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The application of brain tissue deformation values in assessing the safety performance of ice hockey helmets

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Abstract

Dynamic impact responses and brain tissue deformations from helmeted centric and non-centric headform impacts were assessed with respect to suggested concussive injury thresholds from the literature. Results from six commercially available ice hockey helmets were compared statistically. It is proposed that the current centric impact standards for ice hockey helmets measuring linear acceleration have effectively eliminated traumatic head injuries in the sport, but that angular acceleration and brain tissue deformation metrics are more sensitive to the conditions associated with concussive injuries which continue to be a major injury in the sport.

Keywords

Brain injury, Ice hockey, Concussion, Finite element modeling, Injury reconstruction, Helmet testing

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INTRODUCTION

Ice hockey is characterized by high skating velocities (25-45 km/h) with athletes who are big, strong and aggressive^{1,2}. The incidence of concussions (mTBI) has remained the same over the past 10 years and in some cases increased³. They represent 18% of all reported ice hockey injuries⁴ and often require time away from the game⁵. Ice hockey helmets are primarily designed to mitigate traumatic brain injuries (TBI) generally referred to as severe head injuries⁶. While this resulted in a dramatic decrease in traumatic brain injuries, minor traumatic brain injuries did not decrease in frequency or severity^{5,7,8}. Safety standards typically require the helmet to limit peak linear acceleration during an impact below a specific number. Unfortunately even though rotational accelerations have been identified as an important predictor of risk for concussive injuries no standard to date includes rotational accelerations as part of their test protocol^{9,10,11}. The majority of the impacts causing concussion in ice hockey involve impacts between two players^{5,7,12}. This

type of impact is not reflected in the present drop test used to measure the safety of ice hockey helmets.

One possible explanation for this continued high incidence of concussion is the method in which helmets are designed and tested. Currently, helmets are required to adhere to certification standards which use peak resultant linear acceleration as the dependent variable measuring protective performance². This kinematic variable was chosen as an indicator of injury due to cadaveric research linking damaging intracranial pressure and skull fracture to high levels of linear acceleration. As a result, a peak value of 275 g is commonly used as the pass/fail limit for the safety certification of hockey helmets¹³. The research which led to the use of this pass/fail variable was concerned with the prevention of traumatic brain injury and as a result TBI has largely disappeared from the sport of ice hockey¹³. The use of linear acceleration is associated with traumatic brain injury; however it is not fully descriptive of all brain injuries, such as concussion.

An impact to the head can be characterized by the resulting linear and rotational components of the resulting motion. While linear accelerations can cause damaging pressure gradients and skull fracture, concussion has been described as a rotationally dominant injury, and such rotations are not reflected by linear acceleration measurement³. Rotationally induced injury was first theorized by Holbourn¹⁴ and has since been refined.

Rotation of the head is thought to cause the brain to rotate within the skull which can cause focal point stresses and strains to tissue^{15,16}, and diffuse shearing of the brain matter. This diffuse shearing is accentuated in locations of the brain where materials of different densities interact^{17,18}. This diffuse shearing of brain tissue is thought to be the main mechanism of injury for concussion, which was further confirmed by work of Gennarelli et al.¹⁹ who induced concussion in monkeys without any linear acceleration. Currently no certification standard includes both linear and rotational accelerations to evaluate sport helmet performance², as a result helmets are not created with the intention of reducing this type of impact induced motion.

Previous research measuring the ability of ice hockey helmets to manage linear and rotational acceleration have reported that while they may be similar for linear impacts, they show differences in the rotational response⁴. These differences are also present in centric and non-centric testing protocols of American football helmets⁵. Attempts to predict brain injuries using peak linear and angular accelerations have not been particularly successful. Although the result of any impact to the head can be quantified using linear and rotational acceleration, the influence of these motions on damage to brain tissue has yet to be elucidated. As a result of this need to understand the influence of post impact head

kinematics on brain tissue deformation and injury advanced computational models have been developed.

