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The effects of pad geometry and material properties on the biomechanical effectiveness of 26 commercially available hip protectors

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ABSTRACT

Wearable hip protectors (padded garments) represent a promising strategy to decrease impact force and hip fracture risk during falls, and a wide range of products are currently marketed. However, little is known about how design features of hip protectors influence biomechanical effectiveness. We used a mechanical test system (simulating sideways falls) to measure the attenuation in femoral neck force provided by 26 commercially available hip protectors at three impact velocities (2, 3, and 4 m/s). We also used a materials testing machine to characterize the force–deflection properties of each device. Regression analyses were performed to determine which geometric (e.g., height, width, thickness, volume) and force–deflection properties were associated with force attenuation. At an impact velocity of 3 m/s, the force attenuation provided by the various hip protectors ranged between 2.5% and 40%. Hip protectors with lower stiffness (measured at 500 N) provided greater force attenuation at all velocities. Protectors that absorbed more energy demonstrated greater force attenuation at the higher impact velocities (3 and 4 m/s conditions), while protectors that did not directly contact (but instead bridged) the skin overlying the greater trochanter attenuated more force at velocities of 2 and 3 m/s. At these lower velocities, the force attenuation provided by protectors that contacted the skin overlying the greater trochanter increased with increasing pad width, thickness, and energy dissipation. By providing a comparison of the protective value of a large range of existing hip protectors, these results can help to guide consumers and researchers in selecting hip protectors, and in interpreting the results of previous clinical trials. Furthermore, by determining geometric and material parameters that influence biomechanical performance, our results should assist manufacturers in designing devices that offer improved performance and clinical effectiveness.

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1. Introduction

Hip fractures (i.e., fractures of the proximal femur) are a major public health problem for older adults. The lifetime risk for hip fracture in the USA is 17% for Caucasian women and 6% for Caucasian men (Cummings and Melton, 2002). While bone density is a major determinant of fracture risk, the majority of hip fractures occur in persons who do not suffer from osteoporosis (Dargent-Molina et al., 1996; Taylor et al., 2004). Instead, fall mechanics and the resulting loads applied to the proximal femur during impact are the factors most closely associated with the risk of suffering a hip fracture (Cummings and Nevitt, 1994). Sideways falls increase hip fracture risk by 5-fold when compared to forwards or backwards falls (Hayes et al., 1993); the risk increases by 32-fold when direct impact to the greater trochanter occurs (Nevitt and Cummings, 1993). Accordingly, protective devices that reduce the force applied to the proximal femur during fall-related impacts have the potential to reduce hip fracture risk.

Wearable hip protectors (padded garments) are a promising strategy for decreasing hip fracture risk by reducing the loads applied to the proximal femur during fall-related impacts. Until recently there were no established guidelines for assessing the biomechanical and clinical effectiveness of these devices (Cameron et al., 2010; Robinovitch et al., 2009). Consequently, there are currently more than two dozen commercially marketed

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hip protectors in North America that utilize a surprisingly diverse array of design philosophies, materials, and geometry. The first generation of hip protectors used 'hard shell' domes, which bridged over the greater trochanter, to shunt energy away from the proximal femur during impact (Kannus et al., 2000; Lauritzen et al., 1993). More recently, soft shell hip protectors have become more common. These products reduce the force applied to the proximal femur by absorbing energy in the pad material, and reducing the local stiffness over the greater trochanter through a "springs-in-series" mechanism (Laing and Robinovitch, 2008a, 2008b).

Researchers have shown that the biomechanical effectiveness (i.e. force attenuation capacity) of hip protectors is influenced by external factors including impact velocity, soft tissue properties, and pelvic surface geometry (Kannus et al., 1999; Laing and Robinovitch, 2008b; Mills, 1996; van Schoor et al., 2006). Presumably, pad geometry (thickness, surface area) and material properties (stiffness, damping) also affect force attenuation. In controlled experiments, Robinovitch et al. (1995) found that increasing the thickness of 'horse-shoe' shaped pads (of identical surface area) from 18 to 38 mm increased force attenuation by 66%. However, the force–deflection and geometric properties of commercially marketed hip protectors vary widely, and it is not known whether specific biomechanical variables govern (or explain) between-product variations in force attenuation. Such information should guide users in the selection of products, and manufacturers in the design of a new generation of hip protectors with increased biomechanical effectiveness.

The goals of this study were therefore to test the hypotheses that the force attenuation provided by a range of hip protectors when positioned correctly over the greater trochanter would significantly associate with their geometry (e.g. width, thickness) and force–deflection properties (e.g. stiffness, energy absorption). Towards these ends, we used a mechanical test system to measure the force attenuation provided by 26 commercially available hip protectors. For each hip protector, we also used a materials testing system to measure the force–deflection properties, and digital calipers to measure pad geometry. Regression analyses were used to test whether force attenuation depended on material and geometric properties.

2. Methods

2.1. Hip protector brands and characteristics

We used a combination of literature review and Internet searching (using search terms such as "hip protector", "hip pad", "seniors", and "hip fracture") to identify 26 commercially available hip protectors from distributors in nine different countries for testing (Table 1; Fig. 1). We used a binary variable ($material_{type}$) to categorize the products according to their dominant material type (hard versus soft shell). Specifically, 'soft shell' protectors consisted primarily of foam and fabric (21 models), while 'hard shell' protectors contained a relatively stiff material that bridged over the greater trochanter (5 models). We also categorized the protectors based on their dominant geometry type ($geometry_{type}$), which describes the nature of the interface between the hip protector surface and the skin overlying the lateral pelvis. In particular, we categorized 21 models as 'touching', and 5 models as 'not touching' the skin directly overlying the greater trochanter. Basic information on the material types used in the each protector is available as Appendix A in supplementary website material.

