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# Evaluation of dynamic response and brain deformation metrics for a helmeted and non-helmeted Hybrid III headform using a monorail centric/non-centric protocol

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## **Abstract**

Head injuries and concussion in particular has become a source of interest in the sport of ice hockey. This study proposes a monorail test methodology combined with a finite element method to evaluate ice hockey helmets in a centric/non-centric protocol with performance metrics more closely associated with risk of concussion. Two conditions were tested using the protocol a) helmeted vs no helmet, and b) vinyl nitrile lined hockey helmet vs expanded polypropylene lined hockey helmet. Results indicated that the impact velocities and locations produced distinct responses. Also, the protocol distinguished important design characteristics between the two helmet liner types with the vinyl nitrile lined helmet producing lower strain responses in the cerebrum. Furthermore, it was discovered that low risk of injury peak linear and rotational acceleration values can combine to produce much higher risks of injury when using brain deformation metrics. In conclusion, the use of finite element modeling of the human brain along with a centric/non-centric protocol provides an opportunity for researchers and helmet developers to observe how the dynamic response produced from these impacts influence brain tissue deformation and injury risk. This type of centric/non centric physical to finite element modeling methodology could be used to guide innovation for new methods to prevent concussion.

**Keywords:** Ice hockey, Helmets, Standards, Concussion

## **1.0 Introduction**

Mandatory protective headgear in impact and contact sports help protect athletes against traumatic brain injuries (TBI) including intracranial bleeds and skull fractures. However, mild traumatic brain injuries (mTBI), such as concussions, are still common with studies reporting helmets not effective in managing the risk of mTBI [1,2]. The National Hockey League (NHL) report an increase in mTBIs' over the last decade accounting for 18% of all hockey injuries [4]. These statistics suggest that changes in the game including improved helmet technology have had little effect on the incidence of concussion.

Present helmet technology is designed to minimize peak linear acceleration during a direct impact [6]. Linear acceleration was chosen as the performance metric for evaluating helmets as this measure has been associated with TBI [6;7;8]. As a result, linear dominant impact conditions have been utilized in standards to evaluate helmets [9;10]. These standards typically use a headform and monorail system for primarily centric (defined as the impact vector passing through the center of gravity of the head) impacts. However, rotational acceleration has also been identified as an important factor in the incidence of concussion and must also be measured. Higher rotational acceleration responses tend to result from non-centric impacts (defined as impacts whose vector does not pass through the centre of gravity of the head) [11;12]. These rotations cause shear stress within the brain which has been proposed as a predictor for mTBI [11;12]. Current helmet standards do not consider rotational acceleration when assessing helmet performance despite several studies associating rotational acceleration to risk of sustaining a

concussion [5;12;13]. However, definitive thresholds of injury for concussion using linear and rotational acceleration have yet to be elucidated; this difficulty has been identified by researchers to be due to the kinematics not accounting for the interaction between the impact induced motions and the brain tissue [15;16]. As a result, advanced computational models have been developed to better understand the effect of impact head kinematics on brain tissue damage [13;14;15]. Measuring brain tissue deformation using finite element models of the human brain is considered an effective method in evaluating risk of sustaining an mTBI [16].

Finite element modeling of the brain during impact allows for the examination of the effect of complex loading curves on brain tissue deformations. The characteristics of these linear and rotational acceleration loading curves can then be used as input parameters into complex brain models which can then simulate the deformation of tissue resulting from the kinematics of an impact event [17;18]. Past research has shown how this method can predict the effect of linear and rotational accelerations on the stresses and strains imparted to the brain through car crash analysis as well as hockey and football helmet impacts [13;19;20]. As a result, finite element models for the head and brain provide an opportunity to use brain deformation values to evaluate the ability of a hockey helmet to reduce the risk of brain injury [20].

There is presently no standard which uses a centric/non-centric impact method coupled with finite element analysis to measure brain deformations from helmeted impacts. If such a method was developed it may aid in supplying more information on helmet performance using linear and rotational acceleration as well as brain deformation metrics [13;14;15;21]. Previous research has investigated this type of protocol using a linear impactor system, which was created to replicate player to player collisions [5]. This linear impactor method is different from current drop tower methods used by certification bodies to certify helmets. The development of this type

of protocol using the monorail drop system to include centric and non-centric impacts may allow for easier adoption this new protocol using current test equipment.

The objective of this study was to use a monorail centric/non-centric impact methodology to compare the dynamic responses of a helmeted and un-helmeted Hybrid III headform. In addition, VN and EPP helmets were tested determine if there is any difference in the management of linear and rotational acceleration between these impact absorbing liners using the proposed protocol.

