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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

This research was undertaken to examine a new method for assessing the performance of ice hockey helmets. It has been proposed that the current centric impact standards for ice hockey helmets measuring peak 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 common injury. Ice hockey helmets were impacted using both centric and non-centric impact protocols at 7.5 m/s using a linear impactor. Dynamic impact responses and brain tissue deformations from the helmeted centric and non-centric head form impacts were assessed with respect to proposed concussive injury thresholds from the literature. The results of the helmet impacts showed that the method used was sensitive enough to distinguish differences in performance between helmet models. The results have shown that peak linear acceleration yielded low magnitudes of response to an impact, but peak angular acceleration and brain deformation metrics consistently reported higher magnitudes reflecting a high risk for incurring an mTBI.

Keywords

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

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INTRODUCTION

The majority of the impacts causing concussion in ice hockey involve impacts between two players[1,2,3]. 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 with 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 performance[4]. 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 helmets[5]. 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 hockey[5]. The use of linear acceleration is associated with traumatic brain injury and is not fully descriptive of all brain injuries, such as concussion.

An impact to the head can be characterized by the 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 measurement[6]. Rotationally induced injury was first theorized by Holbourn[7] 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[8,9] and diffuse shearing of the brain matter. This diffuse shearing is accentuated in locations of the brain where materials of different densities interact[6,10]. 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[11] who induced concussion in monkeys without any linear acceleration. Currently no certification standard includes both linear and rotational accelerations to evaluate sport helmet performance; 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 response[12]. These differences are also present in centric and non-centric testing protocols of American football helmets[13]. 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 characterized. However, rotational acceleration has been correlated with concussion level brain deformation from laboratory reconstructions using finite element modelling[14,15]. 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 injury[16]. 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 element 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 risk.

METHODOLOGY

A pneumatic linear impactor was used to impact certified ice hockey helmets at 7.5 m/s[17] in centric and non-centric conditions (Table 1; Figure 1). Of the impact conditions, only site 3 is centric, while the remainders are non-centric in nature. The temperature of the laboratory was 22°C ± 2.0°C and relative humidity was below 55%.

Table 1. Testing impact locations.

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

5	Posterior intersection of the mid-sagittal and absolute transverse planes	-45° rotation in the transverse plane

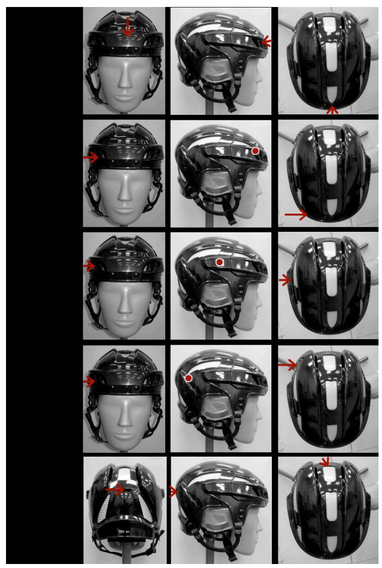

Figure 1. Impact sites on the Hybrid III headform.

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 2).

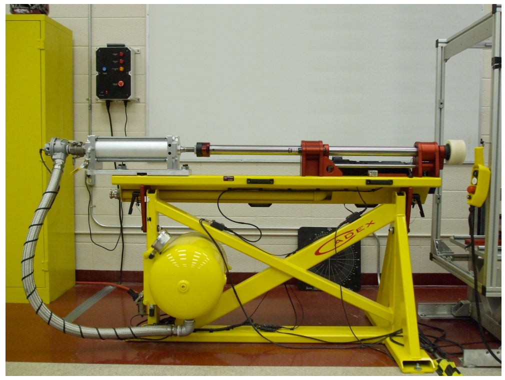

Figure 2. Linear impactor.

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 3). A spring loaded braking system was used to stop the Hybrid III table after the impact.