2.2. Impact force attenuation tests

We used the Simon Fraser University hip impact simulator (Fig. 2) to measure the biomechanical effectiveness of each hip protector. The system and test method have been described in detail previously (Laing and Robinovitch, 2008b, 2009), and are generally compatible with guidelines from an international team of biomechanics and clinical experts (Robinovitch et al., 2009). The system consists of an impact pendulum and surrogate pelvis released from an inclined position by an electromagnet to strike the ground in a horizontal position. The surrogate pelvis is comprised of foam-rubber soft tissues and an instrumented proximal femur (Sawbones, Vashon, WA, USA). Surface geometry and local variation in soft tissue stiffness match average measurements from older women to within one standard deviation (Laing and Robinovitch, 2008b). The surrogate pelvis is connected to the pendulum via leaf springs that simulate the compliance of the

Table 1
Manufacturer, general design approach, and geometric variables measured from 26 hip protectors.

Hip protector name	Company	$Material_{type}$	$Geometry_{type}$[a]	Height (mm)	Width (mm)	$Thick_{pad}$ (mm)	$Thick_{wearing}$ (mm)	Volume (mm^3)
Alimed® Hip Shield	AliMed®	Soft	Y	170	190	14	14	452,200
Anatech	Anatech	Soft	Y	160	210	13.5	14	453,600
Bort	Bort Medical	Soft	Y	170	150	15	19	382,500
Caresse	Remploy Healthcare	Hard	Y	230	140	22	22	708,400
ComfiHips™	ComfiHips LLC	Soft	Y	195	145	17.5	17	494,813
FallGard	FallGard	Soft	Y	175	120	17.5	19	367,500
Hip Guard	HipGuard™ Ltd.	Hard	N	155	165	20.4	19	521,730
Hip Shield	Promedics	Soft	Y	200	150	16	17	480,000
HipEase	Patterson Medical/Sammons Preston	Soft	Y	170	150	32	31	816,000
HIPS	Qvortrup Medical A/S	Hard	N	160	110	5	25	88,000
Hipsaver®	Hipsaver®	Soft	Y	220	200	18	19	792,000
Hornsby Comfy Hip	Hornsby Comfy Hips Pty Ltd	Soft	Y	195	165	18	21	579,150
Impactwear® - Flexible	Impactwear	Soft	Y	155	120	16	19	297,600
KPH®	Medlogics	Hard	N	190	95	15	31	270,750
LYDS	Comfortable	Soft	Y	205	135	6.5	7	179,888
Pelican Super Soft	Pelican Manufacturing PTY LTD	Soft	Y	170	160	21	22	571,200
Pelican Washable	Pelican Manufacturing PTY LTD	Soft	Y	160	150	17	18	408,000
Pelican 179P	Pelican Manufacturing PTY LTD	Soft	Y	155	140	23	24	499,100
Posey Heavy Duty	J.T. Posey company	Soft	Y	170	140	14.5	14	345,100
Posey Regular	J.T. Posey company	Soft	Y	165	140	13	13	300,300
ProtectaHip+Plus®	Plum Enterprises INC	Soft	Y	180	180	18	20	583,200
ProtectaHip®	Plum Enterprises INC	Soft	Y	180	180	14.5	16	469,800
Safehip® Air-X™	TYTEX A/S	Soft	N	210	185	16	16	621,600
Safehip® Classic	TYTEX A/S	Hard	N	160	115	8.6	24	158,240
Secure	Personal Safety Corporation	Soft	Y	165	145	16	16	382,800
WonderHip™	Vital Base AS	Soft	Y	193	137	15	16	396,615

[a] $Geometry_{type}$ indicates whether the protector touches (Y) or does not touch (N) the skin directly overlying the greater trochanter.

