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Impact attenuation provided by older adult protective headwear products during simulated fall-related head impacts

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Abstract

Introduction: While protective headwear products (PHP) are designed to protect older adults from fall-related head injuries, there are limited data on their protective capacity. This study's goal was to assess the impact attenuation provided by commercially available PHP during simulated head impacts.

Methods: A drop tower and Hybrid III headform measured the decrease in peak linear acceleration (g_{atten}) provided by 12 PHP for front- and back-of-head impacts at low (clinically relevant: 3.5 m/s) and high (5.7 m/s) impact velocities.

Results: The range of g_{atten} across PHP was larger at the low velocity (56% and 41% for back and frontal impacts, respectively) vs. high velocity condition (27% and 38% for back and frontal impacts, respectively). A significant interaction between impact location and velocity was observed ($p < .05$), with significantly greater g_{atten} for back-of-head compared to front-of-head impacts at the low impact velocity (19% mean difference). While not significant, there was a modest positive association between g_{atten} and product padding thickness for back-of-head impacts ($p = .095$; $r = 0.349$).

Conclusion: This study demonstrates the wide range in impact attenuation across commercially available PHP, and suggests that existing products provide greater impact attenuation during back-of-head impacts. These data may inform evidence-based decisions for clinicians and consumers and help drive industry innovation.

Keywords

Older adults, head injury, protective devices, impact biomechanics, falls, gmax

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Introduction

Falls and fall-related injuries are a serious public health concern for older adults across the globe.[1–3] Fall-related, and more specifically traumatic brain injuries (TBI), comprise a substantial portion of these injuries[4–7] which are twice as common in older adults, cause more than half of the fall-related deaths in older adults, and the rate of these injuries is increasing.[8,9] Even within older adults, the risk of fall-related TBI increases with age; those over the age of 85 are 6 times more likely to be hospitalized for a fall-related TBI than those aged 65–74.[10] When compounded with the ageing demographics being experienced in many countries,[1,2,5] there is a clear need to develop and evaluate interventions for preventing fall-related TBI in older adults.[6]

Although the physiological and neurological pathways of TBI are still being investigated, impacts to the head are the most

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Figure 1. Photographs of all protective headwear products (PHP) tested; PHP are presented with their identifying letter.

common cause of TBI.[11] Impact loading scenarios can cause both linear and rotational acceleration of the brain, leading to fluctuations of pressure within the skull and straining of the brain tissue, resulting in damage to both blood vessels and axons.[12–18] The type of damage as well as injury severity are affected by the properties of the surface on which the head impacts[19–21] which can be influenced by engineered prevention products (e.g. helmets and playground surface materials[22]).

Intervention approaches have been developed to help prevent fall-related head injuries in older adults. For example, novel compliant flooring and bedside mats products have demonstrated substantial impact attenuation and energy absorption during simulated head impacts.[23,24] Other proposed fall-related injury interventions include indirect methods, such as exercise and environmental modifications,[25] as well as as wearable airbag systems that deploy during a fall,[26] and commercially available protective headwear products (PHP-Figure 1). Helmets used in sporting scenarios, such as in cycling,[27,28] rugby,[29] skiing and snowboarding,[30] have been shown to effectively

Table 1. List of the 12 protective headwear products (PHP) that were tested, including information about the company, model, price and padding thickness measured at the three impact sites.

Product code	Company	Model	Price (CAD)	Padding thickness (cm)		
				Back	Front	Side
A	Danmar	9829 Full Coverage Helmet	$167.75	1.59	1.59	1.59
B	Ice Halo	Ice Halo HD	$95.00	1.59	1.27	1.90
C	Danmar	9820 Soft Shell Helmet	$123.00	1.59	1.59	1.59
D	Head Buddy	Shell	$130.00	5.08	3.18	3.18
E	Ice Halo	Trucker Hat	$80.00	0.95	0.95	0.95
F	Ice Halo	Standard Toque	$80.00	0.95	0.95	0.95
G	Danmar	9708 Soft Comfy Cap	$96.00	2.54	2.54	1.59
H	Ice Halo	Ice Halo Fleece	$49.95	2.54	2.54	3.18
I	AliMed	Economy Helmet	$24.75	0.95	1.27	1.27
J	Ice Halo	Baseball Cap	$60.00	0.64	-	0.64
K	Crasche	Summer Knit Hat	$28.95	1.27	1.27	1.27
L	Hip Saver Canada	Headsaver	$95.00	1.59	1.27	1.27

prevent head injuries by dissipating, or shunting, impact energy. This suggests there may be value to this approach for reducing the risk of head injuries in older adults during fall-related impacts.

