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The effect of positioning on the biomechanical performance of soft shell hip protectors

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ABSTRACT

Wearable hip protectors represent a promising strategy for reducing risk for hip fracture from a sideways fall. However, small changes in pad positioning may influence their protective benefit. Using a mechanical hip impact simulator, we investigated how three marketed soft shell hip protectors attenuate and redistribute the impact force applied to the hip, and how this depends on displacement from their intended position by 2.5 or 5 cm superiorly, posteriorly, inferiorly or anteriorly. For centrally-placed protectors, peak pressure was reduced 93% below the unpadded value by a 16 mm horseshoe-shaped protector, 93% by a 14 mm horseshoe protector, and 94% by a 16 mm continuous protector. In unpadded trials, 83% of the total force was applied to the skin overlying the proximal femur (danger zone). This was lowered to 19% by the centrally placed 16 mm horseshoe protector, to 34% by the 14 mm horseshoe, and to 40% by the 16 mm continuous protector. Corresponding reductions in peak force delivered to the femoral neck (relative to unpadded) were 45%, 38%, and 20%, respectively. The protective benefit of all three protectors decreased with pad displacement. For example, displacement of protectors by 5 cm anteriorly caused peak femoral neck force to increase 60% above centrally-placed values, and approach unpadded values. These results indicate that soft shell hip protectors provide substantial protective benefits, but decline in performance with small displacements from their intended position. Our findings confirm the need for correct and stable positioning of hip protectors in garment design.

1. Introduction

Hip fracture is a common, serious injury for the elderly, and over 90% of hip fractures are caused by falls (Grisso et al., 1990). Hip protectors are wearable devices that are placed over 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 surrounding soft tissues.

While there is controversy about the efficacy of hip protectors from clinical studies (Koike et al., 2009; Kiel et al., 2007; Parker et al., 2006; Forsen et al., 2004; Birks et al., 2004; O'Halloran et al., 2004; van Schoor et al., 2003; Meyer et al., 2003; Harada et al., 2001; Kannus et al., 2000; Chan et al., 2000; Lauritzen et al.,1993), this is thought to be due largely to poor adherence in wearing the device, and there is general agreement that they reduce fracture risk if worn at the time of a fall (Cameron et al., 2001, 2010, 2003; Robinovitch et al., 2009; Ekman et al., 1997). However, additional evidence is required to understand the factors that influence both adherence and biomechanical performance.

Previous biomechanical studies have reported that, when tested in mechanical systems that simulate a fall on the hip, hip protectors attenuate the peak force at the femoral neck by between 17% and 89%, and in some cases, reduce it to a value below the average fracture threshold (3100 ± 1200 N) of elderly women (Laing and Robinovitch, 2008; van Schoor et al., 2006; Kannus et al., 1999; Parkkari et al., 1995). However, it remains unclear how hip protectors redistribute pressure over the hip region biomechanically.

Furthermore, little attention has been directed towards understanding how the positioning of a hip protector influences its biomechanical performance. Most hip protectors are provided by manufacturers as a unit that consists of a tight fitting undergarment, with hip protectors integrated into the lining (or in pockets), and worn so 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).

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While the exact amount of shifting that may occur is poorly understood, Minns et al. reported that the center point of a hip protector may displace as much as 2 cm from its initial position on the skin when the hip rotated from full extension to 90° of flexion. Derler et al. (2005) found from biomechanical testing that a 3 cm displacement of the protector in the anterior direction caused the peak force at the femoral neck to increase by up to 23%.

In the present study, we investigated how hip protectors redistribute the pressure applied to the hip region 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.