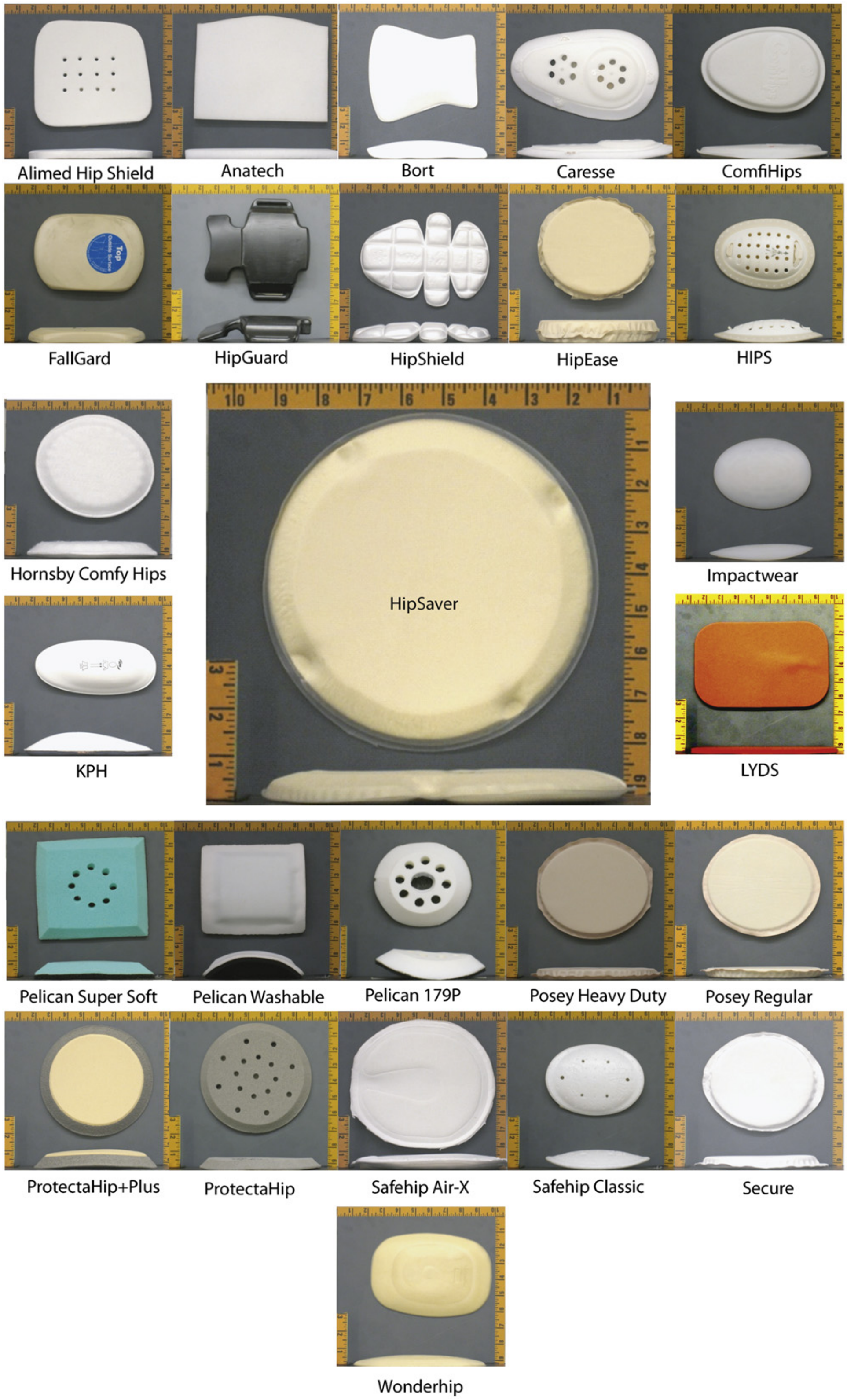

Fig. 1. Photographs showing top and side views of the 26 hip protectors tested. Scale of individual photograph frames is 26.7 cm horizontal × 24.1 cm vertical (10.5 in. × 9.5 in.).

pelvis producing a total effective stiffness of 42.2 kN/m (Laing and Robinovitch, 2010; Robinovitch et al., 1997). The effective mass of the system (28.0 kg) is within one standard deviation of the mean measured from women during lateral falls on the hip (Robinovitch et al., 1997). Impact velocity is varied by adjusting the angle of the pendulum before release, and is measured by a rotary variable inductance transducer (Shaevitz RVIT 15-1201). The force applied to the femoral neck is measured by a load cell (Kistler Model 9712A5000, Amherst, NY, USA), and the total impact force is measured with a force plate (model 2535-08, Bertec Corp., Columbus, OH, USA). During test sessions, hip protectors were positioned according to manufacturer instructions with the aid of a laser that indicated the location of the greater trochanter within the surrogate pelvis.

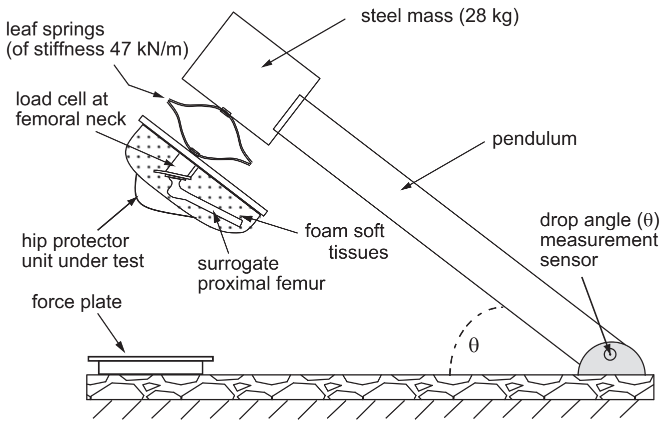

Fig. 2. Simon Fraser University hip impact simulator. The impact pendulum is connected in series with a surrogate pelvis. A load cell located in the femoral neck measures the force applied to the proximal femur during simulated sideways falls.

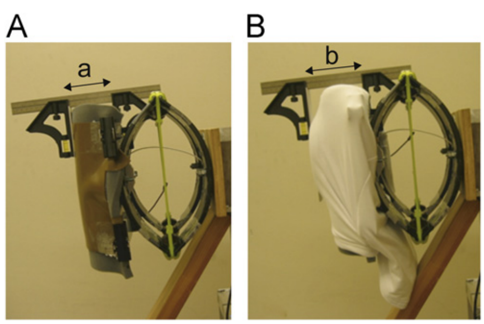

Fig. 3. Illustration of the protocol for measuring the increase in pelvic width in the frontal plane when wearing the protector ($thick_{wearing}$). This variable was calculated as the distance from the medial aspect of the surrogate pelvis to the lateral surface of the hip protector (measure 'b' in subplot B) minus the distance between the medial and lateral aspects of the unpadded surrogate pelvis (measure 'a' in subplot A).

Based on findings that the impact velocity during unexpected sideways falls averages 3.0 m/s (SD 1.0 m/s) (Feldman and Robinovitch, 2007), we simulated mild (2 m/s), moderate (3 m/s), and severe (4 m/s) falls. For each impact velocity, we collected three unpadded trials, followed by two sequential trials with each hip protector. The hip protectors were tested in a random order. This number of repeated impacts is smaller than the five recommended in recent guidelines (Robinovitch et al., 2009), but the compromise was considered reasonable given the large number of hip protectors tested, and the observation that the difference in peak force between repeated trials was only 0.7%, on average. A 3 min refractory period was inserted after each trial. All hip protectors were preconditioned for at least 24 h in the test facility using a thermostat to maintain temperature at approximately 21 °C. Sensors were sampled for 2 s at 1000 Hz. Force data were filtered with a dual-pass fourth-order Butterworth low pass filter with a 35 Hz cut-off frequency (Labview 6.1, National Instruments, Austin, TX, USA). For each trial, we identified the peak femoral neck force (F_{neck}). For each hip protector, the attenuation in femoral neck force (F_{neck_atten}) was calculated as the average percentage decrease in F_{neck} in the padded trials compared to the unpadded conditions.

