FACTORS AFFECTING THE EFFICACY OF HIP
PROTECTORS DURING FALLS

by

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BHS, PT, Yonsei University, 2001

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ABSTRACT

Hip protectors represent a promising strategy for preventing fall-related hip fractures in the elderly. However, fractures still occur when wearing a protector. From a biomechanical perspective, the protective benefit of a hip protector should depend on the wearer's body habitus, the fall orientation, and the position of the hip protector relative to the greater trochanter. This thesis is comprised of two studies designed to test this hypothesis. In the first study, I conducted experiments with human subjects which demonstrate that the reduction in peak magnitude and change in pressure distribution provided by a hip protector depends on body habitus and fall direction. In the second study, I conducted experiments with a hip impact testing system to show that force magnitude and distribution are affected by the position of the hip protector relative to the greater trochanter.

Keywords: Hip protectors; falls; ageing; hip fractures; pressure distribution; BMI; displacement; falling direction

Subject Terms: Biomechanics; impact mechanics; biomedical engineering
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CHAPTER 1 GENERAL INTRODUCTION

Falls are a major cause of injury, including over 90% of hip fractures (Grisso 1991). Wearable hip protectors represent a promising strategy to prevent hip fracture. These devices are usually integrated into tight-fitting undergarments, and are designed to reduce fracture risk by attenuating the total impact force and/or shunting it away from the greater trochanter (GT). Unfortunately, user compliance in wearing hip protectors has traditionally been low [Parker 2006]. In an effort to improve compliance, manufacturers have recently focused on more comfortable soft-shell designs. Additional research is required to understand how these devices influence the magnitude and distribution of force to the underlying skeletal structures, and to determine whether this depends on the body habitus of the faller, the landing configuration, and the exact position of the hip protector. My thesis addresses these questions through experiments with human subjects and with a mechanical hip impact simulator.

Clinical studies have shown that fracture risk decreases with increasing body mass index (BMI) [la Vecchia 1991], which may be due to the force attenuation (or “natural” hip protector effect) provided by a higher thickness of soft tissue over the GT [Robinovitch 1995, Lauritzen 1992]. Soft tissue thickness may influence the additional force attenuation provided by a wearable hip protector, providing little benefit in those who already possess sufficient natural padding, and considerable benefit to those who do not. Chapter 2 describes the
results of falling experiments with high BMI individuals, designed to test this hypothesis.

The experiments described in Chapter 2 also examine how the direction of the fall (anterolateral, lateral, or posterolateral) affects the force attenuation provided by soft shell hip protectors. This is an important issue, since evidence suggests that the fracture strength of the femur is lowest during falls onto the posterolateral aspect of the hip [Pinilla 1996, Keyak 2001, Nankaku 2005]. Accordingly, an important but previously unexplored question is whether hip protectors reduce impact force during falls in the posterolateral direction.

Chapter 3 focuses on how displacement of a hip protector relative to the greater trochanter can influence its protective benefit. Hip protectors are designed and tested with the assumption that they will remain in place in their intended location relative to the GT and femoral diaphysis. However, in normal everyday use, the protector may slide over the skin surface [Minns 2007, Derler 2005], or be displaced due to movement of the greater trochanter under the skin during hip rotation. Improved understanding is required of how this relative movement affects the force attenuation provided by the protector.

In summary, my thesis determines how the magnitude and distribution of force applied to the hip region during falls depends on the type of soft shell hip protector worn, the BMI of the faller, the configuration (angle) of the body during impact, and the relative position of the hip protector with respect to the GT. My specific research questions were as follows:
Research Question 1: How does a hip protector redistribute pressure over the hip region during a fall on the hip?

Research Question 2: Does the effect of hip protectors on force magnitude and pressure distribution depend on (1) the body mass index of the faller? and (2) the orientation of the pelvis at impact?

Research Question 3: Does the protective effect change if a hip protector is displaced relative to the GT?

My specific hypotheses were as follows:

Hypothesis 1: A hip protector will be more effective in reducing the peak force and pressure over the GT for underweight than overweight individuals who have greater natural “padding”.

Hypothesis 2: The force attenuation provided by a hip protector will be greater for falls in the lateral direction, than for falls in the anterolateral or posterolateral directions.

Hypothesis 3: The force attenuation provided by a hip protector will be maximized when the hip protector remains placed in its intended (manufacturer-specified) location, with the GT located at the center of the protector.
CAHPTER 2 EFFECT OF HIP PROTECTORS, FALLING ANGLE AND BODY MASS INDEX ON PRESSURE DISTRIBUTION OVER THE HIP REGION DURING SIMULATED FALLS

2.1 Abstract

Introduction Hip protectors are recommended for preventing fall-related hip fractures in the elderly. In the current study we examined how hip protectors affect the magnitude and distribution of force to the hip during simulated falls, and how this depends on the direction of the fall and the amount of soft tissue padding over the hip.

Methods Fourteen young women with either high or low Body Mass Index (BMI) participated in a “pelvis release experiment” that simulated falls resulting in either lateral, anterolateral or posterolateral impact to the pelvis. For each configuration we measured magnitude of peak pressure, location of peak pressure (d, Θ with respect to the GT), total impact force, and percent force applied to each of four different hip regions. Simulated falls were repeated for the unpadded condition and while wearing a commercially available soft shell hip protector.

Results The soft shell hip protector reduced peak pressure by 70% on average, over all tested conditions. The protective effect was twice greater in low BMI than high BMI individuals, and shunted the peak pressure distally along the shaft of the femur (d = 52 ± 22 mm, Θ = -21 ± 49° in the unpadded trials vs d = 81 ± 23
mm, $\Theta = -10 \pm 35^\circ$ in the padded trials). Wearing the hip protector had no effect on the peak total force. Peak force averaged 12% greater in posterolateral, 17% lower in anterolateral than lateral falls.

