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Protective Capacity of Ice Hockey Goaltender Helmets for Three Events Associated with Concussion

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Abstract

The purpose of this study was to assess the protective capacity of an ice hockey goaltender helmet for three concussive impact events. A helmeted and unhelmeted headform was used to test three common impact events in ice hockey (fall, puck impacts and shoulder collisions). Peak linear acceleration, rotational acceleration and rotational velocity as well as maximum principal strain and von Mises stress were measured for each impact condition. The results demonstrated the tested ice hockey goaltender helmet was well designed to manage fall and puck impacts but does not consistently protect against shoulder collisions and an opportunity may exist to improve helmet designs to better protect goaltenders from shoulder collisions.

Key Words: Brain injury, Impact biomechanics, Dynamic response, Finite element modeling, Brain tissue response

Introduction

In ice hockey, concussions are one of the most common injuries sustained by athletes across all levels of play and age groups (Kelly et al. 1991; Cantu 1996; Goodman et al. 2001; Pashby et al. 2001; Powell 2001; Marchie and Cusimano 2003; Flik et al. 2005; Agel et al. 2007). Protective headgear and helmets have decreased the risk of traumatic brain injury (TBI) in sports (Wennberg and Tator 2003). As such, skull fractures and other TBIs have largely disappeared from sports, however concussion incidents remain common (Wennberg and Tator 2003).

Ice hockey poses a high risk for concussion due to high speeds of players and pucks, physical contact and rigid objects (Delaney 2004). Players in ice hockey can skate at speeds up to 48 km/h (30 mph) and reach gliding speeds of up to 24 km/h (15 mph) (Sim and Chao 1977; Bancroft 1993; Goodman et al. 2001). This combination of player movement and permissible physical contact can create hazardous collision scenarios. Goaltenders are primarily tasked with intercepting goalward-bound pucks, and in the process can face pucks travelling up to 193 km/h (120 mph). Given the mass of the puck (0.17 kg), this equates to 2522 N of force at when the puck reaches the goal (Sim and Chao 1977). As a result of this environment, ice hockey goaltenders have suffered concussions from collisions with players, falls to the ice and puck impacts in which player-to-player collisions are the most common cause of concussion (LaPrade et al. 2009). Research has shown shoulder collisions, falls and puck impacts to the head in ice hockey create unique impact conditions that influence the direction and magnitude of the head kinematic response and subsequent brain tissue stresses and strains (Kendall et al. 2012, 2014; Clark 2015; Ouckama and Pearsall 2014; Rousseau 2014; Rousseau et al. 2014; Nur et al. 2015; Post et al. 2012; Rousseau and Hoshizaki 2015; Smith et al. 2015). Unique impact parameters such as impact site, mass, velocity, angle of impact, and compliance of impactor create different impact loading conditions for shoulder collisions, falls, and puck impacts (Hoshizaki et al. 2014). Differences in loading conditions would suggest that helmet designs should be designed to account for impact conditions that produce high magnitude accelerations with short impact durations and low magnitude accelerations with long durations. At present, it is unknown how an ice hockey goaltender helmet functions to protect the head and brain under these different types of loading conditions.

Considerable research has been conducted in an effort to decrease the incidence of concussion; however, this research has focused on skaters (Hoshizaki et al. 2012; Walsh et al. 2012; Post et al. 2013, 2014; Ouckama and Pearsall 2014), with little research focusing on goaltenders (Nur et al. 2015). Presumably, research has focused on ice hockey skaters as annual concussion rates for ice hockey forwards and defensemen have between reported between 4 and 73 concussions per year (Mölsä et al. 1997; Benson et al. 2011; Hutchison et al. 2015a). Annual concussion rates for goaltenders have been reported between 1 and 5 concussions per year which occurred as a result of falls, player-to-player collisions and puck impacts (Mölsä et al. 1997; LaPrade et al. 2009; Benson et al. 2011; Hutchison et al. 2015a). Concussion has been reported as the second most prevalent injury in National Collegiate Athletic Association (NCAA) intercollegiate ice hockey goaltenders with an incidence of 1.7 per year (LaPrade et al. 2009). The design of skater and goaltender helmets in ice hockey differ in attempts to manage the unique risk of each particular activity. Skater's helmets in ice hockey consist of a plastic shell of polyethylene and either vinyl nitrile (VN) or expanded polypropylene (EPP) foam liners. Ice hockey goaltender helmets on the other hand are typically made of a stiffer shell material such as carbon and Kevlar composite, fibreglass or polycarbonate and have a VN foam liner. With a view to inform helmet safer designs, this paper seeks to understand how an ice hockey goaltender helmet performs under representative injury conditions.

The purpose of this study was to assess the protective capacity of an ice hockey goaltender helmet for the three most common impact events associated with concussion. It was hypothesised that an ice hockey goaltender helmet would reduce peak head kinematic and brain tissue response for falls and puck impacts to levels that are below the reported ranges associated with head and brain injury (Newman et al. 2000; Zhang et al. 2004; Doorly and Gilchrist 2006; Kleiven 2007; Rowson et al. 2012; Post, Hoshizaki, et al. 2015, Post, Kendall, et al. 2015) but that it would not do so for shoulder collisions.

Methods

Equipment

The National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform with a mass of 4.85 ± 0.01 kg was attached to an unbiased neckform (Walsh and Hoshizaki 2012) with a mass of 2.11 ± 0.01 kg and used for all impact conditions. The mass of and dimensions of the unbiased neckform were designed to match that of the Hybrid III neckform. The unbiased neckform had four centred and unarticulated rubber butyl disks of radius 68.0 mm and height 21.5 mm. The disks sit slightly recessed and serially inside aluminium disks measuring 85.6 mm in radius and 12.8 mm in height. The unbiased was designed to respond symmetrical in all axes and therefore did not bias the results of the impact. Nine calibrated single-axis Endevco7264C-2KTZ-2–300 accelerometers (Endevco, San Juan Capistrano, CA) were fixed in an orthogonal position on a block near the centre of gravity of the headform in a ‘3–2-2–2’ accelerometer array developed to measure and calculate three-dimensional motion during an impact (Padgaonkar et al. 1975). Hardware anti-alias filtering was conducted with a low pass 4-pole Butterworth filter at 40 kHz. Signals from the nine accelerometers were collected at 20 kHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered through a 300 Hz low pass Butterworth filter using a SAE J211 Class 180 protocol (SAE, 2007).