2.3. Geometric variables

We used digital callipers to measure the maximum dimensions of each protector in the inferior–superior (*height*) and anterior–posterior (*width*) directions. We also measured the outer- to inner-surface thickness of the portion of the pad that overlied the greater trochanter ($thick_{pad}$), and the product of *height*, *width*, and $thick_{pad}$ (*volume*). Finally, we used a customized clamp (Fig. 3) to measure the increase in pelvic width in the frontal plane when wearing the protector ($thick_{wearing}$, in mm resolution). In some cases, $thick_{wearing}$ was smaller than $thick_{pad}$, due to the tight fit of the garment compressing the soft tissues of the surrogate pelvis. For protectors that bridged over the greater trochanter, a general approximation of the thickness of air gap between the medial aspect of the protector and the skin surface over the greater trochanter is given by subtracting $thick_{pad}$ from $thick_{wearing}$ (except for Hip Guard, for which the compression of the tissues over the greater trochanter was greater than the pad thickness in that area).

2.4. Force–deflection variables

We measured the force–deflection properties of each hip protector (Fig. 4) through dynamic indentation tests with a servohydraulic testing system (FastTrack™ 8874, Instron Corporation, Canton, MA, USA). We used a rigid hip-shaped indenter that matched the pelvic surface geometry of a female model of body

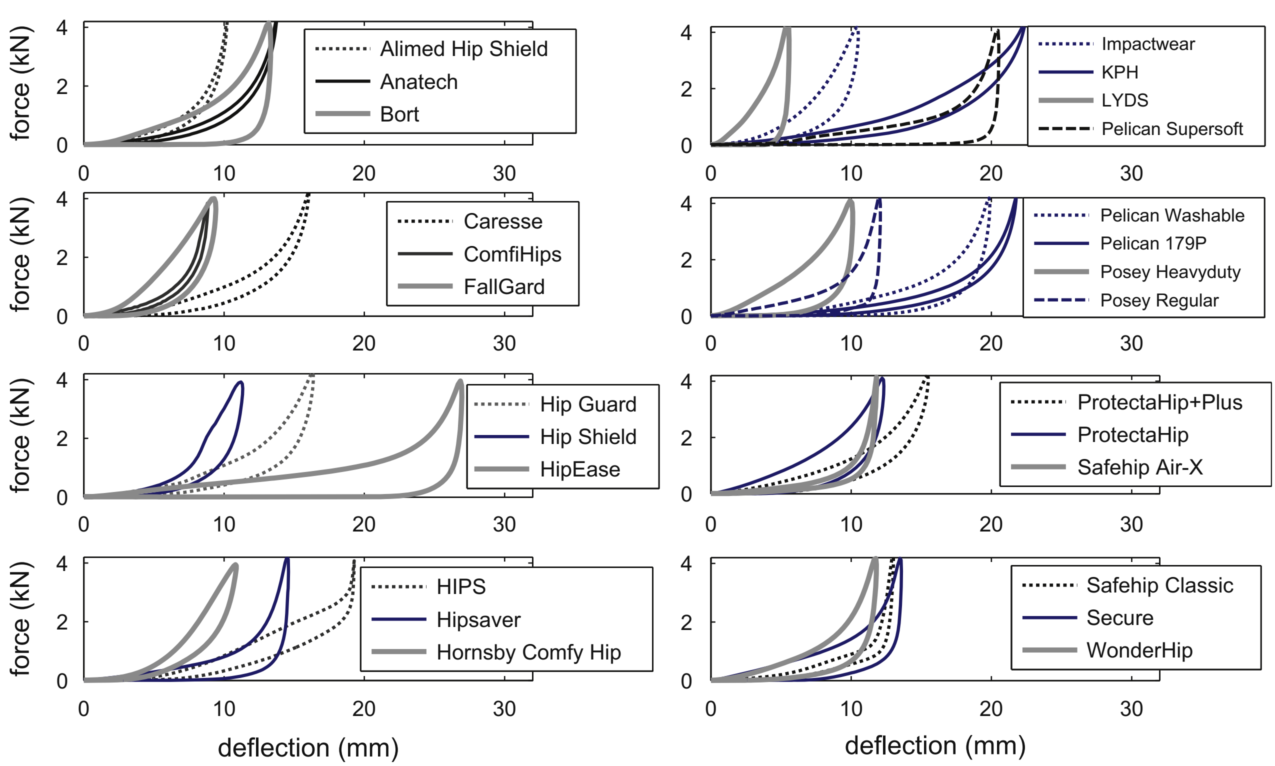

Fig. 4. Force–deflection profile for each of the 26 hip protectors tested.

mass 49.1 kg and height 1.55 m (Laing et al., 2006). Trials were conducted using ramp loading and unloading rates of 35 mm/s to a peak force of 4 kN, the approximate peak load applied to the hip during a sideways fall from standing (Laing and Robinovitch, 2009; Laing et al., 2006). The data acquisition rate was 1000 Hz.

A custom software routine (Matlab v2006a, The Mathworks, Natick, MA, USA) was used to calculate the force–deflection outcomes of interest. Stiffness was measured as the tangent of the force–deflection curve during the compression phase at 500 N (k_{500}), 2000 N (k_{2000}), and 3000 N (k_{3000}). Energy absorbed (E_{abs}) was calculated as the area under the force–deflection curve during the compression phase. Absolute energy dissipated (E_{dis}) was calculated by subtracting from E_{abs} the area under the curve during the decompression phase. Relative energy dissipation ($E_{dis\%}$) was defined as $(E_{dis}/E_{abs}) \times 100\%$

2.5. Statistics

We conducted separate linear regression analyses for the data at 2, 3, and 4 m/s to determine whether the biomechanical effectiveness (F_{neck_atten}) of the hip protectors was associated with general material and geometry type (material$_{type}$, geometry$_{type}$), geometric parameters (*height*, *width*, *thick$_{pad}$*, *volume*, *thick$_{wearing}$*) or force–deflection properties (k_{500}, k_{2000}, k_{3000}, E_{abs}, E_{dis}, $E_{dis\%}$). We initially conducted univariate analyses to identify the parameters (those with $p < 0.10$) to include in the multivariate regression models. We then used backwards stepwise regression (removal from model: $p > 0.1$) to determine the final predictive models. When we observed $geometry_{type}$ as a significant predictor we performed subset analyses (using the same univariate/multivariate approach described above) to identify significant predictors of F_{neck_atten} for devices that did versus did not touch the skin overlying the greater trochanter. All analyses were performed with statistical analysis software (SPSS Version 17.0, Chicago, IL, USA).