Commercially available PHP products (Figure 1) aim to reduce the severity of head impacts by absorbing and shunting energy from the impact site. Many of these products are specifically designed for older adults and are marketed as injury preventing devices during fall-related head impacts. However, unlike many application-specific helmets (e.g. bicycle, football), these products are not associated with formal test standards, which may (in part) contribute to the variety of design philosophies and materials employed. Currently available products differ greatly in design, both in material selection and product shape. Consequently, products may differ in protective capability, as the materials used may have rate dependencies, and differences in product geometries may result in different levels of impact attenuation provided to different parts of the head. Unfortunately, there is a paucity of evidence, including standardized biomechanical effectiveness assessments, to assist clinicians, policy-makers and consumers with decisions regarding older adult PHPs. Such data could also have industry relevance towards accelerating innovation cycles and developing newer generation products with improved protective capacity. Mechanical impact simulators are commonly employed to assess the protective capacity of engineered injury prevention devices. These testing systems are designed to induce (and measure) abrupt headform accelerations and associated impact loads for different types of head impacts.[24,28,31,32] Standardized test methods exist for characterizing the impact attenuation properties for a variety of applications.[33,34] Similar methods could be used to characterize the biomechanical effectiveness of older adult PHPs.

The overarching goal of this study was to characterize the range of biomechanical effectiveness, in terms of impact attenuation, provided by commercially available protective headwear products for older adults. The primary objectives were to test the hypotheses that attenuation in impact severity metrics would: 1) vary substantively across a sample of 12 commercially available older adult protective headwear products, 2) be influenced by headform impact location and 3) be influenced by impact velocity. Towards an initial investigation of the influence of design factors, our fourth hypothesis was that a positive association would exist between impact attenuation and the thickness of padding materials incorporated within the PHPs (specific to front and back aspects of the products). A secondary objective was to develop and implement a preliminary product ranking metric to help support evidence-based decisions from clinicians, industry and consumers.

Methods

Protective headwear products

Based on discussions with healthcare practitioners, industry representatives and internet searches, 12 commercially available protective headwear products were investigated in this study (Figure 1, Table 1). The products fell within three general design categories. The first was termed 'full coverage', and utilized materials to explicitly cover the front, back, sides and top of the head. The second category was 'headband' style products, which covered the front, back and sides of the head with a ring of padding; these could be stand-alone headbands, or imbedded within a different

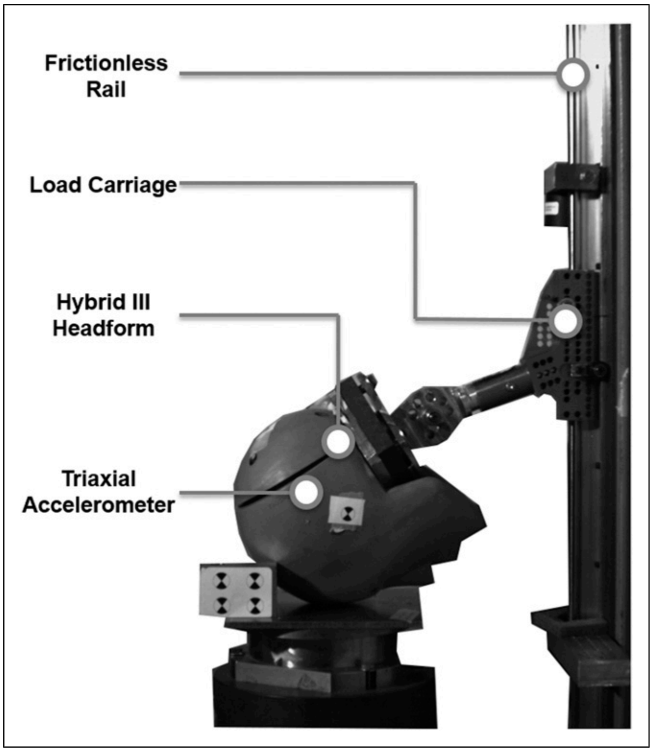

Figure 2. Vertical rail drop tower system with Hybrid III headform; the headform is instrumented with a triaxial accelerometer.

