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Free-fall drop test with interchangeable surfaces to recreate concussive ice hockey head impacts

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Abstract

Ice hockey has one of the highest concussion rates in sport. During collisions with other players, helmets offer limited protection. Various test protocols exist often requiring various types of laboratory equipment. A simplified test protocol was developed to facilitate testing by more researchers, and modifications to certification standards. Measured kinematics (acceleration vs. time trace shape, peak accelerations, and impact duration) of a Hybrid III headform dropped onto different surfaces were compared to published laboratory representations of concussive impacts. An exemplary comparison of five different helmets, ranging from low (US\$50) to high cost (US\$300), covering a range of helmet and liner designs, was also undertaken. Different impact conditions were created by changing the impact surface (Modular Elastomer Programmer pad, or 24 to 96 mm of EVAZOTE-50 foam with a Young's modulus of ~ 1 MPa), surface orientation (0 or 45°), impact site, and helmet make/model. With increasing impact surface compliance, peak accelerations decreased and impact duration increased. Impacts onto a 45° anvil covered with 48 mm of foam produced a similar response to reference concussive collisions in ice hockey. Specifically, these impacts gave similar acceleration vs. time trace shapes, while normalized pairwise differences between reference and measured peak acceleration and impact duration, were less than 10% (difference/maximum value), and mean (± SD) of accelerations and duration fell within the interquartile range of the reference data. These results suggest that by modifying the impact surface, a free-fall drop test can produce a kinematic response in a helmeted headform similar to the method currently used to replicate ice hockey collisions. A wider range of impact scenarios, i.e., fall onto different surfaces, can also be replicated. This test protocol for ice hockey helmets could facilitate simplified testing in certification standards and research.

Keywords Concussion · Brain injury · Impact test · Helmet · Protective equipment

1 Introduction

Concussions are a public health concern [1, 2]. A concussion is a traumatic brain injury induced by excessive head accelerations, typically causing temporary impairment of neurological function [3]. A history of concussions is associated with an increased risk of long-term neurological and psychological health problems [1, 4–8]. In professional ice hockey, concussions place financial burdens on teams (and their insurers) [9, 10], and can cause players to take extended breaks, or end their careers early [11, 12].

Ice hockey has one of the highest concussion rates in sport [13]. Reported concussion rates are 11.8 concussions/100 games, making up 14% of injuries at a professional level [11, 14–16]. These recorded concussion rates may be an underestimation, as symptoms are often under-reported by players [17, 18]. Causes for concussions in

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ice hockey are collisions with an opponent (88%), falls onto the ice or side-boards (7%), and collisions with a teammate (5%) [19, 20]. Falls onto the ice or side-boards produce higher magnitude, shorter duration head accelerations than collisions between players [21–23]. During collisions between players, a player's head impacts several layers of textiles, protective equipment, and the other player's body, which are typically less stiff than helmet liners [21, 24, 25]. Indeed, laboratory tests suggest that a helmet liner will only compress by a small amount during impacts with another player [21, 24]. Consequently, current helmets offer limited protection during common collisions with other players (i.e., compliant surfaces). Even at lower accelerations (below 50 g [26]), increased impact durations have been associated with increased concussion risks, as predicted by finite element (FE) brain trauma models [27, 28] and observed in American Football [29, 30].

Certification test standards (DIN EN ISO 10256 [31], ASTM F1045-16 [32], CSA Z262.1 [33]) ensure helmets offer a minimum level of protection. These standards include linear, guided drop tests onto a perpendicularly aligned, stiff polymer surface (Modular Elastomer Programmer (MEP) pad, Hardness: 60 ± 2 Shore A), with a pass/fail criterion of 275 g, which was reduced from 300 g in 2016, indicative of a trend toward stricter regulations and more effective helmets. Standardization has helped to nearly eliminate skull fractures and catastrophic head injuries in ice hockey [31–34]. Without angular acceleration measures, as introduced for American Football helmet standards [35, 36], or compliant impact surfaces, standards do not assess helmet performance in some potentially concussive impacts [37].

Various tests, including drop tests [21, 26], pneumatic rams [21, 26, 38], pendulum swings [39], and projectile shooters [40], have been used to assess ice hockey helmet performance. A recent protocol [26] requires extensive laboratory equipment that is not available in many research centers nor test houses. Simpler test protocols requiring less equipment could advance ice-hockey helmet research and development. Various injury risk criteria exist [41–46], focusing on linear, angular, and impact duration metrics. Established criteria, like the Head Injury Criterion (HIC), inform pass/fail thresholds in standard tests [41], while the Rotational Injury Criterion (RIC) was developed to estimate injury risk caused by angular acceleration [42]. Brain strains predicted by FE brain trauma models have been shown to have the highest correlation with field data [47, 48].

Knowledge around the cause of concussion and reliable field data is still limited, and it is currently unclear which test methods and injury criteria are preferable [49]. Our understanding is continuously evolving due to in-field data collections using instrumented helmets [50, 51] and mouthguards [52–54], video footage analysis [19, 20], and simulation techniques [55–57]. Development of simple, adaptable tests that replicate a range of head impacts could improve helmets and certification standards.

