at the highest 5 cm drop height ( $F_{dif} = -0.3\%$ ), which suggests that the non-linearities in pelvic stiffness are limited to the low force region. The models did a poor job of characterizing  $T_{max}$ , indicating that mass-spring models more effectively describe force magnitudes than the temporal aspects of impact dynamics.

This study adds to the literature that reports on the compressive stiffness of the pelvis during lateral impact. My linear estimate of effective stiffness based on a mass-spring model of free vibration ( $k_{vibe} = 23696$  (7165)) was quite similar to that observed by Robinovitch et al. (1997a) in five women in a relaxed lying position ( $k = 28231$  (8307) N/m). My linear estimate based on the force-deflection data was slightly lower ( $k_{1st} =$

20922 (3482) N/m), but the linear portion of  $k_{combo}$  (31784 (10266) N/m) was similar to Robinovitch et al.'s value. The current study does not support the validity of a pelvis with completely non-linear force-deflection properties, as the 2<sup>nd</sup> order regression polynomials only marginally improved the curves fit to the force-deflection data. Furthermore, my mass-spring model consistently over-predicted  $F_{max}$  using the non-linear stiffness coefficient.

An important property of my mathematical model is its ability to estimate the force applied to the hip during falls from standing height. I simulated falls with impact velocities of 2, 3, and 4 m/s, which correspond to the mean (SD) hip impact velocity of 3.0 (1.0) m/s observed for unexpected falls from standing initiated by a sudden floor translation (Feldman and Robinovitch, 2007). The mass-spring model with  $k_{combo}$  was the most accurate for my highest drop height condition; consequently, I used this stiffness value in these higher energy fall simulations. Using model inputs averaged across my participants,  $F_{max}$  was predicted as 1846, 2649, and 3434 N for falls with hip impact velocities of 2, 3, and 4 m/s, respectively. The 4 m/s value is slightly above the mean (SD) femoral fracture threshold for elderly women of 3140 (1240) reported by Cheng et al. (1997), which supports assertions that the impact forces associated with worst-case falls on the hip are sufficient to fracture the proximal femur in approximately 50% of elderly women (van den Kroonenberg et al., 1995). This suggests that devices aimed at reducing the magnitude of impact force could have substantial potential benefits as interventions to reduce hip fracture risk.

Accordingly, these results have significant implications for the design of mechanical systems that test the force attenuation provided by protective devices

including hip protectors and compliant floors. At least eight such systems have been reported in the literature, and have incorporated varying combinations of effective mass, stiffness, and impact velocity (Derler et al., 2005, Kannus et al., 1999, Mills, 1996, Nabhani and Bamford, 2002, Nabhani and Bamford, 2004, Parkkari et al., 1995, Robinovitch et al., 1995a, van Schoor et al., 2006, Wiener et al., 2002). Some systems ignore effective stiffness beyond that supplied by soft tissues (Derler et al., 2005, van Schoor et al., 2006, Wiener et al., 2002). Consequently, these systems either produce unrealistically high peak forces or employ low effective masses or velocities to achieve more realistic impact responses. Unfortunately, a mass-spring model in series with a compliant element representing a low-stiffness floor or hip protector predicts that unrealistically stiff systems will overestimate the force attenuation provided by such devices (figure 2.6). For example, the model predicts that a compliant protective element (of stiffness  $k_f = 13$  kN/m) in-series with a pelvis (of stiffness  $k_b = 25$  kN/m) reduces peak force by 35% compared to the unpadded condition. However, if  $k_b$  is increased to 75 kN/m the predicted force attenuation increases to almost 60%, which is similar to the peak force reduction reported for hip protectors measured in relatively rigid mechanical test systems (Derler et al., 2005, Nabhani and Bamford, 2002, van Schoor et al., 2006). Thus, the field requires international standards that prescribe acceptable limits for design parameters including effective stiffness and impact velocity (see Chapter 4) in order to ensure that the marketplace is provided with accurate and valid estimates of the biomechanical effectiveness of commercial products.

The current study was associated with several limitations. First, the drop heights I examined were associated with relatively low impact velocities ranging from means of

0.087 (SD 0.039) m/s for the impending impact condition to 0.917 (SD 0.061) m/s in the 5 cm drop height condition (table 2.2). However, members of our lab have observed hip impact velocities as low as 1.0 during unexpected falls from standing when healthy young participants absorb energy through eccentric lower limb muscle contraction and break the fall with the outstretched hand (Feldman and Robinovitch, 2007).

Consequently, I regard the pelvis release paradigm as an effective means of simulating low severity, but clinically relevant falls. Furthermore, I was unable to examine higher drop heights in-vivo as I observed applied forces in excess of 1200 N, which is similar to the lowest fracture force observed for elderly women with low bone mineral density (Cheng et al., 1997). For the same reason, I was unable to enrol elderly women as study participants. However, the differences in force-deflection properties between elderly women at highest risk of hip fracture and my healthy young participants may be partially offset by the latter's low BMI (average (SD) = 20.9 (1.6) kg/m<sup>2</sup>), which correlates with smaller amounts of natural soft tissue padding over the proximal femur (Maitland et al., 1993). Finally, I only considered a single model configuration in my simulations. However, I'm confident that my mass-spring models provide a comprehensive range of predictions as no other support models (including Voigt, standard linear solid and Maxwell) have demonstrated more accurate predictions of impact dynamics during simulated falls on the hip (Robinovitch et al., 1997b).

This is the first study to measure the force-deflection properties of the hip in-vivo during low-energy sideways falls, and substantially advances our understanding of the dynamic compressive properties of the hip region during sideways falls. The results demonstrate that the pelvis does not possess significant non-linear stiffness characteristics

during impact. Instead, peak force is best predicted by a mass-spring model using a stiffness parameter that is non-linear at low forces (from 0 to 300 N), and linear at forces above this transition threshold. These data provide essential information for expanding the use of mathematic models to assess other aspects of fall dynamics e.g. floor stiffness (see Chapter 5). Furthermore, this detailed characterization of pelvic stiffness will inform ongoing efforts towards the development of international standards for mechanical test systems that assess the force reduction provided by protective devices including hip protectors and compliant floors.

**Table 2-1: Average (SD) characteristics of parameter values used as inputs in the mass-spring model. With the exception of impact velocity, all parameters were determined from data in the impending impact (0 cm drop height) condition.**

	Average (SD)
$m$ (kg)	24.1 (3.3)
$k_{vibe}$ (N/m)	23696 (7165)
$k_{1st}$ (N/m)	20922 (3482)
$k_{2nd}$ A constant	800478 (730112)
$k_{2nd}$ B constant	9306 (8099)
$k_{combo}$ linear phase (N/m)	31784 (10266)
impact velocity 0 cm (m/s)	0.087 (0.039)
impact velocity 1.25 cm (m/s)	0.456 (0.04)
impact velocity 5 cm (m/s)	0.917 (0.061)

**Table 2-2: Average (SD) actual values and model predictions of  $F_{max}$  and  $T_{max}$  across drop heights.**

	Drop height (cm)		
	0	1.25	5
$F_{max}$ (N)			
Experimental	433 (76)	576 (83)	1004 (115)
Mass-spring with $k_{vibe}$	482 (66)	647 (91)	951 (157)
Mass-spring with $k_{1st}$	481 (65)	631 (75)	918 (121)
Mass-spring with $k_{2nd}$	621 (92)	883 (128)	1456 (299)
Mass-spring with $k_{combo}$	517 (58)	679 (62)	995 (113)
$T_{max}$ (s)			
Experimental	0.089 (0.007)	0.047 (0.007)	0.028 (0.005)
Mass-spring with $k_{vibe}$	0.096 (0.017)	0.074 (0.013)	0.064 (0.009)
Mass-spring with $k_{1st}$	0.101 (0.010)	0.076 (0.008)	0.067 (0.007)
Mass-spring with $k_{2nd}$	0.080 (0.013)	0.057 (0.011)	0.046 (0.010)
Mass-spring with $k_{combo}$	0.088 (0.013)	0.068 (0.010)	0.060 (0.008)

Figure 2-1: Experimental schematic. A sling and electromagnet were used to raise and suddenly release the participant's pelvis from heights ( $h$ ) of 0 (impending impact), 1.25, and 5 cm above the floor surface, simulating the impact stage of a sideways fall on the hip. The time-varying force applied to the hip region during impact was measured with a force plate.

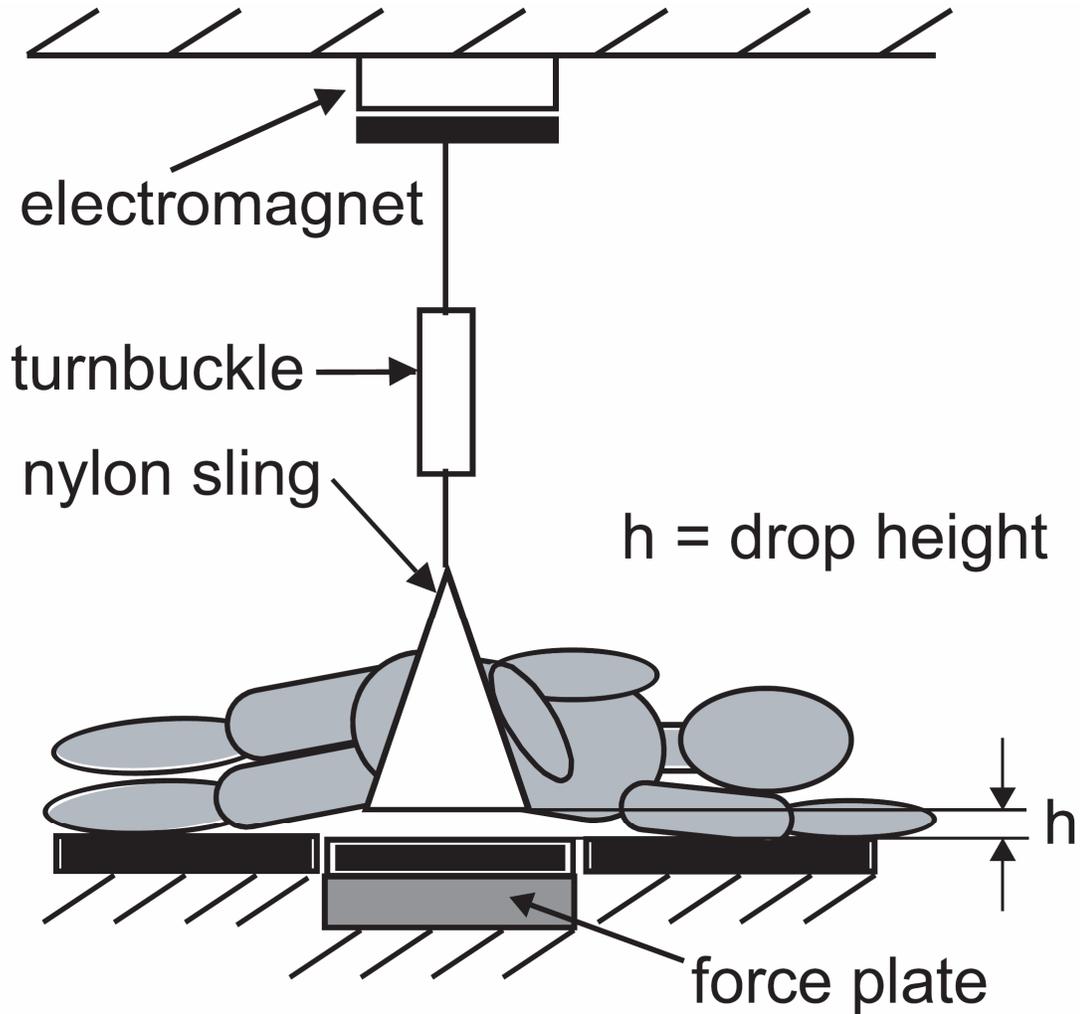
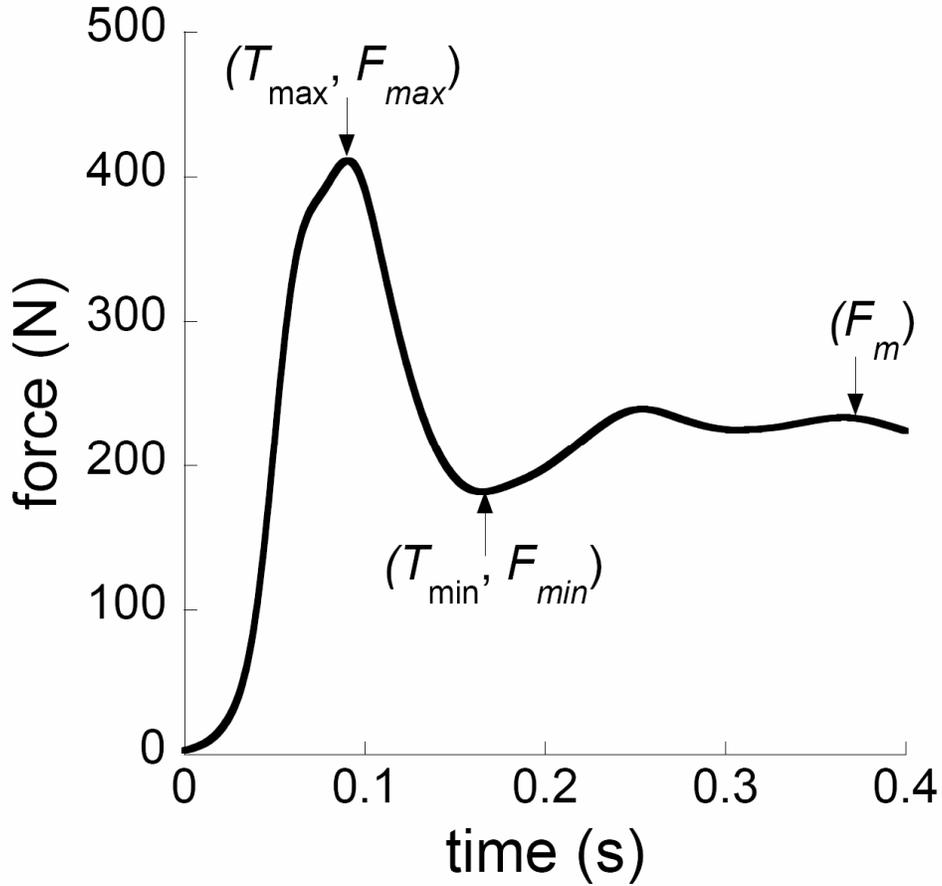
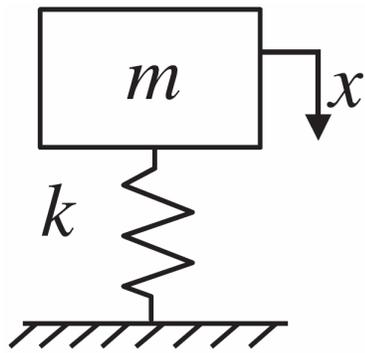


Figure 2-2: A) Example force vs. time trace observed in the impending impact drop height condition. The points indicated were used in the free vibration response approach of characterizing parameter values for a (B) mass-spring model.  $T = 0$  corresponds to the instant of impact ( $T_{imp}, F_{imp}$ ).

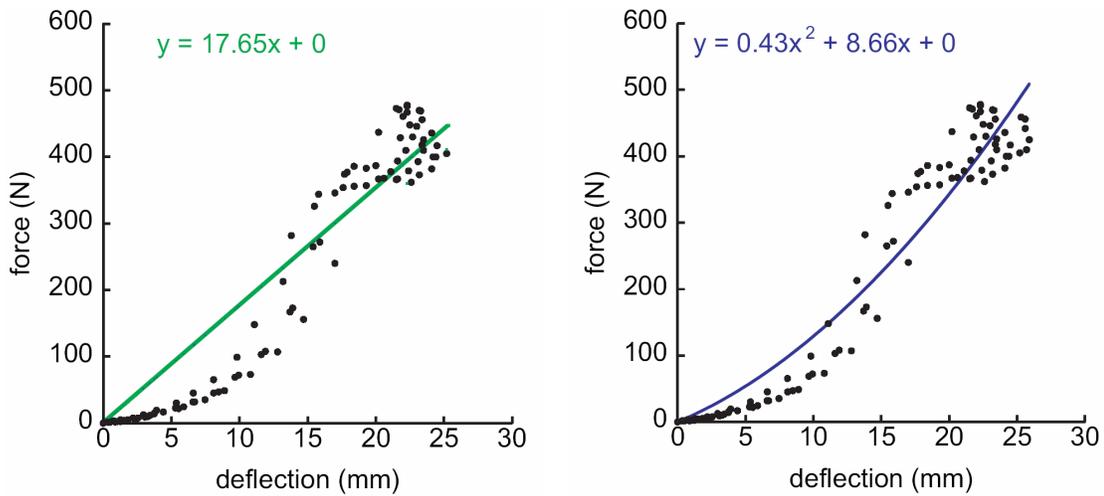


A)



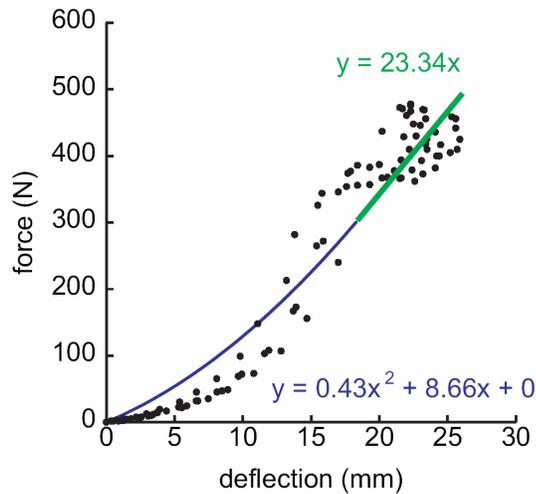
B)

**Figure 2-3: Example force vs. deflection data for a participant in the 0 cm drop height condition with: A) 1<sup>st</sup> order and B) 2<sup>nd</sup> order polynomial fits, and C) the combination fit approach which used the 2<sup>nd</sup> order polynomial for forces up to 300 N, and was equal to the tangent at 300 N for all forces above this transition point.**



**A)**

**B)**



**C)**

Figure 2-4: Example force and deflection vs. time trace observed in the 5 cm drop height condition. Both peak force and peak deflection followed a similar time course occurring at approximately 40 ms after the start of impact.

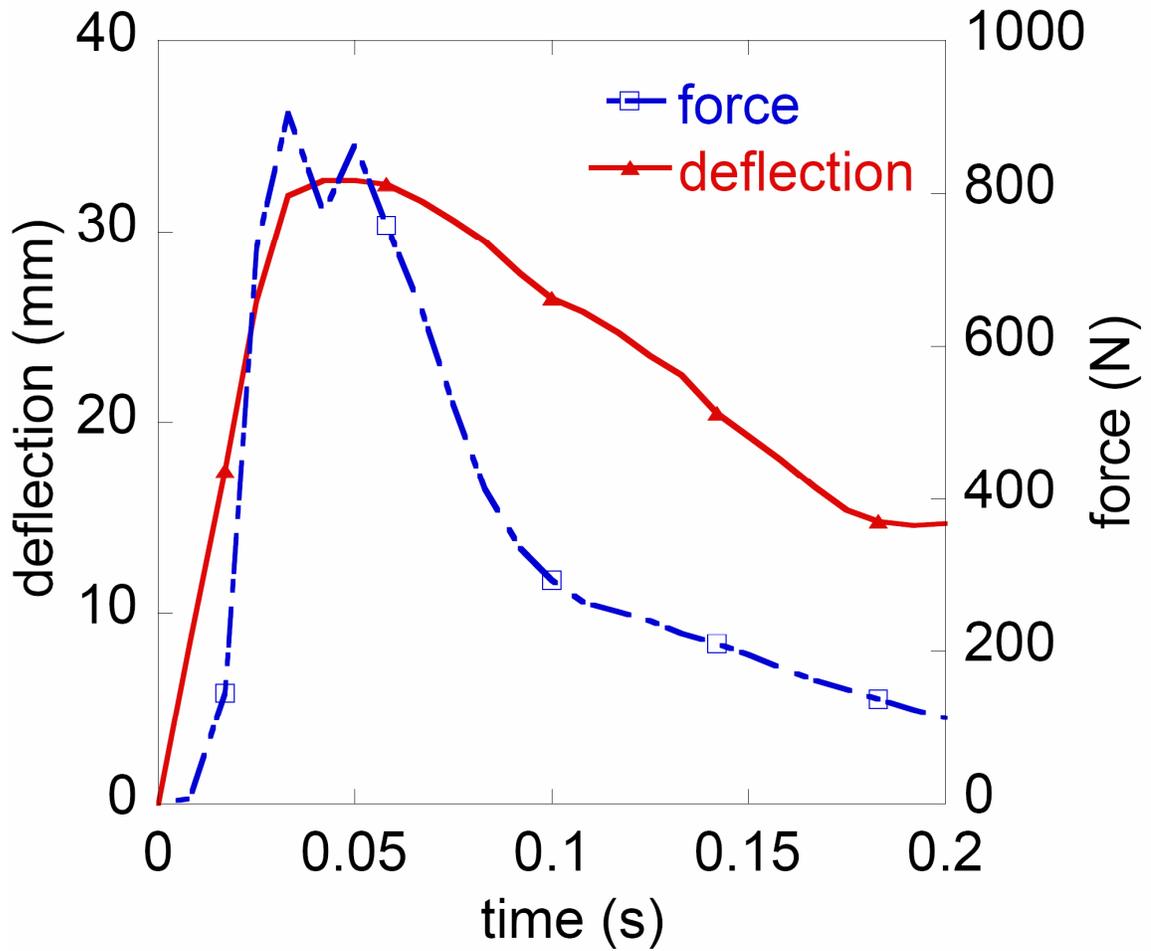
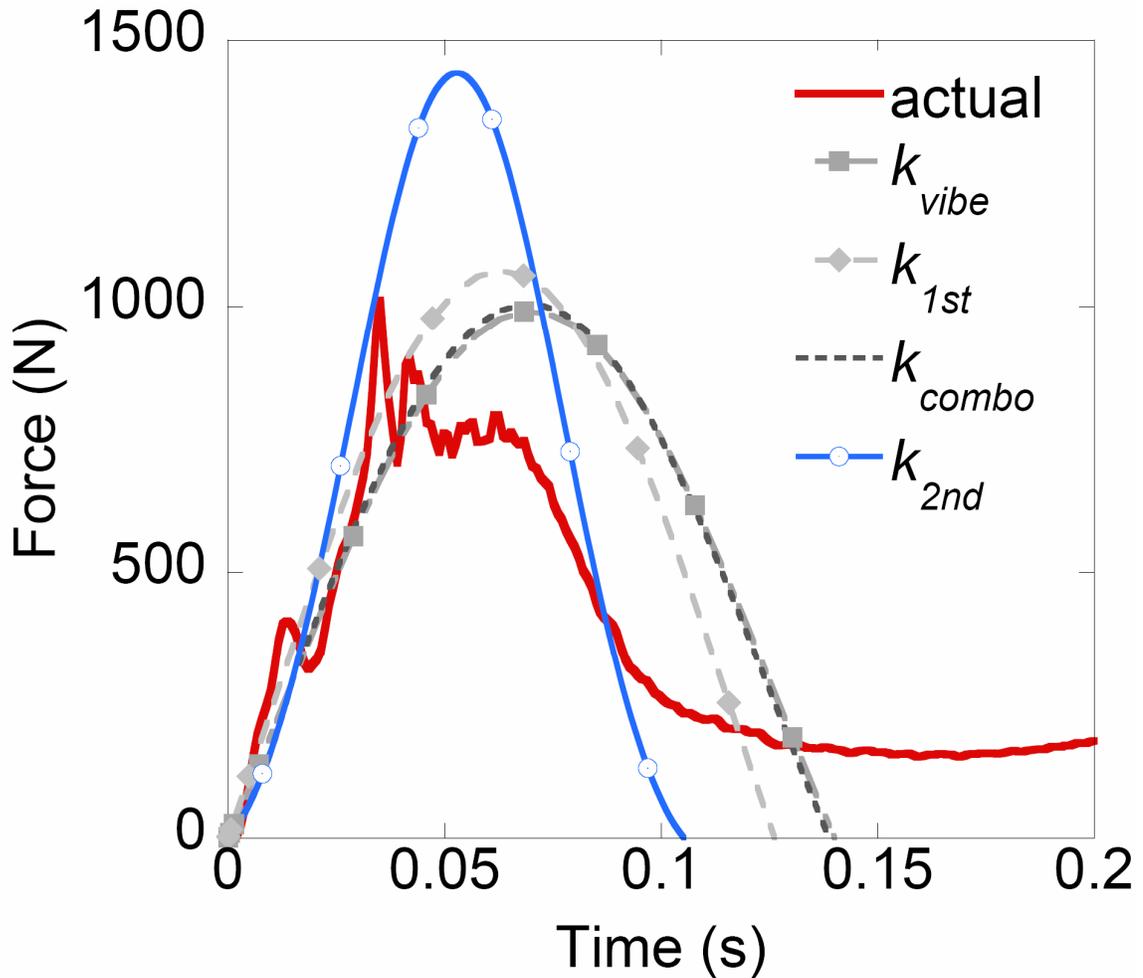
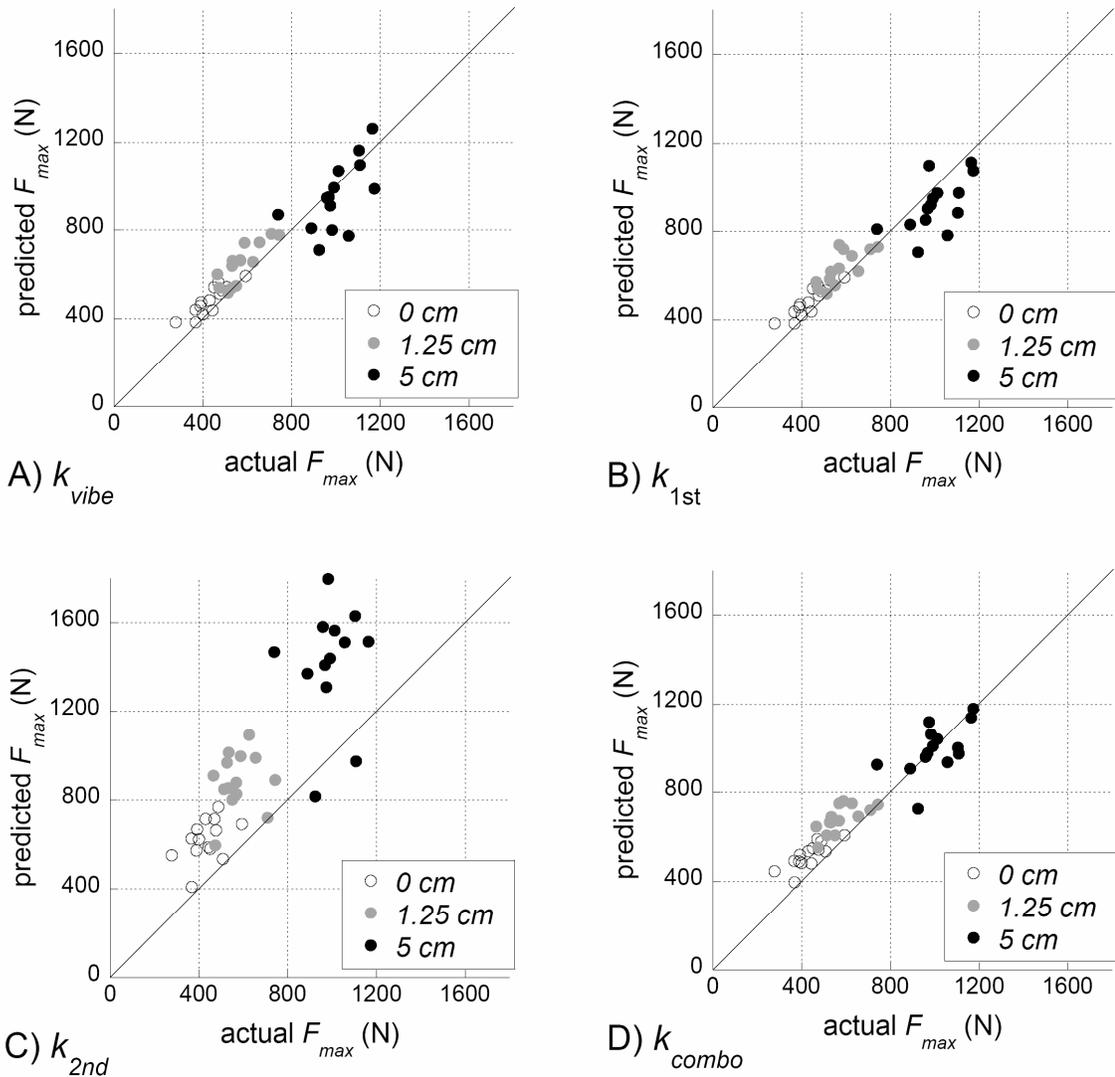


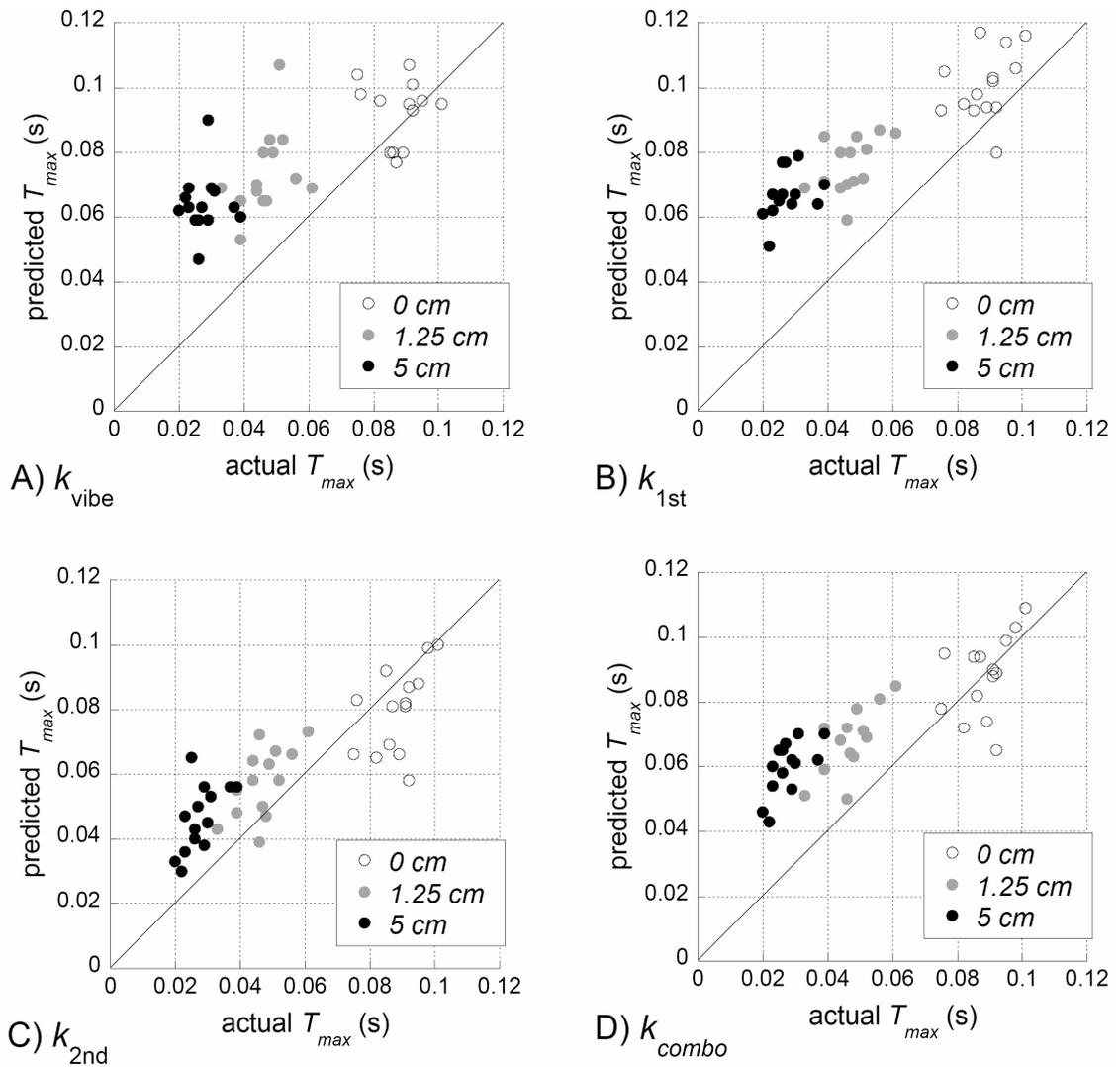
Figure 2-5: Example force vs. time trace observed in the 5 cm drop height condition for experimentally observed data and mass-spring model predictions utilizing effective stiffness estimated from: the free vibration response ( $k_{vibe}$ ), first ( $k_{1st}$ ) and second ( $k_{2nd}$ ) order polynomials fit to the force-deflection data, and a combination approach ( $k_{combo}$ ) which used the 2<sup>nd</sup> order polynomial for forces up to 300 N and the tangent value at 300 N for all forces above this transition point.



