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

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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.

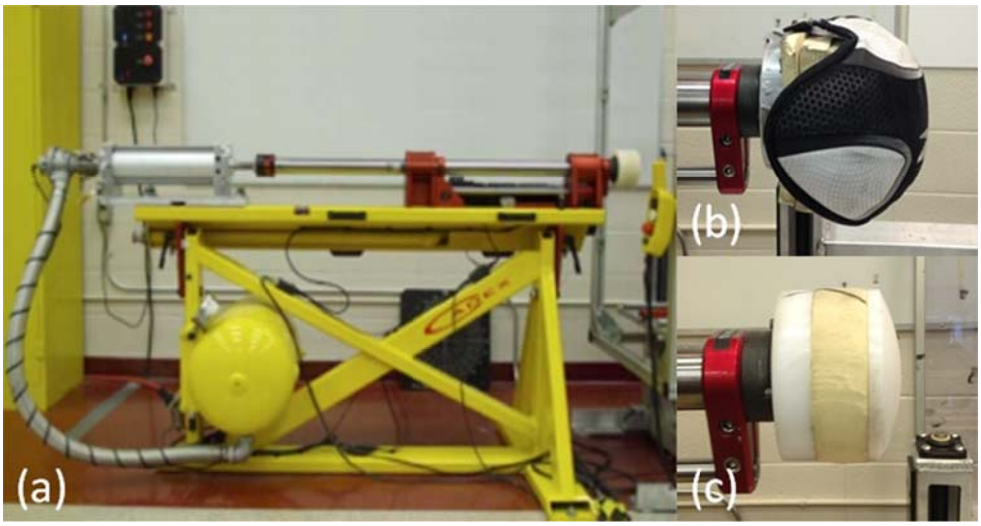

Fig. 3. Pneumatic Linear Impactor: (a) frame supporting the impacting arm, (b) shoulder pad striker, (c) stiff elbow striker.

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:

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

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 β = 145 s^{-1} [41]. The hyperelastic law was expressed using the following equation:

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

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/m³)*	*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 α=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.

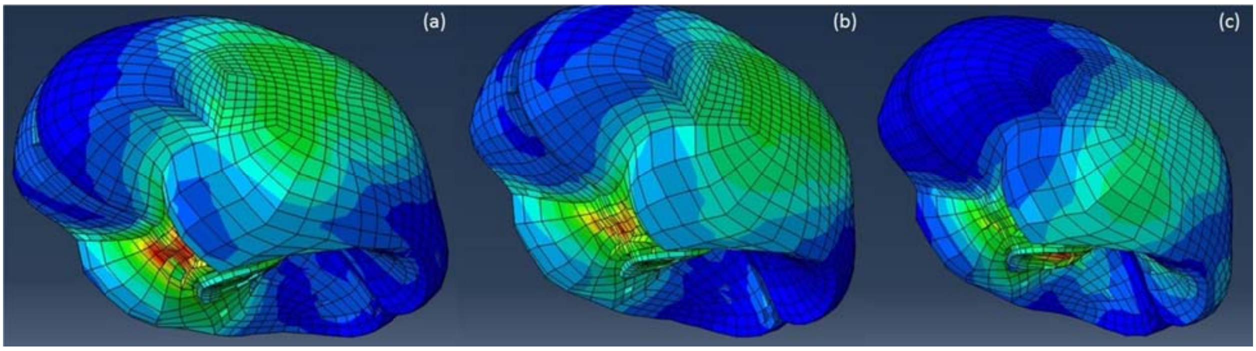

Fig. 4. Strain distributions at maximum for one case of each of the three classified injury risk groups: (a) Low risk group, (b) Concussion group, (c) TBI group.

