Medical Engineering & Physics xxx (2011) xxx–xxx

Contents lists available at SciVerse ScienceDirect

Medical Engineering & Physics

journal homepage: www.elsevier.com/locate/medengphy

The influence of headform orientation and flooring systems on impact dynamics during simulated fall-related head impacts

Alexander D. Wright, Andrew C. Laing*

Injury Biomechanics and Aging Laboratory, Department of Kinesiology, University of Waterloo, Waterloo, Ontario, Canada

ARTICLE INFO

Article history:
Received 25 March 2011
Received in revised form
13 November 2011
Accepted 15 November 2011

Keywords:
Traumatic brain injury
Falls
Compliant floors
Injury Prevention
Head impact biomechanics
Aging

ABSTRACT

Novel compliant flooring systems are a promising approach for reducing fall-related injuries in seniors, as they may provide up to 50% attenuation in peak force during simulated hip impacts while eliciting only minimal influences on balance. This study aimed to determine the protective capacity of novel compliant floors during simulated 'high severity' head impacts compared to common flooring systems.

A headform was impacted onto a common Commercial-Carpet at 1.5, 2.5, and 3.5 m/s in front, back, and side orientations using a mechanical drop tower. Peak impact force applied to the headform (F_{max}), peak linear acceleration of the headform (g_{max}) and Head Injury Criterion (*HIC*) were determined. For the 3.5 m/s trials, backwards-oriented impacts were associated with the highest F_{max} and *HIC* values ($p < 0.001$); accordingly, this head orientation was used to complete additional trials on three common floors (Resilient Rubber, Residential-Loop Carpet, Berber Carpet) and six novel compliant floors at each impact velocity. ANOVAs indicated that flooring type was associated with all parameters at each impact velocity ($p < 0.001$). Compared to impacts on the Commercial Carpet, Dunnett's post hoc indicated all variables were smaller (25–80%) for the novel compliant floors ($p < 0.001$), but larger for Resilient Rubber (31–159%, $p < 0.01$).

This study demonstrates that during 'high severity' simulated impacts, novel compliant floors can substantially reduce the forces and accelerations applied to a headform compared to common floors including carpet and resilient rubber. In combination with reports of minimal balance impairments, these findings support the promise of novel compliant floors as a biomechanically effective strategy for reducing fall-related injuries including traumatic brain injuries and skull fractures.

© 2011 IPEM. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Fall-related injuries in adults over the age of 65 are a major public health issue in Canada, and are associated with direct annual costs of over $2 billion [1]. A substantial portion of this figure may be attributed to fall-related traumatic brain injuries (TBI), which are precipitated by falls in up to 90% of cases [2]. Seniors are hospitalized twice as often as the general population for fall-related TBI, while over half of all fall-related deaths in older adults are due to TBI [3]. The incidence of fall-induced TBI and associated deaths has been rising at alarming rates, increasing by over 25% between 1989 and 1998 [4]. The risk for fall-related TBI increases substantially with age; persons over the age of 85 are hospitalized for fall-related TBI over twice as often as those aged 75–84, and over 6 times as often as those aged 65–74 [5]. Although initial improvements in health outcomes are common following TBI, these types of injuries often lead to residual disability. Thus, prevention remains the optimal approach for reducing associated injury and disability [4]. Considering the ageing Canadian population [6], it is imperative that effective intervention strategies be designed and implemented to stem the social and economic impact of the anticipated rise in fall-related TBI incidence over the coming decades.

Development of effective intervention strategies necessitates an understanding of the cause of TBI. While the exact pathway between mechanical insult and cognitive deficit is not yet fully understood [7], it is generally recognized that the majority of fall-related TBI occur as a result of the head directly striking another surface [8,9]. Even without fracture of the skull, direct impact can cause linear and rotational accelerations of the brain within the brain cavity, creating pressure fluctuations and shear strains that may lead to the tearing of small blood vessels and widespread disruption of axons [8,10–13]. The type and severity of intracranial

* Corresponding author at: Department of Kinesiology, Faculty of Applied Health Sciences, University of Waterloo, 200 University Avenue West, Waterloo, ON, Canada N2L 3G1. Tel.: +1 519 888 4567x38947.
E-mail address: actlaing@uwaterloo.ca (A.C. Laing).

1350-4533/$ – see front matter © 2011 IPEM. Published by Elsevier Ltd. All rights reserved.
doi:10.1016/j.medengphy.2011.11.012

injuries resulting from direct head impact, including intracranial haemorrhaging and diffuse axonal injuries, is highly influenced by the mechanical properties of the impact surface [14–16]. Indeed, previous research reports that unsuitable surfacing has been found to account for between 79 and 100% of severe head injuries in playground environments [17].

Towards the goal of reducing fall-related TBI in older adults, one promising approach entails the installation of novel compliant flooring systems. Novel compliant flooring systems (NCFs) are generally designed to provide a dual-stiffness response characterized by minimal deflection during locomotion, and a transition to increased compliance at the higher loads associated with fall-related impacts. For example, one design type incorporates a continuous top surface overlaying an array of rubber columns that buckle once a critical load threshold is reached. Certain models of these commercially available products have been shown to attenuate the impact force applied to the proximal femur by up to 50% during simulated lateral falls compared to commercial-grade vinyl [18], suggesting a significant protective capacity against hip fractures. This degree of force attenuation is far greater than levels that have been reported for common single-stiffness surfaces including wooden floors (7%), carpets (15%), and carpets with underpadding (24%) [19–21]. However, no independently obtained information is currently available with respect to the influence of common floors versus novel compliant flooring systems on impact dynamics during simulated head impacts.

Evaluation of head impact dynamics is commonly accomplished using mechanical impact simulators. Such tests have found widespread use in the development of safety standards for devices including helmets, airbags, and playground surfaces. Many headforms have been developed to match the anthropometric characteristics of 'average' human heads, including the Hybrid III and FOCUS headforms. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) has also developed biofidelic headforms, which include a glycerin-filled 'brain cavity' to optimally simulate the behaviour of the human head in response to impact [22,23]. Decades of head impact research have produced risk curves and associated injury thresholds for skull fracture and TBI following impact based on force and acceleration profiles, as well as derived injury criteria such as the Head Injury Criterion (*HIC*) [24–28]. Simulated head impacts have been widely used to evaluate head injury risk, including during falls on taekwondo mats [29], falls onto playground surfaces [27], and impacts during athletic competition [30]. Despite the widespread use of simulated head impacts using headforms, the effect of headform orientation, and consequent impact location, has rarely been reported.

Accordingly, our objectives in the current study were to determine: (a) the 'high severity' orientation for simulated head impacts using a biofidelic surrogate human headform based on measures associated with risk for skull fracture and TBI including peak resultant acceleration of the headform centre of gravity (g_{max}), Head Injury Criterion score (*HIC*), and peak impact force applied to the headform (F_{max}); and (b) the influence of 10 flooring surfaces on these outcome variables during 'high severity' impacts, relative to a common compliant flooring surface (commercial-grade carpet with underpadding). We hypothesized that the added compliance associated with the headform's ear (during side impacts) and nose (during front impacts) would lead to reductions in the magnitudes of all outcome variables compared to impacts of the back of the head. Furthermore, we hypothesized that during impacts in the 'worst case' head orientation, impacts onto novel compliant flooring systems would result in lower applied forces and accelerations (e.g. g_{max}, *HIC*, and F_{max}) compared to impacts onto a commercial-grade carpet. Finally, we also hypothesized that the commercial carpet would provide significant force and acceleration attenuation relative to a commercial-grade resilient rubber floor.