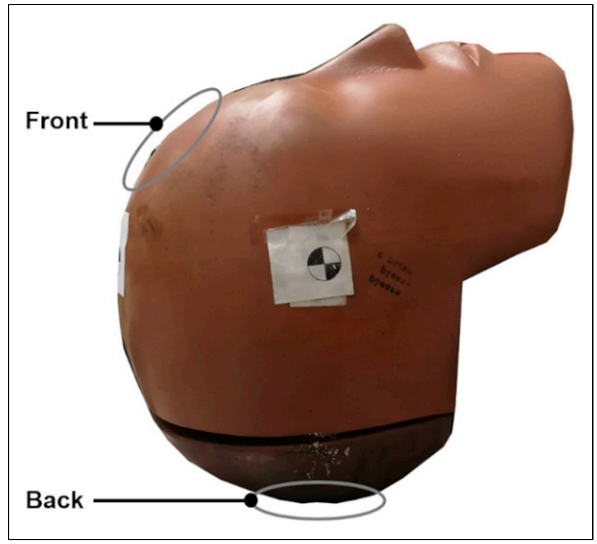

Figure 3. Hybrid III headform impact locations (front and back) used in this study.

style of headwear (such as a knitted hat). The third category involved the addition of padding materials into preexisting headwear products (e.g. a baseball cap). Prior to testing, precision calipers were used to measure each product's material thickness at points corresponding to the primary impact location ($thick_{front}$, $thick_{back}$; Table 1).

Test system

As described previously,[24] a mechanical impact simulator (drop tower) was used to impact a medium-sized Hybrid III (HIII) headform onto a uniaxial load cell mounted beneath an impact anvil (Model 925M113; Kistler Instrument Corporation, Amherst, NY, USA) (Figure 2). HIII headforms have previously been used in experiments comparing unhelmeted and helmeted head impacts,[28,35] and have the benefit of being more resilient against damage during baseline impacts compared to other headforms. Impact velocity was measured with an infrared light gate velocimeter (Model VS300, GHI Systems, Aurora, ON, Canada) positioned immediately above the impact surface. Based on evidence that head impact in older adults is most likely during forward-directed falls,[36] and that head impact velocity averages 3.52 m/s during forward falls in older adults,[37] we used an impact velocity of 3.5 m/s for our low velocity condition trials. Additionally, we employed a higher impact velocity of 5.7 m/s, mimicking the high energy condition used in bicycle helmet standards testing,[38] to simulate a higher severity impact and test the more extreme capabilities of the products. Separate headform orientations were used to create two common impact locations (front of the head, back of the head; Figure 3). A triaxial accelerometer (Model 2707A, Endevco Corporation, San Juan Capistrano, CA, USA) mounted at the headform's centre of mass recorded internal accelerations at impact. Impact force and acceleration data were sampled at 20,000 Hz and saved on a personal computer using a multichannel analogue-to-digital converter card and custom data acquisition software (NIAD 3.0, NI, Austin, Texas, USA).

Experimental protocol

Three repeated baseline impact trials (i.e. no product on the headform) were performed in the two impact location conditions (front, back) for both impact velocities (3.5 m/s, 5.7 m/s), resulting in 12 total baseline trials. Subsequently, trials with all products were conducted; each product was positioned on the headform according to the manufacturer's specifications. Trials were blocked by impact velocity and impact location; two repeated trials were performed for each product before advancing to the next impact location, and the process was repeated at the second impact velocity. This blocked approach (i.e. not randomizing impact velocity and impact location for each product) was chosen to limit between-product variability associated with changes in test system drop height and headform orientation. Two repeated trials were selected for each condition on the basis on high

Table 2. Mean (SD) and Coefficient of Variation (CV) in g_{max} for the three baseline trials in the four combinations of impact velocity and headform impact location.

