Analysis of the protective capacity of ice hockey helmets in a concussion injury reconstruction

Andrew Post, Clara Karton, T. Blaine Hoshizaki, Michael D. Gilchrist

Abstract Concussion unfortunately is a common and serious injury in ice hockey, involving a large number of youth and pro level athletes. The purpose of this research was to examine the efficacy of ten ice hockey helmet models in protecting players. A head impact during a professional ice hockey game that resulted in a documented concussion was reconstructed in the laboratory. The impact parameters were determined from video analysis of the incident that resulted in a concussion. The event was reconstructed in the laboratory using mechanical impacts and finite element modelling. The performance of the ice hockey helmets was measured using peak linear and rotational acceleration. The acceleration time histories were used as input to the University College Dublin Brain Trauma Model to determine the maximum principal strain. The results demonstrated that the shell and liner design of the helmets influence the motion of the head during an impact. Low responses in peak resultant linear acceleration do not reflect low responses in peak resultant rotational acceleration. When comparing the helmets using maximum principal strain as the metric, there were differences between helmet models not reflected in the kinematic data. This suggests that the magnitude of strain is not fully represented by peak resultant linear and rotational acceleration. In addition, while the differences between the helmets' protective capacity were statistically significant, those differences would not have appreciably affected the likelihood of concussive injury in this reconstruction. This suggests that helmets in ice hockey currently may have similar protective capacities in terms of concussive injury.

Keywords Biomechanics, concussion, finite element modeling, helmets, Ice hockey.

I. INTRODUCTION

Concussion unfortunately is a common and serious injury in ice hockey, involving a large number of youth and pro level athletes. [1-2]. Skull fracture and focal brain injury however have become uncommon since the introduction of helmets [3]. While this reduction in skull fracture and focal brain injury has been associated with the introduction of helmets to the sport, the incidence of concussion has not been affected by improvements in helmet technologies [4-5]. It is likely that the lack of efficacy of helmets to protect against concussion is in part because they are not designed to manage rotational accelerations from an impact where rotational acceleration has been correlated with this injury [6-8]. Concussion is an injury that has been demonstrated to be linked to rotationally-induced strains within the brain tissue [9]. These high strains in the brain tissue have been shown to be correlated to rotational acceleration, not linear acceleration [6][9]. Since current ice hockey helmets are not evaluated using rotational acceleration, the ability to manage this type of injurious head motion through helmet technologies has not been investigated or optimized to reduce the risk of concussion. In fact, it is currently unknown if helmets are capable of reducing any metric associated with concussion to the point where a reduction in the risk of injury might be expected.

The authors have investigated the ability of different ice hockey helmet liner materials to attenuate energy under prescribed impact conditions, showing that vinyl nitrile (VN) foams may have some beneficial characteristics in terms of rotation in comparison to expanded polypropylene (EPP) [10]. This study was however limited in that the impact locations used were reflective of high-risk injury sites performed on a Hybrid III headform and not a human subject [11]. In addition, the comparison was between similar VN and EPP ice hockey helmet models. As a result there is a need to investigate how various ice hockey helmet models influence the risk of concussive injury for a simulated impact that elicited an on-ice concussion. This will improve current knowledge and provide a bench mark in performance characteristics of current ice hockey

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helmet design that may aid in the reduction of rotational acceleration in future development. It will also establish the ability, or inability, of current ice hockey helmet designs to influence the risk of concussive injury. The purpose of this research is to examine the efficacy of several different types of ice hockey helmets in protecting a professional ice hockey player for a documented incident where a concussion occurred.