## **2.0 Methodology**

### *2.1 Equipment*

A monorail drop rig was used (Figure 1) to complete the proposed testing protocol for the evaluation of the performance of hockey helmets. For the purpose of this study a 50<sup>th</sup> percentile male Hybrid III head- and neckform (mass  $6.08\text{kg} \pm 0.01\text{kg}$ ) was attached to the drop carriage by the base of the neckform with a special jig designed to ensure a  $90^\circ$  angle between the z-axis of the headform and the monorail (Figure 2). A  $0.46 \pm 0.01\text{m}$  tall anvil extension  $0.104 \pm 0.05\text{m}$  in diameter was firmly fixed to the monorail base. For non-centric impacts the anvil extension was moved horizontally  $6.5\text{cm}$  in line with the x-axis of the headform and secured with C-clamps. Secured on the tip of the impact anvil was a hemispherical nylon pad (diameter  $0.126 \pm 0.01\text{m}$ ) covering a modular elastomer programmer (MEP) 60 Shore Type A ( $0.025 \pm 0.05\text{m}$  thickness) disc (Figure 3). Together the nylon pad and MEP disc weighed  $0.908 \pm 0.001\text{kg}$ . The MEP was chosen as it is a common material used in helmet standards (CSA; NOCSAE). The nylon and MEP disc combination was not designed to reflect any particular impact scenario on the ice.

A 50<sup>th</sup> percentile adult male Hybrid III headform (mass  $4.54 \text{ kg} \pm 0.01\text{kg}$ ) (Figure 4) was used in this study. This type of headform is designed to respond in a reproducible and reliable manner and is primarily used in impact reconstructions [22]. The headform was instrumented with nine single-axis Endevco7264C-2KTZ-2-300 accelerometers according to Padgaonkar's orthogonal 3-2-2-2 linear accelerometer array protocol to measure the three dimensional kinematics of the head from an impact [23]. The headform coordinate system was defined with a left-hand rule. Positive axes were directed toward the anterior, toward the right ear and caudally for  $x$ ,  $y$  and  $z$  respectively. The Hybrid III neck with a mass of  $1.54 \pm 0.05 \text{ kg}$  was composed of 4 butyl rubber discs interlocked between five aluminum plates to simulate human vertebrae. The discs were offset towards the front 0.5cm and were slit to elicit a different response in flexion from that in extension [24].

## *2.2 Data Collection*

Inbound velocity was set using the Cadex Impact v5.7a computer program and recorded using a velocimeter (time gate). The nine mounted single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) were sampled at 20 kHz and the signals were passed through a TDAS Pro Lab system (DTS, Calabasas, CA) prior to being processed by TDAS software.

## *2.3 Procedure*

The Hybrid III was dropped at three different inbound velocities (2, 4 & 6 m/s) in order to examine how the dynamic response changes as velocity increased (Marino and Drouin, 2000). Three impact conditions were chosen for preliminary investigation of non-centric impacts using the monorail drop rig and are shown/listed in Figure 5 and Table 1 [25].

Two models of helmets were tested, with three helmets of each model used for a total of 6 helmets impacted. Each model had identical two piece polyethylene (2PE) shells with either VN or EPP liners. Dimensions of the shell and foam liner are described in Table 2.

The headform and helmeted headform was impacted using a monorail drop rig and each condition tested three consecutive times, which is standard procedure for testing multiple-impact helmets [9;10]. During testing the average time between impacts was  $5 \pm 0.50$  min, which exceeds requirements by current standards [9;10]. Impact site accuracy was ensured by marking the helmet with a permanent marker when it was in contact with the impact cap prior to the first drop. The helmet was reset after each impact to ensure the mark on the helmet was in line with the mark on the impact cap. A different helmet was used for each impact velocity; therefore a total of 162 total helmeted impacts were performed. For the un-helmeted headform condition there was a total of 81 impacts.

#### *2.4 Finite element model (University College Dublin Brain Trauma Model)*

In addition, the resulting three-dimensional loading curve responses (x, y and z) were applied to the University College Dublin (UCDBTM) finite element model to produce brain deformation measurements [26;27]. The UCDBTM model geometries were determined through medical scans of a male cadaver. The head was comprised of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid, grey and white matter, cerebellum and brain stem; totalling 26,000 elements [26]. The validation of the model was found to be in agreement with intracranial pressure and brain motion data taken from select cadaver experiments [28;29].