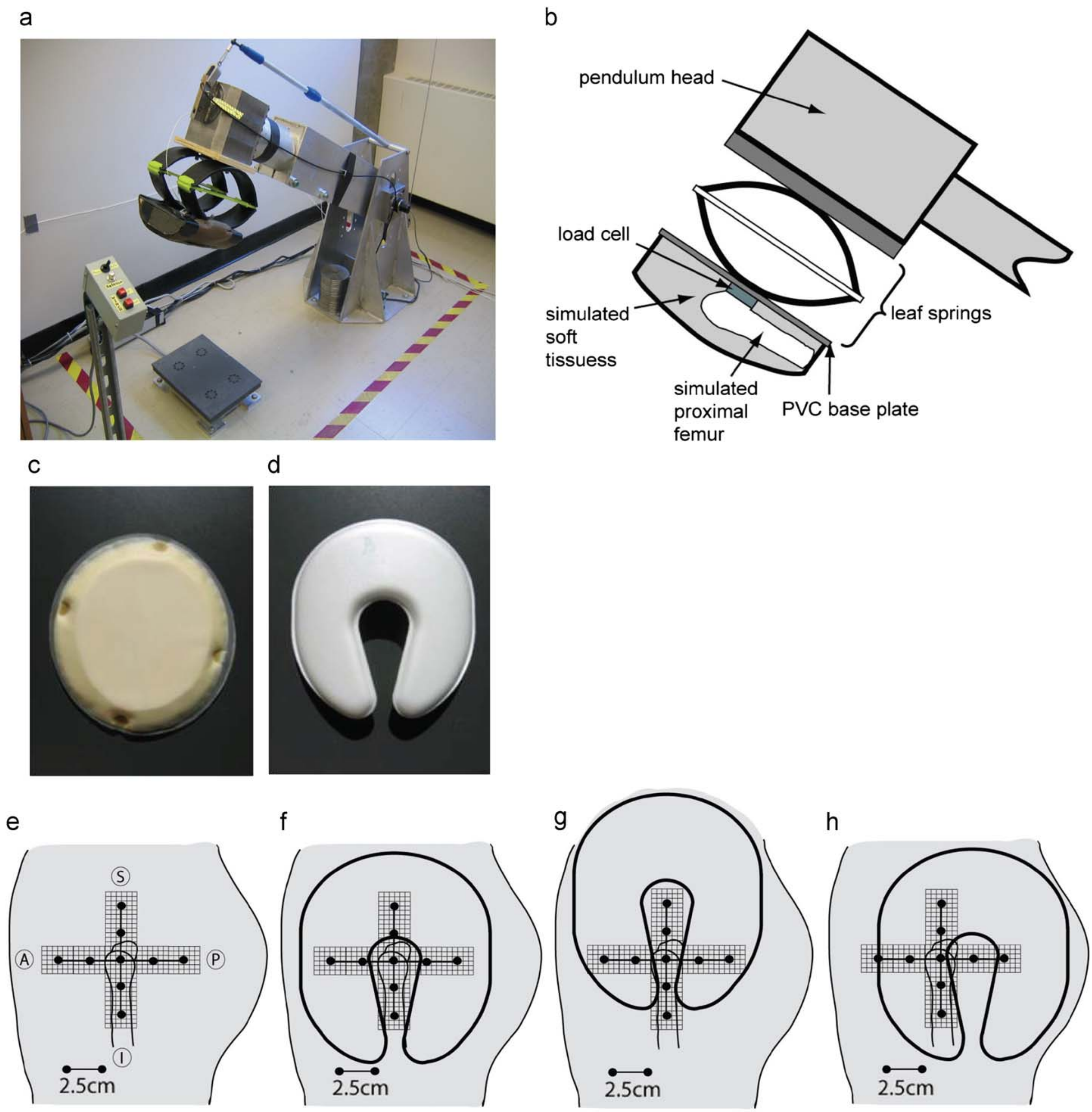

Fig. 1. Biomechanical testing with hip impact simulator. (a) SFU hip impact simulator. (b) Cross sectional diagram indicating placement of load cell to measure femoral neck force. (c) 16 mm continuous soft shell hip protector (*Hipsaver*, 20 cm long and 17 cm wide). (d) 14 or 16 mm thick horseshoe-shaped protector (both from *SafeHip*, 17 cm long and 17 cm wide). (e) Schematic of nine defined hip areas (unpadded condition). (f) Centrally placed horseshoe protector. (g) Protector displaced 5 cm in the superior direction. (h) Protector displaced 2.5 cm in the posterior direction.

2. Methods

2.1. Protocol

Biomechanical impact tests were conducted with the Simon Fraser University hip impact simulator (Fig. 1a and b). The SFU hip impact simulator was designed to measure the total force applied to the surface of a surrogate hip and the femoral neck force transmitted to the femoral neck during simulated sideways falls (Laing and Robinovitch, 2008). It is composed of an impact pendulum and a surrogate hip that mimics the hip geometry and local variation in soft tissue stiffness of typical elderly women.

In our tests, the surrogate hip was dropped onto a dual arrangement of an 2D pressure distribution plate (RSscan International, surface dimension: 40 cm by 60 cm, Olen, Belgium) and a force plate (Bertec, Type 2535-08, Columbus, OH, USA) from drop heights of 5, 10, and 20 cm (trials at higher heights were infeasible due to limits on the RSscan measurement range). While we were most interested in protector performance at the 20 cm drop height, we were also interested in the effect of drop height (fall severity) on the protective value of hip protectors. During each trial, we collected total hip impact force from the force plate, pressure distribution from the RSscan device and femoral neck force from a load cell (Kistler, Type 9712B5, Amherst, NY, USA) 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 3–1270 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.

