Biomechanics of Falls and Hip Fractures in Older Adults

by

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Abstract

Over 90% of hip fractures are due to falls. Yet, only 1-2% of falls result in hip fracture. This suggests that there exist factors that determine injurious and non-injurious falls, but we have limited information on this area. My PhD research addresses this issue through three related studies. My first study examined age-related changes in the dynamic compressive properties (stiffness and damping) of soft tissues over the hip region. My results indicate that the soft tissues of older adults absorb 70% less energy than those of young adults, thereby requiring more energy to be absorbed in the underlying skeletal components, with corresponding increase in fracture risk. My second study determined the effect of hip abductor muscle forces and knee boundary conditions on bone stress at the femoral neck during simulated falls. My results show that physiologically feasible increases in muscle force can reduce peak compressive and tensile stresses by up to 24 and 47%, respectively. These effects are similar to the magnitude of decline in fracture strength associated with osteoporosis. Therefore, muscle contraction at impact may be as important as bone density in determining hip fracture risk during a fall. My third study analyzed the kinematics of real-life falls in older adults, as captured by surveillance cameras, to estimate velocity of the pelvis at impact -- a primary determinant of impact force and fracture risk. Results show that the pelvis impact velocity averages 2.08 m/s, which is 48% below simple free-fall predictions based on fall height ((2gh)^0.5) and 20% below average previously reported for young adults. Results also show that several mechanisms contribute to reducing the pelvis impact velocity, including hand impact and attempts to recover balance by stepping. Collectively, these findings should add important pieces to the puzzle of whether a particular fall will result in hip fracture, and informs future direction for clinical and laboratory-based research on hip fracture prevention.

Keywords: Hip fractures, real-life falls, mechanical properties of soft tissue, muscle force, knee boundary condition, impact velocity
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1. Introduction

This thesis focuses on applying the tools of biomechanics to improve our understanding on the cause and prevention of fall-related hip fractures in older adults. This introductory chapter reviews the epidemiology and biomechanics of these events, and highlights specific knowledge gaps that are addressed in subsequent chapters.

1.1. Epidemiology of fall-related hip fractures in older adults

Hip fractures are a major public health problem for older adults that often result in permanent disability or death. Approximately 20% of older adults hospitalized for a hip fracture die within a year, and about 50% suffer a major decline in independence [1, 2]. In Canada, about one out of seven women and one out of nineteen men will experience a hip fracture if they live to age 50 [3]. About 1.3 million hip fractures occurred worldwide in 1990 [4], and hip fracture risk increases dramatically with age [5]. In Canada, there are approximately 30,000 annual causes of proximal femur fracture (hip fractures) among older adults (aged 65 and older), with related medical costs of about $1.3 billion [6]. In the United States, health care costs related to hip fractures are expected to double (from $17 to $25 billion) between 2005 and 2025 [7].

1.2. Functional anatomy of the hip

The hip joint is a ball-and-socket joint, consisting of the roughly spherical femoral head (ball) which connects to the acetabulum (socket), formed by the junction between the three bones of the pelvis: the ilium, ischium, and pubis (Figure 1-1). Various ligaments span the hip joint, strengthen the joint capsule and prevent excessive range of motion. These include the iliofemoral, ischiofemoral, and pubofemoral ligaments, and the ligamentum teres. For example, the iliofemoral ligament prevents hyperextension and
excessive adduction and lateral rotation of the thigh during activities of daily living. Muscle groups spanning the hip and contributing to hip joint movement include the hip flexors, extensors, abductors and adductors. The hip flexors include the iliopsoas, rectus femoris and sartorius, with the iliopsoas being the prime mover. The hip extensors consist of the gluteus maximus, long head of the biceps femoris, semimembranosus and semitendinosus, with the gluteus maximus being the prime mover. The hip abductors include the gluteus medius, gluteus minimus and tensor fascia lata, and both gluteus medius and minimus are prime movers. The hip adductors include the adductor brevis, adductor longus, adductor magnus, pectineus and gracilis, with the adductor magnus being the prime mover.

The soft tissue layer lateral to the greater trochanter includes the gluteus medius, gluteus minimus and vastus lateralis (Figure 1-1). The gluteus medius originates from the external surface of the ilium, between the posterior and anterior gluteal lines, and inserts on the lateral and superior part of the greater trochanter. The gluteus minimus originates from the external surface of the ilium, between the anterior and inferior gluteal lines, and inserts slightly anterior to the insertion of the gluteus medius. The vastus lateralis (which acts as a knee extensor as part of the quadriceps muscle group) originates from the greater trochanter and the lateral lip of the linea aspera, and inserts on the tibial tuberosity via the patellar tendon. Of note, the portions of gluteus medius and minimus directly lateral to the greater trochanter are the muscle tendons rather than bulk muscle tissue, and these tendons are covered by a thin but strong fascia.

1.3. Biomechanics of fall-related hip fractures

1.3.1. Factor of risk for hip fracture during falls

Several case-control study and cohort studies have examined risk factors for hip fracture in the event of a fall. These have reported that the risk for hip fracture increases approximately 30-fold by impacting to the hip, 6-fold by falling sideways, and 3-fold by a 1 standard deviation decrease in bone density [8-10]. Additional factors influencing fall severity (or peak impact force) have been identified through biomechanical experiments. In general, these studies adopt an engineering concept, “factor of risk”, which defines
the risk for hip fracture during a fall based on the ratio of the peak force applied to the proximal femur during impact, divided by the fracture force of the bone under the same loading conditions [11, 12]. When the factor of risk is equal to or greater than 1, the femur is expected to fracture. In contrast, when the factor of risk is less than 1, the femur will not fracture. While an alternative approach to defining fracture risk might be based on a continuous (rather than dichotomous) scale for fracture probability, such approaches have been rare in considering the biomechanics of hip fracture risk. On the other hand, the magnitude of the factor of risk provides some measure of certainty in the predicted outcome.

In order to provide insight on the factors influencing the “factor of risk”, researchers have examined the factors that influence both the load applied to the bone during a fall, and the failure load of the proximal femur under a fall loading configuration (Figure 1-2). Of note, while over 90% of hip fractures are due to falls [13], only 1 – 2% of the falls result in hip fracture [14, 15]. This is surprising, given that the energy required to fracture the elderly proximal femur is only 5.5 J (SD = 5.0) [16], or less than 1/16th of the potential energy typically available during impact in a fall from standing height, and reflects that the vast majority of this energy is absorbed or dissipated by alternative mechanisms. Several of these factors (i.e., upper limb fall arrest, energy absorption in the lower limbs during descent, simultaneous contact to multiple body parts) are highlighted below.

1.3.2. **Determinants of failure load of the proximal femur**

Several studies have measured the fracture force of the human cadaveric femur in a simulated fall loading configuration [11, 16-29]. Robinovitch et al. recently summarized and reported value of fracture force across 15 studies, which averaged 2,827 N for women and 4,375 N for men [30]. Collectively, these indicate that the fracture force is affected by bone geometry, bone mineral density (BMD), loading configuration, and loading rate. Courtney et al. reported that fracture force was strongly correlated with BMD of the femoral neck ($r^2 = 0.92$), and that the mean fracture force was 52% lower for older than young adults (3,440 N versus 7,200 N) [31]. Similarly, Lochmueller et al. reported approximately 30% lower values of fracture force for proximal femora from older females than males (3,070 N versus 4,230 N) [24]. Courtney et al.
also showed that increases in loading rate (from 2 to 100 mm/s) caused 16% increases in the average fracture force (from 5300 to 6150 N) [18]. Pinilla et al. reported that the fracture force decreased by 24% as the loading angle moved 30 deg posterior to the femoral plane [19]. Manske et al. reported that the cortical cross-sectional area at the femoral neck measured from magnetic resonance imaging (MRI) was associated with fracture force \( r^2 = 0.46 \) [27]. Other investigators have shown that femoral strength is influenced by trabecular bone area, cortical thickness, femoral neck axis length, neck-shaft angle, femoral shaft diameter, and femoral head and neck diameter [24, 28, 29]. Finally, a finite element modeling study found that the estimated strength of the proximal femur was 292% greater under simulated single-limb stance than fall loading (3760 versus 959 N) [32].

1.3.3. **Determinants of applied load during a fall**

Researchers have also investigated the factors that affect the force applied to the hip region during a fall, based on safe falling (“pelvis release”) experiments with young adults [33, 34]. No studies have acquired such measurements in older adults. These studies support the notion that the impact response of the body during a fall on the hip can be simulated accurately by a single-degree-of-freedom mass-spring model [33]. During a sideways fall, this model predicts that the peak force applied to the hip region is

\[
F = v \sqrt{mk},
\]

where \( m \) is the effective mass, \( k \) is the effective stiffness, and \( v \) is the impact velocity [30]. Robinovitch et al. conducted “pelvis release experiments” (low height falls on the hip) with young adults, and found that the effective mass averaged about 50% of body mass, and the effective stiffness averaged 39 kN/m (SD = 16) [34].

No study (to my knowledge) has examined how contraction of the muscles spanning the hip affects peak forces and stresses in the proximal femur during a fall. However, Robinovitch et al. found that when landing in a muscle-active versus muscle-relaxed state (simulating core muscle contraction to maintain the trunk and head raised during impact), the effective mass increased 26%, the effective stiffness increased 86%, and predicted impact forces were 66% higher, suggesting that falling with muscles relaxed is safer [33].
With regard to impact velocity, researchers have used motion capture or video recording to measure pelvis impact velocities during lab-based falling experiment with young adults. Over six studies, vertical velocity at impact averaged 2.58 m/s (SD = 0.53) [35-40]. Feldman and Robinovitch found that pelvis impact velocity decreased about 20% by impacting the hands more than 50 ms prior to the pelvis, and by executing a step during decent. However, no studies have reported pelvis impact velocities during falls in older adults.

1.4. Current gaps in understanding the cause of fall-related hip fractures

1.4.1. Age-related changes in mechanical properties of soft tissue over the hip

Risk for hip fracture increases dramatically with age [5], due in part to age-related declines in bone density and strength, and an increased incidences of falls. An additional contributor to this trend may be age-related changes in the mechanical properties and thickness of soft tissues over the hip region, which act as a natural “shock absorber” for attenuating and distributing impact forces applied to the bone [41, 42]. Supporting this notion, studies have reported that older adults with high body mass index (who tend to possess greater tissue thickness over the hip [42, 43]) have lower risk for hip fracture risk in the event of a fall [44-46], and each standard deviation decrease in soft tissue thickness over the hip caused a 1.8-fold increase in the risk of hip fracture [11].