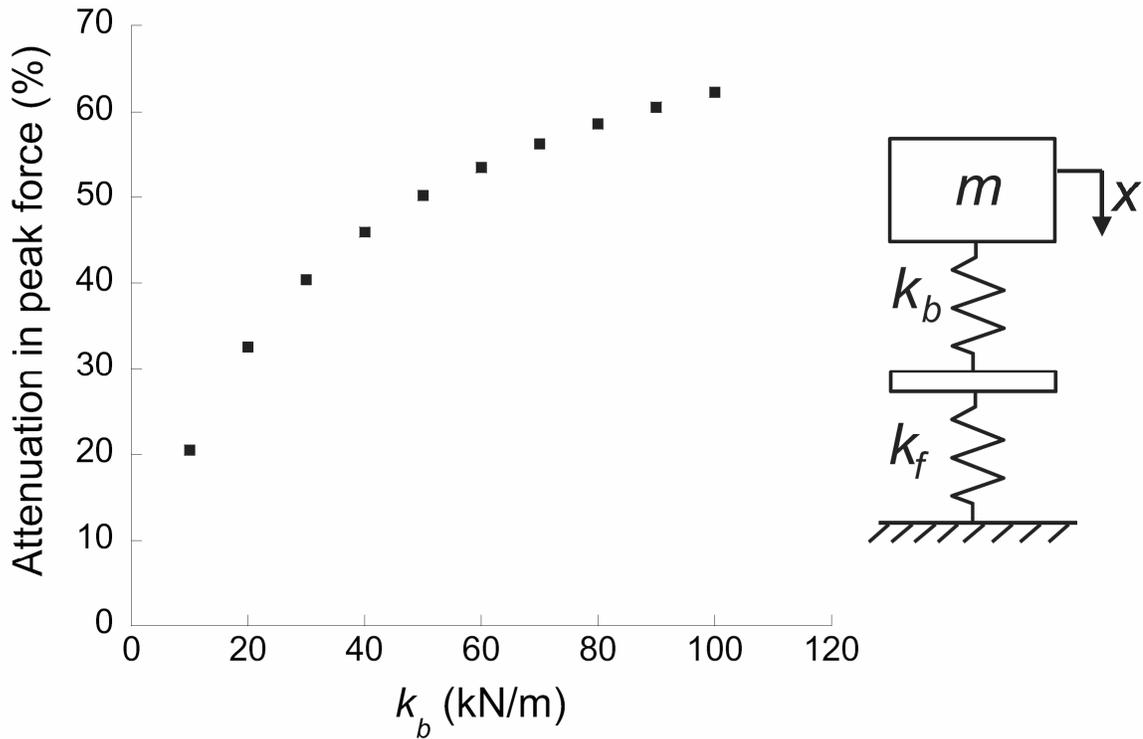
**Figure 2-6: Scatterplots of mass-spring model predictions of  $F_{max}$  vs. actual  $F_{max}$  for all participants at the 0, 1.25, and 5 cm drop heights using: A)  $k_{vibe}$ ; B)  $k_{1st}$ ; C)  $k_{2nd}$ ; and D)  $k_{combo}$ .**



**Figure 2-7: Scatterplots of mass-spring model predictions of  $T_{max}$  vs. actual  $T_{max}$  for all participants at the 0, 1.25, and 5 cm drop heights using: A)  $k_{vibe}$ ; B)  $k_{1st}$ ; C)  $k_{2nd}$ ; and D)  $k_{combo}$ .**



**Figure 2-8: Mass-spring model demonstration of the influence of pelvic stiffness ( $k_b$ ) on the attenuation in peak force provided by a compliant floor or hip protector compared to an unprotected fall.  $m$  is the effective mass of the body,  $u$  is the impact velocity, and  $k_f$  is the stiffness of the protective device. If  $m$  (30 kg),  $u$  (3.1 m/s), and  $k_f$  (13 kN/m), are held constant, the force attenuation provided by a protective device increases as  $k_b$  increases.**



## **CHAPTER 3 EFFECT OF SOFT SHELL HIP PROTECTORS ON PRESSURE DISTRIBUTION TO THE HIP DURING SIDEWAYS FALLS**

### **3.1 Background**

Hip fractures are a major public health problem. The annual worldwide incidence of hip fractures was estimated at 1.6 million in 1990, and is expected to increase to 6.2 million by 2050 (Gullberg et al., 1997). Women with hip fracture are twice as likely to die in the four years following the fracture (Empana et al., 2004b), and three times more likely to be functionally dependent (Norton et al., 2000b) than non-fracture controls having the same baseline health status. Although pharmacologic interventions can increase bone mineral density and reduce fracture risk, their benefits are limited primarily to those with confirmed osteoporosis or previous fractures (Cummings et al., 1998, McClung et al., 2001). Since up to 74% of hip fracture patients are non-osteoporotic (Siris et al., 2004), alternative or supplementary prevention strategies must also be considered. Since 90% of hip fractures are due to falls (Grisso et al., 1991), interventions that reduce the frequency and/or severity of falls should be especially beneficial.

External hip protectors represent a particularly promising strategy for preventing hip fractures in high-risk elderly individuals. These devices consist of soft shell (e.g., foam rubber) or hard shell (e.g., rigid plastic) pads that cover the proximal femur, and are typically imbedded in an undergarment. Hip protectors have been shown to reduce the impact force applied to the proximal femur (Derler et al., 2005, Kannus et al., 1999, Minns et al., 2004b, Parkkari et al., 1994, Robinovitch et al., 1995a, van Schoor et al.,

2006) and substantially decrease the risk for hip fracture if they are worn at the time of a fall (Cameron et al., 2003, Kannus et al., 2000, Lauritzen et al., 1993). However, the clinical effectiveness of the intervention is limited by relatively poor user acceptance and adherence in wearing the device (Parker et al., 2006, Sawka et al., 2005). For example, in a study of 600 frail community dwelling women, Cameron et al. (2003) reported that the relative risk of hip fracture during a fall was reduced more than 4-fold while wearing a hip protector, compared to falls where a hip protector was not worn (RR 0.23; 95% CI 0.08 to 0.67). However, since hip protectors were worn during only 51% of the falls in the intervention group, an intention to treat analysis found no significant difference in the incidence of hip fractures in the intervention and control groups (RR 0.92; 95% CI 0.51 to 1.68). Clearly, there is a need to develop strategies to improve user acceptance and adherence with this intervention.

Soft shell hip protectors have recently become more common in the marketplace, and may be more comfortable and acceptable to users than traditional hard shell designs. Unfortunately, there are conflicting reports on their biomechanical effectiveness. For example, Derler et al. (2005) used a mechanical system to test three hard shell and seven soft shell protectors, and found that peak forces at the femoral neck were slightly higher for hard shell than soft shell protectors at the highest impact energy (75 J). In contrast, van Schoor et al. (2006) found that hard shell protectors provided nearly twice the mean attenuation in peak femoral force as soft shell protectors (76% versus 40%). This discrepancy is probably due to differences between the two research groups in the design of their test systems. Indeed, while there are some human impact data to guide the design of such systems (Robinovitch et al., 1991), difficulty arises in accurately matching the

anatomy and mechanical properties of relevant skeletal structures and soft tissues of the pelvis (Derler et al., 2005, Kannus et al., 1999, Minns et al., 2004b, Robinovitch et al., 1991, Robinovitch et al., 1995a, van Schoor et al., 2006). Accordingly, there is a need for complementary experiments with human subjects to determine the force-attenuating characteristics of soft shell hip protectors.

The current study addresses this need by determining the effect of soft shell hip protectors on the magnitude and distribution of impact force during safe, sideways falls in humans. In particular, I compared the force and pressure attenuation provided by two commercially available soft shell hip protectors: (1) a 16-mm thick oval protector which covers the greater trochanter (GT) and surrounding region, and (2) a 14 mm-thick protector which has a horseshoe shape, and covers the skin adjacent but not directly overlying the GT. I hypothesized that both hip protectors would reduce the total peak force applied to the impact site, due to compression and energy absorption in the foam padding. I also hypothesized that both protectors would alter the distribution of force within the hip region – either by smoothing out the local variation in stiffness through a springs-in-series effect (as in the case of the continuous protector) or by forming an actual bridge over the bone (as in the case of the horseshoe protector). Finally, I hypothesized that the force and pressure attenuations would scale with body mass index (BMI), and be greater for slender individuals due to their high baseline stiffness.

## **3.2 Methods**

### **3.2.1 Participants**

I recruited participants through advertisements posted on the campus of Simon Fraser University. I screened interested individuals to ensure they were 18 years of age or older and free of musculoskeletal conditions such as severe arthritis, or recent sprains, strains, or fractures. Study participants consisted of fifteen women having a mean age of 22.3 years (SD = 3, range = 18 – 29), a mean body mass of 63.5 kg (SD = 6.4, range = 53.2 – 74.4), a mean height of 1.66 m (SD = 0.05, range = 1.59 - 1.74), and a mean BMI ((body mass in kg)/(square of body height in m)) of 23.0 kg/m<sup>2</sup> (SD = 2.5, range = 19.0 - 28.3). All participants provided written informed consent and the study was approved by the Committee on Research Ethics at Simon Fraser University.

### **3.2.2 Pelvis release trials**

The experimental technique (which has been described previously (Robinovitch et al., 1991, 1997a)) involved using a sling and electromagnet to raise and suddenly release the subject's pelvis onto a force plate, simulating the impact stage of a sideways fall on the hip (figure 3.1A). To conduct a trial, the subject lay on her left side, with her pelvis cradled in a nylon sling. A steel cable, the length of which could be adjusted via a turnbuckle, was connected at one end to the sling and at the other end to an electromagnet (model DCA 600-110I; Automatic Equipment Corp., Cincinnati, OH, USA; 90% decay time = 15 ms). I used the turnbuckle to raise the pelvis until a gap of 5 cm (measured with a wooden block) existed between the floor and the skin overlying the GT (for unpadded impacts) or between the floor and the outer surface of the hip protector (for padded impacts). The participant was instructed to relax, and once she confirmed she was ready,

I released the electromagnet causing her to fall onto the landing surface. Throughout the trial, the lateral aspect of the shin and shoulder remained in contact with the floor; only the pelvis was lifted and released. The subject's left arm was extended over her head, with her knees and hips flexed to 30 degrees. I used surgical positioning mats (Olympic Vac Pac models 30 and 31, Olympic Medical Corp., Seattle, WA, USA) to ensure consistent postures between trials.

I used a dual arrangement of a force plate and a load cell to measure how the impact force was distributed to various circular regions centred about the GT (figure 3.1B). The total force applied to the entire hip region was recorded by a force plate (model 4060H; Bertec Corp., Columbus, OH, USA) bolted to the floor. The contact force applied to a circular region centred about the trochanter was recorded with a load cell (model 31; Honeywell Sensotec, Columbus, Ohio, USA) embedded in a wood block secured on top of the force plate. The sensing area of the load cell was varied by bolting an aluminium disc of radius 1.25, 2.5, or 5 cm on top of it. The edges of the disc did not contact the wood block, and the top surface was completely flush. All force signals were sampled at 1000 Hz. In each trial, the subject was positioned so her left GT, located by manual palpation, was directly above the centre of the load cell.

I conducted trials with the participant wearing no hip protector ('unpadded') and while wearing two different commercially available soft shell hip protectors (figure 3.1C). The Hipsaver protector incorporates an oval pad of viscoelastic open-cell foam which is encapsulated in a nylon sleeve and sewn into cotton underwear. The maximum width of the pad is 17 cm, the maximum height is 19 cm, and the thickness is 16 mm. When worn, the centre point of the pad is located directly over the GT. The SafeHip Soft

protector consists of a horseshoe-shaped pad of closed-cell ethylene vinyl acetate (EVA) foam which is sewn into cotton underwear. It has a maximum width of 17 cm, maximum height of 17 cm, and (in the version that I tested) a thickness of 14 mm. The protector has a gap so the foam padding surrounds, but does not directly cover, the GT. I shall refer to these respective protectors as “continuous” and “horseshoe.”

I conducted three trials (the results of which were averaged) for each load cell radius and hip protector condition, for a total of 27 trials per participant. I randomized the order of the conditions, but acquired three trials in each condition consecutively.

In all trials, I found that the temporal variation in impact force was dominated by a single natural frequency and a distinct maximum value (figure 3.2). I will use the label  $F_{total}$  to indicate the peak total force measured by the force plate, and the labels  $F_5$ ,  $F_{2.5}$ , and  $F_{GT}$  to represent the peak force recorded by the load cells of radius 5 cm, 2.5 cm, and 1.25 cm, respectively. I also found that, over the three hip protector conditions (unpadded, continuous protector, and horseshoe protector), changing the load cell radius had no effect on  $F_{total}$  ( $F(2, 28) = 0.535$ ,  $p = 0.578$ ) or time to  $F_{total}$  ( $F(2, 28) = 0.717$ ,  $p = 0.497$ ).

It is important to consider and compare the dynamics governing my pelvis release experiments to those governing actual falls from standing. First, these experiments simulate the impact stage of a sideways fall, where the lateral aspect of the hip impacts the ground, after initial contact to the knee and upper extremity. Of course, sideways falls can produce a wide range of impact configurations, but Feldman and Robinovitch (2007) report that most involve impact to the knee and upper extremity before the hip. Second, due to safety precautions, my drop height of 5 cm is much smaller than the approximately

100 cm which accompanies the typical fall from standing that leads to hip fracture. Higher drop heights were simply deemed unsafe, given that 5 cm was sufficient to produce average peak forces exceeding 1.1 kN, and reports from some subjects of tissue tenderness and slight bruising. Furthermore, the difference in hip impact velocity between pelvis release experiments and falls from standing is not as great as would be expected from simple free fall dynamics, averaging about 1.0 m/s for the former (see Chapter 5) and 3.0 m/s for the latter (Feldman and Robinovitch, 2007). Indeed, members of our laboratory have observed that sideways falls from standing can produce hip impact velocities as low as 0.8 m/s (Feldman and Robinovitch, 2007). This is due to the tendency for humans to utilize protective responses (energy absorption in the lower extremity muscles during descent, and initial impact to the knee and upper extremity (Feldman and Robinovitch, 2007, Robinovitch et al., 2000a, Sandler and Robinovitch, 2001)) to reduce the impact velocity of the hip during sideways falls. Consequently, I regard the pelvis release technique as providing a repeatable, low severity, but clinically relevant experimental model of a sideways fall.

### 3.2.3 Data analysis

I estimated how  $F_{total}$  was distributed into four regions centred about the GT as follows (figure 3.1D): (1) a circle of radius 1.25 cm centred directly over the GT (measured directly as  $F_{GT}$ ), (2) a ring of tissues having an outer radius of 2.5 cm and inner radius of 1.25 cm ( $F_{2.5R} = F_{2.5} - F_{GT}$ ), (3) a ring of tissues having an outer radius of 5 cm and inner radius of 2.5 cm ( $F_{5R} = F_5 - F_{2.5}$ ), and (4) a region outside the circle of 5 cm radius centred about the GT ( $F_{+5} = F_{total} - F_5$ ). I calculated corresponding mean

pressures as  $P_{5R} = F_{5R}/A_{5R}$ ,  $P_{2.5R} = F_{2.5R}/A_{2.5R}$ ; and  $P_{GT} = F_{GT}/A_{1.25}$ , where  $A_{5R} = 58.9 \text{ cm}^2$ ,  $A_{2.5R} = 14.7 \text{ cm}^2$ , and  $A_{1.25} = 4.9 \text{ cm}^2$ .

### 3.2.4 Statistics

I used repeated measures ANOVA to determine whether there were differences between the hip protector conditions in  $P_{5R}$ ,  $P_{2.5R}$ ,  $P_{GT}$ ,  $F_{\text{total}}$ , and  $F_{+5}$ . Where required, I used the Huynh-Feldt epsilon correction factor to adjust for violations of the sphericity assumption. When ANOVA indicated a significant effect, I used paired t-tests to identify differences between conditions. I used Pearson correlation to examine whether observed changes in pressure and force between unpadded and padded conditions associated with the BMI of the participants. I used a significance level of  $\alpha = 0.05$ , and all analyses were performed with statistical analysis software (SPSS Version 14.0, SPSS Inc., Chicago, IL, USA).

### 3.3 Results

There were significant differences in  $P_{GT}$ ,  $P_{2.5R}$ , and  $P_{5R}$  between the hip protector conditions (tables 3.1 and 3.2). Both hip protectors caused  $P_{GT}$  and  $P_{2.5R}$  to decrease, and  $P_{5R}$  to increase. When compared to the unpadded condition, average values of  $P_{GT}$  were 72.9% smaller with the continuous protector and 76.0% smaller with the horseshoe protector (figure 3.3A). Similarly, average values of  $P_{2.5R}$  were 62.0% smaller with the continuous protector and 62.9% smaller with the horseshoe protector. In contrast, when compared to the unpadded condition, average values of  $P_{5R}$  were 29.5% larger with the continuous protector and 36.4% larger with the horseshoe protector.

There were also significant differences in  $F_{total}$  and  $F_{+5}$  between the hip protector conditions (tables 3.1 and 3.2). When compared to the unpadded condition,  $F_{total}$  was 18.9% smaller with the continuous protector and 8.6% smaller with the horseshoe protector. Average values of  $F_{+5}$  were 23.4% higher with the continuous protector and 57.4% higher with the horseshoe protector. In the unpadded condition, 17.1% of the total force on average was applied directly to the GT, 45.5% was applied inside a circular area of radius 2.5 cm centred about the GT, and 53.3% was applied outside the 2.5 cm radius (figure 3.3B). In the padded conditions, only 5.4% of the total force on average was delivered directly to the GT, 17.8% was applied inside the 2.5 cm radius, and 83.2% was applied outside the 2.5 cm radius.

There were no significant differences between the horseshoe and continuous protectors in the magnitudes of  $P_{GT}$ ,  $P_{2.5R}$ , and  $P_{5R}$  (tables 3.1 and 3.2). However, average values of  $F_{total}$  were 12.7% larger, and average values of  $F_{+5}$  were 27.5% larger, for the horseshoe than continuous protector.

BMI correlated significantly with the changes in  $P_{GT}$ ,  $P_{5R}$ , and  $F_{total}$  provided by the continuous protector (figure 3.4), but not with changes in  $P_{2.5R}$  or  $F_{+5}$ . In particular, as BMI decreased, the continuous protector provided more attenuation in  $P_{GT}$  ( $R=0.66$ ;  $p=0.007$ ) and  $F_{total}$  ( $R=0.61$ ;  $p=0.017$ ), and a greater increase in  $P_{5R}$  ( $R=-0.56$ ;  $p=0.031$ ). While similar trends were observed for the horseshoe-shaped protector, these did not reach statistical significance.

### 3.4 Discussion

The goal of this study was to determine the effect of soft shell hip protectors on the magnitude and distribution of impact force during safe sideways falls in humans. These data are essential to determine whether clinical trials are warranted to examine the efficacy of these devices in preventing fractures, and for interpreting the results of those studies. I hypothesized that both the continuous and horseshoe-shaped protectors would reduce the total force and alter the distribution of force within the hip region. I found that  $P_{GT}$  was reduced 76% by the horseshoe and 73% by the continuous protector, while  $F_{total}$  was reduced 9% by the horseshoe and 19% by the continuous protector. The reductions in  $P_{GT}$  were statistically similar for the two protectors, while the reductions in  $F_{total}$  were smaller for the horseshoe than continuous protector, indicating that this protector relies more on energy shunting to reduce  $P_{GT}$ .

I also hypothesized that the attenuation in force and pressure provided by the hip protectors would scale inversely with BMI. I found that the ability of the continuous protector to attenuate  $F_{total}$  and  $P_{GT}$  increased with decreasing BMI. Similar (but statistically non-significant) trends were observed for the horseshoe protector.

My results indicate that soft shell hip protectors reduce  $P_{GT}$  through a combination of energy-absorption and energy-shunting. The energy-shunting capability of the horseshoe protector relates to the fact that it covers the skin adjacent but not directly overlying the GT, and therefore provides a bridge for shunting energy away from the bone and into the surrounding tissues, where presumably it can be absorbed with little risk of proximal femur fracture. This feature allows the horseshoe protector to provide the same reduction in  $P_{GT}$  as the continuous protector, while having a considerably lower

thickness. This more slender profile may improve user compliance in wearing the device, since compliance associates strongly with user impressions of comfort and appearance (Parker et al., 2006, van Schoor et al., 2002). The energy-shunting capacity of the continuous protector relates to its ability to “smooth out” the local variation in soft tissue stiffness over the hip region, by serving as a spring that acts in series with the spring of the pelvis and soft tissues. This “springs in series” mechanism causes the protector to reduce the in-line stiffness over every region that it covers. However, the greatest reduction in stiffness (and pressure) will occur directly over the GT, which has a baseline stiffness which is 2-3 fold greater than in the immediately adjacent regions (Robinovitch et al., 1995a). The same considerations explain the observed association between BMI and GT pressure attenuation for the continuous hip protector: the protector will cause greater reduction in GT pressure in slender individuals who have less natural padding and higher baseline stiffness than individuals with higher BMI.

An unexpected result is that both protectors caused  $P_{5R}$  to increase by over 30%. This seems counter-intuitive, since each protector covered and should decrease local stiffness in this region. This likely arises from the protectors causing a reduction in the total contact area available to support the load, when compared to the unpadded condition. This reduction in contact area dominates over the decrease in stiffness to increase  $P_{5R}$ , while the opposite is true for  $P_{GT}$  and  $P_{2.5R}$ . This observation suggests that soft shell hip protectors should be designed with maximum surface area to reduce pressure over a large region.

Previous studies have utilized mechanical test systems to measure the force attenuation provided by soft shell protectors, and it is useful to compare these data with

the results from my pelvis release experiments with human participants. Parkkari et al. (1994) found that 20 mm thick foam layers of varying density caused a reduction of 22-38% in  $F_{total}$ . These values are higher than the 9-19% attenuation in  $F_{total}$  that I observed. The discrepancy is likely due to the differences in testing methods (human subjects experiments versus mechanical test system), but might also be explained by the increased thickness of their samples. van Schoor et al. (2006) found that the same 16-mm thick continuous soft shell protector (Hipsaver) as used in the current study attenuated peak femoral force by 58% when their surrogate pelvis was covered with a 1-inch thick soft tissue layer, and 46% when their pelvis was covered with a ½-inch thick soft tissue layer. These values are slightly lower than the 73% reduction in  $P_{GT}$  I observed for the Hipsaver protector; again probably due to the differences in testing methods.

It is also informative to compare my force attenuation measures for soft shell protectors to previous measures of the force attenuation provided by traditional hard-shell protectors, which have been shown to reduce fracture risk if worn at the time of a fall. Kannus et al. (1999) reported that peak femoral impact force was reduced 85% by their KPH hard shell protector, and 49.5% by the Safehip protector, in their highest impact energy condition (110 J); both protectors have been observed in clinical trials to reduce hip fracture risk (Norton et al., 2000b). Both Nabhani and Bamford (2002) and van Schoor et al. (2006) found that these protectors reduced peak femoral force between 75 and 85%. On average, these force reductions are similar to the 73% and 76% attenuation in  $P_{GT}$  I observed for the Safehip Soft and Hipsaver protectors, respectively. This suggests that soft shell hip protectors should provide similar force attenuation as traditional hard shell protectors, if worn at the time of a fall.

There are some important limitations to my study. First, I measured the pressure distribution applied to the outer surface of the protectors, and assumed this was equal to that applied to the skin surface. I have little reason for believing this assumption is not justified for soft shell protectors of the type I examined (given their slenderness and relatively low stiffness). Second, I could not determine how the skin surface pressure is distributed to the underlying proximal femur and pelvic bones. This should be studied through careful measures with instrumented cadavers, or with improved mechanical testing systems that accurately duplicate the geometrical and mechanical properties of the pelvis, proximal femur, skin, fat, and muscles. Third, my participants were young healthy women, as opposed to elderly women who are at highest risk for hip fracture, and my falls involved impact velocities approximately one-third the average value measured for sudden sideways falls from standing. I regarded these as essential safety precautions, since my experiments produced direct lateral impact to the hip region, and average impact forces (in the unpadded condition) in excess of 1.1 kN, which is within the range of force found to fracture elderly cadaveric femora (Cheng et al., 1997, Manske et al., 2006). Furthermore, while I do not expect that age in itself would substantially alter the biomechanical variables (effective mass, stiffness, and damping) that govern force generation and the protective value of a hip protector during a fall, it would be useful to confirm this in future studies with elderly participants (although these trials may need to involve a smaller drop height, and thus a safer range of impact force, than the current experiments with young subjects). Finally, I only simulated sideways falls onto the lateral aspect of the hip, and thus cannot comment on the protective value of these protectors when the site of impact is to a more anterior or posterior aspect of the pelvis. I also

ensured that the hip protector was secured in its intended location on the pelvis. Further research is required to determine how the protective value of both soft shell and hard shell protectors are influenced by impact orientation, and by displacement of the protector from its intended location.

In summary, while soft shell hip protectors may be more comfortable and acceptable to users than traditional hard shell designs, improved understanding is required of their ability to attenuate impact force. My results indicate that, through a combination of energy absorption and energy shunting, soft shell protectors reduce total impact force by up to 19%, and pressure over the GT by up to 76%, during sideways falls in humans. While further understanding is required of how surface pressure is distributed to the underlying skeletal structures, these results support the value of clinical trials to establish the effectiveness of soft shell hip protectors in preventing hip fractures, and provide essential data to guide the design of improved hip protectors.

**Table 3-1: Average values and results of repeated-measures ANOVA comparing outcome variables in unpadded and padded conditions (values in parentheses are standard deviations; for all measures, n = 15).**

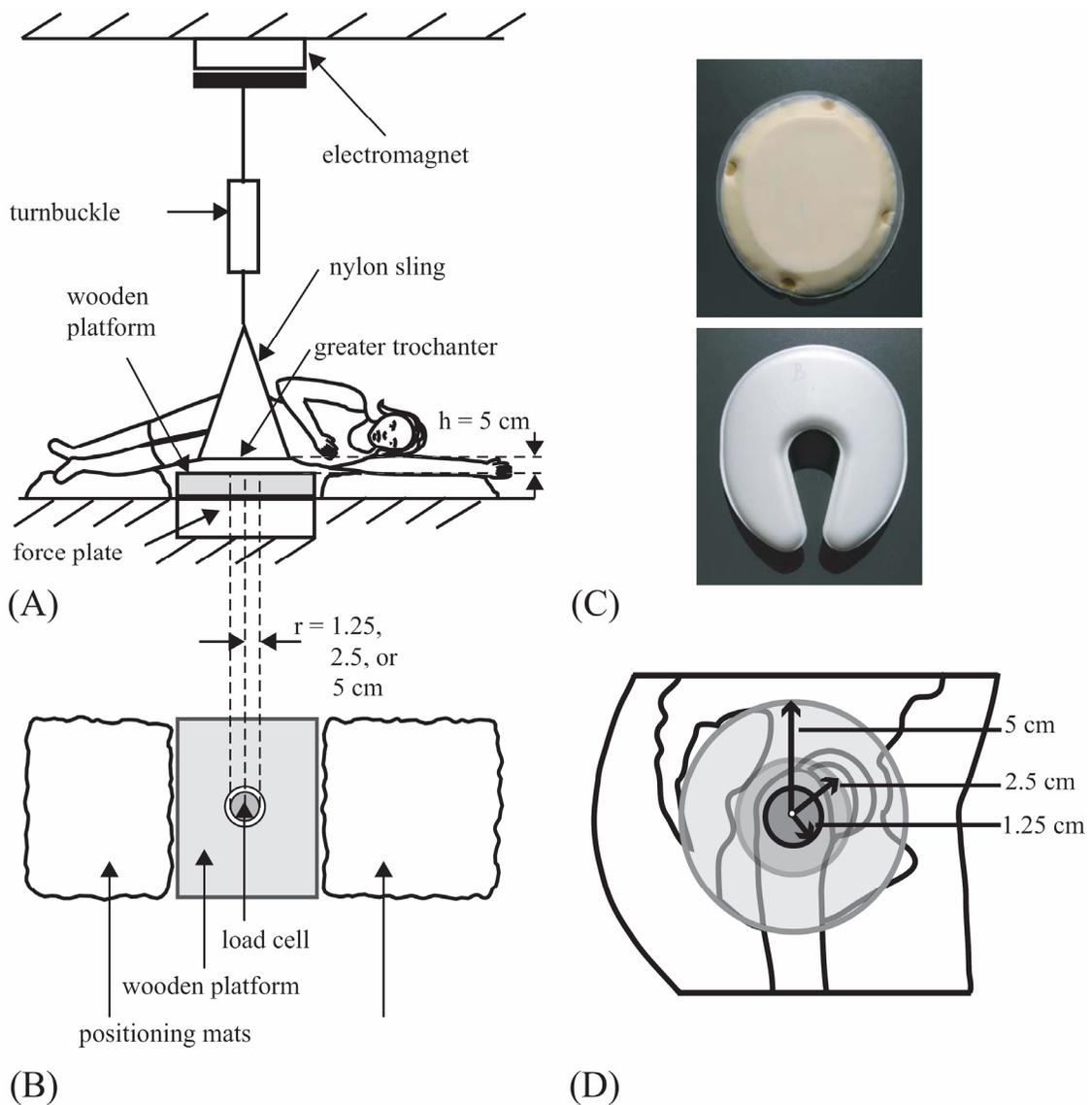
Variable	Unpadded	Continuous protector	Horseshoe protector	df	F	p
$F_{total}$ (N)	1132 (177)	918 (85)	1035 (102)	2, 28	23.6	< 0.001
$F_{GT}$ (N)	202 (126)	55 (11)	49 (33)	2, 28	25.1	< 0.001
$F_{2.5}$ (N)	528 (232)	179 (42)	169 (92)	2, 28	48.6	< 0.001
$F_5$ (N)	784 (202)	513 (68)	524 (93)	2, 28	35.0	< 0.001
$F_{+5}$ (N)	333 (98)	411 (80)	524 (84)	2, 28	37.6	< 0.001
$F_{2.5R}$ (N)	326 (130)	124 (37)	120 (90)	2, 28	30.9	< 0.001
$F_{5R}$ (N)	256 (95)	334 (48)	354 (107)	2, 28	6.2	0.01
$P_{GT}$ (kPa)	413 (257)	112 (22)	99 (68)	2, 28	25.1	< 0.001
$P_{2.5R}$ (kPa)	221 (88)	84 (25)	82 (61)	2, 28	30.9	< 0.001
$P_{5R}$ (kPa)	44 (16)	57 (8)	60 (18)	2, 28	6.2	0.01

**Table 3-2: Pair-wise comparisons of outcome variables in unpadded, continuous protector, and horseshoe protector conditions (n = 15).**

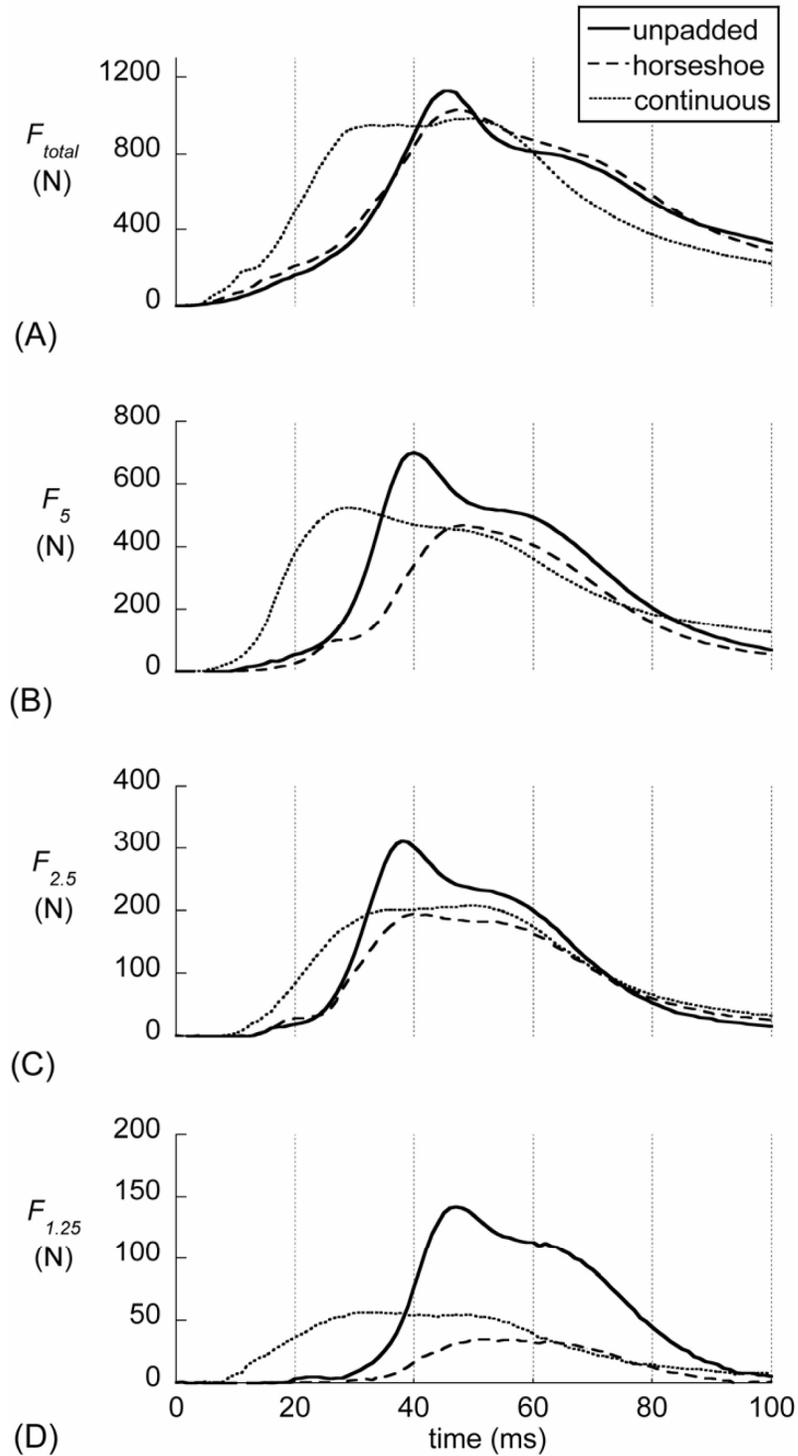
Variable	Reference	Comparison	Mean Difference	<i>p</i> value	95% CI for Difference	
					Lower	Upper
$F_{total}$ (N)	unpadded	continuous	214*	<0.001	121	307
		horseshoe	97*	0.002	44	151
	continuous	horseshoe	-116*	<0.001	-159	-74
$F_{+5}$ (N)	unpadded	continuous	-78*	0.003	-124	-32
		horseshoe	-191*	<0.001	-241	-141
	continuous	horseshoe	-113*	<0.001	-159	-66
$P_{GT}$ (kPa)	unpadded	continuous	301*	<0.001	166	435
		horseshoe	314*	<0.001	190	437
	continuous	horseshoe	13	0.439	-22	49
$P_{2.5R}$ (kPa)	unpadded	continuous	137*	<0.001	92	183
		horseshoe	140*	<0.001	87	192
	continuous	horseshoe	2	0.871	-28	32
$P_{5R}$ (kPa)	unpadded	continuous	-13*	0.004	-21	-5
		horseshoe	-17*	0.018	-30	-3
	continuous	horseshoe	-3	0.477	-14	7

\* Significant at the 0.05 level.