Trials were acquired without a hip protector, and with three different soft shell hip protectors (Figs. 1c and d): 14 and 16 mm thick horseshoe-shaped pads (SafeHip, Tytext A/S, Denmark), and a 16 mm thick continuous pad (Hipsaver, Canton, MA, USA). The pads were taken from hip protector garments and secured to the surrogate hip by means of double-sided tape prior to impact tests. 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 or 5 cm in the superior, posterior, inferior, and anterior directions (Fig. 1e–h). Three trials were acquired for each condition.

2.2. Data analysis

Our main outcome variables were the magnitude of peak pressure, location of peak pressure, femoral neck force, and forces applied to four defined hip regions. Data analysis was conducted with customized Matlab routines. Magnitudes of peak pressure and femoral neck force were taken as the peak values recorded from the pressure distribution array and femoral load cell during the impact event, respectively. Sample pressure distributions at the moment of peak pressure for each protector are provided in Fig. 2b–e. 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 represented direct impact on the GT and femoral diaphysis (Fig. 2a). We anticipated that effective action of a hip protector would reduce the total force applied on this danger zone. We calculated the integrated force applied to each region by summing the product of sensor area multiplied by 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 representing three different subject trials in each condition, we assumed different subjects across all conditions, and we used randomized group ANOVA to test whether each of our outcome variables associated with hip protector type (4 levels), pad displacement (9 levels), and drop height (3 levels). The significance level in all tests was set to α=0.05, and all analyses were conducted in SPSS 16.0.

3. Results

The magnitude of peak pressure on the hip associated with drop height ($p<0.0005$), hip protector type ($p<0.0005$), and hip protector displacement condition ($p<0.0005$). For the 20 cm drops, peak pressure was reduced by 93% by the centrally placed 14 mm horseshoe protector (from 3570 to 234 kPa), by 93% (to 247 kPa) by the 16 mm horseshoe protector, and by 94% (to 204 kPa) by the 16 mm continuous protector (Fig. 2b–e, Fig. 3a). The percent reduction in peak pressure provided by each hip protector increased with fall height, averaging 74% for the 5 cm drop height, 84% for the 10 cm drop height, and 93% for the 20 cm drop height. Displacing the position of hip protectors by 2.5 or 5 cm in any direction caused peak pressure to increase on average by 97% and 158%, respectively (Fig. 3b). Peak pressure increased on average by 194% due to displacement in the anterior direction, and by 129%, 94%, and 93% due to respective displacements in the posterior, superior, and inferior directions. The effect of position was most pronounced for the 14 mm horseshoe protector (where 5 cm displacement caused peak pressure to increase by 234% on average), followed by the 16 mm horseshoe and 16 mm continuous protector (where 5 cm displacement caused increases of 115% and 124%, respectively).

The location of peak pressure 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 when hip protectors were displaced (Table 1), and was completely shunted away from the danger zone by the centrally placed 16 mm horseshoe protector (d=76.8 mm, θ=20.9°), but not by the centrally placed 14 mm horseshoe or by the 16 mm continuous protector (d=4.6 mm, θ=36.8° for both protectors; Fig. 4).

The integrated force on the danger zone 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, the percent of total force applied to the danger zone was 83% in the unpadded condition, 34% for the 14 mm horseshoe, 19% for the 16 mm horseshoe, and 40% for the 16 mm continuous protector when placed centrally (Fig. 5). The force distribution to areas B, C, and D also associated with drop height ($p<0.0005$), hip protector type ($p<0.0005$), and hip protector displacement condition ($p<0.0005$). For 20 cm drops, centrally placed hip protectors redistributed the forces applied on the hip region by lowering and deflecting much of the force away from the danger zone and onto adjacent soft tissue areas B, C, and D (Fig. 6a). This protective effect was reduced by displacements of the protectors (Fig. 6b–d).

Finally, peak femoral neck force associated with drop 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, peak femoral neck force averaged 45% lower with the 16 mm horseshoe protector, 38% lower with the 14 mm horseshoe, and 20% lower with the 16 mm continuous protector, when compared to the unpadded condition (Fig. 7). At the 20 cm drop height, peak femoral neck forces were 60% higher (on average) when the protectors were displaced 5 cm anterior, than when centrally placed. There was a significant interaction between hip protector and displacement, reflecting that peak femoral neck forces were lower for the 16 mm thick horseshoe than the 16 mm continuous protector in 3 of the 8 displacements conditions.