**Discussion** Our results indicate that the soft shell hip protector we tested had a much stronger protective benefit for low BMI than high BMI individuals. Next generation protectors might be developed for improved shunting of pressure away from the proximal femur, improved protection during posterolateral falls, and greater force attenuation for low BMI individuals.

### 2.2 Introduction

Hip fractures are an enormous public health problem for the elderly. Ninety percent of hip fractures are caused by falls. An estimated 1.3 million hip fractures occurred worldwide in 1990 [Parker 2006]. Approximately 20% of older adults hospitalized for hip fracture die within a year, and about 50% suffer a major decline in independence [Empana 2004, Wolinsky 1997]. Fracture risk increases exponentially with age, and given the aging of the population, the global incidence of hip fracture is projected to increase 4-fold to 6 million annual cases by 2050 [Gullberg 1997]. Health care costs for hip fractures are estimated at $7-10 billion annually in the US, and 1.3 billion in Canada [Goeree 1996, Lappe 1998].

Hip protectors represent a promising strategy for preventing hip fractures. They are intended to reduce impact force at the greater trochanter by shunting the force onto the surrounding soft tissues, or by absorbing energy [Robinovitch}
[1995, Wiener 2002, Parkkari 1995, Laing 2006]. Robinovitch et al. reported that total force at the femoral neck was attenuated by 68% by an energy-shunting hip pad. Parkkari et al. reported that an energy-shunting external hip protector attenuated the peak femoral force to below the theoretical fracture threshold (4170 N). In a simulated fall experiment, Wiener et al. asked standing participants to fall sideways on a hard surface while wearing a hip protector. A small piezoelectric film sensor was placed between the hip protector and the skin surface over the hip. They found that only 5% or less impact force was transmitted to the skin sensor. However, the sensor was too small, not covering the whole surface of the protector and therefore did not adequately measure distribution of force or pressure over the contact area. Recently, Laing et al. tested soft shell protectors with human subjects, and found that mean pressure over the GT was reduced by 76% by a 14 mm thick horseshoe shaped protector and by 73% by a 16 mm thick continuous protector. However, their measurements provided only the average pressure over circular areas centered at the GT, and not the exact location of peak pressure. In the current study, we obtained high speed, high resolution maps using a RSscan device to gain new insight on the pressure distribution profile and the benefit of hip protectors.

It is known that people with high body mass index (BMI, weight/height^2) have lower risk for hip fracture in a fall than people with low BMI [La Vecchia 1991] even though higher impact force are often generated by high BMI individuals. One possible reason for this is that individuals with high BMI are likely to have a thicker layer of fat tissue over the greater trochanter, which
provides mechanical shock absorption during a fall [Robinovitch 1995, Lauritzen 1992]. However, no studies have investigated the effect of BMI on pressure distribution during falls with or without a hip protector. We hypothesized that individuals with low BMI would draw relatively greater benefit from pressure reducing hip protectors than individuals with high BMI who have higher baseline padding.

The effect of impact direction on hip fracture risk was examined by Keyak et al. [2001] and Pinilla et al. [1996], who reported that the failure load of the cadaveric femur decreased by 24% as the loading angle changed from lateral to 30 degrees posterolateral, indicating that a greater danger for hip fracture is posed by posterolateral falls. Nankaku et al. [2005] measured impact force and velocity of the greater trochanter during simulated falls in the posterior, posterolateral, and lateral directions, and concluded that posterolateral falls carry the highest risk of hip fracture. These results collectively suggest that posterolateral falls create high risk for hip fracture. However, most hip protectors were designed to protect primarily against sideways falls. An important question is whether they also reduce impact force and redistribute pressure in posterolateral falls.

Against this background, we studied human subjects during falling experiments onto pressure profile sensors and examined how a soft shell hip protector affects the distribution of pressure, and how the protective effect depends on BMI and falling impact configuration.
2.3 Methods

2.3.1 Subjects

Fourteen young women between the ages of 18 and 35 participated. We included only women because older women experience hip fracture more often than men. We excluded individuals with musculoskeletal problems such as arthritis, thoracic outlet syndrome, rotator cuff tears, contracture, sprain, and strain. We measured individuals’ weight, height, and hip girth. Height ranged from 160 to 172 cm. One-half of participants were overweight, possessing a body mass index (BMI) (weight/height^2) greater than 25, and the other half were underweight, having a body mass index less than 18.5. All participants had read and signed a written study information and informed consent form. The study protocol and consent form were approved by the Committee on Research Ethics at Simon Fraser University.

2.3.2 Equipment

During trials, we collected total hip impact force from a force plate (Bertec, model 4060H) and pressure distribution from a RSscan device (RSscan International, surface dimension: 40cm by 60cm) placed on the force plate, at a 500 Hz sampling rate. The RSscan plate had 4096 pressure sensors in 64 by 64 array, that measured pressure with a resolution of 0.01 kPa, a range of 3 to 1270 kPa, and accuracy (maximum error between the actual applied pressure and the value measured by the RSscan plate) of 0.37 kPa, based on in-house calibration.
Reflective markers were placed directly on the skin over the right and left greater trochanters (GT), right and left anterior superior iliac crests (ASIS), sacrum, right posterior inferior iliac crest, left knee, left anterior thigh, and left lateral thigh. The 3D positions of these markers were monitored at 250 Hz with an eight-camera video-based motion measurement system (Eagle System, Motion Analysis Corp.). These marker data were used to construct a virtual left GT marker for use in data analysis, since participants were required to remove the left GT marker just prior to each fall.