3. Results

For each hip protector, geometric parameters are presented in Table 1, while the force–deflection properties and biomechanical effectiveness values are provided in Table 2. F_{neck} during the

Table 2
Force–deflection and biomechanical effectiveness variables measured from the 26 hip protectors.

Hip protector name	k_{500} (kN/m)	k_{2000} (kN/m)	k_{3000} (kN/m)	E_{abs} (J)	E_{dis} (J)	$E_{dis\%}$ (%)	F_{neck_atten} (%)		
							2 m/s	3 m/s	4 m/s
Alimed® Hip Shield	183	963	1991	8.7	2.32	26.7	23.7	20	14.8
Anatech	144	677	1402	11.64	2.64	22.7	22.3	22.6	18.3
Bort	158	509	945	14.71	13.46	91.5	0.3	10.2	8.5
Caresse	151	472	965	15.43	5.91	38.3	30.9	22.1	15.4
ComfiHips™	192	1432	2264	6.72	1.86	27.7	10.1	7.5	11.1
FallGard	349	587	740	13.1	9.03	68.9	2.9	12.4	10.7
Hip Guard	133	449	774	17.6	8.95	50.8	20.6	22.2	18.1
Hip Shield	197	902	795	11.27	6.35	56.4	16.1	12.4	7.3
HipEase	46	433	954	22.91	20.8	90.8	38.4	40	30.1
HIPS	155	207	795	20.03	8.62	43.1	48.3	37.1	26.6
Hipsaver®	95	877	1728	11.4	9.12	80	23.6	23.5	17.6
Hornsby Comfy Hip	270	684	776	12.24	5.39	44	8.2	15.1	8.9
Impactwear® – Flexible	252	668	962	12.23	7.07	57.8	-6.4	2.5	2.4
KPH®	92	276	438	26.1	9.75	37.4	41.5	36.6	23
LYDS	443	889	1569	7.77	7.18	92.4	10.6	10.1	8.2
Pelican Super Soft	68	742	1616	14.08	13.05	92.7	33	30.2	25.3
Pelican Washable	110	497	1047	15.76	10.43	66.2	20	18.9	16
Pelican 179P	92	545	1196	14.97	4.52	30.2	22.5	20.2	15.6
Posey Heavy Duty	266	477	821	13.87	11.42	82.3	3.8	15.4	15
Posey Regular	116	895	1806	10.4	8.69	83.6	26.2	23	15.6
ProtectaHip+Plus®	130	429	769	17.45	8.79	50.4	23.3	24.5	19.7
ProtectaHip®	185	421	752	16.14	11.22	69.5	20.7	22.5	17.5
Safehip® Air-X™	156	1781	3278	6.52	2.84	43.5	34	26.6	18.7
Safehip® Classic	157	1981	3421	8.12	3.49	43	19.4	17.5	8.2
Secure	125	536	1308	14.26	11.31	79.3	24.6	21.3	15.4
WonderHip™	158	647	1270	12.17	8.47	69.6	8	12.2	7.1

Table 3
Univariate regression summary for the dependent variable F_{neck_atten} across all hip protectors. Unstandardized regression coefficients, p values, and unadjusted r values are provided.

	Regression coefficient			p value			Unadjusted r		
	2 m/s	3 m/s	4 m/s	2 m/s	3 m/s	4 m/s	2 m/s	3 m/s	4 m/s
$Material_{type}$	−14.70	−8.492	−3.789	0.021*	0.059	0.257	0.450	0.375	0.231
$Geometry_{type}$	−15.47	−9.609	−4.625	0.015*	0.031*	0.164	0.473	0.424	0.281
k_{500}	−0.093	−0.066	−0.046	0.001*	0.001*	0.001*	0.617	0.634	0.609
k_{2000}	−0.003	−0.006	−0.005	0.620	0.180	0.095	0.102	0.271	0.334
k_{3000}	0.002	−0.001	−0.001	0.664	0.703	0.538	0.089	0.078	0.127
E_{abs}	1.291	1.244	0.866	0.018*	< 0.001*	0.001*	0.461	0.641	0.614
E_{dis}	0.445	0.847	0.700	0.489	0.049*	0.023*	0.142	0.390	0.444
$E_{dis\%}$	−10.88	−1.027	0.889	0.358	0.901	0.882	0.188	0.026	0.031
Height	0.040	−0.024	−0.036	0.753	0.783	0.573	0.065	0.057	0.116
Width	0.017	0.019	0.039	0.860	0.780	0.420	0.036	0.058	0.165
$Thick_{pad}$	0.258	0.405	0.408	0.616	0.250	0.105	0.103	0.234	0.326
$Thick_{wearing}$	1.243	1.036	0.633	0.011*	0.001*	0.010*	0.492	0.592	0.498
Volume	< 0.001	< 0.001	< 0.001	0.414	0.255	0.118	0.167	0.231	0.314

* Indicates significant univariate predictors that were used as inputs in the multivariate models that combined all hip protectors.

Table 4
Univariate regression summary for the dependent variable F_{neck_atten} for only the subset of hip protectors with a $geometry_{type}$ that touched the skin overlying the greater trochanter. Unstandardized regression coefficients, p values, and unadjusted r values are provided.