We aimed to assess whether a free-fall drop test onto surfaces with varying compliance and orientation can recreate common concussive head impacts in ice hockey. Such a test could be replicated with certification standard test equipment that is available to most researchers interested in head impacts. We hope to facilitate representative helmet testing by more researchers, while increasing the feasibility of modifications to certification standards. An exemplary comparison of helmet performance was also undertaken.

2 Methods

An anthropometric headform was dropped without rotation onto various surfaces (Online Resource 1 & Online Resource 2). The headform was a Hybrid III (50th percentile adult male, mass: 4.54 kg; JASTI Co. Ltd), equipped with a sensor system (Slice Nano, Diversified Technical Systems, Inc. (DTS)) with 3 linear accelerometers (ACCEL SLICE, DTS) and 3 angular rate sensors (ARS3 PRO, and DTS) in its center of mass. Before each drop, the headform was positioned in the required orientation and height using strings attached to an "energize to release" electromagnet (Fig. 1). For unhelmeted impacts, the strings were used as slings (Fig. 1a) while for helmeted impacts, the strings were attached to the helmets' ventilation openings (e.g.,—see Fig. 2). String positioning and the length of slings were adjusted to achieve required helmet orientation and height between each test, in a similar approach to previous work [58]. Tests were either unhelmeted, or with one of five different certified [31–33] and commercially available ice hockey helmets. The helmets represented various price ranges, liner materials, and helmet designs (Table 1; Online Resource 3, Fig. S1). The helmets' chin straps were closed, and a tight fit, following manufacturer and retailer recommendations, was checked before every impact.

Two impact surface orientations (Fig. 2), perpendicular (flat) and 45° inclined (oblique), relative to the falling direction, were used. Material layers of different compliances were applied to both anvils. During impacts onto the flat anvil, five different surfaces were used; MEP Pad (1-inch height, 6-inches diameter, 60 ± 2 Shore A hardness, CadexInc) and layered Ethylene-vinyl acetate (EVA) (EVAZOTE-50, algeos.com [59]) foam sheets bonded with double-sided tape giving 24, 48, 72, and 96 mm overall thickness. The headform was dropped onto three centric sites (force vector passing through headform center of mass); Front, Side, and Rear (Fig. 2a–c). During impacts onto the 45° anvil, three-layered EVA foam thicknesses (24, 48, and 72 mm) were used, with two centric (Front and Rear) and

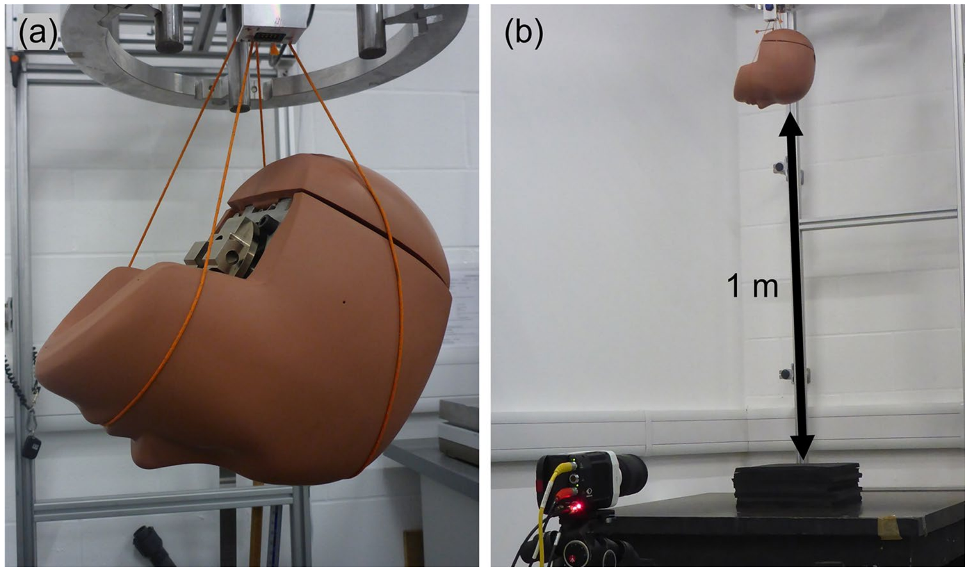


Fig. 1 **a** Hybrid III headform held over the impact surface. **b** Setup used for free-fall drop test onto the flat 96 mm foam (8 layered sheets) surface with a high-speed camera at 1 m distance from the impact location