Researchers have measured the force-attenuating behaviour of cadaveric trochanteric soft tissues (skin, fat, and muscle) during simulated sideways falls [47]. On average, these tissues (of mean thickness 24 mm) attenuated peak femoral impact force by 13% and absorbed 34 J of energy of the total impact energy of 140 J. Furthermore, for each 1 mm increase in tissue thickness, peak force decreased by approximately 71 N and tissue energy absorption increased by 1.7 J. In support of these trends, Lauritzen and Askegaard reported that a 9 mm difference in porcine cadaveric soft tissue thickness produced 60% greater tissue energy absorption during impact [48].
Studies have also examined the compressive properties of trochanteric soft tissues from living humans (which may differ substantially from cadavers, due to post-mortem changes in cell and tissue integrity [49]). Robinovitch et al. used a hand-held indentation device to measure the quasi-static stiffness of soft tissue over the hip region in young women of mean age 24 yrs [50]. They found that tissue stiffness (at a compressive force of 60 N) varied across different pelvis locations, being stiffest directly over the greater trochanter (GT) (at 35 kN/m (SD = 14)) and least stiff at a location 6 cm anterior to the GT (at 9 kN/m (SD = 3)). Laing and Robinovitch used a similar technique with older women of mean age 77 yrs [51], and found that tissue stiffness (at a compressive force of 140 N) was again greatest directly over the greater trochanter (at 34 kN/m (SD = 16)), and lowest at a location 6 cm posterior to the trochanter (at 14 kN/m (SD = 7)). Comparison between these studies is challenging, due to their use of different force magnitudes, and the fact that the load was applied by hand, and thus the rate of loading was not accurately controlled.

Researchers have examined age-related differences in soft tissue compression at sites other than the hip, which appear to reflect location-dependent changes with age in the dynamic compressive behavior of soft tissues. Boyer et al. used a dynamic indentation device to estimate the stiffness and damping of the skin over the anterior forearm (6 cm proximal to the elbow) of young and older adults [52]. They found that, when compared to older adults, the forearm skin of young adults had 50% greater stiffness (42.5 versus 28.4 N/m) and 17% greater damping constant (0.074 versus 0.062 N s/m). However, this study involved very small levels of baseline forces (0.007 N) and tissue deformation (200 µm). In contrast, both Hsu et al. and Kwan et al. reported that the elastic modulus of plantar soft tissue over the big toe and heel was up to 287% greater (122.9 versus 31.7 kPa) in older than young adults [53, 54].

The study described in Chapter 2 extends these findings, to examine age-related changes in the stiffness and damping of soft tissue over the hip in young and older women (in a sideways landing configuration), through dynamic compression. I also examined how these parameters associated with soft tissue thickness, as measured with ultrasound. I hypothesized that tissue stiffness and damping over the hip would be lower in older than in younger women (as observed for the forearm by Boyer et al. (2009)), and would decrease with increasing tissue thickness.
1.4.2. **Effect of muscle force spanning the hip joint and knee boundary condition on the risk of hip fracture**

As discussed above, over 90% of hip fractures in older adults are due to falls [13]. However, only 1 – 2% of falls in older adults result in hip fracture [14, 15]. Improved understanding of the factors that separate injurious versus non-injurious falls should lead to improvement in the prevention of hip fractures in older adults.

In Chapter 3, I examined how a largely unexplored factor - the forces in the muscles spanning the hip joint – affects the forces and stresses in the proximal femur during impact from a fall. I was specifically interested in understanding how peak stresses in the bone are influenced by hip abductor muscle forces, which are estimated to routinely reach 2 - 3 times body weight during daily activities [55, 56]. Based on a simple free body diagram analysis (see Figures 3-1b and 3-3a), I considered that the hip abductor muscle force will generate a bending moment that counteracts the moment created by the impact force, and thereby contribute to a decrease in peak stresses at the femoral neck. However, I also considered the possibility that increases in muscle force may increase the effective stiffness (and mass) of the body, and cause a subsequent increase in total applied force applied, as observed in simulated falls with young adults [33].

I was also interested in examining how knee boundary conditions affect peak bone stresses during a fall. Based on similar free body considerations (Figure 3-3a), I expected that impacting the hip with the knee constrained (as opposed to free in the air) would result in sharing of the total applied load between the knee and hip, and a corresponding decrease in peak bone stress at the femoral neck.

To address these issues, I developed an advanced mechanical device that simulated impact to the hip during a sideways fall. The device allowed me to alter the magnitude of hip abductor muscle forces, and knee boundary conditions, and determine corresponding changes in peak forces and stresses at the femoral neck. I then conducted experiments to test the hypothesis that peak forces and stresses at the femoral neck during impact decrease with (a) increased muscle force, and (b) impacting to the hip with the knee constrained as opposed to free.
1.4.3. Impact velocity and duration of fall in older adults

Fall kinematics can provide indices of impact severity. Hip fracture risk during a fall depends largely on the forces applied to the hip region during impact. These forces depend, in turn, on the velocities and configurations of the body segments at landing. Second, fall kinematics provide information on the time duration of falls, which should be relevant to the faller’s ability to initiate and execute protective responses for safe landing. Finally, fall kinematics provide evidence of the occurrence of such responses, such as attempts to recover balance by stepping or grasping, or to arrest the fall with the outstretched arm [35, 38, 57], and how these might influence impact severity.

Unfortunately, we have little understanding of the kinematics of actual falls in older adults. Our current knowledge is limited to laboratory studies with young adults falling on gym mats [35-40]. Collectively, these suggest that the vertical velocity of the pelvis at impact averages 2.58 m/s (SD = 0.53), and that the time interval between fall initiation to pelvis impact averages 867 ms (SD = 237) [36, 38, 40]. They also suggest that, in landing from a fall, young adults tend to impact the ground with one or both outstretched hands, regardless of fall direction, most often just prior to (but within 50 ms of) impact to the pelvis [36], and that this reduces pelvis impact velocity [40].

However, lab-based falling experiments with young adults may not reflect the range of falling patterns (and impact velocities) occurring in real-life falls among older adults, due to differences in the situational and environmental context of falls, or intrinsic factors such as physical and cognitive function, medication use, and disease diagnoses.

Recently, our research team reported results on the causes and activities associated with falls from analysis of 227 real-life falls captured on video cameras in two long term care facilities [58, 59]. As described in Chapter 4, I conducted kinematic analysis on a subset of fall videos, to estimate pelvis impact velocities, contact times, and body configurations at landing.
1.5. Summary of thesis goals

While considerable efforts have been made to understand the biomechanics of hip fractures in older adults, important gaps exist in our understanding of these events. This PhD thesis addresses three important but unexplored issues concerning the factors that separate injurious and non-injurious falls (Figure 1-2):

1. I conducted experiments with human subjects to measure the stiffness and damping of soft tissue over the greater trochanter, and determine whether these parameters are influenced by age and soft tissue thickness. These results are described in Chapter 2.

2. I conducted mechanical tests with a hip impact simulator to examine whether forces and stresses at the femoral neck were affected by hip abductor muscle force and knee boundary conditions. These results are described in Chapter 3.

3. I conducted kinematic analysis of real-life falls in older adults captured on video, to determine impact velocities and times to impact of the pelvis. These results are described in Chapter 4.

Collectively, these studies add important pieces to the complex puzzle of fall-related hip fractures in older adults. They should be useful for researchers and engineers in understanding the factors that determine whether a specific fall will result in hip fracture, and designing and selecting interventions for hip fracture prevention (such as exercise programs, wearable hip protectors and compliant flooring). They should also help to inform clinicians and health administrators in their approaches to the treatment and prevention of falls and fractures.
The hip joint is a ball-and-socket joint, where the head of femur connects to the acetabulum. Various muscles span the hip joint and contribute to the hip joint movement. GD = gluteus medius, GN = gluteus minimus, AM = adductor magnus, AB = adductor brevis, VL = vastus lateralis, OE = obturator externus, OI = obturator internus, TI = tendon of iliopsoas, GT = greater trochanter, LT = lesser trochanter, S = skin and fat. Accessed from https://www.primalpictures.com on September 26, 2013.
This PhD thesis addresses key knowledge gaps in the model. In particular, Chapter 2 describes the results of experiments to examine age-related changes in the mechanical "shock absorbing" properties of soft tissue over the hip. Chapter 3 examines how femoral neck stresses are affected by muscle forces and knee boundary conditions during falls with a mechanical hip impact simulator. Chapter 4 describes kinematic analysis to determine impact velocities during real-life falls in older adults, captured on video.
2. Age-related changes in dynamic compressive properties of soft tissues over the hip

2.1. Abstract

Hip fracture risk increases dramatically with age, and 90% of fractures are due to falls. During a fall on the hip, the soft tissues overlying the hip region (skin, fat, and muscle) act as a shock absorber to absorb energy and reduce the peak force applied to the bone. I conducted dynamic indentation experiments with young women (aged 19-35; n = 17) and older women (over age 65; n = 17) to test the hypothesis that changes occur with age in the stiffness and damping properties of these tissues.

Tissue stiffness and damping was derived from experiments where subjects lay sideways on a bed with the greater trochanter contacting an 3.8 cm diameter indenter, which applied sinusoidal compression between 5 to 30 Hz with a peak-to-peak amplitude of 1 mm. Soft tissue thickness was measured using ultrasound.

On average, stiffness was 2.9-fold smaller in older than young women (5.7 versus 16.8 kN/m, p = 0.0005) and damping was 3.5-fold smaller in older than young women (81 versus 282 N s/m, p = 0.001). Neither parameter associated with soft tissue thickness.

Our results indicate substantial age-related reductions in the stiffness and damping of soft tissues over the hip region, which reduce their capacity to absorb and dissipate energy before bottoming out during a fall. Strategies such as wearable hip protectors or compliant flooring may compensate for age-related reductions in the shock-absorbing properties of soft tissues and decrease the injury potential of falls.
2.2. Introduction

Hip fractures are a major cause of death and disability in older adults, and over 90% of cases are caused by falls [1, 2, 13]. Risk for hip fracture increases exponentially with age [5], due in part to age-related declines in bone density and strength, and an increased incidences of falls. An additional contributor to this trend may be age-related changes in the mechanical properties and thickness of soft tissues over the hip region, which act as a natural “shock absorber” for attenuating and distributing impact forces applied to the bone [41, 42]. Supporting this notion, clinical studies have reported that older adults with high body mass index (who tend to possess greater tissue thickness over the hip [42, 43]) have lower risk for hip fracture risk in the event of a fall [44-46].

Researchers have measured the force-attenuating behaviour of cadaveric trochanteric soft tissues (skin, fat, and muscle) during simulated sideways falls [47]. On average, these tissues (of mean thickness 24 mm) attenuated peak femoral impact force by 13% and absorbed 34 J of energy of the total impact energy of 140 J. Furthermore, for each 1 mm increase in tissue thickness, peak force decreased by approximately 71 N and tissue energy absorption increased by 1.7 J. In support of these trends, Lauritzen and Askegaard reported that a 9 mm difference in porcine cadaveric soft tissue thickness produced 60% greater tissue energy absorption during impact [48].