Three ice hockey goaltender helmets of one commercially available model (one helmet for each impact event) were used for all helmeted conditions (Figure 1). The helmets were fitted on the headform according to manufacturer's specifications. The shell of the helmets were made of fiberglass and had a carbon steel cage attached. The liner of the helmets consisted of vinyl nitrile (VN) foam. Thickness measurements of the shell and liner were taken from four locations on the helmets. The thickness of the shell was 3.55 ± 0.45 mm and the thickness of the shell plus liner was 14.27 ± 1.87 mm.

Figure 1. Ice hockey goaltender mask used for all helmeted conditions.

A monorail drop rig equipped with a 60 shore A modular elastomer programmer (MEP) anvil was used to simulate falls to the ice (International Organization for Standardization 2003; Canadian Standards Association 2009a, 2009b; American Society for Testing and Materials 2014). The MEP anvil was chosen to simulate the rigidity of the head impacting the ice. The monorail drop rig consisted of a drop carriage which ran along a 4.7 m long rail on ball bushings to reduce the effects of friction on the inbound velocity of the headform. The NOCSAE headform and unbiased neckform were attached to a drop carriage and the desired inbound velocity was entered into Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC). The carriage was raised to the drop height and then dropped. This allowed for the fall conditions to simulate a head and neck being fixed to the torso. A photoelectric time gate placed within 0.02 m of the impact measured the inbound velocity.

A pneumatic linear impactor fitted with a shoulder pad striker was used to reconstruct shoulder collisions (Rousseau and Hoshizaki 2015). The pneumatic linear impactor consisted of three major components: the support/piston frame, the impacting arm and the table housing the NOCSAE headform. The frame supported the impacting arm, the compressed air canister and the piston. The impacting arm (13.1 ± 0.1 kg) was accelerated towards the headform using compressed air. The mass of the impacting arm was calculated for shoulderto- head impacts in ice hockey reconstructions (Rousseau and Hoshizaki 2015). The striking surface consisted of a nylon disc (diameter 13.2 mm) covered with a 142 mm thick layer of vinyl nitrile R338 V foam and a Reebok 11 k shoulder pad was attached to the end of the impact arm. This striker was found to produce a linear acceleration peak and duration similar to that of shoulder impacts performed by ice hockey players to Hybrid III headform at low and high velocities (Rousseau and Hoshizaki 2015).

Puck-to-head impacts were reconstructed using a pneumatic puck launcher. A commercially available puck (0.166 l g) was fired using compressed air down a 0.6200 ± 0.0005 m barrel before it was released from the puck launcher. The mass of the puck was chosen as this is the standard mass for a puck specified within the rules of ice hockey. Using the puck launcher allowed the puck to be free and behave like a projectile in a similar manner to which the puck would behave when taking a shot in ice hockey.

When the pneumatic linear impactor or pneumatic puck launcher was used, the NOCSAE headform was supported by a low-friction sliding table (16.6 ± 0.1 kg). This mass was selected in order to account for the effective torso mass of a struck player in a collision which is typically less than the striking mass (Viano and Pellman 2005; Viano et al. 2007). The table allowed for movement post impact and was stopped using a foam braking system.

A movable locking base on the table was used to attach the headform. This movable locking base allowed the headform to be oriented in five degrees of freedom: fore-aft (x), lateral (y) and up-down (z) translation, as well as fore-aft (y) and axial rotation (z) and remained fixed in position during testing.

Procedure

In order to assess the protective capacity of an ice hockey goaltender helmet, a helmeted and an unhelmeted NOCSAE headform was impacted by three events associated with concussion. The impact protocol used in this study was based on video analysis of real world concussive events occurring to professional ice hockey goaltenders. A search was conducted through professional ice hockey injury reports for diagnosed concussions sustained by goaltenders occurring between the 2000–2001 and 2013–2014 seasons. Cases included in the development of the protocol were those in which a concussion was diagnosed by a medical doctor: a video of the incident was available and the video was of sufficient quality to identify impact parameters. In addition, the concussive events were the result of a fall, puck impact or player-to-player collision. Concussive events which are a result of multiple impacts to the head were excluded from the development of the impact protocol. Forty-three cases were examined in this research and 12 cases were found to be suitable to the development of an impact test protocol. Video analysis was performed using Kinovea 0.8.20 (open source, kinovea.org) in order to determine impact parameters such as velocity, orientation and location, as described by Rousseau (2014) and Post et al. (2014). Impact velocity and orientation were determined using a scaling reference established on the playing surface based on known points and distances (Figure 2). The velocities selected for this study were taken for the case which was determined to have the lowest velocity for each impact event in order to avoid damaging the equipment when impacting an unhelmeted headform. Impact locations for each case were determined using a reference presented in Figure 3. The impact locations selected for this study were those which represented the best coverage of impact possibilities for each individual impact event. A summary of the locations and velocities used in this study are presented in Table 1 and Figure 3. Three trials were completed for each impact condition and peak resultant linear and rotational accelerations of the headform were obtained. Rotational velocity was also determined by integrating the resulting rotational acceleration. The head kinematic response from the impacts served as input into a finite element model to determine von Mises stress (VMS) and maximum principal strain (MPS), within the brain tissue. Both VMS and MPS are common measures used in brain injury biomechanics to quantify the response of brain tissue to an impact (Willinger and Baumgartner 2003; Zhang et al. 2004; Kleiven 2007; Post, Hoshizaki, et al. 2015, Post, Kendall, et al. 2015). Von Mises stress takes the sum of all the tensors involved in the stress of a structure (tensile, compressive and shear stresses) and gives a result in uniaxial stress as one value measured in units of pressure (Da Silva 2005). The equation used to calculate VMS was as follows:

$$\sigma = \sqrt{0.6\left[\left(\sigma_x - \sigma_y\right)^2 + \left(\sigma_y - \sigma_z\right)^2 + (\sigma_z - \sigma_x)^2\right]} + \sqrt{+3\left(\tau_{xy}^2 + \tau_{yz}^2 + \tau_{zx}^2\right)} \qquad [1]$$

where σ is the normal stress in a given axis and τ is the shear stress in a given axis and is measured in Pascals (Pa). Maximum principal strain is described by tissue elongation relative to its original length along one principle axes (Da Silva 2005). The following equation was used to calculate MPS:

$$\varepsilon 1{,}2 = \frac{\varepsilon_x + \varepsilon_y + \varepsilon_z}{3} \pm \sqrt{\left(\varepsilon_x - \varepsilon_y\right)^2 + \left(\varepsilon_y - \varepsilon_z\right)^2 + \left(\varepsilon_y - \varepsilon_z\right)^2} \qquad [2]$$

