Title	Determining the relationship between linear and rotational acceleration and MPS for different magnitudes of classified brain injury risk in ice hockey
Authors(s)	Clark, J. Michio, Post, Andrew, Hoshizaki, Thomas Blaine, Gilchrist, M. D.
Publication date	2015-09-11
Publication information	Clark, J. Michio, Andrew Post, Thomas Blaine Hoshizaki, and M. D. Gilchrist. "Determining the Relationship between Linear and Rotational Acceleration and MPS for Different Magnitudes of Classified Brain Injury Risk in Ice Hockey." International Research Council on the Biomechanics of Injury (IRCOBI), September 11, 2015.
Conference details	International Research Council on Biomechanics of Injury Conference, Lyon, France, 9-11 September 2015
Publisher	International Research Council on the Biomechanics of Injury (IRCOBI)
Item record/more information	http://hdl.handle.net/10197/8682

Determining the Relationship between Linear and Rotational Acceleration and MPS for Different Magnitudes of Classified Brain Injury Risk in Ice Hockey

James Michio Clark, Andrew Post, Thomas Blaine Hoshizaki, Michael D. Gilchrist

Abstract Helmets have successfully decreased the incidence of traumatic brain injuries (TBI) in ice hockey, yet the incidence of concussions has essentially remained unchanged. Current ice hockey helmet certification standards use peak linear acceleration as the principal measuring helmet performance, however peak linear acceleration may not be an appropriate variable to evaluate risk at all magnitudes of brain injury. The purpose of this study is to determine the relationship between linear acceleration, rotational acceleration and maximum principal strain (MPS) for different magnitudes of classified brain injury risk in ice hockey. A helmeted and unhelmeted Hybrid III headform were impacted to the side of the head at two sites and at three velocities under conditions representing three common mechanisms of injury. Resulting linear and rotational accelerations were used as input for the University College Dublin Brain Trauma Model (UCDBTM), to calculate MPS in the brain. The resulting MPS magnitudes were used to separate the data into three groups: low risk; concussion; and TBI. The results demonstrate that the relationship between injury metrics in ice hockey impacts is dependent on the magnitude of classified injury risk and the mechanism of injury.

Keywords ice hockey, traumatic brain injuries, concussion, low risk impacts, impact biomechanics.

I. INTRODUCTION

Severe head injuries have historically been the primary concern of sport officials, resulting in the use of helmets [1]. Since the introduction of helmets, skull fractures and other traumatic brain injuries (TBI) have largely disappeared from sport [2]. The incidence of concussions remains common, however [2-5]. One possible explanation for this continued high incidence of concussion is the fact that current ice hockey helmet standards use a criteria that does not fully reflect the risk of injury. Current ice hockey helmet standards use peak linear acceleration as the principal measure of brain trauma [6-8]. However, peak linear acceleration alone does not reflect all aspects of brain trauma [9-11]. Linear acceleration has been shown to predict the risk of TBI, including subdural hematomas and skull fractures [12-15], whereas rotational acceleration has been associated with concussion and diffuse axonal injury (DAI) [16-18]. Linear and rotational acceleration have been shown to have a low correlation to injury, and brain deformation metrics have been used to bridge the gap between response and injury [10][19]. Research has shown that MPS has a higher correlation with brain injury than peak linear or peak rotational acceleration alone [10][19-21]. Furthermore the strain in the axonal direction has been found to be a better injury predictor than MPS for a concussion data set from the National Football League [22].

The limited ability of peak linear and rotational acceleration to predict the risk of injury has led researchers to suggest that the use of finite element (FE) models to measure brain tissue strains could be a more informative solution [14][23-24]. In efforts to reduce the incidence, head injuries in sports research have examined the relationship between linear and rotational acceleration and brain tissue strains [14][25-29]. Using a simplified 2D head model, Ueno and Melvin [30] showed that linear acceleration influenced the amount of strain, while rotational acceleration was correlated with shear strains. However, Forero Rueda et al. [31] found high strains in the brain tissue correlated with rotational acceleration and not with linear acceleration. These findings are similar to correlations between linear and rotational acceleration and MPS for impacts to ice hockey and American football helmets, which showed rotational acceleration correlated with MPS, whereas linear acceleration did not produce the same correlations [19][28][32-34]. As a result, current ice hockey helmet standards may not be using the appropriate variables to evaluate brain trauma risk. The purpose of this study is

J. M. Clark is M.Sc. student in Biomechanics at the University of Ottawa in Canada (1-613-562-5800 ext. 7210, e-mail: jclar136@uottawa.ca). A. Post is Post-doctoral fellow and T. B. Hoshizaki is Prof. of Biomechanics in the Dept. of Human Kinetics at the University of Ottawa. M. D. Gilchrist is Head of School in the School of Mechanical and Materials Engineering at University College Dublin.

to determine the correlation between linear acceleration, rotational acceleration and MPS for different magnitudes of classified brain injury risk in ice hockey.

II. METHODS

Experimental Testing

To determine the correlation between injury metrics for different magnitudes of classified brain injury risk in ice hockey, an adult 50th percentile Hybrid III headform was used for all impact conditions. The Hybrid III headform (4.54 ± 0.01 kg) was attached to an unbiased neckform (2.11 ± 0.01 kg) [35] and was instrumented with nine single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) in a 3-2-2-2 accelerometer array [36]. The headform was impacted under helmeted and unhelmeted conditions for impacts representing falls, elbow-to-head and shoulder-to-head. Helmeted and unhelmeted conditions were used in order to create a range of different magnitudes of classified brain injury risk. For the helmeted conditions, the headform was equipped with a vinyl nitrile (VN) ice hockey helmet. For each condition the headform was impacted at two sites (centric, non-centric), as shown in Fig. 1. The centric, non-centric sites were chosen because impacts to the side of the head have been reported as common impact locations in ice hockey [36]. Inbound velocities of 3 m/s, 5 m/s and 7 m/s were selected. These velocities were chosen as this represents a low to high range of skating speeds in ice hockey [38]. Signals for the nine accelerometers were collected at 20 KHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered with a CFC class 1000 filter. Three trials were conducted for each condition and peak linear and rotational accelerations of the headform were measured.