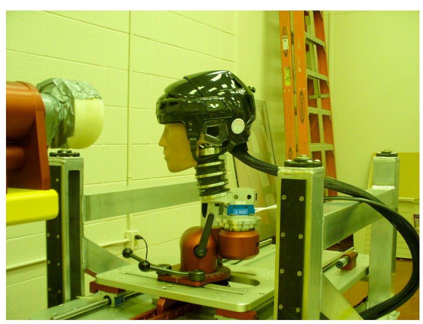

Figure 3. Linear impactor table and helmeted Hybrid III headform.

A 50th percentile male hybrid III headform (mass 4.54 ± 0.01kg) was equipped with accelerometers in a 3-2-2-2 arrangement for measurement of three-dimensional kinematics[18]. 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 (Table 2) were tested under the impact conditions, three impacts per helmet per condition. Six different certified ice hockey helmets were included in this study to determine if the testing protocol and criterion variables were sensitive enough to distinguish between performance differences between the helmets. The companies who participated in this study requested that the results of each model not be reported independently for legal reasons. It is expected that the findings reported in this study will be consistent with similar research comparing different types of helmets in hockey. Hockey helmets are designed for multiple impacts, each helmet was impacted three times per unique test location. This was chosen in accordance with standard Canadian Standards Association[4] test protocols for ice hockey helmet testing. As a result multiple impacts to the same location on an ice hockey helmet produce similar results. The liners of the helmets were vinyl nitrile, expanded polypropylene or a cylindrical engineered structure. Engineered structures are different to foam in that their ability to absorb impact energy is defined by parameters other than density of material, such as wall elasticity and shape of the structure[19]. In this particular case the engineered structure consisted of thermoplastic cylinders arranged to collapse with the impact force absorbing energy. The helmets were installed on the Hybrid III following manufacturers' helmet fitting regulations. A nylon stocking was placed over the headform to allow the helmet to fit more easily.

Table 2. Table of the characteristics of the ice hockey helmets

Helmet	A	B	C	D	E	F
Shell	2PE	1PC	2PE	2PE	2PE	2PE
Liner	3dS	EPP	VN	EPP	EPP	EPP
Mass, kg	0.604	0.318	0.589	0.545	0.567	0.501

2PE: 2 piece polyethylene
1PC: 1 piece polycarbonate
3dS: Three dimensional structure
EPP: Expanded polypropylene
VN: Vinyl nitrile

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 loading curve responses in linear and angular acceleration (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[20,21,22,23].

The University College Dublin Brain Trauma Model (UCDBTM) was the finite element model used to simulate 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[26,27,28,29,30] (Tables 3 and 4). A linearly viscoelastic material model was used to model the brain tissue, the viscoelastic characteristics of which were described for shear as:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor[24]. The hyperelastic material model used for the brain in shear was represented as:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)}$$

where C_{10} and C_{01} are temperature-dependent material parameters, and t is in seconds[31].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 surfaces[32]. The UCDBTM was comprised of a total of approximately 26 000 elements.

Table 3. UCDBTM elastic material properties

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m³)
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 prior cadaver research[33,34,35].

Table 4. Finite element model viscoelastic characteristics for the different regions of brain tissue

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

RESULTS

The dynamic response data of the impacts collected in this research were reported in a previous paper by Walsh et al[36]; sample time histories for each of the five impact sites are found in the appendix (Figures 4 – 8). A finite element simulation for each impact was run using the UCDBTM. The output shows the average brain deformation and the standard deviation of the simulations (Table 5 and 6). For brain tissue deformation data the following results were found. For Von Mises stress the impact condition, site 1 did not result in significant ($p < 0.05$) differences between the six helmet types. Site 2 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 3 helmet C performed better ($p < 0.05$) than all other helmets and helmet B significantly worse. Finally for impact condition 5 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 site 1 impact condition did not distinguish between the six helmets tested, site 2 identified four helmets (B, C, A, E) better ($p < 0.05$) than F and helmet D significantly worse that the other five, the site 3 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 the site 5 condition helmet C performed the best followed by helmet A with helmets E, F and B performing the worst. Impact condition 4 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. The overall performance of the hockey helmets as represented by both the 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). Comparisons between the impact sites with all the helmet data collapsed can be found in table 8.