II. METHODS

Experimental Testing

To examine the efficacy of the different CSA/HECC certified ice hockey helmets, an ice hockey impact reconstruction was conducted where a professional ice hockey player sustained a concussion. To determine the laboratory parameters to conduct the reconstruction, information such as impact location, velocity and angle was determined from available high definition video of the event. In this case two different angles of the event were available and analyzed. These parameters were identified using Kinovea 0.8.2 video analysis software. The impact velocity was determined by applying a perspective grid based on known points and distances on the ice hockey rink, which established the displacement of the player (figure 1). Displacement over time was calculated to determine the contact velocity. The impact angles and location were also determined using this perspective grid (figure 1). The type of helmet worn by the player was determined by video and also by available reports on the injury. An analysis of measurement error of this methodology was conducted and determined to be approximately 5% for velocity and approximately 10 degrees for impact angle [12]. The player was reported by the medical staff to have incurred a grade 2 concussion and missed 24 games (55 days). The initial symptoms of the impact included severe headache and fatigue. From this impact, the player developed long-term persistent concussion symptoms including: nausea, headache, dizziness, seeing spots and depression.

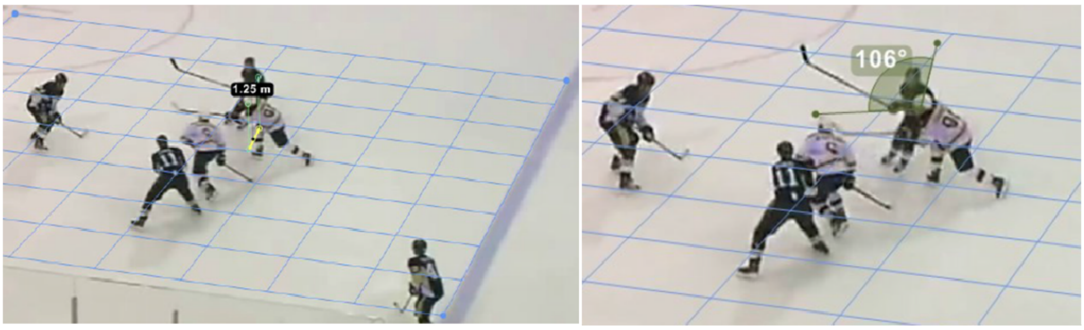

Figure 1. Perspective grid and displacement calculation using Kinovea (left) and angle of impact calculation (right)

The laboratory reconstruction was conducted using the impact parameters determined from the video. As this event was a body to head collision, a linear impactor was used to impact the Hybrid III headform to mimic the characteristics of a collision. A linear impactor was used as it represents the reconstruction of an object to object impact, as is the case for many sport reconstructions [10,11,13]. The linear impactor was outfitted with a vinyl nitrile (VN) 602 cap to simulate the compliance of a shoulder and shoulder pad [12]. The mass of the impactor was 13.1 kg, which is reported in the literature to be similar to the mass of a full body/shoulder impact for this type of impact in ice hockey [12]. The impact velocity was determined to be 6.5 m/s and the impact location is shown in figure 2. Under these impact conditions, 10 different models of ice hockey helmet were impacted (table 1). Five new helmets were used per helmet model, with each impacted once to eliminate the effect of degradation of material. This resulted in a total of five impacts per helmet model. The resulting linear and rotational acceleration time histories from the Hybrid III headform were used as input for determining the brain strains from the reconstruction using the University College Dublin Brain Trauma Model (UCDBTM). The FE modeling software used was ABAQUS (Dassault Systems Simulia Corp, USA).

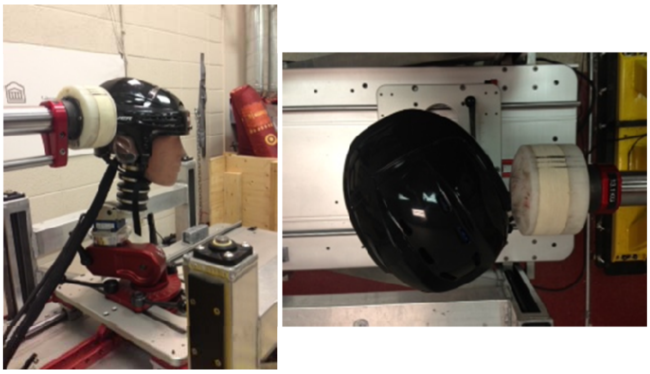

Figure 2. Laboratory reconstruction of the concussion impact.