Headform impact location	Back of head		Front of head	
Impact velocity	Low	High	Low	High
g_{max} mean (SD)	503.2 (10.9)	1114.5 (6.8)	342.2 (8.8)	1348.3 (20.3)
CV (%)	2.18	0.61	2.57	1.51

repeatability of the baseline impact trials (Table 2), and to limit potential changes in the protective properties of the products across impact velocity conditions. All tests were performed using the same two researchers performing the same tasks throughout the experiment.

Data analysis and statistics

Similar to previous work,[24] custom MATLAB routines were used to perform data analysis (MATLAB 2013b, MathWorks, Natick, MA, USA here). ASTM Standard F1292-17a was followed for accelerometer data processing for testing impact attenuation of surfacing materials during simulated head impacts.[33] A fourth-order, dual-pass, low-pass digital Butterworth filter (2077.5 Hz cutoff)[39] was applied before calculating the resultant acceleration from the three orthogonal axes. Peak acceleration (g_{max}), peak impact force (F_{max}) and peak Head Injury Criterion score (HIC_{max}) were extracted for each trial, and the average of both impact trials per condition was computed and used for analyses. However, Pearson product moment correlation analyses revealed that all three outcome variables were highly associated for both impact orientations and velocities, for the unprotected trials (r = 0.699–0.976). Accordingly, the remaining text focuses on $g_{max.}$ Information on F_{max} and HIC_{max}, as well as their associated impact attenuation metrics (F_{atten} and HIC_{atten}) can be found in the Supplementary Material. Both g_{max} and g_{atten} data at the low and high impact velocities were normally distributed, as verified via Shapiro–Wilk's Normality tests**.**

The impact attenuation metric g_{atten} was determined for each trial by calculating the percent reduction in the g_{max} compared to the average baseline (unprotected) trials for the associated impact location and velocity condition. To address our first hypothesis, mean, standard deviation and range of g_{max} and g_{atten} were calculated across all products for each impact velocity and location. Towards the second and third hypothesis, a two-factor ANOVA was conducted to investigate the influence of impact orientation and impact velocity on g_{atten}. Finally, to address our fourth hypothesis, one-tailed Pearson product-moment correlations were performed between product thickness ($thick_{front}$, $thick_{back}$) and g_{atten} in each associated head impact orientation. In all cases, $\alpha < 0.05$ was used as the threshold for significance.

Product ranking

g_{max} results were distilled into a single summary ranking variable ($R_{overall}$). For each of the four impact velocity and impact location combinations, the 12 products were ranked from most (assigned a score of 1) to least (a score of 12) biomechanically effective. If a product's design precluded it from being tested for a certain impact location, it was allocated a rank of 12 (e.g. Product J was not tested in the 'front' condition as the product's brim caused it to dislodge from the headform during impact). For each product, the rankings were summed across the four impact velocities and location combinations to provide a single value ($R_{overall}$) which could range from 4 to 48, with lower values representing more effective impact attenuation. The data in all figures and tables are presented in increasing order of $R_{overall}$, with product A demonstrating the lowest $R_{overall}$, and product L the highest.

Results

The test system produced repeatable results (Table 2). The g_{max} coefficients of variability for the three baseline unprotected trials in each velocity/orientation condition ranged from 0.61% to 2.6%.

While all products reduced g_{max} compared to the baseline unprotected conditions, there was substantial variance in impact severity across products. The range in g_{max} across products was greatest in the back-of-head high velocity condition (436.0–858.7 g), and smallest in the front-of-head low velocity condition (142.4–335.46 g) (Figures 4(a) and 5(a)). Conversely, the range in g_{atten} across products was greatest in the front-of-head low velocity condition (2.0–58.4% attenuation), and smallest in the front-of-head high velocity condition (19.0–46.4% attenuation) (Figures 4(b) and 5(b)).