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

Site	Helmet					
	A	B	C	D	E	F
1	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	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	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	0.302 (NA)	0.356 (0.012)	0.317 (0.008)	0.328 (NA)	0.330 (0.002)	0.323 (0.001)
5	0.268 (0.007)	0.331 (0.009)	0.237 (0.008)	NA	0.306 (0.014)	0.316 (0.004)

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

Site	Helmet					
	A	B	C	D	E	F
1	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	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	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	14.50 (NA)	17.70 (0.82)	15.06 (0.48)	15.61 (NA)	16.15 (0.09)	15.64 (0.17)
5	11.83 (0.34)	15.78 (0.55)	11.08 (0.51)	NA	13.78 (0.77)	14.98 (0.35)

Table 7. Maximum principal strain and Von Mises stress comparison between six ice hockey helmets for all locations. Standard deviation in brackets.

	Brain Tissue Deformation	
	MPS	VMS (kPa)
A	0.227 (0.050)	10.42 (2.57)
B	0.247 (0.075)	10.98 (2.47)
C	0.224 (0.057)	10.39 (2.82)
D	0.247 (0.065)	11.49 (3.28)
E	0.236 (0.066)	10.99 (3.42)
F	0.235 (0.070)	11.01 (3.58)
Mean	0.236 (0.098)	10.88 (0.42)

Table 8. Maximum principal strain and Von Mises stresscomparison between the different impact sites for all helmets. Standard deviation in brackets.

	Brain Tissue Deformation	
Site	MPS	VMS (kPa)
1	0.177 (0.008)	8.0 (0.39)
2	0.237 (0.020)	10.9 (1.07)
3	0.203 (0.014)	9.3 (0.67)
4	0.326 (0.018)	15.8 (1.10)
5	0.292 (0.038)	13.5 (2.01)

When comparing the helmets on linear acceleration alone[36], only helmet C performed better than the other models, while the rest were equivalent and all passed the current standard[4]. This pattern is seen throughout the dynamic 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 rad·s^{-2} to 6 352 rad·s^{-2})[36].

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[36] with helmets B and D having the highest values for brain tissue deformation.

DISCUSSION

When examining helmet performance using the centric and non-centric impact protocols proposed in this research interesting differences are found. The results shown are a result of the complex interaction between shell design and geometry and energy absorbing material. Overall, when the helmets were ranked for each performance metric, helmet C was found to be the best performer, followed by helmet D. Helmet A was found to have the worst results. Helmet C is a vinyl nitrile hockey helmet and as a result the VN material may have been more effective at handling the rotational component of the impact[12,15]. This reduced the dynamic response of the impact and led to reduced maximum principal strain and Von Mises stress results. Helmet D is an expanded polypropylene helmet and performed similarly to helmet C for linear acceleration response but poorer on the angular acceleration response. Helmet A, which performed the worst, employed a three dimensional structure to absorb energy. The poor performance of this particular helmet may be the result of the structures creating a very stiff response to this type of impact as reflected in the high magnitudes. It is possible helmet A would perform better under a higher impact velocity and lower overall system compliance.

Overall, there was not a great deal of difference among the helmets when analysed using linear acceleration[36]. 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 were 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 additional measurement parameters including 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[36]. While notable, this divergence is very important when examining how the dynamic response influences the brain tissue stresses and strains. As reported 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 injuries[7]. It has also been shown in the 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 suggests 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.