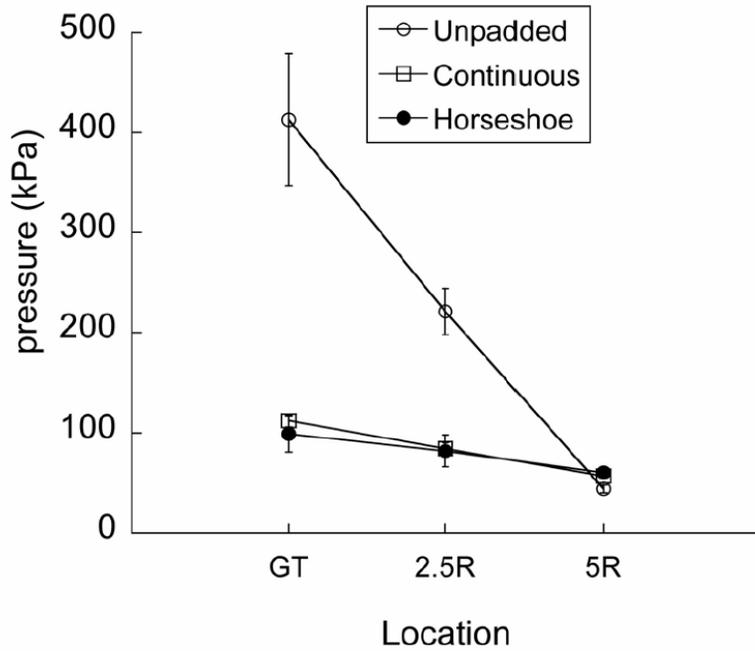
**Figure 3-1: Illustration of experimental methods. (A) Side view and (B) top view of the experimental apparatus. I used a sling and electromagnet to raise and suddenly release the pelvis from a height ( $h$ ) of 5 cm simulating the impact stage of a sideways fall on the hip. A load cell sandwiched between aluminium discs was centred below the greater trochanter (GT) and flush with a wooden platform mounted on top of a force plate. (C) Hip protectors. The 16 mm thick “continuous” protector was centred directly over the GT, while the 14 mm thick “horseshoe” protector covered the region surrounding but not directly over the GT. (D) I calculated the force and pressure in the following regions centred about the GT: a circle of radius 1.25 cm, a ring having an outer radius of 2.5 cm and inner radius of 1.25 cm, and a ring having an outer radius of 5 cm and inner radius of 2.5 cm.**



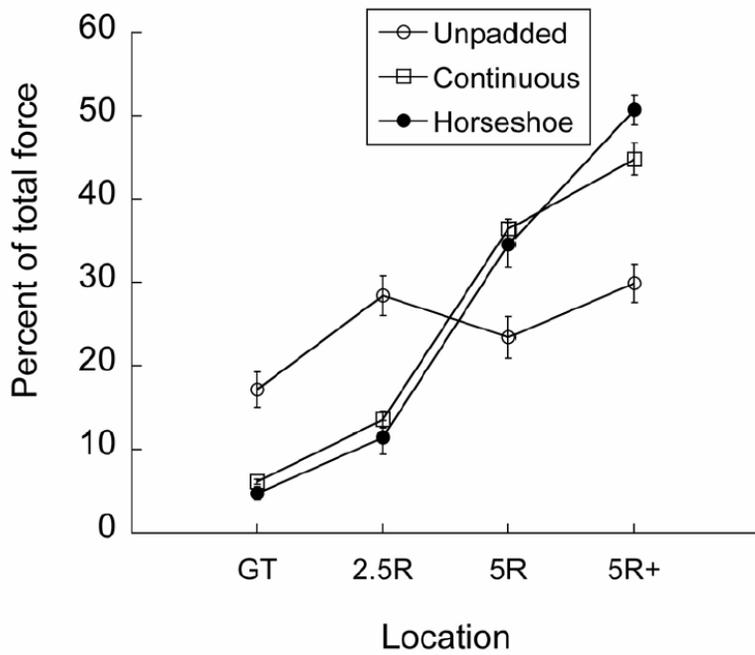
**Figure 3-2: Typical traces of force versus time for a participant in the unpadded, horseshoe, and continuous hip protector conditions. (A) total force measured by the force plate; (B) force measured by the 5 cm radius load cell; (C) force measured by the 2.5 cm radius load cell; (D) force measured by the 1.25 cm radius load cell.**



**Figure 3-3: Effect of hip protectors on (A) the pressure and (B) the percent of total force applied to various hip locations. Both hip protectors caused a dramatic reduction in pressure over the greater trochanter (GT), and redistribution of force to a region at least 2.5 cm outside the GT. Error bars show standard errors.**

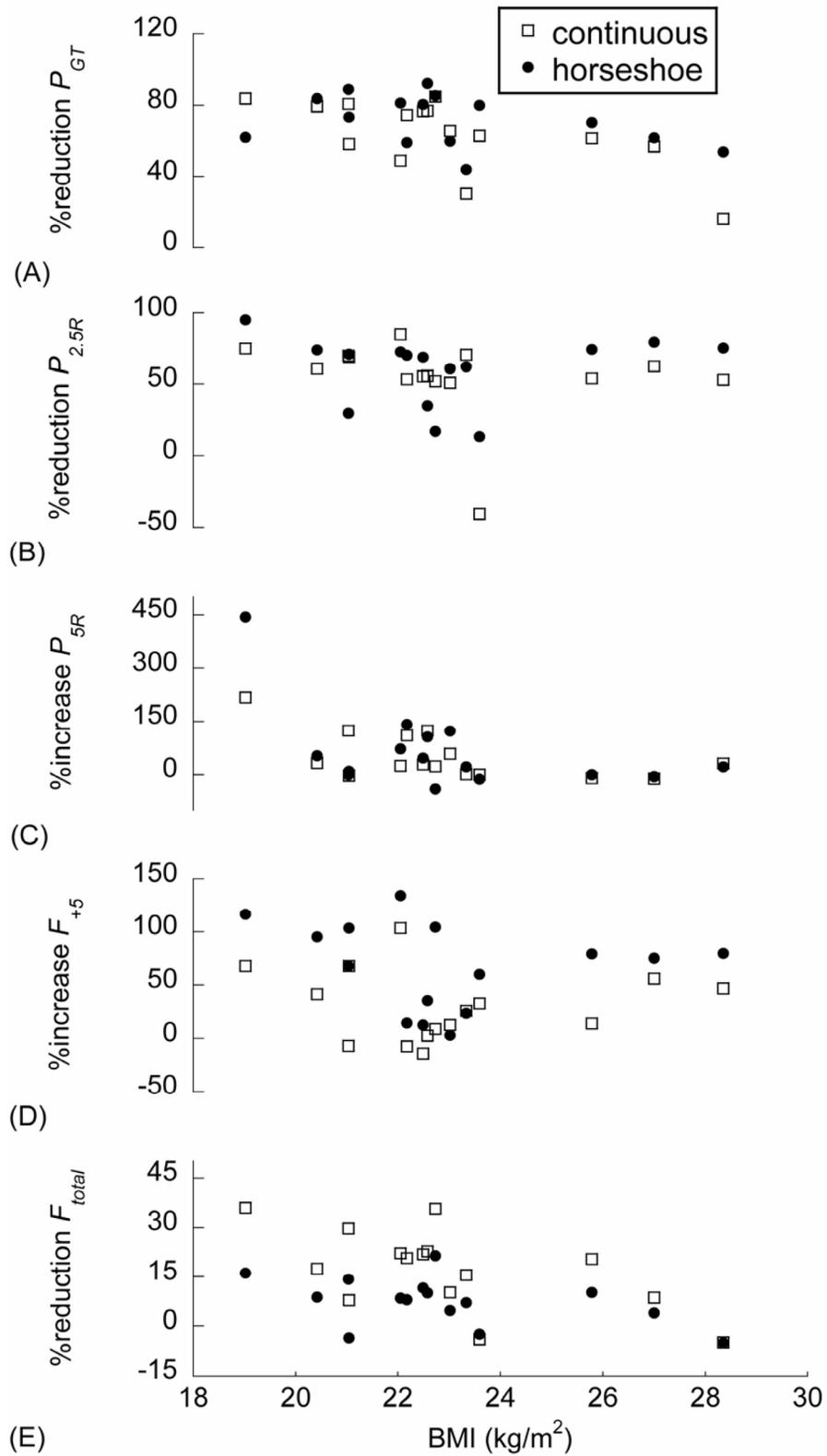


(A)



(B)

**Figure 3-4: Effect of BMI on the percent change in (A)  $P_{GT}$ , (B)  $P_{2.5R}$ , (C)  $P_{5R}$ , (D)  $F_{+5}$ , and (E)  $F_{total}$  provided by the horseshoe and continuous protectors.**



# **CHAPTER 4 THE FORCE ATTENUATION PROVIDED BY HIP PROTECTORS DEPENDS ON IMPACT VELOCITY, PELVIS SIZE, AND SOFT TISSUE STIFFNESS**

## **4.1 Introduction**

Hip fractures are an important world-wide public health issue as they are common, costly, and cause a substantial decline in the quality of life for their sufferers. In Canada alone, there are approximately 23,000 cases of hip fracture every year with associated treatment costs of about \$1 billion (Papadimitropoulos et al., 1997). Approximately 20% of older adults hospitalized for hip fracture will die within one year, and about 50% will suffer a major decline in independence (Empana et al., 2004a, Wolinsky et al., 1997). Furthermore, given the aging of the population and the exponential increase in fracture risk with age, hip fracture incidence is expected to increase 4-fold by the year 2041 (Jaglal et al., 1996, Papadimitropoulos et al., 1997, Wiktorowicz et al., 2001).

External hip protectors (undergarments with protective padding in the pelvic region) represent a promising strategy for reducing hip fracture risk in elderly persons. However, there is a lack of market regulation and contradictory evidence of the clinical value (Kannus et al., 2000, Kiel et al., 2007, Parker et al., 2006, Sawka et al., 2005) and biomechanical performance of hip protectors (Derler et al., 2005, Kannus et al., 1999, Mills, 1996, Nabhani and Bamford, 2002, Robinovitch et al., 1995a, van Schoor et al., 2006). For example, reports of the attenuation in peak force applied to the proximal femur provided by one popular product (Safehip Classic) range from 18.2% (Robinovitch

et al., 1995a) to 83.7% (Nabhani and Bamford, 2002), and for another (Safety pants FI) from 18.7% (van Schoor et al., 2006) to 78.9% (Kannus et al., 1999). While the hip protectors used in these studies were essentially identical, the studies used different test systems or protocols (including different impact energies) for simulating the impact biomechanics of a sideways fall (see section 1.3.1.2.2 and table 1.4). Such discrepancies indicate the need for developing accurate and internationally accepted biomechanical testing standards to guide the market approval and selection of hip protectors for clinical trials.

In the current study, my goals were (a) to develop a testing system that accurately simulated the impact biomechanics of a sideways fall; (b) to utilize this system to determine the force attenuation provided by existing hip protectors; and (c) to determine, through a sensitivity analysis, how the measured force attenuation is influenced by variations in the testing system characteristics. In designing a realistic (or “biofidelic”) test system, and in conducting my sensitivity analysis, I focused on three specific variables that I hypothesized would influence the measured force attenuation. The first was impact velocity, a measure of fall severity which may influence the stiffness and deformation of both the protector and the underlying soft tissues, and therefore the percent of total force delivered to the proximal femur. The second was pelvic size, which may influence force attenuation by altering the contact area, or the energy shunting provided by specific products (e.g. by altering the gap between a rigid shell dome protector and the skin surface (Mills, 1996, Minns et al., 2004a)). Finally, the third variable of interest was the stiffness of the soft tissues over the hip region, an index of

baseline padding which may influence force attenuation through a springs-in-series effect, or by influencing how the protector intrudes into the soft tissues.

## **4.2 Methods**

### **4.2.1 Experimental measures with elderly women**

Study participants consisted of fifteen women with a mean age of 77.5 years (SD 8.5, range 66 to 91), mean body mass of 61.2 kg (SD 7.4, range 48.4 to 72.6), mean height of 1.61 m (SD 0.07, range 1.51 to 1.75), and mean body mass index of 23.6 kg/m<sup>2</sup> (SD 2.1, range 21.2 to 28.9). All participants provided written informed consent and the study was approved by the Committee on Research Ethics at Simon Fraser University.

#### **4.2.1.1 Surface geometry**

I measured pelvic surface geometry with an 8-camera motion measurement system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 120 Hz. This system is accurate to within 1 mm when calibration procedures are followed. The participant wore spandex shorts and a tightly fitted t-shirt. I applied reflective markers to the pelvis in a 10 x 10 grid centred about the left greater trochanter (as determined by manual palpation) and spaced 3.2 cm apart (figures 4.1A and B). I applied one row at the level of greater trochanter, four rows superiorly, and five rows inferiorly; four columns were anterior to the greater trochanter, and five columns were posterior. I also placed markers on the right greater trochanter, and on the sacrum and abdomen at the level of the greater trochanter, and used these to calculate pelvic depth along the frontal midline (sacrum to abdomen), pelvic width along the sagittal midline (left to right greater trochanter), and the pelvic midpoint (intersection of the frontal and sagittal midlines).

During the measures, the participant stood comfortably with her arms crossed over her chest, and the investigator ensured there was no movement or distortion of the spandex shorts and markers from their intended location.

To determine the average hip surface geometry of the 15 subjects, I first used AutoCad software (Autodesk Inc., San Rafael, CA, USA) to determine the straight-line distance from each marker to the pelvis midpoint, and subtracted the radius of the marker to the skin surface. A non-uniform rational B-spline was then fit through the coordinates of the 10 markers in each horizontal row. I then aligned the frontal and sagittal midlines of all subjects, and calculated the average location of the splines (table 4.1) in each transverse plane at 10 angles: the zero or reference angle (along the line connecting the pelvic midpoint to the left greater trochanter), at 10, 20, 30, 40, and 50 degrees posterior to the reference angle, and at 10, 20, 30, and 40 degrees anterior to the reference angle (figure 4.1C). The radius of curvature of a best-fit circle was also calculated at each transverse plane (table 4.1). It was greatest (123.7 mm) for the transverse plane 6.4 cm superior to the greater trochanter, and decreased gradually as the buttock tapered into the upper thigh to a minimum (84.6 mm) for the plane 16 cm inferior to the greater trochanter. The radius of curvature for the coronal plane in line with the greater trochanter was 551 mm. The average width of the pelvis (left to right greater trochanter) was 362 (SD 16) mm; the average depth of the pelvis (sacrum to abdomen) was 266 (16) mm.

#### **4.2.1.2 Soft tissue force-deflection properties**

I determined the stiffness of the soft tissues overlying and adjacent to the greater trochanter using a customized indentation device composed of a 3.8 cm diameter

cylindrical probe which slid inside a guide unit (figure 4.2A) (Robinovitch et al., 1995a). I measured the force applied to the probe with a load cell (model MLP100, Transducer Techniques, Temecula, CA, USA), and the probe deflection with a linear position transducer (model MLT 38000102, Honeywell, Freeport, IL, USA). Both force and deflection were sampled at 1000 Hz. One end of the position transducer was attached to the probe, and the other contacted a plexiglass flange (of 15 x 15 cm surface area), which rested upon the surrounding relaxed skin during indentation. This ensured that the measured deflection was relative to the initial skin surface position, and unaffected by movements of the entire pelvis.

During data collection, the participant stood with her right hip positioned against a rigid wall. In this stance position I assumed there to be a moderate degree of contraction of the gluteal and quadriceps muscles. Stiffness was measured at nine locations, spaced in a three by three grid (6 cm vertical and horizontal spacing between adjacent locations) centred about the left greater trochanter (inset to figure 4.2C). During each trial, the investigator compressed and then relaxed the underlying tissue by applying force via the hand to the probe handle. At each location, three measures were acquired, and the results averaged. I used customized data acquisition and analysis software (Labview 6.1, National Instruments, Austin, TX, USA), and repeated trials that did not meet my inclusion criteria of a loading rate between 135 and 165 N/s and a maximum compressive force between 114 and 160 N. Furthermore, I inserted five minute breaks between trials to allow adequate reperfusion and tissue relaxation. Similar to Robinovitch et al. (1995a), the outcome measure from each trial was the secant stiffness over the force range of 90 –

110 N (figure 4.2B). While this total force is much lower than that accompanying a fall, so is the contact area, thus the pressure or tissue stress may be comparable.

I found that soft tissue stiffness varied substantially across pelvic locations (figure 4.2C), being stiffest directly over the greater trochanter (averaging 34.4 kN/m (SD 15.5)) and least stiff 6 cm posterior to the trochanter (averaging 14.1 kN/m (SD 7.2)).

#### **4.2.2 SFU Hip Impact Simulator**

The Simon Fraser University hip impact simulator (figure 4.3A) is an improved version of the Robinovitch et al. (1995a) test system, and measures the total force applied to the skin overlying the hip region and the force delivered to the femoral neck during a simulated sideways fall. The system consists of an impact pendulum and surrogate pelvis (figure 4.3B), which is released by an electromagnet from an inclined position and strikes the ground in a horizontal position. The surrogate pelvis consists of simulated soft tissues and a proximal femur (Sawbones, Vashon, WA, USA) mounted on a 25.0 x 25.0 x 0.7 cm polyvinyl chloride base plate. The proximal femur was 13 cm in length from its distal end to the centre of the greater trochanter. The femur was fixed to the base plate with a single bolt directly behind the greater trochanter, leaving the distal end free.

The surrogate pelvis was designed to match the average surface geometry and local variation in soft tissue stiffness measured in my elderly participants. The soft tissues were simulated with closed-cell polyethylene foams (Plastazote HD80 of density 80 kg/m<sup>3</sup> and LD45 of density 45 kg/m<sup>3</sup>) directly over the proximal femur, and closed-cell copolymer foam (Evazote EV50) over the regions anterior, posterior, and superior to the femur. These materials were glued together to form a single 21.6 x 24.5 x 8.0 cm block. A 1.2 cm thick layer of open-cell ester foam (SCH180-60E1 of density 29 kg/m<sup>3</sup>) was

secured over the entire outer surface of the pelvis, along with a 1.6 mm layer of gum rubber to simulate skin. The thickness of the soft tissues overlying the greater trochanter was 2.4 cm. Through an iterative process, I determined the combination of material type and thickness that matched to within one SD the mean soft tissue stiffness of my elderly participants at all nine pelvic locations using the same indentation test (figure 4.2C). The accuracy of the match (where a negative value indicates a lower stiffness in the surrogate pelvis) ranged from -11.9% to 26.8%, with an average of 3.2% (SD 10.9).

I acknowledge that damping does exist in the various soft tissues in the pelvis (fat, ligament, muscle, tendon), as indicated by the decay in oscillations following peak force (see Chapters 2 and 5). However, the effect of damping on peak force is not accurately described by a viscous (velocity-dependent) damping force. For falls simulated with a mechanical test system, Robinovitch et al. (1997b) demonstrated that the inclusion of damping elements in Voigt and standard linear solid models did not improve the accuracy of peak force predictions compared to a simple mass-spring model. In addition, in Chapter 2 my mass-spring model was relatively accurate at predicting peak force across three different impact velocity conditions, indicating that the overall effect on peak force of damping mechanisms (versus elastic mechanisms) was relatively minor. For these reasons I felt justified in not explicitly incorporating damping elements into my mechanical hip impact simulator.

The surrogate pelvis is connected to the pendulum via leaf springs which simulate the total effective stiffness of the articulations between the pelvis, trunk, and lower extremities, and the compressive stiffness of the pelvis itself (minus that provided by the peripheral soft tissue layer) (Robinovitch et al., 1997a). The force applied to the femoral

neck is measured by a uni-axial load cell (Kistler Model 9712A5000, Amherst, NY, USA), and the total force applied to the skin surface is measured with a floor-mounted force plate (model 2535-08, Bertec Corp., Columbus, OH, USA). During dynamic loading with the surrogate soft tissues removed the peak force recorded from the load cell is within 2% of that measured by the force plate. Bending moments applied to the load cell are minimal as the system was constructed such that the surrogate pelvis is oriented flush to the force plate surface, and is moving vertically, at the moment of impact. This was confirmed through observations of minimal shear forces in the force plate throughout the impact event. The impact velocity of the surrogate pelvis is varied by adjusting the initial angle of the pendulum before release. The effective mass and stiffness of the entire system, measured from its free vibration response (Robinovitch et al., 1995a, 1997b), are 28.0 kg and 42.2 kN/m, respectively. These values are within one SD of the mean values measured in simulated falls on the hip with young women (Robinovitch et al., 1997a).

#### **4.2.2.1 Experiments with the impact simulator**

I used the hip impact simulator to measure the force attenuation provided by three commercially available hip protectors (figure 4.3C). Safehip Classic (Tytex Inc., Ikast, DE) is a hard shell protector designed to absorb and shunt energy away from the proximal femur during impact. The shell has an inner core of copolymer polypropylene surrounded by expanded polypropylene foam. The total length, width, and height of the shell are 15.8, 11.3, and 2.5 cm, respectively. Safehip Soft (Tytex Inc., Ikast, DE) is a soft shell protector made from closed-cell cross-linked polyolefin. The protector has a horseshoe-shaped geometry with maximum length, width, and thickness of 18.3, 17.0, and 1.6 cm, respectively. The protector is designed to surround, but not directly cover,

the greater trochanter thereby forming a bridge over the bone. Hipsaver SlimFit (Hipsaver Canada, Exeter, ON, Canada) is a soft shell hip protector comprised of an oval-shaped pad of viscoelastic open-cell foam encapsulated in a nylon sleeve. The model I tested has a length, width, and thickness of 19, 17, and 1.6 cm, respectively. All hip protector shells came sewn in undergarments, and were tested with the protectors positioned according to manufacturer directions. In all cases, the surrogate pelvis was of sufficient size to provide a border of at least 3 cm between the outer edges of the hip protector and the boundaries of the soft tissues.

In experiment 1, I measured the force attenuation provided by each hip protector when tested on the biofidelic pelvis. Trials were performed at impact velocities of 1, 2, 3, and 4 m/s. The corresponding impact energies were 14, 56, 126, and 224 J, respectively.

In experiment 2, I examined how the force attenuation provided by each hip protector was affected by pelvis size. I used an ellipse to represent the surface geometry of the pelvis (Feldman and Robinovitch, 2007). Pelvic width (major axis length) was held constant at 362 mm (the average value observed in my participants), while pelvic depth (minor axis length) was varied between 206 mm (small), 266 mm (medium), or 326 mm (large) (figure 4.3D). I positioned the base plate 100 mm lateral to the ellipse centre, resulting in a thickness of soft tissues over the greater trochanter of 30 mm. Trials were performed at an impact velocity of 3 m/s, similar to the average value observed in unexpected sideways falls in young adults (Feldman and Robinovitch, 2007). In all trials the soft tissues were simulated with closed-cell, expanded ethylene vinyl acetate foam (SVA60; density =  $59 \text{ kg/m}^3$ ; stiffness as measured in my indentation protocol = 12.5 kN/m).

In experiment 3, I determined how the force attenuation provided by each hip protector was influenced by pelvic soft tissue stiffness. The tissue stiffness varied between ‘soft’ (open-cell rebond polyurethane foam; density = 96 kg/m<sup>3</sup>; stiffness = 6.5 kN/m), ‘semisoft’ (closed-cell copolymer foam; Ensolite MLC-B; density = 88 kg/m<sup>3</sup>; stiffness = 8.2 kN/m), ‘semifirm’ (closed-cell, expanded ethylene vinyl acetate foam; SVA60; density and stiffness as described above), ‘firm’ (closed-cell polyethylene foam; LD45; density = 45 kg/m<sup>3</sup>; stiffness = 35.8 kN/m), and ‘rigid’ (expanded polystyrene foam; PlastiSpan HD; density = 22 kg/m<sup>3</sup>). For all conditions, I used an elliptical pelvic surface geometry with a major axis length of 362 mm and a minor axis length of 266 mm (average participant values). Trials were performed with an impact velocity of 1 m/s to prevent damage to the test system. Even with this precaution, peak forces in some stiffness conditions were ~200% higher, and were reached ~20% faster, than observed with the biofidelic pelvis.

#### **4.2.3 Data analysis**

For each experimental condition, I collected three unpadded trials, followed by three sequential trials with each hip protector presented in a random order (figure 4.4). Force was collected for two seconds at 1000 Hz, and filtered with a dual-pass fourth-order Butterworth low pass filter with a 35 Hz cut-off frequency (Labview 6.1, National Instruments, Austin, TX, USA). For each trial, I identified the peak total force ( $F_{total}$ ) and peak femoral neck force ( $F_{neck}$ ). For each hip protector, the attenuation in total force ( $F_{total\_atten}$ ) and femoral neck force ( $F_{neck\_atten}$ ) was calculated as the average percentage decrease in  $F_{total}$  and  $F_{neck}$ , respectively, between the padded and unpadded conditions.

#### 4.2.4 Statistics

For experiment 1, I used two-factor randomized groups ANOVA to determine whether  $F_{neck\_atten}$  and  $F_{total\_atten}$  associated with impact velocity (1, 2, 3, 4 m/s) and hip protector condition when tested with the biofidelic surrogate pelvis. Similarly, two-factor ANOVA was used to determine whether  $F_{neck\_atten}$  and  $F_{total\_atten}$  associated in experiment 2 with pelvic size (small, medium, large) and hip protector condition, and in experiment 3 with soft tissue stiffness (soft, semisoft, semifirm, firm, rigid) and hip protector condition. When ANOVA results indicated a significant association, I used independent t-tests to identify differences between conditions. All analyses were performed with statistical analysis software (SPSS Version 15.0, SPSS Inc., Chicago, IL, USA) using a significance level of  $\alpha = 0.01$ .

#### 4.3 Results

When tested with the biofidelic surrogate pelvis (experiment 1), there were significant differences in  $F_{neck\_atten}$  ( $F_{2,24} = 325.8$ ,  $p < 0.001$ ) and  $F_{total\_atten}$  ( $F_{2,24} = 63.0$ ,  $p < 0.001$ ) across hip protectors (table 4.2). Average values of  $F_{neck\_atten}$  were 26.8 (SD 8.0) % for Safehip Soft, 19.0 (4.1) % for Hipsaver, and 16.9 (11.0) % for Safehip Classic. The magnitudes of  $F_{neck\_atten}$ , and the differences between hip protectors, were much larger than those observed for  $F_{total\_atten}$  which averaged 2.8 (1.4) % for Safehip Soft, 4.6 (0.5) % for Hipsaver, and 3.8 (3.4) % for Safehip Classic. The average coefficient of variation across all hip protector and impact velocity conditions was 0.4% for  $F_{total}$  and 1.3% for  $F_{neck}$ .

ANOVA results also indicated that impact velocity associated with  $F_{neck\_atten}$  ( $F_{3,24} = 205.0$ ,  $p < 0.001$ ) and  $F_{total\_atten}$  ( $F_{3,24} = 151.0$ ,  $p < 0.001$ ) (table 4.2). In addition, I noted

significant interactions between hip protector and impact velocity for  $F_{neck\_atten}$  ( $F_{6,24} = 176.6, p < 0.001$ ) and  $F_{total\_atten}$  ( $F_{6,24} = 58.7, p < 0.001$ ). As the impact velocity increased from 1 to 4 m/s,  $F_{neck\_atten}$  decreased for Safehip Classic (from 33.5 to 8.1%) and for Safehip Soft (from 33.5 to 18.9%), but increased for Hipsaver (from 12.9 to 20.6%). As a result, there were different rank orders of hip protectors across impact velocities. Although significant, the differences in  $F_{total\_atten}$  observed within and across hip protectors were less pronounced (table 4.2).

In experiment 2, ANOVA results indicated that pelvic size was significantly associated with  $F_{neck\_atten}$  ( $F_{2,18} = 135.2, p < 0.001$ ), and that its effect on  $F_{neck\_atten}$  differed across hip protectors ( $F_{4,18} = 238.5, p < 0.001$ ) (table 4.3). For Safehip Soft,  $F_{neck\_atten}$  increased from 22.5 in the small condition to 29.5% in the large condition. In contrast, it decreased for Safehip Classic (from 13.6% to 1.7%) and Hipsaver (from 27.5% to 19.0%) between the small and large conditions. Overall, pelvic size did not affect  $F_{total\_atten}$  ( $F_{2,18} = 1.7, p = 0.206$ ), but it did have different effects on  $F_{total\_atten}$  across hip protectors ( $F_{4,18} = 14.5, p < 0.001$ ).

In experiment 3, soft tissue stiffness had a significant influence on  $F_{neck\_atten}$  ( $F_{4,30} = 1318.8, p < 0.001$ ) (table 4.3). The effect differed across hip protectors ( $F_{8,30} = 274.9, p < 0.001$ ).  $F_{neck\_atten}$  for Hipsaver was lowest in the semifirm condition (17.8%), and increased moderately in the soft and rigid conditions (to 35.6% and 38.6%, respectively). From the soft to rigid condition,  $F_{neck\_atten}$  increased more dramatically from 35.8% to 60.2% for Safehip Soft, and from 26.0% to 75.7% for Safehip Classic. Soft tissue stiffness was also associated with  $F_{total\_atten}$  ( $F_{4,30} = 165.3, p < 0.001$ ), and had different

effects across hip protectors ( $F_{8,30} = 91.8, p < 0.001$ ). However, the changes were much smaller than those observed for  $F_{neck\_atten}$  (table 4.3).

#### 4.3.1 General observations

In the unpadded (baseline) condition, increasing impact velocity substantially increased both  $F_{neck}$  and  $F_{total}$  (figure 4.5A). However, only  $F_{neck}$  was substantially influenced by pelvis size and soft tissue stiffness (figures 4.5B and C). Furthermore, increasing the size of the pelvis caused a decrease in the percent of total force delivered through to the femoral neck (from 77% in the small condition to 47% in the large condition). For all of the conditions I studied,  $F_{total\_atten}$  was smaller in magnitude, and varied to a lesser degree, than  $F_{neck\_atten}$ .

