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## **Examination of the relationship of peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts**

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# **Examination of the relationship of peak linear and angular acceleration to brain deformation metrics in hockey helmet impacts**

Ice hockey is a contact sport which has a high incidence of brain injury. The current methods of evaluating protective devices use peak resultant linear acceleration as their pass/fail criteria which are not fully representative of brain injuries as a whole. The purpose of this study is to examine how the linear and angular acceleration loading curves from a helmeted impact influence currently used brain deformation injury metrics. A helmeted Hybrid III headform was impacted in 5 centric and non-centric impact sites to elicit linear and angular acceleration responses. These responses were examined through use of a brain model. The results indicated that when the helmet is examined using peak resultant linear acceleration alone they are similar and protective, but when a 3 dimensional brain deformation response is used to examine the helmets there are risks of brain injury with lower linear accelerations which would pass standard certifications for safety.

Keywords: neurotrauma; impact biomechanics; concussion; ice hockey

## **1. Introduction**

Hockey is a contact sport where collisions and injuries are a common occurrence (Wennberg and Tator 2003). This has led to the development of technologies to prevent injuries from occurring (Hoshizaki and Brien 2004). Larger elbow pads and shin guards have been developed to prevent fractures and helmets are worn to prevent head injuries. Of the injuries which can occur in hockey, head injuries are the most severe as they are associated with a high degree of neurological dysfunction from concussion and, at worst, death. Current research into concussive injuries in particular has shown there may be links between mild traumatic brain injury (mTBI) and long term health problems such as dementia and emotional distress leading to suicide (Gasquoine 1997; Guskiewicz et al. 2005). Hockey helmet design is unique to the sport, with a relatively small helmet being used to aid in the dispersion of energy upon impact. The liner of the

helmet is the component which absorbs the majority of energy during an impact and thus a great deal of research has gone into perfecting these energy absorbing liners to protect against head injuries in hockey (Hoshizaki and Brien 2004; Gimbel and Hoshizaki 2008a, 2008b). Hockey helmets are traditionally composed of two primary types of liner, a vinyl nitrile liner (VN) or an expanded polypropylene liner (EPP). Both these liners are designed for multiple impacts and a range of energies and must meet rigorous test standards. The standards which manage the protective capacity of helmets use peak linear acceleration (g) as the threshold dependent variable (Hoshizaki and Brien 2004; Canadian Standards Association 2009). The pass/fail criterion for ice hockey helmet standards is commonly between 250 and 275 g, which is a value based upon traumatic brain injuries (TBI) and skull fractures (Thomas et al. 1966; Yoganandan and Pintar 2004). As such, skull fractures and other TBIs have largely disappeared from sport, however the incidence of mTBI or concussion remains common (Wennberg and Tator 2003). The long term neurological effects of concussions in sport have recently been identified (Forero et al. 2010; McKee et al. 2010). This research has provided impetus to the sports industry to examine how concussions can be prevented.

While linear acceleration has been the accepted dependent variable to assess the ability of helmets to prevent injury, rotational acceleration can also create brain damage (Holbourn 1943; Hoshizaki and Brien 2004; Gennarelli et al 1983). Rotational acceleration has been linked with diffuse injuries such as concussion and diffuse axonal injury and may be the reason why mTBI is still prevalent in hockey while focal injuries, which are associated with linear accelerations of the head, have largely disappeared. However while there are purely linear and rotational acceleration theories to brain injury, in sport the resulting injurious brain deformation is likely influenced and exacerbated by both types of motion (Ommaya et al 1967; Gurdjian and Gurdjian 1975).

Researchers have attempted to assess the performance of hockey helmets for the attenuation of linear and rotational acceleration (Rousseau et al. 2009a, 2009b). From this research it was discovered that VN and EPP foams perform similarly to reduce linear translations but differently when it comes to managing the rotational component of an impact. As linear and angular acceleration both contribute to brain injury, the influence of these variables on brain tissue deformation has yet to be quantified.