Further analysis of the results was conducted in comparison to mTBI reconstructive thresholds suggested by Zhang and co-workers[20]. There are a number of researchers who have used reconstructive techniques to investigate thresholds of mTBI[22,23]; however since Zhang et al.[20] broke down their analysis into 25%, 50% and 80% risks of injury in their discussion, this was used for the majority of the comparisons. While Zhang et al.[20] used a different FE model of the human brain it may be useful to use their research to frame how the current results compare to the literature in terms of risk of injury. When the helmets were compared across the five impact conditions using peak linear acceleration all helmets produced at least one impact was over the 50% risk of concussion (82 g). In total, the helmets produced values over 50% risk of concussion 7 times for the five impact conditions for linear acceleration. However, none of the helmets reached the 80% risk for a concussion (106 g) as estimated by Zhang and co-workers[20,36]. 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 concussion[20] (5 900 $rad \cdot s^{-2}$) [36]. In total, the helmets produced values over 50% risk of concussion 17 times for the five impact conditions for angular acceleration. Five of the 30 impact conditions tested for the six helmets resulted in a mean over the 80% risk of concussion[20] (7 900 $rad \cdot s^{-2}$) [36].

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 injury[20] (0.225) (Table 5). Fourteen 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 concussion[20] was used (0.244) 12 out of 29 of the impact conditions resulted in values higher than this threshold. The variation in the results is a reflection of the variation induced from the impacting protocol and helmet performance over multiple impacts. When the values for Von Mises stress (VMS) (kPa) were considered the helmets produced values greater than the estimated 50% risk of injury (8.4 kPa) in 22 of the 29 tested sites[23] (Table 6). Unfortunately a value associated with a risk of 80% for Von Mises stress was not presented[23].

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 $rad{\cdot}s^{-2}$ have been suggested to represent a 50 % risk of concussive injury[20]. When these values are used to frame the results of the present research, it becomes evident that using this methodology the helmets consistently perform well for linear acceleration, but they exceed these limits in rotational acceleration, MPS and VMS[23]. 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 finite element model is an approximation of the human system and therefore the results from each simulation must be considered as approximate. The viscoelastic material model used in this study may not be ideal for describing the behaviour of the brain tissue under these impact conditions. 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 input into the model may be in error. Also, using the mTBI threshold data from Zhang et al.[20] may be inappropriate as the sample sizes for their study were small. The impact procedure also has some inherent limitations. The Hybrid III headform used was designed form antero-posterior impacts and not for this kind of centric non-centric impact procedure. As a result the neck and headform will have an effect on the resulting dynamic response. The Hybrid III headform is metal and as a result the curves used as input for the model come from a fully rigid system which is not biofidelic. The Hybrid III neckform is also considered to be stiffer than a human neck in response to impact. Therefore the results presented here may be conservative in nature. The testing procedure on the helmets will produce some variation as after each impact the helmet is likely to move. Although care was taken to maintain proper helmet positioning, it is possible this shifting may add to some variance in the results. Also, although the helmets are multi-impact helmets there will be some minimal change in performance from one hit to the next which is reflected in the standard deviations.

CONCLUSION

This research involved impacting a series of commercially available ice hockey helmet at an impact velocity and compliance that reflect the hazards of the sport. The results show that ice hockey helmets perform similarly when evaluated using linear acceleration. However when rotational acceleration and brain deformation metrics are used to refine the evaluation of the performance of the ice hockey helmets additional differences are found.

These differences are likely a result of unique design characteristics of each helmet, such as geometry and impact liner. This methodology indicates that peak linear acceleration may not be sensitive enough to sufficiently differentiate between these unique design characteristics and guide helmet design with regards to decreasing concussive injury risk.

When the results are examined in context of current injury threshold literature, it was found that linear acceleration consistently reported approximately 50% risk of concussion. However rotational acceleration and the brain deformation metrics used in this research revealed higher risks of injury for the same impacts. As these measures have been suggested as a better predictor of concussive injuries this research shows that a methodology should be considered to evaluate ice hockey helmets using rotational acceleration and brain deformation metrics.