#### 4.4 Discussion

In the current study, I developed a biofidelic hip impact simulator and used it to measure the attenuation in peak femoral neck force ( $F_{neck\_atten}$ ) provided by three commercial hip protectors (Safehip Classic, Safehip Soft, Hipsaver). I found  $F_{neck\_atten}$  was significantly influenced by impact velocity. The most commonly used hip protector in clinical trials (Safehip Classic) attenuated femoral impact force by 33% during a low velocity (1 m/s) fall, but only 8% for a high velocity (4 m/s) fall. Under these higher severity falls, greater force attenuation (of 19 and 21%, respectively) was provided by Safehip Soft and Hipsaver – the two soft shell protectors. I also found that  $F_{neck\_atten}$  was significantly associated with two other test system features – pelvis size, and soft tissue stiffness. In general,  $F_{neck\_atten}$  increased with decreasing pelvis size and increasing soft tissue stiffness. However, significant interaction effects caused the rank order of hip

protectors to depend on testing conditions. Increases in pelvis size caused  $F_{neck\_atten}$  to increase for Safehip Soft, and decrease for Hipsaver and Safehip Classic. Safehip Soft and Safehip Classic provided approximately two-fold greater  $F_{neck\_atten}$  for the rigid vs. low stiffness soft tissues; the increase was less dramatic for Hipsaver.

For all conditions I examined, hip protectors had a much greater influence on  $F_{neck\_atten}$  than on  $F_{total\_atten}$ . When tested with the biofidelic surrogate pelvis, the largest value of  $F_{total\_atten}$  I observed was 9%, while the largest value of  $F_{neck\_atten}$  was 34%. These data imply that force shunting, as opposed to energy absorption, is the primary means of protection, even for continuous soft shell protectors such as Hipsaver.

The complex influence of impact velocity on force attenuation was likely related to the distinct design principles employed by each protector. For the hip protectors which form energy shunting bridges over the greater trochanter (Safehip Soft and Safehip Classic),  $F_{neck\_atten}$  decreased as impact velocity increased. This was likely a consequence of increased intrusion of the protector edges into the soft tissues surrounding the greater trochanter, which compromised the height of the energy shunting bridge. As opposed to forming a bridge, the continuous oval design of the Hipsaver protector attenuates force by absorbing energy and smoothing out the variability in stiffness throughout the hip region (see Chapter 3). Consequently, while  $F_{neck\_atten}$  for Hipsaver was the lowest of all three protectors at 1 m/s, it matched that of Safehip Soft at 4 m/s.

Pelvis size also had a complex effect on the force attenuation provided by hip protectors. For unpadding trials, the percentage of total force transmitted to the femoral neck increased with decreases in pelvis size, likely the result of a decrease in total contact area.  $F_{neck\_atten}$  was further modified by the match between hip protector geometry and the

pelvis surface. For example,  $F_{neck\_atten}$  for Safehip Soft was dependent in part on the pad legs directing energy away from the proximal femur. Since the pad followed the contours of the pelvis, a decrease in pelvis size may have caused a decrease in the contact area of the pad during impact, and a subsequent decrease in  $F_{neck\_atten}$ . The reduction in contact area may have been less dramatic for Hipsaver, as it directly overlies the proximal femur. For Safehip Classic, I was surprised to observe an increase in  $F_{neck\_atten}$  with decreasing pelvis size. To better understand this phenomenon, I examined the interface between the hard shell protector (removed from the garment) and skin surface across pelvis conditions. Although the height of the bridge over the greater trochanter differed only slightly (12 mm for the small and medium pelvises, 16 mm for the large pelvis), the shell's primary point of skin contact, and thus the major route for force shunting, differed substantially across conditions. The large pelvis had a transverse radius of curvature larger than the rigid shell protector. Consequently, the lateral edges of the protector were the primary points of contact with the pelvis surface. In contrast, the decreased radius of curvatures in the medium and small pelvis conditions resulted in the superior and inferior edges of the protector being the primary contact points. This may improve the force attenuation provided by Safehip Classic for smaller pelvises by enhancing the energy shunting pathway superior to the greater trochanter.

For all three protectors, force attenuation increased with increases in soft tissue stiffness. However, the underlying mechanisms differed across hip protectors. For the protectors primarily designed as energy shunters (Safehip Classic and Safehip Soft), this probably arose from decreased intrusion of the edges of the protector into the tissues surrounding the greater trochanter at higher tissue stiffnesses. Hipsaver's force

attenuation was more dependent on the stiffness of the tissues *directly* overlying the greater trochanter (table 4.3), and can be explained through springs-in-series effects. For conditions in which the tissues overlying the greater trochanter stiffened (either due to the foam tissues ‘bottoming out’ in the soft and semisoft conditions, or as a consequence of the higher stiffness materials used in the ‘firm’ and ‘rigid’ conditions), the compliant Hipsaver more effectively decreased the total stiffness of the system, and caused a greater reduction in  $F_{neck}$ .

It is insightful to compare my measures from elderly women to those used in hip protector testing rigs reported in the literature. The mean transverse radius of curvature at the level of the greater trochanter that I observed for my sample of elderly women (116 mm) was larger than the 55 to 100 mm used by some authors (Derler et al., 2005, Mills, 1996, Nabhani and Bamford, 2002), and impossible to compare to other studies which did not specify pelvis size (Kannus et al., 1999, Minns et al., 2004a, Wiener et al., 2002). Unfortunately, even fewer studies have specified the stiffness of the soft tissues incorporated into mechanical test systems. My results are similar to those of Robinovitch et al. (1995a), who reported the regions directly overlying the GT (~35 kN/m) to be two- to three-fold stiffer than the tissues six cm anterior and posterior. In contrast, Mills’ (1996) system utilized substantially stiffer tissues (70 kN/m). Based on my observations of the significant influence of pelvic size and soft tissue stiffness, differences in these characteristics across test systems may in part explain the variation in biomechanical test results reported in the literature (Derler et al., 2005, Kannus et al., 1999, Mills, 1996, Nabhani and Bamford, 2002, Robinovitch et al., 1995a, van Schoor et al., 2006) (see

section 1.3.1.2.2 and table 1.4 for a more detailed comparison of the characteristics and protocols of the mechanical test systems reported in the literature).

The current study has significant strengths and clinical implications. I used a systematic approach to demonstrate that incorporating accurate surface geometry and soft tissue stiffness characteristics into a test system is essential if valid and meaningful estimates of the force attenuation provided by hip protectors are to be produced. By incorporating physiologic pelvic surface geometry, soft tissue stiffness, effective mass, effective stiffness, and impact velocities into my hip impact simulator, I believe it to be the most biofidelic test system reported in the literature. My results suggest that women with average pelvis size and little natural padding (and thus high soft tissue stiffness) should benefit more from Safehip Classic or Safehip Soft than Hipsaver. However, when tested with the biofidelic surrogate pelvis at 4 m/s,  $F_{neck\_atten}$  for the most common clinically tested hip protector (Safehip Classic) was only 8%. Both Safehip Soft (19%) and Hipsaver (21%) performed considerably better for this worst-case fall scenario. However, in general I found the force attenuation provided by hip protectors to be lower than most previous reports (Derler et al., 2005, Kannus et al., 1999, Nabhani and Bamford, 2002, van Schoor et al., 2006). This may in part explain the lack of positive findings in several recent clinical trials of hip protectors (Cameron et al., 2001, Kiel et al., 2007, van Schoor et al., 2003b), and indicates the need to develop a new generation of hip protectors with improved biomechanical effectiveness.

This study also has several important limitations. First, while the characteristics of my biofidelic surrogate pelvis were based on a sample of women with body mass index's slightly below average, there is no consensus on the population upon which mechanical

test systems should be modelled. However, I feel justified modelling this population as women with low body mass index have a higher risk of hip fracture, presumably due to a decreased amount of soft tissues overlying the proximal femur (Keegan et al., 2004). Second, I did not investigate how different levels of muscle activity might influence the stiffness of the pelvic soft tissues. However, I don't believe this to be a substantial confounder as Robinovitch et al. (1997a) found muscle contraction to have no influence on the total effective stiffness of the hip during sideways falls. Third, there is no consensus on the most appropriate materials to use in a mechanical test rig for simulating the mechanical properties of human skin, fat, and muscle. Mills (2007) suggests that oil-extended jelly-like polymers are most appropriate due to their compressive and inertial properties. However, I feel justified in using foam materials based on their match to the force-deflection properties measured from my elderly female participants. Fourth, under my worst case impact condition (where  $F_{total}$  equalled 3.9 kN), the surrogate pelvis experienced a total deflection of approximately 9 cm. While this may appear excessive, biomechanical testing of cadavera confirm the substantial compliance of the pelvis during sideways impacts. In particular, for an individual having a pelvic width of 362 mm (the average measured from my 15 elderly participants), a 9 cm deflection corresponds to 25.4% compression of the pelvis. This is similar to the 27.8% (Cavanaugh et al., 1990) and 23.9% (Etheridge et al., 2005) compressions observed during high-velocity side-impact tests with cadavera that did not result in proximal femur or pelvic fractures. Furthermore, although the time to reach peak force observed for my test system (49 ms for an unpadding condition with a 3 m/s impact velocity) was higher than the ~25 ms reported by Parkkari et al. for pelvic impacts with four healthy young men (1997), it was

similar to that observed during simulated falls with young women (see Chapters 3 and 5). Fifth, while I used a relatively low impact velocity (1 m/s) to investigate the influence of soft tissue stiffness, members of our laboratory have observed values as low as 0.8 m/s in falls from standing (Feldman and Robinovitch, 2007, Robinovitch et al., 2000a). Consequently, I consider my 1 m/s impact velocity condition to simulate a low severity, but clinically relevant, sideways fall. Finally, I acknowledge that my mechanical test system, while representing an improvement on previously described systems, still may not simulate the full range of physiologic characteristics of the human pelvis which influence the force attenuation provided by hip protectors. This indicates the value of additional experiments with living humans (Parkkari et al., 1997) (see Chapter 3) and cadaveric tissues (Bouxsein et al., 2007, Eckstein et al., 2004, Lauritzen and Askegaard, 1992, Robinovitch et al., 1995b) to provide important data for the design and validation of mechanical test systems.

Evidence of the biomechanical basis for hip protectors needs to be enhanced to inform clinical and industry efforts related to hip fracture prevention (Kannus et al., 2003). The current study indicates that the force attenuation provided by three commercially available hip protectors is dependent on the severity of a fall (as modified through impact velocity). For the typical elderly women suffering a worse-case fall, the most commonly used hip protector in clinical trials (Safehip Classic) will attenuate femoral impact force by only 8%. Improved force attenuation is observed in Safehip Soft and Hipsaver, but this does not exceed 21%. Erroneous force attenuations (of up to 76%) and relative rankings will be observed if the pelvis is too large or if the soft tissues are too soft or too stiff. The sensitivity of the measured force attenuation to these various test

parameters in part explains the variability in performance reported by different research groups, and indicates the need for international standards for biomechanical tests systems to evaluate the market suitability of hip protectors. My results also demonstrate how testing systems must be designed to accurately measure force attenuation, a crucial step to developing clinically effective hip protectors.

**Table 4-1: Three dimensional coordinates describing the average pelvic surface geometry of 15 elderly women relative to the greater trochanter (GT).**

Vertical (z) level (mm) <sup>a</sup>	x-y coordinates (mm)										radius of curvature (mm)
	anterior-posterior angle (degrees) <sup>b</sup>										
	+40	+30	+20	+10	0 (GT)	-10	-20	-30	-40	-50	
128	-108, 91	-125, 72	-140, 51	-151, 27	-156, 0	-156, -27	-149, -54	-135, -78	-115, -97	-92, -110	116.5
96	-112, 94	-130, 75	-146, 53	-160, 28	-167, 0	-167, -30	-160, -58	-145, -84	-124, -104	-99, -118	120.2
64	-114, 96	-134, 77	-152, 55	-167, 29	-175, 0	-175, -31	-166, -61	-151, -87	-130, -109	-104, -124	123.7
32	-113, 95	-134, 78	-156, 57	-172, 30	-179, 0	-178, -31	-168, -61	-152, -88	-132, -111	-107, -128	122.8
0 (GT)	-109, 91	-132, 76	-156, 57	-175, 31	-181, 0	-177, -31	-165, -60	-149, -86	-130, -109	-106, -127	116.0
-32	-105, 88	-132, 76	-158, 57	-177, 31	-184, 0	-178, -31	-165, -60	-148, -85	-128, -107	-104, -124	109.1
-64	-103, 86	-133, 77	-159, 58	-177, 31	-184, 0	-180, -32	-166, -60	-145, -84	-123, -103	-98, -117	102.8
-96	-102, 86	-134, 77	-158, 58	-175, 31	-183, 0	-180, -32	-166, -60	-143, -82	-116, -97	-87, -104	95.1
-128	-102, 86	-133, 77	-156, 57	-172, 30	-179, 0	-177, -31	-163, -59	-139, -80	-109, -91	-78, -93	90.2
-160	-100, 84	-129, 74	-151, 55	-165, 29	-172, 0	-171, -30	-156, -57	-132, -76	-102, -85	-68, -81	84.6

<sup>a</sup> positive values are superior and negative values are inferior to the greater trochanter

<sup>b</sup> positive values are measured anterior and negative values are measured posterior to a transverse axis joining the right and left greater trochanters

**Table 4-2: Average (SD) values of  $F_{neck\_atten}$  and  $F_{total\_atten}$  for the biofidelic surrogate pelvis across impact velocity conditions.**

		impact velocity (m/s)				avg (SD)
		1	2	3	4	
$F_{neck\_atten}$ (%)	Safehip Soft	33.5 (1.1) <sup>b</sup>	33.8 (1.0) <sup>b</sup>	20.8 (0.4)	18.9 (0.3)	26.8 (8.0)
	Safehip Classic	32.7 (1.0) <sup>b</sup>	15.8 (1.5) <sup>a,b</sup>	10.9 (0.5) <sup>a,b</sup>	8.1 (0.7) <sup>a,b</sup>	16.9 (11.0)
	Hipsaver	12.9 (2.4) <sup>a</sup>	21.4 (0.7) <sup>a</sup>	20.9 (0.5)	20.7 (0.5)	19.0 (4.1)
$F_{total\_atten}$ (%)	Safehip Soft	4.2 (0.5)	3.7 (0.2) <sup>b</sup>	1.1 (0.1) <sup>b</sup>	2.3 (0.4) <sup>b</sup>	2.8 (1.4)
	Safehip Classic	8.7 (0.7) <sup>a,b</sup>	3.6 (0.2) <sup>b</sup>	1.4 (0.1) <sup>b</sup>	1.5 (0.2) <sup>b</sup>	3.8 (3.4)
	Hipsaver	4.6 (0.6)	5.2 (0.4) <sup>a</sup>	3.9 (0.1) <sup>a</sup>	4.8 (0.7) <sup>a</sup>	4.6 (0.5)

<sup>a</sup> p value for independent t-tests < 0.01 compared to Safehip Soft

<sup>b</sup> p value for independent t-tests < 0.01 compared to Hipsaver

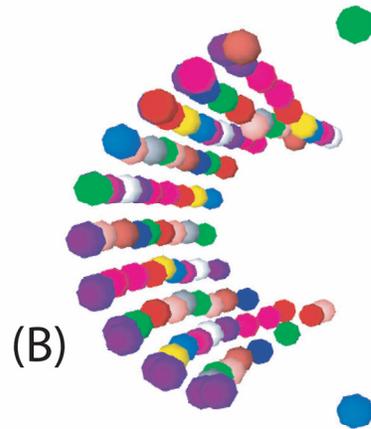
**Table 4-3: Average (SD) values of  $F_{neck\_atten}$  and  $F_{total\_atten}$  across hip protector, pelvis size, and soft tissue stiffness conditions.**

		pelvis size			soft tissue stiffness				
		small	med.	large	soft	semi -soft	semi -firm	firm	rigid
minor axis length (mm)		206	266	326	266	266	266	266	266
Stiffness of tissues anterior & posterior to GT (kN/m)		12.5	12.5	12.5	6.5	8.2	12.5	35.8	∞
Stiffness of tissues overlying GT (kN/m)		16.2	16.2	16.2	26.1	17.2	16.2	31.7	∞
$F_{neck\_atten}$ (%)	Safehip	22.5	25.8	29.5	35.8	38.1	39.3	51.0	60.2
	Soft	(0.1)	(0.4)	(0.4)	(0.9)	(0.3)	(2.5)	(0.6)	(0.4)
	Safehip Classic	13.6 (0.5)	8.9 (0.4)	1.7 (1.5)	26.0 (0.4)	31.5 (1.6)	35.9 (0.1)	52.5 (1.2)	75.7 (0.1)
	Hipsaver	27.5 (0.4)	19.0 (0.1)	19.0 (0.2)	35.6 (0.2)	25.5 (0.3)	17.8 (0.4)	27.9 (1.3)	38.4 (0.6)
$F_{total\_atten}$ (%)	Safehip	2.1	2.8	2.4	15.7	14.6	14.5	10.1	11.0
	Soft	(0.4)	(0.4)	(0.5)	(0.2)	(0.4)	(0.8)	(0.5)	(0.2)
	Safehip Classic	2.3 (0.4)	2.4 (0.3)	3.2 (0.2)	16.0 (0.3)	18.6 (0.4)	20.0 (0.2)	15.2 (0.3)	18.9 (0.1)
	Hipsaver	5.6 (0.4)	3.9 (0.4)	4.0 (<0.1)	22.0 (0.2)	17.7 (0.1)	17.0 (0.3)	16.4 (0.4)	18.2 (0.1)

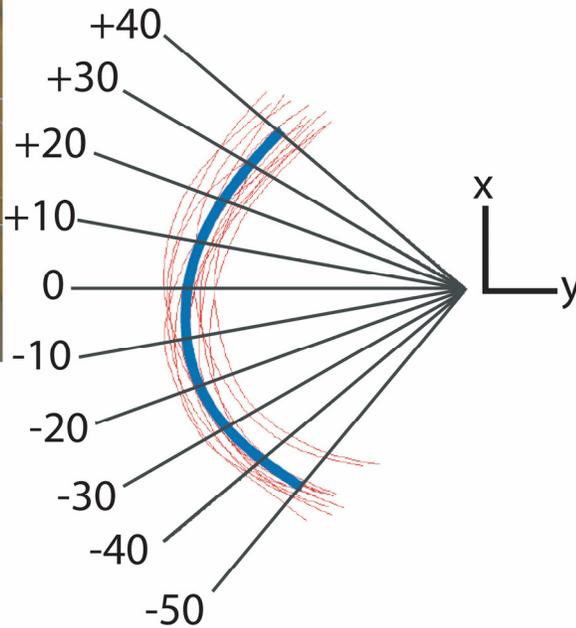
**Figure 4-1: Demonstration of the steps used in determining pelvic surface geometry from the elderly participants. (A) Photograph of the participant during the experiment, showing the 10 by 10 grid of markers recorded by the motion capture system. (B) 3D representation of the markers in the motion capture software. (C) I aligned the frontal and sagittal midlines from each subject, and defined a polar ( $r, \theta$ ) coordinate system having its original at the common body midpoint. For each subject I plotted best-fit splines representing the marker positions at each transverse plane. I then determined the average ( $r$ ) distance from the origin to the spline surface at 10 angles: reference (body midpoint to left GT); 10, 20, 30, 40, and 50 degrees posterior to the reference angle; and 10, 20, 30, and 40 degrees anterior to the reference angle.**



(A)

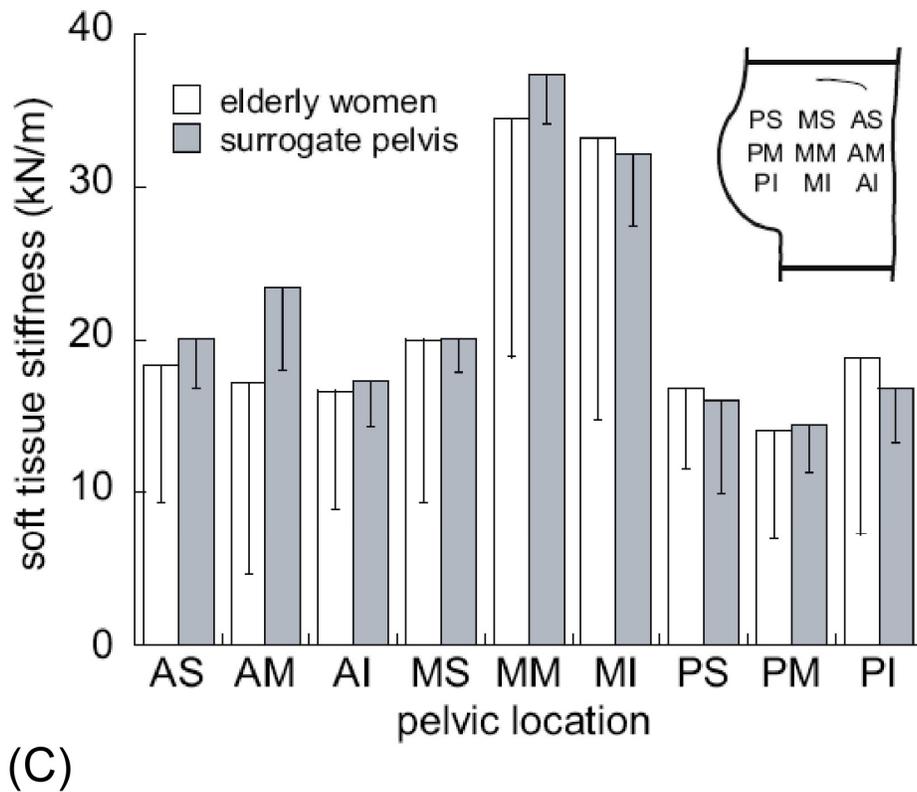
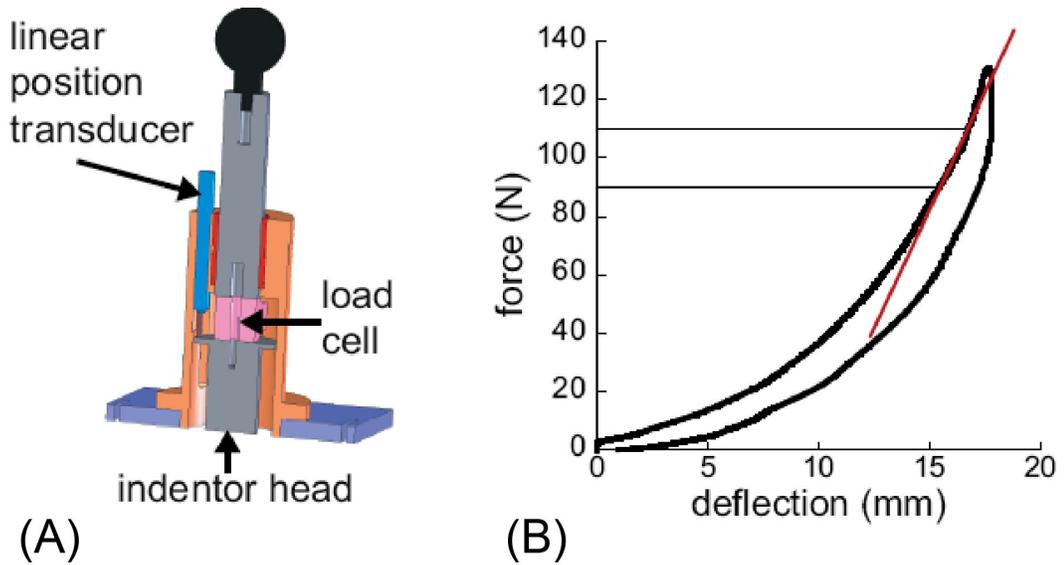


(B)



(C)

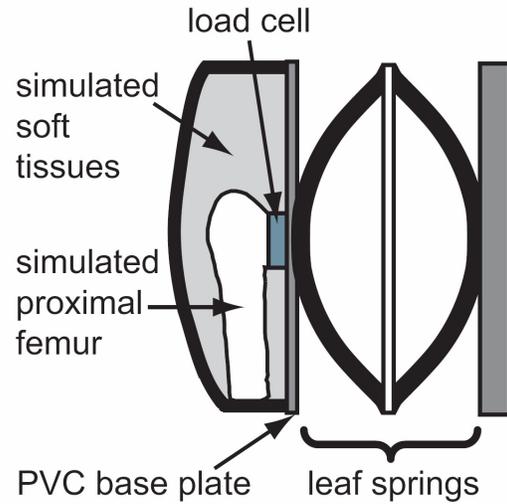
**Figure 4-2: Methods used to determine soft tissue stiffness. (A) Schematic of the indentation device. (B) Sample force – deflection curve; stiffness was determined as the slope of the curve between 90 and 110 N. (C) Average soft tissue stiffness for elderly women (n=15) and the surrogate pelvis. Error bars show one SD.**



**Figure 4-3: Elements of the hip impact simulator and associated test methods. (A) Simon Fraser University hip impact simulator. (B) Schematic of the surrogate pelvis. (C) Photographs of the three hip protectors tested in the current study (from top: Safehip Classic, Safehip Soft, Hipsaver). (D) Schematic of the small, medium, and large pelvis size conditions.**



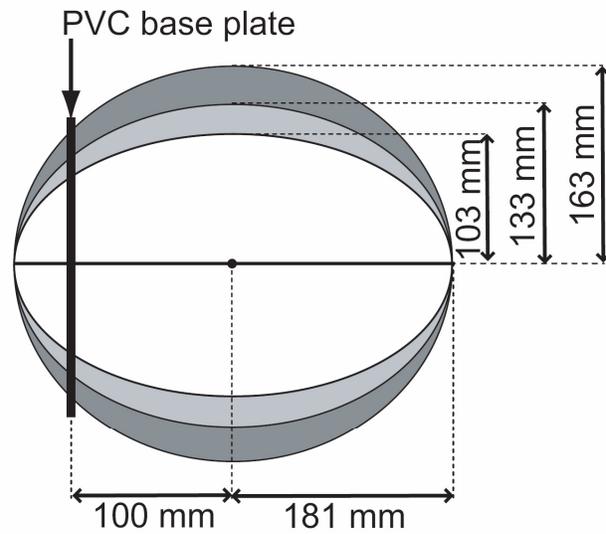
(A)



(B)

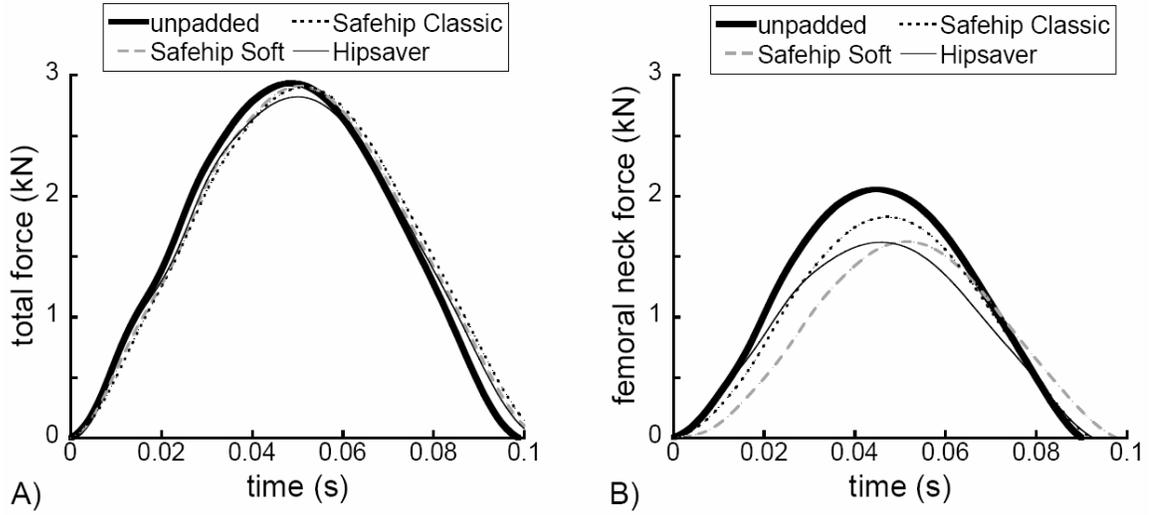


(C)

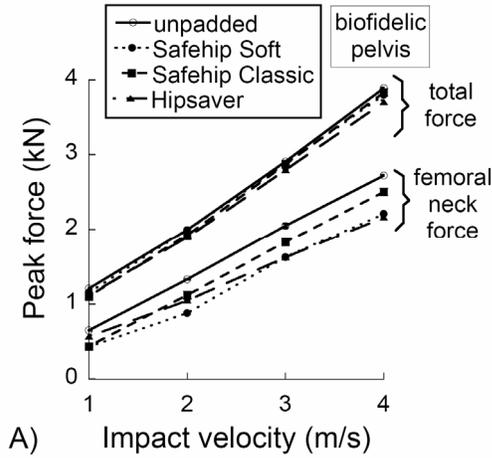


(D)

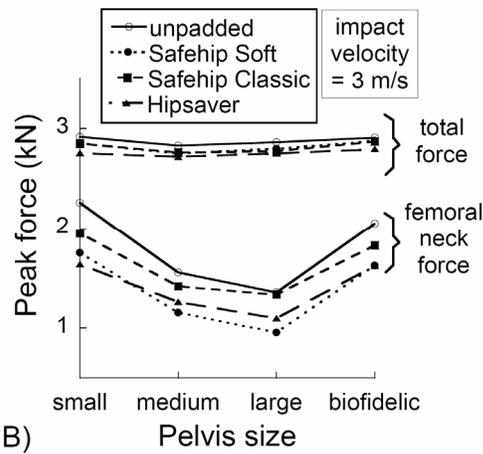
**Figure 4-4: Sample force vs. time traces for unpadded and padded trials with the biofidelic surrogate pelvis at an impact velocity of 3.0 m/s. A) Total force. B) Femoral neck force.**



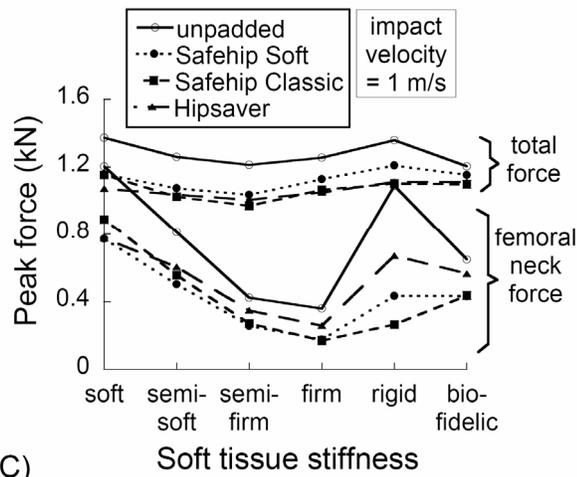
**Figure 4-5:  $F_{total}$  and  $F_{neck}$  in the unpadded (baseline) and hip protector conditions as a function of: (A) impact velocity with the biofidelic surrogate pelvis; (B) pelvic surface geometry; (C) soft tissue stiffness.**



A)



B)



C)

## **CHAPTER 5 EFFECT OF COMPLIANT FLOORING ON IMPACT FORCE DURING FALLS ON THE HIP**

### **5.1 Introduction**

Falls are a major cause of injury in the elderly, and represent the underlying cause of at least 90% of hip fractures (Grisso et al., 1991). In Canada, there are 23,000 hip fractures annually, at a cost of approximately \$1 billion (Papadimitropoulos et al., 1997). Without improvements in prevention, these numbers are expected to increase 4-fold by the year 2041, given the aging of the population and the fact that fracture risk increases exponentially with age (Jaglal et al., 1996, Papadimitropoulos et al., 1997, Wiktorowicz et al., 2001).

Risk for hip fracture during a fall depends on the fracture strength of the proximal femur, and on the forces applied to the hip during impact (Hayes et al., 1996). Thus, interventions that reduce impact force should reduce the risk for hip fracture during a fall. One strategy for achieving this that may be particularly relevant to high-risk environments (such as nursing homes, hospitals, or senior centres) is to reduce the stiffness of the floor. Of course, the design of such floors requires a compromise between the need to reduce floor stiffness enough to attenuate impact force, but not so much that there would be an impairment to balance and increased risk for falls (Redfern et al., 1997, Ring et al., 1989). In the current study, I restricted my attention to the former issue. In particular, my goal was to quantify the effect of floor stiffness on peak impact force

during a fall on the hip. This is an essential step in the design of safe movement environments for older adults.

There are two lines of evidence to suggest that floor stiffness influences impact force during a fall. First, epidemiological studies have reported that, when compared to falling on a hard (concrete or linoleum) surface, falling on a soft (padded carpet, grass, or loose dirt) surface creates reduced risk for hip fracture (Nevitt and Cummings, 1993, Simpson et al., 2004). Second, laboratory studies have found that peak impact forces during simulated falls are reduced between 7% and 23% by compliant floor designs (Casalena et al., 1998a, Gardner et al., 1998, Maki and Fernie, 1990). However, two important limitations of these previous studies are (1) that falls were simulated with a mechanical test system, which may not accurately measure the force attenuation provided by the floor during falls in humans, and (2) that data on the stiffness of the tested floor conditions were not reported.