Finite element modelling of the brain during an impact provides an opportunity to study the influence of complex loading curves on the brain tissue. This allows for the characteristics of the loading curve to be used to predict brain deformations, which have a higher significance in predicting nervous system tissue injury⁶. In the past researchers have shown how this method can predict how linear and rotational accelerations influence the stresses and strains imparted to the brain from football and hockey helmet impacts. This method allows for the measurement of not only linear and rotational acceleration, but also how they interact to create brain injuries. As a result, the development of finite elements models for the head and brain provide an opportunity to use brain deformation values to assess the ability of helmets to manage the risk of brain injury.

This approach is expected to provide more information describing the impact management characteristics of helmets and identify how divergences between linear and rotational acceleration affect brain tissue with respect to injury.

METHODOLOGY

A pneumatic linear impactor was used to impact certified ice hockey helmets at 7.5 m/s in centric and non-centric conditions (Table 1).

Table 1. Testing impact locations.

	Location	Impact Angle
1.FPE15	Anterior intersection of the mid-sagittal and absolute transverse planes	15° elevation in the mid-sagittal plane towards the impactor
2.FBPA	Midpoint between the anterior mid-sagittal and right coronal planes in absolute transverse plane	45° rotation in the transverse plane
3.SCG	Right intersection of the coronal and absolute transverse planes	No vertical or horizontal rotation was applied to the vector
4.RBNA	Midpoint between the posterior mid-sagittal and right coronal planes in absolute transverse plane	-45° rotation in the transverse plane
5.RNA	Posterior intersection of the mid-sagittal and absolute transverse planes	-45° rotation in the transverse plane

The linear impactor is formed of a table housing a helmeted hybrid III headform and the main frame which holds the impacting arm. The mass of the impacting arm was 16.6 ± 0.1 kg and had a 19.05 mm VN600 foam pad with a hemispherical nylon cap affixed to the end (Figure 1).

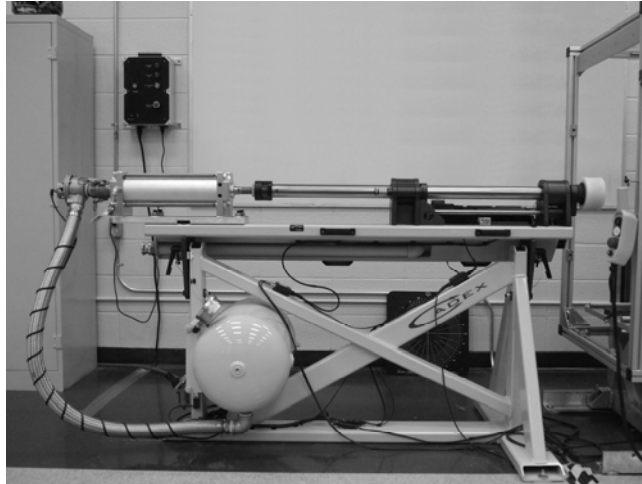


Figure 1. Linear impactor main frame and impacting arm

The impacting arm was propelled by a compressed air mechanism, and the impact velocity was measured by a time gate just prior to impact. The Hybrid III was installed on a sliding device on the table part of the impactor to allow movement of the system post impact (Figure 2). A spring loaded braking system was used to stop the Hybrid III table after the impact.