2.3.3 Protocol

Sideways falls were simulated by releasing the participant from a state of impending impact with the GT raised 5cm above the ground. In the pelvis release experiment, we positioned the subject in a left sideways position with weight supported by the left lower leg and left forearm placed on a mat and with the pelvis cradled in a sling that contacted the upper thigh and the lower rib cage, but not the left greater trochanter and iliac crest (Figure 2-1). A wire cable was attached to the sling at one end and to an electromagnet (model DCA 600-1101; Automatic Equipment Corp., Cincinnati, OH) on the ceiling, and raised the pelvis until a 5 cm gap was measured between the skin surface over the GT and the RSscan plate for the unpadded trials, and between the surface of hip protectors and the RSscan plate for the padded trials. The electromagnet released suddenly, causing the subject to fall onto the RSscan plate. Trials were acquired for three different pelvis orientations: (a) direct impact over the GT, (b) 20°
anterior tilt about the long axis of the body and (c) 20° posterior tilt. Trials were repeated with no hip protector and with a commercial soft shell hip protector (Hipsaver Inc. www.hipsaver.com, Figure 2-2c). Three trials were acquired for each condition. The order of presentation of the conditions was randomized.
Figure 2-1 Schematic of "pelvis release experiment".
2.3.4 Data Analysis

Our main outcome variables were the magnitude of peak pressure, location of peak pressure, total peak force and integrated force applied to each of four different hip regions. Data analysis was conducted with customized Matlab routines. Magnitude of peak pressure was determined by the peak value from the pressure curve over time, where the maximum pressure values from 4096 pressure sensors in the RSscan plate were plotted as a function of time (Figure 2-3a). The location of peak pressure with respect to the GT was expressed by the angle ($\Theta$) from the diaphysis and the distance ($d$) from the GT (Figure 2-2a). We defined four U-shaped regions oriented along the femoral diaphysis and centered at the GT, and named the central area (A) the 'danger zone' since it projected over the femur (Figure 2-2b). We calculated the integrated force applied to each region as the sum of all pressure values measured by each sensor multiplied by the area of the each defined region, and computed percent force defined by the ratio of the integrated force to total integrated force. To determine an anatomical mapping of pressure applied to the hip region, we transformed the coordinate system of the MAC motion analysis system into that of the pressure data from RSscan. This allowed us to estimate the location of the left GT, left ASIS, and orientation of the femoral diaphysis on the RSscan map of pressure distribution (Figure 2-3). The accuracy of this technique, determined by the average X and Y difference between the location of a rigid reflective marker and the peak pressure measured by the RSscan plate when applying pressure to the plate with a rod having the rigid marker at the end, was -1.3 mm in the X axis.
and -0.1 mm in the Y axis. We normalized our outcome variables in order to compare the high and low BMI groups. Peak pressure was multiplied by the product of \((\text{body height (cm)} \times \text{hip girth (cm)}) / \text{body weight (N)}\), force by body weight (N), and distance by body height (cm).

Repeated measure ANOVA was used to test whether the outcome variables were associated with BMI (2 levels), impact orientation (3 levels), and hip protector condition (2 levels).
Figure 2-2 a: The location of peak pressure was expressed as the distance (d) from the greater trochanter and angle (θ) from the diaphysis. b: Four different areas were defined over the hip region; area A (danger zone, light gray) comprised a half-circle of 2.5cm radius centered at the GT plus a rectangle extending distally from the GT, and areas B, C, and D (darker gray) are comprised of C-shaped hollow regions of 10 cm, 15 cm, and 20 cm widths, respectively. c: 16 mm thick soft shell hip protector pad (Hipsaver, 20 cm long and 17 cm width).
Figure 2-3 Pressure profiles in the unpadded and padded condition (all for a single trial). a: Pressure peaked approximately 25 ms after the start of impact in the unpadded condition. We looked at the exact time frame (n) where peak pressure occurred. b: For a low BMI participant pressure aggregated and centered over the GT and the contact area was small in the unpadded condition. c: In the padded condition with the same low BMI participant, pressure was distributed, contact area increased and peak pressure moved away from the GT with decreased magnitude. d: For a high BMI participant in the unpadded condition, peak pressure occurred over the GT and contact area was larger compared to the low BMI participant in the same condition. e: In the padded condition with the same high BMI participant, pressure was fairly evenly distributed over a much larger contact area, and the peak pressure was of lower magnitude and shunted away from the GT. The magnitude of pressure is indicated by a color scale. Pressure unit on a color bar is N/cm^2 (1 N/cm^2 = 10 kPa).
2.4 Results

Peak pressure was associated with hip protector condition, impact configuration, and BMI. On average across all conditions, peak pressure decreased 70% by use of the protector ($F = 13.7$, $p = 0.003$). Peak pressure averaged 38% lower in anterolateral than lateral falls ($F = 4.9$, $p = 0.03$), and 266% higher in low BMI than high BMI participants ($F = 6.7$, $p = 0.024$) (Figure 2-4). Furthermore, there was a significant interaction between hip protector condition and BMI ($F = 7.8$, $p = 0.016$), reflecting a greater protective effect for low BMI than high BMI participants. For example, in the lateral impact configuration, the decrease in peak pressure in the padded versus unpadded conditions was 77% (from 754 to 169 kPa) in low BMI participants, and 48% (from 370 to 180 kPa) in high BMI individuals (Table 2-1 and 2-2). Similar trends were observed in the posterolateral and anterolateral impact configurations.