The current study complements these previous investigations by measuring whether specific changes in surface stiffness affect impact forces during safe, simulated falls on the hip with living human participants. In particular, I utilized the “pelvis release” technique (Robinovitch and Chiu, 1998, Robinovitch et al., 1997a) to determine how impact force during a safe fall on the hip is affected by reductions in floor stiffness. I adjusted floor stiffness by changing the thickness of a layer of foam rubber overlying the force plate, and I used a materials testing machine to quantify the stiffness of each floor condition. I hypothesized that (1) peak impact force during a simulated sideways fall would decrease with decreasing floor stiffness, and (2) force attenuation would not depend on body configuration at impact.

## **5.2 Methods**

### **5.2.1 Participants**

Study participants consisted of fifteen women ranging in age from 21 to 32 years (mean =  $24 \pm 3$  (SD) yrs), in body weight from 46.8 to 77.8 kg (mean =  $59.6 \pm 10.3$  kg), and in height from 1.56 to 1.84 m (mean =  $1.70 \pm 0.09$  m). I chose (in this initial set of experiments) to study women instead of men, since they have a three-fold greater lifetime risk for hip fracture than men (Cummings and Melton, 2002). I also chose to include young women as opposed to elderly, based on safety concerns. Participants were recruited through flyers posted at the University. Respondents were interviewed by phone or in person to ensure they were in good general health, and free of musculoskeletal conditions (such as severe arthritis, or recent sprains, strains, or fractures) that would affect their ability to perform the experiment. All participants provided written informed consent and the study was approved by the Committee on Research Ethics at Simon Fraser University.

### **5.2.2 Experimental protocol**

The experimental protocol (previously termed “pelvis release experiments” (Robinovitch et al., 1997a)) involved using a sling and electromagnet to raise and suddenly release the participant’s pelvis, simulating the impact stage of a sideways fall on the hip (figure 5.1).

To conduct a trial, I first positioned the participant in one of two side fall impact configurations (“upright” or “lying”), with the pelvis cradled in a canvas sling that contacted the upper thigh and iliac crest, but not the greater trochanter. A steel cable, the length of which could be adjusted via a turnbuckle, was connected at one end to the sling

and at the other end to an electromagnet (model DCA 600-110I; Automatic Equipment Corp., Cincinnati, OH, USA). I then used the turnbuckle to raise the pelvis so a gap (measured with a wooden block) of 5 cm existed between the ground and the skin overlying the greater trochanter. I then instructed the participant to completely relax her muscles, and stay relaxed during impact. Once the participant confirmed that she was “ready,” I released the electromagnet (90% decay time = 15 ms), causing the participant to fall onto the landing surface. The time-varying force applied to the hip region during impact was measured with a force plate (model 4060H; Bertec Corp., Columbus, OH, USA) sampling at 960 Hz. Throughout the trial, the lateral aspect of the shin and shoulder remained in contact with the floor; only the pelvis was lifted and released.

In the upright configuration, the participant lay on her left side with her trunk flexed at 68 deg to the horizontal, her left hand contacting the ground, and her left elbow extended. In the lying configuration, the participant lay on her left side with her shoulder flexed overhead. In both conditions, the knees were maintained flexed at 75 deg. Tape marks and goniometers were used to ensure consistent positioning between trials.

For each impact configuration, I conducted trials at five different floor conditions. In the rigid condition, the participant lay directly on the force plate and the adjacent, flush linoleum-covered concrete floor. In the other conditions, the participant lay on a 1.85 m x 0.6 m mat of closed cell, cross-linked ethylene vinyl acetate (EVA) foam (of density 46.6 kg/m<sup>3</sup>) placed over the force plate and surrounding linoleum (product # 4000-686, Mountain Equipment Co-op, Vancouver, BC, Canada). The thickness of the mat was varied between 1.5, 4.5, 7.5, and 10.5 cm; the latter three conditions were achieved by stacking multiple 1.5 cm-thick mats. Each participant underwent three trials (the results

of which were averaged) at each of the two body configurations and five floor conditions, for a total of 30 trials. While the 3 trials for a given condition were acquired consecutively, I randomized the order of presentation of the various body configuration and floor condition combinations.

### **5.2.3 Floor stiffness**

I measured the stiffness of each foam layer ( $k_f$ ) through indentation tests (figure 5.2) with a servohydraulic testing system (FastTrack™ 8874, Instron Corporation, Canton, MA, USA). The indenter was custom designed to match the skin surface geometry of the hip and pelvis region of a 26 year old female of body mass 49.1 kg and height 1.55 m. To construct the indenter, I first made a plaster mold (Gypsona LPL 2, Smith & Nephew, Mississauga, ON, Canada) of the participant's pelvis as she lay on her side with her hips and knees flexed at approximately 10 degrees. The mold extended anteriorly to the front of the thigh, posteriorly to the buttock, superiorly to the iliac spine, and inferiorly to the midpoint of the thigh. After allowing the mold to dry, I filled it with dental stone mix (Precision Stone, Ash Temple Limited, Don Mills, ON, Canada) and inserted a metal bolt for securing the indenter to the Instron machine (see inset to figure 5.2). The indentation tests involved a peak force of 3000 N (similar to the mean value predicted for a fall from standing (Robinovitch et al., 1997a)), a displacement rate of 50 mm/s (slightly less than that expected during a fall, but the maximum value I could achieve without risking damage to the indenter at the lowest foam thickness condition), and a sampling rate of 1000 Hz. Pilot tests at different displacement rates demonstrated minimal rate dependency (or viscous effects) to the force-deflection behaviour of the EVA foam. Since the stiffness varied nonlinearly with force, I fit a second-order

polynomial to the compression phase of each force vs. displacement trace, and then differentiated this equation to arrive at an expression for  $k_f$  as a function of force. Values of  $k_f$  at 1000 N (similar to the peak impact force observed in my pelvis release experiments) were 263 kN/m for the 1.5 cm thick foam later (a condition I shall refer to as “firm”), 95 kN/m for the 4.5 cm thick layer (“semi-firm”), 67 kN/m for the 7.5 cm layer (“semi-soft”), and 59 kN/m for the 10.5 cm thick layer (“soft”).

#### **5.2.4 Data analysis**

In all trials, the measured variation in impact force was dominated by a single natural frequency (figure 5.3). This allowed for easy detection (through a customized MATLAB routine) of the magnitude of peak force ( $F_{max}$ ). I report both raw and normalized (raw/(body mass)) values of  $F_{max}$ , and the percent attenuation in  $F_{max}$  provided by each foam floor condition, relative to the rigid floor condition.

#### **5.2.5 Statistics**

I used a 2-factor repeated measures ANOVA ( $\alpha = 0.05$ ) to determine whether foam thickness and landing configuration affected normalized values of  $F_{max}$ . In these analyses, I used Mauchley’s test to determine whether the assumption of sphericity was violated, and if so, I used the Huynh-Feldt Epsilon correction factor to adjust the degrees of freedom. When ANOVA results indicated a significant effect, I used paired t-tests to test for differences between the various foam thickness conditions. Based on multiple hypothesis tests I applied a Bonferroni correction to the assumed significance levels, and only considered p values less than 0.005 to indicate significant effects between

conditions. All analyses were performed with statistical analysis software (SPSS Version 11.0, SPSS Inc., Chicago, IL, USA).

### 5.2.6 Mathematical modelling

Robinovitch et al. (1997b) previously showed that the peak force and rate of loading applied to the hip during a simulated fall (involving a non-zero impact velocity) is predicted well by a simple mass-spring model. In the current study, I examined whether this model was able to predict the effect of floor stiffness on peak force. My model consisted of a single mass supported on a spring having an effective stiffness  $k$  given by the series combination of the floor spring  $k_f$  (which varied between 59 kN/m and 263 kN/m, as described above) and the effective spring stiffness of the body ( $k_b$ ).

McMahon et al. (1987) showed that the equations describing the time to peak force ( $t_{max}$ ) and magnitude of peak force ( $F_{max}$ ) are given by:

$$t_{max} = 1/\omega_n \left( \pi - \tan^{-1}(u\omega_n/g) \right)$$

and

$$F_{max} = mg \left( (u\omega_n/g) \sin(\omega_n t_{max}) + 1 - \cos(\omega_n t_{max}) \right)$$

where  $u$  is the impact velocity (in m/s),  $m$  is the effective mass (in kg),  $\omega_n$  is the natural frequency (in rad/s), and  $k$  is the effective stiffness (in N/m). The last two parameters are given by

$$\omega_n = \sqrt{k/m}$$

and

$$k = \frac{k_b k_f}{k_b + k_f}.$$

In my simulations, I used average values of  $k_b$  and  $m$  measured by Robinovitch et al. (1997a) in pelvis release experiments with young women. These were  $k_b = 30.4$  kN/m and  $m = 27.3$  kg for the lying condition, and  $k_b = 49.6$  kN/m and  $m = 35.3$  kg for the upright condition.

### 5.3 Results

Experimentally measured values of peak force ( $F_{\max}$ ) decreased with decreasing floor stiffness ( $p < 0.001$ ; table 5.1). The trend was non-linear, with the greatest changes seen between the rigid, firm, and semi-firm conditions (figure 5.3). When compared to the rigid condition, peak forces averaged 8% lower in the firm condition, 15% lower in the semi-firm condition, 16% lower in the semi-soft condition, and 18% lower in soft condition. Paired  $t$ -tests (table 5.2) indicated that  $F_{\max}$  was significantly higher in the rigid than all other conditions, and significantly higher in the firm condition than the semi-firm, semi-soft, and soft conditions. There were no significant differences in  $F_{\max}$  between the semi-firm, semi-soft, and soft conditions.

$F_{\max}$  also associated with body configuration ( $p < 0.001$ ), averaging 30% lower in the upright than lying condition ( $10.9 \pm 0.5$  (SE) versus  $15.6 \pm 0.5$  N/kg, mean difference = 4.7, 95% CI: 3.6 to 5.8 N/kg,  $p < 0.001$ ; table 5.1). Furthermore, there was a

significant interaction between foam thickness and body configuration whereby foam stiffness had a stronger influence on  $F_{max}$  in the lying than upright position ( $p = 0.017$ ).

Our mass-spring model for the lying configuration predicts that, when compared to the rigid condition, values of  $F_{max}$  would be reduced by 3%, 9%, 12%, and 13% percent in the firm, semi-firm, semi-soft, and soft conditions, respectively (figure 5.4). For the upright condition, the model predicts that  $F_{max}$  would be reduced (when compared to the rigid condition) by 6%, 14%, 18%, and 19% percent in the firm, semi-firm, semi-soft, and soft conditions.

## 5.4 Discussion

In this study, I examined how hip impact forces during simulated sideways falls were affected by floor stiffness, which I modified with layers of foam rubber of varying thickness. I found that peak impact force varied nonlinearly with floor stiffness, so a 4.5 cm thick foam mat having a stiffness of 95 kN/m (our semi-firm condition) provided nearly the same force attenuation as a 10.5 cm thick mat having a stiffness of 59 kN/m (our soft condition). I also found that floor stiffness had a slightly greater effect on peak force in the lying configuration than the upright configuration.

Our results are in general agreement with previous studies that employed mechanical systems to estimate the effect of compliant flooring on fall impact force (Casalena et al., 1998a, Gardner et al., 1998), although the comparison is made difficult by the fact that previous studies did not report the stiffness of their floor samples. The 8% attenuation I observed in the firm condition is similar to the 7% reduction observed by

Gardner et al. (1998) for a pile carpet and underpad, and the 11% reduction observed by Maki and Fernie (1990) for 7 mm thick loop carpet. Furthermore, the 18% attenuation I observed in the soft condition matches the 18% reduction observed by Maki and Fernie for carpet plus underpad.

Are the force reductions I observed sufficient to prevent hip fractures in high-risk environments? While clinical trials are required to answer this question definitively, there are at least two encouraging lines of evidence. First, evidence suggests that sideways falls from standing (onto a rigid floor) generate impact forces that are just above the mean value required to fracture the elderly cadaveric femur (Robinovitch et al., 1991, 1997a, van den Kroonenberg et al., 1995). Therefore, even a small reduction in impact force may prevent fractures. Second, while hip protectors have been shown to reduce fracture risk in clinical trials by as much as 50% (Parker et al., 2000), those same devices reduce peak impact force by only 15-20% in laboratory testing (Parkkari et al., 1995, Robinovitch et al., 1995a). This again supports the notion that the relatively modest reductions in peak force observed in the current study may be sufficient to prevent hip fractures.

Of course, in order to be feasible, a floor must not be so compliant that it impairs individuals' balance and mobility to the point that it actually causes falls. How much force attenuation can be achieved by a floor having a stiffness just above this threshold value? To address this question fully, future studies are required to quantify the effect of floor stiffness on older individual's ability to maintain balance during daily activities (e.g., walking, standing, and turning) and recover balance following an unexpected slip or trip. These results must then be compared to the force attenuation data from the current study, in order to identify a range of floor stiffness that represents a reasonable

compromise between force attenuation and postural stability. Other criteria may need to be considered for practical lower bounds on floor stiffness, such as ability to move equipment across the floor. Finally, clinical trials should be conducted to assess the feasibility and efficacy of these candidate designs in preventing fractures.

It is encouraging, however, that even my firm floor condition, which consisted of a 1.5 cm thick foam layer, attenuated peak impact force by 8%. This foam layer is similar in thickness to the foamback layer used in many carpet designs, and to my knowledge, padded carpeting has never been shown to cause falls among the elderly. Indeed, Simpson et al. (2004) found that wooden carpeted floors were associated with nearly half the fracture rate of uncarpeted floors. I estimate that my firm floor would compress about 9 mm, and my semi-firm floor would compress about 16 mm, when an adult of body weight 700 N stood upon it. This assumes that my indentation test results (figure 5.2) are reasonable approximations of the foam force-deflection behaviour during walking, which seems plausible given that the foot has about the same contact area with the ground as the hip during a fall.

How might additional reductions in floor stiffness (beyond the range I examined) affect hip impact force? The results from my model simulations (figure 5.4) provide a means for estimating the attenuation in peak force provided by any floor stiffness. For example, the model predicts that, when compared to falling on a rigid floor, a floor having a stiffness of 10 kN/m would attenuate peak force by 34% in the lying condition, and 42% in the upright condition. Further experiments are needed to verify the accuracy of these predictions. I found that the model tended to under-estimate (by 7 percent on average) the force attenuation provided by compliant flooring in the lying configuration,

and more accurately predict (within 1 percent on average) the force attenuation in the upright configuration. I can think of at least two reasons for discrepancies between experimental data and model predictions. First, the values of effective mass and stiffness for the human body in the model were based on pelvis release experiments with a different group of young women (Robinovitch et al., 1997a). Second, the model used constant values for the stiffness of the human body and the floor, and its predictive accuracy might be improved by instead using force-dependent stiffness functions (especially since the floors I examined had nonlinear stiffness (figure 5.2)). The model's predictive accuracy might also be improved by including damping, through for example a three-element standard linear solid model (Robinovitch et al., 1997b).

There are several important limitations to this study. First, I focused my attention on closed-cell EVA foam rubber, since it appears to represent a practical option for compliant flooring due to its availability, durability, washability, and cost effectiveness. However, other materials may be more effective than EVA in reducing impact force and further studies are warranted in this area. Second, due to safety concerns, I restricted the drop height in my experiment to 5 cm. Even at this relatively low fall height, I observed peak forces approaching 1300 N in the rigid floor condition. In my opinion, peak forces any greater than this would be too near the range of forces observed to fracture proximal femurs from young adult cadavers, which Courtney et al. (1994) measured to range from 5500 to 10000 N. Third, I tested only young women, since women are more likely than men to suffer hip fracture (and it was beyond my current scope to examine gender effects), and safety precautions prevented me from including older women as participants. However, I have little reason to believe that age-related changes in the

structural properties of the pelvis and trochanteric soft tissues would be so great that the general trends I observed would not apply to elderly as well as young women. Fourth, I instructed participants to relax their muscles during impact, but did not monitor muscle activity to ensure they followed this instruction. However, I doubt variations in muscle activation could have affected my results substantially, since most previous studies have found that the state of muscle contraction has an insignificant effect on impact force (Robinovitch et al., 1997a, Sabick et al., 1999). Fifth, I focused on how floor stiffness influences only the peak force  $F_{max}$  applied to the hip, which may not fully capture the protective value of a given floor design. Future studies should examine how compliant floors modify the distribution of pressure over the hip region (which may be particularly profound for a foam rubber surface), and absorb energy during impact to other body parts such as the knees and hands (Kim and Ashton-Miller, 2003, Robinovitch and Chiu, 1998).

In summary, results from my “pelvis release experiments” show that, during impact to the hip, even a thin (1.5 cm thick) layer of foam rubber reduces peak impact force by 8%, and a 4.5 cm thick layer reduces peak force by 15%. Such floor designs would deflect less than 2 cm when a typical adult stood upon them. My values of force attenuation are similar to those observed previously for mechanical test systems, and agree with predictions from a mass-spring model of impact, which provides a means for predicting the force attenuations provided by floor stiffnesses outside the range I measured. These results support the need for laboratory studies to more fully quantify the effect of floor stiffness on postural stability and mobility in the elderly, and clinical trials to determine whether compliant flooring reduces the incidence of hip fracture and other

fall-related injuries in high-risk environments such as nursing homes, hospitals, gymnasiums, and senior centres.

**Table 5-1: Average (one standard error) values of  $F_{max}$  and attenuation of peak force for the various floor conditions and landing configurations.**

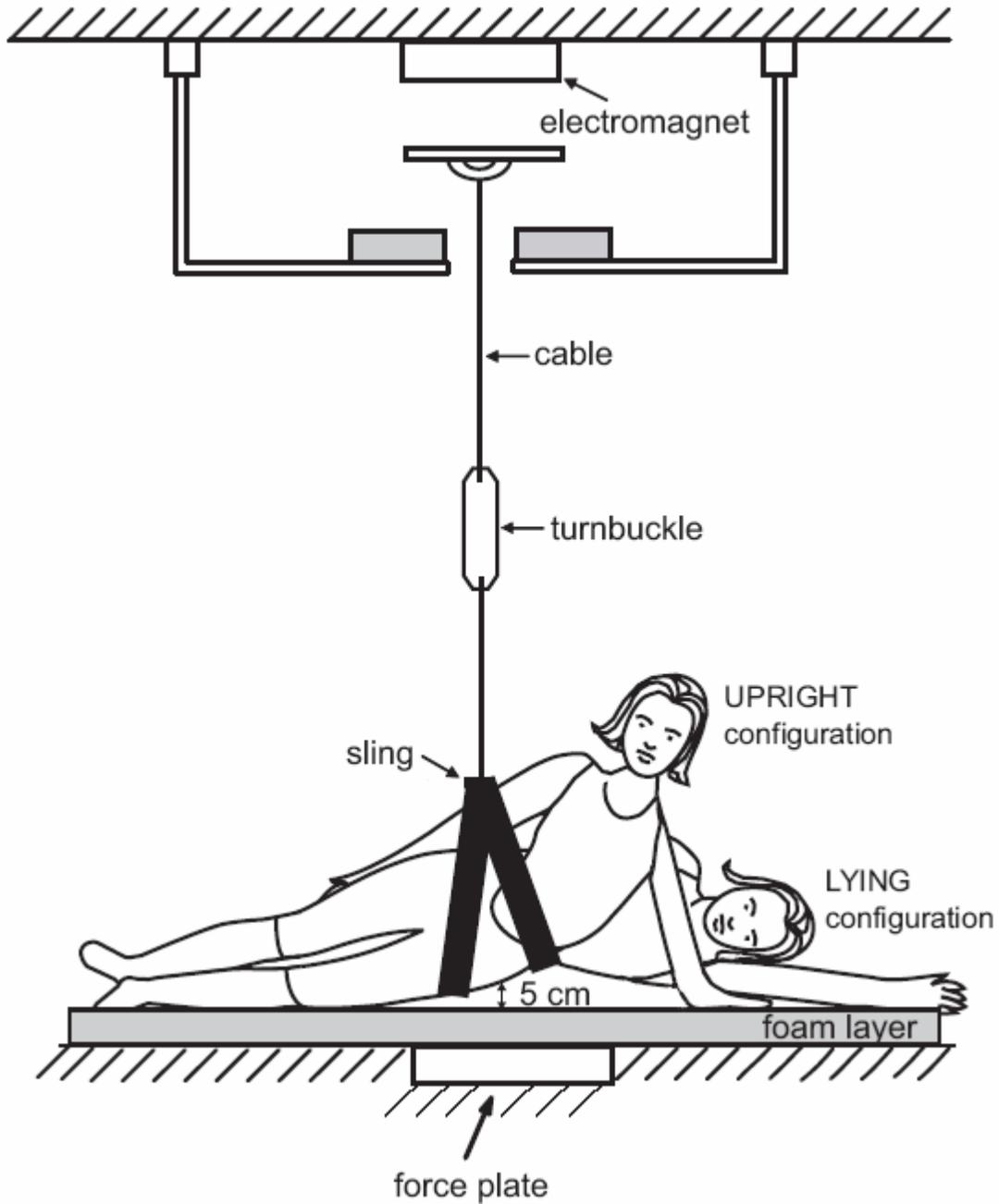
Parameter	Floor Condition				
	rigid	firm	semi-firm	semi-soft	soft
Foam thickness (cm)	0	1.5	4.5	7.5	10.5
Foam stiffness (kN/m)	$\sim\infty$	263	95	67	59
Peak force					
Upright					
Raw value (N)	725 (43)	679 (41)	621 (34)	607 (33)	600 (32)
Normalized value (N/kg)	12.2 (0.6)	11.5 (0.6)	10.5 (0.5)	10.4 (0.6)	10.1 (0.4)
Attenuation (%)	-	6 (3)	14 (2)	15 (4)	16 (2)
Lying					
Raw value (N)	1059 (42)	953 (42)	886 (36)	864 (36)	849 (32)
Normalized value (N/kg)	17.9 (0.6)	16.1 (0.5)	15.0 (0.5)	14.6 (0.5)	14.4 (0.4)
Attenuation (%)	-	10 (2)	16 (2)	18 (2)	19 (2)
Average of upright and lying					
Raw value (N)	892 (37)	816 (36)	754 (31)	735 (28)	724 (29)
Normalized value (N/kg)	15.1 (0.5)	13.8 (0.5)	12.7 (0.4)	12.5 (0.5)	12.2 (0.3)
Attenuation (%)	-	8 (2)	15 (1)	17 (2)	18 (1)

**Table 5-2: Pair-wise comparisons of the main effect (average of upright and lying configurations) of foam thickness on peak force  $F_{max}$  (normalized by body mass).**

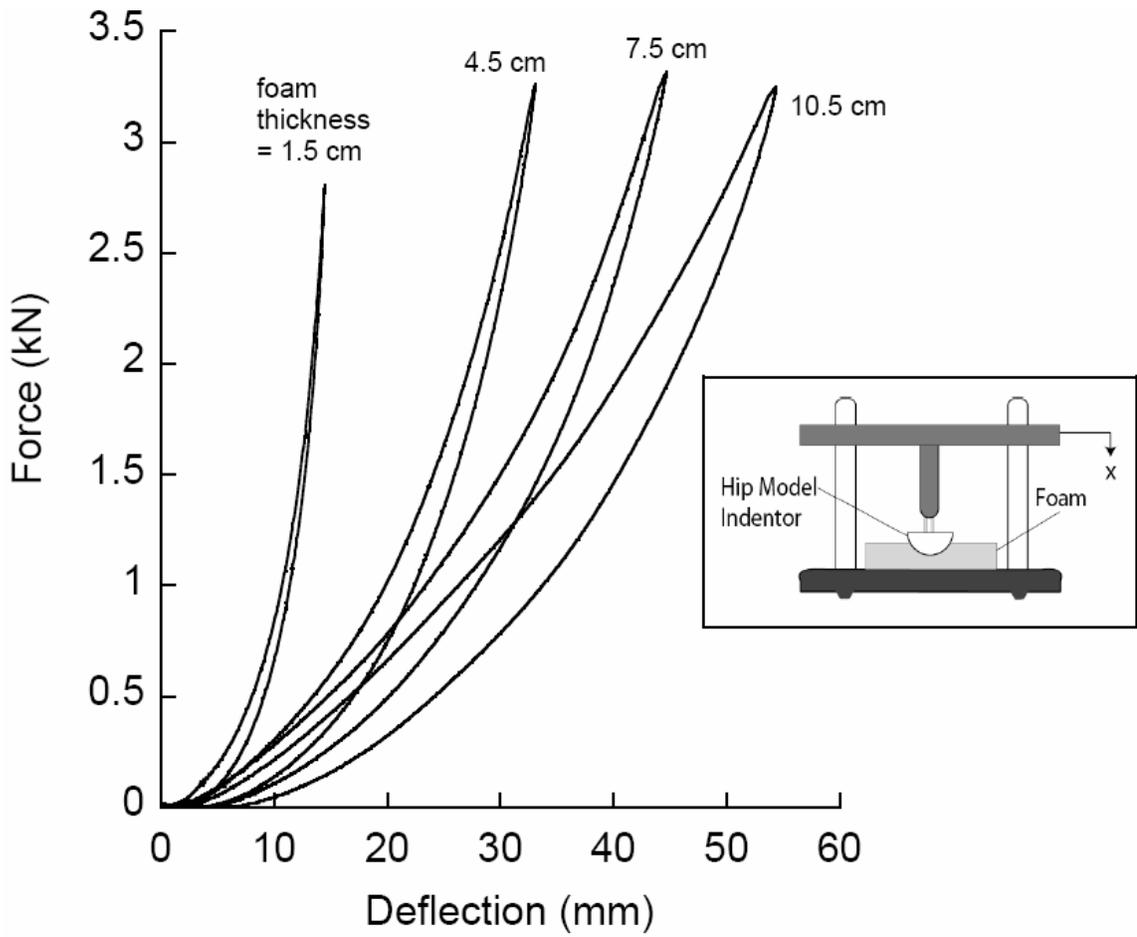
Parameter	Floor Condition		Mean Difference	$p$ value	95% Confidence Interval for Difference	
	Reference	Comparison			Lower	Upper
$F_{max}$ (N/kg)	rigid	firm	1.29 *	<0.001	0.78	1.79
		semi-firm	2.33 *	<0.001	1.90	2.76
		semi-soft	2.58 *	<0.001	1.96	3.19
		soft	2.83 *	<0.001	2.43	3.22
	firm	semi-firm	1.04 *	<0.001	0.72	1.37
		semi-soft	1.29 *	<0.001	0.73	1.84
		soft	1.54 *	<0.001	0.99	2.09
	semi-firm	semi-soft	0.24	0.349	-0.30	0.78
		soft	0.49	0.034	0.04	0.95
	semi-soft	soft	0.25	0.400	-0.37	0.87

\* The mean difference is significant at the 0.005 level.

**Figure 5-1: Experimental schematic.** A sling and electromagnet were used to raise and suddenly release the participant's pelvis from a height of 5 cm in the upright and lying landing configurations, simulating the impact stage of a sideways fall on the hip. The time-varying force applied to the hip region during impact was measured with a force plate.



**Figure 5-2: Force versus deflection traces at a deformation rate of 50 mm/s for the four foam thickness conditions. Inset: Servohydraulic testing system and hip model indenter used to measure foam rubber force-deflection characteristics.**



**Figure 5-3: Force versus time traces for a typical participant in the (A) upright and (B) lying landing configurations for rigid, firm, semi-firm, semi-soft, and soft floor conditions.**

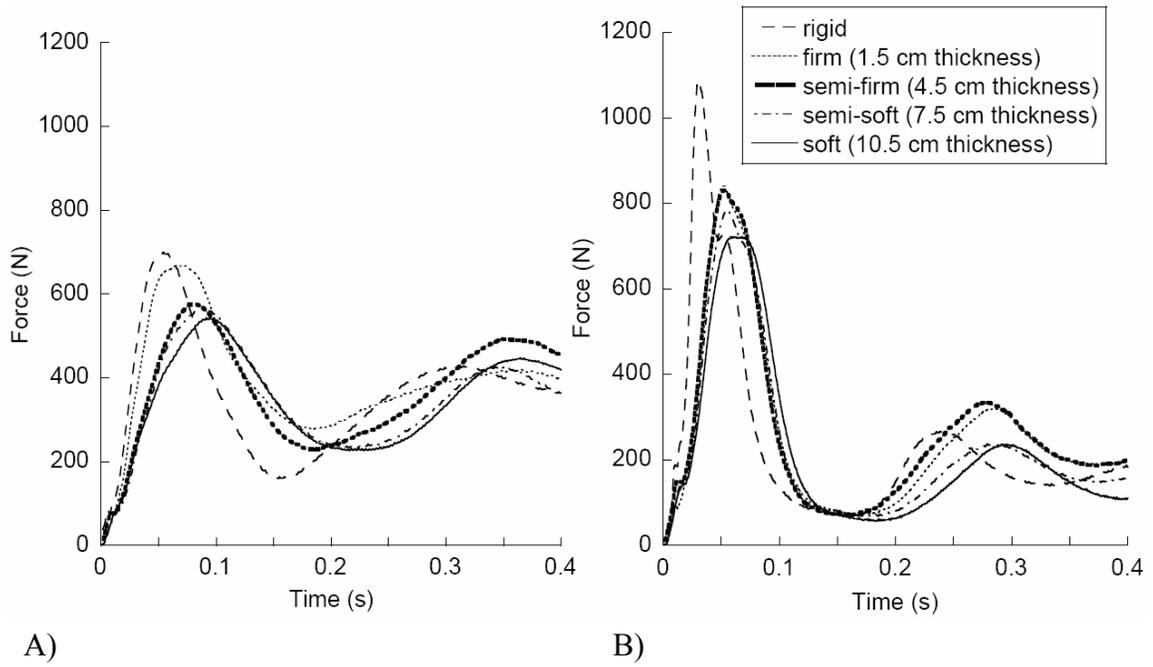
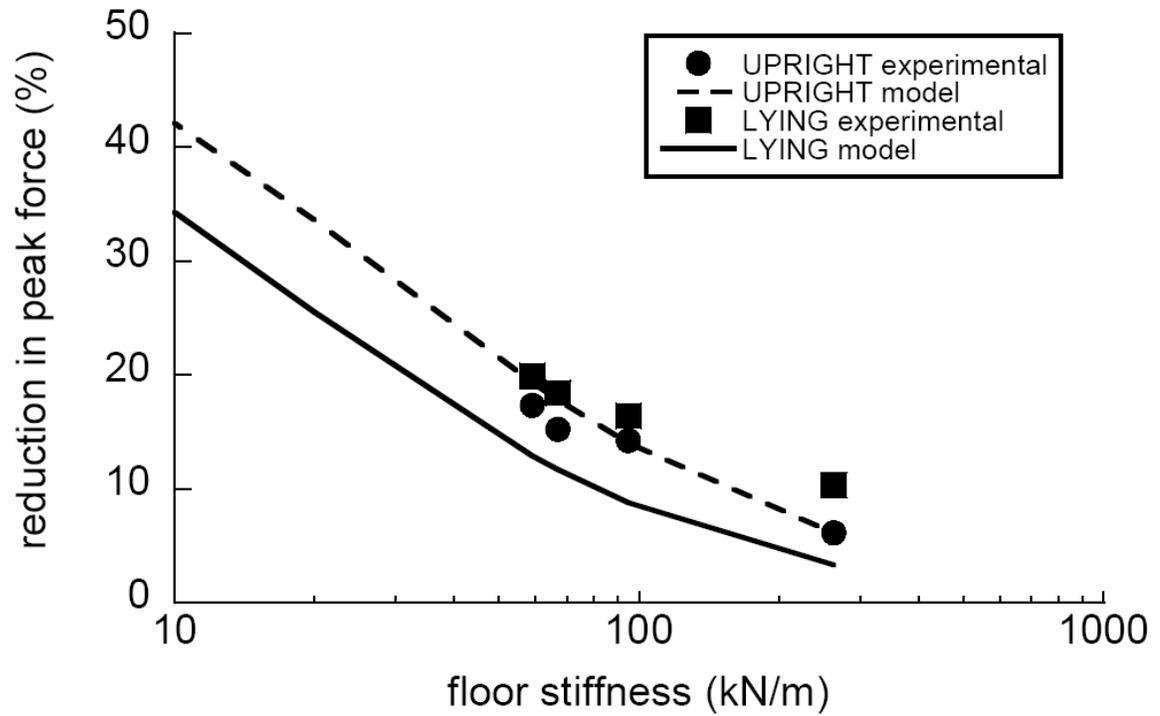


Figure 5-4: Comparison of mean experimental values and model predictions of attenuation in peak force ( $F_{max}$ ) versus floor stiffness ( $k_f$ ) in the upright and lying landing configurations.