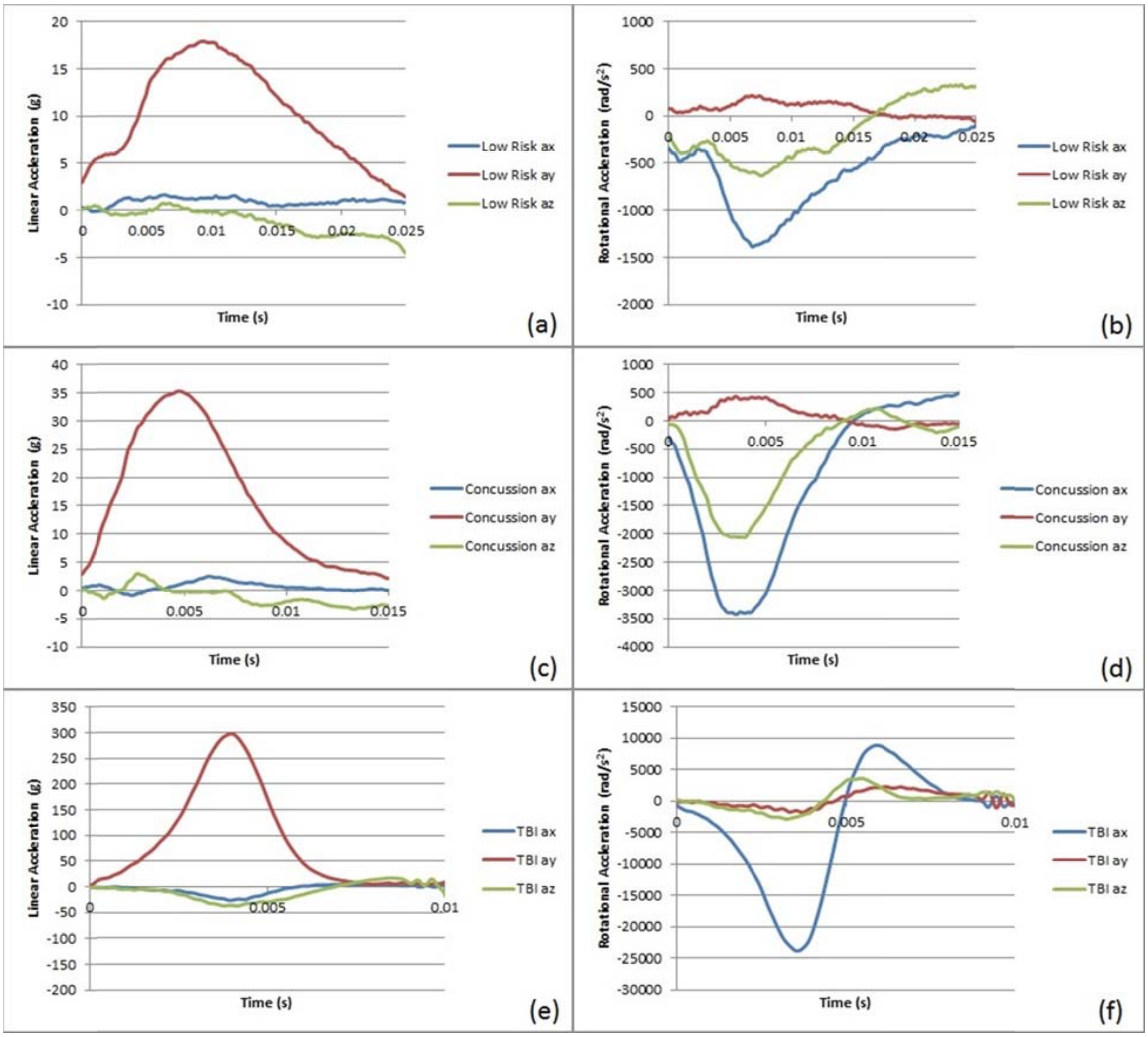

Fig. 5. Six degrees of freedom linear and rotational acceleration traces for one case of each of the three classified injury risk groups: (a) Linear acceleration traces for low risk group, (b) Rotational acceleration traces for low risk group, (c) Linear acceleration traces for concussion group, (d)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

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