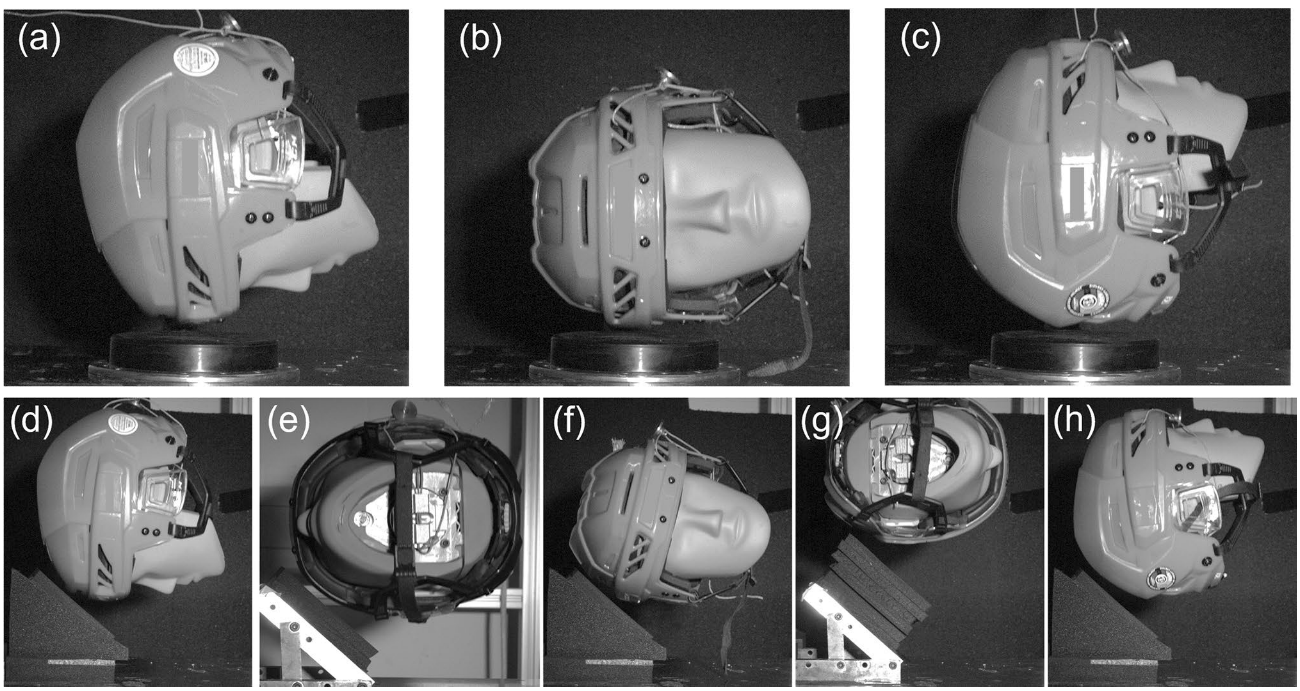


Fig. 2 Impacts onto the (**a**–**c**) Flat and (**d**–**h**) 45° oblique anvil; (**a** and **d**) Front, (**e**) FrontBoss, (**b** and **f**) Side, (**g**) RearBoss, and (**c** and **h**) Rear impact location

three non-centric sites (FrontBoss, Side, RearBoss) impacted (Fig. 2d–h). Similar to certification standards [31–33], the headform was dropped from a height of 1 m (Fig. 1b), resulting in an impact velocity of 4.5 m/s and energy of 51.3–53.8 J, varying with helmet mass (Table 1). For every impact configuration, three trials were carried out. A new helmet was used for each impact surface.

Linear acceleration and angular velocity were measured with a sampling frequency of 100 kHz for 70 ms (20 ms pre-trigger and 50 ms post-trigger), triggered when a 5 g threshold was exceeded in any axis. A CFC 1000 filter, as recommended by Post et al. [60], was applied to each linear accelerometer axis using DTS SLICEWARE (Version 1.08.0868). A 4-pole Butterworth low-pass filter with a cut-off frequency of 200 Hz was applied to each angular velocity axis, chosen based on a frequency analysis using a Fast Fourier Transform [61]. After filtering, the angular velocity data were differentiated to

Table 1 Helmet models with respective price, mass, and liner materials

Helmet	Price [US$]	Mass [grams]	Liner materials
1	44.99	498	Dual-density vinyl nitrile (VN) foam
2	89.99	549	Triple-density foam liner
3	179.99	558	Expanded polypropylene (EPP), shear-thickening polymer (STP)
4	259.99	543	EPP & VN foam, slip-layer
5	299.90	747	VN foam, STP, slip-layer

Prices shown are suggested retail prices during the time of the study. Make and models blinded, with information taken from product guides (also blinded). Images of shells and liner systems are in Online Resource 3, Fig. S1

Table 2 Statistical characteristics of the reference dataset [26]

	PLA [g]	PAA [krad/s^2]	D [ms]
Mean (± SD)	28.8 (± 11.8)	3.44 (± 1.40)	26.2 (± 3.2)
Minimum	7.7	0.69	18.0
Lower quartile (Q1)	20.9	2.58	24.6
Median	26.8	3.55	25.5
Upper quartile (Q3)	35.3	4.00	28.6
Maximum	67.8	7.85	35.8