4. Discussion

The soft shell hip protectors we tested using our hip impact simulator reduced the magnitudes of peak pressure, peak femoral neck 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

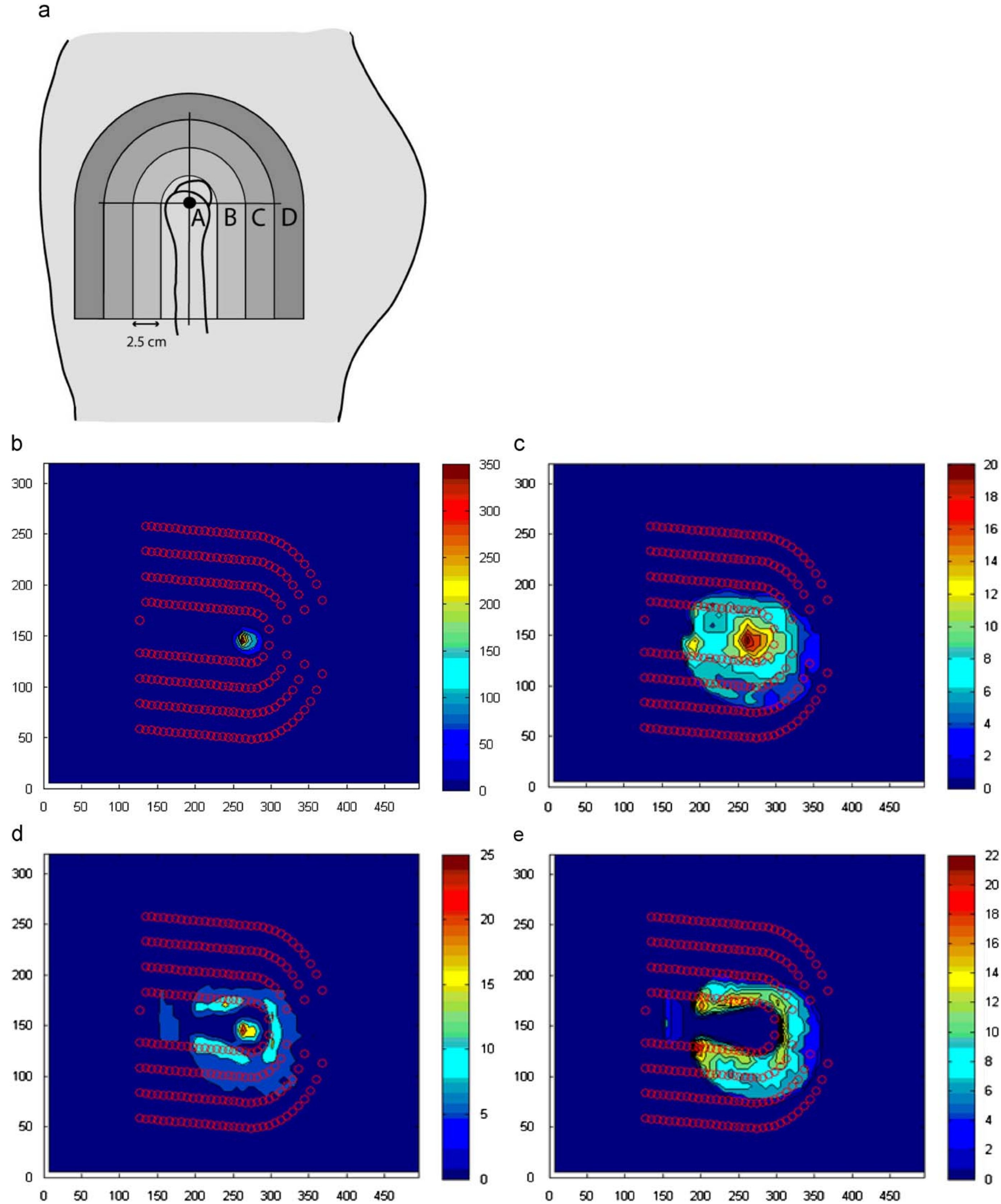

Fig. 2. Pressure distribution profiles with centrally placed hip protectors and 20 cm drop height. (a) Definition of four different areas over the hip region: area A, danger zone (light gray) consisted of a C-shaped region of width 5 cm (sum of a half-circle of radius 2.5 cm centered at the GT and rectangle extending distally 16 cm from the GT); areas B, C, and D (progressively darker gray) consisted of C-shaped hollow regions of width 10, 15, and 20 cm each. (b) In the unpadded condition, a large pressure concentrated on a small contact area at the GT, within the danger zone. Note the large variation in magnitude of pressure (indicated by color scales at right; pressure calibrations are in N/cm^2 ($1\ N/cm^2 = 10$ kPa)). (c) With the 16 mm thick 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 thick 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 thick 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.