With the development of finite element modeling, a new tool is now available for the assessment of the performance of helmet materials in relation to brain injury (Zhang et al. 2001; King et al. 2003). This technique allows for the effects of direction and curve shape of the linear and rotational acceleration loading curves to be represented in dependent variables which are measurements of deformation in specific regions of the brain. Since peak linear and angular acceleration is utilized to describe the motion of the head causing injurious brain deformation, finite element models allow for the measurement of the deformation of the brain in response to these movements (King et al. 2003). Some research to correlate injury to brain deformation parameters has been attempted through injury event reconstructions and cadaver research (Schreiber et al. 1997; Willinger and Baumgartner 2003; Zhang et al. 2004; Kleiven 2007). In this research it has been discovered that Von Mises stress and maximum principal strain have the highest correlation to brain injury, higher than using peak linear or peak angular acceleration alone.

Correlative research using finite element modeling has attempted to link linear or angular acceleration to brain deformation injury metrics such as maximum principal strain and Von Mises stress. Ueno and Melvin (1995) found that linear acceleration was highly influential in the creation of strains while angular acceleration was highly correlated to shear strains. Contrary to these findings, Forero Rueda et al. (2010) found

low correlations for linear acceleration to brain stress and strain (0.2 and 0.52) and higher correlations to angular acceleration (0.72). If reliable and valid correlations could be established between peak resultant kinematic variables and brain deformation response it would allow for the use of properly instrumented head forms to represent the behaviour of brain tissue.

The purpose of this study is to use a combination of physical and finite element models of the human head to examine the relationships between peak linear and angular acceleration and brain deformation that occur in ice hockey head impacts.

## **2. Methodology**

### ***2.1 Equipment***

A pneumatic linear impactor was used to impact the helmeted headform and consists of a frame, impacting arm and a sliding table. The frame houses the compressed air tank and the piston (figure 1). The impacting arm had a mass of  $16.6 \pm 0.1$  kg and was propelled horizontally by the compressed air. The tip of the impactor had a cap consisting of a  $0.677 \pm 0.001$  kg hemispherical nylon pad covering a  $35.71 \pm 0.01$  mm thick modular elastomer programmer (M.E.P.) disc. The sliding table had a mass of  $12.78 \pm 0.01$  kg and housed the Hybrid III headform. The headform had a mass of  $4.54 \pm 0.01$  kg, and the Hybrid III neck weighed  $1.54 \pm 0.01$  kg for a total mass of  $18.86 \pm 0.01$  kg (table and head/neck forms including all screws and brackets). The sliding table allowed for movement post impact, which is comparable to what may happen in a game impact in ice hockey. A spring loaded brake system provided the stopping mechanism following a displacement of  $0.54 \pm 0.01$  m.

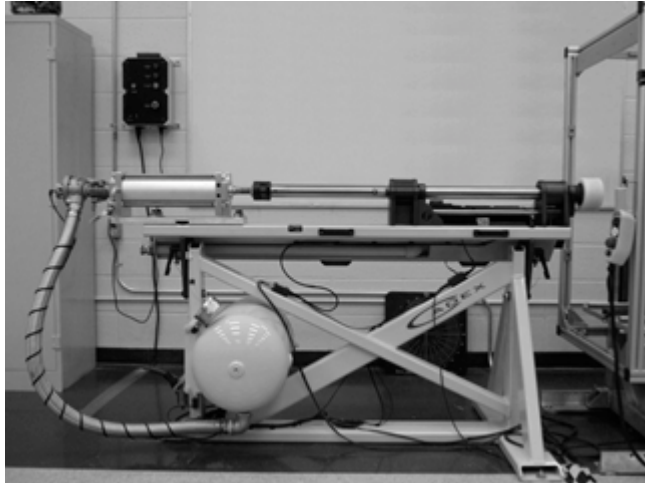


Figure 1. Linear impactor frame

The sliding table supporting the Hybrid III head and neck was built to allow for precision in the location of impact (figure 2). It could be adjusted in five degrees of freedom, including fore-aft (x), lateral (y), and up-down (z) translation, as well as fore-aft (y) and axial (z) rotation of the neck base. The adjustments were lockable and remained fixed throughout the testing.



Figure 2. Linear impactor sliding table and helmeted headform

A 50<sup>th</sup> percentile adult male headform (mass  $4.54 \pm 0.01\text{kg}$ ) and neck was instrumented for measurement of three dimensional kinematics according to

Padgaonkar's 3-2-2-2 accelerometer array (1975) (figure 3). The accelerometers used were Endevco 7264C-2KTZ-2-300.