The location of peak pressure was also associated with hip protector condition ($F = 67.3$, $p < 0.001$ for distance, $F = 7.3$, $p = 0.02$ for angle), with BMI ($F = 5.2$, $p = 0.042$ for angle), and with impact configuration ($F = 21.5$, $p < 0.001$ for angle). Average peak pressure location across all conditions was 52 mm from the GT and $-20.9^\circ$ from the diaphysis in unpadded, and 81 mm from the GT and $-9.9^\circ$ in padded conditions (Figure 2-5). Furthermore, there was a significant interaction between protector and BMI ($F = 8.4$, $p = 0.013$ for distance, $F = 19.8$, $p = 0.001$ for angle), with the hip protector redirecting peak pressure, on average, from 54 mm to 76 mm away from the GT and from $-2.6^\circ$ to $-10.6^\circ$ away from the diaphysis in low BMI individuals, and from 51 to 86 mm and from $-39.1^\circ$ to $-9.2^\circ$
in high BMI individuals. Angle from the diaphysis averaged +4.8° in anterolateral, -18.4° in lateral, and -32.7° in posterolateral falls.

Total peak impact force was associated with impact angle, averaging 12% greater in posterolateral, 17% lower in anterolateral than lateral falls (F = 52.3, p = 0.001), but was not associated with BMI or hip protector condition.

The percent force on the danger zone was associated with impact configuration, averaging 4% greater in anterolateral, 55% lower in posterolateral than lateral falls (F = 11.7, p < 0.0005), but was not associated with BMI or hip protector condition. (Table 2-1 and 2-2).
Figure 2-4 Peak pressures were reduced when a hip protector was used. The protective effect was much stronger for low BMI than high BMI women.
Figure 2-5 The location of peak pressure averaged $52 \pm 22$ mm from the GT and $-20.9 \pm 49^\circ$ from the diaphysis in the unpadded, and $81 \pm 23$ mm from the GT and $-9.9 \pm 35^\circ$ from the diaphysis in the padded condition across all trials.
2.5 Discussion

The soft shell hip protector we tested substantially reduced peak pressure on the GT, and redirected the location of peak pressure distally along the diaphysis. However, it had no effect on total applied force. This suggests that the tested protector had little influence on total stiffness, or consequently total impact force, while causing a substantial change in local stiffness and pressure redistribution. The same mechanical analysis explains our observation that BMI was associated with peak pressure, but not peak force. Laing et al. [2006] reported that peak total force was reduced by 19% by a continuous soft shell hip protector. This discrepancy may be due to different impact configurations. In their experiments, participants lay sideways with trunk horizontal, with their arm extended over the head, while in our experiment the participants' trunk was laterally flexed. This difference in body position may have led to differences in the effective stiffness of the body, and thus in the effect of the hip protector on peak force.

We hypothesized that soft shell Hip protectors would be more effective in reducing peak pressure over the hip region for low BMI individuals, who would tend to have less natural “padding” to act as shock absorber. We found that the soft shell hip protector provided more than twice the peak pressure reduction in low BMI than high BMI participants, and more appropriately shunted pressure away from the GT and diaphysis for low BMI than high BMI individuals, resulting in alleviation of considerable force applied to the proximal femur. These findings collectively indicate that the tested soft shell hip protector has a much stronger
protective effect for low BMI individuals, whereas high BMI individuals have the benefit of substantial natural padding.

We also hypothesized that the efficacy of the hip protector would depend on fall direction. We found that impact angle affected peak pressure and total force, but the hip protectors’ effectiveness did not change with direction of fall. The soft shell hip protector moved the location of peak pressure distally along the proximal femur’s diaphysis by 29 mm, but not away from the bone. Despite its effect on the magnitude of peak pressure, the location of peak pressure and the distribution of force from the danger zone to adjacent regions, wearing the soft shell hip protector had no effect on total peak force. Finally, posterolateral falls generated higher impact forces and it is shown that for such fall angles, the fracture strength of the femur is lower (Pinila 1996, Keyak 2001, Nankaku 2005). Therefore, it is desirable that next generation hip protectors be designed to shunt more force away from the bone for improved protection against posterolateral falls.

In summary, the tested soft shell hip protector reduced peak pressure and redistributed force away from the danger zone more strongly and appropriately in low BMI than high BMI participants. It redirected the location of peak pressure distally along the femoral shaft, caused no change in total force, and provided little protection in posterolateral falls. These drawbacks have to be considered by the next generation hip protector for better hip fracture prevention.
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Table 2-1 Average normalized values in each condition (values in parenthesis are absolute values).
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*significant at 0.05

Table 2-2 ANOVA table
CHAPTER 3  PROTECTIVE EFFECTS OF HIP PROTECTORS DEPEND ON POSITION OF THE PROTECTOR DURING A FALL

3.1 Abstract

*Introduction* Hip protectors are designed to attenuate and redistribute the force applied to the hip region during a fall, and thereby reduce risk for hip fracture. However, little information exists on the effectiveness of hip protectors in achieving these goals, and how this is altered by displacement of the hip protector relative to the GT.

*Methods* We used a hip impact simulator consisting of an impact pendulum and surrogate pelvis, to simulate falls from three different heights (5 cm, 10 cm, and 20 cm). Trials were acquired without a hip protector, and with three different soft shell hip protectors located in their intended location, displaced by 2.5 cm or displaced by 5 cm in the superior, posterior, inferior or anterior directions. Outcome variables were the magnitude and location of peak pressure (measured from an RSscan plate), the force distributed to four different areas over the hip region, and the peak trochanteric force.

*Results* Peak pressure (and trochanteric force) was reduced by 78 % (38 %) with the 14 mm horseshoe, by 83 % (45 %) with the 16 mm horseshoe, and by 88 % (30 %) with the 16 mm continuous protector. The protective effect decreased by 34 % (29 %) when the protectors were displaced 5 cm in the anterior direction.
The location of peak pressure was shunted outside the danger zone by the 16 mm horseshoe protector when it was centrally placed (from \( d = 4.6 \text{ mm} \), \( \Theta = 36.8^\circ \) to \( d = 76.8 \text{ mm} \), \( \Theta = 20.9^\circ \)), but not when it was displaced. In the unpadded condition, 83% of the total force was applied to the danger zone, but only 34% and 19% with the 14 mm and 16 mm horseshoes, and 40% with the 16 mm continuous protector. However, the protective effects decreased with displacement.