2. Materials and methods

2.1. Test system

A mechanical drop tower [31] was used to impact a medium-sized surrogate human head developed by the National Operating Committee for Standards on Athletic Equipment (NOCSAE) onto various flooring surfaces (Fig. 1). Detailed specifications have been reported previously [22,23], but in brief, the headform was comprised on a glycerin-filled acrylonitrile butadiene styrene (ABS plastic) brain cavity, surrounded by separate urethane skull and facial features. An adjustable mechanical release enabled impacts at velocities of 1.5, 2.5, and 3.5 m/s, which were tracked using an infrared light gate velocimeter (Model VS300, GHI Systems, Aurora, ON, Canada) positioned immediately above the impact surface (Fig. 1). The accuracy of the calculated impact velocity values was verified using optical tracking of infrared emitting markers on the headform and headform carriage assembly (Optotrak Certus, Northern Digital Inc., Waterloo, Canada). Using energy conservation principles, the most severe impact velocity theoretically represents a pure free-fall onto the head from a height of 62.5 cm, representing a potential scenario involving a fall out of bed. Although the precise characteristics of typical falls in older adults that lead to head injury are not known, it was anticipated that, during a fall from standing position (initial height of the head ~1.8 m), the impact velocity of the head would be substantially lower than theoretical free-fall calculations (~5.9 m/s) due to protective responses and initial contact of other body parts with the ground. Pilot testing and previous research using mechanical impact simulators to assess impact dynamics of the hip further informed the chosen impact velocities [18]. A triaxial accelerometer (Model 2707A, Endevco Corporation, San Juan Capistrano, CA, USA) mounted at the centre of mass of the headform recorded impact accelerations, while a load cell (Model 925M113, Kistler Instrument Corporation, Amherst, NY, USA) mounted beneath the impact surface measured impact forces. Force and acceleration data were sampled at 20,000 Hz. In all cases, 3 sequential trials were completed for each impact condition. The flooring samples were moved between trials to prevent repeated impacts onto the same location, and in all cases were positioned on top of a plywood interface surface which supported the entire sample and simulated wooden subflooring (Fig. 1).

2.2. Determining the 'high severity' impact orientation

A level loop 'Commercial Carpet' ($Carpet_{comm}$), pile height = 6 mm, face weight = 882 g/m^2 with 6 mm underpad (a common compliant flooring system often found in commercial housing settings) was used as a control condition in this study. In order to determine the 'high severity' impact orientation, trials were conducted onto $Carpet_{comm}$ flooring using three headform orientations (front, side, and back) at three impact velocities (1.5, 2.5, and 3.5 m/s), with order of condition combination randomly determined.

2.3. Novel compliant floors versus common flooring systems

Nine additional flooring conditions were tested in this study under the 'high severity' impact orientation at 1.5, 2.5, and 3.5 m/s impact velocities. These included institutional-grade resilient sheeting, two additional common compliant floors (Residential-Loop Carpet, Berber Carpet) and six NCF conditions (Fig. 1b). The former entailed 2 mm thick resilient rolled sheeting made from a matrix of rubber and natural fillers (*Resilient*) appropriate for installation over concrete or wooden subfloors in institutional settings (Noraplan Classic, Nora Systems Inc, Lawrence,

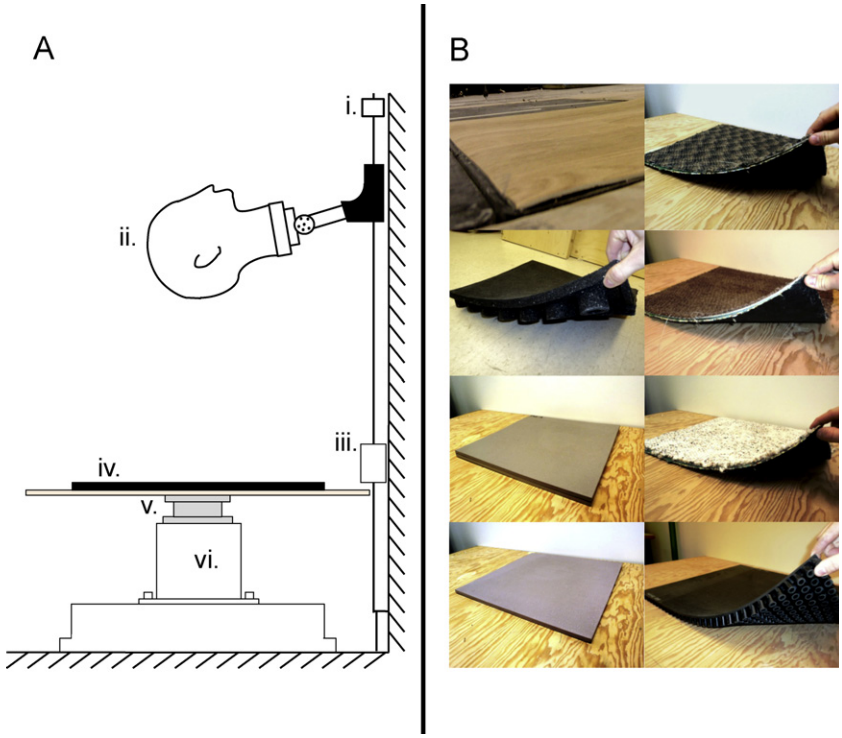

Fig. 1. (A) Schematic of the mechanical head impact simulator with the following elements highlighted: (i) mechanical release; (ii) headform with accelerometer mounted at centre of mass; (iii) light gate velocimeter; (iv) flooring sample positioned on plywood interface surface; (v) load cell; (vi) concrete base. (B) Pictures of the floor conditions tested (clockwise from top left): Resilient (*Resilient*), Commercial Carpet ($Carpet_{comm}$), Residential-Loop Carpet ($Carpet_{res\text{-}loop}$), Berber Carpet ($Carpet_{berber}$), SmartCell (*SmartCell*), 12 mm Kradal ($Kradal_{12\,mm}$), 24 mm Kradal ($Kradal_{24\,mm}$), and SofTile (*SofTile*); not shown in this figure are SmartCell with Resilient overlay ($SmartCell_{resilient}$) and SofTile with Resilient overlay ($SofTile_{resilient}$).