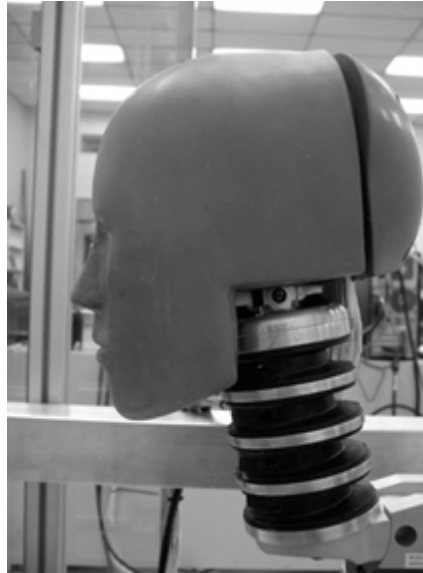


Figure 3. 50<sup>th</sup> percentile Hybrid III headform and neck

Sixteen ice hockey helmets were impacted for this study: 8 had vinyl nitrile (VN) liners and 8 had expanded polypropylene (EPP) liners. The liners both had an offset of ½ inch, and the stiffness of the material was determined through compression testing at 100 mm/min on a universal testing machine as shown in figure 4. They were from the same company and identical models were used to control for shell geometry.

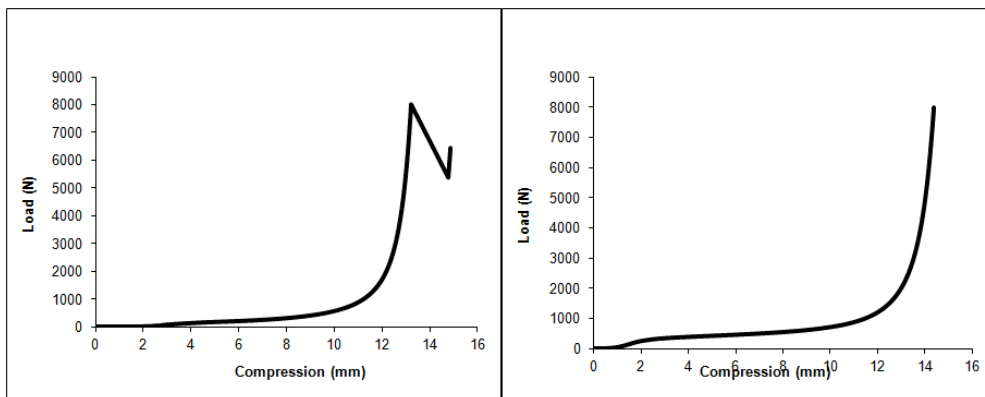


Figure 4. Universal testing machine results of the VN (left) and EPP (right) ice hockey helmet liners

## 2.2 Procedure

To evaluate the protective qualities of the VN and EPP liners each helmet was impacted in five sites designed to elicit both linear and angular accelerations (Table 1). These sites were developed from headform only impacts (Walsh et al. 2011), and are shown in figure 5. The helmets were placed on the headform and impacted once per location at a velocity of 4.5 m/s, in accordance with the CSA monorail drop velocities for helmet certification. The average time between impacts was 7 min  $\pm$  0.45 sec. The impact velocity was measured using a time gate prior to impact. The accelerometers were sampled at 20 kHz and a 15 ms data collection period was triggered when the impact exceeded 3 g. All signals were collected using a Diversified Technical Systems TDAS Pro Lab system and processed by TDAS software. The raw data was filtered using a low pass Butterworth filter at 600 Hz. The resulting response in x, y and z for linear and angular acceleration were then used to drive the finite element model analysis. 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.

Table 1. Testing impact locations.

