The Association among Injury Metrics for Different Events in Ice Hockey Goaltender Impacts

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

Abstract Current ice hockey goaltender helmet standards use a drop test and peak linear acceleration to evaluate performance. However, ice hockey goaltenders are exposed to impacts from collisions, falls and pucks which each create unique loading conditions. As a result, the use of peak linear acceleration as a predictor for brain trauma in current ice hockey standards may not be most appropriate. The purpose of this study was to determine how kinematic response measures correlate to maximum principal strain and von Mises stress for different impact events. A NOCSAE headform was fitted with three ice hockey goaltender helmet models and impacted under conditions representing these three different impact events (fall, puck, collision). Peak resultant linear acceleration, rotational acceleration and rotational velocity of the headform were measured. Resulting accelerations were input into the University College Dublin Brain Trauma Model, which calculated maximum principal strain and von Mises stress in the cerebrum. The results demonstrated that the relationship between injury metrics in ice hockey goaltender impacts is dependent on the impact event and velocity. As a result of these changing relationships, the inclusion of finite element analysis in test protocols may provide a more practical representation of brain loading in evaluating the performance of ice hockey goaltender helmets.

Keywords concussion, goaltenders, finite element modelling, ice hockey, impact biomechanics

I. INTRODUCTION

Concussions have become a particular concern for sporting institutions, since research has indicated that multiple concussions over the period of a player's career could lead to long term disability [1]. This has been an issue in contact sports, including ice hockey, in which concussions are common [2-4]. In efforts to reduce the associated incidence of concussion, researchers have examined which kinematic variables and brain tissue metrics might be better predictors for concussion, with a view to designing better helmets [5-10]. Despite ice hockey goaltenders being the only player on the ice for the entire game, no research has yet examined these parameters for these players. There is no consensus in the literature as to whether forwards and defensemen sustain more concussions than goaltenders [4][11] or vice-versa [12].

Historically, traumatic brain injuries (TBI) were the primary concern in sports such as ice hockey, resulting in mandating the use of helmets [13]. Current ice hockey goaltender helmet standards use peak linear acceleration as the principle measure of brain trauma [14-17]. Linear acceleration has been shown to be associated with head injuries such as skull fracture and brain contusion [5][18-20]. Since the introduction of these standards, skull fractures and brain contusion have largely disappeared from ice hockey sports [2]. Rotational kinematics, as distinct from linear acceleration, have been associated more clearly, albeit indirectly, with concussion [18][21-22]. Finite element analyses of impacts to the human brain have been used to measure brain tissue level responses as this has been found to provide more meaningful information related to concussion [20-22][23-24]. However, such analyses require considerable computing power and time, although they provide insight into the relationship between kinematic variables and brain deformation measures, and can be used to develop improved standards and helmet innovation, thereby reducing the incidence of concussion [8-9][20-24].

The current standards use a drop test to establish impact absorption properties [14-17], but in addition to falls, goaltenders suffer concussions from collisions and puck impacts, in which collisions are the most common cause of concussion [3]. At present, little is known concerning the performance of the helmet under these

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impact conditions. Falls, collisions and puck impacts are defined by unique impact parameters including impact location, mass, velocity, angle of impact, and compliance of impactor [25]. Differences in these impact parameters have been shown to result in unique acceleration response curves [10][26-33], which may influence the relationship between kinematics and brain response. For instance, falls in ice hockey are characterised by the mass of the head impacting a rigid impact surface [25], resulting in high magnitude and short duration linear and rotational accelerations [6-7]. Puck impacts are characterised by low mass and high velocity impact [25], resulting in very short duration linear and rotational acceleration curves [10][34]. Finally, collisions such as shoulder collisions in ice hockey are considered highly compliant and result in low magnitude and long duration linear and rotational accelerations [35]. Subsequently, these unique acceleration response curves induce different brain tissue stresses and strains [6][23][28][31][35]. As a result, the unique loading conditions created by these events can change the relationship among kinematic response measures and maximum principal strain (MPS) and von Mises stress (VMS) [35-37]. As such, the influence of these different kinematics on MPS and VMS needs to be examined to aid the evaluation and development of goaltender helmet technologies. The purpose of this study was to determine how kinematic response measures are correlated to MPS and VMS for different events associated with ice hockey goaltender concussions.