MA, USA). The 'Residential-Loop Carpet' ($Carpet_{res\text{-}loop}$) condition entailed a polypropylene pile-loop carpet (pile height = 9 mm, face weight = 1085 g/m^2) with 6 mm foam-rubber underpadding designed for residential settings. The 'Berber Carpet' ($Carpet_{berber}$) condition was the thickest of the carpets, consisting of a synthetic weave looped polypropylene (pile height = 10 mm, face weight = 1221 g/m^2) also used primarily in residential settings. Six NCF conditions were also tested. '*SmartCell*' (SATech, Chehalis, WA, USA) was a 25 mm tall synthetic, 50-durometer rubber flooring system comprising a continuous surface layer overlying a series of cylindrical rubber columns 14 mm in diameter and 19 mm apart, centre-to-centre. The '*SofTile*' floor (SofSurfaces, Petrolia, ON, Canada) used a similar design, with 50 mm diameter columns spaced at 70 mm intervals; we tested the 50 mm thick model. The *SofTile* and *SmartCell* floors were also tested with a resilient rubber overlay ($SofTile_{resilient}$ and $SmartCell_{resilient}$, respectively), representing a design scenario likely necessary for clinical settings. Two designs from Kradal (Acma Industries Ltd., Upper Hutt, Wellington, New Zealand) comprised the final NCF conditions, including 12 mm ($Kradal_{12\,mm}$) and 24 mm ($Kradal_{24\,mm}$) thick tiles with a relatively stiff top surface over a closed cell polyurethane base layer that incorporates suspended micro-spheres. During testing, the order of floor-velocity combination was randomly determined.

2.4. Data analysis

Accelerometer data was processed according to ASTM Standard F1292-04 for testing impact attenuation of surfacing materials during simulated head impacts [27]. In brief, a fourth-order, dual-pass, low-pass digital Butterworth filter (1000 Hz cutoff) was used before calculating the resultant acceleration from the accelerations in each of the three orthogonal axes. g_{max} was recorded as the single largest value from the resultant acceleration–time history for each impact.

The *HIC* score was also calculated for each impact, according to the following equation [27]:

$$HIC = \max\left[(T_1 - T_0)\left[\frac{1}{T_1 - T_0}\int_{t=T_0}^{T_1} a_r\,dt\right]^{2.5}\right] \quad (1)$$

where a_r is the resultant acceleration profile and T_0 and T_1 define the time interval that maximizes the *HIC* score. F_{max} (Fig. 2) was determined from the force–time profile after filtering using a dual-pass, low-pass 4th order digital Butterworth (500 Hz cutoff).

2.5. Statistics

2.5.1. Determination of the 'high severity' headform orientation

A two-way ANOVA was used to assess the influence of impact orientation and impact velocity on g_{max}, *HIC*, and F_{max}. When significant interactions were found, simple effects were analyzed to determine the influence of impact orientation at each impact velocity, with Tukey's post hoc used to compare across the three orientations.

2.5.2. Floor testing

A two-way ANOVA was used to assess the influence of floor condition and impact velocity on each of the outcome parameters. If a significant interaction was found, simple effects were analyzed to determine the influence of floor condition at each impact velocity. Dunnett's post hoc test (which is appropriate when a baseline comparator condition exists) was used to compare each floor relative to the control condition, $Carpet_{comm}$.

To account for the use of three dependent variables, we used an alpha of 0.0167 (i.e. 0.05/3) for ANOVAs. Post hoc tests were conducted with an experiment-wide significance level of 0.05 using

Please cite this article in press as: Wright AD, Laing AC. The influence of headform orientation and flooring systems on impact dynamics during simulated fall-related head impacts. Med Eng Phys (2011), doi:10.1016/j.medengphy.2011.11.012

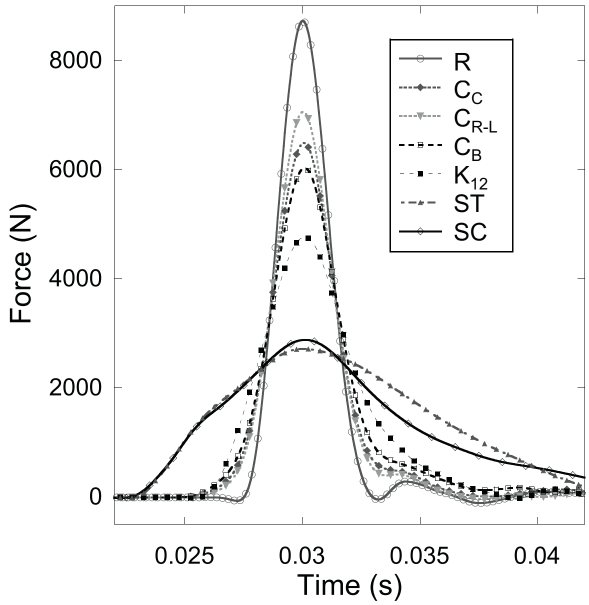

Fig. 2. Representative force versus time profiles from 2.5 m/s impacts onto a subset of the floor conditions tested, including Resilient (R), Commercial Carpet (C_C), Residential-Loop Carpet (C_{R-L}), Berber Carpet (C_B), 12 mm Kradal (K_{12}), SofTile (ST), and SmartCell (SC).

Table 1
Mean (SD) of peak resultant acceleration (g_{max}) and Head Injury Criterion (*HIC*) for impacts onto the front, side, and back of the headform at each impact velocity on the control Commercial Carpet condition. These results informed our decision to use impacts on the back of the headform to compare across floor conditions.

Variable	Orientation	Impact velocity (m/s)		
		1.5	2.5	3.5
g_{max} (g)	*Front*	30.7 (0.4)	62.6 (2.0)	94.1 (5.7)
	Side	62.8 (7.0)*	123.3 (5.3)*	263.0 (9.9)*
	Back	54.7 (3.4)*	122.7 (3.8)*	262.1 (11.1)*
HIC	*Front*	27.0 (1.3)	107.9 (9.1)	250.4 (15.8)
	Side	48.8 (9.3)*	282.8 (58.4)*	827.9 (29.0)*
	Back	39.1 (3.9)	258.0 (23.7)*	1068.0 (40.6)*,[a]

[a] Significantly greater than Side orientation ($p < 0.05$).
* Significantly greater than Front orientation ($p < 0.05$).

SPSS statistical software package (Version 19.0, SPSS Inc., Chicago, IL, USA).

3. Results

3.1. 'High severity' orientation

Results from two-way ANOVAs indicated a significant interaction between impact orientation and impact velocity for all outcome parameters (p always < 0.001). Subsequent ANOVA results indicated a significant effect of orientation for all variables at each impact velocity (p always < 0.001). Tukey's post hoc indicated that *Back* and *Side* impacts consistently yielded higher g_{max}, F_{max}, and *HIC* values when compared to *Front* impacts at all impact velocities (Table 1 and Fig. 3). During impacts at velocities of 1.5 and 2.5 m/s, no differences in any of the outcome parameters were found between *Back* and *Side* orientations. At 3.5 m/s, g_{max} values were not different, however *HIC* and F_{max} values were significantly greater for *Back* impacts.