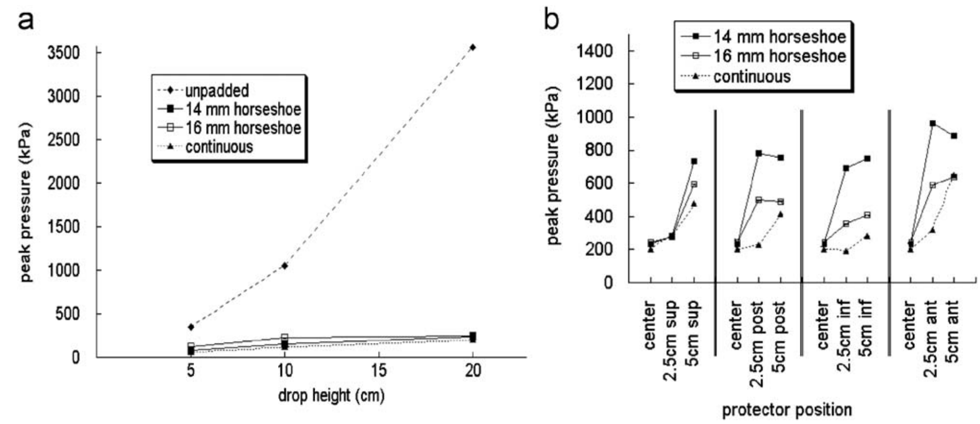

Fig. 3. Peak pressure over the hip region, recorded during impact tests with and without hip protectors. Data points show mean values, with error bars (often small enough to be hidden by the data point symbol) showing ± 1 SE. (a) The magnitude of peak pressure applied to the hip region was reduced by all three hip protectors, and was most pronounced at the 20 cm drop height. (b) Effect of hip protector displacement on peak pressure. All data are from tests at the 20 cm drop height. Pad displacement caused an increase in peak pressure. This was most pronounced for the 14 mm horseshoe protector, but observed even for the 16 mm continuous protector. However, peak pressures in all padded conditions remained well below those measured in the unpadded condition (as shown in a).

Table 1
Mean location of peak pressure (distance from the greater trochanter and angle from the femoral diaphysis) for the three protector types and nine protector positions, at the 20 cm drop height.

Position	Distance (mm)				Angle (deg.)			
	Unpadded	14 mm horseshoe	16 mm horseshoe	16 mm continuous	Unpadded	14 mm horseshoe	16 mm horseshoe	16 mm continuous
Center	4.6	4.6	76.8	4.6	36.8	36.8	20.9	36.8
2.5 cm sup	4.6	22.7	26.1	4.6	36.8	−47.8	−38.3	36.8
5 cm sup	4.6	8.4	22.7	4.6	36.8	−58.8	−47.8	36.8
2.5 cm post	4.6	4.6	4.6	4.6	36.8	−29	−29	36.8
5 cm post	4.6	4.6	4.6	4.6	36.8	36.8	36.8	36.8
2.5 cm inf	4.6	13.8	19.5	11.9	36.8	151	61.4	171.7
5 cm inf	4.6	4.6	11.9	11.9	36.8	36.8	171.7	171.7
2.5 cm ant	4.6	4.6	4.6	4.6	36.8	36.8	36.8	36.8
5 cm ant	4.6	4.6	4.6	4.6	36.8	36.8	36.8	36.8