II. METHODS

Experimental Testing

To examine how kinematic response measures are correlated to MPS and VMS, an event specific impact test protocol based on video analysis of real world ice hockey goaltender concussions was conducted [38]. The event specific impact test protocol was based on the impact parameters of 12 real world concussive events representing falls, puck impacts and shoulder collisions. The laboratory parameters used to define this protocol such as velocity, orientation and location were identified from video of real world ice hockey goaltender concussive events using Kinovea 0.8.2 video analysis software (Kinovea.org), as described by [39] and [40]. Impact velocities and orientations for each case were determined by applying a perspective grid based on known points and distances on the ice. The error within this method was estimated between 5 and 18% for velocity and 10 degrees for impact orientation [39][41]. The error within this method was estimated to be 5% for velocity and 10 degrees for impact orientation [39]. The impact velocities selected for the impact test protocol represent the lower range, mean and upper range velocities for each impact event [38]. Impact locations for each case were determined using a reference presented in Fig. 1 [38-39]. Selected impact locations for the test protocol represent those with the best coverage of impact possibilities for each impact event [38]. Table I represents the impact parameters used in the impact test protocol.

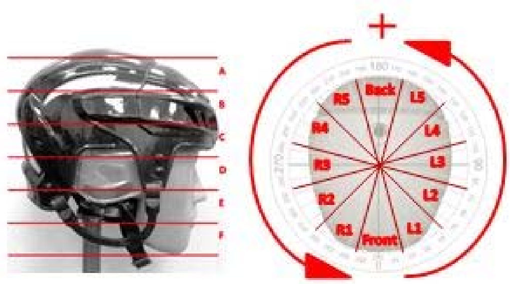

Fig. 1. Top and side view of a head illustrating the 12 sectors (each 30°) and six levels (evenly spaced) used to identify impact location [38-39].

The three impact events assessed were falls, puck impacts and collisions. Falls were simulated with the use of a helmeted headform attached to a monorail drop rig with a 60 shore A modular elastomer programmer (MEP) anvil to simulate the head impacting the ice [14-17]. A pneumatic puck launcher was used to launch pucks at a NOCSAE helmeted headform in order to reconstruct puck impacts to the head. Collisions were reconstructed with a pneumatic linear impactor fitted with a shoulder pad striker, simulating shoulder-to-head impacts [38]. Under this impact protocol three different ice hockey goaltender helmets were impacted. Three trials were completed for each impact condition, resulting in a total of 243 impacts. Peak resultant linear and rotational acceleration were obtained from the headform. Rotational velocity was determined by integrating the resulting components (x,y,z) of a rotational acceleration curve at the centre of gravity of the headform. The resulting linear and rotational acceleration curves served as input into a finite element brain trauma model, which was used to calculate the magnitude of peak MPS and peak VMS in the cerebrum.

TABLE I
CHARACTERISTICS OF THE ICE HOCKEY GOALTENDER HELMETS

Impact Event	Velocities (m/s)	Impact Location		Head Orientation	
		Level	Sector	Y-axis (°)	Z-axis (°)
Fall	3.5, 4.2, 5.0	D	Rear	-	-
		D	L4	-	-
		D	R3	-	-
Puck	29.3, 35.8, 42.3	C	R3	15	90
		E	R2	0	65
		B	R1	15	20
Collision	5.2, 7.3, 9.1	D	Front	0	0
		B	R1	0	0
		D	R3	15	45

Equipment

The monorail drop rig (Fig. 2a) consisted of a 4.7 m long rail which had a drop carriage attached to it. The drop carriage ran along ball bushings to reduce the effects of friction on the inbound velocity of the headform. A NOCSAE headform and unbiased neckform [42] were attached to a drop carriage. Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC) was used to control the velocity and release mechanisms for the impact. The velocity of the impact was measured using a photoelectric time gate.

The puck launcher and linear impactor were attached to a support/piston frame (Fig. 2b,c). The frame supported the compressed air canister, the piston and either the puck launcher or the linear impactor depending on whether a puck impact or collision was being simulated. The pneumatic piston was fired via an electronically controlled solenoid with the air supplied from the compressed air canister which either propelled a puck (0.166l g) down a barrel (0.620 ± 0.001 m) (Fig. 2c) or an impacting arm (13.1 ± 0.1kg) towards the headform. The mass of impacting arm was similar to the calculated effective mass of shoulder-to-head impacts in ice hockey [33]. The striking surface of the impacting arm consisted of a nylon disc (diameter 13.2 mm) covered with 67.79 ± 0.01mm thick layer of vinyl nitrile R338V foam and a Reebok 11k shoulder pad (Fig. 3d), found to produce a linear acceleration peak and duration similar to shoulder impacts performed by ice hockey players to Hybrid III headform at low and high velocities [33]. For both the pneumatic linear impactor and pneumatic puck launcher, the NOCSAE headform and unbiased neckform were attached to a low-friction sliding table. The sliding table had a mass of 12.78 ± 0.01 kg and allowed for movement post impact.