Studies have also examined the compressive properties of trochanteric soft tissues from living humans (which may differ substantially from cadavers, due to post-mortem changes in cell and tissue integrity [49]). Robinovitch et al. used a hand-held indentation device to measure the quasi-static stiffness of soft tissue over the hip region in young women of mean age 24 yrs [50]. They found that tissue stiffness (at a compressive force of 60 N) varied across different pelvis locations, being stiffest directly over the greater trochanter (GT) (at 35 kN/m (SD = 14)) and least stiff at a location 6 cm anterior to the GT (at 9 kN/m (SD = 3)). Laing and Robinovitch used a similar technique with older women of mean age 77 yrs [51], and found that tissue stiffness (at a compressive force of 140 N) was again greatest directly over the greater trochanter (at 34 kN/m (SD = 16)), and lowest at a location 6 cm posterior to the trochanter (at 14 kN/m (SD = 7)). Comparison between these studies is challenging, due to their use of different
force magnitudes, and the fact that the load was applied by hand, and thus the rate of loading was not accurately controlled.

Researchers have examined age-related differences in soft tissue compression at sites other than the hip, which appear to reflect location-dependent changes with age in the dynamic compressive behavior of soft tissues. Boyer et al. used a dynamic indentation device to estimate the stiffness and damping of the skin over the anterior forearm (6 cm proximal to the elbow) of young and older adults [52]. They found that, when compared to older adults, the forearm skin of young adults had 50% greater stiffness (42.5 versus 28.4 N/m) and 17% greater damping constant (0.074 versus 0.062 N s/m). However, this study involved very small baseline forces (0.007 N) and tissue deformation (200 \( \mu \)m). In contrast, both Hsu et al. and Kwan et al. reported that the elastic modulus of plantar soft tissue over the big toe and heel was up to 287% greater (122.9 versus 31.7 kPa) in older than young adults [53, 54].

Against this background, I conducted experiments involving dynamic compression to quantify the stiffness and damping of soft tissue over the hip in young and older women (in a sideways landing configuration). I also examined how these parameters associated with soft tissue thickness, as measured with ultrasound. I hypothesized that tissue stiffness and damping over the hip would be lower in older than in younger women (as observed for the forearm by Boyer et al. (2009)), and would decrease with increasing tissue thickness.

2.3. Methods

2.3.1. Subjects

Participants included 17 young women (of mean age 21.2 (SD = 2.7), height 1.65 m (SD = 0.07), body mass 57.1 kg (SD = 8.8), and body mass index (BMI) 21.1 (SD = 2.8)) and 17 older women (of mean age 69.9 (SD = 4.7), height 1.56 m (SD = 0.07), body mass 62.4 kg (SD = 12.0) and BMI 25.9 (SD = 5.7)). The experimental protocol was approved by the Committee on Research Ethics at Simon Fraser University, and all participants provided written informed consent.
2.3.2. **Protocol**

In the first session, participants lay sideways on a bed having a cut-out at the hip region (Figure 2-1a). A mechanical indenter, having a 3.8 cm diameter circular end plate, contacted the skin over the greater trochanter (GT). The indenter was vibrated sinusoidally by a mechanical shaker (LDS shaker V408, Bruel & Kjaer, Royston, UK), with a peak-to-peak amplitude of 1 mm and frequency sweeping from 5 to 30 Hz at a rate of 1 Hz/s. In all trials, the baseline level of tissue compressive force was 40 N. Time-varying force was measured from a load cell (MLP100, Transducer Techniques, Temecula, CA, USA) and displacement was measured from a linear position transducer (MLT 38000102, Honeywell, Freeport, IL, USA) mounted within the indenter, with sampling rate of 1 kHz. I validated the accuracy of the experimental setup by measuring the force-deflection behaviour of a steel spring, and confirming that the error in calculated stiffness was less than 1% (of the known value of 4.4 kN/m) over a range of frequencies between 2 - 15 Hz.

In the second session, I used B mode ultrasound imaging (SonixRP, Ultrasonix, Richmond, BC, Canada) with a 60 mm linear array probe (model L14-5W/60, Ultrasonix) to measure the thickness of skin, fat, and muscle layers over the GT. Using a technique trained by an experienced sonographer [41, 42], the midpoint of the GT was marked with water soluble ink, where (following initial exploratory measures of the bone surface) the mid-point of ultrasound probe was placed. Care was taken to apply the minimal pressure that allowed full contact between the skin and the ultrasound probe. Using a calibrated cursor, measures were then acquired from screen captures (Figure 2-2) of the thickness of skin (which appeared as a low-intensity white layer), fat (which appeared as a black layer), muscle and fascia (which appeared as a low-intensity white layer), and bone (which appeared as a high-intensity white layer).

2.3.3. **Data Analysis**

To derive tissue stiffness and damping, I fit the force data in the frequency domain (Figure 2-3c) with a single degree of freedom mass-spring-damper model with base excitation [60] (Figure 2-1b). In selecting this model, I considered that previous research has shown that the dynamic response of the body during impact to the hip (in a
simulated sideways fall) is well described by a single degree of freedom, mass-spring-damper model [33, 34]. I also considered (from inspection) that our force traces consistently displayed a local peak between 6 and 9 Hz (Figure 2-3a). This agrees with Robinovitch et al. (1997), who found that the natural frequency of the body during a sideways fall on the hip ranged from 5.7 - 6.2 Hz. Since tissue properties surrounding this resonant frequency reflect their protective value during falls, I focused our analysis over a frequency window extending from 5 Hz (the lowest frequency captured) to 7 Hz beyond the first peak (resonant frequency).

Model parameters were derived from a customized Matlab routine (Version 7.5, MathWorks Inc., Natick, MA, USA) incorporating the least-squares nonlinear curve-fitting routine *lsqcurvefit* to identify, for each trial, values of effective mass ($m$), damping ratio ($\zeta$) and natural frequency ($\omega_n$) that provided a best fit between model and experimental variations in force as a function of the driving frequency ($\omega$), given by:

$$F = m\omega_n^2Yr^2 \frac{1+(2\zeta r)^2}{\sqrt{(1-r^2)^2+(2\zeta r)^2}}$$

(Equation 1)

where $Y$ = amplitude of base oscillation, and $r = \omega/\omega_n$ (see Figure 2-1b and Figure 2-3). To acquire accurate parameter values, I set initial values and upper and lower limits for each parameter based on estimates provided in the previous studies [33, 34]. I also set “90000” for the maximum number of iterations and function evaluations, and “1e-20” and “1e-9” for the termination tolerance on the function value and on $X$, respectively. I confirmed that parameter values from the curve fit optimization were repeatable even when making changes to initial guess within our lower and upper bounds. Acquired parameter values were used to calculate corresponding values of soft tissue stiffness ($k = m\omega_n^2$) and damping ($b = 2\zeta\sqrt{km}$).

For statistical analysis, I used ANOVA to test whether stiffness and damping constants associated with age (as a fixed factor) and tissue thickness (as a covariate). I also used $t$-tests to compare tissue thickness between young and older adults. All analyses were conducted with a significance level of $\alpha = 0.05$ with SPSS 18.0.
2.4. Results

Tissue stiffness was affected by age ($F = 17.0$, $p = 0.0003$; Figure 2-4). The average value of stiffness in young women was 2.9-fold larger than in older women (16.8 (SD = 9.1) versus 5.7 (SD = 5.5) kN/m). However, stiffness did not associate with total soft tissue thickness ($F = 0.4$, $p = 0.5$; Figure 2-5). Nor did it associate with skin thickness, fat thickness or muscle thickness, when total tissue thickness was replaced with each of these variables as a covariate in the ANOVA model.

Similar trends were observed for tissue damping. In particular, damping was also influenced by age ($F = 12.5$, $p = 0.001$; Figure 2-4). On average, damping was 3.5-fold larger in young than older women (282 (SD = 197) versus 81 (SD = 107) N s/m). As with stiffness, again, similar trends were observed when total tissue thickness was replaced with skin, fat or muscle thickness.

Total soft tissue thickness did not differ between young and older adults (32.1 (SD = 7.2) mm in young and 30.4 (SD = 14.9) mm in old; $t = 0.4$, $p = 0.7$; Figure 2-2b). Similarly, there were no differences between young and older adults in skin thickness (1.2 (SD = 0.3) mm in young and 1.4 (SD = 0.5) mm in old; $t = 1.0$, $p = 0.3$), fat (3.5 (SD = 2.1) mm in young and 2.1 (SD = 2.0) mm in old; $t = 1.84$, $p = 0.08$) or muscle thickness (27.2 (SD = 6.0) mm in young and 26.7 (SD = 14.0) mm in old; $t = 0.13$, $p = 0.9$).

2.5. Discussion

I conducted dynamic indentation experiments to quantify the stiffness and damping of soft tissues over the greater trochanter in young and older adults. I found that, on average, compressive stiffness was 2.9-fold greater, and damping was 3.5-fold greater in young than older women.

I expected that age-related differences in tissue stiffness and damping would relate, in part, to differences in tissue thickness. However, the effect of age on tissue properties could not be explained by tissue thickness, which did not differ between young and older adults. Instead, as discussed below, these differences appeared to
arise from age-related changes at the tissue level in the material properties of skin, fat, and muscle.

Our findings of reductions in tissue stiffness and damping among older women agree with Boyer et al (2009), who reported that the dynamic compressive stiffness of skin over the forearm was 50% greater, and damping was 17% greater, in young (aged 18 – 30 yrs) than older women (aged 51 – 70 yrs). On the other hand, our results contrast with both Hsu et al. (2005) and Kwan et al. (2010), who reported age-related stiffening of plantar soft tissues. This likely reflects that aging affects soft tissues mechanical properties in a site-dependant manner [61], due to underlying differences in biological, functional, and environmental (e.g., sun exposure) factors.

When compared to our values, Laing and Robinovitch (2008) reported higher values of average indentation stiffness over the greater trochanter in older women (34 versus 5.7 kN/m). Similarly, Robinovitch et al. (1995a) reported higher average values of compressive stiffness for young women (35 kN/m) than I observed (17 kN/m). These discrepancies may relate to these previous studies using quasi-static loading conditions, and higher force magnitudes (140-160 N in Laing and Robinovitch (2008) and 60 N in Robinovitch et al. (1995a)) than the 40 N used in our study.

A variety of material-level changes in skin, fat, and muscle may have contributed to the approximate 3-fold decline I observed among older women in compressive stiffness and damping. Researchers have shown that declines occur with age in the content of elastin and collagen in skin [62, 63], along with enlargening of collagen fibrils. These changes are associated with changes in keratinocyte protein production and tissue degradation [64]. Declines also occur in the thickness of the subcutaneous fat layer (for which I observed borderline reductions among older adults) [65]. These factors contribute to the increased wrinkling and sagging of skin with age, and reduced ability of skin to absorb energy during impact [66]. Studies have also shown that the compressive stiffness of muscle (in both the relaxed and active state) is considerably reduced in older adults [67], and that older adults have reduced muscle tone [65].