Equipment

The pneumatic linear impactor consisted of two main parts: the frame and the table (figure 3). The frame consisted of the air tank and the impacting arm (13.1 +/- 0.01 kg). The table was where the Hybrid III headform was affixed. The headform was attached by a Hybrid III neckform to a sliding table which allowed for movement post-impact. The sliding table had a mass of 12.78 +/- 0.01 kg. Impact velocity was determined by a photoelectric time gate placed just prior to the impact (0.02 m).

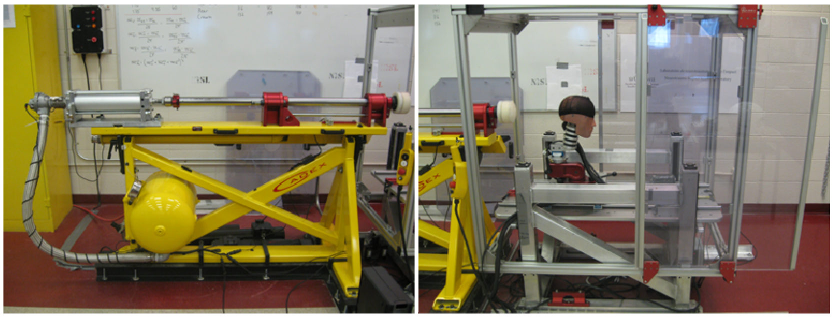

Figure 3. Linear impactor frame (left) and table with Hybrid III headform (right)

A 50th percentile Hybrid III headform (4.54 +/- 0.01 kg) and neckform (1.54 +/- 0.01 kg) was used for this reconstruction. The Hybrid III headform was outfitted with a 3-2-2-2 accelerometer array for the measurement of head motion in x, y and z axes of linear and rotational acceleration [14]. The accelerometers used were Endevco 7264C-2KTZ-2-300 and were sampled at 20 kHz. The signal was collected by Diversified Technical Systems (DTS) Pro lab module and processed using TDAS software. The accelerometer signals were filtered using a CFC class 1000 filter.

A total of ten ice hockey helmet models were used for this reconstruction. Each helmet was placed upon the headform following the manufacturer's fitting guide. Helmet specifications can be found in table 1 and images in Figure 4. The helmets have been labeled to avoid easy identification following the suppliers' request as this research was intended to examine a broad range of helmet designs for their efficacy in reducing parameters linked to concussion, not as a comparative examination.

Table 1. Helmet material and thickness at the impact site. Helmets E and J both had technologies that claimed to reduce rotational acceleration specifically.

	Helmet									
	A	B	C	D	E	F	G	H	I	J
Helmet liner thickness (mm)	18.2	16.5	16.2	18	12.5	13	17.8	14	14.1	14
Helmet liner material	EPP	EPP	VN	EPP	EPP + Rotation	EPP	VN	VN	EPP	3D* structure + Rotation

*3D structure refers to three-dimensional structure that used thermoplastic polyurethane as the material.

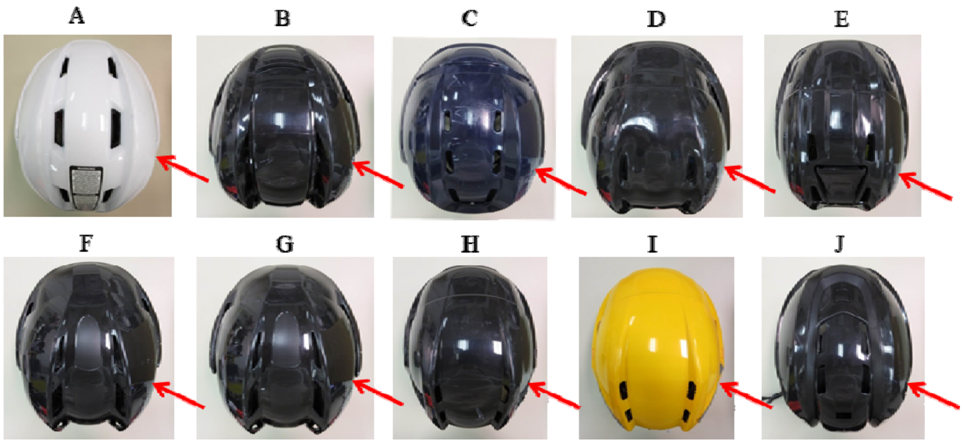

Figure 4. Top down view of the geometry of each helmet. Red arrow denotes impact site on each model.