A medium NOCSAE headform (4.85 ± 0.01 kg) was attached to an unbiased neckform (2.11 ± 0.01 kg) [42] and used for the impact test protocol. The headform was instrumented with nine single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) in a 3-2-2-2 accelerometer array [43]. Signals from the nine accelerometers were collected at 20 KHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered with a CFC 180 filter in accordance with the SAE J211 convention. Three commercially available ice hockey goaltender helmets were tested in the event specific impact test protocol. These three helmet models were chosen as they represented a range of materials commonly used in ice hockey goaltender helmet designs. Each helmet was fitted on the headform according to manufacturer's specifications. Helmet specifications are presented in Table II.

TABLE II
CHARACTERISTICS OF THE ICE HOCKEY GOALTENDER HELMETS

Ice Hockey Goaltender Helmet	Foam Liner Material	Shell Material	Shell of Helmet (mm)	Shell + Padding (mm)
Helmet 1	Vinyl Nitrile	Polycarbonate	3.75 ± 0.23	20.93 ± 0.90
Helmet 2	Vinyl Nitrile	Fiberglass	3.55 ± 0.45	14.27 ± 1.87
Helmet 3	Vinyl Nitrile	Carbon and Kevlar Composite	3.50 ± 0.46	14.19 ± 1.39

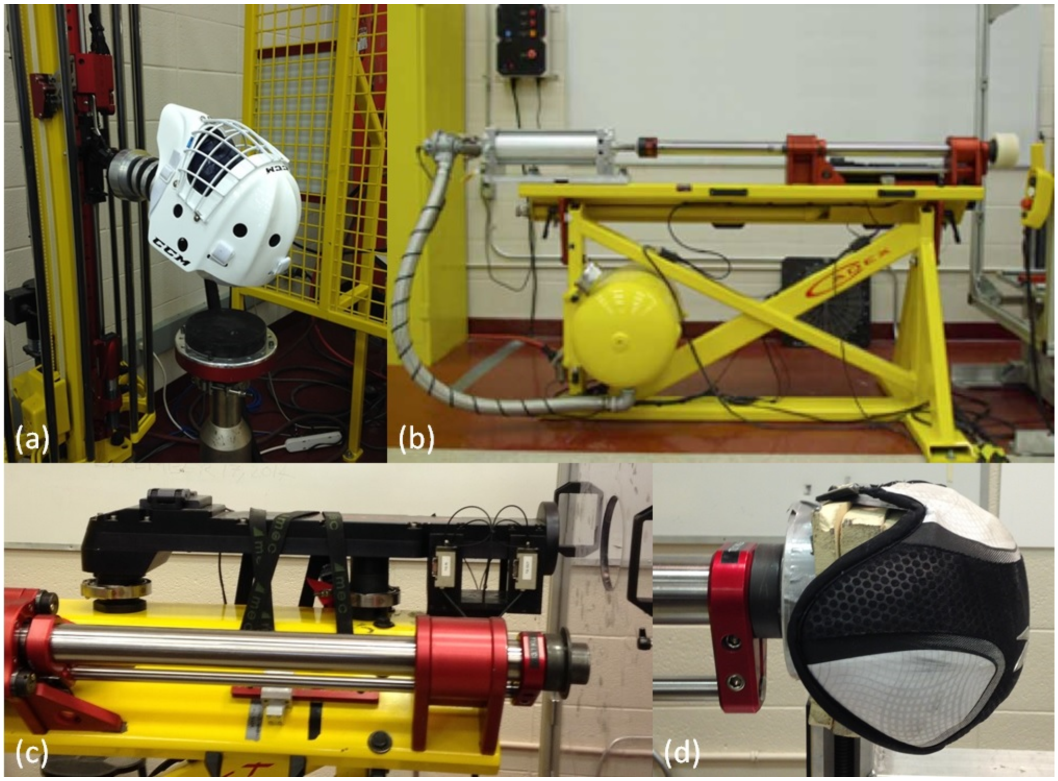

Fig. 2. Impact test equipment: (a) monorail drop rig (b) frame supporting the impacting arm, (c) barrel for puck launcher, (d) shoulder pad striker.