obtain angular acceleration. Peak linear (PLA) and peak angular acceleration (PAA), impact duration (D), time to peak (TTP), and rebound time (RT) were obtained from filtered data. These values were compared quantitatively while the acceleration vs. time trace shapes were compared qualitatively to reference values from concussive head impacts, recreated in a laboratory setting (Table 2, [26, 37]. To enhance visualization, a pairwise distance function, normalized to maximum values obtained in this data collection (linear acceleration/350 g, angular acceleration/11 krad/s^2, and duration/35 ms) was added to Figs. 4 and 5 as a shaded area using a colormap function in MATLAB (R2018a). A 10% pairwise distance corresponded to 35 g, 1.1 krad/s^2, 3.5 m/s, or the Pythagorean equivalent distance from a reference value (i.e., $\sqrt{(\Delta PLA^2 + \Delta D^2)}$ or $\sqrt{(\Delta PRA^2 + \Delta D^2)}$). To assess reliability of repeated measurements, two-way mixed model intra-class correlation coefficients (ICC (3, 1)) with absolute agreement definition, and their respective 95% confidence intervals (CI), were calculated (IBM SPSS 26) for PLA, PRA, and D [62]. Additionally, two measures to assess head impact severity, the Head Injury Criterion (HIC, Eq. 1, [41]), and the Rotational Injury Criterion (RIC, Eq. 2, [42]), were calculated. All obtained values were compared between tested helmets and to the unhelmeted impacts to assess the helmets' impact performance:

$$HIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} a(t)dt \right\}^{2.5} \right]_{max} \quad (1)$$

where *a(t)* is resultant linear acceleration [g]. Time points t_1 and t_2 [s] maximize the obtained HIC value and do not exceed a time interval of 0.015 s [41].

$$RIC = \left[(t_2 - t_1) \left\{ \frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \alpha(t)dt \right\}^{2.5} \right]_{max} \quad (2)$$

where $\alpha(t)$ is resultant angular acceleration [rad/s^2] at the headform's center of mass. Times t_1 and t_2 are similar to those in Eq. 1, but with a maximum time interval of 0.036 s [42].

All impacts were filmed using a high-speed video camera (Phantom Miro R311, Vision Research Ltd., Bedford, UK; resolution, 1024 × 768 pixels, 0.5 mm/pixel; sample rate 2000 fps; exposure, 500 µs; lens, Nikon AF Nikor 24–85 mm). The camera was positioned on a tripod at a distance of 1 m from the impact location, with the field of view perpendicular to the impact surface (Fig. 1b).

3 Results

Linear and angular acceleration vs. time data (Fig. 3) show a single peak, characteristic of collisions between players [37]. Acceleration vs. time traces for Front site impacts (Fig. 3) are similar to other tested impact sites (Online Resource 3, Fig. S2—Fig. S8). Helmeted impacts are pooled together for Figs. 4 and 5 as differences between helmeted impacts are small compared to differences to unhelmeted impacts. The highest impact accelerations and shortest durations were produced during impacts onto the stiff MEP Pad. For impacts onto the flat anvil, helmets reduced peak linear acceleration by up to 71% and increased the impact duration by up to 161% (Fig. 4). Most helmeted impacts were within 10% of the reference values, according to the pairwise distance function, normalized to maximum values, shown as shaded areas. The 10% pairwise distance corresponded to 35 g, 3.5 ms, or the Pythagorean equivalent distance.

For stiff surface impacts onto the oblique anvil, helmets reduced peak linear and angular acceleration by up to 64 and 53%, respectively (Fig. 5), while increasing impact durations

