Comparison of Ice Hockey Goaltender Helmets for Concussion Type Impacts

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Abstract

Concussions are among the most common injuries sustained by ice hockey goaltenders and can result from collisions, falls and puck impacts. However, ice hockey goaltender helmet certification standards solely involve drop tests to a rigid surface. This study examined how the design characteristics of different ice hockey goaltender helmets affect head kinematics and brain strain for the three most common impact events associated with concussion for goaltenders. A NOCSAE headform was impacted under conditions representing falls, puck impacts and shoulder collisions while wearing three different types of ice hockey goaltender helmet models. Resulting linear and rotational acceleration as well as maximum principal strain were measured for each impact condition. The results indicate that a thick liner and stiff shell material are desirable design characteristics for falls and puck impacts to reduce head kinematic and brain tissue responses. However for collisions, the shoulder being more compliant than the materials of the helmet causes insufficient compression of the helmet materials and minimizing any potential performance differences. This suggests that current ice hockey goaltender helmets can be optimized for protection against falls and puck impacts. However, given collisions are the leading cause of concussion for ice hockey goaltenders and the tested helmets provided little to no protection, a clear opportunity exists to design new goaltender helmets which can better protect ice hockey goaltenders from collisions.

Key Terms: Brain injury, Ice hockey, Helmet, Impact biomechanics, Finite element modeling

Introduction

Goaltender masks were introduced into ice hockey to reduce the risk of facial fracture.[17] As technology advanced, the goaltender mask evolved into a helmet/cage combination, rather than a single piece full fiberglass mask. Today, ice hockey goaltender helmets are typically made with a cage of carbon, steel, or titanium, a helmet shell of carbon and Kevlar composite, fiberglass, or polycarbonate and an energy absorbing liner consisting of vinyl nitrile (VN) foam. Ice hockey goaltender helmet designs have progressed to the point where traumatic brain injuries (TBI) have become infrequent in the sport of ice hockey, however concussions remain a common injury.[21, 30, 63] In the National Collegiate Athletic Association (NCAA) concussions have been reported to be the second most common injury sustained by ice hockey goaltenders with an incidence of 1.7 per year.[30] The performance of ice hockey goaltender helmets has been evaluated using certification standards that primarily involve drop tests using measures of peak linear acceleration to establish impact attenuation properties.[1, 6, 7, 22] This has resulted in ice hockey goaltender helmets using materials which are designed to protect against falls. In addition to falls, ice hockey goaltenders can face pucks exceeding 100 mph (161 km/h) and collisions with players traveling at speeds up to 30 mph (48 km/h), which pose a high risk of injury.[60] As a result, it is important that ice hockey goaltender helmets use materials that are not only designed to protect against falls but also against puck impacts and collisions.

Ice hockey goaltenders can suffer concussions from falls, puck impacts and collisions, of which collisions are the leading cause of concussion.[30] Falls, puck impacts, and collisions in ice hockey create unique loading conditions which are applied to the head and brain.[18, 23-25, 37, 38, 53, 58] Differences in impact loading conditions have been shown to affect the protective capabilities of helmets.[3, 5, 10] It is important to gain an understanding of how different materials used in helmets perform under multiple events associated with concussion in order to improve helmet design. Ice hockey goaltender helmets have been compared for their protection from puck impacts as measured by head kinematics,[37] however it is currently unknown how different designs and materials used ice hockey goaltender helmets affect head kinematics and brain response for other impact events associated with concussion. Research in ice hockey helmets has shown that design characteristics such as external shell geometry, shell and liner material, and liner thickness can influence kinematic and brain tissue response.[19, 38, 47, 56, 59] Similar research for goaltender helmets is lacking, and if performed would provide useful information for future design considerations to improve protection. As a result, the purpose of this study was to examine how liner thickness and shell material of ice hockey goaltender helmets would affect the head kinematics and brain strain for the three most common impact events associated with concussion in ice hockey.

Materials and Methods

Procedure

The ice hockey goaltender helmets were impacted under conditions representing falls, puck impacts and collisions. The impact protocol used in this study was based on video analysis of 12 real world ice hockey goaltender concussions.[3, 4] The videos used for this protocol were those in which the event was of sufficient quality to identify impact parameters and the goaltender was diagnosed with a concussion by a medical doctor. These concussive events were the result of a fall, puck, or collision. Video analysis was performed using Kinovea 0.8.2 video analysis software (Kinovea.org), as described by Post, Karton, Hoshizaki and Gilchrist 47, Rousseau 53 and Clark, Post, Hoshizaki and Gilchrist 4 to determine impact parameters such as velocity, orientation and location. A perspective grid based on known points and distances on the ice was applied to determine impact velocities and orientations for each concussive case. The perspective grid allowed for the measurement of distances and angles on the playing surface. Velocity was determined by measuring the distance between the struck player's head and the impacting surface five frames prior to impact (Fig. 1a). Impact orientation was measured by the angle between the struck player's head position and the inbound impactor (Fig. 1b). The error associated with this method was estimated between 5 and 18% for velocity and 10 degrees for impact orientation.[48, 53] The error was determined by measuring skating velocity obtained from Kinovea compared to high speed video.[48] The velocities selected for the event specific impact protocol were taken from the cases which were determined to have the lowest and highest velocity for each impact event. The mean velocity for each impact event was also selected. These velocities were selected for the event specific impact protocol to represent the energy levels associated with each impact event. These velocities are presented in Table 1.[4] Impact locations for each case was determined according to the reference presented in Figure 2.[3, 53] The locations selected for the impact protocol were those which represented the best coverage of possible impact for each event and are shown in Figure 3.[4]