Fig. 1. Impact locations and vectors on the ice hockey helmet, as shown by the red arrows.

Impact Conditions

Falls

A monorail drop rig with a 60 shore A modular elastomer programmer (MEP) anvil was used to simulate falls to the ice. The monorail drop rig consists of a 4.7 m long rail and has a drop carriage in which the Hybrid III headform and an unbiased neckform were attached (Fig. 2). The monorail drop rig was connected to a computer equipped with Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC). The Cadex Software was used to set up the inbound velocity and the velocity was measured using a photoelectric time gate. To avoid unnecessary equipment damage, impacts to an unhelmeted headform at 7 m/s were not completed.

Fig. 2. Monorail drop system used to simulate head impacting the ice (MEP anvil).

Collisions

A pneumatic linear impactor with two different strikers was used to simulate collisions. The pneumatic linear impactor consists of a frame and a table. The frame supports the impacting arm, the compressed air canister and piston (Fig. 3a). The impacting arm (13.01 kg) was propelled by compressed air towards the headform and the impact velocity measured by a laser time gate just prior to impact. The mass of the impacting arm was similar to the mass of shoulder-to-head impacts in ice hockey reconstructions [39]. Two different strikers were fitted to the end of the impacting arm to simulate shoulder and stiff elbow collisions. Shoulder impacts were simulated by fitting the end of the impacting arm with striking surface consisting of a nylon disc (diameter 13.2 mm) covered with 67.79 ± 0.01 mm thick layer of vinyl nitrile R338V foam and a Reebok 11k shoulder pad (Fig. 3b) [39]. To simulate stiff elbow collisions, a striker consisting of a hemispherical nylon pad with a 35.71 ± 0.01 mm thick vinyl nitrile 602 foam disk underneath was used (Fig. 3c). This striker produces similar peak linear and rotational acceleration to that of elbow strikes of ice hockey players [40]. The Hybrid III headform and unbiased neckform were attached to a sliding table (12.78 ± 0.001 kg) to allow for movement post-impact. The headform was affixed to a movable locking base, which allowed for the headform to be oriented in five degrees of freedom and to remain fixed in position during testing.

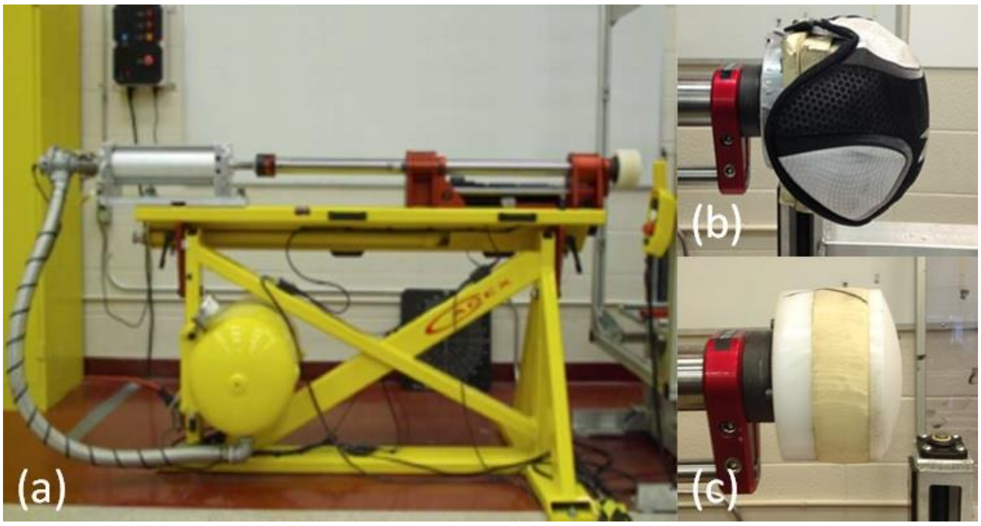

Fig. 3. Pneumatic Linear Impactor: (a) frame supporting the impacting arm, (b) shoulder pad striker, (c) stiff elbow striker.

Computational Modelling

The resulting linear and rotational accelerations served as input to the University College Dublin Brain Trauma Model (UCDBTM) [41-42]. The UCDBTM was used to calculate peak MPS in the cerebrum. The head geometry of the UCDBTM was based on computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver [42]. The UCDBTM had approximately 26,000 reduced integration 8-node hexahedral elements representing the scalp, skull, pia, falx, tentorium, CSF, grey and white matter, cerebellum and brain stem [41-42]. The hourglassing energy remained below the recommended 10% of total energy [43]. The model was validated against cadaveric pressure responses conducted by Nahum et al. [44] and brain motion research conducted by Hardy et al. [43], as well as reconstructions of traumatic brain injuries [46-47]. The shape of the response and the duration of the effect of the model were found to closely approximate the cadaveric pressure responses [44] and brain motion [45] from experimental results [41-42]. As such, the correlation was found to be good and the model was considered to be validated [41-42].