Computational Modelling

The University College Dublin Brain Trauma Model (UCDBTM) was the finite element model used in this study [44-45]. The geometry of the model was based on computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver [45]. The model was composed of approximately 26,000 elements representing the dura, cerebrospinal fluid (CSF), pia, falx, tentorium, grey and white matter, cerebellum and brain stem [44-45]. Validation of the model was performed against intracranial pressure response and brain motion response of previous cadaver research [46-48]. The pressure response of the model was found to match quite well with the experimental results of [46] in terms of shape and duration [44]. Additional intracranial pressure responses were compared to cadaveric research conducted by [47] which involved impacts with both linear and rotational acceleration components. Overall, the model's pressure response was found to be in good agreement with the [47]'s cadaveric pressure responses as the model's pressure has the same general shape and duration. However the magnitudes of response between the model and [47] were found to differ, especially in the case of the occipital lobe [45]. Lastly, the UCDBTM was compared to cadaveric brain motion experiments conducted by [48] and was found to produce similar brain motion traces [45]. Through such comparisons, the response of the UCDBTM was considered to produce a valid response as it showed a good correlation with cadaveric pressure responses conducted and brain motion [44-45]. Reconstructions of real world TBIs were performed to further validate the model and were found to be in good agreement with lesions on CT scans for TBI incidents [49-50].

The material behaviour of the brain tissue was modelled using a linear viscoelastic material model combined with large deformation theory [44-45][51-52]. The compressive behaviour of the brain is considered elastic and the shear characteristics of the 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 [47]. The brain shear response was modelled as hyperelastic and defined by:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-\frac{t}{0.008}} + 1003e^{-\frac{t}{0.15}}(Pa), \tag{2}$$

where C_{10} and C_{01} are the temperature-dependent material parameters [53-54] and t is the time in seconds.

To model the skull brain interface, the cerebral spinal fluid (CSF) was modelled using solid elements with low shear modulus and a high bulk modulus. The contact definitions were assigned no separation and used a friction coefficient of 0.2 [55].

Statistics

To determine how kinematic response measures are correlated to MPS and VMS, Pearson correlation coefficients (r) and r^2 values were calculated. Pearson correlation coefficients were first determined for all the data combined. Data was then separated by impact event and subsequent Pearson correlation coefficients were calculated. Data was further separated by impact velocity and Pearson correlation coefficients were calculated. Strong correlations were considered to be $|r| > 0.700$ and weak correlations were considered to be $|r| < 0.600$. The confidence interval was set at 95% and all data analyses were performed with the statistical software package of SPSS 19.0 for Windows (SPSS Inc, Chicago, IL, USA).

III. RESULTS

All Data Collapsed Together

Pearson correlations conducted on the entire dataset combine (all helmets, impact events, velocities and locations) are presented in Table V. Rotational velocity was found to be significantly ($p = 0.001$) and strongly correlated to MPS and VMS ($r > 0.700$). Significant correlations were also found for linear and rotational acceleration to MPS and VMS ($p = 0.001$). However, the strength of these correlations were lower: specifically, linear acceleration had weak correlations ($r < 0.600$) and rotational acceleration had very weak correlations ($r < 0.400$). Table VI and Fig. 4 shows the distribution of the data across kinematic response measures and MPS and VMS.

TABLE III
PEARSON CORRELATIONS OF KINEMATIC RESPONSE MEASURES
TO MAXIMUM PRINCIPAL STRAIN AND VON MISES STRESS FOR COLLAPSED DATA

Comparison	Pearson Correlation (r)	r^2
Linear Acceleration/MPS	0.553**	0.306
Rotational Acceleration/MPS	0.287**	0.082
Rotational Velocity/MPS	0.847**	0.717
Linear Acceleration/VMS	0.585**	0.342
Rotational Acceleration/VMS	0.339**	0.115
Rotational Velocity/VMS	0.795**	0.632

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

TABLE IV
KINEMATIC AND BRAIN TISSUE RESPONSE FOR DIFFERENT IMPACT EVENTS AND VELOCITIES

Impact Event	Velocity (m/s)	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Rotational Velocity (rad/s)	Maximum Principal Strain	von Mises Stress (kPa)
	3.5	60.6 (13.8)	3099 (1024)	14.7 (4.3)	0.209 (0.049)	6.5 (1.6)
Fall	4.2	85.5 (12.8)	4030 (993)	17.0 (4.7)	0.259 (0.055)	8.2 (1.8)
	5.0	159.6 (22.0)	7814 (1985)	28.2 (8.2)	0.425 (0.104)	14.1 (3.6)
	29.3	37.5 (8.3)	4046 (1191)	5.5 (1.7)	0.124 (0.040)	3.9 (1.0)
Puck	35.8	48.6 (10.4)	5630 (1849)	7.1 (1.7)	0.127 (0.024)	4.0 (0.7)
	42.3	57.7 (16.2)	6730 (2632)	8.1 (2.2)	0.145 (0.035)	4.6 (1.1)
	5.2	20.2 (3.4)	1928 (343)	23.1 (4.6)	0.204 (0.039)	5.9 (1.8)
Collision	7.3	30.9 (5.9)	2829 (606)	26.4 (5.4)	0.269 (0.071)	8.2 (3.0)
	9.1	37.7 (8.1)	3356 (623)	27.6 (6.5)	0.306 (0.093)	9.3 (3.8)