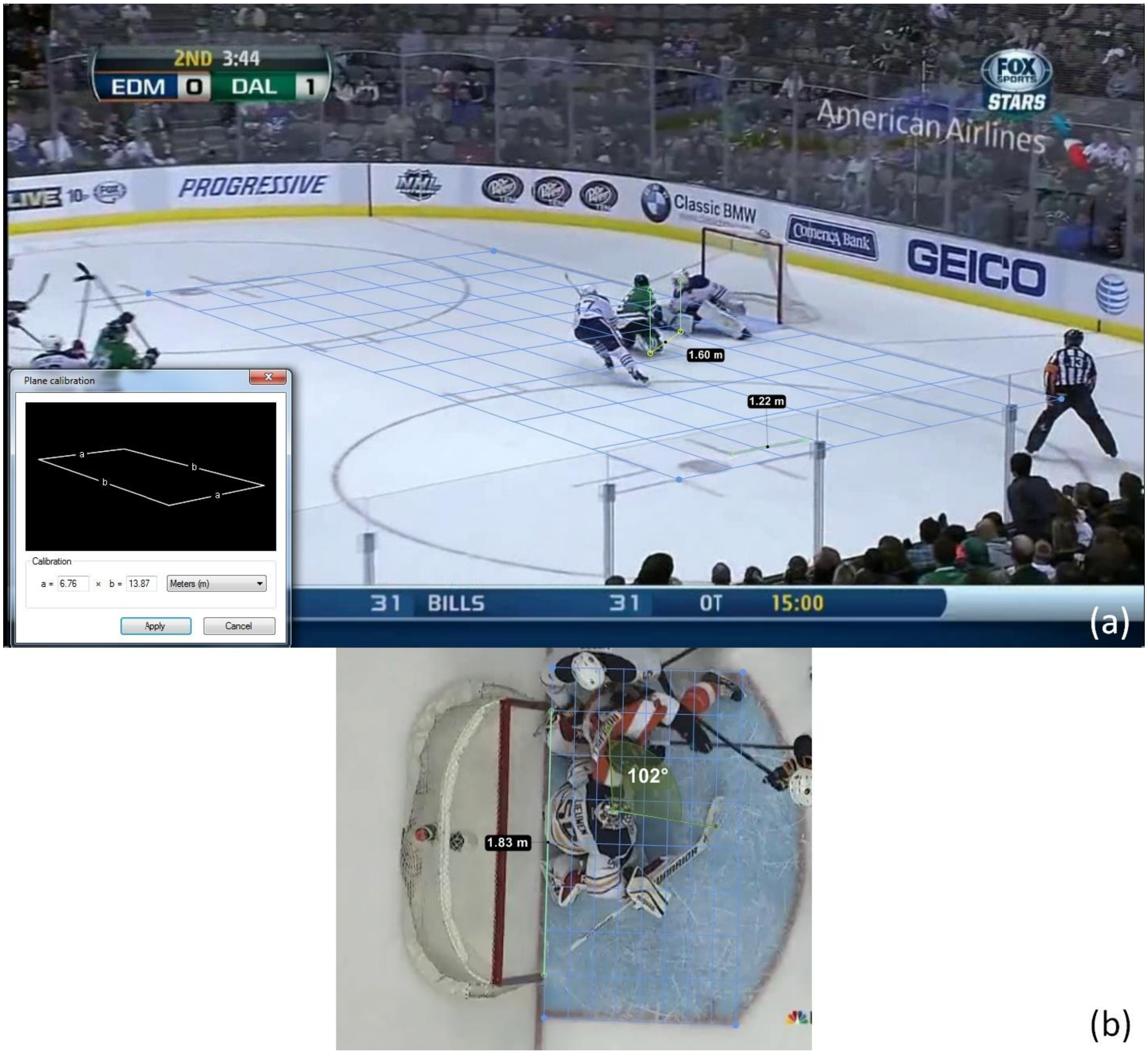

Figure 1. Examples of a perspective grid calibration used in ice hockey to determine: (a) velocity and (b) orientation.

Table 1: Impact velocities used in event specific impact test protocol for ice hockey goaltender helmets determined from video analysis of real world ice hockey goaltender concussion.[4]

Impact Event	**Velocities (m/s)**		
	Lower	**Mean**	**Upper**
Fall	3.5	4.2	5.0
Puck	29.3	35.8	42.3
Collisions	5.2	7.3	9.1

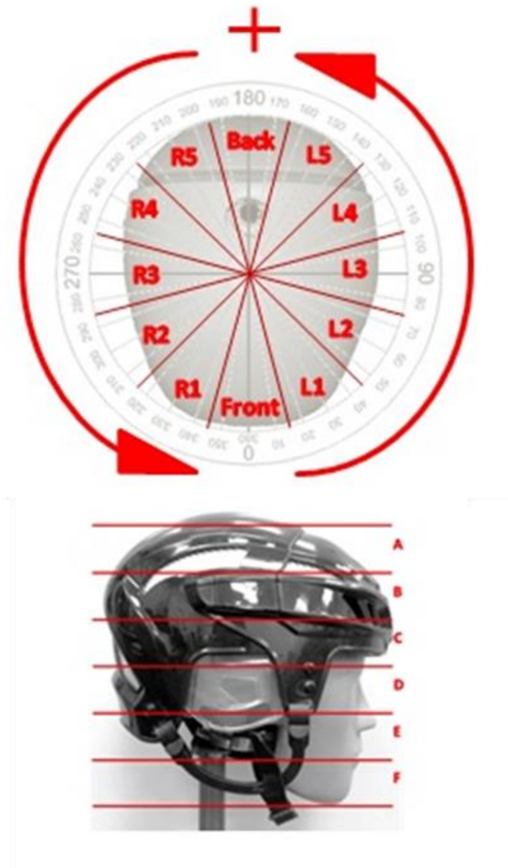

Figure 2. Top and side view of the head illustrating the 12 sectors (each 30°) and six levels (evenly spaced) used to identify impact location.[3, 53]

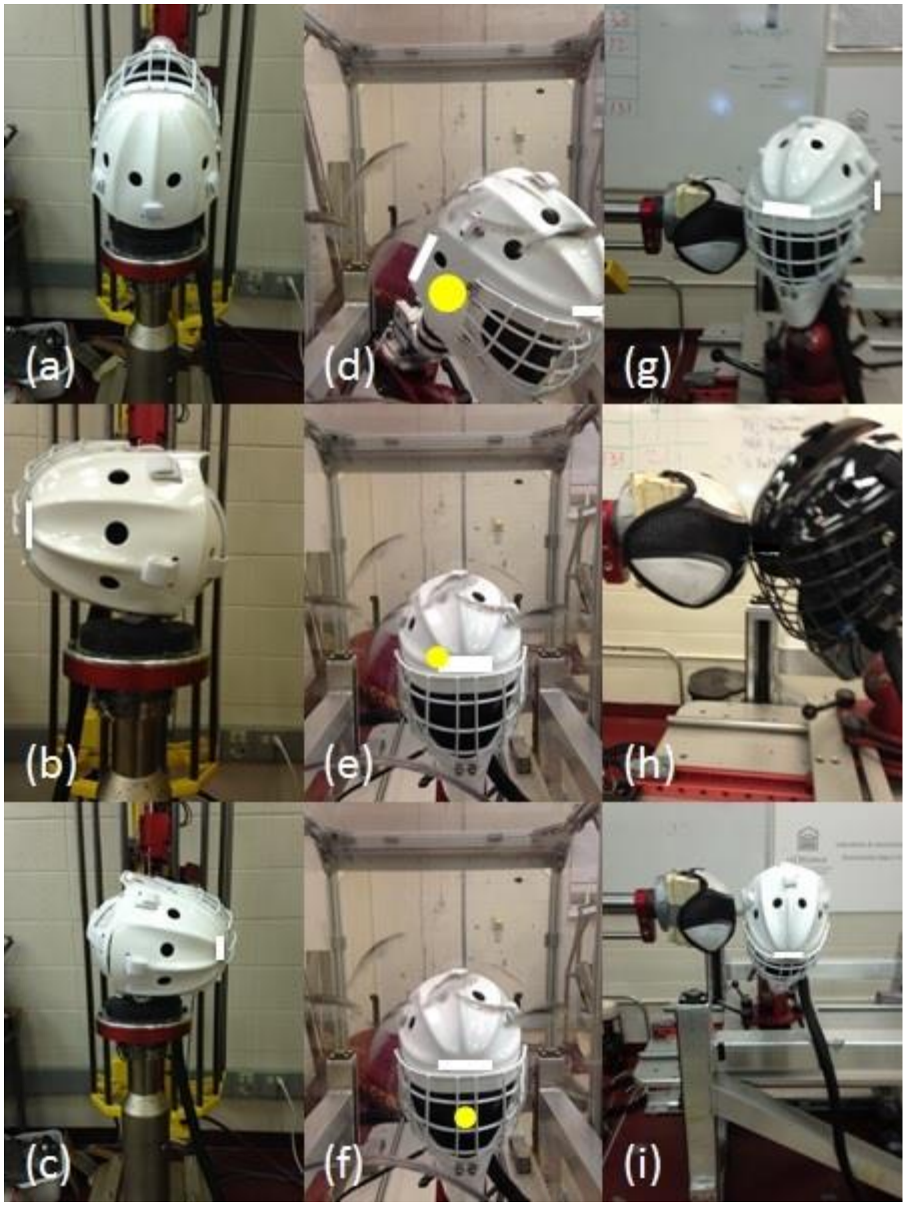

Figure 3. Impact location of an event specific impact test protocol for ice hockey goaltender helmets: (a) fall Rear-D, (b) fall L4-D, (c) fall R3-D, (d) puck impact Front-D, (e) puck impact R1-B, (f) puck impact R3-D, (g) collision R2-E, (h) collision R1-B, (i) collision R3-C.[4]

Nine helmets of each model were impacted under the test conditions; a new helmet was used for each impact event and velocity. Overall, 243 impacts were conducted, in which three trials were performed for each impact condition. The peak linear and rotational acceleration of the headform were obtained for each impact. The linear and rotational acceleration curves were transformed using a rotation matrix prior to inputting the kinematic response into a finite element brain trauma model. Within the NOCSAE headform, the accelerometer blocks are orientated at 25° with respect to the y-axis about the centre of gravity. To align the kinematic response of the NOCSAE headform with the coordinate system used by the finite brain trauma model the following rotational matrix curve was applied to the linear and rotational acceleration curves at the centre of gravity of the headform:

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

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 1 when applied to acceleration-time histories from an accelerometer block orientated at 25° resulted in mean linear and rotational acceleration-time histories which were within the 95% confidence interval of acceleration-time histories from a 0° accelerometer block orientation with minor periods in which the acceleration-time histories differed.[2] The differences were within ± 7 g and ± 700 rad/s^2 and as a result rotated linear and rotational curves were considered to be similar. As the rotated linear and rotational curves were similar, the curves could be input into a finite element brain trauma model for use in head impact biomechanics research.[2] The rotated kinematic response curves were input into a finite element brain trauma model which calculated the magnitude of peak maximal principal strain (MPS) in the cerebrum.