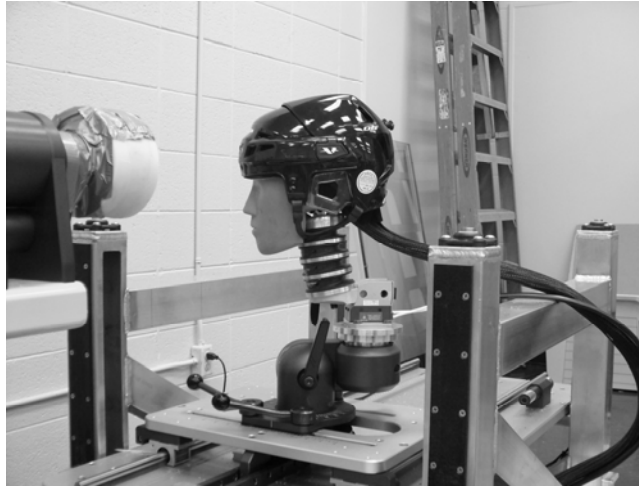


Figure 2. Instrumented Hybrid III headform and sliding table

A 50th percentile male hybrid III headform (mass $4.54 \pm 0.01\text{kg}$) was equipped with accelerometers in a 3-2-2-2 arrangement for measurement of three-dimensional kinematics²⁰. It was attached to the sliding table by a 50th percentile modified Hybrid III neck and lower neck load cell. A solid steel connection replaced the rubber nodding joint in the neck form in order to decrease the measurement variance created when using a rubber nodding joint. The accelerometers used in the Hybrid III were Endevco 7264C-2KTZ-2-300.

Six different ice hockey helmet models were tested under the impact conditions, three impacts per helmet per condition. The liners of the helmets were vinyl nitrile, expanded polypropylene or engineered structure. The accelerometers were sampled at 20 kHz with a

20 ms data collection which was triggered when the curves passed 3 g. The data was collected by a TDAS Pro Lab system (Diversified Technical Systems) and processed by TDAS software. The raw data was filtered with a low pass butterworth filter at 1000 Hz as per the SAE J211 convention. The resulting three dimensional loadig curve responses (x,y and z) were applied to the centre of gravity of the University College Dublin (UCDBTM) finite element model to produce measurements of the brain deformations. The x-axis is defined as facing forward from the head CG, the y-axis to the left of the head and the z-axis vertically upwards. The deformation metrics chosen for this study were selected from previous anatomical and reconstructive research showing correlations to brain injury. As a result of this work, Von Mises stress (VMS) and maximum principal strain (MPS) were selected as variables to measure brain deformation^{11,21,22,23}.

The University College Dublin Brain Trauma Model (UCDBTM) was used to complete the finite element model of the impact deformations of the human brain^{24,25}. The geometry of the model was derived from CT scans of a male participant. The head was comprised of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem. The values describing the material properties of the various brain parts were taken from the literature^{26,27,28,29}. A linearly viscoelastic material model was used to model the brain tissue, and the compressive of the brain was considered elastic.

The brain skull interface was modelled using a sliding boundary condition between the skull, CSF and brain with no space between the cerebrospinal fluid and the pia. A coefficient of friction of 0.2 was used for the sliding surfaces³⁰. The UCDBTM was comprised of a total of approximately 26 000 elements.

Table 2. UCDBTM material properties

Material	Young's modulus (Mpa)	Poisson's ratio	Density (kg/m ³)
Scalp	16.7	0.42	1000
Cortical Bone	15000	0.22	2000
Trabecular Bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx	31.5	0.45	1140
Tentorium	31.5	0.45	1140
CSF	-	0.5	1000
Grey Matter	30	0.49	1060
White Matter	37.5	0.49	1060

The model was validated against intracranial pressure response and brain motion response from cadaver research^{31,32,33}. Further validations were conducted on real life reconstructions with good agreement to clinical data²⁴.

RESULTS

When the helmets were compared across the five impact conditions using peak linear acceleration all helmets produced at least one impact over the 50% risk of concussion (82 g) in total 11 impact conditions out of 30 were above the 50% level. However none of the

helmets reached the 80% risk for a concussion (106 g) as estimated by Zhang and co-workers¹¹ (Table 3). However when peak angular acceleration was included all six tested helmets reported at least one impact condition resulting in a value greater than the 50% risk for concussion¹¹ (5 900 rad·s⁻²) (Table 4). Twenty of the thirty impact conditions resulted in peak angular acceleration values higher than the reported 50% risk. Eight of the 30 impact conditions tested for the six helmets resulted in a mean over the 80% risk of concussion¹¹ (7 900 rad·s⁻²) (Table 4).