In order to estimate how the observed age-related reductions in trochanteric soft tissue stiffness and damping might affect fracture risk during a fall, it is useful to consider
a fall as an “energy management problem.” The trochanteric soft tissue layer can be
modeled conceptually as a parallel spring-damper arrangement, in series with the
underlying proximal femur and pelvis. The energy absorbed and dissipated by the soft
tissues reduces the remaining energy that must be absorbed by the underlying bone (or
through alternative mechanisms). Researchers have previously reported that the total
kinetic energy of the body at the moment of impact from a fall ranges between 70 and
300 J [39]. Furthermore, in experiments that simulate sideways falls on the hip involving
an impact energy of 140 J, an average of about 35 J (or 25% of the total impact energy)
was absorbed in the trochanteric soft tissues [47].

Now, I found that older women had 3-fold lower soft tissue stiffness and damping
than young adults (in the low-force regime), while having similar tissue thickness. To
estimate how this might influence the energy absorbed and dissipated by the soft
tissues, I need to introduce three assumptions to the model. First, I assume that tissue
stiffness and damping is constant at low and moderate levels of deformation, and equal
to the average values observed in our experiments. Second, I assume that, at high
magnitudes of deformation (as occurring during a severe fall), the tissues “bottom out”
(i.e., increase dramatically in stiffness), beyond which they provide a negligible
contribution to energy absorption and force reduction [68]. Third, I assume similar rates
of tissue deformation among young and older women. Under this scenario, the observed
3-fold reductions in stiffness and damping will translate directly to a 3.4-fold reduction in
tissue energy absorption in older women (14.4 versus 4.2 J) (Figure 2-6). This suggests
that, along with age-related declines in the fracture strength of bone [18], and increases
in the frequency of falls, age-related reductions in soft tissue stiffness and damping may
contribute substantially to the dramatic rise with age in the frequency of fall-related hip
fractures.

In summary, our results document 3-fold reductions in the stiffness and damping
of trochanteric soft tissues in older adults, which may increase fall severity and hip
fracture risk. The effect of age on stiffness and damping does not arise from differences
in soft tissue thickness over the hip region, but instead from changes in tissue-level
properties. The decrease with age in the shock-absorbing capacity of soft tissues
supports the value of wearable “hip protectors” or compliant flooring for high-risk
populations.
Figure 2-1.  **Experimental setup and base excitation model.**

(a) Participants lay on a bed, having a cut-out at the pelvis, through which an indenter contacted the skin over the greater trochanter, with a baseline force of 40 N. A mechanical shaker applied 1 mm amplitude of vibration to the skin surface. (b) Single degree of freedom mass-spring-damper model with base excitation (Inman, 2008).

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Figure 2-2. **Soft tissue thickness over the greater trochanter.**
(a) A sample ultrasound image acquired with a 60 mm linear array probe. (b) Average values of skin, fat, muscle, and total tissue thickness in young and older adults (error bars show one standard deviation). There were no statistical differences in tissue thickness between young and older adults.

Figure 2-3. Model fitting of experimental force data.

(a) The displacement (Y) was defined as the average experimental value. (b) The measured variation in force over a window extending from 5 Hz (the lowest
measured frequency) to 7 Hz after the first peak was fit with a mass-spring-damper base excitation model. (c) Sample agreement between experimental and model force traces.

![Sample agreement between experimental and model force traces.](image)

**Figure 2-4. Average values of stiffness and damping**

Average values of stiffness (left) and damping (right) over the greater trochanter (error bars show one standard deviation). Stiffness and damping were 2.9 and 3.5-fold lower, respectively, in older than young women.

![Average values of stiffness and damping](image)

**Figure 2-5. Association between tissue thickness and properties**

![Association between tissue thickness and properties](image)
Relationship between (a) stiffness and soft tissue thickness and (b) damping and soft tissue thickness. Neither stiffness nor damping associated with tissue thickness.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Young</th>
<th>Old</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness (N/m)</td>
<td>16800</td>
<td>5700</td>
</tr>
<tr>
<td>Damping (N s/m)</td>
<td>282</td>
<td>81</td>
</tr>
<tr>
<td>Natural frequency (Hz)</td>
<td>6.4</td>
<td>7.1</td>
</tr>
<tr>
<td>Total tissue thickness over GT (m)</td>
<td>0.032</td>
<td>0.030</td>
</tr>
<tr>
<td>Energy absorbed by tissue stiffness (J)</td>
<td>8.6</td>
<td>2.6</td>
</tr>
<tr>
<td>Energy absorbed by tissue damping (J)</td>
<td>5.8</td>
<td>1.6</td>
</tr>
<tr>
<td>Total energy absorbed (J)</td>
<td>14.4</td>
<td>4.2</td>
</tr>
</tbody>
</table>

Energy absorbed by tissue stiffness ($E_k$)  
$\approx (1/2)kX^2$

Energy absorbed by tissue damping ($E_b$)  
$\approx \int b\dot{x}\,dx = (1/2)\omega bX^2$

**Figure 2-6. Shock-absorbing properties of soft tissue over the greater trochanter in young and older women.**

(a) A conceptual model showing tissue “bottom out” during impact, and expected energy absorption by soft tissue in young and older adults. The energy absorption provided by soft tissue in young and older adults was calculated based on average value of parameters (b) and equations (c).
3. Effect of hip abductor muscle forces and knee boundary conditions on femoral neck stresses during simulated falls

3.1. Abstract

Over 90% of hip fractures are due to falls. Yet, most falls do not result in hip fracture. An important but understudied question concerns the role of muscle forces spanning the hip in affecting fracture risk during falls. Knee boundary conditions during impact from a fall may also affect risk for hip fracture. I simulated sideways falls with a mechanical hip impact simulator, and measured the effect of these variables on peak forces and estimated stresses at the femoral neck.

Peak values of compressive and tensile stresses, shear force, bending moment, and axial force each associated with hip abductor muscle forces and knee boundary conditions (p < 0.0005). When the muscle force increased from 400 to 1200 N, peak compressive and tensile stresses decreased 24% and 47%, respectively. This was due to the tension-band effect of the muscle in reducing the bending moment by 36%. Furthermore, peak compressive and tensile stresses averaged 40% and 47% lower, respectively, in a “free knee” than “fixed knee” condition.

The results indicate that contraction of the hip abductor muscles at the moment of impact during a fall, and landing with the knee free of constraints, substantially reduced peak compressive and tensile stresses at the femoral neck. These findings should contribute to improved approaches for hip fracture risk assessment and prevention.
3.2. Introduction

There are approximately 30,000 annual causes of proximal femur fracture (hip fractures) among older adults (aged 65 and older) in Canada, with related medical costs of about $1.3 billion [6]. About 50% of hip fracture patients will not regain their previous level of independence, and 25% will die in one year [1, 2]. Over 90% of hip fractures in older adults are due to falls [13]. However, only 1 – 2% of falls in older adults result in hip fracture [14, 15]. Improved understating of the factors that separate injurious versus non-injurious falls should lead to improvements in the prevention of hip fractures in older adults.

In general, the “factor of risk” for hip fracture during a fall depends on the ratio of the peak force (or stress) applied to the proximal femur during impact, divided by the force (or stress) that will fracture the bone under similar loading conditions [11]. Previous studies have shown that the force required to fracture the cadaveric proximal femur depends on bone density, bone geometry and loading conditions [11, 20, 27, 29]. Studies have also shown that the magnitude of impact force applied to the hip during a fall depends on the velocity and orientation of the pelvis at impact, and on the stiffness of the soft tissue layer over the bone [41, 42, 51].

In the current study, I examined how a largely unexplored factor - the forces in the muscles spanning the hip joint – affects the forces and stresses in the proximal femur during impact from a fall. I was specifically interested in understanding how peak stresses in the bone are influenced by hip abductor muscle forces, which are estimated to routinely reach 2 - 3 times body weight during daily activities [55, 56]. Based on a simple free body diagram analysis (Figures 3-1b and 3-3a), I considered that the hip abductor muscle force will generate a bending moment that counteracts the moment created by the impact force, and thereby contribute to a decrease in peak stresses at the femoral neck. However, I also considered the possibility that increases in muscle force may increase the effective stiffness (and mass) of the body, and cause a subsequent increase in total applied force applied, as observed in simulated falls with young adults [33].
I was also interested in examining how knee boundary conditions affect peak bone stresses during a fall. Based on similar free body considerations (Figure 3-3a), I expected that impacting the hip with the knee constrained (as opposed to free in the air) would result in sharing of the total applied load between the knee and hip, and a corresponding decreases in peak bone stress at the femoral neck.

To address these issues, I developed an advanced mechanical device that simulated impact to the hip during a sideways fall. The device allowed us to alter the magnitude of hip abductor muscle forces, and knee boundary conditions, and determine corresponding changes in peak forces and stresses at the femoral neck. I then conducted experiments to test the hypothesis that peak forces and stresses at the femoral neck during impact decrease with (a) increased muscle force, and (b) impacting to the hip with the knee constrained as opposed to free.

3.3. Methods

3.3.1. SFU Hip impact simulator

Modifications were made to an existing mechanical hip impact simulator, consisting of a surrogate pelvis and impact pendulum [51], to allow for measurement of 3D forces at the femoral neck, and manipulation of hip abductor muscle forces and supporting conditions at the knee (Figure 3-2). Skeletal elements in the surrogate pelvis were simulated with a synthetic femur and hemi-pelvis (models 3405 and 3406, respectively, Sawbones, Vashon, WA, USA) made of solid fiberglass epoxy without a hollow intramedullary canal. Forces in the hip abductor (gluteus medius and minimus) and hip adductor (adductor magnus) muscles were simulated using 3/32” diameter wire ropes (# 3461T45, McMaster-Carr, Aurora, OH, USA). Each wire attached at one end via a rigid anchor to the anatomical insertion point on the femur, passed through the equivalent muscle origin on the pelvis, and attached to a coil spring (of stiffness 150 kN/m, #9584K82, McMaster-Carr). The muscle force was controlled by adjusting nuts at the end of the wire rope to compress the spring by 2.7 mm (400 N), 5.4 mm (800 N), or 8.1 mm (1200 N), monitored by digital calipers. Polyethylene and polyurethane foams were used to simulate the pelvic soft tissues (skin, fat and muscle). The compressive
indentation stiffness of the tissues over the greater trochanter (GT) was 22.3 kN/m (SD = 1.9), within one standard deviation of the average value for older women (34.4 kN/m (SD = 15.5)) measured by Laing and Robinovitch (2008). The total effective mass of the system (with respect to the hip impact site) was 35 kg.