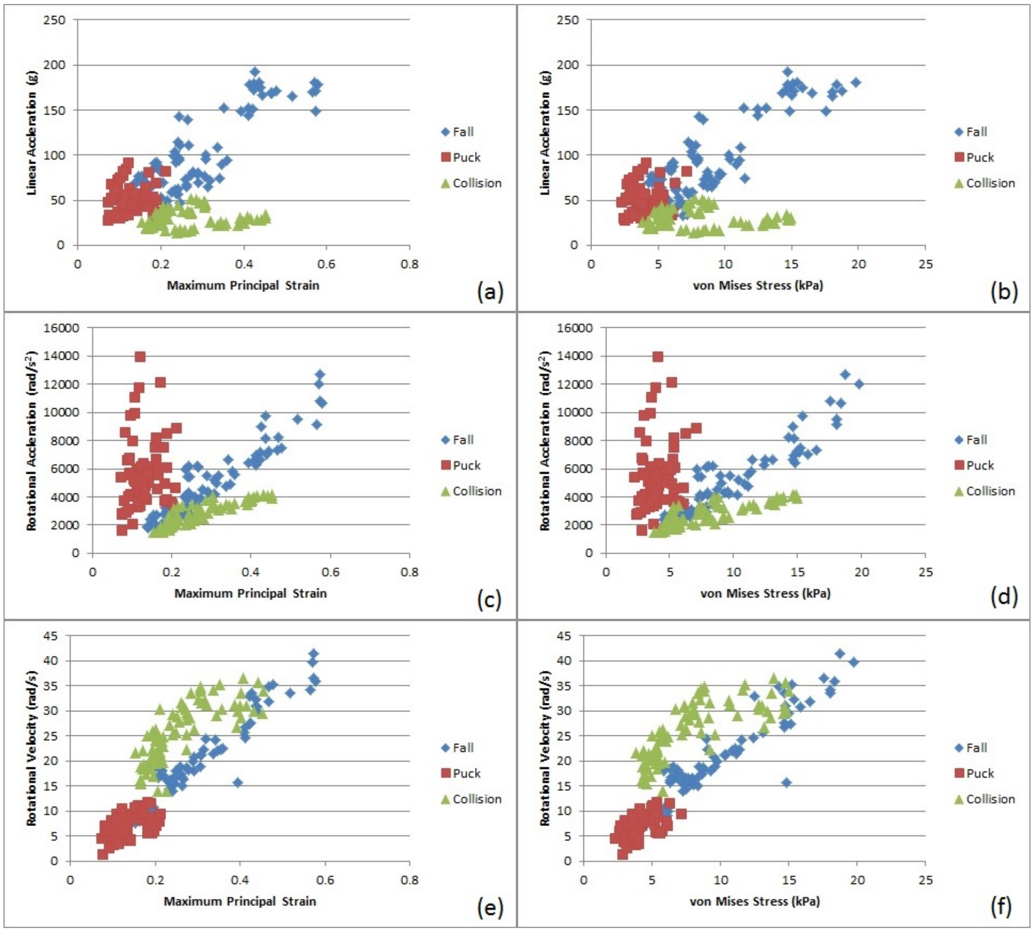

Fig. 3. Scatterplot for all impact events: (a) maximum principal strain (MPS) versus linear acceleration, (b) von Mises stress (VMS) versus linear acceleration, (c) MPS versus rotational acceleration, (d) VMS versus rotational acceleration, (e) MPS versus rotational velocity, (f) VMS versus rotational velocity.

Separated by Impact Event

The effect on correlation between kinematic response measures and MPS and VMS for different impacts when all velocities are considered together is shown in Table V. Falls showed all kinematic response measures had significant and very strong correlations with MPS and VMS ($r > 0.800$). For puck impacts only rotational velocity was significantly correlated with MPS and VMS but these correlations were weak ($r < 0.600$). In examining collisions, rotational acceleration and velocity were found to have significant and strong correlations with MPS and VMS ($r > 0.700$). Linear acceleration had no significant correlation to MPS or VMS for collisions.