Table 3. Mean peak linear accelerations (g) for six ice hockey helmet and five impact sites.

Site	Helmet					
	A	B	C	D	E	F
1.FPE15	74.1 (11.6)	84.5 (8.2)	76.9 (3.8)	76.3 (4.5)	81.8 (2.7)	79.2 (10.0)
2.FBPA	91.6 (8.5)	87.4 (13.7)	82.0 (4.4)	80.8 (4.8)	94.2 (3.5)	88.2 (6.0)
3.SCG	91.0 (7.2)	87.9 (5.0)	80.3 (3.5)	84.6 (5.0)	85.8 (1.4)	90.4 (1.8)
4.RBNA	66.7 (4.2)	65.7 (2.8)	56.7 (1.9)	64.9 (2.7)	67.2 (3.1)	71.4 (2.1)
5.RNA	67.7 (5.4)	78.6 (4.5)	64.1 (1.7)	75.5 (6.6)	72.0 (3.6)	73.4 (3.9)

Table 4. Mean peak angular accelerations (rad·s⁻²) for six ice hockey helmets across five impact sites.

Site	Helmet					
	A	B	C	D	E	F
1.FPE15	4954 (596)	5569 (591)	4912 (351)	5856 (385)	5102 (428)	5603 (828)
2.FBPA	7509 (1928)	7015 (1447)	5065 (399)	5776 (238)	5619 (525)	5830 (450)
3.SCG	8488 (963)	7501 (375)	6387 (309)	6727 (338)	6392 (380)	8430 (394)
4.RBNA	8059 (1041)	8569 (585)	7672 (423)	8560 (538)	8428 (399)	9857 (334)
5.RNA	5802 (1060)	7102 (557)	5314 (447)	7908 (813)	6224 (450)	7229 (657)

Maximum principle strain (MPS) also resulted in all helmets with at least one impact condition resulting in a value above the estimated 50% risk of injury¹¹ (0.225) (Table 5). Seventeen out of the 30 impact conditions resulted in values above the 50% risk of injury using MPS (0.225) (Table 5). When 80% risk of concussion¹¹ was used (0.244) 12 out of 29 of the impact conditions resulted in values higher than this threshold.

Table 5. Mean peak maximum principal strains for six ice hockey helmets across five impact sites

Site	Helmet					
	A	B	C	D	E	F
1.FPE15	0.184 (0.017)	0.186 (0.012)	0.169 (0.010)	0.182 (0.006)	0.175 (0.009)	0.167 (0.009)
2.FBPA	0.226 (0.009)	0.219 (0.007)	0.220 (0.004)	0.268 (0.013)	0.234 (0.007)	0.254 (0.006)
3.SCG	0.195 (0.013)	0.228 (0.007)	0.189 (0.006)	0.209 (0.006)	0.204 (0.015)	0.195 (0.002)
4.RBNA	0.302 (NA)	0.356 (0.012)	0.317 (0.008)	0.328 (NA)	0.330 (0.002)	0.323 (0.001)
5.RNA	0.268 (0.007)	0.331 (0.009)	0.237 (0.008)	NA	0.306 (0.014)	0.316 (0.004)

When the values for Von Mises stress (VMS) (kPa) were considered all helmets produced values greater than the estimated 50% risk of injury (8.4 kPa) in 25 of the 29 tested sites²³

(Table 6). Unfortunately a value associated with a risk of 80% for Von Mises stress does not exist in the literature.

Table 6. Mean peak von Mises stress (kPa) for six ice hockey helmets across five impact sites.