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APPENDIX

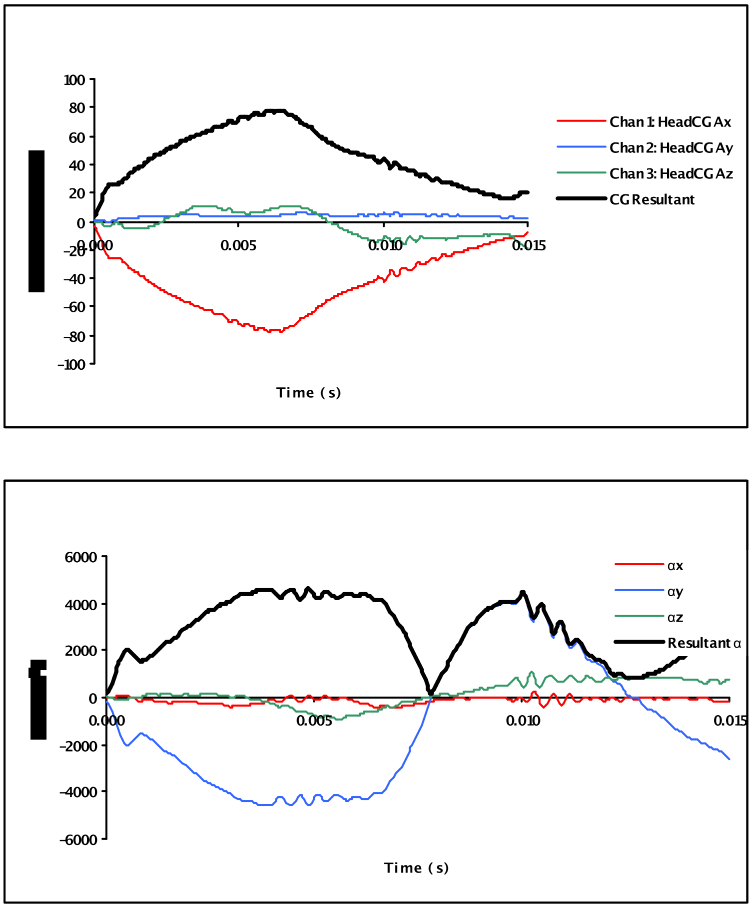


Figure 4. Example linear (top) and angular (bottom) loading curves for the site 1.