Equipment

Headform

For all impact conditions a medium NOCSAE headform (4.85 ± 0.01 kg) was attached to an unbiased neckform.[61] The unbiased neckform is made up of four centred and unarticulated rubber butyl disks of radius 68.0 mm and height 21.5 mm. The rubber disks fit serially and slightly recessed inside aluminium disks measuring 85.6 mm in radius and 12.8 mm in height. The design allows the unbiased neckform to respond symmetrical in all axes and eliminate potential biased effects of the Hybrid III neckform.[61] Nine single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) were fixed in the headform in a 3-2-2-2 accelerometer array.[39] Signals from the nine accelerometers were collected at 20 KHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered through a CFC 180 filter.

Monorail

A monorail drop rig equipped with a 60 shore A modular elastomer programmer (MEP) anvil was used in this study to simulate falls to the ice (Fig. 4).[7] The NOCSAE headform and an unbiased neckform were attached to a drop carriage of the monorail drop rig. The drop carriage ran along a 4.7 m long rail on bushings to reduce the effects of friction on the inbound velocity of the headform. The monorail drop rig was connected to a computer equipped with Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC), which controlled the velocity and release mechanism for the drop carriage. A photoelectric time gate was used to measure the inbound velocity of the impact within 0.02 m of the impact.

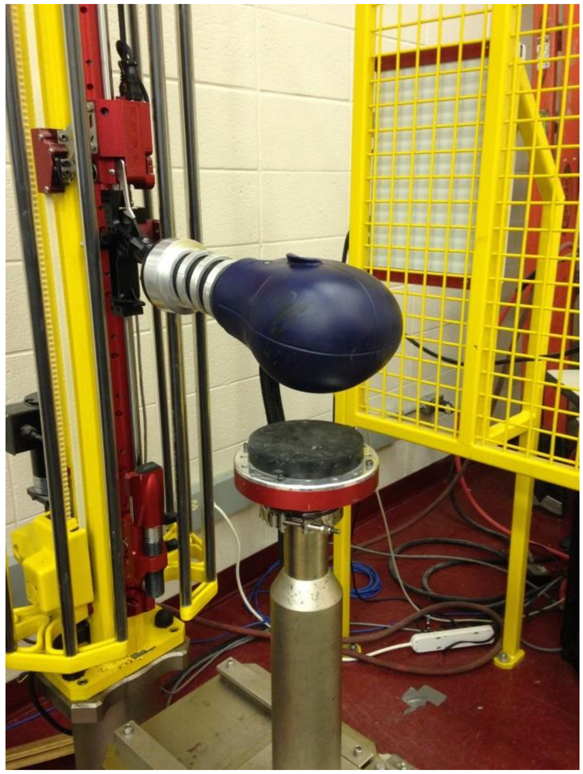

Figure 4. Monorail drop rig with NOCSAE headform attached.

Linear Impactor and Puck Launcher

A pneumatic linear impactor was used with a shoulder pad to represent collision,[54] and a pneumatic puck launcher cannon was used for puck impacts. The pneumatic linear impactor and puck launcher cannon were attached to a support/piston frame. The support frame held a compressed air canister and piston. The impacting arm of the linear impactor or a puck from the puck launcher was propelled towards the headform by compressed air. The mass of the impacting arm was 13.1 ± 0.1 kg which was similar to the calculated effective mass of shoulder-to-head collisions in ice hockey reconstructions.[54] 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.[54] The striker produces a similar linear acceleration peak and duration to that of shoulder collisions performed by volunteer ice hockey players impacting a Hybrid III headform at low and high velocities.[54] When either the linear impactor or puck launcher was used, the headform and neckform were attached to a movable locking base. The moveable locking base allowed for the headform to be orientated 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 remain fixed in position during impacts. The movable part ran along a low low-friction sliding table which allowed for movement post impact.

Goaltender Helmets

Three commercially available ice hockey goaltender helmets were tested in this study (Fig. 5). Each of the helmets was fitted according to manufacturer's specifications on the NOCSAE headform. Descriptions of the helmet's characteristics are presented in Table 2. The three models chosen in this study were selected to represent a range of materials commonly used in ice hockey goaltender helmets. The specific liner density and shell stiffness values have not been identified and remain confidential, following the supplier's request in accordance with research agreements between the supplier and research group. Helmet 3 has the stiffest shell allowed by Helmet 2 and Helmet 1 has the softest shell. The shells of the helmets were all made from the same model designed to have the same thickness with manufacturing variance. The external shell geometry of the three helmets were the same in order to remove any influence of external shell geometry of the results.[59]

Table 2: Ice hockey goaltender helmets characteristics.

Ice Hockey Goaltender Helmet	Cage Material	Shell Material	Shell of Helmet (mm)	Foam Liner Material	Shell + Liner (mm)	Mass (g)
Helmet 1	Carbon	Polycarbonate	3.75 ± 0.23	Vinyl Nitrile 602	20.93 ± 0.90	1.334 ± 0.004
Helmet 2	Stainless Steel	Fiberglass	3.55 ± 0.45	Vinyl Nitrile 600, 602 and 740	14.27 ± 1.87	1.172 ± 0.012
Helmet 3	Titanium	Carbon and Kevlar Composite	3.50 ± 0.46	Vinyl Nitrile 600, 602 and 740	14.19 ± 1.39	1.246 ± 0.059