## **CHAPTER 6 LOW STIFFNESS FLOORS CAN ATTENUATE FALL-RELATED FEMORAL IMPACT FORCES BY UP TO 50% WITHOUT SUBSTANTIALLY IMPAIRING BALANCE IN OLDER WOMEN**

### **6.1 Introduction**

Hip fracture prevention is a current public health priority due to their high frequency, cost, and negative impact on quality of life. There are approximately 23,000 cases of hip fracture every year in Canada with associated treatment costs of about \$1 billion (Papadimitropoulos et al., 1997). Hip fractures cause increased mortality (Empana et al., 2004a, Meyer et al., 2000, Wolinsky et al., 1997), and declines in mobility, physical activity, and functional independence (Norton et al., 2000a, Wolinsky et al., 1997). Without improvements in prevention, hip fracture incidence is expected to increase 4-fold by the year 2041 due to demographic shifts towards a more aged population and the exponential increase in fracture risk with advancing age (Jaglal et al., 1996, Papadimitropoulos et al., 1997, Wiktorowicz et al., 2001). Although osteoporosis screening and pharmaceutical interventions may temper the expected increase in hip fracture rate (Jaglal et al., 2005), they will not assist the substantial portion of hip fracture risk who have normal bone strength (Siris et al., 2004, Wainwright et al., 2005). Consequently, complementary preventive strategies are required to reduce the incidence of hip fractures.

Low stiffness floors have the potential to decrease hip fracture risk by attenuating the force applied to the proximal femur in the event of a fall. Two lines of evidence

support their effectiveness. First, epidemiological studies have reported that falling onto a soft landing surface (padded carpet, grass, or loose dirt) significantly reduces hip fracture risk when compared to a hard surface (concrete or linoleum) (Healey, 1994, Nevitt and Cummings, 1993, Simpson et al., 2004). Second, laboratory studies have demonstrated that energy-absorbing floors can substantially reduce the peak force applied to the hip during the impact phase of a fall. Studies with human participants suggest that foam-rubber materials can decrease peak impact force by up to 20% (Sran and Robinovitch, 2008) (see Chapter 5) and reduce the peak pressure applied to the tissues overlying the greater trochanter by up to 76% (see Chapter 3). Mechanical fall simulators have been used to demonstrate more modest force attenuation for common energy-absorbing floors including wood (7%) and carpets (15%) (Gardner et al., 1998, Maki and Fernie, 1990, Simpson et al., 2004). Additional studies are warranted to determine the force attenuation provided by other floors specifically designed for the purpose of injury prevention e.g. playground surfaces, anti-fatigue mats (Casalena et al., 1998a).

Very large reductions in floor stiffness would likely be counter-productive due to their negative effects on gait and balance. Such floors could reduce impact force sufficiently to prevent hip fracture, but are infeasible due to the limits they would place on the safe performance of activities of daily living (an extreme example is children's inflatable jumping gyms). The mechanisms by which such low stiffness floors could affect balance include impairments in somatosensory feedback systems (Betker et al., 2005, Lord and Menz, 2000, Ring et al., 1989), reduced toe clearance during locomotion, or increased time to develop maximum ground reaction forces. With a few exceptions (Redfern et al., 1997), low stiffness floors such as carpet have not been associated with

impaired balance (Dickinson et al., 2002, Dickinson et al., 2001) or increased fall risk (Healey, 1994). However, to date no study has simultaneously evaluated both the force attenuation and potential balance impairments associated with a wide range of floor stiffnesses in order to identify feasibility limits below which the benefits of reduced impact forces are outweighed by coincident increases in fall risk.

I had several objectives in the current study. My first aim was to determine the force attenuation provided by a range of commercially available low stiffness floors during simulated sideways falls on the hip. My second aim was to test whether these floors influence balance across a range of static and dynamic activities of daily living in a sample of healthy elderly women. Such measures are an essential step in the design of safe movement environments for older adults in high risk settings including residential housing and assisted living settings. This information could also inform social policy changes related to improved housing codes for these facilities, whose numbers are expected to substantially increase in the coming decades in response to our aging population.

## **6.2 Methods**

### **6.2.1 Floor conditions**

I investigated five flooring types selected to provide a wide range of floor stiffness conditions (figure 6.1). The ‘rigid’ control floor was a 2 mm thick layer of slip-resistant dense natural rubber intended for use in commercial and institutional settings (Noraplan Classic, Nora Systems Inc, Lawrence, MA, USA). SmartCell (SATech, Chehalis, WA, USA) is a synthetic rubber (density = 1120 kg/m<sup>3</sup>) floor system

comprising a continuous surface layer overlying an array of cylindrical rubber columns 14 mm in diameter, and spaced at 19 mm intervals. The version I tested had a height of 2.5 cm. The SofTile floor (SofSurfaces, Petrolia, ON, Canada) consists of square tiles of synthetic rubber typically used as playground surfaces. Each 60 x 60 x 10 cm tile comprises a continuous top surface overlying compliant rubber columns 5 cm in diameter spaced at 7 cm intervals, and interfaces with adjacent sections via interlocking flanges. Two additional conditions were comprised of open cell polyurethane foams from gymnasium crash mats. The Firm Foam condition was 11 cm thick with a density of 32 kg/m<sup>3</sup>, while the Soft Foam was 10 cm thick with a density of 22.2 kg/m<sup>3</sup> (The Foam Shop, Vancouver, Canada). The Indentation Load-Deflection (ILD) test is a standard measure of foam firmness in which samples (100 mm thick by 500 mm by 500 mm) are statically loaded with a flat circular indenter (10.1 cm radius). The *ILD25* value is the load associated with a sample strain of 25%, and was 400 N for the Firm Foam and 195 N for the Soft Foam (as reported by the manufacturer).

I measured the force-deflection properties of each floor through indentation tests with a servohydraulic testing system (FastTrack™ 8874, Instron Corporation, Canton, MA, USA) using rigid hip-shaped and foot-shaped indenters (figures 6.1B and C). The hip indenter was custom designed to match the pelvic surface geometry of a 26 year old female (see Chapter 5). The foot indenter matched the surface geometry of a typical walking shoe sole (female sized 9). Trials were conducted using a displacement rate of 30 mm/s to a peak force of 3-4 kN (the approximate peak load applied to the hip during a sideways fall with an impact velocity of 3 m/s) and sampled at 1000 Hz.

### 6.2.2 Force attenuation tests

The SFU hip impact simulator (figure 6.2) is an improved version of the Robinovitch et al. (1995) test system that matches the pelvic soft tissue stiffness and surface geometry of elderly women (see Chapter 4). It measures the total force applied to the skin overlying the hip region and the force delivered to the femoral neck during a simulated sideways fall. The system consists of an impact pendulum and surrogate pelvis, which are released by an electromagnet from a raised position and strike the ground horizontally. The surrogate pelvis (figure 6.2B) consists of simulated soft tissues and a proximal femur (Sawbones, Vashon, WA, USA) mounted on a 25.0 x 25.0 x 0.7 cm polyvinyl chloride base plate. These elements are connected to the pendulum via leaf springs which simulate the effective stiffness of the pelvis (Robinovitch et al., 1997a). The force applied to the femoral neck is measured by a load cell (Kistler Model 9712A5000, Amherst, NY, USA), and the total force applied to the skin surface is measured with a floor-mounted force plate (model 2535-08, Bertec Corp., Columbus, OH, USA). Impact velocity is varied by adjusting the initial angle of the pendulum before release. The effective mass and stiffness of the entire system, measured from its free vibration response (Robinovitch et al., 1995a, 1997b) are 28.0 kg and 42.2 kN/m, respectively. These values are within one SD of the mean values measured in simulated falls on the hip with young women (Robinovitch et al., 1997a).

Based on Feldman and Robinovitch's (2007) observation that the mean hip impact velocity during unexpected falls from standing is 3 m/s (SD = 1), I conducted trials at impact velocities of 2, 3, and 4 m/s. As the floors were of varying thickness, I

shifted the vertical height of the pendulum's rotational axis as required to ensure that the surrogate pelvis was perpendicular to each floor at impact.

The test protocol at each impact velocity involved three trials on the rigid floor followed by three sequential trials with the low stiffness floors presented in a random order. I collected force for two seconds at 1000 Hz, and filtered with a dual-pass fourth-order Butterworth low pass filter with a 35 Hz cut-off frequency (Labview 6.1, National Instruments, Austin, TX, USA). For each trial, I identified the peak femoral neck force ( $F_{neck}$ ). The attenuation in femoral neck force provided by each floor ( $F_{neck\_atten}$ ) was calculated as the average percentage decrease in  $F_{neck}$  compared to the rigid floor condition.

### **6.2.3 Balance tests with elderly women**

Study participants consisted of fifteen women ranging in age from 65 to 90 years (mean = 75.0, SD = 8.1 yrs), in body mass from 57.6 to 93.0 kg (mean = 70.6, SD = 11.3 kg), in height from 1.50 to 1.80 m (mean = 1.60, SD = 0.10 m), and in body mass index from 22.3 to 31.3 kg/m<sup>2</sup> (mean = 27.0, SD = 3.0 kg/m<sup>2</sup>). All participants provided written informed consent and the study was approved by the Committee on Research Ethics at Simon Fraser University.

Each participant underwent three general balance assessment tasks on each floor. The tasks, and the floor conditions within each task, were presented in a random order, with five successive trials performed for each condition. For safety purposes, subjects wore a chest harness attached to the ceiling via a tether. This system did not impede

natural responses but prevented the participant from contacting the ground in the event of a fall (which never occurred).

#### **6.2.3.1 Get Up and Go test**

The participant performed the Timed Get Up and Go test which involved her rising from a chair, walking forward 3 m at a self-selected speed, turning around, walking back to the chair and reseating herself (Podsiadlo and Richardson, 1991). The time from the moment of rising to the moment of return seated posture ( $GUG_{time}$ ) was measured with a stop-watch. The participant also completed a restricted Get Up and Go test in which she was required to keep her feet within a 15 cm wide gait path and 50 x 50 cm turnaround zones marked on the floor surface with tape (figure 6.3A). To avoid the participant sacrificing task completion speed for foot placement accuracy in the restricted test, she was required to complete the trial in 7 to 13 seconds. Each trial was assessed a binary score of ‘success’ or ‘failure’ with success indicating no foot placement errors and an acceptable task completion time. The outcome success was calculated as the proportion of the five trials completed successfully on each floor.

#### **6.2.3.2 Sway during quiet stance**

Sway during quiet stance was assessed with the participant’s eyes open and closed. The floor surfaces were positioned on a rigid platform mounted to a force plate (model 4060-15, Bertec Corp., Columbus, OH, USA). Each trial was 10 s in duration. The participant stood with her feet shoulder width apart and arms hanging naturally at her sides. She was not allowed to move her feet, or to touch the investigator for support, during the trial.

During each trial I sampled from the force plate at 960 Hz and filtered the data with a dual-pass fourth-order Butterworth low pass filter using a 5 Hz cut-off frequency. The anterior-posterior (AP) and medial-lateral (ML) locations of the centre of pressure (COP) were then determined, taking into account the height of the floor samples. I calculated four measures of postural steadiness in the AP and ML directions: maximum range of COP, root mean square distance of sway from the mean COP location, mean velocity of COP, and mean frequency of COP (Mackey and Robinovitch, 2005, Prieto et al., 1996). Due to significant correlations between outcomes, this paper will report only on sway range (*range*) and velocity (*vel*).

#### **6.2.3.3 Backwards floor translation**

The participant stood on top of a wheeled perturbation platform for the backwards floor translation experiment (figure 6.3B). The platform was connected to a linear motor (Triology T4D with Compumotor GV6K driver/controller) and the movement profile was controlled using a custom routine (acceleration = 5 m/s<sup>2</sup>, velocity = 0.2 m/s). During each trial, the platform was displaced 26.5 cm posteriorly over two seconds, decelerated, paused for two seconds, and then returned slowly to the original position. The participant stood on the floor surface with her feet shoulder width apart and her arms crossed in front of her chest. To limit the influence of anticipation, I randomly initiated the perturbation between 0 and 10 seconds after the participant indicated she was 'ready'. The subject was required to maintain her balance during and following the perturbation without moving her feet. Hip flexion and arm movements were allowed. I calculated the outcome *success* as the proportion of the five trials completed successfully on each floor.

#### 6.2.3.4 Subjective ratings

The participant provided subjective feedback on each floor following the balance tests. She rated her confidence in maintaining balance during daily activities (*confidence*) on a scale ranging from 1 to 10, with 1 corresponding to ‘no confidence, this floor will definitely cause me to fall’, and 10 corresponding to ‘confident, a fall is no more likely than on a typical rigid floor’. She also rated the perceived utility or practicality of each floor for everyday use (*utility*), with 1 corresponding to ‘terrible, I would never want to install this in my home’, and 10 corresponding to ‘great, I would love to install this in my home’.

#### 6.2.4 Statistics

For the force attenuation experiment, I used two-factor randomized groups ANOVA to determine whether  $F_{neck\_atten}$  associated with impact velocity and floor condition. For the balance experiments, I used logistic regression to test whether the proportion of successful trials (*success*) was associated with floor and task (restricted Get Up and Go test and backwards floor translation). I also used 1-factor repeated measures ANOVA to test the effect of floor on  $GUG_{time}$ , *confidence*, and *utility*, and 2-factor repeated measures ANOVA to assess the influence of floor and vision on sway variables. When results indicated significant main effect associations, I used independent t-tests ( $F_{neck\_atten}$ ), Wald Chi-Square (*success*), or paired t-tests ( $GUG_{time}$ , *confidence*, *utility*, sway variables) to identify differences between conditions. All analyses were performed with statistical analysis software (SPSS Version 15.0, SPSS Inc., Chicago, IL, USA or SAS Version 9.1, SAS Institute Inc., Cary, NC, USA) using a significance level of  $\alpha = 0.01$ .

## 6.3 Results

### 6.3.1 Force attenuation

Floor type was significantly associated with  $F_{neck\_atten}$  ( $F_{3,24} = 6093.4$ ,  $p < 0.001$ ) (table 6.1, figure 6.4). Average values of  $F_{neck\_atten}$  were 24.5 (SD 8.4) % for SmartCell, 47.2 (3.5) % for SofTile, 76.6 (11.3) % for Firm Foam, and 52.4 (14.1) % for Soft Foam (figure 6.5).

ANOVA results also indicated that impact velocity was associated with  $F_{neck\_atten}$  ( $F_{2,24} = 220.0$ ,  $p < 0.001$ ). In addition, I noted significant interactions between floors and impact velocity ( $F_{6,24} = 536.0$ ,  $p < 0.001$ ). As the impact velocity increased from 2 to 4 m/s,  $F_{neck\_atten}$  increased for SmartCell (from 17.3 to 33.7%) and SofTile (from 44.9 to 51.2%), but decreased for the Firm Foam (from 87.0 to 64.5%) and Soft Foam (from 66.1 to 37.9%) conditions.

### 6.3.2 Balance measures

There was a significant influence of floor condition on  $GUG_{time}$  ( $F_{4,56} = 19.1$ ,  $p < 0.001$ ). Paired t-tests demonstrated no significant differences between the rigid, SmartCell, and SofTile floors;  $GUG_{time}$  was significantly higher for the Firm Foam and Soft Foam conditions (by 19.7% for both) compared to the rigid floor (table 6.1, figure 6.5).

ANOVA results indicated a significant effect of floor on sway range and velocity (table 6.1 and figure 6.5;  $range\_ap$   $F_{4,56} = 87.3$ ,  $p < 0.001$ ;  $range\_ml$   $F_{4,56} = 50.9$ ,  $p < 0.001$ ;  $vel\_ap$   $F_{4,56} = 38.4$ ,  $p < 0.001$ ;  $vel\_ml$   $F_{4,56} = 41.2$ ,  $p < 0.001$ ). Paired t-tests

indicated that SmartCell was significantly different from the rigid floor for only *range\_ap* and *vel\_ap* in the eyes open condition. In contrast, the rigid floor was significantly different than SoftTile for all sway variables except for *range\_ml* in the eyes open condition, and significantly different from the Firm Foam and Soft Foam conditions for all sway outcomes in both vision conditions.

Loss of vision had significant effects on sway range and velocity (table 6.1; *range\_ap*  $F_{1,14} = 37.8$ ,  $p < 0.001$ ; *range\_ml*  $F_{1,14} = 20.6$ ,  $p < 0.001$ ; *vel\_ap*  $F_{1,14} = 44.6$ ,  $p < 0.001$ ; *vel\_ml*  $F_{1,14} = 25.9$ ,  $p < 0.001$ ). Outcome magnitudes were increased by an average of 49 (SD 13) % in the eyes closed condition.

I also observed significant interactions between floor and vision for sway range and velocity characterized by an increased effect of floor in the eyes closed condition (table 6.1; *range\_ap*  $F_{4,56} = 18.8$ ,  $p < 0.001$ ; *range\_ml*  $F_{4,56} = 19.8$ ,  $p < 0.001$ ; *vel\_ap*  $F_{4,56} = 15.5$ ,  $p < 0.001$ ; *vel\_ml*  $F_{4,56} = 17.3$ ,  $p < 0.001$ ).

Logistic regression indicated that *success* was significantly influenced by floor ( $\chi^2_{4,52} = 107.9$ ,  $p < 0.001$ ; figure 6.5) and task ( $\chi^2_{2,52} = 97.0$ ,  $p < 0.001$ ), and that a significant interaction existed ( $\chi^2_{4,52} = 13.9$ ,  $p = 0.008$ ) (table 6.1). In the restricted Get Up and Go test, there were no significant differences in *success* between the rigid, SmartCell, and SoftTile floors, while *success* was significantly lower for the Firm Foam and Soft Foam. In the backwards floor translation task, *success* was similar for all floors except the Firm Foam condition.

### 6.3.3 Subjective Ratings

Floor was significantly associated with participant ratings of *confidence* ( $F_{4,56} = 28.0, p < 0.001$ ) and *utility* ( $F_{4,56} = 57.0, p < 0.001$ ). There were no significant differences between the rigid, SmartCell, or SofTile floors, but both *confidence* and *utility* were significantly lower for the Firm Foam and Soft Foam conditions (table 6.1).

## 6.4 Discussion

In the current study, I examined the influence of floor stiffness (5 conditions) on the peak force applied to the proximal femur during simulated falls with a mechanical test system, and on measures of balance in elderly women across a range of activities of daily living. My results suggest that currently available flooring systems (e.g. SmartCell and SofTile floors) can attenuate femoral neck impact force by up to 47%, while causing only minimal effects on postural stability in elderly women. Further reductions in floor stiffness, while leading to greater force attenuation, caused substantial negative effects on postural sway, balance recovery ability, and balance confidence, and were rated as impractical by the study participants.

The effects of these floors on force attenuation and balance maintenance are directly related to their force-deflection properties (figures 6.1B and C). Of the four energy-absorbing floors, SmartCell is the stiffest and thinnest, and consequently its force attenuation was lowest (mean  $F_{neck\_atten} = 24.5\%$ ). However, it deflects less than 1 mm at forces associated with a typical 1000 N footfall (figure 6.1C), which explains its limited influence on balance measures. SofTile's slightly higher compliance enhances its force attenuation properties (mean  $F_{neck\_atten} = 47.2\%$ ), but at the cost of slightly increased

local deflection during a footfall (~ 4 mm). The stiffness of the foam floors are highly non-linear, dramatically increasing at relative strains exceeding 0.9. The larger area under the force-deflection curve for the Firm Foam indicates that it absorbs more energy than the Soft Foam before ‘bottoming out’, which corresponds to its higher  $F_{neck\_atten}$  at all impact velocities. The non-linear characteristics of the foam floors are likely responsible for the differential influence of impact velocity across floors. Specifically,  $F_{neck\_atten}$  decreased with increasing impact velocity for the Firm Foam and Soft Foam conditions which suggests that the foam materials bottomed out during the 2 m/s impacts, and thus provided little additional protection during the higher energy impacts. I observed the opposite trend for SmartCell and SofTile (i.e.  $F_{neck\_atten}$  increased with increasing impact velocity), which suggests that additional floor deflection (and energy absorption) was still available in these floors during the higher impact energy conditions. The initial low stiffness phase of the Soft Foam and Firm Foam materials may ideally suit their role as impact force attenuators in gymnasium mats, but the large local deflections preclude their use in settings where locomotion and mobility are essential.

It is insightful to examine the force attenuation provided by these floors compared to that reported in the literature. Mechanical fall simulators have demonstrated that force attenuation ranges up to 7% for wooden floors, 15% for carpets, and 24% for carpets in conjunction with common underpadding (Gardner et al., 1998, Maki and Fernie, 1990, Simpson et al., 2004). Although one group has reported force reductions as high as 73 and 56% for PVC foam underlying carpet and vinyl floors (Minns et al., 2004c, Nabhani and Bamford, 2002, 2004), they likely overstate the protective value of these floors as their mechanical test system did not account for the natural compliance of the pelvic

region during impact. Using a systematic design approach, Casalena et al. (1998a, 1998b) developed a floor that attenuated the peak force applied to the proximal femur by 15% while deflection less than 1 mm at forces associated with walking. In general, the force attenuation provided by the floors I examined exceeded those reported in the literature, which suggests that these floors have benefits beyond standard energy-absorbing flooring options such as carpets and rugs.

Our results also contribute to the inconclusive literature that examines the effects of common energy-absorbing floors on balance. Redfern et al. (1997) reported that anterior-posterior sway increased for elderly persons standing on thick carpet when exposed to a moving visual environment, and used this to suggest that the force attenuation properties of energy-absorbing floors might be outweighed by an increase in fall risk. However, as floor stiffness only influenced sway during conditions of deliberate sensory conflict, the external validity of these results is questionable. In contrast, Dickenson et al. (2002, 2001) found no difference in sway on rigid compared to carpeted floors. Reports of preferred gait velocity are similarly inconclusive with one study finding a decrease of 5% on carpet compared to parquetry (Stephens and Goldie, 1999) and another reporting an increase of 20% for carpet compared to vinyl (Willmott, 1986). My results indicate that SofTile had an observable influence on sway during quiet stance, but I observed no impairments for  $GUG_{time}$ , *success*, or *confidence*. SmartCell influenced balance to an even lesser degree - outcomes were significantly different from the rigid floor for only two of the thirteen possible balance variable / condition combinations (table 6.1). Overall, these results indicate that low stiffness floors can substantially attenuate impact force with only minimal coincident impairments in balance.

An important question is the economic feasibility of installing low stiffness flooring in high risk environments. To address this issue, I performed a basic cost-benefit analysis to evaluate the attractiveness of investing in this technology. Similar to Zacker et al. (1998), my approach modeled the use of low stiffness floors in four residential care rooms with a total footprint 92.9 m<sup>2</sup>. Based on discussions with manufacturers the approximate purchase and installation costs for floors similar to SmartCell or SofTile are 161 CAD / m<sup>2</sup>, compared to 27 CAD / m<sup>2</sup> for typical hospital-grade vinyl (the ‘standard’ condition). I assumed that the energy-absorbing floor would be covered by this vinyl layer, and thus would require no additional maintenance costs. Consequently, differential cost for the energy-absorbing floor was 15000 CAD. Regarding potential benefits, I assumed that: existing screening practices could identify eight high-risk persons to live in the intervention rooms, the energy-absorbing floor would have no influence on fall rate, and that each high-risk person would experience five falls per year in their rooms (a total of 40 falls) (Zacker and Shea, 1998). I further assumed that the intervention floor would halve the standard fracture rate of 2 hip fractures per 100 falls (Nevitt et al., 1991, Tinetti et al., 1988), resulting in a differential incidence of 0.4 hip fractures per year. Using an estimated direct cost for hip fracture treatment of 26,500 CAD (Wiktorowicz et al., 2001), this analysis predicts the energy-absorbing floor to have direct benefits of 10,600 CAD per year and a pay-off period of approximately 1.5 years. These results suggest that low stiffness floors can reduce the burden associated with hip fractures in a fiscally responsible manner even without including indirect cost savings associated with morbidity and mortality.

The study's major strength was my simultaneous examination of the two competing demands of low stiffness floors: force attenuation and balance impairment. I used a validated hip impact simulator (see Chapter 4) to measure the force attenuation provided by floors during simulated sideways falls, and enhanced the generalizability of the results by simulating a range of fall severities. I also conducted a comprehensive balance assessment on all floors including tests of balance maintenance, postural adjustment to voluntary movements, and ability to respond to external perturbations (Chiu et al., 2003). Furthermore, the tasks chosen represented those for which fall incidence is relatively high (e.g. gait and transitional tasks such as sit-to-stand) and / or those which are significantly associated with fall risk. Finally, I conducted my balance tests using a sample of older women which match the likely target population for low stiffness flooring in high risk environments such as hospitals and nursing homes.

There were also potential limitations associated with my study. First, I did not examine the effects on balance of transitioning from rigid to low stiffness floors, which when unexpected, can result in trunk instability (Marigold and Patla, 2005). To minimize this possibility, compliant floors could be installed using well-marked transitional gradients between areas of high and low stiffness to allow for pre-programmed motor program adjustments (Ferris et al., 1998, Kerdok et al., 2002, MacLellan and Patla, 2006). Second, this study did not incorporate feasibility assessments from the perspective of residential care staff. I acknowledge that such input is essential in selecting floors for use in practice. Third, as I examined surrogates of fall risk, rather than fall incidence itself, it is possible that the floors I examined have influences on human balance that were not captured in the current study. Additional studies (both in laboratory and clinical

settings) could enhance the robustness of fall-risk predictions by building upon the cadre of balance and stability tests performed in this study. Finally, I confined my focus to the influence of energy-absorbing floors on hip fractures. Critical fall height (CFH – the fall height above which more than 55% of fallers will sustain a serious head injury) data supplied by the manufacturers support my observations of increased force attenuation provided by SofTile (CFH = 3.05 m) compared to SmartCell (CFH = 1.07 m). However, a clearer picture of the potential benefits of low stiffness floors could be provided through a comprehensive evaluation of other fall-related injuries including wrist fractures and concussions.

Preventive measures are required to stem the expected increase in hip fractures in the coming decades. Biomechanical tests in our laboratory demonstrate that commercially available hip protectors can reduce the force applied to the proximal femur by up to 35% during simulated sideways falls (see Chapter 4). The current study demonstrates that low stiffness floors (an alternative or complimentary intervention option) offer approximately 2-fold greater force attenuation than hip protectors, and are unlikely to increase fall risk compared to typical rigid floors. Unlike wearable hip protectors which require active user compliance to reduce fracture risk, low stiffness flooring is a passive intervention strategy that is not dependent on user compliance. Furthermore, my basic cost-benefit analysis indicates that these floors are a fiscally responsible investment, with a projected pay-off period of only 1.5 years. Overall, this comprehensive study suggests that low stiffness floors are a promising intervention for reducing hip fracture risk, and supports the development of clinical trials to test their effectiveness in high risk settings including hospitals, rehabilitation centers, and

residential housing settings. The results also indicate the need for biomechanical test standards to guide market approval of this technology.

**Table 6-1: Average (SD) of the force attenuation, balance, and participant rating outcomes across floors.**

		Rigid	SmartCell	SofTile	Firm Foam	Soft Foam
$F_{neck\_atten}$ (%) <sup>a</sup>	2 m/s	0	17.3 (0.9)	44.9 (0.1)	87.0 (0.1)	66.1 (0.4)
	3 m/s	0	22.5 (0.3)	45.5 (1.0)	78.3 (0.8)	53.1 (2.2)
	4 m/s	0	33.7 (0.2)	51.2 (0.3)	64.5 (0.9)	37.9 (0.6)
$GUG_{time}$ (s)		11.7 (1.9)	11.4 (1.9)	12.0 (2.4)	14.0 (3.2) <sup>b</sup>	14.0 (3.1) <sup>b</sup>
$Success$ <sup>c</sup>	BFT	0.79 (0.36)	0.85 (0.32)	0.79 (0.33)	0.45 (0.42) <sup>b</sup>	0.67 (0.38)
	GUG	0.65 (0.40)	0.53 (0.36)	0.52 (0.38)	0.27 (0.34) <sup>b</sup>	0.25 (0.27) <sup>b</sup>
$range_{ap}$ (mm) <sup>d</sup>	EO	15.1 (4.0)	18.7 (5.5) <sup>b</sup>	20.3 (4.4) <sup>b</sup>	33.3 (6.2) <sup>b</sup>	23.4 (5.5) <sup>b</sup>
	EC	17.4 (5.3)	19.8 (5.8)	25.1 (7.3) <sup>b</sup>	54.8 (17.8) <sup>b</sup>	31.6 (8.1) <sup>b</sup>
$range_{ml}$ (mm) <sup>d</sup>	EO	6.3 (2.3)	7.5 (2.9)	8.3 (2.7)	26.7 (10.8) <sup>b</sup>	12.4 (5.5) <sup>b</sup>
	EC	7.1 (2.3)	7.5 (2.5)	10.9 (4.6) <sup>b</sup>	44.2 (22.5) <sup>b</sup>	15.2 (8.1) <sup>b</sup>
$vel_{ap}$ (mm/s) <sup>d</sup>	EO	8.0 (2.4)	9.1 (3.2) <sup>b</sup>	9.3 (2.3) <sup>b</sup>	20.3 (6.5) <sup>b</sup>	12.8 (2.4) <sup>b</sup>
	EC	11.5 (3.6)	12.1 (4.0)	15.1 (5.9) <sup>b</sup>	40.3 (18.8) <sup>b</sup>	21.1 (6.6) <sup>b</sup>
$vel_{ml}$ (mm/s) <sup>d</sup>	EO	3.6 (0.9)	3.9 (1.1)	4.4 (1.1) <sup>b</sup>	11.0 (4.5) <sup>b</sup>	6.7 (2.1) <sup>b</sup>
	EC	4.4 (1.6)	4.7 (1.3)	5.8 (2.0) <sup>b</sup>	19.8 (10.4) <sup>b</sup>	8.3 (3.3) <sup>b</sup>
$confidence$		9.5 (1.1)	9.6 (0.6)	9.3 (1.5)	5.4 (2.8) <sup>b</sup>	5.5 (3.1) <sup>b</sup>
$Utility$		9.1 (2.1)	8.5 (1.8)	8.2 (2.4)	2.6 (2.6) <sup>b</sup>	1.9 (2.4) <sup>b</sup>

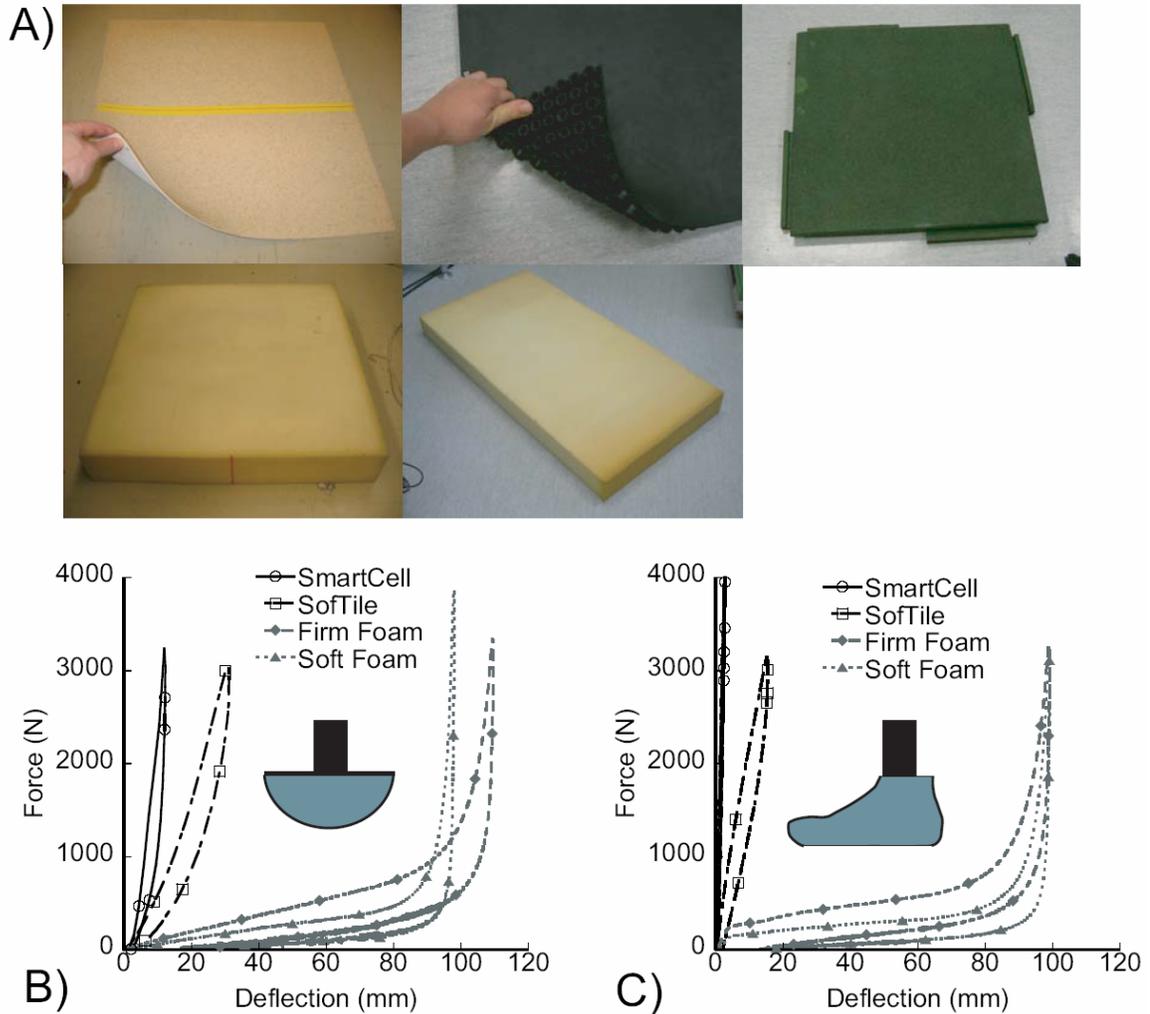
<sup>a</sup> all floors significantly different when averaged across impact velocities ( $p < 0.001$ )