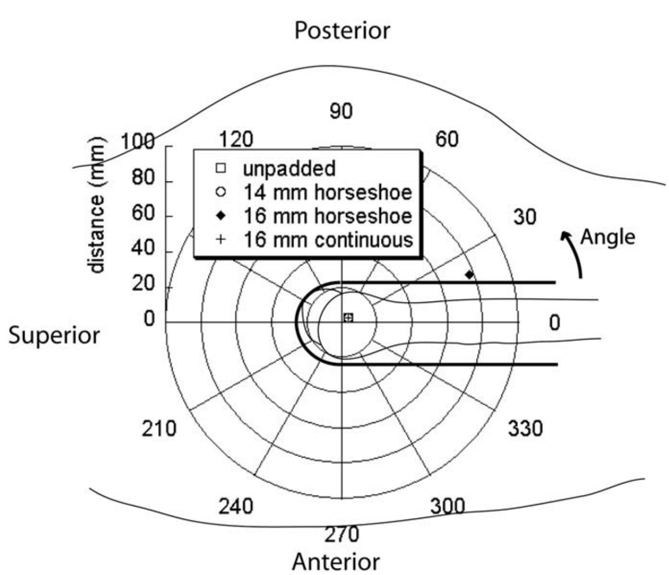

Fig. 4. Changes in the location of peak pressure as function of hip protector types (all for 20 cm drop height and placed in their intended location). Peak pressure was shunted away from the danger zone by the 16 mm horseshoe protector, but with both 14 mm horseshoe and 16 mm continuous protectors the peak pressure remained within the danger zone and its location did not change. Note that the unpadded, 14 mm horseshoe and 16 mm continuous data points all overlap and are centered on the GT.

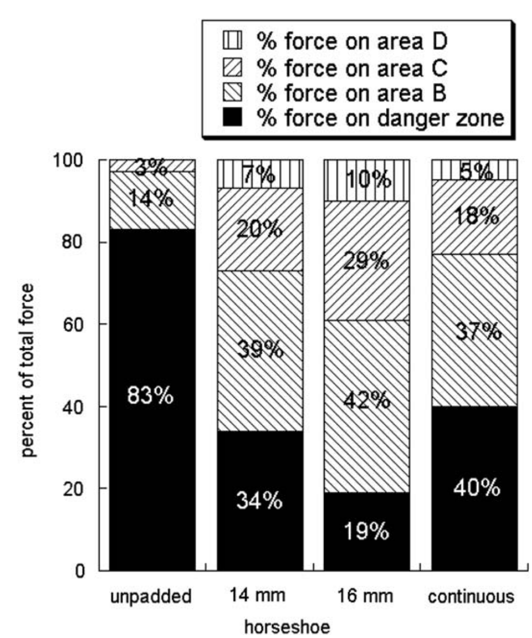

Fig. 5. Force distribution to four different areas over the hip. For a 20 cm drop height, the force applied to the danger zone was 83% of the total force in the unpadded condition, but only 19% of the total force with the centrally placed 16 mm horseshoe protector.