Computational Modeling

The finite element model used for this research was the University College Dublin Brain Trauma Model (UCDBTM) [15-16]. The geometry of the model was based upon CT scans of a male cadaver and, as such, does not represent the typical 50th percentile of the human male. This model is comprised of the dura, cerebrospinal fluid (CSF), pia, falx, tentorium, grey and white matter, cerebellum and brain stem. The UCDBTM had approximately 26,000 elements and was validated against Nahum et al's [17] and Hardy et al's [18] cadaver impact research. Real life reconstructions were conducted to further validate the model and were found to be in good agreement with lesions on CT scans for TBI incidents [19-20].

The material properties of the model are presented in tables 2 and 3. The brain characteristics were derived from Zhang et al [21]. A linear viscoelastic material model combined with large deformation theory was used to model the brain tissue [15-16][22-23]. The compressive behaviour of the brain was considered elastic, and the shear characteristics of the brain were defined as:

$$G(t) = G\infty + (G0 - G\infty)e\text{-}\beta t$$

where $G\infty$ is the long-term shear modulus, G0 is the short-term shear modulus and β is the decay factor [15]. The hyperelastic material model which was used to model the brain tissue is expressed as

$$C\_10 (t)=0.9C\_01 (t)=620.5+1930e\text{^}(t/0.008)+1103e\text{^}(t/0.15) \quad (Pa)$$

where C10 and C01 are temperature-dependent material parameters and t is the time in seconds. The brain-skull interaction was accomplished by modeling the CSF as solid elements with a high bulk modulus and low shear modulus; the contact definitions allowed for no separation and used a friction coefficient of 0.2 [24].

Table 2. Material characteristics of the UCDBTM

Material	Poisson's Ratio	Density (kg/m^3)	Young's Modulus (Mpa)
Dura	0.45	1130	31.5
Pia	0.45	1130	11.5
Falx	0.45	1140	31.5
Tentorium	0.45	1140	31.5
CSF	0.5	1000	-
Grey Matter	0.49	1060	Hyperelastic

White Matter	0.49	1060	Hyperelastic

Table 3. Material characteristics of the brain tissue used for the UCDBTM

Material	Shear Modulus (kPa) G_0	G_∞	Decay Constant (Gpa)	Bulk Modulus (s^{-1})
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

Statistics

The variables used to compare the helmets were peak resultant linear acceleration, peak resultant rotational acceleration and maximum principal strain (MPS). Statistical significance was determined by ANOVA with a Tukey post hoc when significant main effects are found. The confidence interval was set at 95%.

III. RESULTS

The results of the reconstruction in peak resultant linear acceleration, peak resultant rotational acceleration and maximum principal strain are presented in table 4. The helmet designated H was the helmet that was reported to have been worn by the player at the time of the incident. Significant main effects were found between the helmets for peak resultant linear and peak resultant rotational acceleration as well as MPS ($p<0.05$). The helmets that had the lowest response in peak linear acceleration were helmets B (73.4 g), C (73.1 g), D (73.6 g), I (75.0 g) and J (79.4 g) ($p<0.05$). The helmets that produced the lowest magnitude rotational acceleration from the concussion reconstruction were helmets C (6.0 krad/s^2), D (6.6 krad/s^2) and J (6.1 krad/s^2) ($p<0.05$). When comparisons were made using MPS calculated from the UCDBTM, the following helmets produced the lowest responses: A (0.457), C (0.408), D (0.478), F (0.445), G (0.476), H (0.445), I (0.450) and J (0.412) ($p<0.05$) with only two helmets with significantly larger magnitudes (B: 0.509 and E: 0.496) ($p<0.05$).

Table 4. Peak resultant accelerations and maximum principal strain results for the ice hockey concussion reconstruction. Standard deviation in parentheses.

