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Effect of hip protectors, falling angle and body mass index on pressure distribution over the hip during simulated falls

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ABSTRACT

Background: We examined how a soft shell hip protector affects the magnitude and distribution of force to the hip during simulated falls, and how the protective effect depends on the fall direction and the amount of soft tissue padding over the hip.
Methods: Fourteen young women with either high or low body mass index participated in a "pelvis release experiment" that simulated falls resulting in either lateral, anterolateral or posterolateral impact to the pelvis with/without a soft shell hip protector. Outcome variables were the magnitude and location of peak pressure (d, theta) with respect to the greater trochanter, total impact force, and percent force applied to four defined hip regions.
Findings: The soft shell hip protector reduced peak pressure by 70%. The effect was two times greater in low than high body mass index individuals. The protector shunted the peak pressure distally along the shaft of the femur ($d = 52$ mm (SD 22), theta = $-21°$ (SD 49) in the unpadded trials versus $d = 81$ mm (SD 23), theta = $-10°$ (SD 35) in the padded trials). Peak force averaged 12% greater in posterolateral and 17% lower in anterolateral than lateral falls.
Interpretation: Our results indicate that the hip protector we tested had a much stronger protective benefit for low than high body mass index individuals. Next generation protectors might be developed for improved shunting of pressure away from the femur, improved protection during posterolateral falls, and greater force attenuation for low body mass index individuals.

1. 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 (Johnell and Kanis, 2004). Approximately 20% of older adults hospitalized for hip fracture die within a year, and about 50% suffer a major decline in independence (Empana et al., 2004; Wolinsky et al., 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 et al., 1997). Health care costs for hip fractures are estimated at $12.1 billion in 2005, and projected to grow incurring $25.3 billion by 2025 (Burge et al., 2007).

Hip protectors represent a promising strategy for preventing hip fractures. They are intended to reduce impact force at the greater trochanter (GT) by shunting the force onto the surrounding soft tissues, or by absorbing energy. Robinovitch et al. (1995a) reported that total force at the femoral neck was attenuated 68% by an energy-shunting hip pad. In a simulated fall experiment, Wiener et al. (2002) asked standing participants to fall sideways on a hard surface while wearing a hip protector. A piezoeletric film sensor was placed between the hip protector and the skin over the hip. They found that only 5% or less impact force was transmitted to the skin sensor. However, the sensor did not cover the whole surface of the protector and therefore did not measure distribution of force or pressure over the entire contact area. Recently, Laing and Robinovitch (2008a) tested soft shell protectors with human subjects, and found that the mean pressure over the GT was reduced 76% by a 14 mm thick horseshoe-shaped protector and 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 two-dimensional pressure distribution device (RSscan) 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 et al., 1991). One possible reason is

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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 et al., 1995b; Lauritzen et al., 1993). However, no studies have investigated the effect of BMI on pressure distribution during falls with or without a hip protector.

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° posterolateral, indicating a greater danger for hip fracture 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 found that impact force was highest in posterolateral falls. These results collectively suggest that posterolateral falls create high risk for hip fracture. However, most hip protectors are 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. Methods

2.1. Subjects

Fourteen young women between the ages of 18 and 35 participated. We included only women because hip fractures are approximately 3-fold more common in older women than men (Chevalley et al., 2007; Bjorgul and Reikeras, 2007; Lonnroos et al., 2006). We excluded individuals with musculoskeletal problems such as arthritis, thoracic outlet syndrome, or recent rotator cuff tears, contracture, sprain, and strain. We measured individuals' weight, height, and hip girth. Height ranged from 160 to 172 cm. Participants were selected so that one-half possessed a body mass index ($BMI = weight/height^2$) greater than 25, and the other half had a body mass index less than 18.5. Average body weight and height were 47 kg (SD 4) and 162 cm (SD 5) in the low BMI group, and 75 kg (SD 9) and 163 cm (SD 5) in the high BMI group. All participants provided written informed consent. The study protocol and consent form were approved by the Committee on Research Ethics at Simon Fraser University.

2.2. Equipment

During each trial, we collected total hip impact force from a force plate (Bertec, model 4060H, Columbus, OH, USA) and pressure distribution from a 2D scanning plate (RSscan International, surface dimension: 40 cm by 60 cm, Olen, Belgium) placed on the force plate, at a 500 Hz sampling rate. The RSscan plate had 4096 pressure sensors in a 64 by 64 array, that measured pressure with a resolution of 0.01 kPa, a range of 3–1270 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 spine (ASIS), sacrum, right posterior inferior iliac spine (PIIS), 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 and associated software package (Eagle camera system with EVaRT 5.0 software, Motion Analysis Corp., Santa Rosa, CA, USA). We used the "Joint Virtual" tool in EVaRT to construct a virtual left GT marker, since participants were required to remove the left GT marker just prior to impact. Briefly, this technique assumes that the pelvis markers (at the sacrum, and the right and left ASIS, GT and PIIS) form a rigid body with consistent relative distances between markers (established from data collected just prior to release, when all markers are present), allowing estimation of the coordinates of the missing GT from the remaining pelvis markers.

2.3. Protocol

Sideways falls were simulated through "pelvis release experiments," which involved releasing the participant from a state of impending impact with the GT raised 5 cm above the ground. This technique allows for precise control of impact position, and generates peak impact forces that, while safe for our young participants, are within the range of force observed to fracture the elderly femur (Laing and Robinovitch, 2008a). During the trials, we positioned the subject lying on her left side with the shin, lower thigh, forearm and hand contacting the ground, and 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 (Fig. 1). A wire cable attached the sling to an electromagnet (model DCA 600–110I; Automatic Equipment Corporation, Cincinnati, OH, USA) mounted on the ceiling. A turnbuckle located in-line with the cable was used to raise 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 the hip protector and the RSscan plate for the padded trials. The electromagnet was then suddenly released, causing the subject to fall onto the RSscan plate. Trials were acquired for three different impact configurations of the pelvis: (a) direct impact to the lateral aspect of the GT, (b) impact to the pelvis when rotated (about the long axis of the body) 20° anterior to the frontal plane and (c) impact to the pelvis when rotated 20° posterior to the frontal plane. In each configuration, trials were conducted with no hip protector and with a commercial soft shell hip protector (of thickness 16 mm and surface area 20×17 cm; Hipsaver Inc., Canton, MA, USA; Fig. 2c). Three trials were acquired for each condition. The order of presentation of the conditions was randomized. Surgical positioning mats (Vac-Pac, Olympic Medical, Seattle, WA, USA) were placed under the shin and lower thigh, and under the forearm and hand, to ensure consistent positioning of the participant between successive trials in a given impact configuration.

2.4. 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 defined hip regions. Data analysis was conducted with customized Matlab routines. The 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 (Fig. 3a). The location of peak pressure with respect to the GT was expressed by the angle (θ) from the diaphysis and the distance (d) from the GT (Fig. 2a).

We defined four 2.5 cm wide 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 (Fig. 2b). We calculated the integrated force applied to each region as the sum of all pressure values measured by the corresponding sensors multiplied by the area of the each defined region, and computed percent force defined by the ratio of the region's integrated force to total integrated force. To determine an anatomical mapping of pressure applied to the entire hip region, we transformed the coordinate system of the MAC motion analysis system into that