The material characteristics of the model are presented in Tables 1 and 2. The brain characteristics were taken from the anatomical research conducted by Zhang et al. [48]. The brain tissue was modelled using a linearly viscoelastic model combined with large deformation theory [41-42][49-50]. The behaviour of the tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [41]. The compression of the brain tissue was considered elastic. The shear characteristics of the viscoelastic brain were defined using the following equation:

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

Where G_∞ is the long-term shear modulus, G_0 is the short-term shear modulus and β is the decay factor [41]. A Mooney–Rivlin hyperelastic material model was used for the brain to maintain these properties in conjunction with a viscoelastic material property in ABAQUS, giving the material a decay factor of $\beta = 145\ s^{-1}$ [41]. The hyperelastic law was expressed using the following equation:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-\frac{t}{0.008}} + 1103e^{-\frac{t}{0.15}}(Pa), \tag{2}$$

where C_{10} is the mechanical energy absorbed by the material when the first strain invariant changes by a unit step input, C_{01} is the energy absorbed when the second strain invariant changes by a unit step [49-50] and t is the time in seconds. The modelling of the CSF was conducted using solid elements with the bulk modulus of water and a low shear modulus [41-42].

The contact interaction at the skull–brain interface was assigned no separation and used a friction coefficient of 0.2 [51].

TABLE 1
MATERIAL PROPERTIES OF UCDBTM

Material	*Poisson's Ratio*	*Density (kg/m^3)*	*Young's Modulus (Mpa)*
Scalp	0.42	1000	16.7
Cortical Bone	0.22	2000	15000
Trabecular Bone	0.24	1300	1000
Dura	0.45	1130	31.5
Pia	0.45	1130	11.5
Falx	0.045	1140	31.5
Tentorium	0.45	1140	31.5
CSF	0.5	1000	Water
Grey Matter	0.49	1060	Hyperelastic
White Matter	0.49	1060	Hyperelastic

TABLE 2
MATERIAL CHARACTERISTICS OF THE BRAIN TISSUE FOR THE UCDBTM

	Shear Modulus (kPa)			
Material	G_0	G_∞	Bulk Modulus (s^1)	Decay Constant (Gpa)

Cerebellum	10	2	2.19	80
Brain Stem	22.5	4.5	2.19	80
White Matter	12.5	2.5	2.19	80
Grey Matter	10	2	2.19	80

Analysis

Impact conditions were categorised into three groups based on mean peak MPS values. The three groups were low risk, concussion and TBI and were separated based on classified risks of injury as determined by football and hospital injury reconstruction and anatomical experiments [10][19][22][47][51-52]. The low risk group was defined as MPS values less than 0.190 [10][19][22][51-52]. The concussion group was defined as subjects who incurred an impact to the head that resulted in the symptomology of concussion with no evidence of TBI lesions on MRI or CT scans. The concussion group consisted of impact conditions that resulted in MPS values between 0.190 [10][19][22][51-52] and 0.387 [45]. The TBI group was defined as impact conditions resulting in MPS values of 0.388 or more [47]. The types of TBI injuries included subdural hematoma, subarachnoid hemorrhage and brain contusions [47]. Maximum principle strain values were chosen as the variable to separate impact conditions into groups because strain has been shown to be a likely mechanism of brain tissue injury that results in concussion [9-11].

To determine the correlation between linear acceleration, rotational acceleration and MPS, Pearson correlation coefficients (r) and r^2 values were calculated. Pearson correlation coefficients were calculated for when all the data was combined. Data was then separated by classified injury risk group and Pearson correlation coefficients were calculated. The probability of making a type 1 error for all comparisons was set at $\alpha=0.05$. All data analyses were performed using the statistical software package of SPSS 19.0 for Windows.

III. RESULTS

Strain distributions at maximum for one case of each of the three classified injury risk groups are shown in Fig. 4. Fig. 5 demonstrates the six degrees of freedom linear and rotational acceleration traces for one case of each of the three classified injury risk groups.

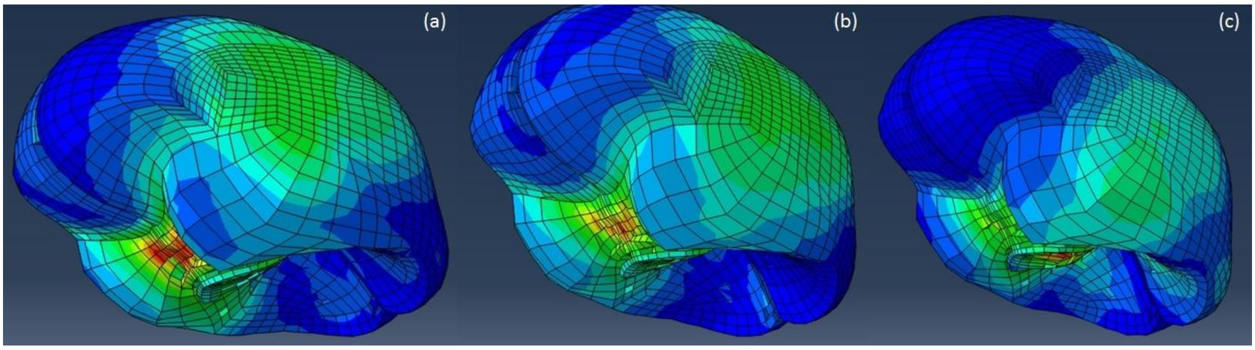

Fig. 4. Strain distributions at maximum for one case of each of the three classified injury risk groups: (a) Low risk group, (b) Concussion group, (c) TBI group.