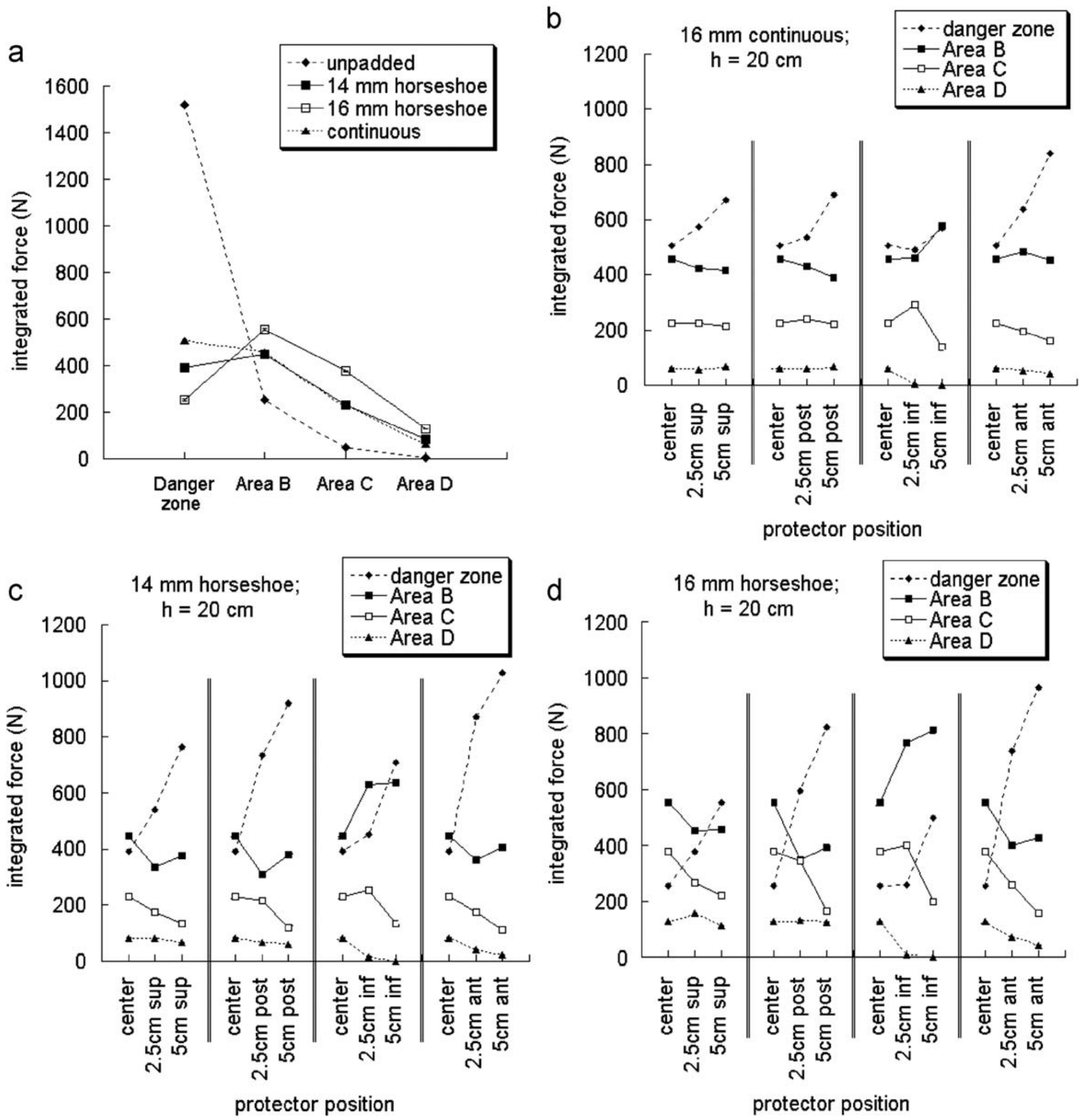

Fig. 6. Total force applied to four distinct regions over the hip (as defined in Fig. 2a). All data are from the 20 cm drop height. Data points show mean values, with error bars (often small enough to be hidden by the data point symbol) showing ± 1 SE. (a) Hip protectors lowered the force applied to the danger zone, in part by increasing that delivered to adjacent areas B, C, and D. (b–d) The ability of hip protectors to shunt force was impaired by displacement of the protector away from its intended (center) location. This effect was most pronounced for the horseshoe-shaped protectors, and for displacement in the anterior direction. Refer to part (a) for unpadded values.

for protective efficacy and support the notion that by wearing any of these soft shell hip protectors, the risk for hip fracture is reduced. While the 16 mm continuous protector caused the greatest overall reduction in peak pressure, both 14 and 16 mm horseshoe protectors provided greater reduction in the femoral neck force and percent force to the danger zone, and were more effective in shunting the location of peak pressure away from the GT. Furthermore, only the 16 mm horseshoe shunted the location of peak pressure outside the danger zone. These collective observations suggest that, when positioned in their intended location, the horseshoe protectors are better than the continuous protector in reducing the risk of hip fracture in falls, and the 16 mm thick version provides better protection than the 14 mm horseshoe design.

We also found that the biomechanical performance of all three protectors depended on their position over the hip region. The horseshoe protectors were especially sensitive to positioning: displacement in any direction caused increases in peak pressure, peak femoral neck force, and percent of total force applied to the danger zone. Furthermore, for all three protectors we tested, 5 cm displacement in the anterior direction caused peak femoral neck forces to approach or even exceed values observed in the unpadded condition, implying minimal protective value or even a slight increase in fracture risk under these conditions—an issue of clear importance to garment design. Our results agree with and extend the findings of Derler et al. (2005), who conducted biomechanical tests on ten different hip protectors, placed in their intended location, and displaced 3 cm anteriorly. These authors reported that anterior displacement caused peak femoral neck force to increase for seven of the tested products, and to decrease for the remaining three.