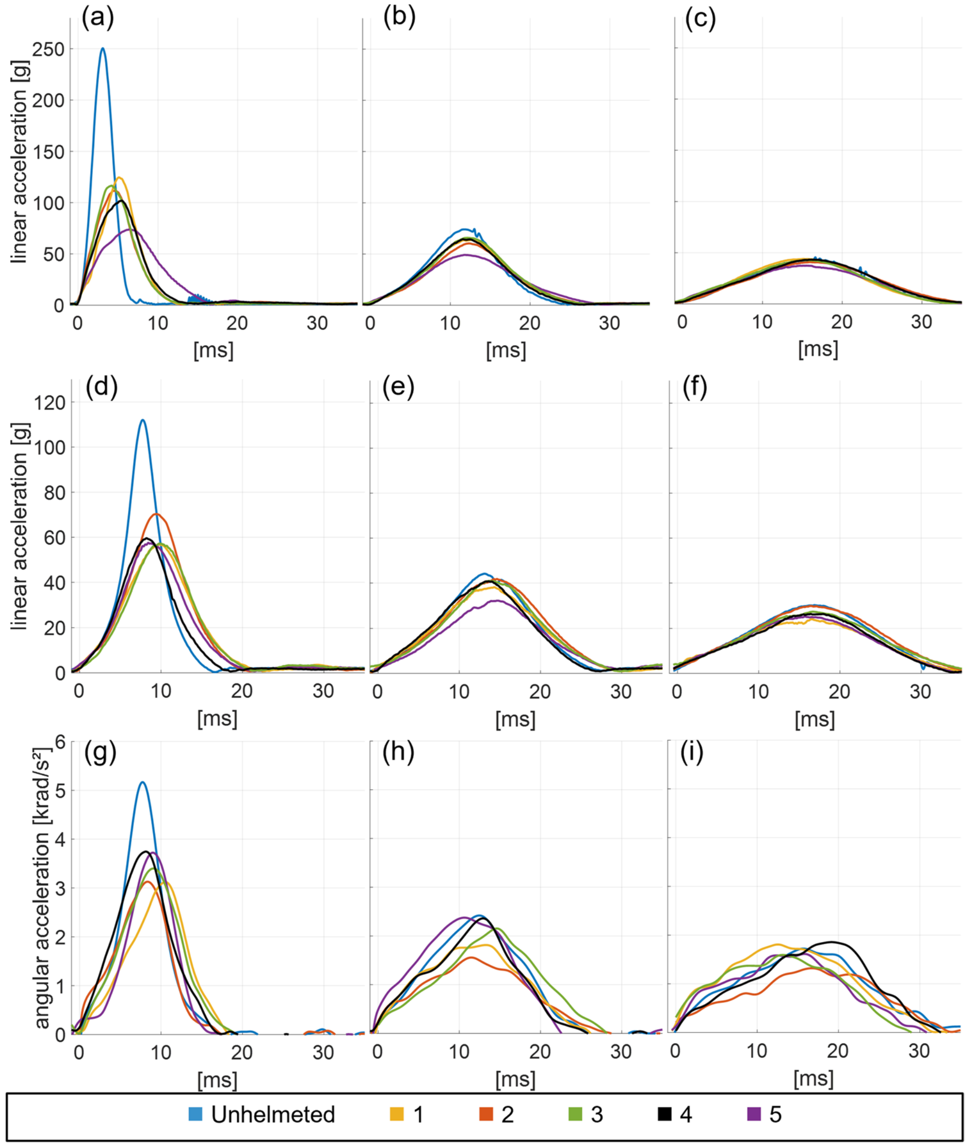


Fig. 3 **a**–**f** Linear and (**g**–**i**) angular acceleration vs time traces for (**a**–**c**) flat and (**d**–**i**) oblique surface, Front site impacts onto the (**a**) MEP Pad, (**d and g**) 24 mm foam layer, (**b**, **e** and **h**) 48 mm foam layer, (**f** and **i**) 72 mm foam layer, and (**c**) 96 mm foam layer

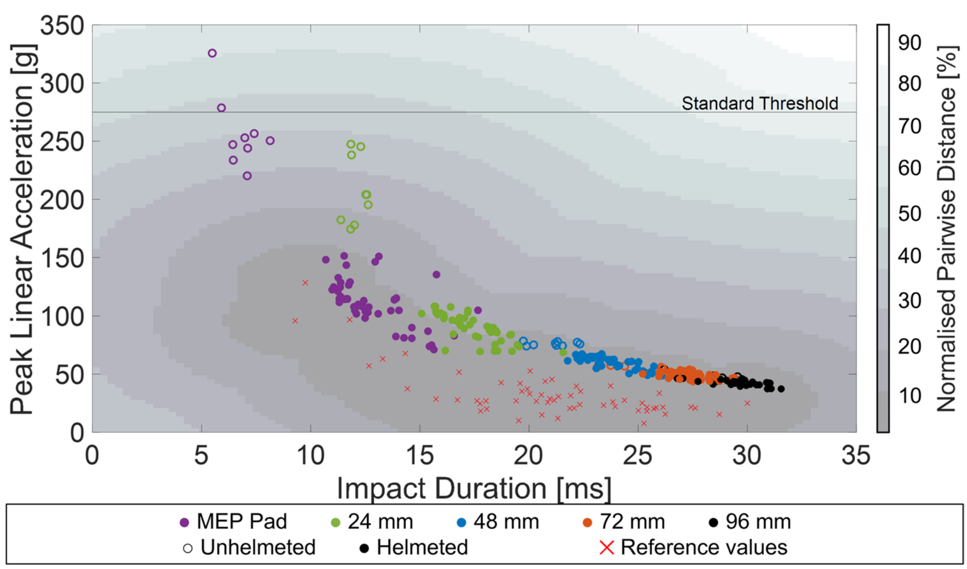


Fig. 4 Peak linear acceleration vs. impact duration for all impacts onto the flat surface; filled markers represent helmeted impacts and unfilled markers represent unhelmeted impacts. Shaded areas represent a normalised (acceleration/350 g and duration/35 ms) pairwise distance from the reference values in percent. Reference values of concussive impacts recreated in laboratory environment obtained from [26]