	Helmet A	B	C	D	E	F	G	H	I	J
Helmet liner thickness at site (mm)	18.2	16.5	16.2	18	12.5	13	17.8	14	14.1	14
Linear acceleration (g)	82.2 (7.3)	73.4 (3.0)	73.1 (2.2)	73.6 (1.8)	87.7 (1.8)	82.4 (2.0)	86.8 (2.5)	80.2 (1.6)	75.0 (0.5)	79.4 (2.9)
Rotational acceleration (krad/s^2)	7.4 (1.1)	7.3 (0.4)	6.0 (0.5)	6.6 (0.7)	9.5 (0.3)	8.0 (0.4)	8.6 (0.9)	7.4 (0.4)	7.3 (0.2)	6.1 (0.4)
Maximum principal strain (mm/mm)	0.457 (0.05)	0.509 (0.01)	0.408 (0.05)	0.478 (0.06)	0.496 (0.02)	0.445 (0.01)	0.476 (0.02)	0.445 (0.02)	0.450 (0.01)	0.412 (0.01)

IV. DISCUSSION

The results of the reconstruction for this ice hockey concussion event indicated magnitudes of response that would suggest a risk of brain injury [25-30]. The linear acceleration responses for the impacts to the Hybrid III headform had proposed magnitudes in the region of 25% risk of incurring a concussion for the helmets with the lowest responses [25]. The helmets that produced the largest magnitude response in the region of 80 g or more crossed into the range of 50% risk of injury as reported in the literature from impact reconstructions [25]. These comparisons are consistent with on-field helmet sensor data, where magnitudes of 70 to 80 g have been shown to produce concussions for impacts in American football [26-27]. The rotational acceleration response also indicated a high risk of concussion with all helmets having magnitudes above a 50% risk, and the worst performing helmets above an 80% risk of brain injury [25]. This trend continues for the MPS results, with strains in excess of 40%, indicating greater than 50% risk of concussion [25][28-32]. When the different helmets were compared using linear acceleration as the performance metric, five helmets performed similarly and had lower magnitudes of response in comparison to the remainder. The liners of these helmets were EPP (3 helmets), VN (1 helmet) and three-dimensional structure with rotation technology (1 helmet). Liner thickness for the helmets that produced the lowest linear magnitude results were varied and not always thicker than those helmets with higher values, indicating liner thickness does not necessarily improve performance under these impact conditions. Other factors such as shell geometry may have had an effect upon these linear acceleration responses (figure 4).

Peak resultant rotational acceleration responses were the lowest in only three of the ten helmets. The three helmets were: C (VN), D (EPP) and J (three-dimensional structure and rotational components). These helmets produced lower magnitude linear accelerations in addition to lower rotational accelerations. The remaining two helmets (B and I) that produced lowest linear accelerations did not produce lower results when rotational acceleration was used as the metric. This indicates that a helmet with a low linear acceleration result does not necessarily produce a low rotational acceleration result. Interestingly, helmets C and D are the same model but VN and EPP versions, and thus have the same shell geometry. It is possible that the shell geometry in combination with the VN and EPP liner may have an effect on improving the rotational acceleration responses for this particular ice hockey impact location and condition. Helmet J had technology that was designed to specifically reduce rotational acceleration and used three-dimensional structures as a linear acceleration managing component. For this particular impact it appears that this helmet produces lower results than most conventional helmet models, except helmets C and D. Both C and D are larger helmets overall, with thicker liners than helmet J, suggesting that a slightly bigger helmet (shell geometry and liner) may perform better regardless of technology used if optimized for ice hockey impacts (Figure 4). While many of the other helmets were also bigger in dimensions (shell and liner thickness), they may have not had an interaction between ice hockey helmet geometry and liner that C and D benefitted from.