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

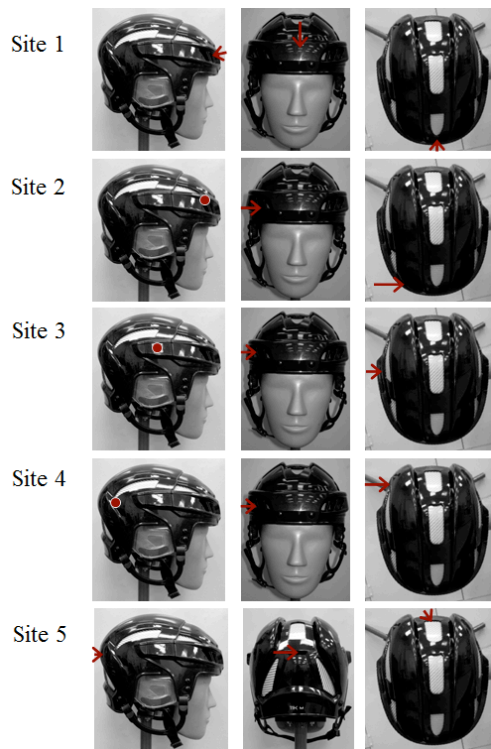


Figure 5. Impact locations and vectors on the helmet as shown by arrows

### ***2.3 Finite element model***

The model used in this research was developed in Dublin and is known as the University College Dublin Brain Trauma Model (UCDBTM). The geometric parameters of the model were from a male cadaver as determined by medical imaging techniques (Horgan and Gilchrist 2003, 2004). The head and brain are comprised of ten parts: the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem (table 2).

Table 2. Finite element model material properties

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m <sup>3</sup> )
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

Validation of the model was accomplished through comparisons against intracranial pressure data from Nahum et al. (1977) cadaver impact tests and brain motion against Hardy et al.'s (2001) research. Further validations accomplished comparing real world brain injury events to the model reconstructions with good agreement.

A linearly viscoelastic material model combined with large deformation theory was chosen to model the brain tissue. The behaviour of this tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus. The compressive behaviour of the brain was considered elastic. The CSF layer was modeled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF and brain was used. The model was comprised of 7,318 hexahedral elements representing the brain, and 2,874 hexahedral elements representing the CSF layer.

### 3. Results

The effects of liner type on maximum principal strain (MPS) and Von Mises stress (VMS) are presented in tables 3 and 4. The means were compared using an ANOVA with a p value of 0.05.

Table 3. Peak linear and angular acceleration and maximum principal strain.

Site	material	Peak Acceleration		Grey Matter	Maximum Principal Strain		
		Linear (g)	Angular (rad/s <sup>2</sup> )		White Matter	Cerebellum	Brain Stem
1	VN	113.6 (12.9)	5614 (863)	0.220 (0.019)	0.189 (0.022)	0.042 (0.004)	0.108 (0.005)
	EPP	105.1 (20.2)	5910 (1233)	0.216 (0.023)	0.164 (0.029)	0.044 (0.005)	0.105 (0.009)
2	VN	100.3 (7.3)	7843 (493)	0.249 (0.022)	0.209 (0.026)	0.09 (0.004)	0.146 (0.007)
	EPP	100.9 (7.9)	9179 (598)	0.289 (0.033)	0.191 (0.017)	0.103 (0.008)	0.148 (0.007)
3	VN	116.8 (7.4)	7888 (569)	0.301 (0.017)	0.151 (0.009)	0.072 (0.006)	0.134 (0.007)
	EPP	128.2 (19.7)	9745 (804)	0.331 (0.025)	0.158 (0.020)	0.095 (0.011)	0.147 (0.013)
4	VN	70.3 (6.7)	8890 (973)	0.328 (0.031)	0.177 (0.022)	0.079 (0.008)	0.123 (0.007)
	EPP	78.5 (3.4)	10410 (890)	0.356 (0.030)	0.178 (0.009)	0.093 (0.007)	0.136 (0.013)
5	VN	68.2 (5.6)	4949 (801)	0.293 (0.035)	0.159 (0.029)	0.069 (0.012)	0.135 (0.008)
	EPP	67.2 (6.3)	6077 (492)	0.305 (0.019)	0.160 (0.007)	0.064 (0.007)	0.138 (0.011)

Table 4. Peak linear and angular acceleration and Von Mises Stress

Site	material	Peak Acceleration		Grey Matter	Von Mises Stress (Pa)		
		Linear (g)	Angular (rad/s <sup>2</sup> )		White Matter	Cerebellum	Brain Stem
1	VN	113.6 (12.9)	5614 (863)	10556 (1087)	9311 (1209)	2101 (109)	4949 (286)
	EPP	105.1 (20.2)	5910 (1233)	10195 (1355)	7917 (1434)	2082 (266)	4908 (477)
2	VN	100.3 (7.3)	7843 (493)	12420 (1111)	9699 (1451)	4326 (256)	6627 (396)
	EPP	100.9 (7.9)	9179 (598)	14550 (1717)	8609 (904)	4941 (360)	6738 (414)
3	VN	116.8 (7.4)	7888 (569)	15108 (1023)	6819 (337)	3325 (315)	6242 (517)
	EPP	128.2 (19.7)	9745 (804)	17107 (1324)	7369 (738)	4454 (478)	6704 (874)
4	VN	70.3 (6.7)	8890 (973)	16338 (1864)	8488 (1326)	3771 (489)	5456 (482)
	EPP	78.5 (3.4)	10410 (890)	18121 (1956)	8318 (712)	4607 (607)	6034 (832)
5	VN	68.2 (5.6)	4949 (801)	14397 (2117)	7522 (1632)	3209 (684)	5648 (356)
	EPP	67.2 (6.3)	6077 (492)	14957 (1294)	7367 (480)	2898 (418)	5641 (641)