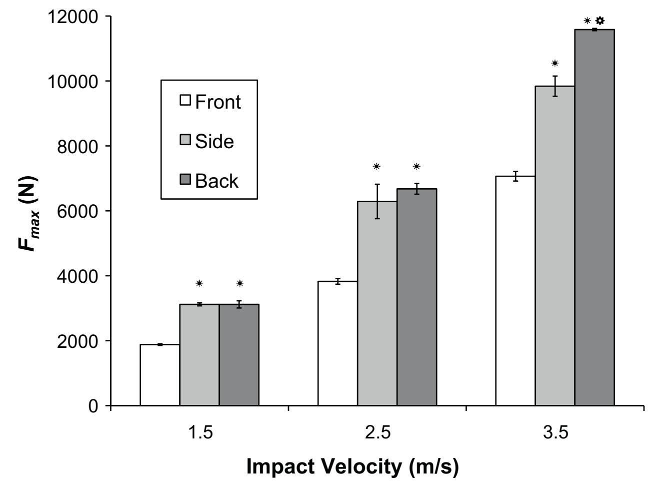

Fig. 3. Mean (SD) of peak force (F_{max}) for impacts onto the front, side, and back of the headform at each impact velocity for the control Commercial Carpet condition. * indicates significant ($p < 0.05$) increase relative to front, and ✽ indicates significant ($p < 0.05$) increase relative to side.

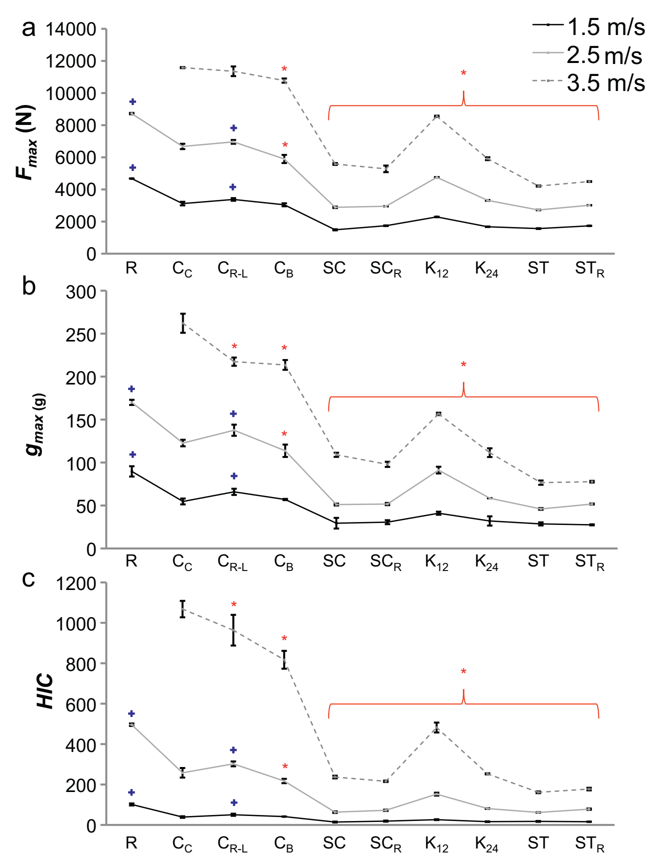

Fig. 4. Mean (SD) for: (a) peak impact force (F_{max}); (b) peak resultant acceleration (g_{max}), and (c) Head Injury Criterion (*HIC*) score for simulated head impacts across floor condition at each impact velocity. Floor conditions include Resilient (R), Commercial Carpet (C_C), Residential-Loop Carpet (C_{R-L}), Berber Carpet (C_B), SmartCell (SC), SmartCell with Resilient overlay (SC_R), 12 mm Kradal (K_{12}), 24 mm Kradal (K_{24}), SofTile (ST), and SofTile with Resilient overlay (ST_R). * indicates significant ($p < 0.05$) decrease compared to C_C floor, and + indicates significant ($p < 0.05$) increase compared to C_C floor for a given impact velocity. Note that all variables were significantly reduced for impacts onto novel compliant flooring systems at all impact velocities.

3.2. NCFs versus common flooring surfaces

Based on the results in Section 3.1, all additional testing was completed using impacts to the *Back* of the headform. The data is summarized below and in Fig. 4. It should be noted that impacts at 3.5 m/s were not conducted for the *Resilient* floor in order to protect the mechanical integrity of our testing system.

3.2.1. Peak acceleration (g_{max})

Peak accelerations ranged from 54 to 262 *g* for impacts onto the carpet conditions, from 90 to 170 *g* onto the *Resilient* floor, and from 27 to 157 *g* on the NCFs (Fig. 4). ANOVA indicated a significant interaction between floor condition and impact velocity ($F = 137.6$, $p < 0.001$). At 1.5 m/s, there was a significant effect of floor ($F = 92.6$, $p < 0.001$). Dunnett's post hoc demonstrated that, compared to $Carpet_{comm}$, peak accelerations were lower for all of the NCF conditions (p always ≤ 0.002). Similar effects were found for impacts at 2.5 m/s ($F = 431.4$, $p < 0.001$) and 3.5 m/s ($F = 558.7$, $p < 0.001$), whereby peak accelerations were consistently lower for the NCF conditions compared to $Carpet_{comm}$ (p always < 0.001). Across all three impact velocities, g_{max} was attenuated by at least 25% and up to 70% for impacts onto NCFs compared to $Carpet_{comm}$.

During impacts at both 1.5 m/s and 2.5 m/s, g_{max} was significantly larger for impacts onto *Resilient* (64% and 39% larger, respectively (p always < 0.001)) relative to $Carpet_{comm}$. We also observed this effect for impacts onto $Carpet_{res\text{-}loop}$ at 1.5 m/s and 2.5 m/s (20% larger, $p = 0.01$; and 12% larger, $p = 0.001$, respectively) relative to $Carpet_{comm}$. However, at 3.5 m/s, g_{max} was 17% lower for $Carpet_{res\text{-}loop}$ relative to *CC* ($p < 0.001$). $Carpet_{berber}$ impacts were not different from $Carpet_{comm}$ at 1.5 m/s ($p = 0.975$), but yielded reduced g_{max} values at 2.5 m/s (7%, $p = 0.047$) and 3.5 m/s (18%, $p < 0.001$).

3.2.2. Head Injury Criterion (HIC)

HIC scores ranged from 39 to 1068 for carpeted conditions across all tested impact velocities, between 101 and 496 onto the *Resilient* floor (not tested at 3.5 m/s), and from 14 to 482 onto NCF conditions (Fig. 4). A significant interaction was found between floor condition and impact velocity whereby the attenuation in *HIC* scores provided by NCFs increased as impact velocity increased ($F = 268.3$, $p < 0.001$). Subsequent one-way ANOVAs indicated that floor condition was associated with *HIC* at each impact velocity ($F = 236.5$, 640.2, 356.5 at 1.5, 2.5, and 3.5 m/s respectively, p always < 0.001). Dunnett's post hoc revealed that *HIC* scores were consistently lower for impacts onto NCFs relative to $Carpet_{comm}$. NCFs reduced *HIC* scores by 33–63% at 1.5 m/s, by 41–76% at 2.5 m/s, and by 55–85% for impacts at 3.5 m/s (p always < 0.001).