When comparisons were made between the helmets using MPS, many of the differences found for peak linear and rotational acceleration disappeared. Helmets that performed well rotationally (C, D and J) also produced low results for MPS, but were not different from helmets A, F, G, H and I. Overall, the lowest results were 0.408 to 0.478 maximum principal strain. Many of these helmets did not produce lowest magnitude results for linear or rotational acceleration, but were equivalent to other helmets that did produce low kinematic results. This result would suggest that the metrics affecting the maximum principal strain variable are not necessarily just peak magnitudes of linear and rotational acceleration, but perhaps other aspects of the linear and rotational acceleration curve such as slope or duration. This has been reported to be likely by the authors for ice hockey impacts [33] and brain injury simulations [34-35] in addition to other research in the literature [36] and seems to be supported by these results.

Finally, helmet H was the helmet reported to have been worn by the player at the time of the collision. As indicated previously, this helmet presented in the high risk category based on all response metrics resulting from the event reconstruction. This may explain the long-term symptoms the ice hockey player experienced as a result of this impact, causing loss of game time for approximately 55 days.

The results indicate that the shell geometry and liner designs that are unique to ice hockey helmets do have an influence on the resulting head motion from the impact. This is represented by the significantly different responses between the certified helmets for all metrics used in this research, suggesting that future improvements in helmet design will affect the risk of brain injury in ice hockey. These results are however specific to the particular case and impact site and the same helmets may not produce the lowest magnitudes if the impact was a different location on the head. It is likely, however, that there would still be differences between these helmets at other locations based on differences in shell geometry and liner configuration and type.

When the results of these impacts are examined from a biological standpoint, while the magnitudes of response between helmets are significantly different, they are not necessarily sufficiently different in magnitude to be considered to have had an effect on the resulting concussive injury. This is significant as these helmets represent a large cross-section of the currently available helmets for ice hockey, and none would have protected this player from concussive injury for this impact. Expanding on this point, while this research represents only one concussive impact it would be reasonable to assume that different helmet models offer essentially similar protection for this type of injury. The magnitudes of response reported by this research support the notion that helmet designers have yet to create dramatically safer helmets. It is clear that shell design, material and liner thickness contribute greatly to the safety of a helmet. The varied designs examined in this research that reflect improvement in magnitudes of response linked to concussion reflect an opportunity to optimize ice hockey helmets to improve player safety.

V. CONCLUSIONS

Overall, the ice hockey concussion reconstruction conducted in the laboratory produced results that were consistent with a concussion, confirmed by the amount of time that the player was suffering symptoms keeping him out of the game (55 days). The results of this ice hockey concussion reconstruction demonstrate that the shell and liner design of ice hockey helmets influence the motion of the head from an impact, confirming that the design of the ice hockey helmet worn does influence the risk of injury from a head impact. Notably, low responses in linear acceleration do not transfer to low responses in rotational acceleration. When comparing the helmets using maximum principal strain as the metric, there were few differences among their designs that were indicated with the peak kinematic responses, which may be a result of their design being based on linear acceleration. This result indicates that the magnitude of strain may be influenced by linear and rotational acceleration curve parameters and not represented by the peak resultant magnitudes. Finally, while the results were statistically significant between helmet types, they may not necessarily affect the risk of concussion. This reinforces the notion that to truly affect the rates of concussive injury significantly better helmets need to be developed in conjunction with game management, athlete education and rule changes to eliminate impacts to the head.

Limitations

The research conducted in this study is subject to certain limitations. The data collection method has described tolerances for error in the data collection and must be considered when examining the results. In addition, the Hybrid III headform is not a human head; however, it has been shown to produce responses within the range that would be expected for cadavers [37] and has been used in many brain injury reconstruction studies, producing results consistent with anatomical literature. The UCDBTM uses material characteristics that specify linear viscoelasticity for the brain. This is a common practice in current impact brain modeling, even though there are brain tissues that are nonlinear viscoelastic in material definition. Incorporating these nonlinear viscoelastic properties once they are properly defined in finite element brain models would be a necessary future improvement. Adding these parameters for material characteristics would likely change the magnitudes of maximum principal strain in this research, but it is doubtful that the relationships between the helmets or overall results would change significantly. In addition to this limitation, the UCDBTM and all other models are validated to the same cadaveric impact data. Cadaver brain tissue exhibits different responses to that of live tissues, and these differences should be considered when considering the simulation results.

VI. ACKNOWLEDGEMENTS

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