#### 3.1 EPP and VN liner comparison

In three sites (2, 3 and 4) the EPP liner produced significantly higher magnitude maximum principal strain and Von Mises Stress when compared with the VN helmet liner in the grey matter ( $p < 0.05$ ). Sites 1, 2 and 5 showed no significant difference

between the performances of the liner types for the grey matter ( $p < 0.05$ ). The values in maximum principal strain and Von Mises Stress were lower for the VN liner for site 3 with the EPP outperforming the VN liner in sites 1 and 2. There was no difference between liners in the remaining sites. The maximum principal strain and Von Mises stress values for the cerebellum and brain stem showed little difference between the two helmet liners ( $p < 0.05$ ). The VN liner produced lower values at the cerebellum for sites 2 and 3 and there were no differences between the liners at the brain stem ( $p < 0.05$ ). In the model of the brain the grey matter had the highest stresses and strains (0.36, 18.6 kPa), while the cerebellum had the lowest (0.043, 2.2 kPa).

### ***3.2 Influence of impact site on kinematic and brain deformation responses***

When examining the effect of impact site on the performance of the helmets, there are significant differences between all sites except site 1 to site 2 and site 4 to site 5 for linear acceleration ( $p < 0.05$ ). For angular acceleration all sites produced significantly different results from each other except for site 1 to site 5, site 2 to site 3, site 3 to site 4 and site 2 to site 4. When examining impact site by MPS, all sites were significantly different from each other except for site 4 to site 3 and site 3 to site 5 ( $p < 0.05$ ). When examining site by VMS, all sites were significantly different except site 3 to site 4, site 3 to site 5, and site 2 to site 5 ( $p < 0.05$ ).

Overall, the impact site with the largest magnitudes in linear acceleration was site 3, producing values between 114 to 128g. The largest magnitudes of angular acceleration were produced by the site 4 impact, producing values between 8890 to 10410  $\text{rad/s}^2$ . When examining brain tissue deformation, the site with the largest MPS and VMS was site 4, producing values ranging from 0.328 to 0.356 and 16 to 18 kPa respectively.

### 3.3 Correlations

Significant correlations were found between peak linear and angular acceleration and maximum principal strain and Von Mises stress. When all data was collapsed (table 5), linear acceleration had a low negative correlation to maximum principal strain (-0.239). Peak angular acceleration had a moderately positive correlation to maximum principal strain and Von Mises stress (0.638, 0.677).

Table 5. Global correlations of linear and angular acceleration to maximum principal strain and Von Mises stress

	Peak Linear Acceleration	Peak Angular Acceleration	Maximum Principal Strain	Von Mises Stress
Peak Linear Acceleration	-	0.182	-0.239*	-0.171
Peak Angular Acceleration	0.182	-	0.638*	0.677*
Maximum Principal Strain	-0.239*	0.638*	-	0.995*
Von Mises Stress	-0.171	0.677*	0.995*	-

\* Correlation is significant at the 0.05 level (2-tailed).

When correlations were broken down to regional peaks per brain tissue type (tables 6 and 7), a significant low negative relationship was found between peak linear acceleration and maximum principal strain in the white matter. Correlations varying from low positive to high positive between angular acceleration and maximum principal strain were found in the grey matter, cerebellum and brain stem (0.638, 0.771, 0.452). Low positive correlations of linear acceleration to Von Mises stress were found in the brain stem (0.346). Correlations varying from moderate to high positive were found for angular acceleration to Von Mises stress for the grey matter, cerebellum and brain stem (0.677, 0.78, 0.557).