Similarly to the results for g_{max}, the *HIC* scores for impacts onto the *Resilient* floor were 159% larger than those onto $Carpet_{comm}$ at 1.5 m/s ($p < 0.001$), and 92% larger at 2.5 m/s ($p < 0.001$). *HIC* was significantly larger for impacts onto $Carpet_{res\text{-}loop}$ compared to $Carpet_{comm}$ at 1.5 m/s ($p = 0.001$) and 2.5 m/s ($p < 0.001$), but was reduced at 3.5 m/s ($p = 0.008$). Compared to $Carpet_{comm}$, *HIC* was not different for impacts onto $Carpet_{berber}$ at 1.5 m/s ($p = 0.92$), but was significantly lower at both 2.5 m/s and 3.5 m/s ($p < 0.001$).

3.2.3. Peak force (F_{max})

Peak impact force across impact velocities ranged from 3045 to 11,583 N for impacts onto the carpet conditions, from 4676 to 8721 N for the *Resilient* condition, and between 1487 and 8552 N onto the NCFs (Fig. 4). We observed a significant interaction between floor condition and impact velocity ($F = 395.7$, $p < 0.001$). During impacts at 1.5, 2.5, and 3.5 m/s, floor was significantly associated with F_{max} ($F = 1085$, 1252, 1522 respectively, p always < 0.001). Post hoc analysis provided that, compared to $Carpet_{comm}$, F_{max} was always significantly lower for impacts onto NCFs (p always < 0.001). At 1.5 m/s, peak force attenuation provided by the NCFs ranged from 27 to 52%, similar to that at 2.5 m/s (29–59%) and 3.5 m/s (26–64%).

$Carpet_{comm}$ provided some force attenuation relative to *Resilient*; F_{max} values were 50% larger for impacts onto *Resilient* at 1.5 m/s, and 31% larger for impacts at 2.5 m/s. F_{max} was 8% larger for impacts onto $Carpet_{res\text{-}loop}$ at 1.5 m/s ($p < 0.001$), 4% larger at 2.5 m/s ($p = 0.023$), but not significantly different at 3.5 m/s ($p = 0.251$). Compared to $Carpet_{comm}$, impacts onto $Carpet_{berber}$ produced peak forces that were not significantly different at 1.5 m/s ($p = 0.493$); F_{max} was, however, 12% lower at 2.5 m/s and 7% lower at 3.5 m/s ($p < 0.001$).

4. Discussion

In the current study, we first examined the influence of headform orientation on indices of skull fracture and TBI risk and found that impacts onto the back of the headform represented the 'high severity' orientation based on resultant acceleration and force profiles. We then assessed the influence of flooring type on head impact dynamics during these 'high severity' impact scenarios. Our hypothesis that the headform would experience lower forces and accelerations during impacts onto novel compliant floors (NCFs) than onto the Commercial Carpet was supported in 54 of 54 possible comparisons (6 floors × 3 impact velocities × 3 variables (F_{max}, g_{max}, *HIC*)). Regarding our second hypothesis, we observed that impacts onto Commercial Carpet yielded significantly lower values for all outcome variables compared to *Resilient* in six of six possible comparisons (2 impact velocities × 3 variables). Although not compared statistically, it can be inferred that the outcomes for the NCFs would also be substantially reduced compared to *Resilient* based on their relationship to the Commercial Carpet. Interestingly, an interaction effect between floor condition and impact velocity was observed for all three outcome parameters. This interaction was generally characterized by increased attenuation in outcomes in the NCF conditions as impact velocity increased, suggesting that the protective capacity of these floors may be greater as impact severity increases. Overall, these results indicate that the NCFs tested in this study are capable of substantially reducing indices of skull fracture and TBI risk compared to common flooring materials during simulated falls involving head impacts.

Several possible explanations exist for our observation that backwards headform orientation was the most severe impact orientation we tested. First, the test system used in this study was rigidly mounted to the vertical guide rail, and so did not incorporate a biofidelic neck component (Fig. 1). As such, the differential range of motion and stiffness properties for neck flexion, extension, and lateral bending were not simulated. However, a consistent orientation of the headform with respect to the mounting system was used in all conditions, which allowed us to isolate the influence of headform characteristics on impact dynamics. Accordingly, another possible explanation for the high severity impacts being represented by the backwards orientation may relate to the construction of the NOCSAE headform itself. The headform is comprised of a high durometer urethane skull covered with a lower durometer urethane that forms the skin and anatomical features of the head (such as the nose, ears and lips). A mass-spring model of the headform-floor system suggests that impact orientations with the lowest effective stiffness will result in lower peak forces and accelerations. Hooke's law states that an element's stiffness is proportional to its cross-sectional area divided by length (in our case equivalent to the headform-floor contact area and headform urethane thickness, respectively). A Hertzian contact model suggests that the large radii of curvature for the back of the headform results in a larger contact area compared to the nose, while the thickness of the urethane is largest for the nose and lowest in the occipital region. These characteristics are predictive of increasing effective

Please cite this article in press as: Wright AD, Laing AC. The influence of headform orientation and flooring systems on impact dynamics during simulated fall-related head impacts. Med Eng Phys (2011), doi:10.1016/j.medengphy.2011.11.012

stiffness across the front, side, and back of the headform, respectively, which corresponds to our observation of increasing impact severity across these orientation conditions.

Our definition of the back of the headform as a 'high severity' impact orientation is specific to our test system, and is not intended to contribute to the discussion regarding the effect of impact location/direction on head injury risk during real-world falls involving head impact. Early studies suggested that real-world impacts to the lateral aspect of the human head are most likely to lead to concussion [32], which corresponds to finite-element models demonstrating a lower tolerance for lateral impacts compared to anterior–posterior or axial impacts [33–35]. In an analysis of head impacts experienced by collegiate football players, 46% of concussive impacts occurred to the top of the head, whereas frontal, lateral, and occipital impacts were responsible for 31%, 15%, and 8% of observed concussions, respectively. However, no significant correlation was found between impact location and clinical outcome severity [36]. The finding in the current study that impact severity was substantially affected by headform orientation during impact suggests that this factor should be considered and reported in future research involving simulated impacts with headforms. Furthermore, although the headform used in this study was chosen based on its high level of biofidelity including a glycerin-filled cavity to simulate brain movement [23], further development of headforms that aim to mimic the orientation-sensitive response of the human head might be warranted.