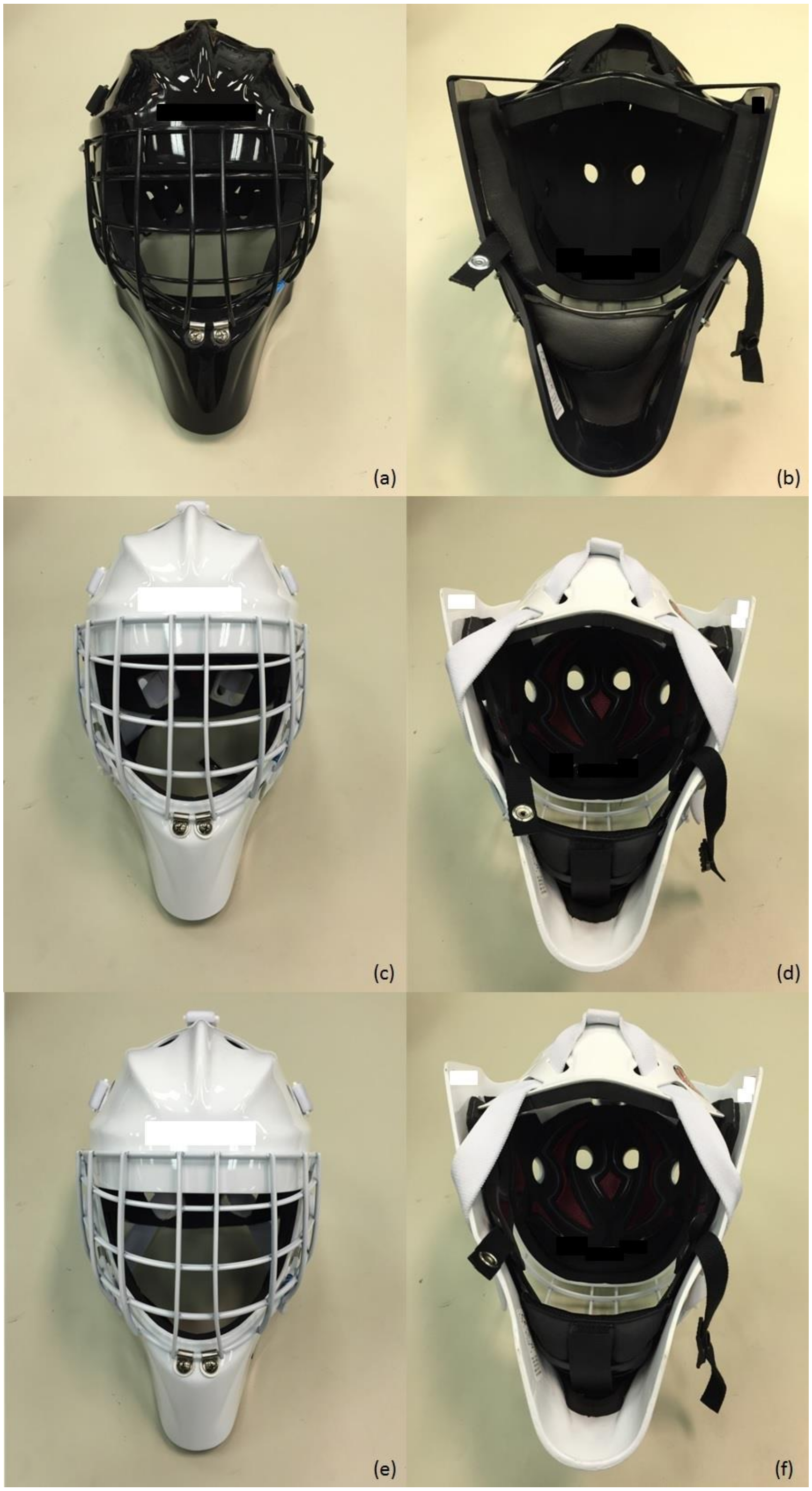

Figure 5. Three commercially available ice hockey goaltender helmets (a,b) Helmet 1; (c,d) Helmet 2; (e,f) Helmet 3.

Finite Element Model

The University College Dublin Brain Trauma Model (UCDBTM) was the finite element model used for this research.[15, 16] The geometry of the UCDBTM was extracted for computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver.[15] The model included the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem consists of approximately 26,000 elements.[15, 16] The material parameters used in the UCDBTM were taken from the literature and are presented in Tables 3 and 4.[29, 57, 64, 66, 68] The material behaviour of the brain tissue was modelled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus.[15] The compressive behaviour of the brain tissue was considered elastic. The shear characteristics of the viscoelastic behaviour of the brain were defined as:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t} \qquad [2]$$

where G_∞, is the long term shear modulus, G_0, is the short term shear modulus and β is the decay factor [15]. The UCDBTM modeled the CSF layer using solid elements with low shear modulus and a high bulk to create a sliding boundary condition between the pia and CSF. The contact interaction between the pia and CSF allowed no separation and used a friction coefficient of 0.2.[31]

Table 3: Material properties for the University College Dublin Brain Trauma Model.[15]

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

Table 4: Material characteristics of the brain tissue used in the University College Dublin Brain Trauma Model.[15]

Material	Shear modules (kPa)		Decay Constant (s^{-1})	Bulk Modulus (Gpa)
	G_0	G_∞		
Cerebellum	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19

Model validation was performed by comparing the UCDBTM's response to cadaveric intracranial pressure data from Nahum, Smith and Ward 32 and brain motion research using neutral density targets (NDT's) in cadaver impact conducted by Hardy, Foster, Mason, Yang, King and Tashman 14. The model's response was found to closely approximate the cadaveric pressure responses[32] and brain motion[14] from experimental results in both shape and duration.[15, 16] The response of the model was further examined using reconstructions of real world events in which there was good agreement with simulation results and lesions on CT scans for TBI incidents.[9, 44]

Statistics

To compare the response of ice hockey goaltender helmets one-way ANOVAs were conducted for each impact condition on mean peak linear acceleration, rotational acceleration and maximal principal strain (MPS). Peak linear acceleration, rotational acceleration and MPS were selected as dependant variables as they are common measures used to describe the severity of an impact to the head in brain injury research.[9, 25, 41, 45, 46, 67] When significant main effects were found post hoc Tukey tests were performed. For all comparisons α was set to 0.05. Statistical analysis was performed using the statistical software package of SPSS 19.0 for Windows (SPSS Inc, Chicago, IL, USA).

Results

The effect of different tested ice hockey goaltender helmet models on the kinematic response and MPS are presented in Figures 6-8.

Falls

Significant main effects were found for linear acceleration of falls across all velocities and locations ($p < 0.05$) except for 5.0 m/s at impact location L4-D and R3-D. For rotational acceleration produced by falls, significant main effects of helmet model were found across all location and velocities ($p < 0.05$), with the exceptions of 3.5 m/s at L4-D, 4.2 m/s at R3-D and 5.0 m/s at Rear-D. Falls were found to have significant main effects for MPS at all velocities and locations ($p < 0.05$) expect for locations Rear-D and L4-D at 3.5 m/s. Figure 6 present the results of the post hoc tests for kinematic response and MPS of falls.

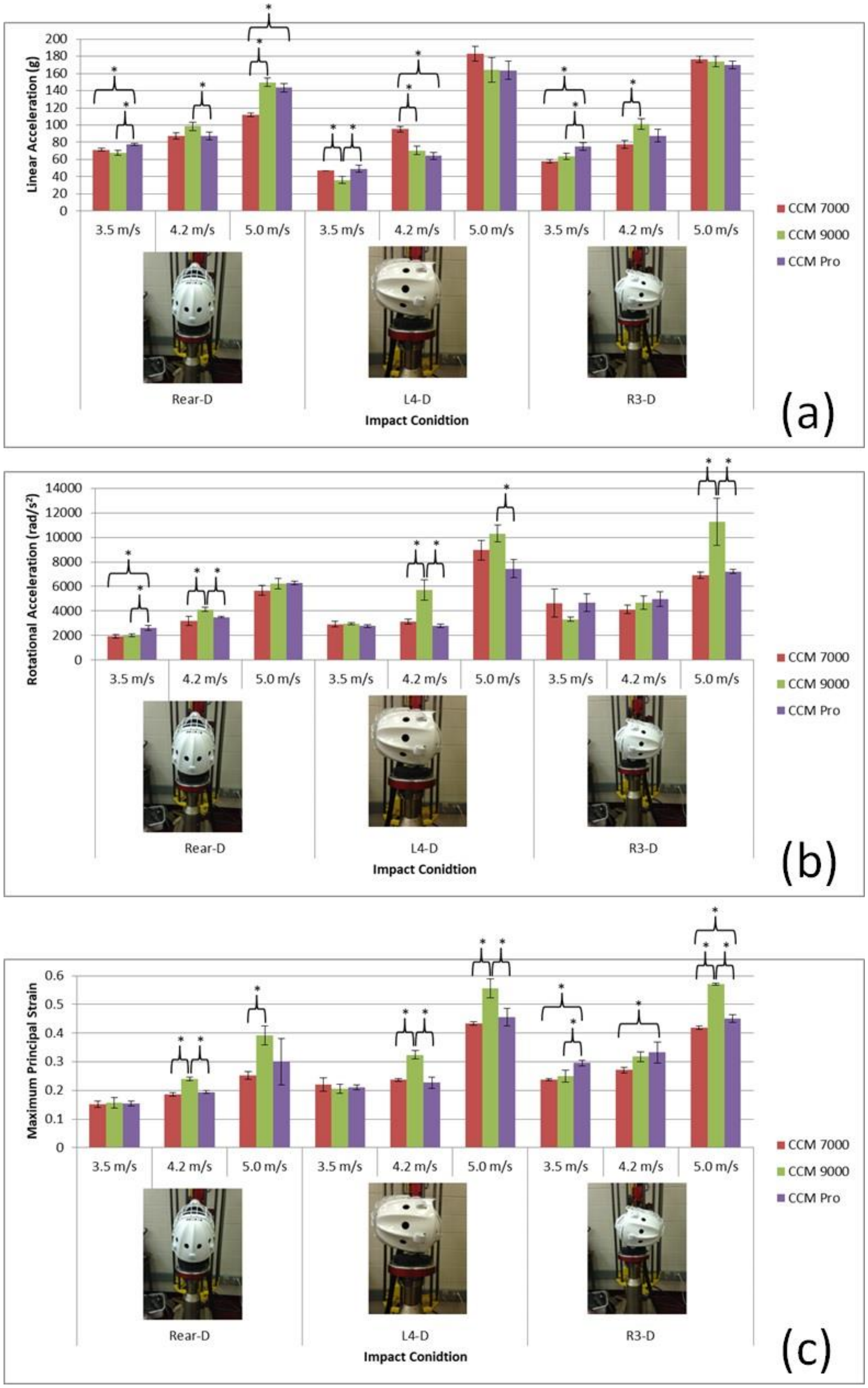

Figure 6. Mean peak kinematic response and maximum principal strain of falls for three ice hockey goaltender helmets: (a) linear acceleration; (b) rotational acceleration; (c) maximum principal strain.

