Protective Capacity of Ice Hockey Helmets against Different Impact Events

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Abstract

In ice hockey, concussions can occur as a result of many different types of impact events, however hockey helmets are certified using a single injury scenario, involving drop tests to a rigid surface. The purpose of this study is to measure the protective capacity of ice hockey helmets for different impact events in ice hockey. A helmeted and unhelmeted Hybrid III headform were impacted simulating falls, elbow, shoulder and puck impacts in ice hockey. Linear and rotational acceleration and maximum principal strain (MPS) were measured. A comparison of helmeted and unhelmeted impacts found significant differences existed in most conditions ($p<0.05$), however some shoulder and puck impacts showed no significant difference ($p>0.05$). Impacts to the ice hockey helmet tested resulted in acceleration levels below reported ranges of concussion and TBI for falls up to 5 m/s, elbow collisions, and low velocity puck impacts but not for shoulder collisions or high velocity puck impacts and falls. The helmet tested reduced MPS below reported ranges of concussion and TBI for falls up to 5 m/s but not for the other impact events across all velocities and locations. This suggests that the ice hockey helmet tested is unable to reduce engineering parameters below reported ranges of concussion and TBI for impact conditions which do not represent a drop against a rigid surface.

Key Words: Ice hockey, Helmet, Concussion, Impact biomechanics, Finite element modeling

Introduction

Concussions are common in ice hockey and can lead to long term disability.[36] The use of helmets has reduced the incidence of traumatic brain injuries (TBI), however concussions are a common injury.[80] This may be related to current standard certification tests for ice hockey helmets employing linear dominant drop tests to a rigid surface[4] that do not fully reflect the complex loading scenarios experienced in sport.[31,69,73] In nature, linear and rotational acceleration seldom exist independent of each other[21,50] and, as such, both forms of loading have been shown to contribute to head injury.[55] Skull fractures have been more commonly associated with linear acceleration[33,47] whereas rotational kinematics are more commonly associated with concussion and other forms of TBI.[17,21,56,74] Additionally, research examining the cause of concussion in the National Hockey League (NHL) reported concussive events due to falls account for only 7% of all concussions while collisions with an opponent accounted for 88%.[26] Of the 88% of concussive collisions, shoulder-to-head impacts were the most common impact event.[27] The remaining concussions were reportedly caused by injuries from puck, stick, helmet, and knee impacts.[27] Similarly, collisions with an opponent are reported as the most common event causing concussion in boy's and men's hockey at the youth, high school and collegiate levels.[2,11,35] In women's hockey, concussions also occur as a result of collisions with an opponent; however, concussion also occurs frequently due to falls onto the ice and into the boards.[16] This demonstrates, across a broad range of ice hockey populations, that concussions do not only occur due to falls, but often also as a result of other impact events.

Collisions, falls, and puck impacts in ice hockey are characterized by different impact parameters such as impact location, mass, velocity, angle of impact, and compliance of impactor. All of these impact parameters affect the acceleration time curve profile, creating different combinations of linear and rotational acceleration time curve profiles which can influence brain tissue stresses and strains.[12–14,17,20,28,31,32,49,52,57,58,74,85,87] Falls in ice hockey are typically characterized by the mass of the head impacting a rigid impact surface,[25] resulting in high magnitude and short duration linear and rotational acceleration.[12,53,54,59] Such an event is reflected in the current ice hockey certification standard as it aims to replicate the injurious impact events examined by Gurdjian et al.[18] which involved an animal model and cadaver head drops to a rigid surface. Injuries caused by puck impacts are characterized by low mass and high velocity impact.[25] These impacts produce a very short duration of linear and rotational acceleration pulses,[49,66] whereas collisions in ice hockey are largely dependent on the characteristics of the part of the body that impacts the head[25] and have been found to produce lower magnitude and longer duration linear and rotational acceleration.[29,31,64] In addition, each of these events can impact the head in a linear (through the center of mass, or centric) and nonlinear (not through the center of gravity, or obliquely, or non-centric) fashion, which affects the linear and rotational acceleration time curve profiles experienced by the head and brain.[12,41,44,63,79] As a result, current helmet test methods are not designed to simulate conditions of impact that commonly lead to concussion. The purpose of this study was to measure the protective capacity of ice hockey helmets against different impact events. It was hypothesised that an ice hockey helmet would reduce peak accelerations and maximum principal strain (MPS) for fall impacts to levels that are below the reported ranges associated with head and brain injuries when compared to unhelmeted conditions. It was further hypothesised that this protection would not be associated with certain other impact events.

Methods

Equipment

An adult 50th percentile Hybrid III headform (4.54 ± 0.01 kg) was attached to an unbiased neckform (2.11 ± 0.01 kg)[78] and used for all impact conditions. The unbiased neckform was created to match the mass and dimensions of the Hybrid III neckform. The unbiased neckform consists of 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 neckform was designed to respond in the same manner to impacts in all directions and eliminate potential biased effects of the Hybrid III neckform.[78] 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.[50] Signals for the nine accelerometers were collected at 20 kHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered with a 1650 Hz low pass Butterworth filter in accordance with the SAE J211 convention.