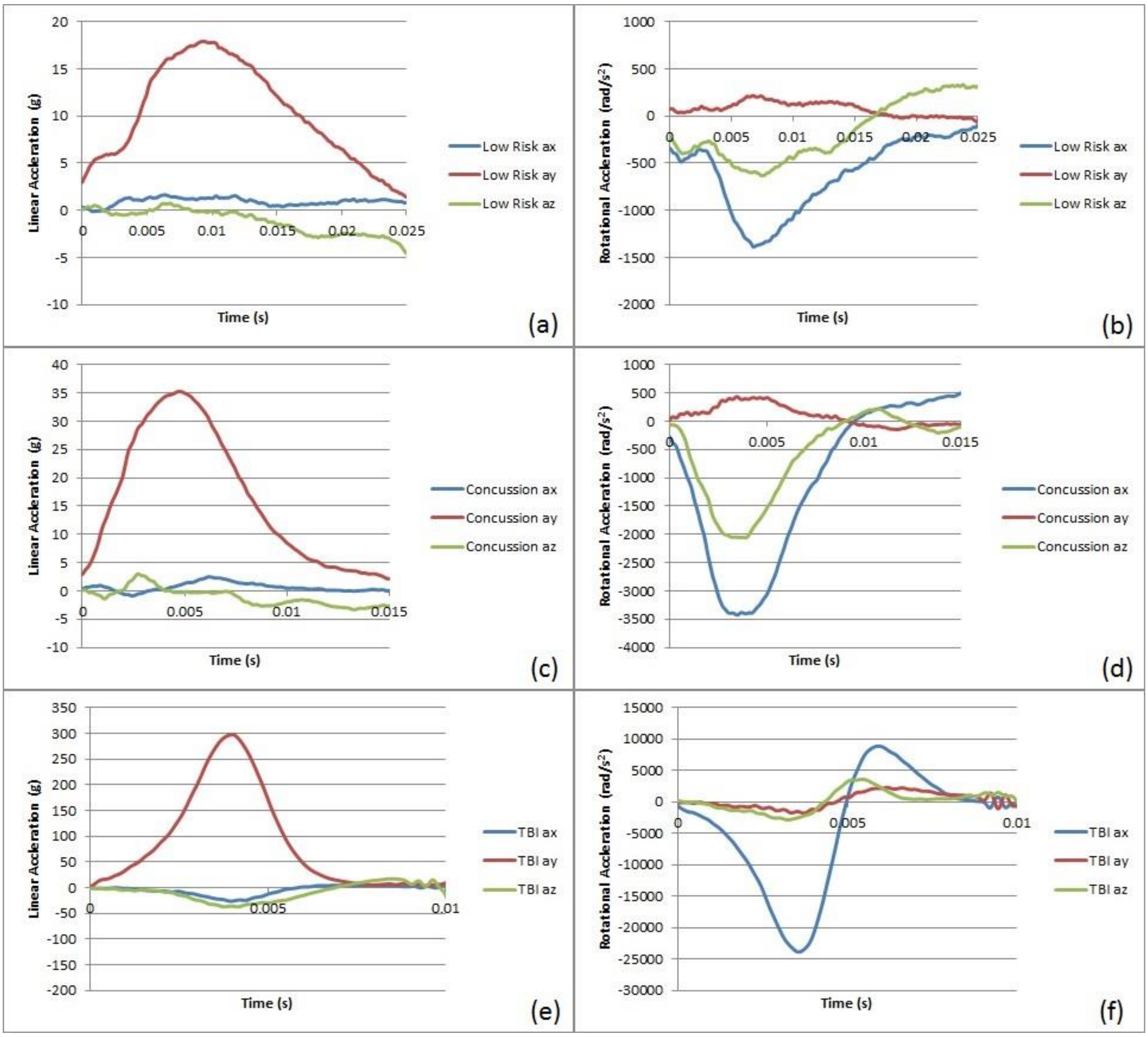

Fig. 5. Six degrees of freedom linear and rotational acceleration traces for one case of each of the three classified injury risk groups: (a) Linear acceleration traces for low risk group, (b) Rotational acceleration traces for low risk group, (c) Linear acceleration traces for concussion group, (d) Rotational acceleration traces for concussion group, (e) Linear acceleration traces for TBI group, (f) Rotational acceleration traces for TBI group.

The classification of impact conditions by magnitudes of brain injury risk are presented in Tables 3–5. The low risk group comprised of all shoulder impacts at 3 m/s. The concussion group consisted of helmeted falls at 3 m/s, elbow impacts at 3 m/s and helmeted impacts for site 2 at 5 m/s and shoulder impact at 5 m/s and 7 m/s. All elbow impacts at 5 m/s and 7 m/s and all falls, except for helmeted impacts to site 2 at 5 m/s, and helmeted falls at 3 m/s were categorised into the TBI group.

TABLE 3

IMPACT CONDITIONS CLASSIFIED AS LOW RISK OF INJURY WITH LINEAR AND ROTATIONAL AND MAXIMUM PRINCIPLE STRAIN (STANDARD DEVIATIONS IN BRACKETS)

Mechanism	Velocity (m/s)	Site	Helmet (Yes/No)	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Maximum Principle Strain
Shoulder	3	1	No	19.2 (0.5)	1479 (47)	0.154 (0.001)
			Yes	17.9 (0.2)	1462 (57)	0.149 (0.006)
		2	No	18.0 (0.5)	1688 (37)	0.176 (0.007)

			Yes	16.3 (0.1)	1206 (38)	0.140 (0.003)

TABLE 4

IMPACT CONDITIONS CLASSIFIED AS A RISK OF CONCUSSION WITH LINEAR AND ROTATIONAL AND MAXIMUM PRINCIPLE STRAIN (STANDARD DEVIATIONS IN BRACKETS)