Table 6. Brain location specific correlations of linear and angular acceleration to maximum principal strain

	Peak Linear Acceleration	Peak Angular Acceleration	Maximum Principal Strain			
			Grey Matter	White Matter	Cerebellum	Brainstem
Peak Linear Acceleration	-	0.182	-0.239*	0.074	-0.021	0.117
Peak Angular Acceleration	0.182	-	0.638*	0.143	0.771*	0.452*

\* Correlation is significant at the 0.05 level (2-tailed).

Table 7. Brain location specific correlations of linear and angular acceleration to Von Mises stress

	Linear Acceleration	Angular Acceleration	Von Mises Stress			
			Grey Matter	White Matter	Cerebellum	Brainstem
Peak Linear Acceleration	-	0.182	-0.171	0.097	-0.001	0.346*
Peak Angular Acceleration	0.182	-	0.677*	0.089	0.78*	0.557*

\* Correlation is significant at the 0.05 level (2-tailed).

## 4. Discussion

### 4.1 EPP and VN differences

The results indicate that the vinyl nitrile and expanded polypropylene liners produce different responses as measured by the dependent variables depending on the impact location. When the performances of the helmets are examined by linear acceleration alone, the liners differ significantly only in site 4 ( $p < 0.05$ ). However, when performance is analyzed using angular acceleration, there are differences in peak angular acceleration in sites 2, 3, 4 and 5 ( $p < 0.05$ ), with the VN liner producing lower magnitudes for each impact condition. These results indicate that while the helmets may be considered similar in protective capacity by linear acceleration, there are certain characteristics of the liners which make the EPP produce larger peak resultant angular accelerations. When performance is evaluated by brain deformation metrics, the VN liner produced statistically significant lower magnitude stresses and strains than the EPP liner in the grey matter in two sites (2 and 3).

## ***4.2 Site analysis***

The areas of the brain which showed the largest magnitudes of stress and strain were the grey and white matter, with the brain stem and then the cerebellum with the lowest magnitudes respectively. The largest values were found for the non-centric impacts for the grey matter, indicating a combination of linear and higher angular acceleration contributes to a greater risk of injury. These results support Bandak and Eppinger's finite element brain model research indicating that a combination of linear and angular acceleration produced larger brain strains than each in isolation (1994). The white matter values were highest for the side impact. These results indicate that the site of injury may be sensitive to the location of the impact, which supports animal research showing the degree and severity can vary according to loading vector (Gennarelli et al. 1983; Adams et al. 1983).

## ***4.3 Correlations***

The correlations found in this study would seem to suggest that these peak resultant angular accelerations may be more predictive of maximum principal strain and Von Mises stress values than peak linear acceleration. These results are consistent with previous research indicating a correlation of 0.72 for helmet impacts (Forero Rueda et al. 2010). Interesting observations can be found when the results are analyzed by brain tissue type. For example, a high correlation was found in the grey matter and cerebellum for MPS with peak resultant angular acceleration and not for the other two brain regions (white matter and brainstem). This could be further evidence that suggests that the brain tissue response is sensitive to brain tissue type and location. Strength of correlations also varied with the type of acceleration, with linear acceleration having negative correlations to brain deformation metrics and angular acceleration having positive

relationships. This phenomenon can be described from two perspectives, the finite element model and from the testing methodology. From the finite element modelling approach, these correlations suggest that brain tissue has varying responses specific to the type of loading. One possible reason for these differences in response seen across impact site and brain deformation could be attributed to the geometry and material characteristics of each type of brain tissue in the model (Prange et al. 2002). This would be expected due to the different material characteristics for each brain tissue as is defined in the UCDBTM. These properties could also have an effect on the different brain deformation responses seen under various loading conditions. In this study the material property characteristics governing grey matter, white matter, the brainstem, and cerebellum would subject them to variable deformation responses at impact.

These correlations are also influenced by the methodology used. The impact sites chosen for this study were primarily non-centric, meaning the impact vectors reside outside the centre of gravity of the headform. This would result in rotational acceleration dominant responses, which would then have more influence on the development of brain deformation than linear acceleration. This would, in part, explain why there can be a negative correlation for linear acceleration to brain deformation metric, as the impact sites can produce low linear acceleration and high rotational acceleration responses which produce large brain deformations. Similarly, if the headform was impacted through the centre of gravity it would be expected that the linear acceleration would become the dominant response and rotational acceleration less so. This type of impact would likely produce high correlations of linear acceleration to brain deformation and less so for the rotational acceleration.