	Regression coefficient			p value			Unadjusted r		
	2 m/s	3 m/s	4 m/s	2 m/s	3 m/s	4 m/s	2 m/s	3 m/s	4 m/s
k_{500}	−0.082	−0.059	NA	0.001*	0.001*	NA	0.676	0.670	NA
k_{2000}	−0.005	−0.012	NA	0.636	0.115	NA	0.110	0.354	NA
k_{3000}	0.006	<0.001	NA	0.313	0.973	NA	0.231	0.008	NA
E_{abs}	1.307	1.493	NA	0.073*	0.002*	NA	0.399	0.635	NA
E_{dis}	0.628	0.972	NA	0.303	0.018*	NA	0.236	0.509	NA
$E_{dis\%}$	−0.443	6.320	NA	0.969	0.435	NA	0.009	0.180	NA
Height	0.044	−0.033	NA	0.733	0.718	NA	0.079	0.084	NA
Width	0.199	0.148	NA	0.058*	0.049*	NA	0.420	0.434	NA
$Thick_{pad}$	1.186	1.038	NA	0.022*	0.004*	NA	0.498	0.607	NA
$Thick_{wearing}$	0.892	0.887	NA	0.106*	0.020*	NA	0.362	0.502	NA
Volume	<0.001	<0.001	NA	0.003*	0.001*	NA	0.610	0.653	NA

* Indicates significant univariate predictors that were used as inputs in the multivariate models that included hip protectors that touched the skin overlying the greater trochanter ($thick_{wearing}$ at 2 m/s included as it was of borderline significance). Analyses for this subset of hip protectors were not conducted for F_{neck_atten} at 4 m/s as univariate analyses including all hip protectors (Table 3) found no significant effect of $geometry_{type}$ at this impact velocity.

Table 5
Multivariate regression summary for the final model predictors of F_{neck_atten} across all hip protectors. Unstandardized regression coefficients and p values for dependent variables and the overall adjusted r^2 for each model are provided.

	Regression coefficient	p value	Adjusted r^2 for model
2 m/s			
$Geometry_{type}$	−12.28	0.018	0.476
k_{500}	−0.083	0.001	
3 m/s			
$Geometry_{type}$	−6.01	0.058	0.582
k_{500}	−0.043	0.010	
E_{abs}	0.788	0.011	
4 m/s			
k_{500}	−0.032	0.015	0.480
E_{abs}	0.607	0.013	

Table 6
Multivariate regression summary for the final model predictors of F_{neck_atten} for the subset of hip protectors with a $geometry_{type}$ that touched the skin overlying the greater trochanter. Unstandardized regression coefficients and p values for dependent variables and the overall adjusted r^2 for each model are provided. No information is provided for the subset of hip protectors that did not touch the skin overlying the greater trochanter (N=5) as there were no significant predictors of F_{neck_atten} for this group.

	Regression coefficient	p value	Adjusted r^2 for model
2 m/s			
Width	0.209	0.006	0.615
$Thick_{pad}$	5.63	0.001	
$Thick_{wearing}$	−4.69	0.003	
3 m/s			
E_{dis}	0.787	0.003	0.789
Width	0.179	<0.001	
$Thick_{pad}$	3.09	<0.001	
$Thick_{wearing}$	−2.48	0.004	

baseline unpadded conditions averaged 1392, 2100, and 2698 N for the 2, 3, and 4 m/s impact velocity conditions, respectively. F_{neck} during the padded condition ranged from 726 to 1478 at 2 m/s, 1271 to 2040 N at 3 m/s, and 1889 to 2630 N at 4 m/s. Corresponding values of F_{neck_atten} ranged from 38.4% to −6.4% (mean (SD)=20.2 (13.1)%) at 2 m/s, from 40% to 2.5% (mean (SD)=20.2 (9.1)%) at 3 m/s, and from 30.1% to 2.4% (mean (SD)=15.2% (6.6)%) at 4 m/s.

Results from univariate regression analyses are summarized in Tables 3 and 4. When all hip protectors were combined (Table 3) the variables that associated significantly with F_{neck_atten} differed with impact velocity ($material_{type}$, $geometry_{type}$, k_{500}, E_{abs}, and $thick_{wearing}$ at 2 m/s; $geometry_{type}$, k_{500}, E_{abs}, E_{dis}, and $thick_{wearing}$ at 3 m/s; k_{500}, E_{abs}, E_{dis}, and $thick_{wearing}$ at 4 m/s). As $geometry_{type}$ was a significant predictor of F_{neck_atten} in the 2 and 3 m/s conditions, we performed subset analyses at these impact velocities. There were no significant univariate predictors of F_{neck_atten} for the pads that bridged over the greater trochanter (likely related to the small sample size of 5), so multivariate regression analyses were not performed for this subset of hip protectors. In contrast, for pads that touched the skin overlying greater trochanter (Table 4) there were six common univariate predictors of F_{neck_atten} in the 2 and 3 m/s datasets (k_{500}, E_{abs}, *width*, $thick_{pad}$, $thick_{wearing}$, and *volume*), in addition to E_{dis} for the 3 m/s condition.

The final predictive models from the multivariate regression analyses are presented in Tables 5 and 6. When all hip protectors were combined (Table 5), lower k_{500} values (i.e. lower stiffness at 500 N) were associated with greater force attenuation at all impact velocities (Fig. 5). Specifically, a 100 kN/m decrease in k_{500} was associated with an absolute increase in F_{neck_atten} of 8.3% at 2 m/s, 4.3% at 3 m/s, and 3.2% at 4 m/s. In addition, hip protectors that absorbed more energy (E_{abs}) were associated with greater attenuation in femoral neck force (F_{neck_atten}) in the 3 and 4 m/s conditions. Pads that did not contact the skin overlying the greater trochanter attenuated (on average) 1.9-fold more force at 2 m/s, and 1.5-fold more force at 3 m/s compared to pads that directly covered the greater trochanter.

The results of the multivariate regression analyses for the subset of protectors that touch the skin overlying the greater trochanter are presented in Table 6. Significant predictors of F_{neck_atten} included increased *width*, $thick_{pad}$, and $thick_{wearing}$ for the 2 m/s data, and E_{dis}, *width*, $thick_{pad}$, and $thick_{wearing}$ for the 3 m/s condition.

4. Discussion

In the current study, we used regression analyses to determine whether the force attenuation (biomechanical effectiveness) provided by 26 commercially available hip protectors could be predicted by factors including general design characteristics, geometric parameters or force–deflection properties. We observed a wide range among these devices in biomechanical performance: at an impact velocity of 3 m/s, force attenuation ranged from 2.5% to 40%. We found that hip protectors with lower initial stiffness attenuated greater force at all impact velocities.