Discussion Hip protectors showed protective effects, and the 16 mm thickness horseshoe protector was best. But the benefit was reduced by protector displacements as small as 2.5 cm. Our findings indicate a need for improved designs of hip protectors and methods for correct placement.

3.2 Introduction

Hip fracture is a common, serious injury for the elderly, and over 90% of hip fractures are caused by falls [Grisso 1991]. Hip protectors are promising wearable devices that cover the greater trochanter (GT) and proximal femur, and are intended to reduce the risk for hip fracture by attenuating impact forces applied to the femur by either absorbing energy or shunting energy to the surrounding tissues. Previous biomechanical studies indicate that hip protectors can attenuate the impact force at the skin surface over the hip by 91% [Wiener 2002], and reduce peak force at the femoral neck by between 42% and 89% during a simulated severe fall, resulting in forces below the average fracture threshold (3100 N ± 1200 N) of elderly women with a mean age of 71 years [Robinovitch 1995, van Schoor 2006, Parkkari 1995, Kannus 1999]. However, it

Furthermore, little attention has been directed towards understanding how hip protector positioning influences biomechanical performance. Most hip protectors are designed to protect the GT from external impact during a fall, and the protective effect is maximized when the center of the hip protector pad is located over the GT. However, the pad may not always remain centered at the GT during activities of daily living either because the pad itself may shift, or because the location of the GT can shift depending on the orientation of the femur. Minns et al [2007] recognized this issue and reported that the center point of a hip protector was displaced by 0.64 cm from its initial position on the skin when the hip rotated from full extension to 90° of flexion. Derler et al [2005] conducted biomechanical testing, and found that a 3 cm displacement of the protector in the anterior direction caused the total impact force and trochanteric force to increase 9 % and 23 % with a hard shell hip protector, and to decrease 4 % and 11 % with a soft shell hip protector.
In the present study, we investigated how hip protectors redistributed pressure applied to the hip region through a standardized biomechanical test using a hip impact simulator that enabled us to test greater drop heights and more conditions than would be possible with human subjects. We evaluated the extent to which benefits provided by hip protectors depend on protector positioning as well as the design of the protector. Based on our results, we discuss here improved designs for next generation hip protectors.

3.3 Methods

3.3.1 protocol

Biomechanical impact tests were conducted with the Simon Fraser University hip impact simulator (Figure 3-1a and 3-1b). The SFU hip impact simulator was designed to measure total force applied to the surface of a surrogate hip and trochanteric force transmitted to the femoral neck during simulated falls [Laing 2008]. It is composed of an impact pendulum and a surrogate hip that mimics the hip geometry and distribution in soft tissue stiffness of typical elderly women. The surrogate hip was dropped onto a dual arrangement of an RSscan plate (RSscan International, surface dimension: 40cm by 60cm) and a force plate (Bertec, Type 2535-08) from fall heights of 5 cm, 10 cm and 20 cm (that produced the maximum possible impact force within the RSscan measurement range). Trials were acquired without a hip protector, and with three different soft shell hip protectors: 14 mm and 16 mm thick horseshoe-shaped pads (SafeHip, Tytext A/S), and a 16 mm thick continuous pad.
(Hipsaver) (Figure 3-1c and 3-1d). Each pad was taken from a hip protector garment and secured to the surrogate hip by means of double sided tape prior to the impact test. For each drop height, each protector was tested in nine positions: located centrally in its intended location over the GT, and displaced by either 2.5 cm or 5 cm in the superior, posterior, inferior, and anterior directions (Figure 3-1e ~ 3-1h). Three trials were acquired for each condition.

During each trial, we collected total hip impact force from the force plate, pressure distribution from the RSscan device and trochanteric force from a load cell (Kistler, Type 9712B5) mounted beneath the femoral component of the surrogate pelvis. All measures were acquired with a 500 Hz sampling rate. The RSscan plate had 4096 pressure sensors (64 by 64 array), a resolution of 0.01 kPa, range of 3 to 1270 kPa and accuracy (maximum error between the actual applied pressure and the value measured by RSscan plate) of 0.37 kPa, based on in-house calibration.
Figure 3-1 Biomechanical testing with hip impact simulator. a: SFU hip impact simulator. b: cross sectional diagram indicating placement of load cell. c: 16 mm continuous soft shell hip protector pad (Hipsaver, 20 cm long and 17 cm width). d: 14 mm or 16 mm horseshoe pad (both from SafeHip, 17 cm long and 17 cm width). e: schematic of nine defined hip areas (unpadded condition). f: centrally placed horseshoe protector. g: displaced 5 cm in the superior direction. h: displaced 2.5 cm in the posterior direction.
3.3.2 Data Analysis

Our main outcome variables were the magnitude of peak pressure, location of peak pressure, trochanteric force and force applied to four defined hip regions. Data analysis was conducted with customized Matlab routines. Magnitudes of peak pressure and trochanteric force were taken as the peak values recorded from the pressure array and femoral load cell during the impact event, respectively. The location of peak pressure with respect to the GT was expressed as the distance (d) from the GT and the angle (θ) from the femoral diaphysis. We defined four C-shaped regions over the hip centered about the GT, and named the central area (area A) the ‘danger zone’ since it projected over the GT and femoral diaphysis (Figure 3-2). We anticipated that the effective action of each hip protector would be to reduce the force applied on this danger zone. We calculated the integrated force applied to each region by summing the product of the sensor area multiplied by the pressure measured by each sensor within the area of interest.