TABLE V

PEARSON CORRELATIONS OF KINEMATIC RESPONSE MEASURES TO MAXIMUM PRINCIPAL STRAIN AND VON MISES STRESS FOR DIFFERENT IMPACT EVENTS

Comparison	Fall		Puck		Collision	
	R	r^2	r	r^2	r	r^2
Linear Acceleration/MPS	0.830**	0.689	0.000	0.000	0.110	0.012
Rotational Acceleration/MPS	0.936**	0.876	0.105	0.011	0.877**	0.769
Rotational Velocity/MPS	0.952**	0.906	0.488**	0.238	0.784**	0.615
Linear Acceleration/VMS	0.857**	0.734	0.054	0.003	-0.015	0.000
Rotational Acceleration/VMS	0.931**	0.867	0.156	0.024	0.829**	0.687
Rotational Velocity/VMS	0.931**	0.867	0.546**	0.298	0.768**	0.590

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

Separated by Impact Event and Velocity

Tables VI – VIII presents the effect on correlations between kinematic response measures and MPS and VMS for different impact events when the effect of increasing velocity is eliminated. For falls, all correlations were significant ($p < 0.05$) except for linear acceleration between MPS and VMS at 3.5 m/s and 4.2 m/s. Rotational acceleration and velocity had strong correlations to MPS and VMS ($r > 0.700$). Across all velocities, puck impacts only had significant correlations between rotational velocity measures and MPS and VMS ($p < 0.05$), which were found to be weak ($r < 0.600$). Collisions showed all kinematic response measures had significant correlations between kinematic response measures and MPS and VMS ($p < 0.05$) across all velocities. Linear acceleration was found to be negatively correlated to MPS and VMS, whereas rotational acceleration and velocity had strong positive correlations to MPS and VMS ($r < 0.700$).

TABLE VI
PEARSON CORRELATIONS OF KINEMATIC RESPONSE MEASURES TO MAXIMUM PRINCIPAL STRAIN AND VON MISES STRESS FOR DIFFERENT FALL VELOCITIES

Comparison	3.5 m/s		4.2 m/s		5.0 m/s	
	r	r^2	r	r^2	r	r^2
Linear Acceleration/MPS	-0.121	0.015	0.030	0.001	0.717**	0.514
Rotational Acceleration/MPS	0.775**	0.601	0.834**	0.696	0.831**	0.691
Rotational Velocity/MPS	0.813**	0.660	0.943**	0.889	0.923**	0.852
Linear Acceleration/VMS	-0.057	0.003	0.224	0.050	0.803**	0.645
Rotational Acceleration/VMS	0.770**	0.593	0.840**	0.706	0.803**	0.645
Rotational Velocity/VMS	0.759**	0.576	0.891**	0.794	0.877**	0.769

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

TABLE VII
PEARSON CORRELATIONS OF KINEMATIC RESPONSE MEASURES TO MAXIMUM PRINCIPAL STRAIN AND VON MISES STRESS FOR DIFFERENT PUCK VELOCITIES

Comparison	29.3 m/s		35.8 m/s		42.3 m/s	
	r	r^2	r	r^2	r	r^2
Linear Acceleration/MPS	-0.228	0.052	-0.236	0.056	-0.132	0.017
Rotational Acceleration/MPS	0.093	0.009	-0.159	0.025	0.001	0.000
Rotational Velocity/MPS	0.453*	0.205	0.451*	0.203	0.469*	0.220
Linear Acceleration/VMS	-0.252	0.063	-0.245	0.060	-0.056	0.003
Rotational Acceleration/VMS	0.063	0.004	-0.121	0.015	0.067	0.004
Rotational Velocity/VMS	0.453*	0.205	0.507**	0.257	0.545**	0.297

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

TABLE VIII
PEARSON CORRELATIONS OF KINEMATIC RESPONSE MEASURES TO MAXIMUM PRINCIPAL STRAIN AND VON MISES STRESS FOR DIFFERENT COLLISION VELOCITIES

Comparison	5.2 m/s		7.3 m/s		9.1 m/s	
	r	r^2	r	r^2	r	r^2
Linear Acceleration/MPS	-0.589**	0.347	-0.607**	0.368	-0.467*	0.218
Rotational Acceleration/MPS	0.804**	0.646	0.898**	0.806	0.839**	0.704
Rotational Velocity/MPS	0.718**	0.516	0.812**	0.659	0.752**	0.566
Linear Acceleration/VMS	-0.638**	0.407	-0.637**	0.406	-0.540**	0.292
Rotational Acceleration/VMS	0.794**	0.630	0.890**	0.792	0.812**	0.659
Rotational Velocity/VMS	0.702**	0.493	0.798**	0.637	0.712**	0.507