Mechanism	Velocity (m/s)	Site	Helmet (Yes/No)	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Maximum Principle Strain
Fall	3	1	Yes	49.7 (0.4)	3470 (203)	0.290 (0.004)
		2	Yes	54.4 (3.9)	3473 (380)	0.301 (0.007)
Stiff Elbow	3	1	No	71.0 (1.1)	7606 (151)	0.353 (0.014)
			Yes	17.9 (0.2)	4028 (58)	0.299 (0.013)
		2	No	45.0 (0.6)	3737 (55)	0.302 (0.003)
			Yes	27.1 (0.2)	2161 (159)	0.199 (0.008)
	5	2	Yes	47.1 (0.2)	3814 (94)	0.344 (0.012)
Shoulder	5	1	No	32.0 (0.3)	2572 (50)	0.266 (0.003)
			Yes	28.6 (0.4)	2500 (121)	0.261 (0.004)
		2	No	30.4 (0.3)	2980 (33)	0.276 (0.001)
			Yes	26.6 (0.6)	1971 (125)	0.212 (0.016)
	7	1	No	48.3 (0.8)	2709 (95)	0.316 (0.021)
			Yes	45.2 (0.5)	2666 (68)	0.304 (0.027)
		2	No	47.2 (0.2)	4037 (26)	0.372 (0.003)
			Yes	38.8 (0.6)	3107 (189)	0.278 (0.016)

TABLE 5

IMPACT CONDITIONS CLASSIFIED AS A RISK OF TBI WITH LINEAR AND ROTATIONAL AND MAXIMUM PRINCIPLE STRAIN (STANDARD DEVIATIONS IN BRACKETS)

Mechanism	Velocity (m/s)	Site	Helmet (Yes/No)	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Maximum Principle Strain
Fall	3	1	No	184.7 (4.5)	13049 (1147)	0.549 (0.040)
		2	No	124.1 (1.6)	8332 (155)	0.516 (0.011)
	5	1	No	338.9 (15.6)	10145 (981)	0.822 (0.031)
			Yes	135.9 (10.9)	11304 (228)	0.554 (0.011)
		2	No	228.3 (2.1)	16788 (260)	0.815 (0.018)
			Yes	117.1 (5.1)	8569 (379)	0.562 (0.017)
	7	1	Yes	310.9 (10.8)	25544 (1279)	0.920 (0.021)
		2	Yes	250.8 (9.7)	19869 (666)	0.970 (0.014)
Stiff Elbow	5	1	No	107.2 (2.0)	11304 (228)	0.481 (0.002)
			Yes	67.3 (3.5)	6250 (384)	0.418 (0.008)
		2	No	67.5 (2.0)	6327 (409)	0.487 (0.016)
	7	1	No	129.4 (1.0)	14238 (297)	0.713 (0.016)
			Yes	90.8 (0.6)	9293 (158)	0.538 (0.007)
		2	No	91.2 (2.4)	10057 (290)	0.608 (0.004)
			Yes	67.6 (1.0)	6397 (281)	0.525 (0.006)

When correlations were conducted on all the data together, significant ($p<0.05$) and very strong correlations ($r>0.900$) were found between all injury metrics (Table 6). Table 7 shows that when data was separated by magnitude of classified injury risk, all correlations were significant ($p>0.05$) except for low risk linear acceleration/MPS. Impacts within the risk of concussion showed a strong correlation for linear/rotational acceleration ($r>0.800$) and a moderate correlation for linear acceleration/MPS ($r>0.700$). Conditions associated with a risk of TBI showed strong correlations between all injury metrics ($r>0.800$).

TABLE 6
PEARSON CORRELATIONS BETWEEN LINEAR ACCELERATION AND MAXIMUM PRINCIPAL STRAIN FOR COLLAPSED DATA

Comparison	Pearson Correlation (r)	r^2
Linear/Rotational Acceleration	0.976**	0.952
Linear Acceleration/MPS	0.916**	0.839
Rotational Acceleration/MPS	0.947**	0.947

** Correlation is significant at the 0.01 level (2-tailed).

TABLE 7
PEARSON CORRELATIONS BETWEEN LINEAR ACCELERATION AND MAXIMUM PRINCIPAL STRAIN FOR DIFFERENT MAGNITUDES OF CLASSIFIED INJURY RISK

Classified Injury Risk	Comparison	Pearson Correlation (r)	r^2
Low Risk	Linear/Rotational Acceleration	0.652*	0.425
	Linear Acceleration/MPS	0.435	0.189
	Rotational Acceleration/MPS	0.935**	0.874
Concussion	Linear/Rotational Acceleration	0.811**	0.658
	Linear Acceleration/MPS	0.761**	0.579
	Rotational Acceleration/MPS	0.669**	0.447
TBI	Linear/Rotational Acceleration	0.960**	0.922
	Linear Acceleration/MPS	0.862**	0.743
	Rotational Acceleration/MPS	0.905**	0.819

* Correlation is significant at the 0.05 level (2-tailed).
** Correlation is significant at the 0.01 level (2-tailed).

IV. DISCUSSION

The purpose of this study was to determine the correlation between linear acceleration, rotational acceleration and MPS for different magnitudes of classified brain injury risk in ice hockey. The results demonstrate that when all conditions were collapsed, very strong correlations were found between all injury metrics. However, the relationship between injury metrics for ice hockey impacts was found to be dependent on the magnitude of classified injury risk. The TBI showed strong correlations across all variables, whereas the low risk and concussion groups were found to have low to strong correlations. These results suggest that using linear acceleration as the principal measure of brain trauma may not be appropriate in every situation. As a result, this study demonstrates the importance of selecting appropriate injury metrics to reflect trauma for each injury group.