We determined the location of the GT and the orientation of the femoral diaphysis on the RSscan coordinate system by removing the soft tissue cover on the surrogate pelvis, and measuring the pressure distribution while slowly lowering the pendulum so contact with the RSscan, occurred at only two locations: the GT and a pin temporarily inserted along the midline of the femoral diaphysis.

For statistical analysis, we regarded the three repeated trials in each condition as three different people’s trials in each condition and assumed
different people across all conditions, and used randomized group ANOVA to test whether each of our outcome variables was associated with drop height (3 levels), hip protector type (4 levels), and pad displacement (9 levels). The significance level in all tests was set to $\alpha = 0.05$, and all analysis were conducted in SPSS 16.0.
Figure 3-2 Pressure distribution profiles with centrally placed hip protectors and 20 cm fall height. **a:** definition of four different areas over the hip region: area A, danger zone (gray) consisted of a C-shaped region of width 5 cm (sum of half-circle of radius 2.5 cm centered at GT and rectangle extending distally from the GT); areas B, C, and D (darker gray) consisted of C-shaped hollow regions of width 10 cm, 15 cm, and 20 cm each. **b:** In the unpadded condition, pressure concentrated at the GT within the danger zone and the contact area was small. The magnitude of pressure is indicated by color scale at right. Pressure calibrations on color bars are in N/cm² (1 N/cm² = 10 kPa). **c:** With the 16 mm continuous protector, the pressure was fairly evenly distributed over the hip region and the contact area was large, but considerable pressure was still applied to the danger zone. **d:** With the 14 mm horseshoe protector, the pressure was unevenly distributed and the contact area was large. Whereas much of the pressure was deflected away from the danger zone, considerable pressure was still applied directly on the GT. **e:** With the 16 mm horseshoe protector most of the pressure was redistributed away from the danger zone, over a large contact area covering soft tissues. Note that essentially no pressure was applied directly onto the GT.
3.4 Results

The magnitude of peak pressure was associated with drop height (p < 0.0005), hip protector type (p < 0.0005) and hip protector displacement condition (p < 0.0005). On average, peak pressure was reduced by 78\% with the 14 mm horseshoe pad (from 1660 kPa to 370 kPa), by 83\% (to 280 kPa) with the 16 mm horseshoe pad, and by 88\% (to 200 kPa) with the 16 mm continuous pad (Figure 3-3a). The peak pressure was on average 34\% greater for hip protector displacements of 5 cm in the anterior direction, compared to centrally placed pad location (716 kPa vs 535 kPa) (Figure 3-3b).

The location of peak pressure was associated with drop height (p < 0.0005 for both distance and angle), hip protector type (p < 0.0005 for both distance and angle) and hip protector displacement condition (p < 0.0005 for both distance and angle). For example, for 20 cm drops, the location of peak pressure in the unpadded condition was on average 4.6 mm from the GT and 36.8° from the diaphysis. The peak pressure moved closer to the GT by hip protector displacements (Figure 3-3c and 3-3d), and was shunted outside the danger zone by the 16 mm horseshoe protector centrally placed (d = 76.8 mm, Θ = 20.9°), but not by the centrally placed 14 mm horseshoe or 16 mm continuous protector (d = 4.6 mm, Θ = 36.8° for both protectors) (Figure 3-4).

The integrated force applied to the danger zone was associated with drop height (p < 0.0005), hip protector type (p < 0.0005) as well as hip protector displacement condition (p < 0.0005). For 20 cm drops, 83\% of the total force
was applied to the danger zone in the unpadded condition, but the percent force was reduced to 34% and 19% with 14 mm and 16 mm horseshoe protectors, and to 40% with the 16 mm continuous protector (Figure 3-5a). The force reduction to the danger zone was greatest when the 16 mm horseshoe hip protector was placed centrally, and lowest with 5 cm displacement in the anterior direction (% force reduction of 58% vs 18%). This trend was observed with the 14 mm horseshoe, but not with the 16 mm continuous protector (Figure 3-5b).

The force distribution to areas B, C and D was also associated with fall height ($p < 0.0005$), hip protector type ($p < 0.0005$) and hip protector displacement condition ($p < 0.0005$). For 20 cm drops, hip protectors redistributed force applied to the hip region by lowering and delivering the force applied to the danger zone onto adjacent areas B, C and D, and this effect was reduced by displacements of the hip protector (Figure 3-6).