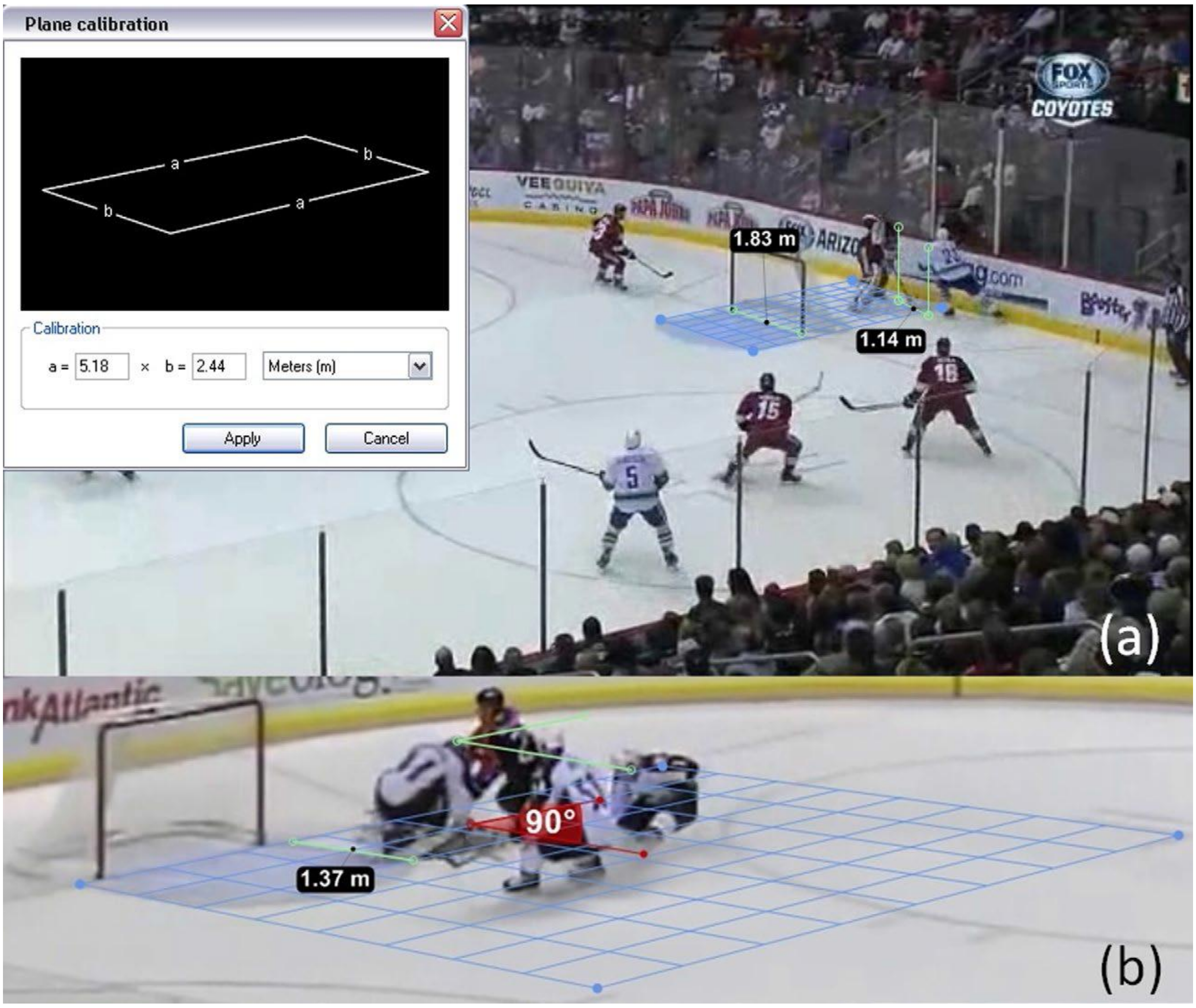

Figure 2. Examples of a perspective grid calibration used in ice hockey to determine: (a) velocity and (b) orientation.

Table 1: Impact location and velocities selected for impact protocol as determined by video analysis of real world concussive events.

Mechanism	Velocity (m/s)	Location		Head Rotation	
		Sector	Level	Y-axis	Z-axis
		Rear	D	-	-
Fall	3.5	L4	D	-	-
		R3	D	-	-
		R3	C	15°	90°
Collison	5.2	R2	E	0°	65°
		R1	B	15°	20°
		Front	D	0°	0°
Puck	29.3	R1	B	0°	0°
		R3	D	15°	45°

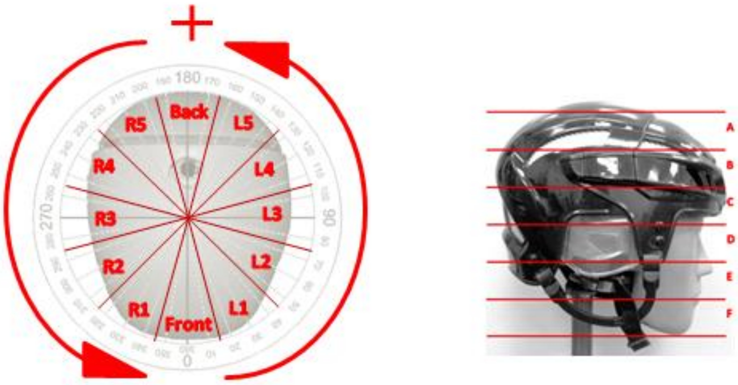

Figure 3. Top and side view of a head illustrating the 12 sectors (each 30°) and six levels (evenly spaced) used to identify impact location (Rousseau 2014).

Figure 4. Impact location of an event specific impact test protocol for ice hockey goaltender masks (Clark 2015): (a) fall Rear-D, (b) fall L4-D, (c) fall R3-D, (d) puck impact Front-D, (e)

puck impact R1-B, (f) puck impact R3-D, (g) collision R3-C, (h) collision R2-E, (i) collisions R1-B.