Site	Helmet					
	A	B	C	D	E	F
1.FPE15	8.03 (0.62)	8.48 (0.16)	7.86 (0.11)	8.23 (0.62)	7.86 (0.46)	7.34 (0.40)
2.FBPA	10.28 (0.51)	9.88 (0.38)	10.02 (0.20)	12.54 (0.77)	10.97 (0.39)	11.84 (0.39)
3.SCG	8.85 (0.24)	10.49 (0.41)	8.62 (0.28)	9.58 (0.29)	8.98 (0.43)	9.23 (0.12)
4.RBNA	14.50 (NA)	17.70 (0.82)	15.06 (0.48)	15.61 (NA)	16.15 (0.09)	15.64 (0.17)
5.RNA	11.83 (0.34)	15.78 (0.55)	11.08 (0.51)	NA	13.78 (0.77)	14.98 (0.35)

Table 7. Dynamic impact response and brain tissue deformation comparison between six ice hockey helmets for all locations.

	Dynamic Impact Response		Brain Tissue Deformation	
	Linear (g)	Rotational (rad/s ²)	MPS	VMS (kPa)
A	78.2 ± 13.4	6 962 ± 1 787	0.227 ± 0.050	10.42 ± 2.57
B	80.8 ± 11.2	7 151 ± 1 241	0.247 ± 0.075	10.98 ± 2.47
C	72.0 ± 10.5	5 869 ± 1 113	0.224 ± 0.057	10.39 ± 2.82
D	76.4 ± 8.1	6 965 ± 1 218	0.247 ± 0.065	11.49 ± 3.28
E	80.2 ± 10.2	6 352 ± 1 220	0.236 ± 0.066	10.99 ± 3.42
F	80.7 ± 9.4	7 393 ± 1 722	0.235 ± 0.070	11.01 ± 3.58
Mean	78.1 ± 10.9	6780 ± 1487	0.236 ± 0.098	10.88 ± 0.42

The overall performance of the helmets as represented by both the dynamic response and brain tissue deformation measures is presented in table 7. The brain deformation metrics resulted in mean values in a range of 10.39 to 11.49 kPa for Von Mises stress (Table 7) and 0.224 to 0.247 for maximum principal strain (Table 7).

Each dependent variable was analyzed for significance for each impact condition. The following are the results for the dynamic response data. Peak linear acceleration revealed no significant differences for the FPE15 condition; helmet D performed significantly better than the other four helmets and helmets A and E the worse for FBPA; helmet C performed significantly better than the other helmets and helmet A significantly worse for SCG; helmet C performed significantly better and helmet F significantly worse than the other helmets for RBNA and finally helmet C significantly better followed by helmet A with helmets E and F next, followed by D and lastly helmet B at condition RNA. For peak angular acceleration helmets C and A performed the best followed by helmets E, B and F and finally helmet D the lowest for FPE15; helmet C performed the best followed by E, D, F and B and A the lowest for FBPA; helmets C, E and D performed the best followed by helmet B and finally helmets F and A for condition SCG; Helmet C performed the best with no significant difference for the other five helmets for condition RBNA and finally helmets C and A performed significantly better, followed by helmets E, B and F followed by helmet D for condition RNA.

For brain tissue deformation data the following results were found. For Von Mises stress the impact condition, FPE15 did not result in significant ($p < 0.05$) differences between the six helmet types. FBPA identified helmets B, C, A, and E as performing better

($p < 0.05$) than helmet F which in turn performed better than helmet D. For impact condition SCG helmet C performed better ($p < 0.05$) than all other helmets and helmet B significantly worse. Finally for impact condition RNA helmets C and A performed significantly better than helmets E and F with helmet B performing the worse. Using maximum principle strain values for the FPE15 impact condition did not distinguish between the six helmets tested, FBPA identified four helmets (B, C, A, E) better ($p < 0.05$) than F and helmet D significantly worse than the other five, the SCG condition identified helmets C, F, A all significantly better than helmets E and D which in turn was significantly better than helmet B finally for RNA condition helmet C performed the best followed by helmet A with helmets E, F and B performing the worst. Impact condition RBNA produced some of the highest tissue deformation values but unfortunately not all of the loading curves could be analyzed and therefore the significance of the mean differences was not reported.