Low Risk Correlations

The low risk group was found to have correlations of various degrees. A moderate correlation between linear and rotational acceleration was found. Walsh et al. [15] also found that linear and rotational acceleration had a moderate correlation for low risk impacts. This suggests that at low risk of injury, linear and rotational accelerations are related but may not accurately reflect trauma in one another [15][29]. Linear acceleration and MPS were found to have no significant correlation, whereas rotational acceleration and MPS were strongly correlated. These findings are constant with previous research in sport impacts demonstrating rotational acceleration is highly correlated to MPS, while linear acceleration is not [28][31][33-34][56]. Supporting low energy impacts allows for the difference between linear acceleration and MPS to become evident [57], while at low risk of injury rotational accelerations are effective at representing brain tissue strain [12-13][17][58]. Therefore, rotational acceleration may be a more appropriate injury metric for head impact counters to reflect trauma for impacts at low risk of injury.

Concussion Correlations

The concussion group was found to produce strong correlations between all injury metrics. As a result, the concussion group was found to produce higher correlations compared to the low risk group. The strong correlations observed in this study for the concussion group are similar to previous research examining the correlations between peak linear and rotational acceleration and MPS [57][59-60]. However, the strong relationships found in this study and in previous research are likely due to the influence of a large range of impact parameters and increases in energy [11]. The concussion group is comprised of impact ranging in velocity from 3 m/s to 7 m/s, representing all three mechanisms of injury (fall, elbow and shoulder impacts). Increases in velocity across a 4 m/s impact range have been found to produce very strong correlations among linear and rotational acceleration and MPS [61], confirming that an increase in velocity would result in an increase in linear and rotational acceleration and MPS. In addition, mechanism of injury has been shown to influence the correlations among linear and rotational acceleration and MPS [57][60]. As falls and stiff elbow impacts produced high magnitude responses, whereas shoulder impacts produced low responses, this causes high correlations among injury metrics. In contrast, the low risk group was solely comprised of shoulder impacts at 3 m/s. When controlling for impact parameters, rotational acceleration has been found to be highly correlated with MPS, but linear accelerations do not demonstrate the same correlation [28][33-34]. The concussion group had higher correlations than the low risk group due to the influence of velocity and mechanism of injury.

TBI Correlations

The results indicate that at magnitudes of brain injury associated with the risk of TBI, all injury metrics are very strongly correlated ($r > 0.9$). The TBI group was found to produce higher correlations among injury metrics compared to the concussion group. An explanation for the high correlation observed between accelerations and MPS at the TBI risk level could be due to high energy levels [57]. The TBI group was found to consist of high energy impacts, which are associated with high magnitude responses and, as a result, cause high correlations. Previous research using TBI cases for falls has also shown a significant positive correlation between linear and rotational acceleration [62]. This suggests that for impacts associated with a risk of TBI, a reduction in linear acceleration would result in a decrease in the rotational acceleration [59][63]. Thus, helmet safely standards that solely use linear acceleration as their pass-fail metric [1] are able to appropriately reflect trauma for impacts associated with a risk of TBI.

Limitations

The present research should be considered according to its limitations. The three groups of low risk, concussion and TBI were not represented by all mechanisms. As such, this may produce a bias between groups. However, the impact conditions chosen represent a wide range of ice hockey impact characteristics [37-39] and would represent the risk of injury associated with each mechanism. The MPS thresholds for concussion and TBI used to separate the three groups were based on American football and hospital injury reconstructions and anatomical experiments [10][19][22][51-52]. These thresholds may be specific to helmet-to-helmet collisions resulting in concussion for American football, falls resulting in TBI for the hospital setting and anatomical experiments; they may not accurately reflect the risk of injury in ice hockey. However the thresholds chosen for the groups are based upon literature of multiple experimental methods and as such can be used for comparative purposes. It should be noted that the research used to define the TBI group may have had subjects that had the symptomology of concussion as well. The Hybrid III headform is not biofidelic, but it does produce results that are within those expected for cadaveric impacts [64]. The response of the UCDBTM is dependent on the material characteristics that specify linear viscoelasticity for the brain. As such, the response of the UCDBTM is meant to be representative of how the brain may deform under the loading scenarios and may not represent the exact motion of the brain.

V. CONCLUSIONS

This study examined the correlation between linear acceleration, rotational acceleration and MPS for different magnitudes of classified brain injury risk in ice hockey. The results indicate that the relationship between injury metrics in ice hockey impacts is dependent on the magnitude of classified injury risk. The MPS for the low risk group was found to be highly correlated to rotational acceleration and not correlated to linear acceleration, while the concussion group showed strong correlations across all variables due to the influence of velocity and mechanism of injury. The TBI was found to produce the strongest relationships for high energy impacts. This research demonstrates that it is important for helmet standards to select the appropriate injury metric to reflect risk of specific injuries associated with the different magnitudes of classified head injury risk in ice hockey.