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

IV. DISCUSSION

Combined Data

The purpose of this study was to describe the association between kinematic response measures and MPS and VMS for different events associated with ice hockey goaltender concussions. When all impact events were collapsed together, only rotational velocity showed strong correlations to MPS and VMS. This is due to the influence of impact events on correlations between kinematic response and brain tissue response, as observed in this study and previous research [35-37]. Each impact event created a unique acceleration curve, which affects the magnitude of kinematic response measures and brain tissue response [10][31-33][35][37]. Falls result in high magnitude and short duration linear and rotational acceleration [6-7] which are reported to produce high rotational velocities and brain stresses and strain in ice hockey goaltender impacts [38]. Collision impacts in ice hockey, on the other hand, result in low magnitude and long duration acceleration curves [33] which lead to high rotational velocities and brain stresses and strain [38]. Finally, puck impacts are characteried by short duration acceleration curves [10][34] and lead to low rotational velocities and brain stresses and strains [38]. Such differences create situations in which both low and high magnitude linear and rotational accelerations can be coupled with high brain stresses and strains. Additionally, high magnitude peak linear and rotational accelerations can be coupled with low brain stresses and strains. Whereas, magnitudes of rotational velocity tend to reflect brain stress and strain levels (Fig. 3). This results in a stronger correlation between rotational velocity and brain tissue response than acceleration response when all impact events are considered together. Similar correlations have been found by [22], when examining correlations between kinematic response measures and brain strain for a large variety of impact events in American football collected by the Head Impact Telemetry (HIT) System.

Falls

Falls were found to produce the highest correlations among kinematic response measures and brain tissue response when compared with puck impacts and collisions. Previous research examining falls where patients suffered TBIs and persistent post-concussive syndrome (PCS) also showed a significant positive correlation among injury metrics [32][35]. However, the strong correlations observed in this study were found to be a result of increases in energy. As the velocity of the impact increases, the magnitude of kinematic response and brain tissue response increases, which has been shown to result in strong correlations among linear and rotational acceleration and MPS [56]. As seen in this study, linear acceleration is not correlated to MPS and VMS for falls at 3.5 and 4.2 m/s, while rotational kinematics maintains strong correlations. Similar results have been found in examinations using different finite element brain trauma models in which MPS and VMS are strongly correlated to rotational kinematics but have lower correlations with linear acceleration [20][23][57-58]. This demonstrates at lower energy levels that rotational kinematics are more effective at representing brain stresses and strain than linear acceleration [8-9][18][20][23][37]. When high energy falls at 5.0 m/s were assessed, all kinematic response measures had strong correlations to MPS and VMS. These strong correlations observed are likely due to high energy levels, resulting in both high magnitude head and brain responses [36-37]. In a group of impacts classified as a risk of TBI, the group was found to consist of high energy impacts which were associated with high magnitude responses and, as a result, had high correlations [37]. These would suggest that, for high energy falls in ice hockey, a reduction in any kinematic variable would reflect a decrease in MPS and VMS, but for lower energy falls likely only reductions in rotational kinematics will reflect a reduction in MPS and VMS. Therefore, solely using peak linear acceleration as a pass-fail metric for drop tests linked to TBI [14-17] may be an appropriate measure to evaluate ice hockey goaltender helmets, but this is not the case of concussions.

Puck Impacts

In contrast to falls, the correlation between kinematic response measures and MPS and VMS were much lower for puck impacts. Only rotational velocity was found to have a correlation with MPS and VMS, however these correlations were rather poor. These low correlations between kinematic response and brain tissue response are likely attributed to the design of ice hockey goaltender helmets. Ice hockey goaltender helmets are designed with a thick and stiff shell which deflects most of the energy for a puck impact [34]. Such behaviour has been found to mitigate the effect of velocity for MPS and VMS across a 29.3 m/s to 42.3 m/s range [38]. However, kinematic response measures were found to increase with increasing velocity [38]. These results would suggest that, although kinematic magnitudes may increase with increasing velocity, the ice hockey goaltender helmets deflects the puck in a manner which changes the kinematic response but the brain experiences similar stresses and strains. Creating a situation in which the range in response of MPS and VMS is much smaller than the response of kinematic variables. This leads to low to no correlation between the kinematic response and brain tissue response. In addition, it was observed that velocity had a minimal effect on the relationship between kinematic response and brain tissue response, as these correlations remained poor. This suggests that kinematic response measures do not correlate with MPS and VMS for puck impacts to ice hockey goaltender helmets. Additionally, these relationships suggest puck impacts are less likely to cause injury, which has been reflected in the low rate of injury from puck impacts in ice hockey [2-4].