Finite Element Model

Before the resulting acceleration response was input into the finite brain trauma model a rotational matrix was applied. The accelerometer block of the NOCSAE headform was oriented on a 25° with respect to the y-axis about the centre of gravity. To align the acceleration response of the NOCSAE headform with the frame of reference used by the finite brain trauma model, the linear and rotational acceleration responses at the centre of gravity were rotated using the following rotation matrix:

$$R_y(\theta) = \begin{bmatrix} \cos(\theta) & 0 & \sin(\theta) \\ 0 & 1 & 0 \\ -\sin(\theta) & 0 & \cos(\theta) \end{bmatrix} \qquad [3]$$

where θ is the angle between the frame of reference used by the NOCSAE headform and the finite element brain trauma model. The rotation matrix described in Equation (3) was validated for head impacts using impact sites from the uOTP5 (Walsh et al. 2012) at 5 m/s. It was demonstrated that applying the rotation matrix to acceleration- time histories from an accelerometer block oriented at 25° produced mean linear and rotational acceleration- time histories which remained within the 95% confidence interval of acceleration-time histories from a 0° accelerometer block orientation with only small periods in which the mean acceleration-time histories differed. These differences were seen on the magnitude of ± 7 g and ± 700 rad/s2. As a result, the rotated linear and rotational acceleration-time histories consisted to have similar loading curves and the rotational matrix described in Equation (1) was considered valid for use in head impact biomechanics research.

The finite element model used in this study was the University College Dublin Brain Trauma Model (UCDBTM) (Horgan and Gilchrist 2003, 2004).The head geometry of the UCDBTM was extracted from computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver (Horgan and Gilchrist 2003). The model of the head and brain includes the scalp, skull, pia, falx, tentorium, CSF, grey and white matter, cerebellum and brain stem consists of 26, 000 elements (Horgan and Gilchrist 2003, 2004). The material characteristics of the model were taken from the literature (Ruan 1994; Willinger et al. 1995; Zhou et al. 1995; Kleiven and von Holst 2002).

Model validation was performed against intracranial pressure response and brain motion response of previous cadaver research (Nahum et al. 1977; Trosseille et al. 1992; Hardy et al. 2001). The shape of the response and the duration of the pressure response were found to match quite well with the experimental results of Nahum et al. (1977) (Horgan and Gilchrist 2003). Intracranial pressure response was also compared to experiments conducted by Trosseille et al. (1992) which involved impacts with both rotational and translational acceleration components. The general shape and duration of the model's pressure response was found to agree with the cadaveric pressure responses of Trosseille et al. (1992), however the magnitudes were found to differ, especially in the case of the occipital lobe (Horgan and Gilchrist 2004). Finally, the model was compared to the brain motion research conducted by Hardy et al. (2001) and found to have similar brain motion traces (Horgan and Gilchrist 2004). As such, the correlation of the model's response to cadaveric pressure responses conducted and brain motion research was found to be good and the model was considered to be validated (Horgan and Gilchrist 2003, 2004). Further comparisons were conducted by Doorly and Gilchrist (2006) and Post, Hoshizaki et al. (2015) using reconstructions of real world traumatic brain injury incidents with results that were in agreement with lesions on CT scans and values reported in the literature from falls.

Analysis

To assess the protective capacity of an ice hockey goaltender helmet, helmeted and unhelmeted impacts were compared in order to determine if the helmet was able to reduce head kinematic response and brain stress and strain. In order to make comparisons, the data was separated by impact event and location conditions. This created 9 distinct groups in which 3 measurements were taken while the headform was wearing an ice hockey goaltender helmet and 3 without a helmet. Independent sample t-tests were then used to compare helmeted and unhelmeted conditions for each group. The probability of making a type 1 error was set at $\alpha = 0.05$ for all comparisons. The statistical software package of SPSS 19.0 for Windows (SPSS Inc, Chicago, IL, USA) was used for all data analyses. Further comparisons were between helmeted and unhelmeted conditions to determine if the significant reduction resulted in head kinematic response or brain stresses and strains which were below reported ranges of TBIs and concussion in the literature (Newman et al. 2000; Zhang et al. 2004; Doorly and Gilchrist 2006; Kleiven 2007; Rowson et al. 2012; Post, Hoshizaki, et al. 2015, Post, Kendall, et al. 2015). The proposed probability of sustaining a concussion values for linear acceleration, rotational acceleration, MPS and VMS, by Zhang et al. (2004) and rotational velocity values from Rowson et al. (2012) are presented in Table 3. These values are used to distinguish between low and high proposed probabilities of sustaining a concussion.

Results

A description of the 12 cases used in the development of an event specific impact test protocol for ice hockey goaltender helmets as are presented in Table 2. The protective capacity of the tested ice hockey goaltender helmet to reduce peak linear and rotational acceleration, rotational velocity, VMS and MPS for different impact events are presented in Figures 5–9 and Table 4. All helmeted falls were found to produce significantly lower peak linear and rotational acceleration, rotational velocity, VMS and MPS than unhelmeted falls. Puck impacts to a helmeted headform produced significantly lower peak linear and rotational acceleration, rotational velocity, VMS and MPS values compared to puck impacts to an unhelmeted headform. Shoulder collisions to a helmeted headform resulted in significantly lower peak linear acceleration, rotational velocity, VMS and MPS compared to shoulder collisions to a bare headform for impacts to location R1-B. Peak linear acceleration, rotational velocity and VMS were significantly lower for helmeted shoulder collisions compared to unhelmeted collisions at location R3-C. There was no significant difference between helmeted and unhelmeted shoulder collisions to R3-C for peak rotational acceleration and MPS. Helmeted shoulder collisions to impact location R2-E produced significantly lower rotational acceleration and velocity, MPS and VMS compared to unhelmeted shoulder collisions, whereas linear acceleration did not significantly differ.

Table 2: Case description of the 12 concussive events used in the development of an event specific impact test protocol for ice hockey goaltender helmets.

Case	**Anthropometrics**		**Mechanism**	**Location**		**Head Rotation**		**Velocity (m/s)**
	Height (m)	**Mass (kg)**		**Sector**	**Level**	**Y-axis**	**Z-axis**	
1	1.91	100	Fall	R5	D	-	-	-
2	1.88	92	Fall	L4	D	-	-	5.0
3	1.80	88	Fall	Rear	D	-	-	3.5
4	1.80	84	Puck	R3	D	15°	45°	-
5	1.93	98	Puck	Front	D	0°	0°	29.3
6	1.91	113	Puck	R1	B	0°	0°	42.3
7	1.88	82	Collisions	R3	B	90°	90°	9.1
8	1.88	92	Collisions	L2	E	0°	297°	8.8
9	1.93	98	Collisions	R3	C	15°	90°	5.7
10	1.91	97	Collisions	L3	B	15°	265°	8.0
11	1.96	84	Collisions	L3	A	45°	258°	5.2
12	1.91	98	Collisions	R1	B	15°	20°	5.2

Table 3: Proposed values for the probability of sustaining a concussion by Zhang, Yang and King (2004) and Rowson, Duma, Beckwith, Chu, Greenwald, Crisco, Brolinson, Duhaime, McAllister and Maerlender (2012).