VI. REFERENCES

[1] Hoshizaki, T. B. and Brien, S. E. The science and design of head protection in sport. *Neurosurgery*, 2004, 55(4):956–67.
[2] Wennberg, R. A. and Tator, C. H. National hockey league reported concussions, 1986–87 to 2001–02. *Canadian Journal of Neurological Sciences*, 2003, 30(3):206–9.
[3] Pellman, E. J., Powell, J. W., et al. Concussion in professional football: epidemiological features of game injuries and review of literature. Part 3. *Neurosurgery*, 2004, 54:81–97.
[4] Pellman, E. J. and Viano, D. C. Concussion in professional football: summary of the research conducted by the National Football League's Committee on mild traumatic brain injury. *Neurosurgical Focus*, 2006, 21(4):e12.
[5] Casson, I. R., Viano, D. C., Powell, J. W., Pellman, E. J. Twelve years of National Football League concussion data. *Sports Health*, 2010, 2(6):471–83.
[6] Canadian Standards Association. 2009. Ice Hockey Helmets. Z262.1-09. Mississauga, Ontario, Canada.
[7] Sports Legacy Institute. 2013. Head Impact Counter Standard for Sport Helmets. SLI 001. Available from: http://www.sportslegacy.org.
[8] Sports Legacy Institute. 2013. Universal Application Standard for Head Impact Counters. SLI 002. Available from: http://www.sportslegacy.org.
[9] Zhang, L., Yang, K. H. and King, A. I. Biomechanics of neurotrauma. *Neurological Research*, 2001, 23(2–3):144–56.
[10] Kleiven, S. Predictors for traumatic brain injuries evaluated through accident reconstruction. *Stapp Car Crash Journal*, 2007, 51:81–114.
[11] Post, A. and Hoshizaki, T. B. Rotational Acceleration, Brain Tissue Strain, and the Relationship to Concussion. *Journal of Biomechanical Engineering*, 2015, 137:1–8.
[12] Holbourn, A. H. Mechanics of head injuries. *Lancet*, 1943, 2:438–41.
[13] Ommaya, A. K. and Hirsch, A. E. Tolerances for cerebral concussion from head impact and whiplash in primates. *Journal of Biomechanics*, 1971, 4:13–21.
[14] King, A. I., Yang, K. H., Zhang, L. and Hardy, W. Is head injury caused by linear or angular acceleration? *Proceedings of IRCOBI Conference*, 2003, Lisbon (Portugal), 1–12.
[15] Walsh, E. S., Rousseau, P. and Hoshizaki, T. B. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. *Journal of Sports Engineering*, 2011, 13(3):135–43.
[16] Ommaya, A. K. and Gennarelli, T. A. Cerebral concussion and traumatic unconsciousness - correlation of experimental and clinical observations on blunt head injuries. *Brain*, 1974, 97(Dec):633–54.
[17] Gennarelli, T. A., Thibault, L. E., Adams, H., Graham, D. I., Thompson, C. J., Marcincin, R. P. Diffuse Axonal Injury and Traumatic Coma in the Primate. *Annals of Neurology*, 1982, 12(6):564–74.
[18] Gennarelli, T. A. Head injury in man and experimental animals: clinical aspects. *Acta Neurochirurgica*, 1983, 32:1–13.
[19] Zhang, L., Yang, K. H. and King, A. I. A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering*, 2004, 126:226–36.
[20] Schreiber, D. I., Bain, A. C., Meaney, D. F. In vivo thresholds for mechanical injury to the blood brain barrier. *Proceedings of the 41st Stapp Car Crash Conference*, 1997, Lake Buena Vista, FL (USA).

[21] Willinger, R. and Baumgarthner, D. Human head tolerance limit to specific injury mechanisms. *International Journal of Crashworthiness*, 2003, 8(6):605–17.

[22]Giordano, C. and Kleiven, S. Evaluation of axonal strain as a predictor for mild traumatic brain injuries using finite element Modeling. *Stapp Car Crash Journal*, 2014, 58:29-61.

[23]Post, A. and Hoshizaki, T. B. Mechanical Properties Describing Brain Impact Injuries: A Review. *Trauma*, 2012, 14(4):327–49.

[24]Post, A., Hoshizaki, T. B. and Gilchrist, M. D. Finite Element Analysis of the Effect of Loading Shape on Brain Injury Predictors. *Journal of Biomechanics*, 2012, 45(4):679–83.

[25] Thomas, L. M., Roberts, V. L. and Gurdjian, E. S. Experimental intracranial pressure gradients in the human skull. *Journal of Neurology, Neurosurgery & Psychiatry*, 1966, 29:404–11.

[26] Yoganandan, N. and Pintar, F. A. Biomechanics of Temporo-Parietal Skull Fracture. *Clinical Biomechanics*, 2004, 19:225–39.

[27] Rousseau, P., Post, A. and Hoshizaki, T. B. A comparison of peak linear and angular headform accelerations using ice hockey helmets. *Journal of ASTM International*, 2009, 6(1):1–11.

[28] Post, A., Oeur, A., Hoshizaki, B. and Gilchrist, M. D. Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts. *Computer Methods in Biomechanics and Biomedical Engineering*, 2011:1–9.

[29] Oeur, A., Zanetti, Z. and Hoshizaki, T. B. Angular acceleration responses of American football, lacrosse and ice hockey helmets subject to low-energy impacts. *Proceedings of the IRCOBI Conference*, 2014, Berlin (Germany), 81–92.

[30] Ueno, K. and Melvin, J. W. Finite Element Model Study of Head Impact Based on Hybrid III Head Acceleration: The Effects of Rotational and Translational Acceleration. *Journal of Biomechanical Engineering*, 1995, 117(3):319–29.

[31] Forero Rueda, M. A., Cui, L. and Gilchrist, M. D. Finite element modeling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. *Computer Methods in Biomechanics and Biomedical Engineering*, 2011, 14(12):1021–31.

[32] Kleiven, S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *International Journal of Crashworthiness*, 2006, 11(1):65–79.

[33] Post, A., Oeur, A., Hoshizaki, T. B. and Gilchrist, M. D. The Influence of Centric and Non-Centric Impacts to American Football Helmets on the Correlation Between Commonly used Metrics in Brain Injury Research. *Proceedings of IRCOBI Conference*, 2012, Dublin (Ireland), 419–29.

[34] Post, A., Oeur, A., Hoshizaki, T. B. and Gilchrist, M. D. An Examination of American Football Helmets Using Brain Deformation Metrics Associated With Concussion. *Materials and Design*, 2013, 45:653–62.

[35] Walsh, E. S. and Hoshizaki, T. B. Comparative analysis of the Hybrid III neckform to unbiased neckforms using a centric and non-centric impact protocol. *ASTM Symposium on the mechanism of concussion in sports*, 2012, Atlanta (USA).

[36] Padgaonkar, A. J., Krieger, K. W. and King, A. I. Measurement of angular acceleration of a rigid body using linear accelerometers. *Journal of Applied Mechanics*, 1975, 42(30):552–6.