Collisions

Strong relationships were found between rotational kinematics and MPS and VMS for collisions. When all collisions were considered together, the brain tissue response showed a strong correlation with rotational kinematics but no correlation to linear acceleration. Similar relationships have also been reported for collisions using different finite element models of the brain [20][23][57-58] highlighting the importance of measuring rotational kinematics to assess brain trauma [18-21]. However it should be noted that in Fig. 3e rotational velocity reaches a plateau which may represent a limitation in using rotational velocity to predict MPS for severe shoulder collisions. Rotational velocity likely reaches a plateau as for high severity collisions because the area under the rotational acceleration may not change. However, the shape of the rotational acceleration curve may change. Changes in the acceleration curve shape have been shown to influence brain stresses and strains [31]. As a result, MPS can continue to increase despite rotational velocity reaching a plateau. As a result MPS had a slightly lower correlation with rotational velocity than rotational acceleration. A 75% probability of sustaining a concussion has been reported between 27.7 and 30.8 rad/s [59-60] and therefore this plateau also represents an upper range in which a concussion is more than likely. Additionally, in Fig 3e collisions were generally associated with larger rotational velocities for a given level of MPS than falls. This is a result of falls and collisions creating different rotational acceleration curves [6-7][38][43] in which the area under the rotational acceleration curve, as represented by rotational velocity, must be greater in collisions than falls to produce a given level of strain within the brain. It was also observed that these correlations between kinematics response measures and brain stresses and strain were affected by increases in velocity. When correlations were conducted on each collision velocity separately, rotational kinematics remained strongly correlated to MPS and VMS, however linear acceleration was found to be negatively correlated to MPS and VMS. These results are likely a reflection of the fact that rotational motions and not linear motions cause shear strains in the brain tissue that result in concussion [18][26-27] and this is reflected by the responses in FE models [21-22][61]. In addition, the impact sites that resulted from the video analysis created conditions that created higher rotations and less linear translations. This kind of impact scenario would create a situation where the MPS would follow the rotational acceleration responses, and not the linear responses. Since the linear responses would not be as affected by this kind of impact, the correlations would be low, or negative. Therefore, this study would suggest due to the influence of impact event and velocity on the relationships between kinematic response and brain tissue response, the use of finite element analysis in test procedures maybe a more practical method to quantify brain loading in the evaluation of ice hockey goaltender helmets.

Limitations

The NOCSAE headform may not imitate the dynamic properties of a human head, but it does produce results that are within those expected for cadaveric impacts [62]. This headform is widely accepted and used as a human head surrogate in the certification of football and lacrosse helmets. The neck constraint forces of the unbiased neckform remained constant throughout all impacts. In real world events the neck constraint forces acting on the head for the different scenarios considered in this study could be different, which may affect the response of the head. The response of the UCDBTM is meant to be a representation of how the brain may respond. Assumptions are made surrounding the boundary conditions and material properties of the model cadaveric and other anatomical testing. As such, this may not reflect the exact motion of the brain. Analysis using the UCDBTM was limited to the cerebrum as the brain stem has not yet been validated. The event specific impact test protocol used in this study was based on real world concussive events which occurred in professional ice hockey goaltenders and may not represent impacts of other age groups and skill levels. However, by limiting the development of protocol to professional ice hockey goaltenders it allows for access to injury reports and high quality game film for video analysis.

V. CONCLUSIONS

This study examined how kinematic response measures are correlated to MPS and VMS for different concussive impact events sustained by ice hockey goaltenders. The results demonstrated that the relationship between kinematic response measures and MPS and VMS is dependent on the type of impact event as well as the impact velocity. Falls showed strong relationships between kinematic response measures and brain tissue response when velocity conditions were considered together. However, each velocity was analysed separately, MPS and VMS were only correlated to linear acceleration for the highest fall velocity. For puck impacts, on the other hand, only rotational velocity showed low correlations to MPS and VMS. When assessing collisions, only rotational kinematics were related to MPS and VMS; however, when separated by velocity, linear acceleration was negativity correlated to MPS and VMS. As a result of these changing relationships, test protocols may have a greater benefit to include the use of finite element analysis to quantify the load which the brain experiences when evaluating the performance of ice hockey goaltender helmets.

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