<sup>b</sup> significantly different from the rigid floor ( $p < 0.001$ )

<sup>c</sup> BFT refers to the backwards floor translation task; GUG refers to the restricted Get Up and Go task

<sup>d</sup> EO and EC refer to the Eyes Open and Eyes Closed conditions, respectively

**Figure 6-1: Details of the floor conditions tested. A) Pictures (clockwise from top left): Rigid, SmartCell, SofTile, Firm Foam, Soft Foam. B) Force-deflection properties using a hip-shaped indenter. C) Force-deflection properties using a foot-shaped indenter.**



**Figure 6-2: Simon Fraser University Hip Impact Simulator. A) Entire system indicating floor samples positioned above the force plate. B) Surrogate pelvis with soft tissues removed to illustrate the proximal femur and location of the femoral neck load cell.**

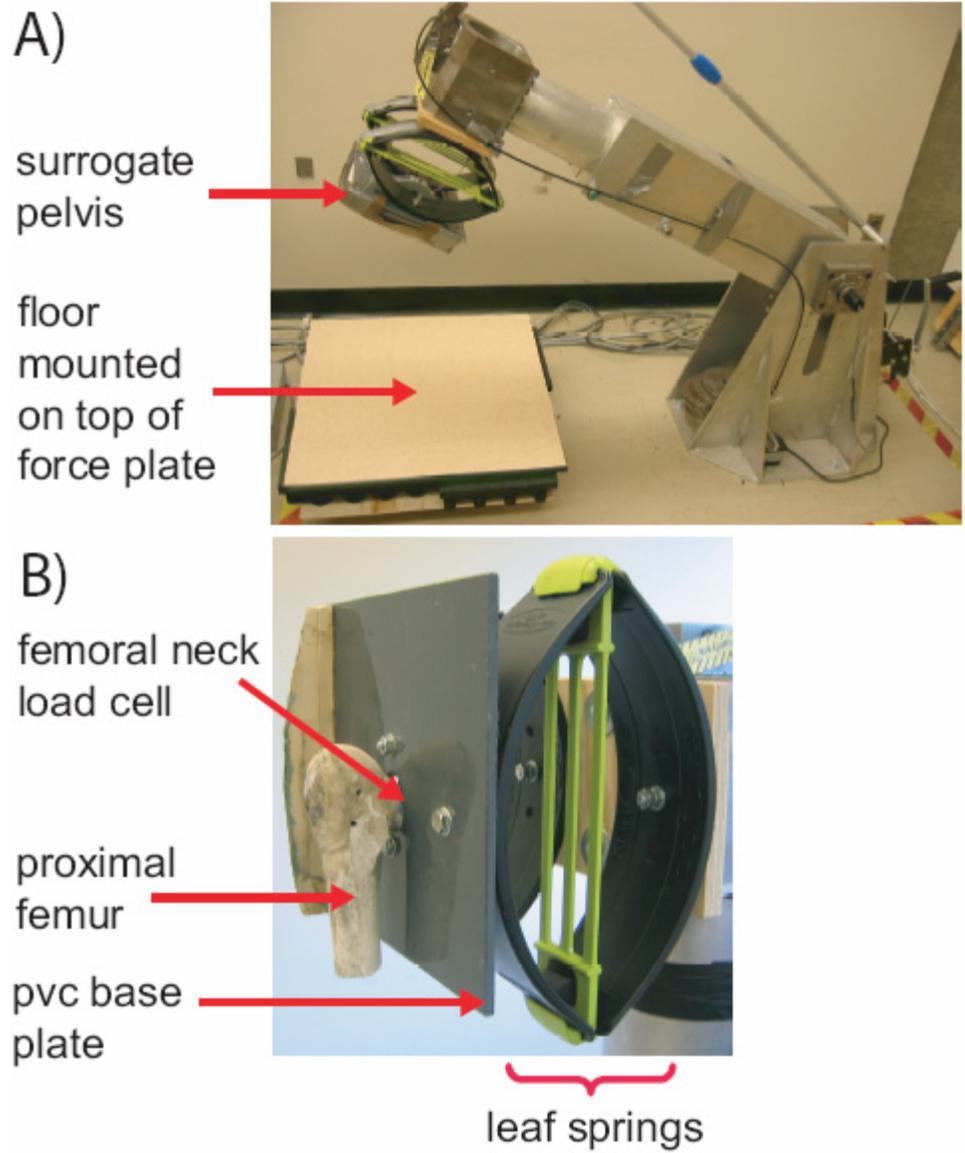
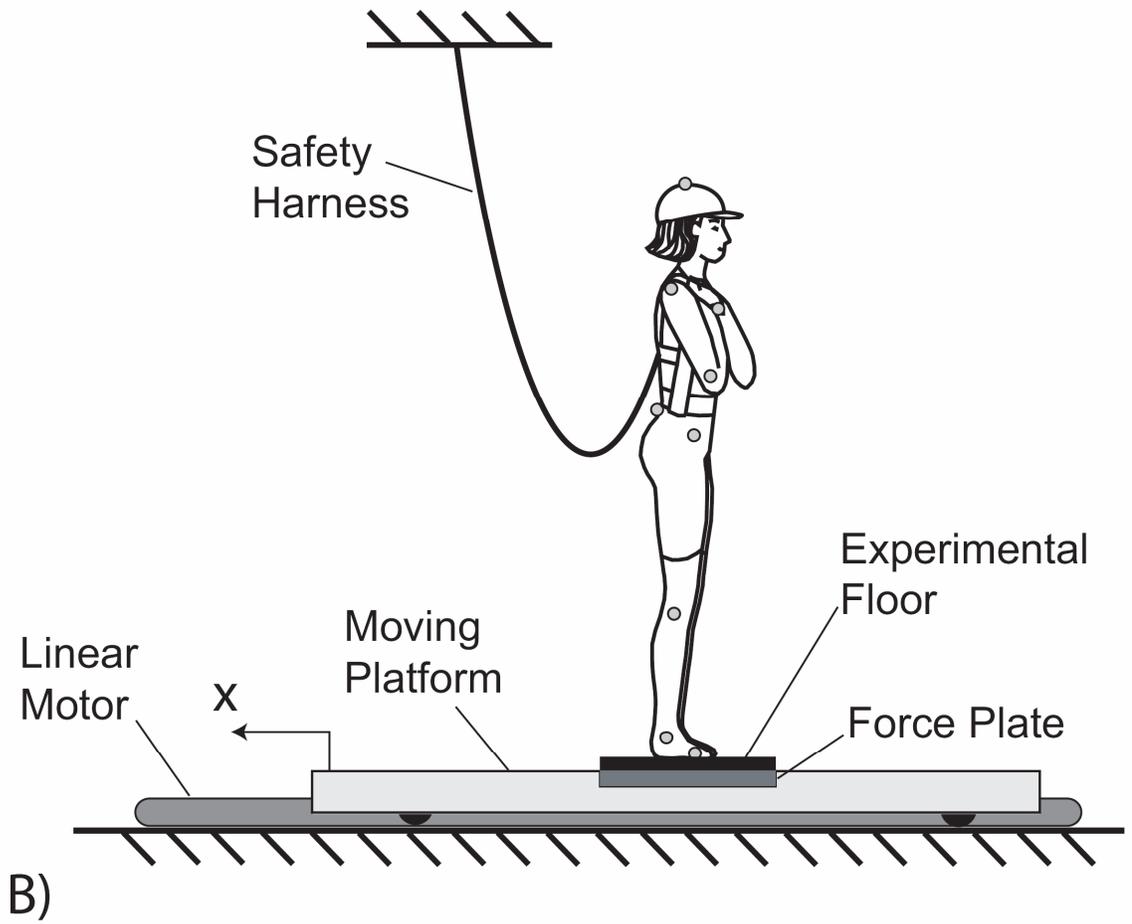
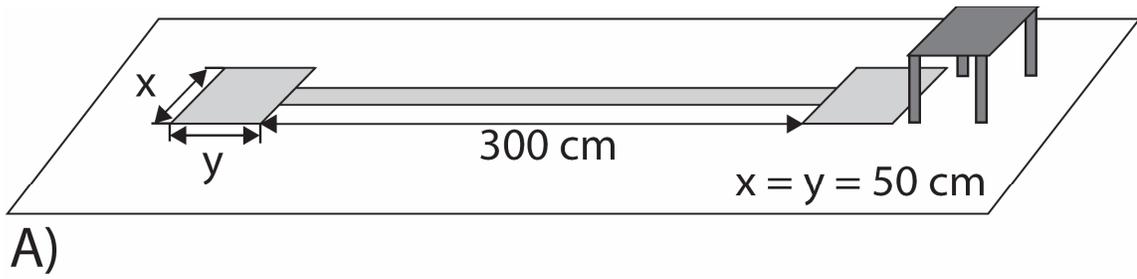
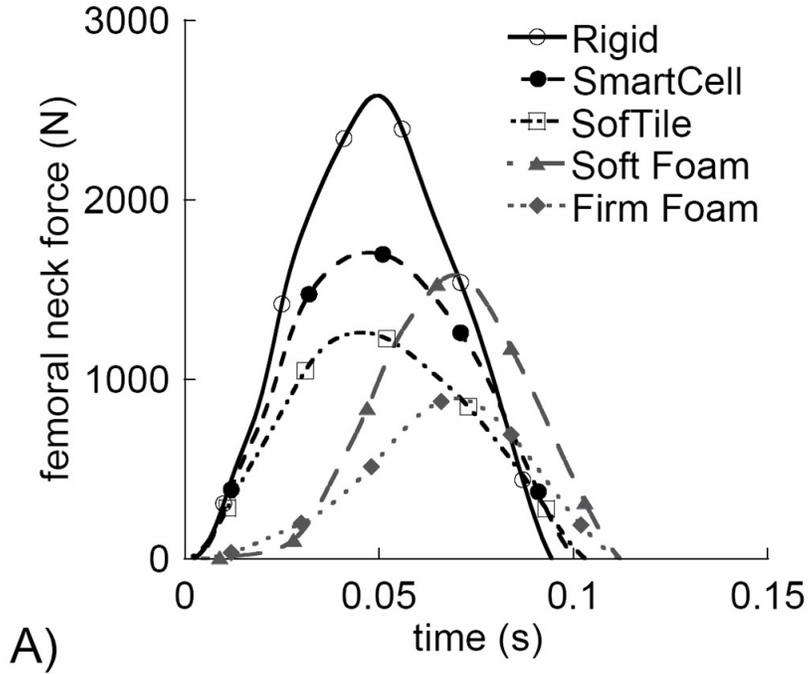


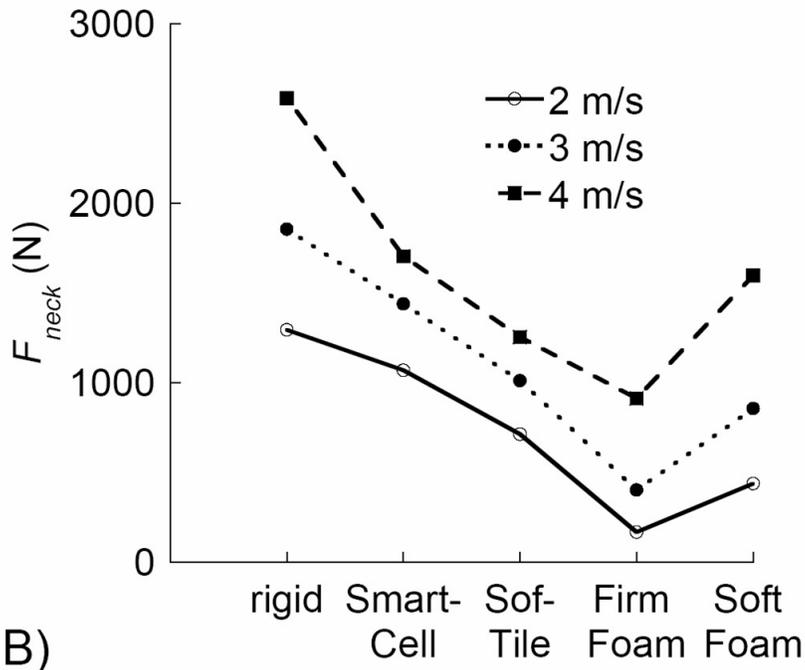
Figure 6-3: Schematic of: A) acceptable gait path and turn-around zones for restricted Get Up and Go Test; and B) backwards floor translation task.



**Figure 6-4: Impact force measures across floor conditions. A) Femoral neck force vs. time at an impact velocity of 4 m/s. B)  $F_{neck}$  at 2, 3, and 4 m/s impact velocities.**



A)



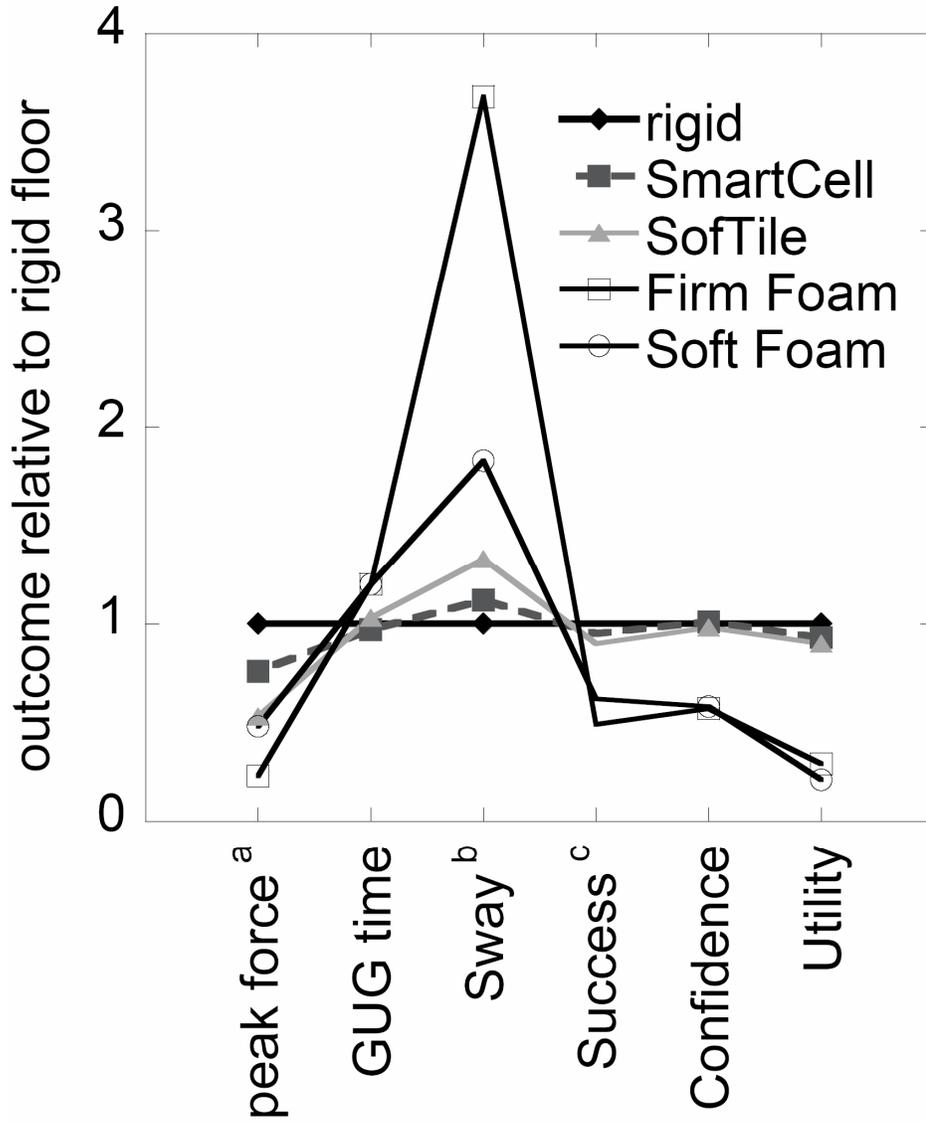
B)

**Figure 6-5: Results from the low stiffness floor conditions relative to the rigid floor. Hip fracture risk would be minimized by a floor with a score of 0 for ‘peak force’ values of 1 for all other variables.**

<sup>a</sup> averaged across all impact velocity conditions

<sup>b</sup> averaged across restricted Get Up and Go and floor translation tasks

<sup>c</sup> averaged across sway range and velocity in both vision conditions



## **CHAPTER 7 THESIS SYNTHESIS AND CONCLUSION**

The motivation underlying my doctoral research is to develop improved strategies for slowing or reversing the projected increase in hip fracture incidence among older adults in the coming decades (Jaglal et al., 1996, Papadimitropoulos et al., 1997, Wiktorowicz et al., 2001). Hip fractures and the resulting rehabilitation period are painful for the sufferers, who in most cases never fully regain their pre-fracture health status and independence. Hip fractures also generate enormous financial demands on health care systems (Papadimitropoulos et al., 1997). Similar demographic trends in the USA, Europe, and Asia underscore the global importance of hip fracture prevention.

In acknowledgement of the multi-factorial etiology of hip fractures, prevention strategies have targeted a wide range of risk factors. Pharmacologic interventions may decrease risk for persons with low bone mineral density by maintaining or increasing bone strength, but are inappropriate for the majority of hip fracture patients who do not suffer from osteoporosis (Dargent-Molina et al., 1996, Taylor et al., 2004). Ninety percent of hip fractures are the result of falls, and there is considerable evidence of the effectiveness of multi-factorial prevention and management programs (including exercise) in reducing fall rates (Chang et al., 2004). However, to date there is little evidence to support the value of such programs in reducing fracture rates. This may be due in part to insufficient study power, or the possibility that these programs are less effective in reducing falls in the highest-risk elderly or in preventing those falls that create greatest risk for fracture. Based on the limitations associated with current

preventive strategies, it is apparent that the continued development of alternative intervention approaches is required.

As hip fracture risk is directly dependent on the force applied to the proximal femur, wearable hip protectors and compliant floors represent promising approaches for attenuating impact force and reducing fracture risk in the event of a fall. Initial trials utilizing cluster randomization of long-term care facilities provided evidence of the effectiveness of hip protectors in reducing hip fracture incidence in clinical practice (Ekman et al., 1997, Harada et al., 2001, Kannus et al., 2000, Lauritzen et al., 1993). However, more recent clinical studies based on individual randomization have failed to replicate these positive findings. This may be due in part to the use of hip protectors with questionable design features and force attenuation properties (Kiel et al., 2007). Consequently, there have been recent calls for an increased biomechanical basis for the design and selection of products tested in clinical trials (Kannus et al., 2003). My thesis addresses this need by using biomechanical tools to advance our understanding of how force is generated and distributed to the hip region during a fall, and how this is affected by hip protectors and compliant flooring.

My initial study provides the first available measures of in-vivo force-deflection properties of the human hip during lateral impact, to complement previous estimates obtained through analysis of the body's free vibration response. The results demonstrate that peak force is most accurately predicted by a single degree of freedom mass-spring model with a stiffness element that is non-linear for forces below 300 N, and linear above this transition threshold. Furthermore, model simulation of worst-case falls suggest that peak impact force for an average woman is approximately 3400 N, sufficient to fracture

the proximal femur in approximately 50% of elderly women. These results improve our ability to predict the risk for hip fracture associated with a given fall, and inform the development of mathematical and mechanical systems for the design and testing of hip protectors and compliant flooring.

In my second study I conducted additional experiments with young women to test whether currently available soft shell hip protectors alter the magnitude and distribution of force applied to the hip region during sideways falls. My results demonstrate that through a combination of energy-shunting and energy-absorption mechanisms, soft shell hip protectors substantially reduce (by up to 76%) the pressure over the greater trochanter, while only modestly reducing (by up to 19%) the total force applied to the skin contact area. This study is the first to examine how force is distributed throughout the hip region during sideways falls, and document how this distribution is altered by hip protectors. Furthermore, the striking reductions in peak pressure support the value of clinical trials to test whether soft shell protectors prevent hip fractures in vulnerable populations.

Expanding on the results of these low-energy falling experiments with humans, my third study focussed on the improved design of a mechanical test rig to simulate realistic (and potentially injurious) falls. Although wearable hip protectors represent a promising strategy for preventing hip fractures, there is lack of agreement on biomechanical testing standards to guide market approval and product selection for clinical trials. I addressed this issue by designing a fall impact simulator which incorporates a “biofidelic” surrogate pelvis which matches the surface geometry and soft tissue stiffness measured in elderly women. I then used this system to measure the

attenuation in peak femoral neck force provided by two commercially available soft shell protectors (Safehip Soft and Hipsaver) and one rigid shell protector (Safehip Classic). Finally, I examined how the force attenuation provided by each protector was influenced by systematic changes in fall severity (impact velocity), body size (pelvis size), and soft tissue stiffness. My results indicate that, under biofidelic testing conditions, the soft shell hip protectors I examined generally provided greater force attenuation (averaging up to 27%) than the hard shell protector. However, measured values of force attenuation and the rank ordering of performance between the protectors varied with impact velocity, pelvic size, and pelvic soft tissue stiffness. This study provides the first documentation of pelvis geometry and soft tissue stiffness data from a sample of vulnerable aged women, and demonstrates that test systems whose characteristics deviate from these values will provide erroneous estimates of the force attenuation provided by hip protectors. These results are already being utilized by other groups to guide the development of international testing standards for hip protectors.

Compliant flooring represents a promising but understudied strategy for reducing impact force and hip fracture risk due to falls in high-risk environments such as nursing homes, hospitals, gymnasiums, and senior centres. In my fourth study I conducted additional “pelvis release experiments” with young women to test whether peak hip impact force is influenced by floor stiffness, and whether this relationship depends on body configuration at impact. Trials were conducted for rigid floor conditions, and with layers of EVA foam rubber overlying the floor that I regarded as firm (1.5 cm thick; stiffness = 263 kN/m), semi-firm (4.5 cm thick; stiffness = 95 kN/m), semi-soft (7.5 cm thick; stiffness = 67 kN/m), and soft (10.5 cm thick; stiffness = 59 kN/m). When

compared to the rigid floor condition, peak hip impact force averaged 8% lower in the firm condition and 15% lower in the semi-firm condition. Peak forces were not significantly different between the semi-firm, semi-soft, and soft floor conditions, indicating that a 4.5 cm thick foam mat provides nearly the same force attenuation as a 10.5 cm thick mat. These results indicate that common foam-rubber materials have the potential to reduce hip fracture risk by reducing the force applied to the hip region during falls regardless of body configuration at impact. They also support further laboratory studies to examine the effect of a wider range of floor stiffnesses on force attenuation during falls, and on balance and mobility during daily activities.

I extended this line of enquiry in my final study by examining whether a range of common compliant floors (e.g. playground surfaces, anti-fatigue mats, gymnasium mats) could effectively reduce the impact loads applied to the femur without coincident impairments in balance and mobility in older women. I used the hip impact simulator developed in my third study to assess the attenuation in femoral neck force provided by four floors (SmartCell, SofTile, Firm Foam, Soft Foam), when compared to a standard rigid floor. In addition, I assessed the influence of these floors on performance during a variety of static and dynamic balance tests with fifteen elderly women. The results demonstrate substantial differences in the mean attenuation in peak femoral neck force provided by the SmartCell (24.5%), SofTile (47.2%), Firm Foam (76.6%), and Soft Foam (52.4%) floors. However, there were no significant differences between the rigid, SmartCell, and SofTile floors in trial success, Get Up and Go time, balance confidence or utility ratings. Furthermore, the rigid floor was significantly different from SmartCell for only two of the eight sway outcome/vision combinations, compared to seven or more

significant differences for the SofTile, Firm Foam and Soft Foam conditions. A basic cost-benefit analysis predicted a pay-off period of 1.5 years for implementing these floors in a high-risk residential housing setting. These results suggest that two commercially available low stiffness floors can attenuate femoral impact force by up to 50% without substantially impairing balance in older women. This comprehensive study contributes to the literature by systematically examining how compliant flooring affects two competing dimensions of hip fracture risk: applied force and fall risk. The results significantly enhance the biomechanical evidence base surrounding compliant floors, and support the development of clinical trials to test their effectiveness at reducing fall-related injuries in high risk settings, such as hospitals and nursing homes.

In addition to advancing our understanding of the value of hip protectors and compliant flooring in reducing hip fracture risk, this thesis raises important questions to address through additional studies. First, my experimental approaches investigated only sideways falls with direct impact to the hip. Additional studies are required to determine the full range of impact configurations that are observed clinically, and to examine the force attenuation properties of protective devices (hip protectors in particular) for these conditions. Second, as hip fracture risk is markedly increased for women with low body mass index, there is great interest in the influence of body habitus on the effectiveness of these interventions. My second study indicates that soft shell hip protectors may attenuate impact force more effectively for slender persons who have less natural soft tissue padding over the proximal femur. However, additional studies are required to more systematically evaluate this issue and determine if certain styles of hip protectors are most effective for persons with specific anthropometric characteristics. Third, in contrast

to low stiffness flooring (a passive intervention), the clinical effectiveness of hip protectors is dependent on user compliance (both initial acceptance and adherence). Compliance in clinical trials has averaged only 50% (van Schoor et al., 2002), and appears to be dependent on characteristics of the user (e.g. dementia, dependency on staff, fear of falling), residence (e.g. size of nursing home), and staff (e.g. training and education). Issues of compliance need to be further examined from multiple perspectives (pad materials, garment design, instructions and training) to develop strategies to maximize the clinical effectiveness of these devices. Finally, although not examined in this thesis, compliant floors have the potential to reduce the risk of other fall-related injuries, including vertebral fractures, concussion, and wrist fracture. Consequently, additional biomechanical studies are required to provide a comprehensive overview of the total protective benefit associated with this intervention. Such information would support enhanced economic analysis models to more accurately estimate the potential financial benefits of investing in this technology.

In summary, this thesis significantly enhances the biomechanical evidence-base surrounding the use of hip protectors and compliant flooring to reduce hip fracture risk. The robustness of the research findings were enhanced through the combined use of experimental and mathematical modelling approaches. In addition to exploring fundamental research questions surrounding human impact dynamics, this thesis reports on the biomechanical effectiveness of specific intervention strategies. The substantial force attenuation (up to 50%) measured for some compliant floors with only minimal coincident impairment in balance (in the tests we conducted) indicates a promising avenue for preventive efforts. This thesis also demonstrates that, despite practical

limitations in their thickness and surface area, wearable hip protectors can attenuate the force applied to the hip during falls by up to 35%. Overall, these data support the initiation of clinical trials to test the effectiveness of compliant floors and soft shell hip protectors in practice, and provide a biomechanical evidence-base for the selection of intervention products. Of equal importance, this thesis demonstrates the need for developing international standards for the biomechanical testing and market approval of these devices. These are essential steps for increasing the quality of hip protectors and compliant floors available in the marketplace, and consequently, for enhancing their overall effectiveness at reducing hip fracture risk in vulnerable populations.

## REFERENCES

- Askegaard, V., Lauritzen, J. B., 1995. Load on the hip in a stiff sideways fall. *European Journal of Musculoskeletal Research*. 4, 111-116.
- Betker, A. L., Moussavi, Z. M., Szturm, T., 2005. On modeling center of foot pressure distortion through a medium. *IEEE Trans Biomed Eng*. 52, 345-52.
- Birks, Y. F., Hildreth, R., Campbell, P., Sharpe, C., Torgerson, D. J., Watt, I., 2003. Randomised controlled trial of hip protectors for the prevention of second hip fractures. *Age Ageing*. 32, 442-4.
- Birks, Y. F., Porthouse, J., Addie, C., Loughney, K., Saxon, L., Baverstock, M., Francis, R. M., Reid, D. M., Watt, I., Torgerson, D. J., 2004. Randomized controlled trial of hip protectors among women living in the community. *Osteoporos Int*. 15, 701-6.
- Black, D. M., Cummings, S. R., Karpf, D. B., Cauley, J. A., Thompson, D. E., Nevitt, M. C., Bauer, D. C., Genant, H. K., Haskell, W. L., Marcus, R., Ott, S. M., Torner, J. C., Quandt, S. A., Reiss, T. F., Ensrud, K. E., 1996. Randomised trial of effect of alendronate on risk of fracture in women with existing vertebral fractures. Fracture Intervention Trial Research Group. *Lancet*. 348, 1535-41.
- Black, D. M., Greenspan, S. L., Ensrud, K. E., Palermo, L., McGowan, J. A., Lang, T. F., Garnero, P., Bouxsein, M. L., Bilezikian, J. P., Rosen, C. J., 2003. The effects of parathyroid hormone and alendronate alone or in combination in postmenopausal osteoporosis. *N Engl J Med*. 349, 1207-15.
- Body, J. J., Diel, I., Bell, R., 2004. Profiling the safety and tolerability of bisphosphonates. *Seminars In Oncology*. 31, 73-8.
- Bouxsein, M. L., Szulc, P., Munoz, F., Thrall, E., Sornay-Rendu, E., Delmas, P. D., 2007. Contribution of trochanteric soft tissues to fall force estimates, the factor of risk, and prediction of hip fracture risk. *J Bone Miner Res*. 22, 825-31.
- Burl, J. B., Centola, J., Bonner, A., Burque, C., 2003. Hip protector compliance: a 13-month study on factors and cost in a long-term care facility. *J Am Med Dir Assoc*. 4, 245-50.
- Burr, D. B., Turner, C. H. (2003) Primer on the Metabolic Bone Diseases and Disorders of Mineral Metabolism, 5th ed., Chapter 8: Biomechanics of Bone. IN *FAVUS*, M. J. (Ed.) Washington, D.C., American Society for Bone and Mineral Research. 58-64.
- Cameron, I. D., Cumming, R. G., Kurrle, S. E., Quine, S., Lockwood, K., Salkeld, G., Finnegan, T., 2003. A randomised trial of hip protector use by frail older women living in their own homes. *Injury Prevention*. 9, 138-41.

- Cameron, I. D., Venman, J., Kurrle, S. E., Lockwood, K., Birks, C., Cumming, R. G., Quine, S., Bashford, G., 2001. Hip protectors in aged-care facilities: a randomized trial of use by individual higher-risk residents. *Age and Ageing*. 30, 477-81.
- Campbell, A. J., Borrie, M. J., Spears, G. F., 1989. Risk factors for falls in a community-based prospective study of people 70 years and older. *J Gerontol*. 44, M112-7.
- Casalena, J. A., Badre-Alam, A., Ovaert, T. C., Cavanagh, P. R., Streit, D. A., 1998a. The Penn State Safety Floor: Part II--Reduction of fall-related peak impact forces on the femur. *J Biomech Eng*. 120, 527-32.
- Casalena, J. A., Ovaert, T. C., Cavanagh, P. R., Streit, D. A., 1998b. The Penn State Safety Floor: Part I--Design parameters associated with walking deflections. *J Biomech Eng*. 120, 518-26.
- Cauley, J. A., Seeley, D. G., Ensrud, K., Ettinger, B., Black, D., Cummings, S. R., 1995. Estrogen replacement therapy and fractures in older women. Study of Osteoporotic Fractures Research Group. *Ann Intern Med*. 122, 9-16.
- Cavanaugh, J. M., Walilko, J. T., Malhotra, A., Zhu, Y., King, A. I. (1990) Biomechanical response and injury tolerance of the pelvis in twelve sled side impacts. *In: Proceedings of the 34th Stapp Car Crash Conference*. Warrendale, PA, Society of Automotive Engineers.
- Chan, D. K., Hillier, G., Coore, M., Cooke, R., Monk, R., Mills, J., Hung, W. T., 2000. Effectiveness and acceptability of a newly designed hip protector: a pilot study. *Arch Gerontol Geriatr*. 30, 25-34.
- Chang, J. T., Morton, S. C., Rubenstein, L. Z., Mojica, W. A., Maglione, M., Suttrop, M. J., Roth, E. A., Shekelle, P. G., 2004. Interventions for the prevention of falls in older adults: systematic review and meta-analysis of randomised clinical trials. *BMJ*. 328, 680.
- Cheng, X. G., Lowet, G., Boonen, S., Nicholson, P. H., Brys, P., Nijs, J., Dequeker, J., 1997. Assessment of the strength of proximal femur in vitro: relationship to femoral bone mineral density and femoral geometry. *Bone*. 20, 213-8.
- Chiu, A. Y., Au-Yeung, S. S., Lo, S. K., 2003. A comparison of four functional tests in discriminating fallers from non-fallers in older people. *Disabil Rehabil*. 25, 45-50.
- Courtney, A. C., Wachtel, E. F., Myers, E. R., Hayes, W. C., 1994. Effects of loading rate on strength of the proximal femur. *Calcified Tissue International*. 55, 53-8.
- Courtney, A. C., Wachtel, E. F., Myers, E. R., Hayes, W. C., 1995. Age-related reductions in the strength of the femur tested in a fall-loading configuration. *J Bone Joint Surg Am*. 77, 387-95.
- Cryer, C., Knox, A., Stevenson, E., 2006. Factors associated with the initial acceptance of hip protectors amongst older people in residential care. *Age Ageing*. 35, 72-5.
- Cumming, R. G., Nevitt, M. C., Cummings, S. R., 1997. Epidemiology of hip fractures. *Epidemiol Rev*. 19, 244-57.