It is worthwhile to consider the observed F_{max} and *HIC* scores in context with proposed injury thresholds. Using free-falling impactors, the skull fracture thresholds of various cranial bones have been estimated by several groups. For example, Nahum and colleagues estimated a minimal force tolerance level of 3560–7117 N for the frontal bone [37]. More recently, through the use of acoustic emission sensors, Cormier et al. have suggested that forces between 1885 and 2405 N are associated with a 50% risk of frontal bone fracture [38]. While the peak forces observed in the current study were much greater than either of these proposed thresholds, the surface area of the impact interface must be considered before making inferences related to fracture risk. The impactors used in the aforementioned studies had a contact area of 6.45 cm^2. While the exact contact area between headform and flooring surface was not measured in this study, firmer floors (Resilient, Commercial Carpet) would have likely undergone smaller deformations following impact, leading to much smaller contact areas compared to the NCFs tested. Based on animal and cadaveric data, the expanded Prasad–Mertz curves suggest that a *HIC* score of 1000 is associated with a non-zero risk of a fatal head injury, an 18% probability of severe head injury, a 55% probability of a serious injury, and a 90% probability of moderate head injury for the average adult [25,26]. For impacts at 3.5 m/s onto Commercial Carpet, *HIC* scores exceeded this threshold (mean (SD) = 1068.0 (40.6)). Furthermore, the *HIC* for the *Resilient* floor condition was 495.9 (6.2) at an impact velocity of only 2.5 m/s. In contrast, the largest mean *HIC* value from 3.5 m/s impacts onto any of the NCFs was less than 500 ($Kradal_{12\,mm}$: 482.0 (24.5)), with all other NCF surfaces yielding average *HIC* scores of less than 300. In clinical terms, these results suggest that the risk of moderate head injury for an average adult is 5–25% for a fall involving 3.5 m/s head impact onto the novel compliant floors we tested, compared to an 80–90% risk onto the common compliant floors. For the NCFs, outcomes corresponded with floor thickness. For example, at 3.5 m/s the *HIC* was 161.5 (4.3) for the 50 mm *SofTile* product (likely the least appropriate for indoor implementation), compared to 237.0 (6.6) for the 25 mm *SmartCell*, and 482.0 (24.5) for the 12 mm $Kradal_{12\,mm}$ floor. Additional studies should be considered which investigate the design features that are most predictive of biomechanical effectiveness during head impact, in addition to practical factors including usability, durability, and ease of implementation.

Our results are in accordance with previous reports of the force attenuative properties of specific novel and common compliant flooring systems. Maki et al. [20] used a mechanical fall simulator to determine peak deceleration and peak force during simulated hip impacts onto common flooring surfaces (although they did not specify the impact velocity achieved). They report that, in comparison to impacts onto a vinyl floor similar to the *Resilient* condition used in the current study, padded carpets provided the greatest level of impact attenuation (up to 23%). Others have reported force attenuative values as high as 56% and 73% when incorporating PVC underlay beneath vinyl and carpet floors, respectively [39–41]; however, these values may overestimate the protective capacity of these flooring conditions as the effective compliance of the pelvic region was not incorporated into their testing system. Most recently, Laing and Robinovitch [18] reported that the same SmartCell floor tested in this study attenuated peak femoral impact force by 17.3% compared to a commercial-grade vinyl for impacts at 2 m/s, and 22.5% at 3 m/s, while a 100 mm thick SofTile product provided peak force attenuation of 44.9% and 45.5% at 2 and 3 m/s, respectively. In the current study, Commercial Carpet lowered peak forces by 30% compared to the *Resilient* floor for impacts at 2.5 m/s. Impacts onto the *SmartCell* floor produced peak forces that were at least 51% lower than the Commercial Carpet (thus, 80% lower than *Resilient*). *SofTile* reduced peak forces by at least 59% compared to Commercial Carpet (~90% lower than peak forces onto *Resilient*). These values are higher than those reported for simulated falls on the hip as our headform is likely much stiffer than the pelvis' effective stiffness of approximately 40 kN/m [42,43]. These data, in conjunction with additional studies that have assessed the influence of floor stiffness during falls on the upper limb [44] and buttocks [45], suggest that compliant floors have the potential to decrease the risk of a wide range of fall-related injuries.

For novel compliant floors to be an effective intervention strategy in reducing fall-related injuries, they must have the capacity to decrease impact loads and accelerations while having minimal concomitant influences on the balance and mobility of the target users. Numerous reports have established that some compliant surfaces may decrease postural stability and consequently increase the likelihood of falling. Compared to rigid surfaces, compliant foam surfaces have been associated with increased postural sway during quiet stance [46–49], as well as a lowered trajectory of the whole-body centre-of-mass (COM), reduced toe clearance, and increased step length, step width, and step width variability during gait [50,51]. Regarding common compliant flooring systems tested in this study, thick carpet has been shown to increase anterior–posterior sway for older adults when visual fields are altered [46], although these effects are not observed under normal vision conditions [52,53]. Regarding novel compliant flooring systems, Laing and Robinovitch [18] found that medial-lateral postural sway on a SmartCell floor was not different than on a rigid surface for community-dwelling elderly women, and that scores on the Timed Up and Go test (a predictor of fall risk [54–56]) were not different for SmartCell, SofTile and a rigid floor condition. Furthermore, Wright and Laing [57] found that the displacement profiles of both the centre of mass (a balance indicator) and the underfoot centre-of-pressure (a balance control variable) were not affected by SmartCell and SofTile floors in community-dwelling elderly women during backwards perturbations. Despite these encouraging findings, further research is needed to ascertain if and how balance control is affected on these floors during activities of daily living for older adults residing in high-risk settings (retirement homes, nursing homes, hospitals) where NCFs are most likely to be installed.

There were several limitations associated with this study, the majority of which are specific to the test apparatus. First, while little

Please cite this article in press as: Wright AD, Laing AC. The influence of headform orientation and flooring systems on impact dynamics during simulated fall-related head impacts. Med Eng Phys (2011), doi:10.1016/j.medengphy.2011.11.012

conclusive information is available with respect to the characteristics of 'typical' falls and subsequent head impacts experienced by older adults [58], it is unlikely that all injurious real-world falls involving head impact are characterized by the purely vertical cranial trajectory that our test system simulated. Although the relative importance of linear versus rotational accelerations in TBI pathology is still a matter of debate, rotational accelerations are nonetheless hypothesized to be a primary risk factor [59,60], and such rotational effects were minimized in our simulated impacts. However, the test method we used is similar to those used for national standards on assessing the protective capacity of playground surfaces [27] and sports helmets [22], allowing for comparisons of the protective capacity offered by these differing intervention strategies. Second, the Head Injury Criterion outcome that we reported is reliant on measurements of external linear kinematics of the head, and is not specific to direction of impact, nor is it able to reflect the response of the brain within the cranial cavity [12,61]. However, the complex etiology of head injuries makes it immensely difficult to establish accurate injury criteria and associated injury thresholds, so despite its limitations, the *HIC* represents the best and most widely used risk prediction tool currently available [16,59]. Lastly, the impact velocities used in this study may not reflect those experienced during 'typical' head impacts, which may be greater than 3.5 m/s. Pilot testing at higher impact velocities caused damage to our mechanical test system, including a rupture of one headform's glycerin-filled brain cavity. A compromise between headform biofidelity and external validity could be achieved through the use of a more durable headform (e.g. Hybrid III, FOCUS) to assess the protective capacity offered by flooring surfaces at higher impact velocities. However, future research is needed to characterize the inputs that should be incorporated into a biofidelic test method for simulating fall-related head impacts in older adults (e.g. distributions of head orientation, impact velocities, and load trajectories). In addition, the risk of injury across these loading scenarios must be characterized, similar to research being conducted for sports-related head impacts [33,36,59,61–66].