ANOVA indicated a significant main effect of impact location on g_{atten} characterized by a 19.5% decrease in impact attenuation for front of the head (mean (SD) = 31.11 (15.14)%) compared back of the head impacts (mean (SD) = 50.61 (14.75)%) (F = 22.03, p < .001). No main effect of velocity was observed (F = 0.54, p = .468). As an interaction between impact velocity and location was observed (F = 7.03, p = .0113), comparisons were performed at each

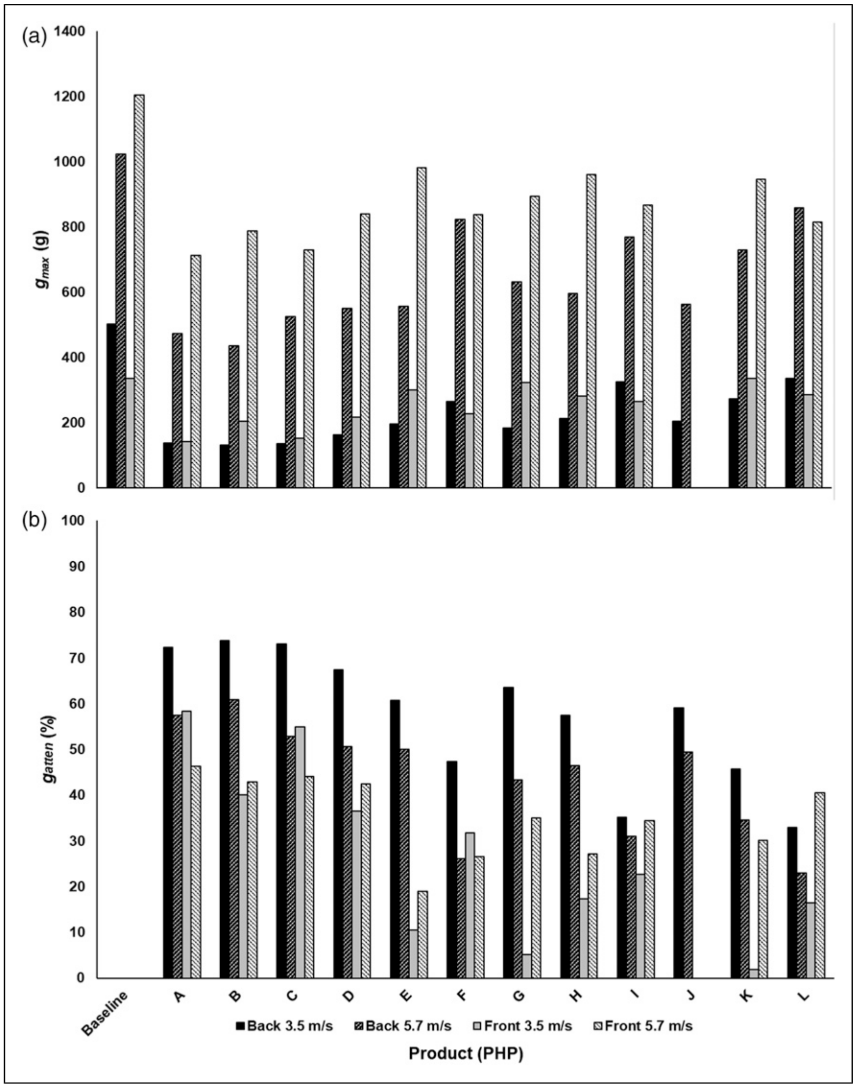

Figure 4. Comparison of all products in each head configuration for variables A) *gmax* and B) *gatten*.

impact velocity. At the low impact velocity, g_{atten} was significantly higher for back of the head impacts compared to frontal impacts (57.4 (14.2)% vs. 26.9 (19.1)%, respectively; F = 19.12, p < .001). However, there was no effect of impact location on g_{atten} at the higher impact velocity (F = 3.54, p = .074).