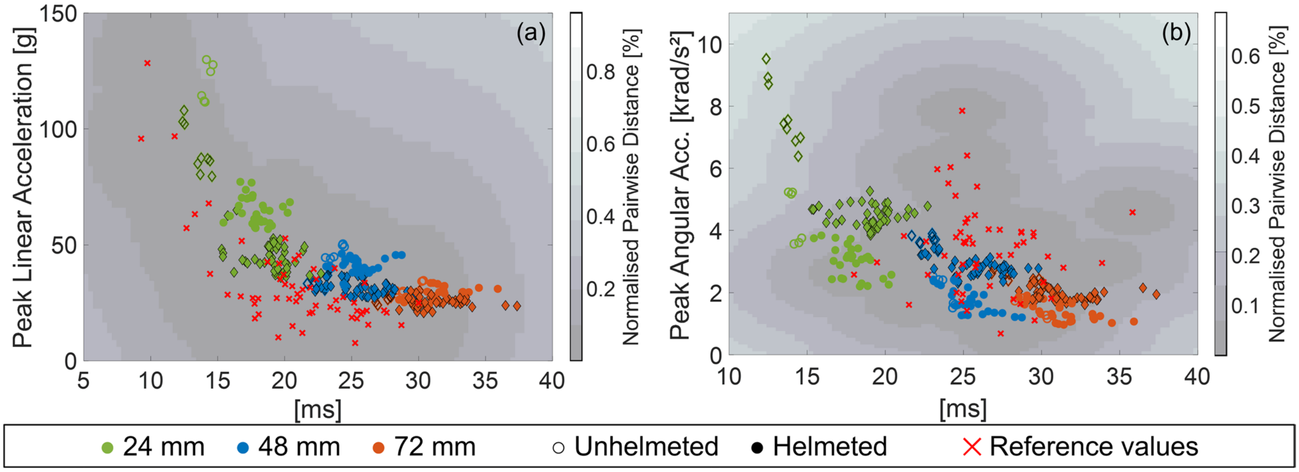
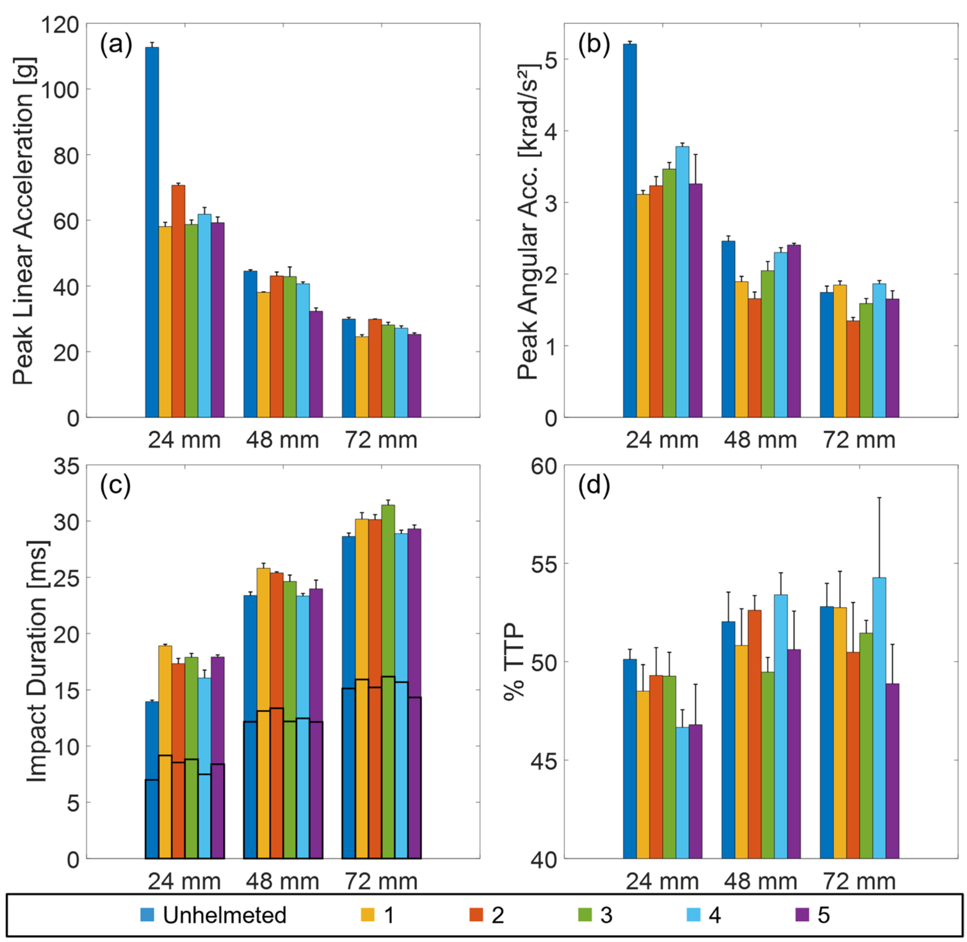


Fig. 5 Peak (**a**) linear and (**b**) angular acceleration vs. impact duration for all impacts onto the oblique surface; filled markers represent helmeted impacts and unfilled markers represent unhelmeted impacts, circles represent centric impacts and diamonds with a black outline represent non-centric impacts. Shaded areas represent a normalised (lin. acceleration/350 g, ang. acceleration/1.1 krad/s^2, and duration/35 ms) pairwise distance from the reference values in percent. Reference values of concussive impacts recreated in laboratory environment obtained from [26]

Fig. 6 Mean peak (**a**) linear and (**b**) angular accelerations, (**c**) mean impact durations with horizontal bars indicating the proportion of time to peak (bottom half) and rebound time (top half), and (**d**) percentage of time to peak of the total impact duration for all oblique surface, showing Front site impacts with other sites in Online Resource 3, Fig. S9—Fig. S11 & Fig. S13—Fig. S16



by up to 66% (Fig. 6c). Peak acceleration decreased with increasing impact surface compliance, while impact duration increased, as expected [21]. Again, most helmeted impacts were within 10% (35 g, 1.1 krad/s^2, and 4 ms) of the normalized pairwise distance function shown as shaded areas (Fig. 5).