Puck Impacts

For puck impact significant main effects were found in linear acceleration at Front-D and R3-D across all velocities ($p < 0.05$). No significant main effects were found for R1-B ($p > 0.05$) except at 29.4 m/s. Significant main effects for rotational acceleration of puck impacts were found at Front-D and R3-D at all velocities ($p < 0.05$) except for Front-D at 42.3 m/s. Impact location R1-B showed no significant main effects ($p < 0.05$) with the exception of 42.3 m/s. For MPS, puck impacts were found to have significant main effects for 29.3 m/s and 35.8 m/s at Front-D at, and 35.8 and 42.3 m/s at R1-B ($p < 0.05$) but not for other impact conditions. The post hoc tests results for kinematic response and MPS of puck impacts are presented in Figure 7.

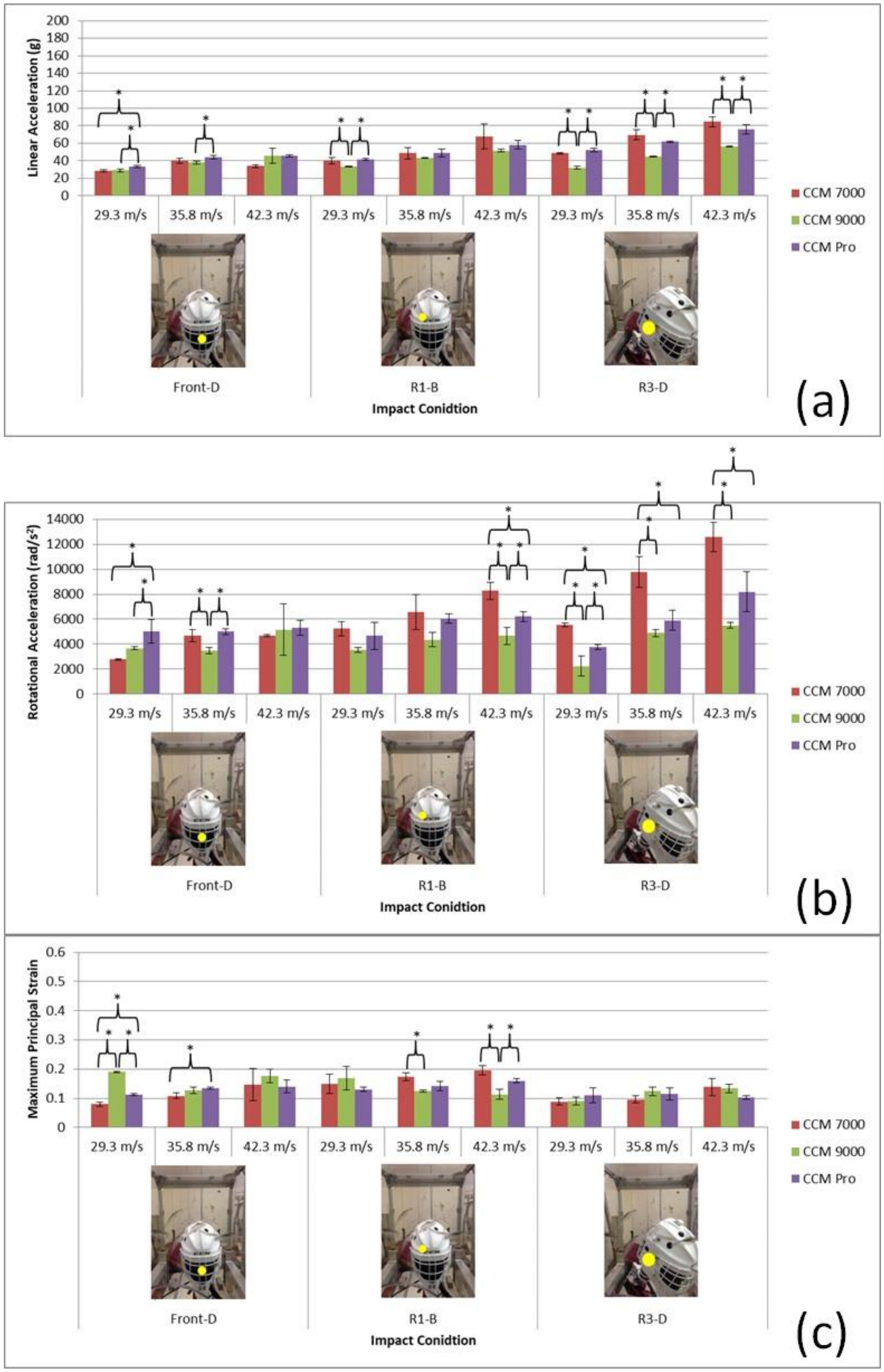

Figure 7. Mean peak kinematic response and maximum principal strain of puck impacts for three ice hockey goaltender helmets: (a) linear acceleration; (b) rotational acceleration; (c) maximum principal strain.

Collisions

Collisions were found to have significant main effects for linear acceleration at R3-C and R2-E across all velocities ($p < 0.05$) with the exception of R3-C at 9.1 m/s. Impact location R1-B showed no significant main effects ($p > 0.05$) except at 9.1 m/s. Significant main effects were found for rotational acceleration of collisions at R3-C and R1-B at all velocities ($p < 0.05$) except for R3-C at 7.3 m/s. No significant main effects were found for R2-E ($p > 0.05$) with the exception of 7.3 m/s. For MPS produced by collisions no significant main effects were found across all locations and velocities ($p > 0.05$) expect for R3-C and R2-E at 7.3 m/s. Figure 8 demonstrate the results of the post hoc tests for kinematic response and MPS of collisions.