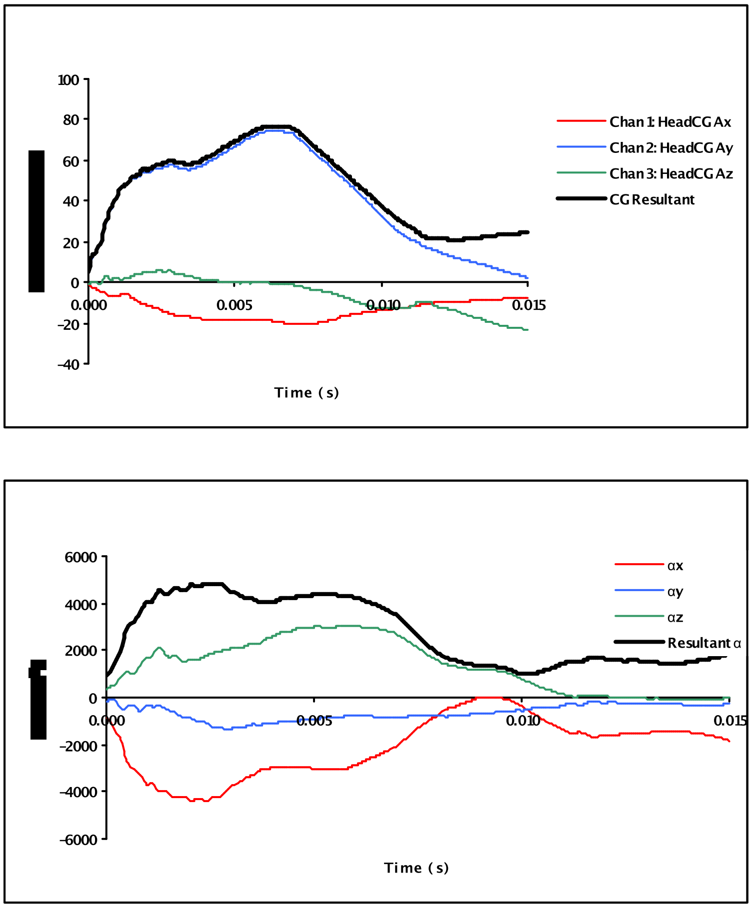


Figure 5. Example linear (top) and angular (bottom) loading curves for the site 2.

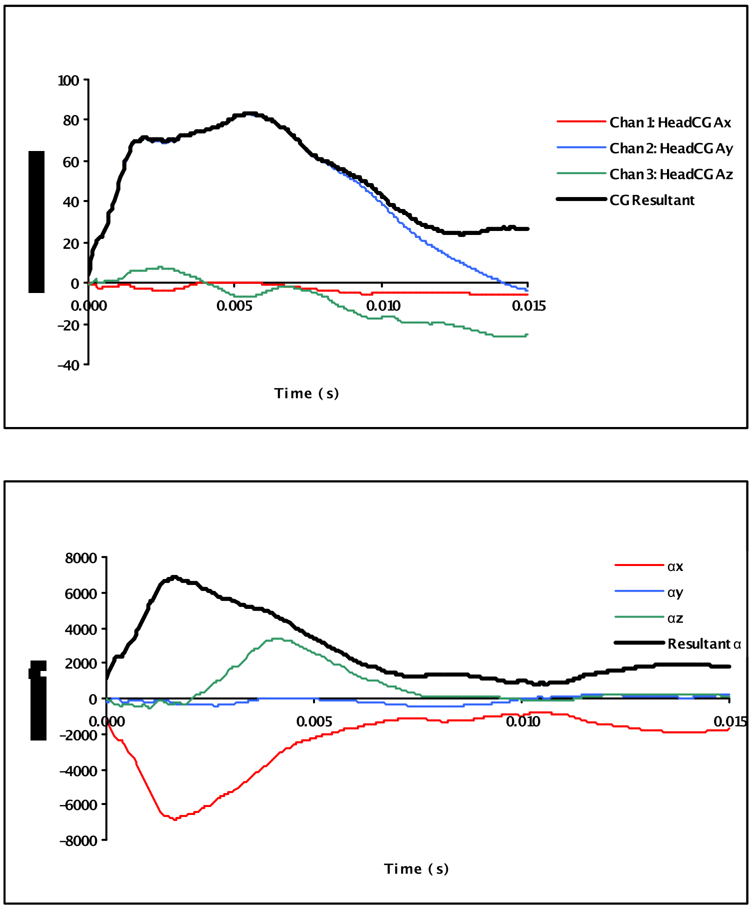
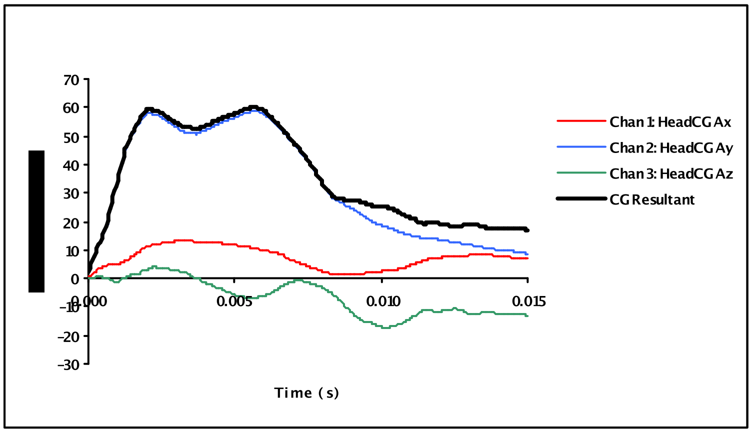


Figure 6. Example linear (top) and angular (bottom) loading curves for the site 3.