For some compliant surface impacts (72 mm and above), the helmets did not reduce peak accelerations but still increased impact durations (Figs. 4 and 5). During oblique impacts, centric impact sites produced higher linear accelerations and lower angular accelerations than non-centric impact sites (Fig. 5). For non-centric impact sites angular accelerations were higher than linear accelerations. Impact durations and the difference between unhelmeted and helmeted tests were typically smaller for centric impact sites (Fig. 5).

Differences in linear (Fig. 6a); Online Resource 3, Fig. S9 & Fig. S13) and angular (Fig. 6b); Online Resource 3, Fig. S14) peak accelerations between tested helmets were seen in some impacts. No trend of a helmet producing lower peak accelerations compared to other helmets was observed. Helmets increased impact duration for all impact surface compliances (Fig. 6c); Online Resource 3, Fig. S10 & Fig. S15). The proportion of time to peak (TTP) and rebound time (RT) increased from 48 to 52% with increasing surface compliance (Fig. 6d); Online Resource 3, Fig. S11 & Fig. S16). Clearly observable differences between helmets were obtained for the HIC (Fig. 7a); Online Resource 3, Fig. S12 & Fig. S17) and the RIC (Fig. 7b); Online Resource 3, Fig. S18).

Mean values, standard deviations (SD), ICC and their respective 95% CI suggest excellent reliability (Table 3). To mitigate the effect of the large true score variance in the dataset, additional calculations, where measurements were grouped by impacted surface and where unhelmeted impacts were excluded, were carried out (Online Resource 3, Table S1 & Table S2). The lowest obtained ICC was 0.838 with lower and upper limits of a 95% CI of 0.619 and 0.941, respectively, suggesting excellent reliability.

Table 3 Mean values, standard deviations (SD), ICC and their respective 95% CI for flat surface PLA and D, and oblique surface PLA, PAA, and D

	Mean (±SD)	ICC	95% confidence interval
Flat, PLA [g]	70.6 (±28.1)	0.995	0.991→0.997
Flat, D [ms]	22.1 (±6.3)	0.996	0.994→0.997
Oblique, PLA [g]	38.3 (±13.2)	0.991	0.987→0.994
Oblique, PAA [krad/s2]	2.66 (±1.10)	0.997	0.995→0.998
Oblique, D [ms]	25.1 (±5.5)	0.990	0.984→0.993

Unhelmeted impacts were excluded in the calculations of the ICC values shown. ICC values for measures grouped by the impacted surface and with unhelmeted impacts included are shown in Online Resource 3, Table S1 & Table S2

4 Discussion

A broad range of headform kinematic responses were obtained. As expected, peak accelerations decreased (Fig. 6a and b) while impact durations increased (Fig. 6c) with surface compliance (Figs. 4 and 5) [21, 24, 38]. Increasing surface compliance had a greater effect on the unhelmeted headform than the helmeted one, as expected [21, 24, 25, 38]. As such, the difference between helmeted and unhelmeted impacts decreased up to a point where a fitted helmet made no measurable difference to the peak accelerations, but impact durations still increased (Figs. 4 and 5). This effect of decreasing helmet effectiveness with increasing surface compliance, also shown previously [21, 24], suggests that the free-fall drop test with interchangeable surfaces can

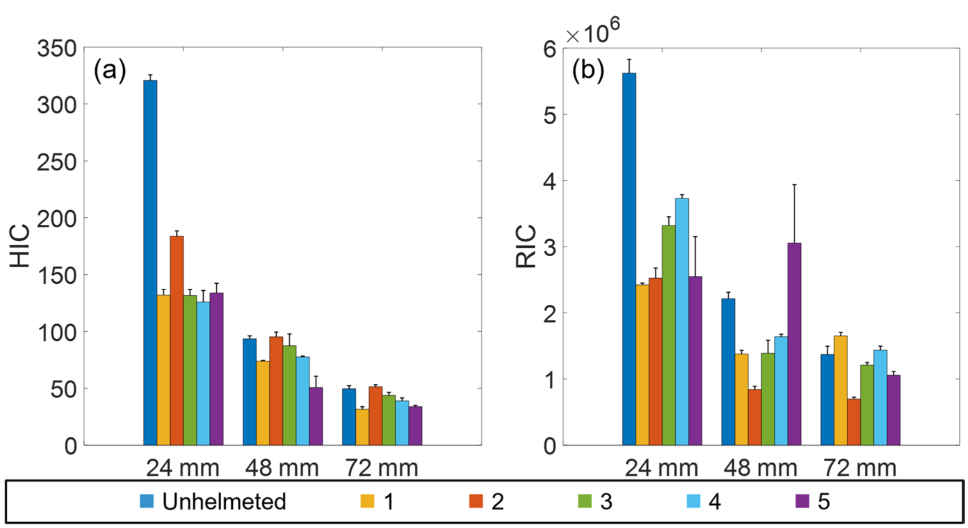


Fig. 7 Mean (**a**) HIC and (**b**) RIC for all oblique surface, showing Front site Impacts with other sites in Online Resource 3, Fig. S12, Fig. S17, and Fig. S18

replicate ice hockey shoulder and elbow to head impacts. ICC and SDs (Table 3, Figs. 6 and 7) suggest excellent reliability of repeated measures [63]. Due to the large true score variance in the dataset, ICC should be considered cautiously [64]. However, due to the ICC still suggesting excellent reliability when measures were grouped by surface and when unhelmeted impacts were not considered (Online Resource 3, Table S1), the test method (within-subjects) appears to have high reliability [62].