Finally, trochanteric impact force was associated with fall height ($p < 0.0005$), hip protector type ($p < 0.0005$) and hip protector displacement condition ($p < 0.0005$). For 20 cm falls with hip protectors centrally placed, trochanteric force averaged 45% lower with the 16 mm horseshoe protector, 38% lower with the 14 mm horseshoe, and 30% lower with the 16 mm continuous protector, compared to the unpadded condition (Figure 3-7). The trochanteric force was 29% higher for 5 cm displacement in the anterior direction, compared to centrally placed pad. There was a significant interaction between hip protector and displacement, indicating that both the 14 mm and the 16 mm thick horseshoe
protectors outperformed the 16 mm continuous protector in all but 3 conditions of displacements.
Figure 3-3 a: Peak pressure reduction provided by three hip protectors. The magnitude of peak pressure applied to the hip region was reduced by all these hip protectors. Hip protector displacement effect on the magnitude (b) and location (c and d) of peak pressure across all three hip protectors (all for 20 cm fall height). With displacement of the protector, the magnitude of peak pressure increased and the location of peak pressure moved towards the GT.
Figure 3-4 Changes in the location of peak pressure as function of hip protector types (all for 20 cm fall height). Peak pressure was shunted away from the danger zone by the 16 mm horseshoe protector when centrally placed (b), but with both 14 mm horseshoe (a) and 16 mm continuous protectors (c) the peak pressure remained within the danger zone and did not change.
Figure 3-5 a: Force distribution to four different areas over the hip (1). For a 20 cm fall height, the force applied to the danger zone was 83% of the total force in the unpadded condition, but only 19% of the total force with the centrally placed 16 mm horseshoe protector. b: Percent force reduction in the danger zone provided by hip protectors with respect to eight displaced locations (averaged across all fall heights). The % force reduction was defined by subtracting the percent force applied to the danger zone in each condition from the percent force applied to the danger zone in the unpadded. For a 20 cm fall height, for example, with the 16 mm horseshoe protector placed centrally the % force reduction in the danger zone would be 64% = 83% - 19%. There was a reduction of 64% in the force applied to the danger zone when the 16 mm horseshoe protector was located centrally over the GT, but only 22% when the protector was displaced 5 cm anteriorly.
Figure 3-6 Force distribution to four different areas over the hip (2). a: Hip protectors lowered the force applied to the danger zone and delivered the force onto adjacent areas B, C and D. The force distribution was affected by displacement of the (b) 16 mm continuous (c) 14 mm horseshoe and (d) 16 mm horseshoe protector, respectively. The force distribution was greatest when hip protectors were placed centrally and decreased when displaced. The horseshoe protectors were more influenced by the displacements than the continuous protector.
Figure 3-7 Effect of the protector positioning on the trochanteric force (20 cm drop height). On average, the trochanteric force was reduced 38% by the 14 mm horseshoe protector, 45% by the 16 mm horseshoe protector, and 30% by 16 mm continuous protector, and there was a significant interaction between displacement and protector type, indicating that the 14 and 16 mm horseshoe protector were influenced more by displacement than the 16 mm continuous protector. The 16 mm continuous protector performed better when displaced 2.5 cm and 5 cm in the inferior direction compared to centrally placed.
3.5 Discussion

The soft shell hip protectors we tested using our hip impact simulator reduced the magnitudes of peak pressure, peak trochanteric force and the percent of total force applied to the danger zone overlying the proximal femur, and shunted the location of peak pressure away from the GT, in nearly all of the conditions we examined. These findings provided direct biomechanical evidence to support the notion that by wearing any of these soft shell hip protectors, risk for fracture is reduced. While the 16 mm continuous protector caused the greatest overall reduction in peak pressure, both 14 mm and 16 mm horseshoe protectors provided even better reduction in the trochanteric force and percent force to the danger zone when compared to the continuous protector, and were more effective in shunting the location of peak pressure away from the GT. Furthermore, the 16 mm horseshoe provided approximately 4 % more attenuation in trochanteric force, 8 % more attenuation in danger zone force than the 14 mm across all conditions, and the location of peak pressure was shunted out of the danger zone with the 16 mm horseshoe, but not with the 14 mm horseshoe protector. These collective observations suggest that the horseshoe protectors are better than the continuous protector in reducing risk for hip fracture, and the 16 mm thick version provides better protection than the 14 mm horseshoe design.

The protective benefit of the 16 mm horseshoe protector tested declined when the protector was displaced away from the GT, with increases in the peak pressure, the peak trochanteric force and the percent of total force applied to the
danger zone, and with decreases in distance between the location of peak pressure and the GT. Similar trends were observed for both the 14 mm horseshoe and the 16 mm continuous protector in most cases. For example, for the 14 mm horseshoe protector the location of peak pressure was shunted away from the GT when displaced 5 cm in the superior direction and 2.5 cm in the superior and inferior direction, but not when centrally placed (Figure 3-3c). Similarly, for the 16 mm continuous protector the peak pressure location moved away from the GT when displaced 2.5 cm in the inferior direction, and the percent force reduction to the danger zone increased with displacements (Figure 3-5b), and the peak trochanteric force decreased when displaced 5 cm in the inferior direction and 2.5 cm in the posterior and inferior direction. Furthermore, when both the 14 mm and 16 mm horseshoe protectors were displaced by 5 cm in the anterior direction the trochanteric force actually exceeded the value measured in the unpadded condition.

Derler et al. [2005] tested ten different hip protectors by placing them correctly on a hip model, and by displacing by 3 cm in the anterior direction in order to investigate protector displacement effect on the efficacy of the protector, and reported that with a protector displacement the peak trochanteric force decreased in three hip protectors, and increased in seven hip protectors. In the present study, the trochanteric force increased when the 14 mm and 16 mm horseshoe protectors were displaced away from their intended position, and for the 16 mm continuous protector the trochanteric force increased with all but three displacement conditions.
These results collectively indicate that the effectiveness of the horseshoe shaped hip protectors is relatively more influenced by displacement of the protector from their intended position than for the continuous hip protector.

Hip protector garments are designed to place and hold protective pads over the GT, and their protective effect is maximized when the protector is located in its intended position, especially a horseshoe shaped protector. In practical use, however, the optimum position may not be achieved, so the effectiveness of hip protectors can be compromised due to displacement of the protector. Protector displacement can be caused by either poor fit or movement of the garment over the skin surface during daily activities. Minns et al. [2007] reported that the position of hip protectors relative to the GT was highly variable, and the protector pad was displaced posteriorly 0.45 cm from its initial position due to walking and sitting, and 0.64 cm when the hip was flexed from full extension to 90 degrees of flexion. While these displacements were small when compared to displacements used in this study, in practice protector displacement can become much greater when combined with displacement resulting from poor garment fit. Protector displacements can also be caused by movement of the GT under the skin with joint rotation. The GT moves depending on orientation of the femur, and certain amount of hip flexion and internal/external rotation from its initial position when standing is expected at the impact stage of a fall, resulting in protector displacement when the impact force applied to the hip region.