The material properties of the model were taken from the literature (Table 3, 4) [30 – 34]. The material model used for the brain was linearly viscoelastic combined with a large deformation theory. The behaviour of this brain was characterized as viscoelastic in shear with a

deviatoric stress rate dependent on the shear relaxation modulus [26]. The compressive nature of the brain was elastic. The shear characteristics of the viscoelastic behaviour were described thus:

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t} \quad (1)$$

where  $G_{\infty}$  is the long term shear modulus,  $G_0$  is the short term shear modulus and  $\beta$  is the decay factor [26]. The hyperelastic material model used for the brain in shear was expressed thus:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)} \quad (2)$$

where  $C_{10}$  and  $C_{01}$  are temperature-dependent material parameters, and  $t$  is in seconds [35]. A sliding boundary between the CSF and brain was used to simulate the brain skull interface. The CSF was modelled using solid elements with a high bulk and low shear modulus. This was done so that the CSF could behave similar to a fluid. The contact interaction was specified as no separation and used a friction coefficient of 0.2 [36].

The loading curves from the physical reconstructions were input at the centre of gravity of the model and brain deformations calculated. Von Mises stress and maximum principal strain were the metrics used to measure the resulting brain deformation in the cerebrum. The cerebrum was chosen as the region of interest for this study as it remains the only part of the brain to have been validated for the use of finite element modelling. The deformation metrics chosen for this study were selected from previous anatomical and reconstructive research showing correlations to brain injury [13 – 15]. The null hypothesis is that there would be no difference in response between the centric and non-centric impacting conditions. The second null hypothesis is that there would be no difference in response between the ice hockey helmets with different foam liners (VN and EPP). Statistical test for significance between the impact sites (centric vs non-centric) and helmet foam liners (VN and EPP) were conducted by 2-way ANOVA with a 95% confidence interval. Post hoc analyses were conducted using a Tukey HSD test.

There are some limitations to the research as presented in this study. Using a rigid physical head form to measure the response to an impact was chosen to increase the control over impact vector and helmet interfaces. While this facilitates these interfaces, the Hybrid III is not a replication of a real human head, and thus the loading curves generated would not be those of a biofidelic system, however it has been used extensively in impact research. In addition, 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 [20;40]. 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 dynamic 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 characteristics of human brain tissue which would influence the results. A subset of elements across the tentorium was removed as they were undergoing slightly abnormal restrictions (Doorly, 2007), however the rest of the cerebrum underwent a rigorous validation against intracranial pressure and brain motion and was in agreement with experimental data. As a result, the 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. In addition injury data from this model is compared to literature, however it would be more ideal to compare it to injury reconstruction research using the same version of the UCDBTM.

### **3.0 Results**

The results for the impacts are shown in tables 5, 6, and 7. Across all sites and velocities, the use of a VN or EPP lined ice hockey helmet reduced the dynamic response and brain deformation ( $p < 0.05$ ). When comparing the different helmet liners, the VN liner had lower magnitude linear and rotational acceleration at 2 m/s ( $p < 0.05$ ) at the FBPA site than the EPP liner. At the SCG impact site, the VN liner had lower magnitude linear and rotational acceleration at 2 m/s as well as just rotationally at 4 m/s ( $p < 0.05$ ). The VN liner had lower magnitude responses at all three velocities at the RBNA site ( $p < 0.05$ ). The VN helmet produced lower linear and rotational acceleration values at 2 and 4 m/s and linearly at 6 m/s ( $p < 0.05$ ).

When examining the results of the brain deformation metrics, the VN helmet had lower responses than the EPP helmet at the FBPA site at 2.0 m/s, and 6.0 m/s for both maximum principal strain (MPS) and von Mises stress (VMS). At the 4.0 m/s velocity condition, the VN helmet had lower values than the EPP helmet for MPS but not VMS ( $p < 0.05$ ). For the SCG impact site, the helmets had similar values at 2.0 m/s and 6.0 m/s for MPS and VMS ( $p < 0.05$ ). At the 4.0 m/s impact velocity, the VN liner had lower values than the EPP helmet when using MPS and VMS as a metric ( $p < 0.05$ ). For the RBNA impact site the VN yielded lower magnitudes of MPS and VMS at 4.0 m/s and 6.0 m/s. At the 4.0 m/s impact velocity the VN and EPP helmets had equivalent values ( $p < 0.05$ ).