[37] Hutchison, M. G., Comper, P., Meeuwisse, W. H. and Echemendia, R. J. A systematic video analysis of National Hockey League (NHL) concussions, part II: how concussions occur in the NHL. *British Journal of Sports Medicine*, 2013, 00:1–5.

[38] Formenti, F. and Minetti, A. E. Human locomotion on ice: the evolution of ice-skating energetics through history. *The Journal of Experimental Biology*, 2007, 210:1825–33.

[39] Rousseau, P. and Hoshizaki, T. B. Defining the effective impact mass of elbow and shoulder strikes in ice hockey. *Sports Biomechanics*, 2015, in press.

[40] Coulson, N. R., Foreman, S. G. and Hoshizaki, T. B. Translational and rotational accelerations generated during reconstructed ice hockey impacts on a Hybrid III headform. *Journal of ASTM International*, 2009, 6(2):1–8.

[41] Horgan, T. J. and Gilchrist, M. D. The creation of three-dimensional finite element models for simulating head impact biomechanics. *International Journal of Crashworthiness*, 2003, 8(4):353–66.

[42] Horgan, T. J. and Gilchrist, M. D. Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. *International Journal of Crashworthiness*, 2004, 9(4):401–18.

[43]Horgan, T. J. A finite element model of the human head for use in the study of pedestrian accidents. *PhD Thesis*, 2003, University College Dublin, Ireland.

[44]Nahum, A. M., Smith, R. and Ward, C. C. Intracranial pressure dynamics during head impact. *Proceedings 21st Stapp car crash conference*, 1977, New Orleans (USA).

[45] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I. and Tashman, S. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar X-ray. *Stapp car crash journal*, 2001, 51:17–80.

[46] Doorly, M. C. and Gilchrist, M. D. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. *Computer Methods in Biomechanics and Biomedical Engineering*, 2006, 9(6):371–7.

[47] Post, A., Hoshizaki, T. B., Gilchrist, M. D., Brien, S., Cusimano, M. D. and Marshall, S. Traumatic brain injuries: The influence of the direction of impact. *Neurosurgery*, 2015, 76(1):81–91.

[48] Zhang, L., Yang, K., et al. Recent advances in brain injury research: A new human head model development and validation. *Stapp Car Crash Journal*, 2001b, 45:369–93.

[49] Shuck, L. Z. and Advani, S. H. Rheological response of human brain tissue in shear. *Journal of Basic Engineering*, 1972, 94(4):905–12.

[50] Zhou, C., Khalil, T. B. and King, A. I. A new model for comparing responses of the homogeneous and inhomogeneous human brain. *Proceedings of the 39th Stapp Car Crash Conference*, 1995, 121–36.

[51]Bain, A. C. and Meaney, D. F. Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury, *Journal of Biomechanical Engineering*, 2000, 122:615–622.

[52]Morrison, B., Carter, H. L., et al., A tissue level criterion for living brain developed with an in vitro model of traumatic mechanical loading, *Stapp Car Crash Journal*, 2003, 47:93-105.

[53]Mendis, K., Stalnaker, R. and Advani, S. A constitutive relationship for large deformation finite element modeling of brain tissue. *Journal of Biomechanical Engineering*, 1995, 117(4):279–85.

[54] Miller, K. and Chinzei, K. Constitutive modelling of brain tissue: Experiment and theory. *Journal of Biomechanics*, 1997, 30(11):1115–21.

[55] Miller, R., Margulies, S., et al. Finite element modeling approaches for predicting injury in an experimental model of severe diffuse axonal injury. *Proceedings of the 42nd Stapp Car Crash Conference*, 1998, Tempe (USA).

[56] Forero Rueda, M. A., Halley, W. L. and Gilchrist, M. D. Fall and injury incidence rates for jockeys while racing in Ireland, France and Britain. *Injury*, 2010, 41:533–9.

[57] Zanetti, K. A., Hoshizaki, T. B., Gilchrist, M. D. The association between peak resultant linear acceleration and brain tissue deformation in American football-related head Impacts. *7th World Congress of Biomechanics*, Boston (USA).

[58] Margulies, S. S. and Thibault, L. E. A proposed tolerance criterion for diffuse axonal injury in man. *Journal of Biomechanics*, 1992, 25(8):917–23.

[59] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R. and Waekerle, J. F. Concussion in Professional Football: Reconstruction of Game Impacts and Injuries. *Neurosurgery*, 2003, 53(4):799–812.

[60] Post, A., Kendall, M., et al. Characterization of Persistent Concussive Syndrome Through Injury Reconstruction and Finite Element Modelling. *Journal of the Mechanical Behavior of Biomedical Materials*, 2015b, 51:325–35.

[61] Post, A., Oeur, A., Hoshizaki, T. B. and Gilchrist, M. D. Differences in Region Specific Brain Tissue Stress and Strain due to Impact Velocity for Simulated American Football Impacts. *Journal Sports Engineering and Technology*, 2014:1–11.

[62] Post, A., Hoshizaki, T. B., Gilchrist, M. D., Brien, S., Cusimano, M. D. and Marshall, S. The influence of dynamic response and brain deformation metrics on the occurrence of subdural hematoma in different regions of the brain. *Journal of Neurosurgery*, 2013:1–9.

[63] Rowson, S., Duma, S. M., et al. Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion. *Annals of Biomedical Engineering*, 2012, 40(1):1–13.

[64] Kendall, M., Walsh, E. S. and Hoshizaki, T. B. Comparison between Hybrid III and Hodgson-WSU headforms by linear and angular dynamic impact response. *Journal of Sports Engineering and Technology*, 2012, 0(0):1–6.