Variable	Probability of Sustaining a Concussion			Reference
	25 %	**50 %**	**80 %**	
Linear Acceleration (g)	66	82	106	Zhang et al. (2004)
Rotational Acceleration (rad/s^2)	4600	5900	7900	Zhang et al. (2004)
Rotational Velocity (rad/s)	25.8	28.3	30.8	Rowson et al. (2012)
Maximum Principal Strain	0.14	0.19	0.24	Zhang et al. (2004)
Von Mises Stress (kPa)	6	7.8	10	Zhang et al. (2004)

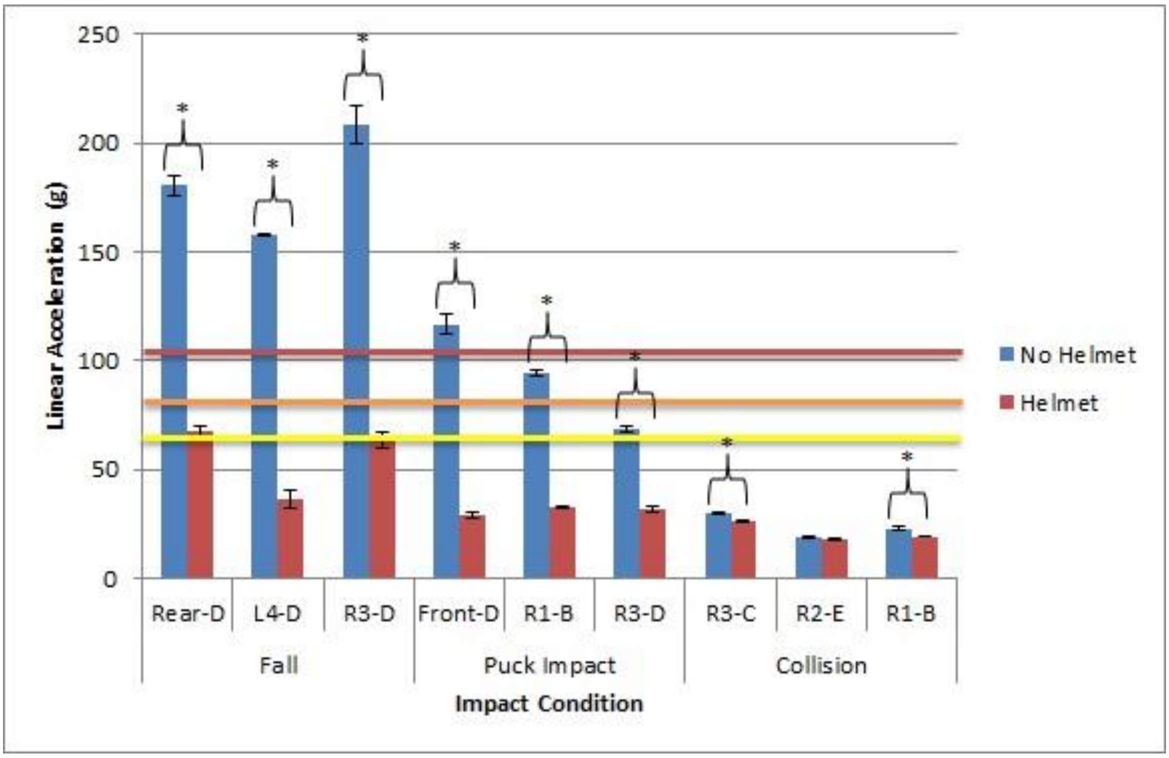

Figure 5. Mean peak resultant linear acceleration for helmeted and unhelmeted impacts of each event. The three lines represent a 25 (yellow), 50 (orange) and 80% (red) probability of sustaining a concussion (Zhang et al. 2004).