- Cummings, S. R., Black, D. M., Thompson, D. E., Applegate, W. B., Barrett-Connor, E., Musliner, T. A., Palermo, L., Prineas, R., Rubin, S. M., Scott, J. C., Vogt, T., Wallace, R., Yates, A. J., LaCroix, A. Z., 1998. Effect of alendronate on risk of fracture in women with low bone density but without vertebral fractures: results from the Fracture Intervention Trial. *JAMA*. 280, 2077-82.
- Cummings, S. R., Melton, L. J., 2002. Epidemiology and outcomes of osteoporotic fractures. *Lancet*. 359, 1761-7.
- Cummings, S. R., Nevitt, M. C., 1994. Non-skeletal determinants of fractures: the potential importance of the mechanics of falls. Study of Osteoporotic Fractures Research Group. *Osteoporos Int*. 4 Suppl 1, 67-70.
- Cummings, S. R., Nevitt, M. C., Browner, W. S., Stone, K., Fox, K. M., Ensrud, K. E., Cauley, J., Black, D., Vogt, T. M., 1995. Risk factors for hip fracture in white women. Study of Osteoporotic Fractures Research Group. *N Engl J Med*. 332, 767-73.
- Dargent-Molina, P., Favier, F., Grandjean, H., Baudoin, C., Schott, A. M., Hausherr, E., Meunier, P. J., Breart, G., 1996. Fall-related factors and risk of hip fracture: the EPIDOS prospective study. *Lancet*. 348, 145-9.
- Derler, S., Spierings, A. B., Schmitt, K. U., 2005. Anatomical hip model for the mechanical testing of hip protectors. *Medical Engineering & Physics*. 27, 475-85.
- Dickinson, J. I., Shroyer, J. L., Elias, J. W., 2002. The influence of commercial-grade carpet on postural sway and balance strategy among older adults. *Gerontologist*. 42, 552-9.
- Dickinson, J. I., Shroyer, J. L., Elias, J. W., Hutton, J. T., Gentry, G. M., 2001. The effect of selected residential carpet and pad on the balance of healthy older adults. *Environment and Behavior*. 33, 279-295.
- Eckstein, F., Wunderer, C., Boehm, H., Kuhn, V., Priemel, M., Link, T. M., Lochmuller, E. M., 2004. Reproducibility and side differences of mechanical tests for determining the structural strength of the proximal femur. *J Bone Miner Res*. 19, 379-85.
- Ekman, A., Mallmin, H., Michaelsson, K., Ljunghall, S., 1997. External hip protectors to prevent osteoporotic hip fractures. *Lancet*. 350, 563-4.
- Empana, J. P., Dargent-Molina, P., Breart, G., 2004a. Effect of hip fracture on mortality in elderly women: the EPIDOS prospective study. *Journal of the American Geriatrics Society*. 52, 685-90.
- Etheridge, B. S., Beason, D. P., Lopez, R. R., Alonso, J. E., McGwin, G., Eberhardt, A. W., 2005. Effects of trochanteric soft tissues and bone density on fracture of the female pelvis in experimental side impacts. *Ann Biomed Eng*. 33, 248-54.
- Feldman, F., Robinovitch, S. N., 2007. Reducing hip fracture risk during sideways falls: evidence in young adults of the protective effects of impact to the hands and stepping. *Journal of Biomechanics*. 40, 2612-2618.

- Ferris, D. P., Louie, M., Farley, C. T., 1998. Running in the real world: adjusting leg stiffness for different surfaces. *Proc Biol Sci.* 265, 989-94.
- Finkelstein, J. S., Hayes, A., Hunzelman, J. L., Wyland, J. J., Lee, H., Neer, R. M., 2003. The effects of parathyroid hormone, alendronate, or both in men with osteoporosis. *N Engl J Med.* 349, 1216-26.
- Forsen, L., Sandvig, S., Schuller, A., Sogaard, A. J., 2004. Compliance with external hip protectors in nursing homes in Norway. *Inj Prev.* 10, 344-9.
- Gardner, T. N., Simpson, A. H., Booth, C., Sprukkelhorst, P., Evans, M., Kenwright, J., Evans, J. G., 1998. Measurement of impact force, simulation of fall and hip fracture. *Med Eng Phys.* 20, 57-65.
- Gill, J., Allum, J. H., Carpenter, M. G., Held-Ziolkowska, M., Adkin, A. L., Honegger, F., Pierchala, K., 2001. Trunk sway measures of postural stability during clinical balance tests: effects of age. *J Gerontol A Biol Sci Med Sci.* 56, M438-47.
- Greenspan, S. L., Myers, E. R., Maitland, L. A., Resnick, N. M., Hayes, W. C., 1994. Fall severity and bone mineral density as risk factors for hip fracture in ambulatory elderly. *Journal of the American Medical Association.* 271, 128-33.
- Grisso, J. A., Kelsey, J. L., Strom, B. L., Chiu, G. Y., Maislin, G., O'Brien, L. A., Hoffman, S., Kaplan, F., 1991. Risk factors for falls as a cause of hip fracture in women. The Northeast Hip Fracture Study Group. *N Engl J Med.* 324, 1326-31.
- Groen, B. E., Weerdesteyn, V., Duysens, J., 2006. Martial arts fall techniques decrease the impact forces at the hip during sideways falling. *J Biomech.*
- Gullberg, B., Johnell, O., Kanis, J. A., 1997. World-wide projections for hip fracture. *Osteoporos Int.* 7, 407-13.
- Harada, A., Mizuno, M., Takemura, M., Tokuda, H., Okuizumi, H., Niino, N., 2001. Hip fracture prevention trial using hip protectors in Japanese nursing homes. *Osteoporos Int.* 12, 215-21.
- Hayes, W. C., Myers, E. R., Morris, J. N., Gerhart, T. N., Yett, H. S., Lipsitz, L. A., 1993. Impact near the hip dominates fracture risk in elderly nursing home residents who fall. *Calcified Tissue International.* 52, 192-8.
- Hayes, W. C., Myers, E. R., Robinovitch, S. N., Van Den Kroonenberg, A., Courtney, A. C., McMahon, T. A., 1996. Etiology and prevention of age-related hip fractures. *Bone.* 18, 77S-86S.
- Hayes, W. C., Piazza, S. J., Zysset, P. K., 1991. Biomechanics of fracture risk prediction of the hip and spine by quantitative computed tomography. *Radiol Clin North Am.* 29, 1-18.
- Healey, F., 1994. Does flooring type affect risk of injury in older in-patients? *Nurs Times.* 90, 40-1.
- Hsiao, E. T., Robinovitch, S. N., 1998. Common protective movements govern unexpected falls from standing height. *J Biomech.* 31, 1-9.

- Hubacher, M., Wettstein, A., 2001. Acceptance of hip protectors for hip fracture prevention in nursing homes. *Osteoporos Int.* 12, 794-9.
- Jaglal, S. B., Sherry, P. G., Schatzker, J., 1996. The impact and consequences of hip fracture in Ontario. *Canadian Journal of Surgery.* 39, 105-11.
- Jaglal, S. B., Weller, I., Mamdani, M., Hawker, G., Kreder, H., Jaakkimainen, L., Adachi, J. D., 2005. Population trends in BMD testing, treatment, and hip and wrist fracture rates: are the hip fracture projections wrong? *J Bone Miner Res.* 20, 898-905.
- Kannus, P., Parkkari, J., Khan, K., 2003. Hip protectors need an evidence base. *Lancet.* 362, 1168-9.
- Kannus, P., Parkkari, J., Niemi, S., Pasanen, M., Palvanen, M., Jarvinen, M., Vuori, I., 2000. Prevention of hip fracture in elderly people with use of a hip protector. *New England Journal of Medicine.* 343, 1506-13.
- Kannus, P., Parkkari, J., Poutala, J., 1999. Comparison of force attenuation properties of four different hip protectors under simulated falling conditions in the elderly: an in vitro biomechanical study. *Bone.* 25, 229-35.
- Kannus, P., Parkkari, J., Sievanen, H., Heinonen, A., Vuori, I., Jarvinen, M., 1996. Epidemiology of hip fractures. *Bone.* 18, 57S-63S.
- Kannus, P., Uusi-Rasi, K., Palvanen, M., Parkkari, J., 2005. Non-pharmacological means to prevent fractures among older adults. *Ann Med.* 37, 303-10.
- Keegan, T. H., Kelsey, J. L., King, A. C., Quesenberry, C. P., Jr., Sidney, S., 2004. Characteristics of fallers who fracture at the foot, distal forearm, proximal humerus, pelvis, and shaft of the tibia/fibula compared with fallers who do not fracture. *Am J Epidemiol.* 159, 192-203.
- Kerdok, A. E., Biewener, A. A., McMahon, T. A., Weyand, P. G., Herr, H. M., 2002. Energetics and mechanics of human running on surfaces of different stiffnesses. *J Appl Physiol.* 92, 469-78.
- Kiel, D. P., Magaziner, J., Zimmerman, S., Ball, L., Barton, B. A., Brown, K. M., Stone, J. P., Dewkett, D., Birge, S. J., 2007. Efficacy of a hip protector to prevent hip fracture in nursing home residents: the HIP PRO randomized controlled trial. *JAMA: the journal of the American Medical Association.* 298, 413-22.
- Kim, K. J., Ashton-Miller, J. A., 2003. Biomechanics of fall arrest using the upper extremity: age differences. *Clinical Biomechanics (Bristol, Avon).* 18, 311-8.
- Lauritzen, J. B., 1997. Hip fractures. Epidemiology, risk factors, falls, energy absorption, hip protectors, and prevention. *Dan Med Bull.* 44, 155-68.
- Lauritzen, J. B., Askegaard, V., 1992. Protection against hip fractures by energy absorption. *Dan Med Bull.* 39, 91-3.
- Lauritzen, J. B., Petersen, M. M., Lund, B., 1993. Effect of external hip protectors on hip fractures. *Lancet.* 341, 11-3.
- Leavitt, F. J., 2003. Can martial arts falling techniques prevent injuries? *Inj Prev.* 9, 284.

- Lee, H. S., Kim, M. J., 1997. [A study on fall accident]. *Taehan Kanho*. 36, 45-62.
- Liberman, U. A., Weiss, S. R., Broll, J., Minne, H. W., Quan, H., Bell, N. H., Rodriguez-Portales, J., Downs, R. W., Jr., Dequeker, J., Favus, M., 1995. Effect of oral alendronate on bone mineral density and the incidence of fractures in postmenopausal osteoporosis. The Alendronate Phase III Osteoporosis Treatment Study Group. *N Engl J Med*. 333, 1437-43.
- Liu-Ambrose, T., Khan, K. M., Eng, J. J., Janssen, P. A., Lord, S. R., McKay, H. A., 2004. Resistance and agility training reduce fall risk in women aged 75 to 85 with low bone mass: a 6-month randomized, controlled trial. *J Am Geriatr Soc*. 52, 657-65.
- Lochmuller, E. M., Miller, P., Burklein, D., Wehr, U., Rambeck, W., Eckstein, F., 2000. In situ femoral dual-energy X-ray absorptiometry related to ash weight, bone size and density, and its relationship with mechanical failure loads of the proximal femur. *Osteoporos Int*. 11, 361-7.
- Lord, S. R., Clark, R. D., Webster, I. W., 1991. Postural stability and associated physiological factors in a population of aged persons. *J Gerontol*. 46, M69-76.
- Lord, S. R., Menz, H. B., 2000. Visual contributions to postural stability in older adults. *Gerontology*. 46, 306-10.
- Lotz, J. C., Hayes, W. C., 1990. The use of quantitative computed tomography to estimate risk of fracture of the hip from falls. *J Bone Joint Surg Am*. 72, 689-700.
- Lundin-Olsson, L., Nyberg, L., Gustafson, Y., 1998. Attention, frailty, and falls: the effect of a manual task on basic mobility. *J Am Geriatr Soc*. 46, 758-61.
- Mackey, D. C., Robinovitch, S. N., 2005. Postural steadiness during quiet stance does not associate with ability to recover balance in older women. *Clin Biomech (Bristol, Avon)*. 20, 776-83.
- MacLellan, M. J., Patla, A. E., 2006. Adaptations of walking pattern on a compliant surface to regulate dynamic stability. *Exp Brain Res*. 173, 521-30.
- Maitland, L. A., Myers, E. R., Hipp, J. A., Hayes, W. C., Greenspan, S. L., 1993. Read my hips: measuring trochanteric soft tissue thickness. *Calcif Tissue Int*. 52, 85-9.
- Maki, B. E., Fernie, G. R., 1990. Impact attenuation of floor coverings in simulated falling accidents. *Applied Ergonomics*. 21, 107-14.
- Maki, B. E., Holliday, P. J., Fernie, G. R., 1990. Aging and postural control. A comparison of spontaneous- and induced-sway balance tests. *J Am Geriatr Soc*. 38, 1-9.
- Maki, B. E., Holliday, P. J., Topper, A. K., 1994. A prospective study of postural balance and risk of falling in an ambulatory and independent elderly population. *J Gerontol*. 49, M72-84.
- Manske, S. L., Liu-Ambrose, T., de Bakker, P. M., Liu, D., Kontulainen, S., Guy, P., Oxland, T. R., McKay, H. A., 2006. Femoral neck cortical geometry measured

- with magnetic resonance imaging is associated with proximal femur strength. *Osteoporos Int.* 17, 1539-45.
- Marigold, D. S., Patla, A. E., 2005. Adapting locomotion to different surface compliances: neuromuscular responses and changes in movement dynamics. *J Neurophysiol.* 94, 1733-50.
- Mathias, S., Nayak, U. S., Isaacs, B., 1986. Balance in elderly patients: the "get-up and go" test. *Arch Phys Med Rehabil.* 67, 387-9.
- McClung, M. R., Geusens, P., Miller, P. D., Zippel, H., Bensen, W. G., Roux, C., Adami, S., Fogelman, I., Diamond, T., Eastell, R., Meunier, P. J., Reginster, J. Y., 2001. Effect of risedronate on the risk of hip fracture in elderly women. Hip Intervention Program Study Group. *N Engl J Med.* 344, 333-40.
- McMahon, T. A., Greene, P. R., 1979. The influence of track compliance on running. *J Biomech.* 12, 893-904.
- McMahon, T. A., Valiant, G., Frederick, E. C., 1987. Groucho running. *J Appl Physiol.* 62, 2326-37.
- Melton, L. J., 3rd, 1993. Hip fractures: a worldwide problem today and tomorrow. *Bone.* 14 Suppl 1, S1-8.
- Melton, L. J., 3rd, 2000. Who has osteoporosis? A conflict between clinical and public health perspectives. *J Bone Miner Res.* 15, 2309-14.
- Meyer, H. E., Tverdal, A., Falch, J. A., Pedersen, J. I., 2000. Factors associated with mortality after hip fracture. *Osteoporosis International.* 11, 228-32.
- Mills, N. J., 1996. The biomechanics of hip protectors. *Proceedings Of The Institution Of Mechanical Engineers. Part H, Journal Of Engineering In Medicine.* 210, 259-66.
- Mills, N. J. (2007) *Polymer Foams Handbook: Engineering and Biomechanics Applications and Design Guide*, Oxford, UK, Elsevier Ltd.
- Minns, J., 2006. How safe are hip protectors? Accessed via [http://www.profane.eu.org/profane\\_documents/how\\_safe\\_are\\_hip\\_protectors.pdf](http://www.profane.eu.org/profane_documents/how_safe_are_hip_protectors.pdf) July 26, 2006.
- Minns, J., Dodd, C., Gardner, R., Bamford, J., Nabhani, F., 2004a. Assessing the safety and effectiveness of hip protectors. *Nursing Standard (Royal College Of Nursing (Great Britain)).* 18, 33-8.
- Minns, J., Dodd, C., Gardner, R., Bamford, J., Nabhani, F., 2004b. Assessing the safety and effectiveness of hip protectors. *Nurs Stand.* 18, 33-8.
- Minns, J., Nabhani, F., Bamford, J. S., 2004c. Can flooring and underlay materials reduce hip fractures in older people? *Nurs Older People.* 16, 16-20.
- Nabhani, F., Bamford, J., 2002. Mechanical testing of hip protectors. *Journal of Materials Processing Technology.* 124, 311-318.
- Nabhani, F., Bamford, J., 2004. Impact properties of floor coverings and their role during simulated hip fractures. *Journal of Materials Processing Technology.* 153-154, 139-144.

- Nankaku, M., Kanzaki, H., Tsuboyama, T., Nakamura, T., 2005. Evaluation of hip fracture risk in relation to fall direction. *Osteoporos Int.* 16, 1315-20.
- Neer, R. M., Arnaud, C. D., Zanchetta, J. R., Prince, R., Gaich, G. A., Reginster, J. Y., Hodsman, A. B., Eriksen, E. F., Ish-Shalom, S., Genant, H. K., Wang, O., Mitlak, B. H., 2001. Effect of parathyroid hormone (1-34) on fractures and bone mineral density in postmenopausal women with osteoporosis. *N Engl J Med.* 344, 1434-41.
- Nevitt, M. C., Cummings, S. R., 1993. Type of fall and risk of hip and wrist fractures: the study of osteoporotic fractures. The Study of Osteoporotic Fractures Research Group. *Journal of the American Geriatrics Society.* 41, 1226-34.
- Nevitt, M. C., Cummings, S. R., Hudes, E. S., 1991. Risk factors for injurious falls: a prospective study. *J Gerontol.* 46, M164-70.
- Nguyen, T. V., Eisman, J. A., Kelly, P. J., Sambrook, P. N., 1996. Risk factors for osteoporotic fractures in elderly men. *Am J Epidemiol.* 144, 255-63.
- Norton, R., Butler, M., Robinson, E., Lee-Joe, T., Campbell, A. J., 2000a. Declines in physical functioning attributable to hip fracture among older people: a follow-up study of case-control participants. *Disability and Rehabilitation.* 22, 345-51.
- O'Halloran P, D., G, W. C., T, R. O. B., Kernohan, G., Orr, J., Dunlop, L., L, J. M., 2006. Factors affecting adherence to use of hip protectors amongst residents of nursing homes-A correlation study. *Int J Nurs Stud.*
- O'Halloran, P. D., Murray, L. J., Cran, G. W., Dunlop, L., Kernohan, G., Beringer, T. R., 2005. The effect of type of hip protector and resident characteristics on adherence to use of hip protectors in nursing and residential homes--an exploratory study. *Int J Nurs Stud.* 42, 387-97.
- Okuizumi, H., Harada, A., Iwata, H., Konishi, N., 1998. Effect on the femur of a new hip fracture preventive system using dropped-weight impact testing. *Journal of Bone and Mineral Research.* 13, 1940-5.
- Okumiya, K., Matsubayashi, K., Nakamura, T., Fujisawa, M., Osaki, Y., Doi, Y., Ozawa, T., 1998. The timed "up & go" test is a useful predictor of falls in community-dwelling older people. *J Am Geriatr Soc.* 46, 928-30.
- Orwoll, E., Ettinger, M., Weiss, S., Miller, P., Kendler, D., Graham, J., Adami, S., Weber, K., Lorenc, R., Pietschmann, P., Vandormael, K., Lombardi, A., 2000. Alendronate for the treatment of osteoporosis in men. *N Engl J Med.* 343, 604-10.
- Owings, T. M., Pavol, M. J., Foley, K. T., Grabiner, M. D., 2000. Measures of postural stability are not predictors of recovery from large postural disturbances in healthy older adults. *J Am Geriatr Soc.* 48, 42-50.
- Papadimitropoulos, E. A., Coyte, P. C., Josse, R. G., Greenwood, C. E., 1997. Current and projected rates of hip fracture in Canada. *Canadian Medical Association Journal.* 157, 1357-63.
- Parker, M. J., Gillespie, L. D., Gillespie, W. J., 2000. Hip protectors for preventing hip fractures in the elderly. *Cochrane Database Syst Rev.* CD001255.

- Parker, M. J., Gillespie, W. J., Gillespie, L. D., 2006. Effectiveness of hip protectors for preventing hip fractures in elderly people: systematic review. *British Medical Journal*. 332, 571-4.
- Parkkari, J., Kannus, P., Heikkila, J., Poutala, J., Heinonen, A., Sievanen, H., Vuori, I., 1997. Impact experiments of an external hip protector in young volunteers. *Calcified Tissue International*. 60, 354-7.
- Parkkari, J., Kannus, P., Heikkila, J., Poutala, J., Sievanen, H., Vuori, I., 1995. Energy-shunting external hip protector attenuates the peak femoral impact force below the theoretical fracture threshold: an in vitro biomechanical study under falling conditions of the elderly. *Journal of Bone and Mineral Research*. 10, 1437-42.
- Parkkari, J., Kannus, P., Poutala, J., Vuori, I., 1994. Force attenuation properties of various trochanteric padding materials under typical falling conditions of the elderly. *J Bone Miner Res*. 9, 1391-6.
- Pinilla, T. P., Boardman, K. C., Bouxsein, M. L., Myers, E. R., Hayes, W. C., 1996. Impact direction from a fall influences the failure load of the proximal femur as much as age-related bone loss. *Calcif Tissue Int*. 58, 231-5.
- Podsiadlo, D., Richardson, S., 1991. The timed "Up & Go": a test of basic functional mobility for frail elderly persons. *J Am Geriatr Soc*. 39, 142-8.
- Prieto, T. E., Myklebust, J. B., Hoffmann, R. G., Lovett, E. G., Myklebust, B. M., 1996. Measures of postural steadiness: differences between healthy young and elderly adults. *IEEE Trans Biomed Eng*. 43, 956-66.
- Province, M. A., Hadley, E. C., Hornbrook, M. C., Lipsitz, L. A., Miller, J. P., Mulrow, C. D., Ory, M. G., Sattin, R. W., Tinetti, M. E., Wolf, S. L., 1995. The effects of exercise on falls in elderly patients. A preplanned meta-analysis of the FICSIT Trials. *Frailty and Injuries: Cooperative Studies of Intervention Techniques*. *Jama*. 273, 1341-7.
- Redfern, M. S., Moore, P. L., Yarsky, C. M., 1997. The influence of flooring on standing balance among older persons. *Hum Factors*. 39, 445-55.
- Reginster, J. Y., Seeman, E., De Vernejoul, M. C., Adami, S., Compston, J., Phenekos, C., Devogelaer, J. P., Curiel, M. D., Sawicki, A., Goemaere, S., Sorensen, O. H., Felsenberg, D., Meunier, P. J., 2005. Strontium ranelate reduces the risk of nonvertebral fractures in postmenopausal women with osteoporosis: Treatment of Peripheral Osteoporosis (TROPOS) study. *J Clin Endocrinol Metab*. 90, 2816-22.
- Ring, C., Nayak, U. S., Isaacs, B., 1989. The effect of visual deprivation and proprioceptive change on postural sway in healthy adults. *J Am Geriatr Soc*. 37, 745-9.
- Robertson, M. C., Campbell, A. J., Gardner, M. M., Devlin, N., 2002. Preventing injuries in older people by preventing falls: a meta-analysis of individual-level data. *J Am Geriatr Soc*. 50, 905-11.
- Robinovitch, S. N., Chiu, J., 1998. Surface stiffness affects impact force during a fall on the outstretched hand. *J Orthop Res*. 16, 309-13.

- Robinovitch, S. N., Chiu, J., Sandler, R., Liu, Q., 2000a. Impact severity in self-initiated sits and falls associates with center-of-gravity excursion during descent. *Journal of Biomechanics*. 33, 863-70.
- Robinovitch, S. N., Hayes, W. C., McMahon, T. A., 1991. Prediction of femoral impact forces in falls on the hip. *Journal of Biomechanical Engineering*. 113, 366-74.
- Robinovitch, S. N., Hayes, W. C., McMahon, T. A., 1995a. Energy-shunting hip padding system attenuates femoral impact force in a simulated fall. *Journal of Biomechanical Engineering*. 117, 409-13.
- Robinovitch, S. N., Hayes, W. C., McMahon, T. A., 1997a. Distribution of contact force during impact to the hip. *Annals of Biomedical Engineering*. 25, 499-508.
- Robinovitch, S. N., Hayes, W. C., McMahon, T. A., 1997b. Predicting the impact response of a nonlinear single-degree-of-freedom shock-absorbing system from the measured step response. *Journal of Biomechanical Engineering*. 119, 221-7.
- Robinovitch, S. N., Hsiao, E. T., Sandler, R., Cortez, J., Liu, Q., Paiement, G. D., 2000b. Prevention of falls and fall-related fractures through biomechanics. *Exerc Sport Sci Rev*. 28, 74-9.
- Robinovitch, S. N., Inkster, L., Maurer, J., Warnick, B., 2003. Strategies for avoiding hip impact during sideways falls. *J Bone Miner Res*. 18, 1267-73.
- Robinovitch, S. N., McMahon, T. A., Hayes, W. C., 1995b. Force attenuation in trochanteric soft tissues during impact from a fall. *J Orthop Res*. 13, 956-62.
- Sabick, M. B., Hay, J. G., Goel, V. K., Banks, S. A., 1999. Active responses decrease impact forces at the hip and shoulder in falls to the side. *J Biomech*. 32, 993-8.
- Sandler, R., Robinovitch, S., 2001. An analysis of the effect of lower extremity strength on impact severity during a backward fall. *J Biomech Eng*. 123, 590-8.
- Sawka, A. M., Boulos, P., Beattie, K., Thabane, L., Papaioannou, A., Gafni, A., Cranney, A., Zytaruk, N., Hanley, D. A., Adachi, J. D., 2005. Do hip protectors decrease the risk of hip fracture in institutional and community-dwelling elderly? A systematic review and meta-analysis of randomized controlled trials. *Osteoporosis International*. 16, 1461-74.
- Schwartz, A. V., Kelsey, J. L., Sidney, S., Grisso, J. A., 1998. Characteristics of falls and risk of hip fracture in elderly men. *Osteoporosis International*. 8, 240-6.
- Shumway-Cook, A., Brauer, S., Woollacott, M., 2000. Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Phys Ther*. 80, 896-903.
- Simpson, A. H., Lamb, S., Roberts, P. J., Gardner, T. N., Evans, J. G., 2004. Does the type of flooring affect the risk of hip fracture? *Age Ageing*. 33, 242-6.
- Siris, E. S., Chen, Y. T., Abbott, T. A., Barrett-Connor, E., Miller, P. D., Wehren, L. E., Berger, M. L., 2004. Bone mineral density thresholds for pharmacological intervention to prevent fractures. *Arch Intern Med*. 164, 1108-12.

- Sran, M., Robinovitch, S., 2008. Preventing fall-related vertebral fractures: Effect of floor stiffness on peak impact forces during backward falls. *Spine*. Submitted.
- Stephens, J. M., Goldie, P. A., 1999. Walking speed on parquetry and carpet after stroke: effect of surface and retest reliability. *Clin Rehabil.* 13, 171-81.
- Suzuki, T., Yoshida, H., Ishizaki, T., Yukawa, H., Watanabe, S., Kumagai, S., Shinkai, S., Shibata, H., Nakamura, T., Yasumura, S., Haga, H., 1999. [Compliance in use of external protectors for hip fractures among the community elderly in Japan]. *Nippon Ronen Igakkai Zasshi.* 36, 40-4.
- Taylor, B. C., Schreiner, P. J., Stone, K. L., Fink, H. A., Cummings, S. R., Nevitt, M. C., Bowman, P. J., Ensrud, K. E., 2004. Long-term prediction of incident hip fracture risk in elderly white women: study of osteoporotic fractures. *J Am Geriatr Soc.* 52, 1479-86.
- Teasdale, N., Stelmach, G. E., Breunig, A., 1991. Postural sway characteristics of the elderly under normal and altered visual and support surface conditions. *J Gerontol.* 46, B238-44.
- Thapa, P. B., Gideon, P., Brockman, K. G., Fought, R. L., Ray, W. A., 1996. Clinical and biomechanical measures of balance as fall predictors in ambulatory nursing home residents. *J Gerontol A Biol Sci Med Sci.* 51, M239-46.
- Thompson, P., Jones, C., Dawson, A., Thomas, P., Villar, T., 2005. An in-service evaluation of hip protector use in residential homes. *Age Ageing.* 34, 52-6.
- Tinetti, M. E., Doucette, J., Claus, E., Marottoli, R., 1995. Risk factors for serious injury during falls by older persons in the community. *J Am Geriatr Soc.* 43, 1214-21.
- Tinetti, M. E., Speechley, M., Ginter, S. F., 1988. Risk factors for falls among elderly persons living in the community. *New England Journal of Medicine.* 319, 1701-7.
- Topper, A. K., Maki, B. E., Holliday, P. J., 1993. Are activity-based assessments of balance and gait in the elderly predictive of risk of falling and/or type of fall? *J Am Geriatr Soc.* 41, 479-87.
- van den Kroonenberg, A. J., Hayes, W. C., McMahon, T. A., 1995. Dynamic models for sideways falls from standing height. *J Biomech Eng.* 117, 309-18.
- van den Kroonenberg, A. J., Hayes, W. C., McMahon, T. A., 1996. Hip impact velocities and body configurations for voluntary falls from standing height. *Journal of Biomechanics.* 29, 807-11.
- van Schoor, N. M., Asma, G., Smit, J. H., Bouter, L. M., Lips, P., 2003a. The Amsterdam Hip Protector Study: compliance and determinants of compliance. *Osteoporos Int.* 14, 353-9.
- van Schoor, N. M., Deville, W. L., Bouter, L. M., Lips, P., 2002. Acceptance and compliance with external hip protectors: a systematic review of the literature. *Osteoporos Int.* 13, 917-24.

- van Schoor, N. M., Smit, J. H., Twisk, J. W., Bouter, L. M., Lips, P., 2003b. Prevention of hip fractures by external hip protectors: a randomized controlled trial. *JAMA: the journal of the American Medical Association*. 289, 1957-62.
- van Schoor, N. M., van der Veen, A. J., Schaap, L. A., Smit, T. H., Lips, P., 2006. Biomechanical comparison of hard and soft hip protectors, and the influence of soft tissue. *Bone*. 39, 401-7.
- Viano, D. C., Lau, I. V., Asbury, C., King, A. I., Begeman, P., 1989. Biomechanics of the human chest, abdomen, and pelvis in lateral impact. *Accid Anal Prev*. 21, 553-74.
- Wainwright, S. A., Marshall, L. M., Ensrud, K. E., Cauley, J. A., Black, D. M., Hillier, T. A., Hochberg, M. C., Vogt, M. T., Orwoll, E. S., 2005. Hip fracture in women without osteoporosis. *J Clin Endocrinol Metab*. 90, 2787-93.
- Warnke, A., Meyer, G., Bender, R., Muhlhauser, I., 2004. Predictors of adherence to the use of hip protectors in nursing home residents. *J Am Geriatr Soc*. 52, 340-5.
- Watts, N. B., Becker, P., 1999. Alendronate increases spine and hip bone mineral density in women with postmenopausal osteoporosis who failed to respond to intermittent cyclical etidronate. *Bone*. 24, 65-8.
- Wiener, S. L., Andersson, G. B., Nyhus, L. M., Czech, J., 2002. Force reduction by an external hip protector on the human hip after falls. *Clinical Orthopaedics and Related Research*. 157-68.
- Wiktorowicz, M. E., Goeree, R., Papaioannou, A., Adachi, J. D., Papadimitropoulos, E., 2001. Economic implications of hip fracture: health service use, institutional care and cost in Canada. *Osteoporosis International*. 12, 271-8.
- Willmott, M., 1986. The effect of a vinyl floor surface and a carpeted floor surface upon walking in elderly hospital in-patients. *Age Ageing*. 15, 119-20.
- Wolf, S. L., Barnhart, H. X., Kutner, N. G., McNeely, E., Coogler, C., Xu, T., 1996. Reducing frailty and falls in older persons: an investigation of Tai Chi and computerized balance training. Atlanta FICSIT Group. *Frailty and Injuries: Cooperative Studies of Intervention Techniques*. *J Am Geriatr Soc*. 44, 489-97.
- Wolinsky, F. D., Fitzgerald, J. F., 1994. The risk of hip fracture among noninstitutionalized older adults. *J Gerontol*. 49, S165-75.
- Wolinsky, F. D., Fitzgerald, J. F., Stump, T. E., 1997. The effect of hip fracture on mortality, hospitalization, and functional status: a prospective study. *American Journal of Public Health*. 87, 398-403.
- Zacker, C., Shea, D., 1998. An economic evaluation of energy-absorbing flooring to prevent hip fractures. *Int J Technol Assess Health Care*. 14, 446-57.
- Zuckerman, J. D., 1996. Hip fracture. *N Engl J Med*. 334, 1519-25.