3.3.2. Sideways fall simulation

I used the system to simulate sideways falls involving initial forces of 400, 800, or 1200 N in each of the two hip abductor muscles. Given that the angle between the gluteus medius and minimus was approximately 30 degrees, the vector sum of the two forces was about 770, 1540, or 2320 N (or 1 - 3 times body weight for a typical individual). All trials involved a force of 80 N in the adductor muscle. I also conducted trials with three types of knee boundary conditions, including the distal femur either free in space, secured to the pendulum via a wooden block, or secured to the pendulum via a steel spring (of stiffness 13.3 kN/m and resting length 11.1 cm). To prevent component damage, all trials involved an impact velocity of 2 m/s, which has been regarded as a “moderate” fall severity, just below one standard deviation of the average value of 3.01 m/s (SD = 0.83) measured for falls from standing in young adults [40]. The impact velocity was monitored from a rotary variable inductance transducer (RVIT 15-120i, Measurement Specialties, Hampton, VA, USA) mounted at the fulcrum of pendulum. In each trial, 3D forces at the femoral neck were acquired from a 3D piezoelectric load cell (model 9074C, Kistler, Novi, MI, USA) mounted in the neck of femur. Furthermore, the total force applied to the skin surface of the pelvis was measured from a force plate (model 2535-08, Bertec, Columbus, OH, USA) mounted on the ground. Data were acquired at 1000 Hz, and three trials were acquired in each condition.

3.3.3. Data Analysis

I considered both the peak compressive and tensile stresses generated in femoral neck during impact, based on previous studies which suggest that each has an important role in the etiology of hip fractures [69-73].

Force traces recorded during impact (from both the load cell and force plate) typically exhibited three successive peaks (Figures 3-3b, 3-3c and 3-3d), with the
second peak ($F_{\text{max}2}$) consistently exceeding the first ($F_{\text{max}1}$) and third peak ($F_{\text{max}3}$). From each trial, I identified: (1) the peak force from force plate ($F_{\text{applied}}$); (2) the peak axial force ($F = F_x$) and (3) peak shear force ($V = F_x + F_y$) at the femoral neck; (4) the peak bending moment ($M = V \cdot e$) applied to the femoral neck; and the peak estimated values of (5) compressive and (6) tensile stress at the femoral neck, given by:

\[ \sigma_{\text{compressive}} = \sigma_{\text{axial}} + \sigma_{\text{bending}} = \frac{F}{A} + \frac{M_y}{l} = \frac{F_x}{\pi r^2} + \frac{V \cdot e \cdot r}{4 r^4} \quad \text{and} \quad \sigma_{\text{tensile}} = \sigma_{\text{axial}} - \sigma_{\text{bending}} = \frac{F}{A} - \frac{M_y}{l} = \frac{F_x}{\pi r^2} - \frac{V \cdot e \cdot r}{4 r^4} \]

where, the distance (e) between the center of the femoral head and the load cell was 27 mm, and the radius (r) of the femoral neck was 16.5 mm (both measured with calipers directly from the device). I also report average values of these variables in the 10 ms window occurring just prior to impact (Figure 3-3c), which reflect baseline levels of pre-stress in the bone. All data analyses were conducted using a customized routine developed in MATLAB version 7.5 (MathWorks Inc., Natick, Massachusetts, USA).

### 3.3.4. Statistics

Analysis of variance (ANOVA) was used to test whether each outcome variable associated with muscle force (3 levels) and knee boundary conditions (3 levels). When ANOVA results indicated a significant effect, I conducted pair-wise comparisons using a Bonferroni correction. Each analysis was conducted with SPSS 18.0, using a total significance level of $\alpha = 0.05$.

### 3.4. Results

#### 3.4.1. Peak forces and stresses during impact

I observed significant main effects for muscle force ($p < 0.0005$) on each outcome variables (Figure 3-4), except for peak total force ($p = 0.2$). On average, when the muscle force increased from 400 to 1200 N, the peak compressive stress at the femoral neck decreased 24% (9.2 (SD = 2.4) versus 7.0 (SD = 1.1) MPa, $p < 0.0005$),
and the peak tensile stress at the femoral neck decreased 47% (6.4 (SD = 2.2) versus 3.4 (SD = 0.4) MPa, p < 0.0005). Over the same range, average values of peak shear force (and bending moment) decreased by 36% (1014.2 (SD = 300.7) versus 645.0 (SD = 105.3) N, p < 0.0005), and the peak axial force increased 45% (1268.9 (SD = 131.9) versus 1843.9 (SD = 260.1) N, p < 0.0005). Each of these variables scaled with muscle force in a relatively linear manner (Figure 3-4). In particular, compressive stress was well described by the equation $10.3 - 0.0028 \times \text{muscle force}$ ($r^2 = 1.0$), and tensile stress by $7.7 - 0.0038 \times \text{muscle force}$ ($r^2 = 0.98$).

I also observed a significant effect of knee boundary conditions (p < 0.0005) on each outcome variable (Figure 3-5). On average, when compared to the “fixed knee” conditions, the “free knee” condition yielded 40% lower values of peak compressive stress (6.0 (SD = 0.4) versus 10.0 (SD = 1.7) MPa, p < 0.0005), 47% lower values of peak tensile stress (3.3 (SD = 0.4) versus 6.2 (SD = 2.2) MPa, p < 0.0005), 44% lower values of peak shear force (580.5 (SD = 67.5) versus 1041.5 (SD = 272.2) N, p < 0.0005) and bending moment, and 25% lower values of peak axial force (1343.1 (SD = 175.9) versus 1793.8 (SD = 320.3) N, p < 0.0005). Peak total force recorded by the force plate was also significantly influenced by knee boundary condition (p < 0.0005), being 5% lower in the “free knee” than “fixed knee” condition (3295.9 (SD = 93.3) versus 3465.3 (SD = 143.6) N).

For each outcome, I also observed significant interactions between muscle force and knee boundary condition (p < 0.0005). In general, muscle force had a slightly greater influence on peak force and stress in the femoral neck in the fixed than free knee condition. Furthermore, the effect of knee boundary conditions on peak total force was most pronounced at low magnitudes of muscle force.

### 3.4.2. Peak forces and stresses prior to impact

The forces and stresses at the femoral neck prior to impact were also influenced by hip abductor muscle forces (p < 0.0005; Figure 3-4). On average, when muscle force increased from 400 to 1200 N, peak compressive stress increased from 0.2 (SD = 0.1) to 4.6 (SD = 0.5) MPa, peak tensile stress increased from 0.16 (SD = 0.07) to 2.76 (SD =
0.33) MPa, shear force increased from 24.3 (SD = 15.6) to 481.9 (SD = 55.6) N, and axial force increased from 25.3 (SD = 50.2) to 795.1 (SD = 167.2) N.

Forces and stresses were also lower in the “free knee” than “fixed knee” condition (p < 0.012), but not for tensile stress. On average, peak compressive stress was 17% lower (2.4 (SD = 1.8) versus 2.9 (SD = 2.1) MPa), shear force was 11% lower (260.5 (SD = 203.5) versus 294.7 (SD = 215.3) N), and axial force was 41% lower (335.0 (SD = 273.1) versus 567.0 (SD = 410.2) N). There was a significant interaction between muscle force and knee condition for axial force (p = 0.006), but not for compressive stress, tensile stress, and shear force (p > 0.4).

3.5. Discussion

In this study, I employed a mechanical hip impact simulator to examine how hip abductor muscle forces and knee boundary conditions affect the peak forces and stresses in the femoral neck during a fall. I found strong support for my hypothesis that peak forces and stresses would decrease with increases in hip abductor muscle force. A three-fold increase in muscle force (from 400 to 1200 N in each of the gluteus medius and minimus) caused a 24% reduction in peak compressive stress, and a 47% reduction in peak tensile stress at the femoral neck during impact. As hypothesized, this followed from a 36% reduction in peak shear force and bending moment. Furthermore, these variables scale with muscle force in a linear manner, with peak compressive stress decreasing 280 kPa, and peak tensile stress decreasing 380 KPa, on average, per 100 N increase in hip abductor muscle force.

In contrast, I did not find support for my hypothesis that peak forces and stresses in the femoral neck would decrease by impacting the hip with the knee constrained as opposed to free. In contrast, peak forces and stresses were substantially lower in the “knee free” than “fixed knee” condition (reductions of 40% and 47% in peak compressive and tensile stress, and 44% and 25% in peak shear and axial force). This appeared to be due in part to increases in the “fixed knee” condition in total effective stiffness (and perhaps effective mass), and a corresponding increase of 5% in peak total force recorded by the force plate, which offset the benefit of the knee reaction force in sharing
this load. These results imply that falls with knee fixed at impact (to the ground or contralateral leg) may create substantially greater fracture risk than falls with the knee free in the air.

Our observation of a protective effect of hip abductor muscle forces in reducing peak bone stress complements results from clinical cohort studies, which indicate that increased lower extremity muscle strength decreases hip fracture risk in older adults in the event of a fall, independent of bone density [74, 75]. These results highlight the potential value of including assessment of hip abductor muscle strength in screening hip fracture risk in older adults, and muscle strengthening exercises in hip fracture prevention.

To place our results in context, it is useful to compare our measures of peak bone stress to the range of stress that has been shown to fracture cadaveric samples of elderly cortical or trabecular bone (Table 3-1). In our worse-case condition (400 N muscle force and “fixed knee”), peak compressive stress averaged 11.9 MPa, and peak tensile stress averaged 8.8 MPa. These values are close to the range that has been shown to cause tensile failure (10 - 272 MPa), but not compressive failure (40 - 365 MPa) of cortical bone, and well within the range shown to cause failure of trabecular bone (0.4 - 27.3 MPa under compression, and 1.3 - 3.5 MPa under tension). This suggests that, even under our moderate fall impact velocity (2 m/s), peak stresses in the femoral neck approach the failure limits for elderly bone. It is also interesting to note that our values of peak stress due to muscle contraction in the pre-impact stage of our trials (which ranged from 0.2 to 4.6 MPa for compressive stress and 0.16 to 2.76 MPa for tensile stress) were within the range of failure stress for trabecular bone, but far below those for cortical bone.

It is also useful to compare the combined effect on peak applied stress of the two fall-related variables I explored, to document age-related changes in the force or stress that cause bone fracture. Peak stresses in our best-case condition (1200 N muscle force and “knee free”) were 5.6 MPa for compressive and 3.1 MPa for tensile, or 2.1 and 3.0-fold lower than in our worst-case condition. In comparison, Burstein et al. (1976) showed that the ultimate compressive stress of human femoral cortical bone decreased 1.16-fold (from 209 to 180 MPa) from the twentieth to the eightieth, and Ebbesen et al. (1999)
found that the ultimate compressive stress of human vertebral trabecular bone decreased 4.6-fold (from 7 to 1.5 MPa) from the thirtieth to the tenth decade. These comparisons indicate the crucial role of both fall mechanics and bone strength on hip fracture risk. Our results should help to inform improved strategies for reducing fall severity, including better approaches for assessing the biomechanical performance of wearable hip protectors [30] and compliant flooring [76], and the development and training of strategies for “safe landing” in the event of a fall [77].