Limited associations were observed between padding thickness ($thick_{front}$, $thick_{back}$) and g_{atten}. More specifically, while a modest positive correlation was observed for back-of-head impacts (r = 0.349; p = .095), there was little evidence of a relationship for front-of-head impacts (r = 0.100; p = .658). Subset analyses indicated no significant associations between padding thickness and g_{atten} at the high impact velocity condition (r = 0.331, p = .320 vs r = 0.403, p = .194 for front and back-of-head impacts, respectively) or low impact velocity condition (r = 0.010, p = .977 vs r = 0.390, p = .210, front and back).

Based on overall ranking ($R_{overall}$; Table 3), the three most effective products were A, B and C ($R_{overall}$ values of 7, 8 and 9, respectively); g_{atten} averaged 56.4% (SD = 12.0%) for these products. In contrast, the three lowest ranked products were J, K and L ($R_{overall}$ = 37, 37 and 38, respectively), which were associated with mean (SD) g_{atten} of 27.8 (20.0)%.

Discussion

The overarching goal of this study was to characterize the impact attenuation of a sample of 12 commercially available protective headwear products for older adults. In support of our first hypothesis, while all products reduced metrics of impact severity, impact attenuation varied substantively across products and between conditions. For example, the product with the best overall rankings reduced g_{max} between 58.4% to 72.3% at the low impact velocity, and 46.4% to 57.4% at the high impact velocity. Comparatively, the product with the lowest overall ranking reduced g_{max} by 16.4% to 33.0% and by 23.0% to 40.6% at the low and high impact velocities, respectively. There was partial support of our second and third hypotheses. At the lower impact velocity, the products (on average) provided 30.5% greater impact attenuation for impacts to the back versus front of the head. However, these differences in impact location were not observed at the higher impact velocity. Similarly, while impact severity was higher for the high impact velocity condition (Figure 4(a)), the relative impact attenuation captured by g_{atten} was not affected by impact velocity (Figure 4(b)). Finally, there was little support for our fourth hypothesis as there was only a modest positive correlation between g_{atten} and product thickness for the back of head impacts (r = 0.35), and even less association between these variables for front of head impacts ($r = 0.10$). Overall, the findings from this study could assist in driving innovation in older adult protective headwear design, in addition to assisting clinical and consumer stakeholders with purchase decisions.

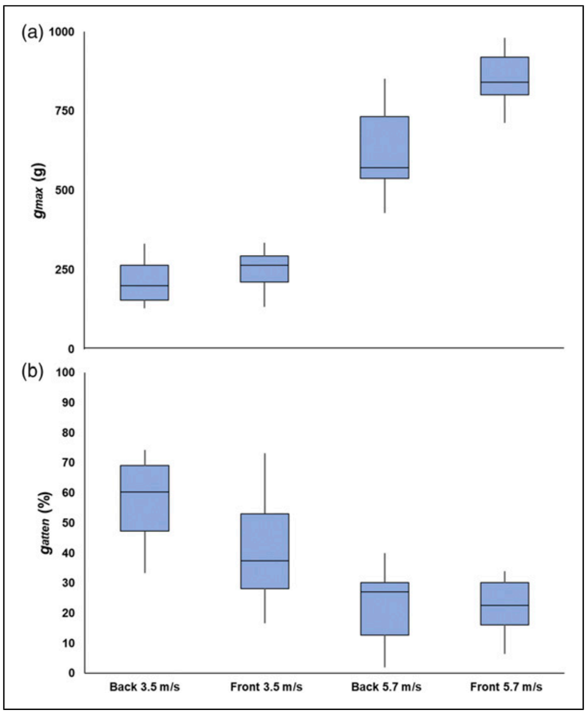

Figure 5. Box and whisker plots showing the median (horizontal bar), middle quartiles (blue box) and range (vertical lines) in both *gmax* and *gmax* across the four conditions (front- and back-of-the-head impacts at the low and high impact velocities).