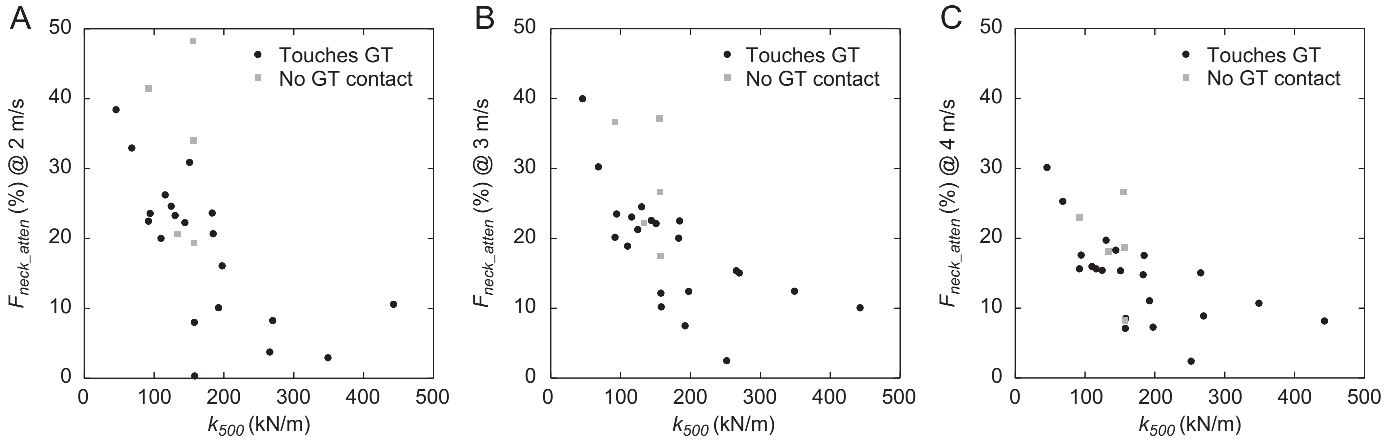

Fig. 5. Scatterplots of F_{neck_atten} versus k_{500} broken down by *geometry*$_{type}$ (i.e. protectors that do or do not touch the skin overlying the greater trochanter) at the: (A) 2 m/s, (B) 3 m/s, and (C) 4 m/s impact velocity conditions.

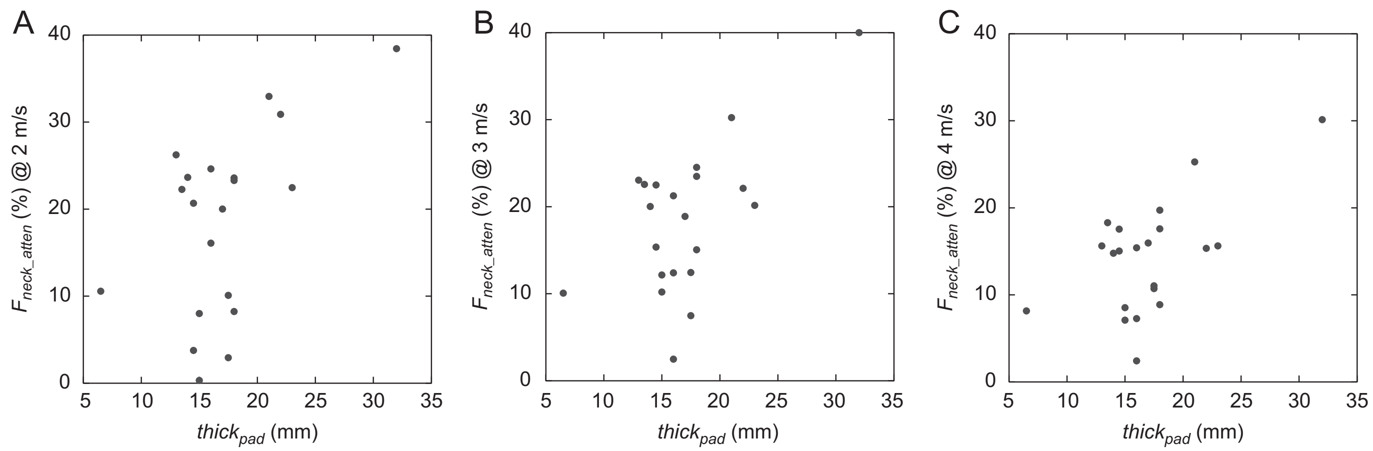

Fig. 6. Scatterplots of F_{neck_atten} versus $thick_{pad}$ at the: (A) 2 m/s, (B) 3 m/s, and (C) 4 m/s impact velocity conditions for the subset of hip protectors that directly touch the skin overlying the greater trochanter.

Protectors that absorbed more energy demonstrated greater force attenuation at the higher impact velocities (3 and 4 m/s conditions), while protectors that did not directly cover (but instead bridged) the skin overlying the greater trochanter attenuated up to 1.9-fold more force in the mild and moderate fall conditions (2 and 3 m/s impact velocities). Subset analyses at these lower velocities demonstrated that force attenuation for protectors that rested against the skin overlying the greater trochanter increased with increasing pad width, thickness, and energy dissipation.

Our results suggest that, among the range of hip protectors we tested, biomechanical effectiveness depends more on the general geometry type than the general material type. For example, force attenuation did not associate with *material*$_{type}$, with the top four performing protectors consisting of two soft shell designs (HipEase and Pelican Super Soft) and two hard shell products (KPH and HIPS). On the other hand, hip protectors that bridged over the proximal femur were associated with greater force attenuation at low-to-moderate impact velocities, likely due to their enhanced ability to shunt energy away from the proximal femur and into the iliac crest (superiorly) and surrounding soft tissues (anteriorly and posteriorly). These results agree with Kannus et al. (1999) and van Schoor et al. (2006) who also observed superior performance of energy-shunting hip protectors for low and/or moderate energy levels. However, we found that these designs provided no benefit over those that directly covered the proximal femur for higher severity falls (*geometry*$_{type}$ dropped out of the final model for the 4 m/s condition), perhaps due to a 'bottoming-out' of the protective bridge over/around the proximal femur. These findings agree with Derler et al. (2005) who reported that, while no general advantage was evident for energy-shunting vs. energy-absorbing designs, the three energy-shunting hip protectors they tested were associated with slightly higher femoral neck loads at high impact energies. Additional studies are warranted to assess whether increasing the thickness of these protectors enhances their biomechanical effectiveness at higher impact velocities.