This study had several important limitations. First, I used a mechanical hip impact simulator to examine the isolated effect of changes in hip abductor muscle forces on peak forces and stresses at the femoral neck during a sideways fall. While I examined a range of force in the abductors typical of daily activities (1-3 times body weight), experimental data are not currently available on the state of muscle activation during the impact stage of falls in humans, to confirm the accuracy of the scenarios I examined (e.g., relative activation of hip abductors versus hip adductors). Second, our experimental trials simulated impact to the lateral aspect of the pelvis, with zero internal/external rotation and abduction/adduction of the proximal femur. However, studies have shown that the orientation of the femur at impact from a fall influences both the total applied force [41] and femoral fracture strength [19, 32]. Future studies should expand the ability in our impact simulator to examine a range of impact positions. Finally, each of our trials involved an impact velocity of 2 m/s, representing a fall of moderate severity. While I expect that similar trends would be observed at higher impact velocities, additional experiments are required to verify this, including use of the 3.4 m/s value suggested by recent guidelines for the biomechanical assessment of wearable hip protectors [30].

In summary, I found that the state of stress in the femoral neck at impact during simulated sideways falls depends strongly on the magnitude of hip abductor muscle force. This complements epidemiological evidence that lower extremity muscle strength affects hip fracture risk during a fall [74, 75]. I also found that knee boundary conditions strongly influenced peak stresses, profoundly affecting fracture risk. These results should help to inform improved approaches for assessing the biomechanical performance of wearable hip protectors and compliant flooring, screening of hip fracture risk in older adults based on hip abductor muscle strength (and risk reduction through
resistance training), and the development and training of strategies for “safe landing” in the event of a fall.

**Figure 3-1. Predicted states of stress in the femoral neck.**

(a) During the single-legged stance phase of gait, the inferior aspect of the femoral neck receives compressive stress, while the superior aspect experiences tensile stress. (b) During impact to the greater trochanter in a sideways fall, the state of stress is reversed, with the inferior aspect under tension, and the superior aspect under compression.
Figure 3.2. *Hip impact simulator used in simulated sideways falling experiments.*

A load cell recorded axial and shear forces at the femoral neck. Wire ropes were used to simulate hip abductor muscle forces. Foam polymer simulated soft tissues. The distal femur was either free, or attached to the shaft of pendulum via a coil spring (shown) or wooden block.
Figure 3-3. Experimental measures and calculated outcome variables.

(a) Free body diagram and stress analysis at the femoral neck during impact. Sample force and stress traces for (b) 400 N muscle force and “fixed knee” condition; (c) 1200 N muscle force and “fixed knee” condition; and (d) 1200 N muscle force and “free knee” condition. Increased muscle force caused peak axial force (F) to increase, but peak shear force (V) and compressive stress to decrease (b versus c). Peak axial force (F), shear force (V) and compressive stress were further reduced in the “free knee” than “fixed knee” condition (d versus c).
During impact, increased muscle force caused significant reductions in peak values of compressive and tensile stress (top left) and shear force (middle left), and increases in peak axial force (also shown in middle left). Before impact, increased
muscle force caused significant increases in all stresses and forces in the femoral neck (top and middle right).

Figure 3-5. Association between knee boundary condition and outcome variables during impact.

All stresses and forces in the femoral neck were lower in the “knee free” than “fixed knee” condition, and of intermediate magnitude in the “spring knee” condition.
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4. Kinematic analysis of pelvis impact velocities from video-captured falls by older adults in long-term care

4.1. Abstract

Falls are the cause of 90% of hip fractures in older adults. Risk for hip fracture depends largely on the forces generated during impact, which depend, in turn, on the velocity of the body segments at landing. While previous studies have reported pelvis impact velocities during laboratory-based falls in young adults, no study has examined impact velocities during falls in older adults.

I addressed this issue by analyzing video footage of real-life falls, captured through networks of digital cameras in two long-term care facilities. I estimated the time-varying position of the pelvis by digitizing pelvis landmarks from fall initiation through impact, using calibration data specific to each fall location, fall orientation, and camera view in a direct linear transformation algorithm. I then differentiated position data to determine vertical and horizontal velocities at impact. In laboratory-based falling experiments, the measurement error was deemed acceptable for forward and backward falls, with a mean bias of 0.16 m/s (which I applied as a correction), and a standard deviation (SD) of 0.23 m/s. Accuracy was deemed unacceptable for sideways falls (SD = 0.59 m/s), which were excluded from analysis.

From subsequent analysis of 21 backward and 4 forward real-life falls experienced by 23 older adults (of mean age 80 (SD = 10)), the impact velocity of the pelvis averaged 2.08 m/s (SD = 0.62). The time interval between the moment of imbalance and impact of the pelvis averaged 1271 ms (SD = 648). The average vertical descent distance of the pelvis during the fall was 0.81 m (SD = 0.17). Our vertical velocities were, on average, 47.8 % below simple free-fall predictions based on fall height ((2gh)^0.5), and 20% below average values previously reported for young adults.
in laboratory falling experiments. Several mechanisms appeared to contribute to reducing pelvis impact velocity, including hand impact and attempts to recover balance by stepping.

Our results should be useful for falls researchers and engineers in understanding the injury potential of falls, and designing interventions (such as exercise programs, wearable hip protectors and compliant flooring) for hip fracture prevention.

4.2. Introduction

Falls are the number one cause of injuries in older adults, including 90% of hip fractures [13]. However, only a very small portion of falls (1-2%) cause hip fracture. An understanding of the factors that determine “fall severity” - and separate injurious and non-injurious falls - is important for researchers and engineers in designing and selecting interventions for fracture prevention (such as wearable hip protectors and compliant flooring), and for clinicians and health administrators in treating and preventing fall-related injuries. In this regard, a consideration of the kinematics of falls is important for several reasons.

First, kinematic analysis can yield estimates of the impact velocities of contacting body parts [35-40]. Hip fracture risk during a fall depends largely on the forces applied to the hip region during impact. These forces depend, in turn, on the velocity of the pelvis at landing [33]. Second, kinematic analysis can provide information on the time duration of falls, which should be relevant to the faller’s ability to initiate and execute protective responses, such as arresting the fall with the upper limbs [57]. Finally, kinematic analysis can provide evidence on the frequency of these responses, and how they influence impact velocity [38, 40, 57].

Unfortunately, there is currently little understanding of the kinematics of falls in older adults. Current knowledge is limited to laboratory studies with young adults falling on gym mats [35-40]. Collectively, these studies suggest that the vertical velocity of the pelvis at impact averages 2.58 m/s (SD = 0.53), and that the time interval between fall initiation to pelvis impact averages 867 ms (SD = 237) [36, 38, 40]. These studies also suggest that, in landing from a fall, young adults tend to avoid head impact, and
distribute impact energy through near-simultaneous impacts to the pelvis and outstretched hands [36, 40].

However, lab-based falling experiments with young adults may not reflect the range of falling patterns (and impact velocities) occurring in real-life falls among older adults, due to differences in the situational and environmental context of falls, or intrinsic factors such as physical and cognitive function, medication use, and disease diagnoses.

Recently, our research team reported results on the causes and activities associated with falls from analysis of 227 real-life falls captured on video cameras in two long-term care (LTC) facilities [58, 59]. In the current study, I conducted kinematic analysis on a subset of fall videos, to estimate pelvis impact velocities and contact times. I compare our findings for older adults to previous results for young adults. I also compare our vertical impact velocities to theoretical free-fall predictions, and explore how impact velocity was affected by fall direction and upper limb impact.

4.3. Methods

4.3.1. Real-life falls library

As part of a study ongoing since 2007 on the cause and circumstances of falls in older adults, we have partnered with two LTC facilities in the Vancouver area (Delta View, a 312-bed facility in Delta, BC, and New Vista, a 236-bed facility in Burnaby, BC) to capture video footage of real-life falls in older adults [58]. Delta View had a network of 216 digital cameras, while New Vista had 48.

Between April 2007 and February 2013, I captured 813 falls experienced by 306 individuals. When a fall occurred, facility staff completed an incident report and contacted our team so that I could collect video footage. Each video was recorded with a frame rate of 30 frames per second and a resolution of 640 by 480 or 480 by 720 pixels. The study was approved by the Office of Research Ethics at SFU. Each resident provided written consent to the facilities for video capture of their images, and these data were shared as secondary data with the research team. Additional written consent was secured from individuals to share their images.
Each fall video was analyzed by a team of at least three experts using a structured, validated questionnaire [59] to determine (a) the cause of fall, (b) the activity at the time of the fall, (c) the initial direction of the fall (forward, backward, sideways, and straight down) (d) the landing configuration (forward, backward, or sideways), (e) the occurrence (if any) of stepping responses, (f) the occurrence of impact to the hand(s) or knee(s), (g) the estimated time instant of the onset of imbalance (leading to the fall), and (h) the estimated time instant of the initiation of the fall (defined as one frame after the last recovery step, if any). As described below, I used these latter two parameters, along with the pelvis impact time, to calculate the duration of the fall.

Each fall video was reviewed and cases were excluded (Figure 4-1) that did not involve pelvis impact (n = 63); where the pelvis was occluded from camera view during descent or impact (n = 137); where the fall occurred far from the camera and difficult to recognize the pelvis (n = 21); where the fall involved initial impact to objects other than the floor (n = 18); and where the faller fell directly toward or away from camera and difficult to digitize the pelvis (n = 66). In selecting and analyzing the remaining cases, I considered the results of experiments (described below) to measure the accuracy of our approach for estimating impact velocities from planar video, through comparison with "gold standard" estimates from 3D motion capture.

4.3.2. Two-dimensional direct linear transformation (2D DLT)

Our approach for estimating impact velocities involved digitizing points of interest in the 2D image plane, and applying a two-dimensional direct linear transform (2D DLT) process to reconstruct those points in a corresponding plane in 3D space, via a transformation matrix obtained through a calibration process. In this process, I utilized MATLAB software (run on version 7.5, MathWorks Inc., Natick, Massachusetts, USA) developed and freely shared by Hedrick and colleagues [78] and Meershoek [79]. The 2D DLT technique has been widely used for planar kinematic analyses of video footage, particularly to estimate joint angles and velocities in sports activities [80-85]. While the calibration process can correct for lens distortion, an important limitation of the 2D DLT technique is the potential for “perspective errors,” arising when the digitized points of interest move outside the calibrated image plane.
In an attempt to minimize the perspective errors, I calibrated each video specifically to the fall orientation. This process involved recording images (from the surveillance camera that captured the fall) of a flat calibration panel, having dimensions 160 cm x 160 cm, and contained a 5 x 5 grid of 10 cm diameter markers spaced 40 cm apart (see far-right images in Figure 4-2). The panel was placed with the bottom surface flush to the ground, centered at the midpoint of the faller's feet (at the moment of fall initiation), and lined between the feet and the location of the pelvis at impact. From the resulting images, the positions of the 25 markers were digitized [78], and the 8 DLT coefficients were computed [79].

4.3.3. Laboratory measures of accuracy

I tested the accuracy of our velocity estimates through laboratory falls with a human volunteer, and with an inverted pendulum (aluminum rod) of 1.57 m length, connected to a low-friction ball joint at the floor. Each trial was captured with a surveillance camera identical to the type most common in the LTC sites (model WVC210, Cisco Systems, San Jose, CA) recording at 30 Hz, and an 8 motion capture camera system (Motion Analysis Corp., Santa Rosa, CA, USA) recording at 250 Hz. The surveillance camera was placed at a height of 2.6 m and horizontal distance of 5 m from the site of the fall.