There is value in comparing the impact severities we observed with injury risk tolerances reported in the literature. For example, Zhang et al. (2004) reported that 66, 82 and 106 g of translational acceleration were associated with 25, 50 and 80 percentile probabilities (respectively) of suffering mild traumatic brain injury (mTBI).[40] In our study, low velocity impacts at both locations resulted in baseline g_{max} values (mean (SD) = 411.9 (87.8) g) in excess of Zhang et al.'s 80% mTBI probability threshold. While our test system was likely stiffer than actual human head and neck segments, our low velocity g_{max} values were within 20% of Caccese et al.[41] who incorporated a Hybrid III neckform into their test system and performed head impact trials at 3.58 m/s (g_{max} = 422 g and 410 g for their back and front of

Table 3. Product rankings based on g_{max} in the three impact orientations; each product rated within impact orientation, and ultimately ranked overall ($R_{overall}$) based on the sum of the three orientation-specific ranks.

	Low impact velocity				High impact velocity				
	Back impact orientation		Front impact orientation		Back impact orientation		Front impact orientation		
Product code	g_{max}	Rank order	g_{max}	Rank order	g_{max}	Rank order	g_{max}	Rank order	$R_{overall}$
A	139.2	3	142.4	1	474.3	2	713.1	1	**7**
B	131.6	1	204.9	3	436.0	1	786.9	3	**8**
C	135.7	2	154.0	2	525.1	3	728.8	2	**9**
D	163.8	4	217.4	4	550.7	4	840.7	6	**18**
E	265.2	9	228.4	5	823.0	11	837.3	5	**30**
F	197.6	6	300.7	9	556.8	5	980.4	11	**31**
G	183.5	5	324.4	10	631.1	8	894.0	8	**31**
H	213.7	8	283.0	7	596.2	7	961.4	10	**32**
I	326.3	11	264.7	6	769.0	10	867.6	7	**34**
J	205.8	7	N/A	12	563.4	6	N/A	12	**37**
K	272.9	10	335.5	11	858.7	12	814.8	4	**37**
L	337.2	12	285.9	8	728.5	9	946.3	9	**38**

head conditions, respectively). The current findings do generally support that the older adult falls on which we modelled our 3.5 m/s impact velocity[37,40] were associated with substantial impact loads, and correspond to the high incidence of fall-related concussions and mTBI observed clinically in older adults.[8,10,42] Interestingly, due to the severe nature of these impacts, even the most effective PHP we tested (product A) was unable to reduce the impact severity below the 80%ile mTBI threshold for any impact location (g_{max} = 139.2 g). At lower impact velocities, or with a more biofidelic neck segment, the impact magnitude and injury threshold distributions may more closely align; it is possible that the PHPs we tested might more effectively reduce injury risk (or severity) under such conditions. As evidenced by the high g_{max} values in our baseline (unprotected) conditions, next generation products may wish to target even greater impact attenuation given the severe nature of fall-related head impacts in older adults.

Impact attenuation is likely influenced by factors including the material and geometric properties of energy absorbing materials, and where they are incorporated within the product. For example, the padding for the product with the best $R_{overall}$ score (product A) had a thickness of 1.59 cm in its front and back aspects. Interestingly, other products (e.g. H and K) incorporated similar padding thickness but were less effective in attenuating g_{max}. The density of foam materials likely played a role in these findings. For example, product B (Ice Halo HD) incorporated a higher density vinyl nitrile foam compared to the Ice Halo Fleece (product H). Despite product B being thinner, the foam's higher material density (and stiffness) likely resulted in greater energy absorption before bottoming out, which translated into more effective energy management and lower g_{max} values**.** Accordingly, the properties of the padding materials (e.g. density, stiffness, hysteresis), in addition to their geometry, influence the impact attenuative properties of these headwear products. The properties of common padding materials (e.g. expanded polystyrene, expanded polypropylene), vinyl nitrile foams) and novel rate sensitive materials (e.g. carbon nanotube foams, aluminum foams, encapsulated low shear oils) can be tuned to match the characteristics of anticipated impact events. Explicit exploration of material effects is an area for future research.