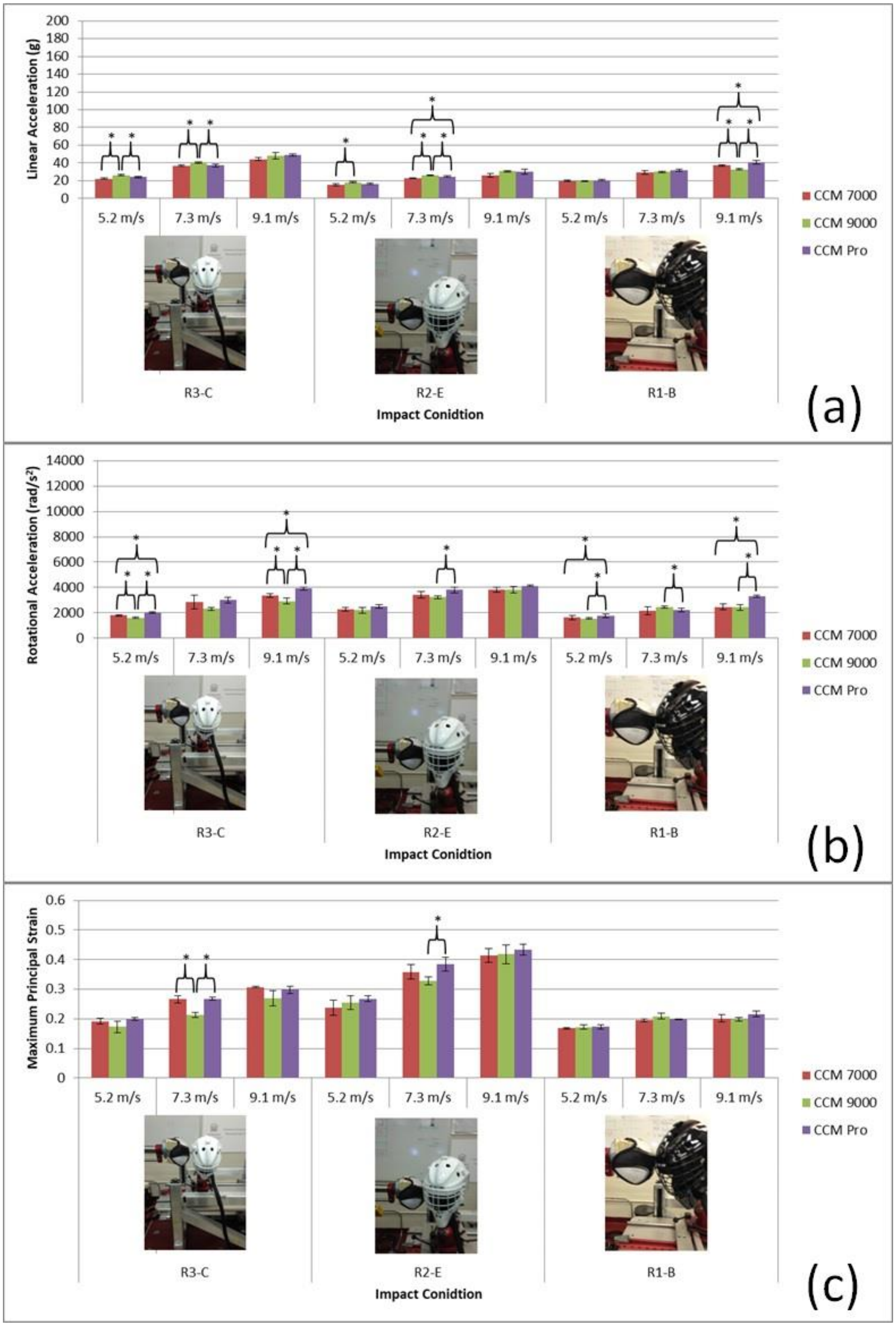

Figure 8. Mean peak kinematic response and maximum principal strain of collisions for three ice hockey goaltender helmets: (a) linear acceleration; (b) rotational acceleration; (c) maximum principal strain.

Discussion

Falls

The results for falls indicate that the tested ice hockey goaltender helmets perform differently depending on the dependent variable being measured. Overall, there was not a large difference in performance of the tested ice hockey goaltender helmets when analysed using linear acceleration for falls. This was expected as ice hockey goaltender helmets are certified using a drop test and peak linear acceleration to determine impact absorption properties.[1, 6, 7, 22] As a result the materials used in tested ice hockey goaltender helmets are designed to reduce linear accelerations for falling conditions and offered similar protection across models. However kinematic response and MPS values remained within reported ranges of concussion.[13, 28, 33, 40, 53, 67] These results are similar to those found for studies comparing player ice hockey helmets, as different helmet models have been found to produce similar linear accelerations but different rotational accelerations.[50, 55, 62] A reduction in linear acceleration does not necessarily result in lower rotational accelerations[12, 27, 50, 51, 67] and as a result, although the tested helmets may reduce linear acceleration similarly, the different design aspects may not decrease the rotational energy of an impact in a similar manner. When examining rotational acceleration and MPS for falls at 4.2 and 5.0 m/s, Helmet 1 and Helmet 3 generally outperformed Helmet 2. These differences in performance are a reflection of the different design characteristics for the tested ice hockey goaltender helmets. Helmet 1 has a thicker and single low density liner compared to Helmet 2 which has a thinner and dual density liner. Helmet 3 has a stiffer shell than Helmet 2. It is likely the thicker and single low density liner of Helmet 1 and stiffer shell of Helmet 3 compared to Helmet 2 resulted in reduced rotational acceleration and MPS. The thicker single low density liner would allow for more compression allowing for greater energy attenuation, while the stiffer shell material could allow the impact force to be spread across a greater portion of the liner.[10-12, 46, 47]

Puck Impacts

When examining the results for puck impacts, tested ice hockey goaltender helmets were found to produce different responses as measured by the dependent variables depending on the impact location. Impact location Front-D represents an impact to the cage of ice hockey goaltender helmet. For this location linear acceleration values were found to be below reported ranges of concussion but for rotational acceleration and MPS values most impact conditions were found to be within reported ranges of concussion.[13, 28, 33, 40, 67] Despite each helmet tested using a different material for the cage; the tested helmets generally produced similar linear and rotational accelerations and MPS, demonstrating that each of the cages has similar energy attenuation levels. Demonstrating the different materials commonly used for the cage of ice hockey goaltender helmets do not improve performance for puck impacts. When examining locations R1-B and R3-D most linear accelerations values were below reported ranges of concussion whereas most conditions had rotational acceleration and MPS values within reported ranges of concussion.[13, 28, 33, 40, 67] For these locations Helmet 2 was found to perform the best when examining linear and rotational accelerations. Helmet 2 has a stiffer shell than Helmet 1 and a thicker dual density liner than Helmet 3. The combination of these characteristics caused Helmet 2 to have reduced linear and rotational acceleration by 12.8 – 145.9 %.

For projectile impacts, having a stiff shell is desirable as this can help to distribute the impact over a larger contact area of the energy attenuating foam liner.[46] However when MPS was examined, all tested helmets were found to produce similar values in most conditions. This suggests each tested ice hockey goaltender helmet may deflect and/or attenuate energy for a puck impact differently due to differences in liner thickness and shell material resulting in different peak accelerations.[37] However the different design considerations may not influence other aspects of the acceleration curves such as slope and duration which contribute to brain strain. In order for the tested ice hockey goaltender helmets to influence the level of MPS produced from puck impacts the material and design considerations may need to consider other aspects of the acceleration curves. Similar findings have been reported for ice hockey skater helmets[47, 52] and brain injury simulations[43, 45, 65] which would support these results.

Collisions

Collisions are the leading cause of concussions for ice hockey goaltenders[30] and as a result improving helmet design for protection against collisions could reduce the incidence of injury. The resulting head kinematic and brain strain values produced by collisions in this study were found to be within the range of reported concussive collisions in ice hockey reconstructions.[23, 42, 53] The tested ice hockey goaltender helmets examined in this study were found to produce similar kinematic response and brain strain values for collisions. The similar results among the tested ice hockey goaltender helmets for collisions suggest that unlike falls and puck impacts a thicker liner and stiff shell material does not offer any protective advantage. For highly compliant impacts such as shoulder collisions a relativity small amount of energy is attenuated by the helmet.[3, 5, 10, 20] The materials currently used in the tested ice hockey goaltender helmets are stiffer than the shoulder and therefore have minimal influence on the resulting kinematic response and brain strain values as the materials do not compress enough to absorb a sufficient amount of the impact energy. As collisions are the leading cause of concussion for ice hockey goaltenders[30], other helmet design characteristics which can reduce kinematic response and brain strain values should be considered. Potential liner material and design considerations which may aid in reducing head kinematic and brain tissue responses for collisions are the use of 3-D liner structures, layered foam liners or functionally graded liners.[8, 11, 12, 47, 49] Research in equestrian helmet designs has shown layered foam liners and functionally graded liners perform better than traditional uniform liners across a 4.4 to 7.7 m/s velocity range.[8, 10, 11] In American Football and ice hockey research 3-D liner structures have been shown to result in a reduction of kinematic and brain tissue response values for collisions compared to helmets using VN and expanded polypropylene foam liners. These materials may provide an effective design solution to manage head kinematic and brain tissue responses for the loading conditions created by collisions.

Limitations

The results of the present study should be considered according to its limitations. The NOCSAE headform used in this study may not imitate the dynamic properties of a human head, however the response of the headform has been found to be within those expected for cadaveric impacts.[26] Additionally the NOCSAE headform is widely accepted as a human head surrogate and is used in the certification of American football, lacrosse and ice hockey helmets.[34-36] Three ice hockey goaltender helmets were tested in this study. These three helmet models may not characterise all design characteristics in goaltender helmets and other design characteristics may cause differences in helmet performance.[37, 38, 55, 56] The finite element model used in this study, the UCDBTM, makes assumptions surrounding the boundary conditions and material properties of the model cadaveric and other anatomical testing. As a result the response of the model is meant to be a representation of how the brain could react to an impact and may not represent the exact motion of the brain.