Trials were conducted at five falling angles with respect to the surveillance camera position: 30 deg (toward the camera), 60 deg, 90 deg (perpendicular to the camera axis), 120 deg, and 150 deg (away from the camera). For the human participant, trials were also conducted for three falling directions: forward, backward, and sideways. A single trial was acquired for each combination of camera angle and fall direction. Reflective markers were placed on both greater trochanters on the human, and on the midpoint of the rod.

From each video, I conducted frame-by-frame digitization of either the greater trochanter or posterior aspect of the sacrum (depending on which was deemed more visible to the camera view throughout the fall), throughout the interval starting at least two frames before fall initiation, and ending at least five frames after initial impact of the pelvis appeared to occur. I transformed these results to position data using the above-
described 2D DLT procedure, using calibration coefficients specific to each fall orientation. I then used finite difference to estimate time-varying vertical and horizontal velocities. The resulting velocity – time traces were fit with a fifth-order polynomial (which provided the best fit to predictions from the inverted pendulum) using the Matlab `polyfit` function (Figure 4-3). The vertical impact velocity of the pelvis was defined as the peak value of the curve fit. I also report horizontal velocities occurring at the instant of peak vertical velocity (which I refer to as “horizontal impact velocity”).

The accuracy of the method was expressed as the mean value, and the standard deviation (SD) across the five fall orientations, in the difference between impact velocity estimates based on the surveillance camera data, and the 3D motion capture data [86]. Results were considered separately for forward, sideways, and backward falls.

### 4.3.4. Analysis of video-captured falls in older adults

These accuracy results were used to guide the inclusion of video-captured falls in older adults for analysis, based on fall direction. In particular, my threshold for “acceptable” accuracy, for a given fall direction, was a standard deviation in the error of our laboratory-based estimates of vertical impact velocity of 0.25 m/s or less, which would reflect a 95% confidence interval of 1.0 m/s (a clinically meaningful difference). I also excluded falls involving significant rotation during descent (n = 208), and falls occurring “straight down” (n = 119). Fall videos in our older adult dataset meeting these additional inclusion criteria were analyzed using the same digitization and calibration protocol as described above.

I provide descriptive results (means and standard deviations) for both vertical and horizontal impact velocities of the pelvis. However, I focus primarily on vertical velocity as a more relevant indicator of fall severity. I also report fall durations, defined either as (a) the interval between the onset of imbalance and the estimated instant of initial pelvis impact, or (b) the interval between the onset of fall initiation (as defined above) and initial pelvis impact. Finally, I report the vertical descent distance of the pelvis from imbalance to impact (or fall height $h$) for each fall, and compare measured vertical velocities to theoretical estimates based on free fall of a falling mass (where
impact velocity = \((2*9.81*h)^{0.5}\) or an inverted pendulum with uniformly-distributed mass (where impact velocity = \((3/2*9.81*h)^{0.5}\)).

4.4. Results

4.4.1. Laboratory falls

In my laboratory experiments with the inverted pendulum, I observed strong agreement between the 2D DLT and 3D motion capture techniques. The average difference in the vertical impact velocity was \(0.008\) m/s, and the SD was \(0.04\). The average difference in the horizontal impact velocity was \(-0.11\) m/s (SD = 0.07). In my laboratory-based falls with a human participant, I observed acceptable accuracy in my impact velocity estimates for forward and backward falls, but not for sideways falls (Figure 4-4). In forward falls, the mean difference in vertical impact velocity between the 2D DLT and 3D motion capture techniques was \(-0.26\) m/s (where a negative mean reflects overestimation from the 2D DLT technique), and the SD was \(0.21\) m/s. For backward falls, the mean difference was \(-0.06\) m/s, and the SD was \(0.21\) m/s. In sideways falls, the mean difference averaged \(-1.01\) m/s and the SD was \(0.59\) m/s. The mean difference in horizontal impact velocity was \(-0.15\) (SD = 0.32) for forward falls, \(-0.16\) (SD = 0.18) for backward falls, and \(-0.13\) (SD = 0.54) m/s for sideways falls. Based on the high observed SD for sideways falls, sideways falls in our older adult dataset (n = 152) were excluded from further consideration. Since the error was not different between forward and backward falls \(p > 0.1\), and identical for vertical and horizontal velocity (at \(-0.16\) m/s), my baseline 2D DLT impact velocity estimates were adjusted by subtracting the combined mean bias (overestimate) of \(0.16\) m/s. The pooled SD for forward and backward falls of \(0.23\) for vertical velocity (and \(0.24\) for horizontal velocity) reflects a 95% confidence interval of +/- 0.46 m/s in our estimated vertical impact velocities, and +/- 0.48 m/s in our estimated horizontal impact velocities.

4.4.2. Falls by older adults

Our final analysis included 25 video-captured falls in 23 older adults (Table 4-1). The average age of the faller was \(80.3\) yrs (SD = \(9.8\)), and \(56\%\) \((n = 14)\) were female.
The average fall height was 81.5 cm (SD = 17.0). Two cases (video ID #23 and #24) occurred while using an assistive device (walker or wheelchair). Twenty-one cases (84%) involved backward falls, and 4 cases (16%) involved forward falls. Hand impact occurred before pelvis impact in 21 cases (all 4 forward falls, and 17 of 21 backward falls). The knee(s) impacted before the pelvis in all 4 forward falls, and no backward falls. Head impact occurred in 12 cases (all 4 forward falls, and 8 of 21 backward falls), always after pelvis impact. The most common cause of imbalance was incorrect weight shifting (12 of 25 cases), followed by hit/bump (7 cases). Trips, collapses, and loss-of-support each accounted for 2 falls. The most common activity at the time of falling was standing (14 of 25 cases), followed by walking (8 cases), and transferring from standing to sitting (3 cases). Stepping after the onset of imbalance occurred in 16 of 25 falls.

Over all 25 falls, the average vertical impact velocity of the pelvis was 2.08 m/s (SD = 0.62) and the average horizontal impact velocity was 1.16 m/s (SD = 1.41). When compared to forward falls, backward falls involved 20% higher vertical impact velocities (2.13 m/s (SD = 0.64) versus 1.77 m/s (SD = 0.34); Figure 4-5a), and 55% lower horizontal impact velocities (0.96 m/s (SD = 1.41) versus 2.16 m/s (SD = 1.01)). In backward falls, the average vertical impact velocity exceeded the horizontal velocity by 122%, while in forward falls, the average value of vertical impact velocity was 18% lower than the average horizontal velocity (Figure 4-5a).

The average time interval between the onset of imbalance and impact to the pelvis was 1271 ms (SD = 648), while the average time interval between fall initiation and pelvis impact was 593 ms (SD = 255). The time interval from the moment of imbalance to impact was 44% greater in forward than backward falls (1706 ms (SD = 914) versus 1188 ms (SD = 578); Figure 4-5b). Similarly, the time interval between initiation and impact was 71% greater in forward than backward falls (911 ms (SD = 391) versus 532 ms (SD = 175)).

Hand impact, stepping responses, cause of imbalance, and activity at the time of falling had no clear effect on vertical impact velocities or fall durations, which varied considerably within each response category (Table 4-1). The vertical velocity averaged 2.03 m/s (SD = 0.58) in trials where one or both hands impacted before the pelvis versus 2.31 m/s (SD = 0.86) in falls not involving hand impact, and 2.04 m/s (SD = 0.57) in trials
where steps occurred after imbalance versus 2.15 m/s (SD = 0.72) in falls not involving steps. The vertical impact velocity averaged 2.26 m/s (SD = 0.70) for falls from standing, 1.82 m/s (SD = 0.59) for falls while transferring, and 1.86 m/s (SD = 0.36) for falls from walking. Finally, the average vertical velocity was 2.11 m/s (SD = 0.38) in falls due to incorrect transfer, 2.04 m/s (SD = 0.70) in falls due to bumps, and 2.34 m/s (SD = 1.07) in falls due to collapses.

When compared to theoretical predictions based on free-fall from each measured fall height (Table 4-1 and Figure 4-6), our vertical impact velocities averaged 47.8% (SD = 14.2) lower than predictions from a falling mass model (which averaged 3.98 m/s (SD = 0.41)), and 39.7% (SD = 16.4) lower than predictions from an inverted pendulum model (which averaged 3.45 m/s (SD = 0.36)). Furthermore, regression analysis (using SPSS 18.0) indicated that our measured vertical impact velocities were predicted poorly by a linear curve fit \(v = 0.012h + 1.116, \ R^2 = 0.106, \ p = 0.112\) and nonlinear curve fit \(v = (h)^{0.523} + 0.202, \ R^2 = 0.13, \ p = 0.074\) by the fall heights I measured.

### 4.5. Discussion

In this pilot study involving kinematic analysis of 25 real-life falls in older adults captured on video, I found that the vertical impact velocity of the pelvis averaged 2.08 m/s. On average, this was 48% lower than theoretical predictions (based on fall height) from a falling mass model, and 40% lower than predictions from an inverted pendulum model. Furthermore, our vertical impact velocities average 20% lower than the mean value (2.58 m/s (SD = 0.53)) reported across six laboratory-based falling studies involving young adult participants [35-40]. This suggests that, among the falls I analyzed, specific mechanisms were utilized to absorb energy during descent, and reduce the vertical velocity of the pelvis. These included contacting the pelvis with the trunk relatively upright and squatting during descent (both of which appeared to be common in backward falls), impacting the ground with the outstretched hands before pelvis impact, and attempts at balance recovery by stepping after the onset of imbalance.

I used two methods to describe the duration of the fall, to account for the confounding influence of stepping after the onset of imbalance (which occurred in 64%
of falls). The interval between pelvis impact and the apparent onset of imbalance averaged 1271 ms. This reflects the total time available to the faller to execute balance recovery and safe landing responses. The time interval from pelvis impact to fall initiation (defined as one frame after foot contact during the last step) averaged 593 ms. This reflects the typical time interval, after attempted balance recovery, available to the faller to execute safe landing. Previous laboratory studies of falls in young adults have typically involved sudden perturbations, and have reported the average time interval between the onset of the perturbation and pelvis impact as 867 ms (SD = 237) [36, 38, 40], which lies between our two values.

Previous studies have also reported that the time required for older adults to move their hands into a protective position to arrest a fall averages 615 ms (SD = 88), which is well within our average fall duration of 1271 ms [57]. Indeed, I found that one or both hands impacted the ground before the pelvis in 100% of forward falls, and in 84% of backward falls. Pelvis impact velocities averaged 12% lower in falls involving hand impact (compared to no hand impact), and 5% lower in falls involving stepping (compared to no stepping). Stepping responses also increased total fall duration by 78%. These findings agree with results from laboratory-based falling experiments with young adults, where pelvis impact velocities were decreased 22% by taking a step, and 18% by impacting the hands prior to the pelvis [40].