However, we also observed cases where displacement of the 16 mm continuous and 14 mm horseshoe pads from their intended location actually improved their performance. For example, inferior displacement of the 16 mm continuous pad improved its ability to decrease peak femoral neck force, shunt

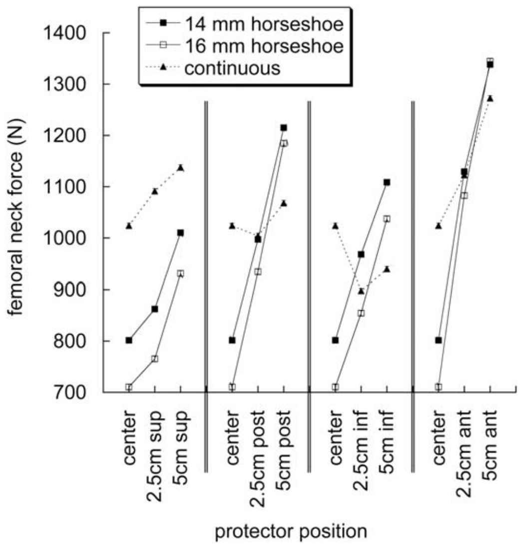

Fig. 7. Effect of protector position on the peak force measured by the load cell embedded in the femoral neck. All data are from the 20 cm drop height condition. Data points show mean values, with error bars (often small enough to be hidden by the data point symbol) showing ± 1 SE. When hip protectors were centrally located, peak femoral neck force was lower for the horseshoe than continuous protectors. However, when displaced 5 cm in the posterior, inferior, or anterior directions, peak force was lower for the continuous than horseshoe protector. Furthermore, 5 cm displacement in the anterior direction caused peak forces to approach or even exceed values observed in the unpadded condition. A rare beneficial effect of displacement is observed in the reduction in peak force due to inferior displacement of the continuous protector.

peak pressure away from the GT, and reduce total force to the danger zone. Furthermore, the location of peak pressure was shunted away from the GT when the 14 mm horseshoe protector was displaced superiorly, but not when it was centrally placed.

Our evidence regarding the effect of pad positioning on biomechanical performance highlights the importance of well-designed and well-fitting garments to house these pads. All three of the protectors we tested are designed to be integrated in tight fitting undergarments. In practical use, however, one obvious source of protector displacement is shifting of the garment over the skin surface during daily activities. Minns et al. (2007) reported that the position of a well-fitting hip protector relative to the GT displaced 0.45 cm posteriorly from its initial position due to walking and sitting, and up to 2 cm when the hip was flexed from full extension to 90° of flexion. Clearly, displacements may be considerably greater for poor-fitting garments. An additional source of relative pad displacement (revealed anecdotally during the course of the current study) is movement of the GT under the skin with flexion and internal rotation of the hip, which is common when the hip impacts the ground from a sideways fall (Hsiao and Robinovitch, 1998; Feldman and Robinovitch, 2007). Future work should quantify how hip joint rotations affect the position of the proximal femur relative to the skin and a wearable hip protector.

There are two important limitations to the current study. First, we tested only soft shell hip protectors and not hard shell designs. While hard shell protectors have traditionally been the focus of clinical trials (Kannus et al., 1999, 2000; Lauritzen et al., 1993), soft shell designs are currently receiving increased attention due to the potential for improved user acceptance and adherence (Choi et al., 2009; Bentzen et al., 2008; Laing and Robinovitch, 2008). Furthermore, the experimental technique we used – of measuring pressure distribution over the external surface of the pad (and surrounding skin) with a rigid pressure sensing plate – is biomechanically relevant for soft shell designs, since it reflects the pressure applied through to the soft tissues and bone, but less so for a hard shell "dome" designs, where the pad may contact the plate only at the apex of the dome. In such case, a device located between the pad and skin (such as a pressure sensing cloth) is required to measure the relevant pressure distribution. Second, our impact experiments involved smaller drop heights (5, 10, and 20 cm) and created lower impact forces than might be expected during typical falls from standing. However, we could not exceed a 20 cm drop height without risking damage to our RSscan pressure measurement system. This limitation should be addressed in future studies when devices become available with a larger measurement range.

In summary, the three soft shell hip protectors we tested showed protective effects against external impact when positioned in their intended location over the proximal femur. The horseshoe shaped protectors provided greater reduction in femoral neck force than the continuous protector, and the 16 mm thick horseshoe pad was the only one that shunted peak pressure outside the danger zone overlying the proximal femur. Furthermore, our results indicate that there are optimal positions for these pads over the hip region in order to maximize their protective benefit. For the horseshoe shaped protectors, performance was optimal when placed in the central intended location, while for the continuous protector, performance was enhanced when displaced as much as 5 cm inferiorly from the intended location. The performance of all three tested protectors declined when displaced anteriorly. These findings highlight the crucial role of the design of the pad itself, and the proper fit of the garment in which the pads are integrated.

Conflict of interest statement

Robinovitch is a consultant to Tytex A/S, manufacturer of the Safehip soft shell protectors included in the current study. Choi and Hoffer have no financial or personal relationships with any hip protector manufacturer.

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