Summary

The purpose of this study was to examine how design characteristics (liner thickness and shell material) of ice hockey goaltender helmets affect head kinematics and brain strain for head impact events in ice hockey. The results demonstrated that helmet shell stiffness and liner thickness had no clinically significant effect on peak linear acceleration for falls and remains within reported ranges of concussion. However, they had a positive reducing effect on rotational acceleration and MPS. For puck impacts, the tested helmets with thicker liners and stiffer shells reduced the linear and rotational accelerations but had no significant effect on MPS. This suggests that a thick liner and stiff shell material are desirable design characteristics to protect against falls and puck impacts. For collisions however, such design characteristics had minimal effect on resulting head kinematic and brain tissue responses. This is likely a result the materials used in the tested ice hockey helmets being stiffer and as a result the helmet materials do not compress enough to absorb a sufficient amount of the impact energy, minimizing performance differences among the tested ice hockey goaltender helmets. This suggests that given the tested ice hockey goaltender helmets can be optimized for protection from falls and puck impacts. However, as collisions are the leading cause of concussion for ice hockey goaltenders and the tested helmets provided little to no protection, new helmets capable of reducing kinematic response and brain tissue strain should be considered.

Acknowledgments

Helmets for this research were supplied by Reebok-CCM and funding was received from the Ontario Graduate Scholarship in Science and Technology (OGSST). Two of the authors (J.M.C and M.D.G.) receive funding from the European Union’s Horizon 2020 research and innovation programme under the Marie Skiodowska-Curie grant agreement No. 642662.

Conflict of Interest

The University of Ottawa holds research agreements with the helmet manufacture for testing and development of ice hockey goaltender masks.

References

1. ASTM. Standard Specification for Head and Face Protective Equipment for Ice Hockey Goaltenders. In: *F1587-12a*. West Conshohocken, PA: American Society for Testing and Materials International, 2014.
2. Clark J. M. Evaluation of the Protective Capacity of Ice Hockey Goaltender Masks for Three Accident Events using Dynamic Response and Brain Stress and Strain. Université d'Ottawa/University of Ottawa, 2015.
3. Clark J. M., T. B. Hoshizaki and M. D. Gilchrist. Protective capacity of an ice hockey goaltender helmet for three events associated with concussion. *Computer methods in biomechanics and biomedical engineering* 20: 1299-1311, 2017.
4. Clark J. M., A. Post, T. B. Hoshizaki and M. D. Gilchrist. The Association among Injury Metrics for Different Events in Ice Hockey Goaltender Impacts. In: *IRCOBI Conference*. Malaga, Spain: 2016, p. 205-216.
5. Clark J. M., A. Post, T. B. Hoshizaki and M. D. Gilchrist. Protective capacity of ice hockey helmets against different impact events. *Annals of biomedical engineering* 44: 3693-3704, 2016.
6. CSA. Face Protectors for Use in Ice Hockey. In: *Z262.2-09*. Mississauga, Ontario, Canada: Canadian Standards Association, 2009.
7. CSA. Ice Hockey Helmets. In: *Z262.1-09*. Mississauga, Ontario, Canada: Canadian Standards Association, 2009.
8. Cui L., M. A. Forero Rueda and M. Gilchrist. Optimisation of energy absorbing liner for equestrian helmets. Part II: Functionally graded foam liner. *Materials & Design* 30: 3414-3419, 2009.
9. Doorly M. C. and M. D. Gilchrist. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. *Computer methods in biomechanics and biomedical engineering* 9: 371-377, 2006.
10. Forero Rueda M. A. Equestrian Helmet Design: A Computational and Head Impact Biomechanics Simulation Approach. University College Dublin, 2009.
11. Forero Rueda M. A., L. Cui and M. Gilchrist. Optimisation of energy absorbing liner for equestrian helmets. Part I: Layered foam liner. *Materials & Design* 30: 3405-3413, 2009.
12. Forero Rueda M. A., L. Cui and M. D. Gilchrist. Finite element modelling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. *Computer methods in biomechanics and biomedical engineering* 14: 1021-1031, 2011.
13. Giordano C. and S. Kleiven. Evaluation of axonal strain as a predictor for mild traumatic brain injuries using finite element modeling. *Stapp car crash journal* 58: 29, 2014.
14. Hardy W. N., C. D. Foster, M. J. Mason, K. H. Yang, A. I. King and S. Tashman. Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-ray. *Stapp car crash journal* 45: 337-368, 2001.

15. Horgan T. J. and M. D. Gilchrist. The creation of three-dimensional finite element models for simulating head impact biomechanics. *International Journal of Crashworthiness* 8: 353-366, 2003.
16. Horgan T. J. and M. D. Gilchrist. Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. *International Journal of Crashworthiness* 9: 401-418, 2004.
17. Hoshizaki T. B. and S. E. Brien. The science and design of head protection in sport. *Neurosurgery* 55: 956-967, 2004.
18. Hoshizaki T. B., A. Post, R. A. Oeur and S. E. Brien. Current and future concepts in helmet and sports injury prevention. *Neurosurgery* 75: S136-S148, 2014.
19. Hoshizaki T. B., E. Walsh, A. Post, P. Rousseau, M. Kendall, C. Karton, A. Oeur, S. Foreman and M. D. Gilchrist. The application of brain tissue deformation values in assessing the safety performance of ice hockey helmets. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 1754337112448765, 2012.
20. Hunt H. and N. Mills. The protection of horse riders in impacts with the ground. In: *IRCOBI*. Stockholm, Sweden: 1989, p. 157–168.
21. Hutchison M. G., P. Comper, W. H. Meeuwisse and R. J. Echemendia. A systematic video analysis of National Hockey League (NHL) concussions, part I: who, when, where and what? *British journal of sports medicine* 49: 547-551, 2015.
22. IOS. Head and Face Protection for Use in Ice Hockey. In: *ISO 10256*. Geneva, Switzerland: International Organization for Standardization, 2003.
23. Kendall M. Comparison and Characterization of Different Concussive Brain Injury Events. Université d'Ottawa/University of Ottawa, 2016.
24. Kendall M., A. Post, P. Rousseau and T. Hoshizaki. The effect of shoulder pad design on reducing peak resultant linear and rotational acceleration in shoulder-to-head impacts. In: *Mechanism of Concussion in Sports*ASTM International, 2014.
25. Kendall M., A. Post, P. Rousseau, A. Oeur, M. Gilchrist and B. Hoshizaki. A comparison of dynamic impact response and brain deformation metrics within the cerebrum of head impact reconstructions representing three mechanisms of head injury in ice hockey. In: *IRCOBI Conference*. Dublin, Ireland: International Research Council on the Biomechanics of Injury, 2012, p. 430-440.
26. Kendall M., E. S. Walsh and T. B. Hoshizaki. Comparison between Hybrid III and Hodgson–WSU headforms by linear and angular dynamic impact response. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 1754337112436901, 2012.
27. Kleiven S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *International Journal of Crashworthiness* 11: 65-79, 2006.
28. Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstructions. *Stapp Car Crash J* 51: 81-114, 2007.
29. Kleiven S. and H. von Holst. Consequences of brain size following impact in prediction of subdural hematoma evaluated with numerical techniques. In: *Proceedings of the International Research Council on the Biomechanics of Injury conference*. Munich, Germany: International Research Council on Biomechanics of Injury, 2002, p. 161-172.