#### **4.0 Discussion**

The impact protocol used in this study was designed to evaluate helmets under linear dominant (centric) and rotationally dominant (non-centric) impacts using the standardized monorail drop rig. The results indicate, as in previous studies, that the material used in the liners of ice hockey helmets has a significant effect on the dynamic response of the Hybrid III headform. The vinyl

nitrile helmet performed significantly better than the EPP helmet at 2 m/s in both linear and angular acceleration. Vinyl nitrile was also significantly better in managing angular acceleration at 4 m/s. At impact site FBPA the VN liner was significantly better in linear and angular acceleration at 2 m/s. The vinyl nitrile liner was also significantly better at SCG 2 m/s in both linear and angular acceleration. Also at SCG, the VN liner performed better angularly at 4 m/s. At RBNA, the VN liner performed better in linear and angular acceleration for 2 and 4 m/s drops. It also managed linear acceleration better at 6 m/s than the EPP liner. Overall, the results indicate that the VN liner was more effective at reducing the brain deformation metrics associated with concussion over the impact sites and velocity ranges tested within this research which is similar to previous literature [20]. These results show that it is possible to influence the brain deformation metrics using protective devices such as ice hockey helmets.

Previous research addressing the potential risk of injury has been conducted by various researchers employing methods in anatomical and reconstructive areas of brain tissue deformation. Using similar physical to finite element modeling methods, peak linear and rotational acceleration risks of injury have been proposed as: 82 g and 5900 rad/s<sup>2</sup> for 50%; 106 g and 7200 rad/s<sup>2</sup> for 80% risk of concussion [13]. The MPS values indicating a 80% risk of injury is from 0.225 to 0.244 for 80% [13; 14; 15]. While direct comparisons can't be made as the material properties of the models are different, it can be used to place the results into context of the literature. When using this research to frame the results of this study, interesting relationships between dynamic response and brain deformation metrics emerge. At the 2 m/s impact condition the helmets all perform well below the 50% risk of injury for these injury metrics. When the velocity is increased to 4 m/s the dynamic results would indicate a risk of injury below 80% whereas the brain deformation metrics used would indicate a risk well in

excess of 80%. These results demonstrate how the linear and rotational acceleration loading curves combine to create larger deformations in brain tissue. This also indicates that when evaluating performance in terms of risk of injury, using brain deformation metrics to evaluate safety of helmets may be beneficial [16].

As previous literature has shown, linearly dominant motion is more likely to cause focal injury and intracranial pressure [37], while rotationally dominant motions are more likely to cause diffuse shear strains of the brain [38;39]. These diffuse shear strains of brain tissue are thought to be related to concussive injuries [11], which has been a major issue for the game of ice hockey. Finite element modeling research has also shown that when combinations of both linear and angular acceleration loading curves are input into the model, the resulting risk of injury is often more severe. The protocol proposed in this study was successful in creating centric and non-centric impact condition which produced results showing differences in the linear and rotational performance between the ice hockey helmets. The results indicate that measuring linear acceleration for testing standards alone may not properly evaluate the performance of ice hockey helmets when attempting to reduce the incidence of concussion.

## **5.0 Conclusion**

The results of this research show the importance of evaluating the performance of protective helmets over a range of impact conditions designed to produce both linear and angular acceleration. The use of brain deformation metrics in this research demonstrates how lower linear and rotational acceleration magnitudes can create higher risks of injury in terms of brain tissue stresses in the cerebrum. The use of finite element modeling of the human brain along with a centric/non-centric protocol provides an opportunity for researchers and helmet developers to

observe how the dynamic response produced from these impacts influence brain tissue deformation and injury risk. This type of centric/non centric physical to finite element modeling methodology could be used to guide innovation for new methods to prevent concussion.

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**Table 1.** The centric and non-centric impact conditions.

FBPA	Midpoint between the anterior mid-sagittal and right coronal planes in the transverse plane	45° rotation in the transverse plane
SCG	Right intersection of the coronal and transverse planes	No vertical or horizontal rotation was applied to the vector
RBNA	Midpoint between the posterior mid-sagittal and right coronal planes in the transverse plane	-45° rotation in the transverse plane

**Table 2.** Hockey helmet characteristics

Helmet	Liner	Mass (kg)	Shell Thickness Front (mm)	Foam Thickness (mm)	
				Front	Front Boss
1	EPP	0.565	3.71	10.64	11.68
2	VN	0.587	3.60	12.04	12.08

**Table 3.** 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	15 000	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	Water	0.5	1000
Grey Matter	Hyperelastic	0.49	1060
White Matter	Hyperelastic	0.49	1060
Bridging veins	9.5	-	-

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

	Shear Modulus (kPa)		Decay Constant	Bulk Modulus
	$G_0$	$G_\infty$	( $s^{-1}$ )	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

**Table 5.** Dynamic response and brain deformation results for impacts to an unhelmeted and helmeted (VN/EPP) headform for the three impact sites at 2 m/s. Brackets denote standard deviation.