While two impact locations were investigated in this study, it was clear that some products were not designed to provide equal degrees of protection to both aspects of the head. A clear example was product J, which was marketed as a classic "baseball cap" with foam embedded within the posterior and lateral lining, but no energy absorptive materials incorporated into the front of the product. This resulted in moderate rankings (7 and 6 out of 12) for low and high velocity back of the head impacts, but the lowest ranking (12 out of 12) for the front impact location. However, orientation-specific design approaches may have merit based on fall mechanics and impact sites. For example, while the aspect of the head struck was not reported, Schonnop et al. found the odds ratio for head impact in long term care (LTC) residents was at least 2.7-fold larger for initial fall directions that were forwards compared to backwards, sideways or straight down.[36] Similarly, of the LTC falls that Choi et al. examined, head impacts were observed in 100% of forward falls vs. 38% for backwards falls.[37] In addition to matching most likely impact locations, orientation-specific design approaches may result in more streamlined products which could enhance user acceptance and compliance. Future studies may wish to consider differential weightings across impact locations once more detailed impact location patterns for older adults are reported in the literature.

There are several important clinical messages that arise from this study. First, the PHPs we tested all reduced impact severity to some extent, and so if worn properly, have the potential to attenuate linear impact loads during fall-related head impacts. Given the link between angular accelerations and TBI,[32,43] in future studies there is value in complementing the current results with consideration of rotational effects. Second, while all products conferred benefits, there was a wide range in biomechanical effectiveness. Third, the impact attenuation provided differed across head orientation, with the products conferring (on average) 19.5% greater impact attenuation for impact to the back of the head, compared to impacts to the front of the head. Accordingly, this study provides initial data (Table 3) that could assist clinicians/consumers in identifying products that may confer the greatest impact attenuation for their given needs. Finally, initial acceptance of (and adherence to) engineered products is paramount for clinically effective injury risk reduction strategies.[44,45] Accordingly, stakeholders may need to balance the biomechanical effectiveness conferred by a product with factors that may influence end use (e.g. size, aesthetics, comfort), with a potential goal of tailoring products to end user needs and desires.

There were several limitations associated with this study. Firstly, the stability in biomechanical effectiveness of the protective headwear products was not assessed over repeated impacts. As decrements in protective capacity over multiple trials have been reported for products including bicycle helmets and hip protectors,[32,46] future studies should characterize these potential effects on commercially available older adult PHPs to account for potential impact-induced and long-term changes due to product ageing. Secondly, our test protocol utilized two impact velocities (3.5 m/s and 5.7 m/s) towards simulating both typical and more extreme impacts (the lower velocity matched the characteristics observed in real-life video capture of falls in older adults,[37] with the higher velocity being taken directly from a cycling helmet testing standard[38]). However, impact velocity can vary greatly depending on the direction of the fall and the configuration of the body during the fall.[37] For example, a study by Hajiagmamemar and colleagues used a Hybrid III ATD to quantify head impact characteristics of standing height falls with different fall directions and body configurations reported head impact velocities ranging from 2.9 m/s to 6.75 m/s.[47] Future studies may consider employing a range of impact velocities towards characterizing potential velocity-dependencies in the protective capacity of the products. Third, while the test system employed was similar to those reported in the literature,[28,35] the lack of a surrogate neck increased system stiffness. While this likely contributed to the relatively high g_{max} values we observed, it is unlikely the relative ranking of PHPs was affected. In addition, only linear accelerations were considered; based on the link between TBI and rotational accelerations,[32,43] in future work there is value in measuring the effects of PHPs on these important outcomes. Finally, as discussed above our ranking system allocated equal weighting to the impact attenuation observed in each headform impact location. Future studies may wish to consider alternative approaches once more information becomes available on the relative incidence (and clinical outcomes) of head impact locations during falls in older adults.

Conclusion

In summary, this study highlights the wide range of biomechanical effectiveness across 12 commercially available older adult protective headwear products. The results indicate that the impact reduction conferred by these products is a complex issue and differs across impact sites. The overall product ranking of biomechanical effectiveness, in addition to the impact site-specific data, could help inform stakeholders in their choice of Protective Headwear Products tailored to the needs of end-users. This study is helping to inform the development of biomechanical test standards and minimum performance requirements for older adult protective headwear products.

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Supplemental Material

Supplemental material for this article is available online.

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