## 5. Summary and conclusions

The dependent variables used in the study were chosen due to the research in injury reconstruction and anatomical stress testing indicating their correlation to injury. Previous research applying axial stretching to cadavers and animal neuronal tissues have identified that values in maximum principal strain above 0.14 and Von Mises stress of 8.0 kPa leads to mechanical failure (Table 8). Research using three dimensional linear and rotational acceleration responses generated from on-field sport reconstructions and motorcycle accidents resulting in brain injury have identified values for maximum principal strain above 0.19 and values of 7.8 kPa for Von Mises stress as leading to possible brain injury (Table 9).

Table 8. Tolerance values for anatomic tissue testing

Injury threshold value	Tissue type	Method	Reference
20 Mpa stress, 0.2 strain	Cerebral blood vessels	Cerebral cadaver arteries	Chalupnik et al. (1971)
0.51 to 0.55 strain	Parasagittal bridging veins	Cadaver bridging veins	Lee and Hault (1989)
0.19 strain, 6-11 kPa Von Mises	Cortex	Cadaver tissue	Schreiber et al (1997)
8-16 kPa for DAI	CNS tissue	Sheep model	Anderson et al. (1999)
0.14 to 0.34 strain	White matter axons	Male guinea pig optic nerve stretch	Bain and Meaney (2000)
0.31 strain	Cerebral blood vessels	Human cerebral blood vessels from surgery	Monson et al. (2003)

Table 9. Tolerance values for mTBI calculated from injury event reconstructions.

mTBI Threshold value (50% chance)	Dependent variable	Reference
0.21-0.26	Max. Principal Strain	Kleiven (2007)
0.19	Max. Principal Strain	Zhang et al. (2004)
8.4 kPa	Von Mises	Kleiven (2007)
7.8 kPa	Von Mises	Zhang et al. (2004)
18 kPa	Von Mises	Willinger and Baumgartner (2003)

When these values are used for benchmarks against which the VN and EPP hockey helmets can be compared interesting relationships result. Both the grey and white matter indicates a significant chance of brain injury from the impacts using a linear impactor at 5.5 m/s, which is a slow velocity for the game of ice hockey. Also, both helmets perform well when just linear acceleration is examined, with values far below those of the current CSA standard (275 g).

These results could shed light on the current incidences of mTBI in hockey. The helmets used currently in hockey are designed to prevent against TBI and as such such injuries have largely disappeared, however mTBI remains prevalent in the sport (Wennberg and Tator 2003). These results indicate that when the helmet is examined for linear acceleration alone they are similar and protective, but when a 3 dimensional brain deformation response is used to examine the helmets there are possibilities of brain injury with lower linear accelerations which could be considered 'safe'. As a result of this possible relationship it may be prudent to evaluate helmets on the basis of a 3 dimensional kinematic response to an impact instead of just linear acceleration. This type of analysis may aid in the prevention of mTBI in sports.

The correlation results indicate that peak resultant acceleration may be predictive of global peak brain deformation. However the strength of the correlations is influenced by the type of acceleration (linear or angular), region of brain tissue (grey matter, white matter, brainstem, or cerebellum) and the impact site. As a result of the influential nature of these parameters on overall correlations, it could be concluded that it may be inappropriate to use peak resultant acceleration correlations as predictors of brain deformation response. This study supports the use of brain deformation metrics for evaluating helmet performance in eliciting its protective capacity to brain injury rather than only using linear acceleration alone.

## ***5.1 Limitations***

There are some limitations to the approach used in this study. The method of using a rigid physical head form to record three dimensional responses to an impact was chosen to increase the control over impact vector and helmet interfaces. These conditions create a response which has increased error, since the human head is not a fully rigid system. Also the Hybrid III system has only been validated for impacts in the anterior/posterior direction and the results should be interpreted while considering this limitation. The Hybrid III neck has also been shown to influence the response of the headform depending on impact condition as has been shown by Foreman and Hoshizaki (2010) and Rousseau and Hoshizaki (2009c). This neck influence is also a limitation which must be considered when interpreting the results. As such, the results in this study are not meant to replicate a human injury in the game of ice hockey, but rather create a three dimensional response to be collected and the dependent variables compared in a controlled situation. The finite element model that has been used, while partially validated, makes assumptions concerning the behaviour of human brain tissue and their characteristics which would influence the results. Therefore the results and conclusions from the relationships found between the dependents variables in this study would be more definitive if improved constitutive data were used in the finite element models.

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