		Peak Resultant Acceleration		Brain Deformation Metric	
		Linear (g)	Rotational ( $krad/s^2$ )	Maximum Principal Strain	Von Mises Stress (kPa)
FBPA	Un-helmeted	110.7 (0.6)	11.2 (0.08)	0.348 (0.006)	13.4 (0.31)
	EPP	40.0 (4.7)	3.26 (0.48)	0.154 (0.001)	4.54 (0.27)
	VN	31.8 (1.2)	2.43 (0.08)	0.126 (0.001)	3.64 (0.08)
SCG	Un-helmeted	105.2 (5.8)	10.7 (0.63)	0.259 (0.012)	8.96 (0.44)
	EPP	30.3 (0.9)	2.63 (0.14)	0.115 (0.002)	3.47 (0.02)
	VN	24.2 (0.1)	2.069 (0.04)	0.115 (0.001)	3.52 (0.10)
RBNA	Un-helmeted	74.8 (0.8)	9.29 (0.07)	0.243 (0.010)	8.19 (0.38)
	EPP	31.8 (3.2)	3.46 (0.41)	0.157 (0.004)	4.46 (0.08)
	VN	23.4 (0.4)	2.61 (0.09)	0.152 (0.003)	4.30 (0.07)

**Table 6.** Dynamic response and brain deformation results for impacts to an unhelmeted and helmeted (VN/EPP) headform for the three impact sites at 4 m/s. Brackets denote standard deviation.

		Peak Resultant Acceleration		Brain Deformation Metric	
		Linear (g)	Rotational (krad/s <sup>2</sup> )	Maximum Principal Strain	Von Mises Stress (kPa)
FBPA	Un-helmeted	244.5 (2.3)	26.1 (0.42)	0.819 (0.015)	34.2 (0.49)
	EPP	93.1 (6.6)	7.86 (0.77)	0.328 (0.010)	11.0 (0.57)
	VN	97.8 (10.7)	6.66 (0.98)	0.282 (0.011)	9.60 (0.32)
SCG	Un-helmeted	259.4 (6.9)	25.4 (0.39)	0.569 (0.005)	18.9 (0.09)
	EPP	82.4 (6.9)	8.68 (0.97)	0.310 (0.019)	9.58 (0.59)
	VN	81.4 (5.7)	6.05 (0.63)	0.252 (0.015)	7.70 (0.41)
RBNA	Un-helmeted	179.1 (2.1)	22.6 (0.24)	0.591 (0.005)	19.0 (0.12)
	EPP	80.3 (5.6)	9.71 (1.22)	0.448 (0.026)	13.4 (0.88)
	VN	65.9 (4.4)	7.1 (0.53)	0.382 (0.016)	11.2 (0.57)

**Table 7.** Dynamic response and brain deformation results for impacts to an unhelmeted and helmeted (VN/EPP) headform for the three impact sites at 6 m/s. Brackets denote standard deviation.

		Peak Resultant Acceleration		Brain Deformation Metric	
		Linear (g)	Rotational (krad/s <sup>2</sup> )	Maximum Principal Strain	Von Mises Stress (kPa)
FBPA	Un-helmeted	372.6 (15.8)	37.8 (2.78)	1.279 (0.115)	46.5 (3.30)
	EPP	238.9 (19.2)	19.5 (1.45)	0.673 (0.009)	26.2 (0.91)
	VN	259.0 (8.4)	18.6 (0.57)	0.580 (0.019)	21.5 (1.30)
SCG	Un-helmeted	379.2 (5.9)	35.5 (1.05)	0.941 (0.018)	30.5 (0.51)
	EPP	198.8 (28.1)	20.4 (3.78)	0.683 (0.069)	23.5 (2.74)
	VN	224.6 (35.9)	18.4 (2.36)	0.623 (0.023)	19.8 (0.62)
RBNA	Un-helmeted	289.7 (5.6)	37.3 (3.73)	1.053 (0.021)	34.0 (0.95)
	EPP	176.7 (8.5)	21.8 (1.80)	1.056 (0.096)	31.6 (3.18)
	VN	158.4 (7.0)	19.5 (0.89)	0.873 (0.017)	26.7 (0.15)



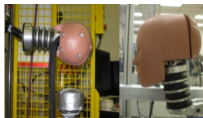
**Fig 1.** Monorail drop rig.



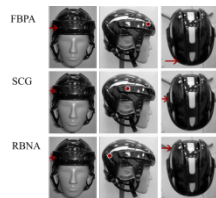
**Fig 2.** 90° Jig Attachment for Hybrid III head and neckform



**Fig 3.** MEP Nylon Cap on an anvil extension



**Fig 4.** Hybrid III neck and headform.



**Fig 5.** Images depicting the impact locations on an ice hockey helmet.