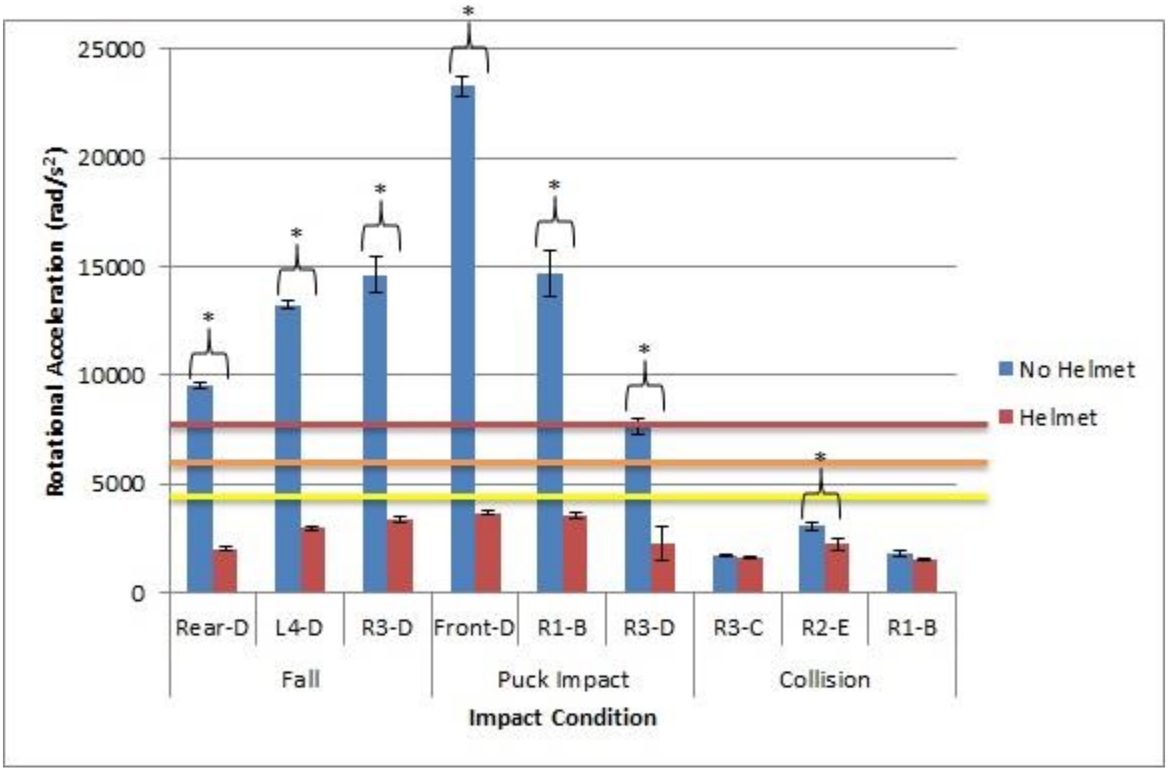

Figure 6. Mean peak resultant rotational accelerations for helmeted and unhelmeted impacts of each event. The three lines represent a 25 (yellow), 50 (orange) and 80% (red) probability of sustaining a concussion (Zhang et al. 2004).

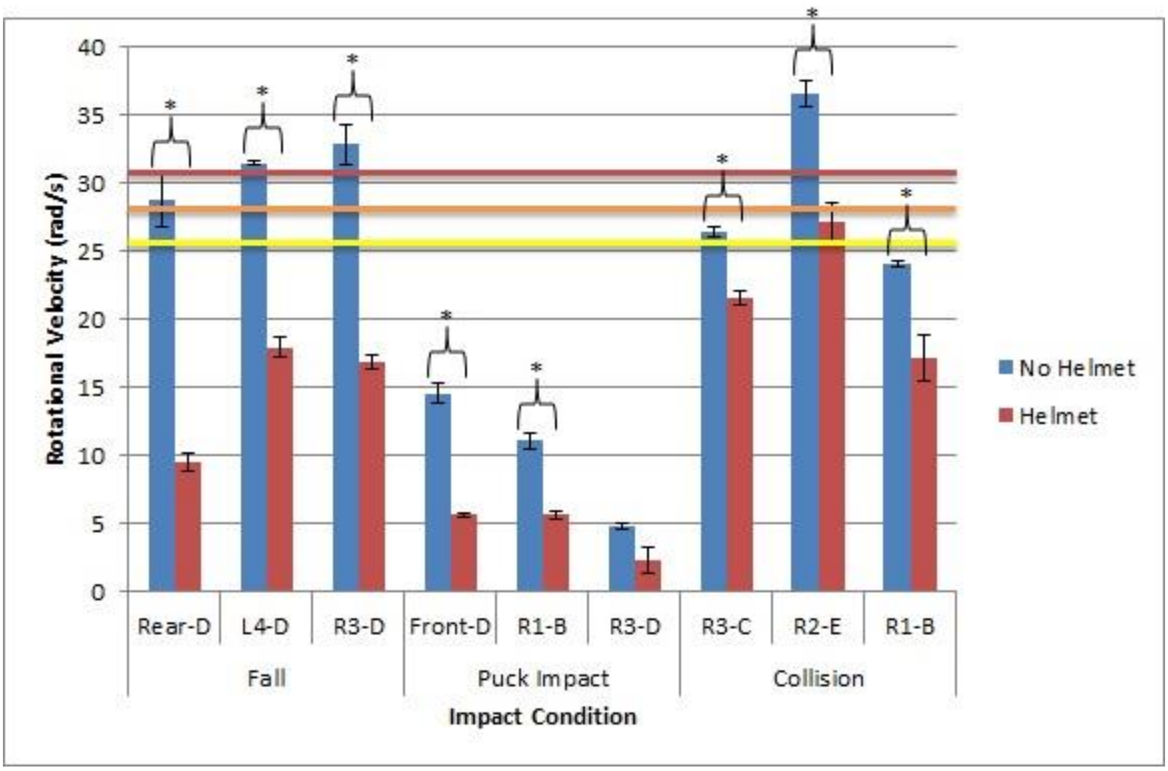

Figure 7. Mean peak rotational velocities for helmeted and unhelmeted impacts of each event. The three lines represent a 25% (yellow), 50% (orange) and 80% (red) probability of sustaining a concussion (Rowson et al. 2012).