Comparing impacts with reference values published by Post et al. [26], the closest representation was achieved with oblique, 48-mm foam (Fig. 5), non-centric impacts. Flat surface impacts (Fig. 4) generally produced higher linear accelerations than the reference data. Impacting foam layers of 72 mm, or thicker, produced longer duration impacts than the reference data (Figs. 4 and 5).

Impacting centric impact sites during oblique impacts resulted in higher linear accelerations and lower angular accelerations than the reference data (Fig. 5). Impacting non-centric locations produced magnitudes in an acceptable range of the reference data (mean difference < 3 g and 0.62 krad/s^2). Due to the spread and skew in the reference data, and the lower variation and relatively normal distribution in the data we collected, this free-fall drop test can only replicate a proportion of the dataset. For 48-mm foam, non-centric impacts mean PLA (31.5 ± 2.8 g) is between the median and Q3, mean PAA (2.8 ± 0.3 krad/s^2) is between Q1 and the median, and mean D (25.8 ± 1.8 ms) is between Q1 and Q3 of the reference data (Table 2), respectively. Hence, presented results lie in a range of potentially concussive real-world ice hockey collision head impacts (Fig. 5, [26]), and while they do not cover the whole range within the dataset, they are more precise (less variable). Future work could modify the angle of the oblique impacted surface to tailor the headform's peak accelerations to different areas of the reference data, to improve representative accuracy.

Obtained linear and angular acceleration vs. time traces (Fig. 3) were similar to reported shoulder/head collisions [37]. Commonly used guided drop tests [49] produce acceleration vs. time traces with an initial high, short-duration peak followed by a longer duration, lower magnitude peak [37] similar to falls. In head impact research, the single peak shape, characteristic for collisions is generally achieved using a horizontal impactor [26, 37, 49]. We have produced a collision type acceleration vs. time trace shape with a drop test.

Comparing measured peak accelerations between helmets shows differences in some impacts. Most (538 of the 540) impacts produced lower peak accelerations than the 275 g threshold used in the standards [31–33], with two tests onto the MEP pad exceeding the threshold (Fig. 4). We used a different test setup—a free-fall drop test—so the pass/fail threshold should only be indicatively applied. No helmet consistently outperformed all other helmets. Greater differences between helmets were found for the calculated Head Injury Criterion and the Rotational Injury Criterion (Fig. 7), that both consider a maximum time interval [41, 42]. Helmets that produced high linear acceleration (or HIC) generally produced low angular acceleration (or RIC), and vice versa (Figs. 6a and b and 7). However, no conclusions can be made about which materials or design features are preferential in an ice hockey helmet based on the obtained data. The proportion of time to peak of the total impact duration increased with increasing surface compliance (Fig. 6d), which is likely caused by the compressed surface and not the helmet.

The reference dataset was obtained from laboratory recreated head impacts [26], as an in-field head injury data set is not publicly available. Despite being considered the best available estimate, a dataset of measured concussive impacts in ice hockey could increase the confidence in head impact research, including this study. Variations in measured linear accelerations, angular velocities, and head impact measures for some impact conditions along with the analysis of the high-speed video footage show that it is difficult to hit the predefined impact site precisely. While ICC (within-subject) suggest excellent reliability [63], future work could modify the test setup to use a drop carriage similar to Meehan et al. [65], carry out more repeated measures per impact scenario, and assess interrater reliability to further assess variation [62, 64].

It is possible that the EVA foam, used to produce different impact surface compliances, degraded between tests. Future testing could consider the durability of the impacted surface. An anthropometric headform such as the 50th percentile Hybrid III used here has limited biofidelity and only partially represents head geometry, helmet fit, and friction between head and helmet [66, 67]. The Hybrid III headform is, however, widely accepted and used in head impact research [49]. Further work could use different headforms, include an anthropometric neckform, and study predicted brain-stresses and -strains using finite element models. The five different helmets fitted onto the headform in this study were chosen to represent the range of helmet designs and the price range of commercially available helmets. In future work, adding additional helmets could increase confidence in the dataset and would give further insights into the effectiveness of different helmet designs.

5 Conclusion

This study demonstrated that by modifying the impacted surface, a free-fall drop test can produce kinematic responses similar to the method currently used to replicate ice hockey collisions. A wide range of head impact scenarios, representative of falls onto ice and collisions with other players, can be replicated using this method. One meter drops onto a 48 mm layered EVAZOTE-50 foam surface, aligned at 45°, gave peak linear and angular accelerations, and impact durations, within 10% of those obtained by current best practice methods. These findings facilitate a simpler test protocol for ice hockey helmets, either for adoption in certification standards or research.

Supplementary Information The online version contains supplementary material available at https://doi.org/10.1007/s12283-023-00416-6.

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Data availability Data will be stored on Sheffield Hallam University's Research Archive for five years, and will be made available upon reasonable request.

Declarations

Conflict of interest The authors declare that they have no conflict of interest.

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