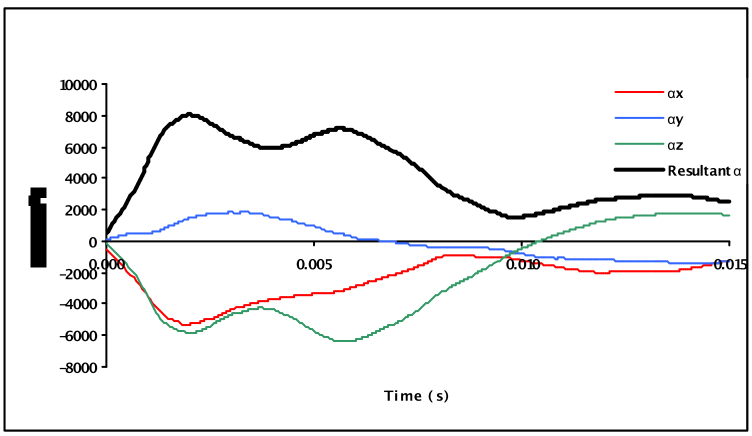


Figure 7. Example linear (top) and angular (bottom) loading curves for the site 4.

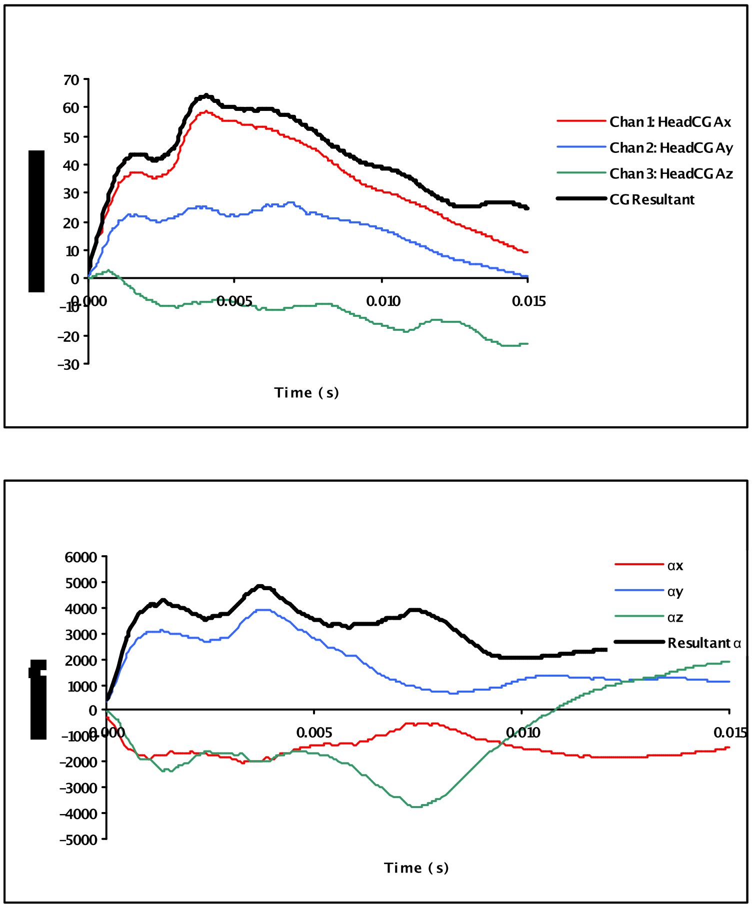


Figure 8. Example linear (top) and angular (bottom) loading curves for the site 5.