There are additional biomechanical issues that need to be studied to fully characterize the potential protective capacity of novel compliant floors during head impacts. For example, additional studies should investigate the potential influence of surface compliance on the rotational accelerations experienced within the brain cavity during oblique head impacts. Furthermore, the deformation of compliant floors around the skull during obliquely oriented head impacts might increase the system coefficient of friction leading to a slowing of the head's horizontal velocity and a concomitant increase in neck bending loads due to the inertia of the body. Future studies should characterize such factors towards the goal of developing flooring systems that optimize potentially competing demands for protection across a range of injury types. Furthermore, research is needed to investigate the host of additional factors that must be considered in determining the clinical feasibility of novel compliant floors, including cost-benefit analyses for new and retrofitted facilities [18,67], any additional demands that might be placed on the facility staff (e.g. while using lift assists or rolling wheelchairs and beds), as well as the viscoelastic behaviour and durability of the floors during prolonged and repeated loading. Nonetheless, the results of the current study are encouraging, indicating that the novel compliant floors tested can substantially reduce the magnitudes of widely used indices of skull fracture and TBI risk compared to common flooring products.

In order to limit the expected increase in the incidence of fall-related TBI (and other fall-related injuries) in seniors over the coming decades, it is imperative that effective intervention strategies be designed and implemented. Novel compliant flooring systems appear to be a promising approach, capable of providing substantial protective capacity against head injury and other fall-related injuries without introducing impairments to balance and mobility [18,55]. The added benefit of being a passive intervention approach precludes the need for active user compliance and adherence to ensure effectiveness, unlike intervention strategies such as exercise, pharmacological agents, and wearable hip protectors. The results of this study further support the development of clinical trials to test the effectiveness of NCFs in high-risk environments such as hospitals, seniors' centres, and residential-care facilities.

Acknowledgements

This research was funded in part by an operating grant obtained by ACL from the Natural Sciences and Engineering Research Council of Canada (NSERC; grant # 386544-2010). ADW was supported by the Ontario Graduate Scholarship program.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.medengphy.2011.11.012.

Conflict of interest statement

The authors have no conflicts of interest to declare. No persons other than ADW and ACL had substantial input into study design, data analysis, or manuscript development.

References

[1] SMARTRISK. The economic burden of injury in Canada. Toronto, ON: SMARTRISK; 2009.
[2] Pickett W, Ardern C, Brison RJ. A population-based study of potential brain injuries requiring emergency care. CMAJ 2001;165(3):288–92.
[3] Thomas KE, Stevens JA, Sarmiento K, Wald MM. Fall-related traumatic brain injury deaths and hospitalizations among older adults—United States, 2005. J Safe Res 2008;39(3):269–72.
[4] Adekoya N, Thurman DJ, White DD, Webb KW. Surveillance for traumatic brain injury deaths—United States, 1989–1998. MMWR Surveill Summ 2002;51(10):1–14.
[5] Coronado VG, Thomas KE, Sattin RW, Johnson RL. The CDC traumatic brain injury surveillance system: characteristics of persons aged 65 years and older hospitalized with a TBI. J Head Trauma Rehabil 2005;20(3):215–28.
[6] Health Canada. Canada's aging population. Government of Canada; 2002.
[7] Feng Y, Abney TM, Okamoto RJ, Pless RB, Genin GM, Bayly PV. Relative brain displacement and deformation during constrained mild frontal head impact. J R Soc Interface 2010;7:1677–88.
[8] Ferrell RB, Tanev KS. Traumatic brain injury in older adults. Curr Psychiatry Rep 2002;4(5):354–62.
[9] McHenry BG. HIC and the ATB. ATB Users' Group; 2004.
[10] Ivancevic VG. New mechanics of traumatic brain injury. Cogn Neurodyn 2009;3(3):281–93.
[11] Singh A, Lu Y, Chen C, Kallakuri S, Cavanaugh JM. A new model of traumatic axonal injury to determine the effects of strain and displacement rates. Stapp Car Crash J 2006;50:601–23.
[12] Hardy WN, Mason MJ, Foster CD, Shah CS, Kopacz JM, Yang KH, King AI, Bishop J, Bey M, et al. A study of the response of the human cadaver head to impact. Stapp Car Crash J 2007;51:17–80.
[13] King AI. Fundamentals of impact biomechanics: Part I—biomechanics of the head, neck, and thorax. Annu Rev Biomed Eng 2000;2:55–81.
[14] McLean AJ, Anderson RWG. Biomechanics of closed head injury. Head Injury 1997:25–37.
[15] Gennarelli TA. Clinical and experimental head injury. Biomech Impact Trauma 1984:103–15.
[16] Cory CZ, Jones MD, James DS, Leadbeatter S, Nokes LDM. The potential and limitations of utilising head impact injury models to assess the likelihood of significant head injury in infants after a fall. Forensic Sci Int 2001;123:89–106.
[17] Mack MG, Sacks JJ, Thompson D. Testing the impact attenuation of loose-fill playground surfaces. Inj Prev 2000;6(2):141–4.
[18] Laing AC, Robinovitch SN. Low stiffness floors can attenuate fall-related femoral impact forces by up to 50% without substantially impairing balance in older women. Accid Anal Prev 2009;41(3):642–50.
[19] Gardner TN, Simpson AH, Booth C, Sprukkelhorst P, Evans M, Kenwright J, Evans JG. Measurement of impact force, simulation of fall and hip fracture. Med Eng Phys 1998;20(1):57–65.