When comparing the helmets on linear acceleration alone, only helmet C performed better than the other models, while the rest were equivalent. This pattern is seen throughout the kinematic dependent variables, with helmet C producing significantly lower values than the majority of the other helmet models. The only helmet to perform similar rotationally was helmet E (5 869 $\text{rad}\cdot\text{s}^{-2}$ to 6 352 $\text{rad}\cdot\text{s}^{-2}$).

When broken down by impact site the helmets performed differently depending on the impact location. Helmet C consistently outperformed the others across impact condition for dynamic response and helmets C and A performed the best using brain tissue values. Helmets B and F tended to result in the highest values for dynamic response with helmets B and D having the highest values for brain tissue deformation.

DISCUSSION

Overall, there was not a great deal of difference among the helmets when analysed using linear acceleration. This was expected as peak linear acceleration is the parameter around which ice hockey helmets are tested and designed, and as such are similarly protective. When other parameters are added to the analysis, such as brain deformation metrics, there are more differences discovered between the helmets. This added sensitivity to the structure and design of the helmets is likely a result of both the impacting protocol and the added parameters which are used in a finite element modelling analysis. The protocol used in this study was designed to evaluate helmets under linear dominant (centric) and rotationally dominant (non-centric) impacts. As a result, the protocol created situations where the linear and rotational accelerations diverge. In some sites the dynamic response of the helmeted head form resulted in relatively low linear accelerations with correspondingly high rotational accelerations. While notable, this divergence becomes very

important when examining how the dynamic response influences the brain tissue stresses and strains. As has been shown in the literature, linearly dominant motion is more likely to cause high intracranial pressures and focal injury, while rotationally dominant motions are more likely to cause diffuse shear strains of brain tissue such as those incurred in concussive injuries¹⁴. It has also been shown through finite element modelling research that when there are combinations of both linear and rotational acceleration loading curves the resulting risk of injury is often more severe than either one in isolation. This speaks to the necessity of having a protocol that examines the performance of protective devices across a range of impacts designed to produce both linear and rotational acceleration. The use of finite element modelling in conjunction with such a protocol provides an opportunity to observe how the dynamic response produced from these impacts influence brain deformation and ultimately the risk of injury.

When these results are put into a framework of injury risk, interesting relationships result. Research involving the risk of injury has been conducted by various researchers employing methods in anatomical and reconstructive areas of brain tissue damage. From this research, maximum principal strain values above 0.225 and Von Mises Stress above 8.4 kPa has been proposed as possible 50% risk of injury threshold for concussion. Similarly, values above 82 g and 5 900 rad.s⁻² have been suggested to represent a 50 % risk

of concussive injury¹¹. When these values are used to frame the results of the present research, it becomes evident that through this methodology the helmets consistently perform well for linear acceleration, but well above the limits in rotational, MPS and VMS²³. These results identify the added sensitivity that these measures have when evaluating helmet performance. This sensitivity is attained because finite element modelling uses additional components of the dynamic response in creating brain deformation measures as opposed to one peak resultant value of peak linear and/or rotational acceleration. Finite element modelling provides a representation of brain tissue densities and potentially reflects how some tissues may be more sensitive to the direction of loading curves.

As is inherent with any work using finite element simulations of the human head, the conclusions presented here are a result of the specific conditions and material definitions described within the model. The UCDBTM, like any model is an approximation of the human system and therefore the results from each simulation must be considered as approximate. In addition, the use of a metal headform for the impacts does not represent the more compliant and deformable skull of the average human. As a result, the loading curves inputted to the model may be in error.

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