In summary, all the soft shell hip protectors we tested showed protective effects against external impact when optimally positioned in their intended
location over the proximal femur. However, the horseshoe shaped protectors we
tested provided superior protective benefit when compared to the continuous
protector, and the 16 mm thick version was better than 14 mm between the
horseshoe protectors. Furthermore, the protective effect was strongly dependent
on correct placement of the protector with respect to the GT. Our findings are
informative for developing more efficacious hip protectors and garments.
CHAPTER 4  THESIS CONCLUSIONS

The aim of this thesis was to examine how hip protectors redistribute the force applied to hip region during a fall, and to test how the protective benefit provided by a protector is affected by the wearer’s BMI, the fall direction, and the position of the protector relative to the GT. In falling experiments with humans, I found that the soft shell hip protectors we tested substantially reduced the peak pressure and the force applied to the “danger zone” (i.e. the skin over the proximal femur). I also found in experiments with a hip impact simulator, which were done for larger fall heights than possible with human subjects, that hip protectors reduced the peak force delivered to the femoral neck, and shunted the peak pressure away from the GT. These findings provide insight on how hip protectors may protect against bone fracture during a fall.

I also found that the benefit provided by hip protectors is greater for low BMI than high BMI individuals who have greater natural ‘padding’, and is usually maximized when the protector is placed centrally rather than displaced from the GT. Furthermore, posterolateral falls generate higher impact force (and no greater attenuation in force by hip protectors), which, combined with the known lower fracture force of the proximal femur in this direction, probably account for the higher risk for hip fracture in posterolateral falls than anterolateral and lateral falls.
I also found that, when compared to a continuous shaped protector, a horseshoe-shaped soft shell hip protector of equivalent (16 mm) thickness provided greater force attenuation and shunting of force away from the proximal femur. Of particular note was the tendency of the continuous protector to shunt force distally from the GT but along the femoral diaphysis.

To understand the role of BMI, future work should investigate the relative contributions of muscle, fat and skin tissue layers over the hip region to force attenuation and pressure distribution during a fall. This is challenging research that will likely require a combination of mathematical modeling and experimental measures (perhaps involving imaging based on magnetic resonance or ultrasound). Future work is also required to carefully examine how hip joint rotations affect the position of the proximal femur relative to the skin and a wearable hip protector. Finally, work is required to develop next-generation hip protectors and garments that provide (a) improved force shunting away from the diaphysis, (b) better protection for posterolateral falls, and (c) greater force attenuation for people who have minimal soft tissue padding. This will probably require a combination of changes in material, thickness, surface geometry (perhaps optimizing the horseshoe geometry), and garment design.
APPENDICES

A Accuracy of the RSscan Results

In order to confirm how accurately the RSscan measures pressure we conducted the following test.

Measures were acquired by placing four different weights (20 lb, 40 lb, 85 lb, and 125 lb) on a flat aluminum disc of surface area 78.5 cm$^2$ on the RSscan in three configurations (with the disc located centrally, 10cm to the right of the midpoint, and 10cm of the left of the midpoint). Three repeated trials were acquired for each condition.

We found that the RSscan underestimated the mean pressure in all trials (Figure A-1). Measured pressure did not vary appreciably between the three disc locations. A linear scaling factor was calculated based on mean data: corrected pressure $= 1/0.28477 \times$ measured RSscan pressure $- 0.4458$, with a goodness of fit of $R = 0.99$ (Figure A-2).
Figure A-1 Measured mean pressure in response to different loads: (a) 20lb, (b) 40lb, (c) 85lb, and (d) 125lb. A red arrow in each figure represents actual mean pressure applied: 1.12 N/cm², 2.24 N/cm², 4.78 N/cm², and 7.03 N/cm² respectively. RSscan Underestimated actual pressure applied. Error bars show the standard deviation for pressure values measured by pressure sensors over the entire contact area.
Figure A-2 Relationship between applied pressure and measured pressure by RSscan. A scaling factor suggested is 0.285.
B Accuracy of the virtual marker technique and marker mapping technique for locating the GT

In the pelvis release experiments, a challenging aspect of data analysis is to estimate the location of the GT during impact. This can not be measured directly, since the left (impacting side) GT and lateral thigh markers are occluded from camera view and must be removed just prior to the fall, in order not to influence the measured pressure. To estimate the GT location during impact, we track the location of a virtual marker constructed by the Eva software (Motion Analysis Corp.). To verify the accuracy of this technique, motion data were acquired while normal walking. Four markers were placed on the pelvis (right and left ASIS and GT). Trials were acquired in leaving all markers in place and removing the left GT marker shortly after starting of walking. The X, Y, and Z position of the actual and virtual left GT markers were then tracked and compared. We found that the virtual marker technique was very promising to estimate the actual GT position (Figure B-1).
Figure B-1 Accuracy of a virtual marker technique. We tested how accurate the virtual marker position from the MAC system software is compared to actual marker position. The graph shows x, y, z position change of the actual and virtual left GT markers while a subject walks normally. LGT: left greater trochanter.
We also applied localized pressure in separate trials to five spots on the RSscan plate (center, bottom left, top left, top right, and bottom right) by pressing down with a 2.2cm diameter rigid reflective marker while measuring the location of the marker with the MAC system. We then examined the agreement between the known and calculated and how closely the location of the rigid marker matched the location of pressure point on a pressure map.

There were approximately ± 5 mm of maximum difference between the location of the rigid marker and pressure point in X and Y direction. However, the difference in each trial was quite evenly distributed around ‘zero’ (Figure B-2).
Figure B-2 Localized pressure was applied to five spots over the RSscan plate with a rigid marker; center, bottom left, top left, top right, and bottom right. The graph shows how accurately the calculated position of the rigid marker matches the location of peak pressure; (a) X variation between a rigid marker and pressure point (rigid marker_x - pressure point_x), (b) Y variation between a rigid marker and pressure point (rigid marker_y - pressure point_y), (c) mean of the x variation, (d) mean of the y variation.
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