Having said that, we did find that several force–deflection properties (which depend in turn on both material and geometrical properties) associated with the force attenuation provided by hip protectors. For example, protectors with lower stiffness (measured at 500 N of applied load) were associated with increased force attenuation. Low initial stiffness likely influences the pad's ability to deflect and absorb energy before stiffening or "bottoming out" at higher forces; energy absorption was in turn observed to associate positively with force attenuation. Interestingly, three of the four protectors with the highest E_{abs} values employed hard shell designs (Table 2). This indicates that hard shell protectors – traditionally considered to confer benefits through energy shunting – provide benefits also through deformation and energy absorption in their stiff dome or shell-like elements during impact from a fall.

We found that pad geometry was an especially important predictor of force attenuation for hip protectors that are in direct contact with the skin overlying the greater trochanter. For this group of 21 hip protectors, regression coefficients indicate that a 1 mm increase in pad thickness (Table 6; Fig. 6) was associated with an absolute F_{neck_atten} increase of 5.6% for 2 m/s impacts, and of 3.1% for 3 m/s impacts (after adjusting for *width*, $thick_{wearing}$, and E_{dis}). Increasing the thickness of protective devices has previously been shown to increase force attenuation during sideways falls on the hip (Laing et al., 2006; Nabhani and Bamford, 2004; Parkkari et al., 1994; Robinovitch et al., 1995), likely due to decreased stiffness and increased energy absorption. The observed effect of anterior–posterior pad width on force attenuation is also logical, as a wider pad allows energy to be shunted further from the proximal femur. Despite these distinct biomechanical benefits, hip protector manufacturers are hesitant in developing products with high thickness and width, which users might perceive as 'bulky'. Clearly, studies are required in various populations (high risk individuals, care providers, and families) to assess how hip protector geometry affects user acceptance and adherence, and determine whether these design trends are justified.

Our results are important for several reasons. First, they should help to guide consumers in selecting between available devices, based on the biomechanical performance. Second, they should assist researchers in selecting devices for future clinical trials, and interpreting the results of previous trials. Finally, by highlighting the geometric and material parameters that influence biomechanical performance, they should assist manufacturers in designing devices that offer improved performance and clinical effectiveness.

There are notable limitations to the current study. First, our sample of 26 hip protectors reflects the range of devices we were able to secure in 2009. Only five of these were classified as 'hard shell', or as protectors that bridged over the greater trochanter, reflecting the popularity of 'soft shell' products in the marketplace. Second, we only assessed the biomechanical effectiveness of hip protectors positioned over the hip according to manufacturer instructions. Studies should further explore the effect of pad 'shifting' on the biomechanical effectiveness of hip protectors (Choi et al., 2010; Schmitt et al., 2004), and whether the predictive factors of force attenuation observed in the current study remain consistent for protectors that are not optimally positioned. Third, we only assessed the ability of hip protectors to attenuate force during sideways falls causing impact to the lateral aspect of the hip. Although falls may produce a range of impact configurations, we focused on sideways impacts as they are linked most closely to hip fracture (Hayes et al., 1993; Nevitt and Cummings, 1993), and most hip protectors are designed primarily to reduce impact force during lateral impacts. Fourth, as the clinical effectiveness of hip protectors depends on user compliance (Forsen et al., 2004; O'Halloran, 2006; van Schoor et al., 2002), additional research is required to determine how the factors we found to influence biomechanical effectiveness affect user acceptance and adherence. Fifth, the resolution of $thick_{wearing}$ and *volume* were relatively coarse. Future research could employ 3D scanning systems to better characterize geometric parameters of hip protectors. Sixth, the peak femoral neck force (F_{neck}) measured during the unpadded condition (a mean of 2698 N for the 4 m/s impact velocity condition) is lower than the 3.5–4.5 kN range included in recent test system recommendations (Robinovitch et al., 2009). However, the theoretical bases for these standards were derived from mathematical predictions of total impact force (incorporating effective pelvis mass, stiffness, and impact velocity), and the peak total impact force measured by the force plate (mean = 3880 N) in the current study was within this recommended range. Finally, this paper does not report on the potential influence of pad displacement (i.e. shifting from the manufacturers' suggested positioning) on the biomechanical effectiveness of each hip protector.

In summary, this study illustrates that there is wide variation in the force attenuation provided by currently marketed hip protectors when positioned correctly over the greater trochanter, and shows that a range of material and geometrical variables affect biomechanical effectiveness. When compared to hip protectors that directly cover the greater trochanter, hip protectors that bridged over the greater trochanter provided 1.5 to 1.9-fold greater force attenuation. Force-attenuation also increased with decreasing stiffness at 500 N, and higher energy absorption. For protectors that directly covered the skin overlying the greater trochanter, force attenuation also increased with increases in pad width and thickness. By comparing the protective value of a large range of existing hip protectors, these results help to guide consumers and researchers in selecting hip protectors, and in interpreting the results of previous clinical trials. Furthermore, by determining geometric and material parameters that influenced biomechanical performance, our results should assist manufacturers in designing devices that offer improved performance and clinical effectiveness.

Conflict of interest

Independent of this project, SNR has received funding from Tytex A/S (the producer of the Safehip line of hip protectors) in support of hip protector research. SNR is also a co-inventor on a patent licensed to Tytex A/S, and forming the basis of the Safehip Soft product tested in this study. In 2006, ACL received funding from Tytex A/S to support travel to a conference. These relationships had no bearing on any aspect of the study design, data analysis or interpretation of results. No person other than the authors had input into the study design and subsequent reporting of data.

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Appendix A. Supplementary materials

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jbiomech.2011.08.016.

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