[20] Maki BE, Holliday PJ, Fernie GR. Aging and postural control. A comparison of spontaneous- and induced-sway balance tests. J Am Geriatr Soc 1990;38(1):1–9.
[21] Simpson AH, Lamb S, Roberts PJ, Gardner TN, Evans JG. Does the type of flooring affect the risk of hip fracture? Age Ageing 2004;33(3):242–6.
[22] NOCSAE. Doc. 001-08m10: standard test method and equipment used in evaluating the performance characteristics of protective headgear/equipment. National Operating Committee on Standards for Athletic Equipment; 2009.
[23] Higgins M, Halstead D, Snyder-Macklet L, Barlow D. Measurement of impact acceleration: mouthpiece accelerometer versus helmet accelerometer. J Athl Train 2007;42(1):5–10.
[24] Gurdjian E, Robert V, Thomas L. Tolerance curves of acceleration and intercranial pressure and protective index in experimental head injury. J Trauma 1966;6(5):600–4.
[25] Prasad P, Mertz HJ. The Position of the United States Delegation to the Iso Working Group 6 on the use of HIC in the automotive environment. In: S.A.E. government industry meeting. 1985. SAE 85124.
[26] Mackay M. The increasing importance of the biomechanics of impact trauma. Sadhana 2007;32(4):397–408.
[27] ASTM. Report F1292-04: standard specification for impact attenuation of surfacing materials within the use zone of playground equipment. American Society for Testing and Materials; 2004.
[28] Funk JR, Cormier JM, Bain CE, Guzman H, Bonugli E, Manoogian SJ. Head and neck loading in everyday and vigorous activities. Ann Biomed Eng 2011;39(3):766–76.
[29] Hrysomallis C, McLaughlin P. Risk of head injury from falls on taekwondo mats. In: 17th international symposium on biomechanics in sports. 1999. p. 373–6.
[30] Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF. Concussion in professional football: reconstruction of game impacts and injuries. Neurosugery 2003;53:799–814.
[31] Dixon JL, Brodie IKR. The New ISO Standards for Ice Hockey Helmets and Face Protectors: Moving Toward International Standards Harmonization and Conformity Assessment. Safety in Ice Hockey: Second Volume, ASTM STP 1212, vol. 2; 1993. p. 192–213.
[32] Hodgson VC, Thomas LM, Khalil TB. The role of impact location in reversible cerebral concussion. In: 27th Stapp car crash conference. 1983. SAE Paper No. 831618.
[33] Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. J Biomech Eng 2004;126(2):226–36.
[34] Zhang L, Yang KH, King AI. Comparison of brain responses between frontal and lateral impacts by finite element modeling. J Neurotrauma 2001;18(1):21–30.
[35] Kleiven S. Influence of impact direction on the human head in prediction of subdural hematoma. J Neurotrauma 2003;20(4):365–79.
[36] Guskiewicz KM, Mihalik JP, Shankar V, Marshall SW, Crowell DH, Oliaro SM, Ciocca MF, Hooker DN. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. Neurosugery 2007;61(6):1244–53.
[37] Nahum AM. The biomechanics of facial bone fracture. Laryngoscope 1975;85(1):140–56.
[38] Cormier J, Manoogian S, Bisplinghoff J, Rowson S, Santago A, McNally C, Duma S, Bolte J. The tolerance of the frontal bone to blunt impact. J Biomech Eng 2011;133(2):021004.
[39] Minns J, Dodd C, Gardner R, Bamford J, Nabhani F. Assessing the safety and effectiveness of hip protectors. Nurs Stand (Roy Coll Nurs, Great Britain) 2004;18(39):33–8.
[40] Nabhani F, Bamford J. Mechanical testing of hip protectors. J Mater Process Technol 2002;124(3):311–8.
[41] Nabhani F, Bamford J. Impact properties of floor coverings and their role during simulated hip fractures. J Mater Process Technol 2004;153–154:139–44.
[42] Laing AC, Robinovitch SN. Characterizing the effective stiffness of the pelvis during sideways falls on the hip. J Biomech 2010;43(10):1898–904.
[43] Robinovitch SN, Feldman F, Wan D, Aziz O, Sarraf T. Video recording of real-life falls in long term care provides new insight on the cause and circumstances of falls in older adults. In: Proceedings of the 19th annual meeting of the international society for posture and gait. 2009.
[44] Robinovitch SN, Chiu J. Surface stiffness affects impact force during a fall on the outstretched hand. J Orthop Res 1998;16(3):309–13.
[45] Sran MM, Robinovitch SN. Preventing fall-related vertebral fractures: effect of floor stiffness on peak impact forces during backward falls. Spine (Phila, PA, 1976) 2008;33(17):1856–62.
[46] Redfern MS, Moore PL, Yarsky CM. The influence of flooring on standing balance among older persons. Hum Factors 1997;39(3):445–55.
[47] Lord SR, Clark RD, Webster IW. Postural stability and associated physiological factors in a population of aged persons. J Gerontol 1991;46(3):M69–76.
[48] Lord SR, Menz HB. Visual contributions to postural stability in older adults. Gerontology 2000;46(6):306–10.
[49] Gill J, Allum JH, Carpenter MG, Held-Ziolkowska M, Adkin AL, Honegger F, Pierchala K. Trunk sway measures of postural stability during clinical balance tests: effects of age. J Gerontol A Biol Sci Med Sci 2001;56(7):M438–47.
[50] MacLellan MJ, Patla AE. Stepping over an obstacle on a compliant travel surface reveals adaptive and maladaptive changes in locomotion patterns. Exp Brain Res 2006;173(3):531–8.
[51] Marigold DS, Patla AE. Adapting locomotion to different surface compliances: neuromuscular responses and changes in movement dynamics. J Neurophysiol 2005;94(3):1733–50.
[52] Dickinson JI, Shroyer JL, Elias JW. The influence of commercial-grade carpet on postural sway and balance strategy among older adults. Gerontologist 2002;42(4):552–9.
[53] Dickinson JI, Shroyer JL, Elias JW, Hutton JT, Gentry GM. The effect of selected residential carpet and pad on the balance of healthy older adults. Environ Behav 2001;33(2):279–95.
[54] Podsiadlo D, Richardson S. The timed "Up & Go": a test of basic functional mobility for frail elderly persons. J Am Geriatr Soc 1991;39(2):142–8.
[55] Lundin-Olsson L, Nyberg L, Gustafson Y. Attention, frailty, and falls: the effect of a manual task on basic mobility. J Am Geriatr Soc 1998;46(6): 758–61.
[56] Chiu AY, Au-Yeung SS, Lo SK. A comparison of four functional tests in discriminating fallers from non-fallers in older people. Disabil Rehabil 2003;25(1):45–50.
[57] Wright AD, Laing AC. The influence of novel compliant floors on balance control in elderly women—a biomechanical study. Accid Anal Prev 2011;43(4): 1480–7.
[58] Klenk J, Becker C, Lieken F, Nicolai S, Maetzler W, Alt W, Zijlstra W, Hausdorff JM, van Lummel RC, et al. Comparison of acceleration signals of simulated and real-world backward falls. Med Eng Phys 2011;33:368–73.
[59] Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. Exerc Sport Sci Rev 2011;39(1):4–11.
[60] Ommaya AK, Gennarelli TA. Cerebral concussion and traumatic unconsciousness. Correlation of experimental and clinical observations of blunt head injuries. Brain 1974;97(4):633–54.
[61] Marjoux D, Baumgartner D, Deck C, Willinger R. Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria. Accid Anal Prev 2008;40(3): 1135–48.
[62] Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosugery 2008;62(4):789–98.
[63] Duma SM, Rowson S. Past, present, and future of head injury research. Exerc Sport Sci Rev 2011;39(1):2–3.
[64] Rowson S, Beckwith JG, Chu JJ, Leonard DS, Greenwald RM, Duma SM. A six degree of freedom head acceleration measurement device for use in football. J Appl Biomech 2011;27(1):8–14.
[65] Rowson S, Brolinson G, Goforth M, Dietter D, Duma S. Linear and angular head acceleration measurements in collegiate football. J Biomech Eng 2009;131(6):061016.
[66] Shain KS, Madigan ML, Rowson S, Bisplinghoff J, Duma SM. Analysis of the ability of catcher's masks to attenuate head accelerations on impact with a baseball. Clin J Sport Med 2010;20(6):422–7.
[67] Zacker C, Shea D. An economic evaluation of energy-absorbing flooring to prevent hip fractures. Int J Technol Assess Health Care 1998;14(3): 446–57.