30. LaPrade R., C. Wijdicks and S. Spiridonov. A prospective study of injuries in NCAA intercollegiate ice-hockey goaltenders. In: *Fifth International Symposium on Safety in Ice Hockey*ASTM International, 2009.
31. Miller R. T., S. S. Margulies, M. Leoni, M. Nonaka, X. Chen, D. H. Smith and D. F. Meaney. Finite element modeling approaches for predicting injury in an experimental model of severe diffuse axonal injury. SAE Technical Paper, 1998.
32. Nahum A. M., R. Smith and C. C. Ward. Intracranial pressure dynamics during head impact. In: *21st Stapp Car Crash Conference*. Warrendale, PA: SAE Technical Paper, 1977, p. 337.
33. Newman J. A., C. Barr, M. C. Beusenberg, E. Fournier, N. Shewchenko, E. Welbourne and C. Withnall. A new biomechanical assessment of mild traumatic brain injury. Part 2: results and conclusions. In: *International Research Council on the Biomechanics of Injury conference*. Montpellier, France: 2000, p. 223-233.
34. NOCSAE. Standard Performance Specification for Newly Manufactured Football Helmets In: *NOCSAE DOC (ND)002-13m15*. Overland Park, Kansas, USA: National Operating Committee on Standards for Athletic Equipment, 2015.
35. NOCSAE. Standard Performance Specification for Newly Manufactured Ice Hockey Helmets. In: *NOCSAE DOC (ND) 030 - 11m16*. Overland Park, Kansas, USA: National Operating Committee on Standards for Athletic Equipment, 2016.
36. NOCSAE. Standard Performance Specification for Newly Manufactured Lacrosse Helmets with Faceguard. In: *NOCSAE DOC (ND) 041- 11m15*. Overland Park, Kansas, USA: National Operating Committee on Standards for Athletic Equipment, 2015.
37. Nur S., M. Kendall, J. M. Clark and T. B. Hoshizaki. A comparison of the capacity of ice hockey goaltender masks for the protection from puck impacts. *Sports biomechanics* 14: 459-469, 2015.
38. Ouckama R. and D. Pearsall. Projectile Impact Testing of Ice Hockey Helmets: Headform Kinematics and Dynamic Measurement of Localized Pressure Distribution. *IRCOBI Proceedings* 62-71, 2014.
39. Padgaonkar A. J., K. Krieger and A. King. Measurement of angular acceleration of a rigid body using linear accelerometers. *Journal of Applied Mechanics* 42: 552-556, 1975.
40. Patton D. A., A. S. McIntosh and S. Kleiven. The Biomechanical Determinants of Concussion: Finite Element Simulations to Investigate Tissue-Level Predictors of Injury During Sporting Impacts to the Unprotected Head. *Journal of applied biomechanics* 31: 2015.
41. Pellman E. J., D. C. Viano, A. M. Tucker, I. R. Casson and J. F. Waeckerle. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery* 53: 799-814, 2003.
42. Post A., J. M. Clark, D. Robertson, T. B. Hoshizaki and M. D. Gilchrist. The effect of acceleration signal processing for head impact numeric simulations. *Sports Engineering* 1-9, 2016.
43. Post A., B. Hoshizaki and M. D. Gilchrist. Finite element analysis of the effect of loading curve shape on brain injury predictors. *Journal of biomechanics* 45: 679-683, 2012.
44. Post A., T. B. Hoshizaki, M. D. Gilchrist, S. Brien, M. Cusimano and S. Marshall. Traumatic brain injuries: the influence of the direction of impact. *Neurosurgery* 76: 81-91, 2015.
45. Post A., T. B. Hoshizaki, M. D. Gilchrist, S. Brien, M. D. Cusimano and S. Marshall. The influence of dynamic response and brain deformation metrics on the occurrence of subdural

hematoma in different regions of the brain: Laboratory investigation. *Journal of neurosurgery* 120: 453-461, 2014.
46. Post A., C. Karton, T. Blaine Hoshizaki, M. D. Gilchrist and J. Bailes. Evaluation of the protective capacity of baseball helmets for concussive impacts. *Computer methods in biomechanics and biomedical engineering* 19: 366-375, 2016.
47. Post A., C. Karton, T. B. Hoshizaki and M. D. Gilchrist. Analysis of the protective capacity of ice hockey helmets in a concussion injury reconstruction. In: *IRCOBI Conference*. Berlin, Germany: 2014, p. 72-80.
48. Post A., D. Koncan, M. Kendall, J. Cournoyer, J. M. Clark, G. Kosiwka, W. Chen, A. de Grau and T. B. Hoshizaki. Analysis of velocity accuracy using a low-cost video analysis software. *Sports Engineering* in press: 2018.
49. Post A., A. Oeur, B. Hoshizaki and M. D. Gilchrist. An examination of American football helmets using brain deformation metrics associated with concussion. *Materials & Design* 45: 653-662, 2013.
50. Post A., A. Oeur, B. Hoshizaki and M. D. Gilchrist. Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts. *Computer methods in biomechanics and biomedical engineering* 16: 511-519, 2013.
51. Post A., A. Oeur, T. Hoshizaki and M. D. Gilchrist. The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research. In: *IRCOBI Conference*. Dublin, Ireland: 2012, p. 10-14.
52. Post A., E. S. Walsh, T. B. Hoshizaki and M. D. Gilchrist. Analysis of loading curve characteristics on the production of brain deformation metrics. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 200-207, 2012.
53. Rousseau P. Analysis of concussion metrics of real-world concussive and non-injurious elbow and shoulder to head collisions in ice hockey. Ottawa, Canada: Université d'Ottawa/University of Ottawa, 2014.
54. Rousseau P. and T. B. Hoshizaki. Defining the effective impact mass of elbow and shoulder strikes in ice hockey. *Sports biomechanics* 14: 57-67, 2015.
55. Rousseau P., A. Post and T. Hoshizaki. A comparison of peak linear and angular headform accelerations using ice hockey helmets. In: *Fifth International Symposium on Safety in Ice Hockey*ASTM International, 2009.
56. Rousseau P., A. Post and T. Hoshizaki. The effects of impact management materials in ice hockey helmets on head injury criteria. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 223: 159-165, 2009.
57. Ruan J. S. Impact biomechanics of head injury by mathematical modeling. Wayne State University, 1994.
58. Smith T. A., P. D. Halstead, E. McCalley, S. A. Kebschull, S. Halstead and J. Killeffer. Angular head motion with and without head contact: implications for brain injury. *Sports Engineering* 18: 165-175, 2015.
59. Spyrou E., D. Pearsall and T. Hoshizaki. Effect of local shell geometry and material properties on impact attenuation of ice hockey helmets. *Sports Engineering* 3: 25-35, 2000.
60. Stuart M. J. and A. Smith. Injuries in Junior A Ice Hockey A Three-Year Prospective Study. *The American journal of sports medicine* 23: 458-461, 1995.
61. Walsh E. S. and T. Hoshizaki. Comparative analysis of the Hybrid III neckform to unbiased neckforms using a centric and non-centric impact protocol. In: *ASTM Symposium on the mechanism of concussion in sports*. Atlanta, GA: 2012.

62. Walsh E. S., A. Post, P. Rousseau, M. Kendall, C. Karton, A. Oeur, S. Foreman and T. B. Hoshizaki. Dynamic impact response characteristics of a helmeted Hybrid III headform using a centric and non-centric impact protocol. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 1754337112442299, 2012.
63. Wennberg R. and C. Tator. National Hockey League reported concussions, 1986-87 to 2001-02. *Canadian Journal of Neurological Sciences/Journal Canadien des Sciences Neurologiques* 30: 206-209, 2003.
64. Willinger R., L. Taleb and C. Kopp. Modal and temporal analysis of head mathematical models. *Journal of neurotrauma* 12: 743-754, 1995.
65. Yoganandan N., J. Li, J. Zhang, F. A. Pintar and T. A. Gennarelli. Influence of angular acceleration–deceleration pulse shapes on regional brain strains. *Journal of biomechanics* 41: 2253-2262, 2008.
66. Zhang L., K. H. Yang, R. Dwarampudi, K. Omori, T. Li, K. Chang, W. N. Hardy, T. B. Khalil and A. I. King. Recent advances in brain injury research: a new human head model development and validation. *Stapp Car Crash J* 45: 369-394, 2001.
67. Zhang L., K. H. Yang and A. I. King. A proposed injury threshold for mild traumatic brain injury. *Journal of biomechanical engineering* 126: 226-236, 2004.
68. Zhou C., T. B. Khalil and A. I. King. A new model comparing impact responses of the homogeneous and inhomogeneous human brain. In: *39th Stapp Car Crash Conference*SAE Technical Paper, 1995, p. 121-137.