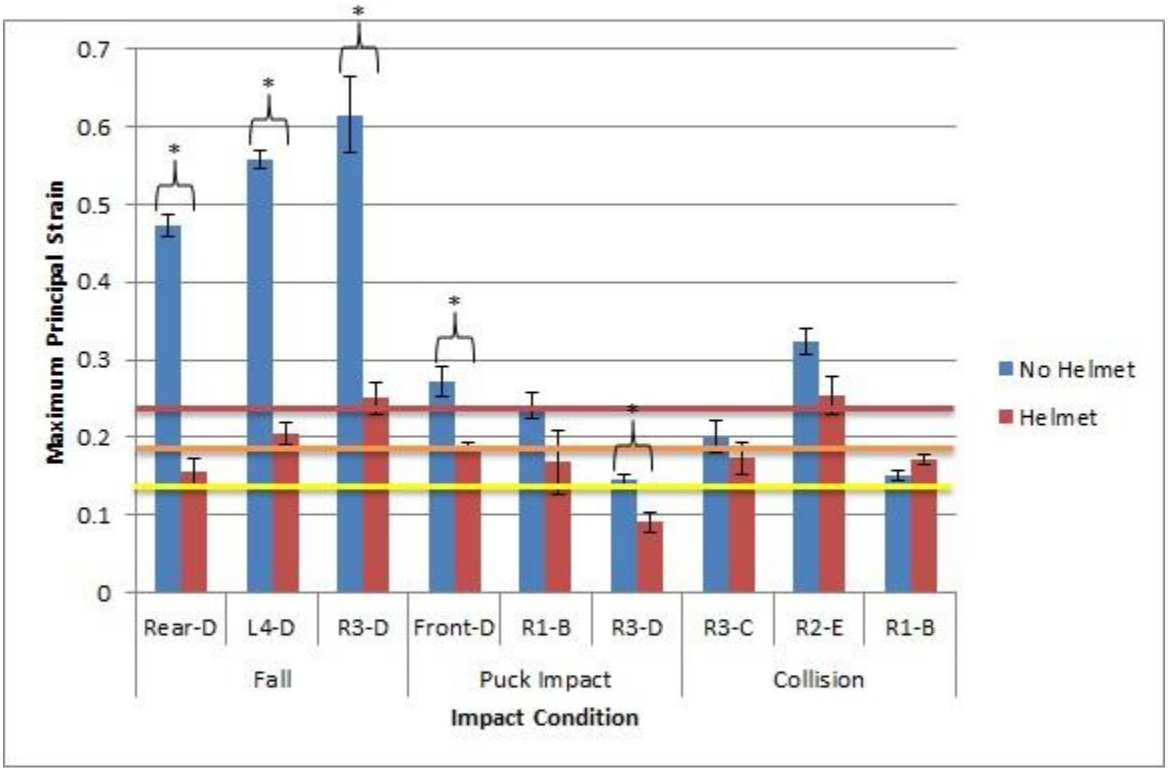

Figure 8. Mean peak maximum principle strain for helmeted and unhelmeted impacts of each event. The three lines represent a 25% (yellow), 50% (orange) and 80% (red) probability of sustaining a concussion (Zhang et al. 2004).

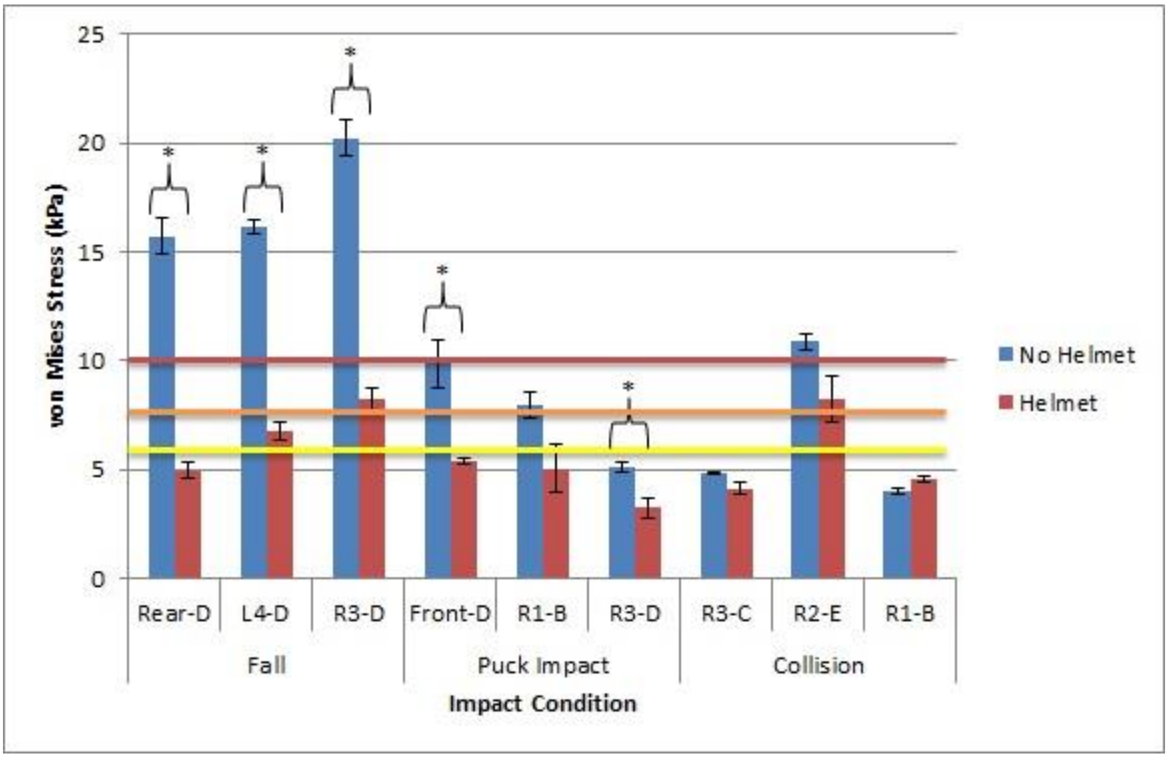

Figure 9. Mean peak von Mises stress for helmeted and unhelmeted impacts of each event. The three lines represent a 25% (yellow), 50% (orange) and 80% (red) probability of sustaining a concussion (Zhang et al. 2004).

Table 4: Percentage reduction between unhelmeted and helmet impacts.

Mechanism	Location	Linear Acceleration (%)	Rotational Acceleration (%)	Rotational Velocity (%)	Maximum Principle Strain (%)	Von Mises Stress (%)
Fall	Rear-D	62.6	78.7	67.1	67.2	68.6
	L4-D	77.1	77.7	43.0	63.1	58.0
	R3-D	69.5	77.2	48.4	59.3	59.4
Puck	Front-D	75.2	84.2	61.1	30.2	45.5
	R1-B	64.8	75.8	49.1	30.2	36.6
	R3-D	53.6	70.6	52.1	38.0	36.2
Collison	R3-C	12.7	5.8	18.6	13.8	14.7
	R2-E	2.5	27.2	25.9	21.5	24.1
	R1-B	15.8	14.3	28.9	-13.8	-12.6

Discussion

The purpose of this study was to assess the protective characteristics of an ice hockey goaltender helmet for three events associated with concussion. When comparing no-helmet to helmet conditions, significant differences were found for most conditions; however the degree to which wearing a helmet reflected a decrease in head kinematic and brain tissue response varied. The tested ice hockey goaltender helmet significantly reduced head kinematic response as well as brain stress and strain measures in the fall and puck impact conditions but not consistently for shoulder collisions and the percent reduction was smaller (Table 4). These reductions in head kinematic response and brain stress and strain were found to be below reported ranges of concussion and TBI in American football and patients admitted to hospital for falls and puck impacts but not for some of the shoulder collision conditions (Newman et al. 2000; Zhang et al. 2004; Kleiven 2007; Rowson et al. 2012; Post, Hoshizaki, et al. 2015, Post, Kendall, et al. 2015). These results demonstrate the tested ice hockey goaltender helmet was well designed to decrease head kinematic and brain tissue response from falls and puck impacts but may not provide adequate protection from shoulder collisions in some conditions.