Analyzing the complex movements of falls from planar video is challenging, due to the out-of-plane motions of the body segments that often accompany during descent. I found that, in laboratory experiments that also involved 3D motion capture, our DLT-based technique was accurate, on average, to within 0.008 m/s (SD = 0.04) in measuring the impact velocity of a 1.57 m inverted pendulum. I also found that it provided acceptable accuracy (with a 95% confidence interval of +/- 0.46 m/s) for forward and backward falls with our human participant (where the trajectory of body parts remained parallel to the calibration plane). However, measurement accuracy was unacceptable for sideways falls (where knee and trunk flexion often caused out-of-plane movement of the pelvis), which were excluded from analysis. Future studies might address this limitation by capturing falls with multiple camera views, and employing 3D analysis techniques.
Several additional limitations exist to our study. Our results are based on falls experienced by residents in LTC, and may not apply to healthier community dwelling older adults, or young adults. Furthermore, I had limited access to medical records, and did not compare the characteristics of falls to physical and cognitive function, medications, or disease diagnoses. Certainly, much is to be learning in future studies that relate the kinematics of the fall to the clinical context.

In summary, I analyzed falls in older adults captured by surveillance cameras in LTC, to determine pelvis impact velocities and fall durations. To my knowledge, this is the first study to report these measures from real-life falls in older adults. The vertical impact velocity of the pelvis averaged 2.08 m/s, or 40% lower than theoretical predictions from an inverted pendulum model. The time interval between the moment of imbalance and impact of the pelvis averaged 1271 ms. The impact velocity averaged 12% lower in falls that involved hand impact, and 5% by stepping. My results should be useful for falls researchers and engineers in understanding the injury potential of falls, and designing interventions (such as exercise programs, wearable hip protectors and compliant flooring) for hip fracture prevention.
Among 813 fall videos captured, 25 forward and backward falls involving the pelvis impact were selected for analysis.
Figure 4-2. Fall video snapshots

(a) forward fall in an older adult; (b) forward fall in a young adult; (c) backward fall in an older adult; and (d) backward fall in a young adult. The far-right panel illustrates the 25-marker calibration panel, placed at the exact location of the fall (in the lab or LTC facility), and oriented in the plane of the fall.
Figure 4-3. Sample pelvis velocity traces

(a) Forward fall in an older adult; (b) backward fall in an older adult; (c) forward fall in a young adult; and (d) backward fall in a young adult. Vertical velocities are shown in red, and horizontal velocities in black.
Figure 4-4. Statistical agreement between the 2D DLT and 3D motion capture

Agreement from laboratory experiments between the 2D DLT technique and 3D motion capture in (a) vertical impact velocity and (b) horizontal impact velocity. Results are shown for the mid-point of a 1.57 m length pendulum, and for the pelvis for a human participant falling in the forward, backward, and sideways directions (at five different camera angles; see text for explanation). Sideways falls in human exhibited larger variability between camera angles (vertical velocity SD = 0.59 m/s, horizontal velocity SD = 0.54 m/s) than forward and backward falls. Among forward and backward falls, the mean difference between the techniques was -0.16 (SD = 0.23) for vertical velocity, and -0.16 (SD = 0.24) m/s for horizontal velocity.
Figure 4-5. Impact velocity and fall duration during a fall

Mean values of (a) pelvis impact velocity and (b) fall duration for forward and backward falls. Fall duration is shown as the time interval between pelvis impact, and either the last step before falling (initiation) or the onset of imbalance.
Figure 4-6. Vertical velocities of the pelvis at impact during a fall compared to theoretical predictions.

The vertical impact velocities were 47.8% (SD = 14.2) and 39.7% (SD = 16.4) lower than free-fall and pendulum fall predictions, respectively.
Table 4-1. Impact velocities and characteristics of selected falls (n = 25) from 23 different residents.

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<th>Pelvis impact</th>
<th>Hand impact</th>
<th>Knee impact</th>
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<th>Activity at the time of fall</th>
<th>Stepping response</th>
<th>Injuries noted</th>
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<th>Peak horizontal velocity (m/s)</th>
<th>Horizontal velocity at peak vertical velocity (m/s)</th>
<th>Impact time A (ms)</th>
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Mean: 2.08 ± 1.58; SD: 0.62 ± 1.92.

Fall direction: F = forward, B = backward; Cause of fall: T = trip/stumble, B = hit/bump, IT = incorrect transfer, LOS = loss of support, C = leg collapse; Activity at the time of fall: W = walking, S = standing, T = transferring; Head injuries included redness/swelling/cuts/scrapes; Impact time A = time interval between the moment of imbalance and impact; Impact time B = time interval between fall initiation and impact; Fall height = vertical descent distance of the pelvis from fall initiation to impact.
5. Synthesis of results and future directions

This PhD thesis addresses key knowledge gaps and provides important new observations to enhance our understanding of the biomechanics of all-related hip fractures in older adults. In particular, in Chapter 2, I describe how age-related softening of soft tissue over the hip region may affect hip fracture risk, based on dynamic compression experiments with young and older women. In Chapter 3, I show how impacting the ground from a fall with the hip abductor muscles contracted is protective, based on mechanical tests using a hip impact simulator. In Chapter 4, I document impact velocities (and times to impact) accompanying real-life falls in older adults captured on video. Below, I summarize my key findings, and discuss future research needs.

In chapter 2, I conducted vibration experiments with young and older women, and measured stiffness and damping constants of soft tissue over the greater trochanter (GT). I also measured soft tissue thickness over the GT using ultrasound. I found that on average stiffness and damping were 2.9-fold and 3.5-fold lower, respectively, in older than young adults. This suggests that, when compared to young adults, older adults’ soft tissue absorb and dissipate about 3-fold less energy during impact to the hip, thereby increasing the energy that must be absorbed through alternative mechanisms, including deformation of the underling bones (proximal femur and pelvis). This may help to explain, along with age-related declines in bone strength, the dramatic increase in hip fracture incidence with age. Surprisingly, the age-related tissue softening effect did not arise from soft tissue thickness. Instead, the effect seems to be explained by age-related changes in tissue properties.

These results should inform hip fracture risk assessment in older adults, perhaps involving the development of clinical tools for measuring tissue stiffness and damping. Such measures may also inform the design, selection, and testing of wearable “hip
protectors” or compliant flooring, or resistance training exercise to increase or maintain muscle tone.

My results suggest several directions for future research. An ultrasound-integrated indenter setup may provide better estimates of the stiffness and damping of individual soft tissue layers. Histology measures should be acquired (through biopsy or cadaver specimens) of the age-related changes in the collagen and elastin content of skin and fat, and muscle fiber geometry, to help understand the mechanisms underlying changes in tissue properties. Clinical trials should examine the effect of resistance training exercise programs on soft tissue stiffness and damping in older adults.

In chapter 3, I developed a hip impact simulator that allowed for variation in hip abductor muscle forces and knee boundary conditions. I used this system to simulate sideways falls, and determined (from measures of 3D forces) peak stresses at the femoral neck. I found that peak compressive stress decreased 24%, and peak tensile stress decreased 47%, when hip abductor muscle forces increased from 400 to 1200 N. I also found that peak compressive stress was 40% lower, and peak tensile stress was 47% lower under “free knee” than “fixed knee” boundary conditions. These effects are similar to the magnitude of decline in fracture strength associated with osteoporosis, and provide insights in risk assessment (hip abductor muscle strength testing) and prevention of hip fracture in older adults. In particular, the results indicate again the potential value of resistance training exercise in hip fracture prevention, and should be help to develop and train “safe landing” strategies (land “muscle contracted”) in falling and protective clothing design.

Again, these findings point toward important directions for future research. I used a mechanical system to examine the isolated effect of hip abductor muscle force on peak bone stress during a sideways fall. However, there are very limited data on the states of muscle activation that tend to accompany the impact stage of falls. Future work should simulate muscle forces, based on measures of muscle activity in actual falls. My results also have interesting implications for so-called spontaneous hip fractures. In particular, I observed that, even in pre-impact stage of falls (when only muscle forces were acting), the compressive and tensile stress at the femoral neck were within the range of failure stress for trabecular bone (although they were far below those for
cortical bone). This draws attention to the potential for muscle forces to create spontaneous fractures, which may be explored through further experiments with our hip impact simulator. Finally, I observed a 16% reduction in peak compressive stress, and a 23% reduction in peak tensile stress, between “fixed knee” and “spring knee” conditions. This points toward the value of further research on the potential value of padding the knee, as well as the hip region, to protect against hip fracture.

In chapter 4, I conducted kinematic analysis of real-life falls in older adults captured through video cameras in long-term care. In particular, I employed a 2D direct linear transformation algorithm to measure velocity of the pelvis at impact from a fall by digitizing the video footage. I also measured duration of falls (decent time of the pelvis) in each fall. I found that the pelvis impact velocities averaged 2.08 (SD = 0.62) m/s, which represented 48% lower than simple free-fall predictions and 20% lower than that of young adults. I also found that the vertical impact velocities can be reduced by 5% by executing a step and by 12% by impacting hands before the pelvis impact. I also found that time interval between the moment of imbalance and impact of the pelvis averaged 1,271 (SD = 648) ms, and this time interval is 2-fold greater than time required for older adults to place their hands to arrest falls (615 ms). These results likely support a notion that older adults have enough time to outstretch hands to avoid head impact or reduce impact severity during a fall. Indeed, 84% of analyzed falls analyzed involved hands impact before the pelvis impacts. These results should be useful for falls researchers and engineers in understanding the injury potential of falls, and designing interventions (such as exercise programs, wearable hip protectors and compliant flooring) for hip fracture prevention.

In terms of future research, improved methods are required (perhaps involving multiple camera views) to analyze sideways falls, which posed challenges to our 2D DLT technique due to out-of-plane motions of body segments. Sideways falls often involve hip impact and are thus particularly relevant to the issue of hip fracture. Analysis techniques should also expand beyond pelvis impact, to examine head impact and hand impact. Larger studies are required to link video evidence of fall characteristics with measures of functional status, medications, and disease diagnoses, along with the occurrence of fall injuries. While video capture of real-life falls in the community is
challenging, future studies might focus on comparing fall characteristics across the lifespan.

In summary, risk of fall-related hip fractures in older adults depends on bone strength as well as applied load transmitted to the proximal femur during impact. Three-fold reduction occurs with age in the stiffness and damping of soft tissue over the hip, which may affect the force and energy absorbed in the bone during a fall. Activation of muscles spanning the hip joint and knee boundary conditions strongly influence peak stresses at the femoral neck during a fall. A wide range of impact velocities accompany falls in older adults. The average value is 48% below free fall predictions, which may help to explain why most falls do not result in hip fracture. Impact velocities were reduced by upper extremity impact and stepping, and stepping also increased the time duration of the fall. These findings add important pieces to the puzzle of whether a particular fall will result in hip fracture, and informs future direction for clinical and laboratory-based research on hip fracture prevention.
References