Falls

The tested ice hockey goaltender helmet significantly reduced head kinematic response as well as brain stress and strain in the simulated fall conditions. These reductions also represented a decrease in head kinematic and brain tissue response from above to below levels of reported concussion and TBI for injurious cases reconstructed from American football and patients admitted to hospital (Newman et al. 2000; Zhang et al. 2004; Kleiven 2007; Rowson et al. 2012; Post, Hoshizaki, et al. 2015, Post, Kendall, et al. 2015). As such, the findings of this study support that the tested ice hockey goaltender helmet can decrease head kinematic and brain tissue response from falls to hard surfaces such as ice. This is reflected in the low incidence of concussions in ice hockey due to falls (7%) when considering examining skaters and goaltenders together (Hutchison et al. 2015a).

Puck Impacts

When examining the results for puck impacts, the tested ice hockey goaltender helmet reduced head kinematic response and brain stress and strain measures. These reductions reflect a decrease in head kinematic and brain tissue response below levels reported to cause brain injury in American football and patients admitted to the hospital (Newman et al. 2000; Zhang et al. 2004; Doorly and Gilchrist 2006; Kleiven 2007). This may help to explain the low incidence of concussion attributed to puck impacts in ice hockey (LaPrade et al. 2009; Hutchison et al. 2015b). Ice hockey goaltender helmets have a thick and stiff shell. The stiff shell design of ice hockey goaltender helmets acts to deflect most of the energy from a puck impact (Nur et al. 2015). This suggests that for puck impacts, having a stiff shell is a desirable design for ice hockey goaltender helmets.

Shoulder Collisions

The tested ice hockey goaltender helmet was found to be less effective at reducing head kinematic response and brain stress and strain measures relative to the other two impact scenarios. Despite significant reductions in most conditions between helmeted and unhelmeted impacts, the differences were very small in magnitude and percent reduction. Unhelmeted and helmeted shoulder collisions resulted in peak linear and rotational acceleration below reported ranges of concussion in American football (Newman et al. 2000; Zhang et al. 2004). These low acceleration values compared to those reported for concussive American football impacts and the other two impact scenarios is due to the higher level of compliance for shoulder collisions. High compliance impacts result in low magnitude accelerations but long impact durations (Gilchrist 2003; Rousseau 2014; Clark et al. 2016; Kendall 2016). Long duration impacts have been suggested to cause high rotational velocities and high brain stresses and strains (Willinger et al. 1994; Gilchrist 2003; Clark 2015). When examining the results for rotational velocity, MPS and von Mises stress, shoulder collisions represented values within the range reported of concussion (Newman et al. 2000; Zhang et al. 2004; Kleiven 2007; Rowson et al. 2012) and the tested ice hockey goaltender mask was unable to reduce these values below this range for some conditions. The limited ability of the tested ice hockey goaltender helmet to effectively reduce head kinematic and brain tissue response for shoulder collisions found in this study may reflect the high compliance of shoulder collisions compared to falls and puck impacts. For impacts with compliant surfaces, the energy that is not transmitted to the head is attenuated by the both the helmet and the impacting surface (Hunt and Mills 1989; Forero Rueda 2009; Clark et al. 2016). As a result, for high compliant impacts such as shoulder collisions, most of the energy is attenuated by the shoulder and very little energy is attenuated by the helmet. As such the tested ice hockey goaltender helmet is less effective at reducing head kinematic and brain tissue response during shoulder collisions. This limited protection from shoulder collisions could explain why 58% of concussions sustained by ice hockey goaltender concussions are due to player-to-player collisions (LaPrade et al. 2009) and highlights the importance of protecting ice hockey goaltenders from player-to-player collisions (Liberman and Mulder 2007). As such, an increased emphasis on enforcement of rules to protect ice hockey goaltenders from player-to-player collisions and/or more severe penalties for players, who commit such infractions, may prove to be more effective in reducing the rate of concussions. Additionally, the inclusion of a shoulder collision simulation in ice hockey goaltender helmet standards could aid to improve goaltender helmet designs from this kind of loading scenario.

Limitations

The present study and its results should be considered according to its limitations. The NOCSAE headform may not imitate the dynamic properties of a human head (Seemann et al. 1986; Deng 1989). However, despite such limitations, the NOCSAE headform is widely accepted and used as a human head surrogate to certify football and lacrosse helmets. Additionally, the unbiased neckform may not imitate the exact response of the human neck. The unbiased neckform was modelled after the Hybrid III neckform but unlike the Hybrid III neckform, the unbiased neckform is designed to response symmetrical in all axes and remove any bias in the results. A single ice hockey goaltender helmet which represents a commonly worn design was tested (Nur et al. 2015) as a representation of how ice hockey goaltender helmets may commonly perform in real world situations. Differences in shell and liner design between helmets can cause differences in helmet performance, however these performance differences are small in magnitude (Rousseau et al. 2009a, 2009b; Ouckama and Pearsall 2014; Nur et al. 2015). The UCDBTM makes assumptions surrounding material characteristics which are based on cadaveric and other anatomical testing. As such, the response of the model UCDBTM is meant to be representative of how the brain may respond and may not represent the exact motion of the brain. Further the UCDBTM is a partially validated model and as a result the analysis was limited to the cerebrum. The error associated with using Kinovea 0.8.2 video analysis software (Kinovea.org), has been estimated to be between 5 and 18% for velocity and 10° for impact orientation (Post et al. in revision, Rousseau 2014). This range of measurement variation is comparable to those reported by Newman et al. (2005) and McIntosh et al. (2000) (11.3 and 13% respectively). The impact velocities selected in this study represented the lowest velocity analyzed to result in concussion. These velocities were selected as to avoid damaging the equipment when impacting an unhelmeted headform. As such, the low velocity impact may result in a response which is represented as a low risk of concussion. The impact testing protocol used for this study was based on real world concussive events which occurred solely in elite ice hockey goaltenders and as such may not represent the protective capacity of ice hockey goaltender helmets for other age levels. However by solely selecting elite ice hockey goaltenders it allows for access to injury reports and high quality game film for video analysis.

Conclusion

Concussions in ice hockey can occur as a result of a number of impact events. This study investigated the protective capacity of an ice hockey goaltender helmet for three events associated with concussion (fall, shoulder collision, puck impacts). When comparing helmeted and unhelmeted impacts, the tested ice hockey goaltender helmet significantly reduced head kinematic response and brain stress and strain measures for falls and puck impacts but not always for shoulder collisions. Overall this study demonstrates the tested ice hockey goaltender helmet was well designed to protect against falls and puck impacts but an opportunity may exist to improve goaltender helmet designs to provide better protection from shoulder collisions.

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Disclosure statement

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