Falls to the ice were reconstructed using a monorail drop rig with a 60 shore A modular elastomer programmer (MEP) anvil to represent ice (Fig. 1).[4] The Hybrid III headform and unbiased neckform was attached to the monorail drop system by a drop carriage. The drop carriage ran along rails on ball bushings to reduce the effects of friction on the inbound velocity of the headform and was released by a pneumatic piston. The monorail drop rig was connected to a computer equipped with Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC), a multi-functional program used to control the velocity and release mechanisms for the impact. The impact velocity was measured using a photoelectric time gate. Falls at 7 m/s were not conducted for an unhelmeted headform to avoid unnecessary equipment damage.

Figure 1. Monorail drop system used to simulate head impacting the ice (MEP anvil)

Collisions were reconstructed using a pneumatic linear impactor with two different strikers. The pneumatic linear impactor consisted of a frame, an impacting arm and a sliding table. The frame supports the impacting arm, the compressed air canister and piston (Fig. 2a). The impacting arm (13.01 kg) was and the impact velocity was measured by a time gate just prior to impact. The mass of the impacting arm is similar to that calculated for shoulder-to-head impacts in ice hockey reconstructions.[66] The end of the impacting arm was fitted with two different strikers to simulate stiff elbow collisions and shoulder-to-head impacts. A striker consisting of hemispherical nylon pad with a 35.71 ± 0.01 mm thick vinyl nitrile 602 foam disk underneath was used to simulate stiff elbow collisions (Fig. 2b). This striker produces similar peak linear and rotational acceleration to that of elbow strikes of ice hockey players.[5] To simulate shoulder collisions, the end of the impacting arm was fitted with a striking surface consisting of a nylon disc (diameter 13.2 mm) covered with 67.79 ± 0.01 mm thick layer of vinyl nitrile R338 V foam and a Reebok 11 k shoulder pad (Fig. 2c). This striker was found to represent a linear acceleration peak and duration of volunteer ice hockey players impacting the temporal region of a Hybrid III headform at low and high velocities.[66]

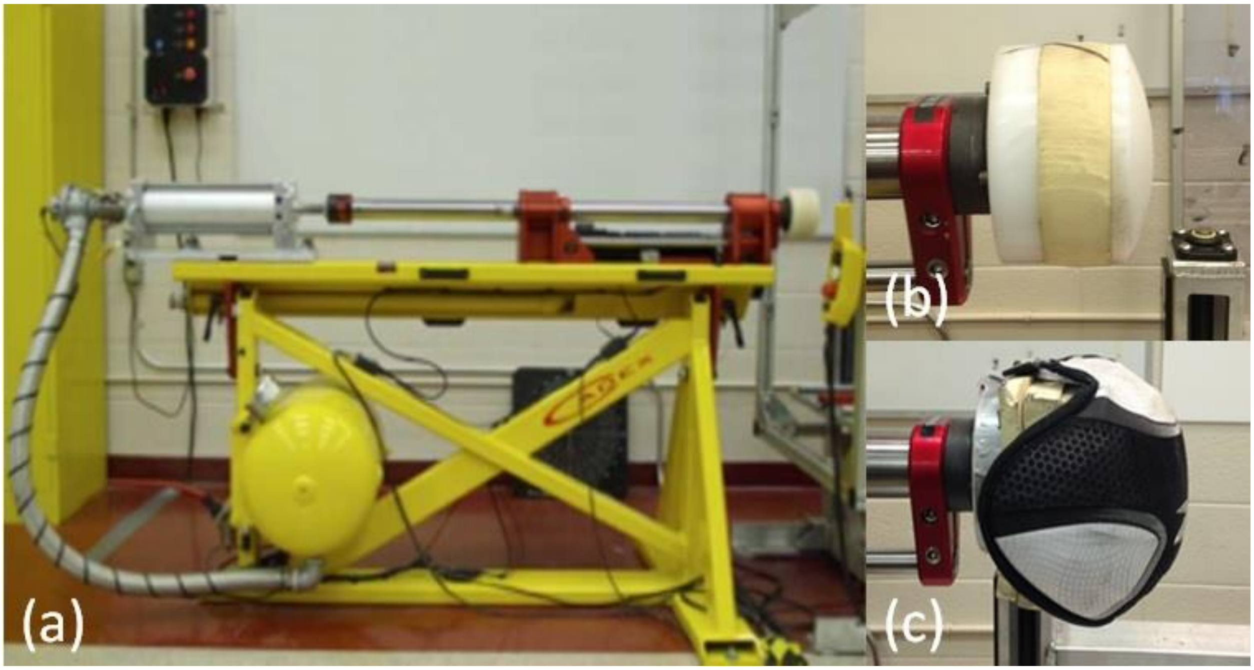

Figure 2. Pneumatic Linear Impactor: (a) frame supporting the impacting arm, (b) stiff elbow striker, (c) shoulder pad striker.

A pneumatic puck launcher (Fig. 3) was used to launch a puck in order to simulate puck to head impacts. The puck was fired by compressed air which is operated by an electronic control. The puck travels 0.6200 ± 0.0005 m down the barrel of the puck launch before it was released.

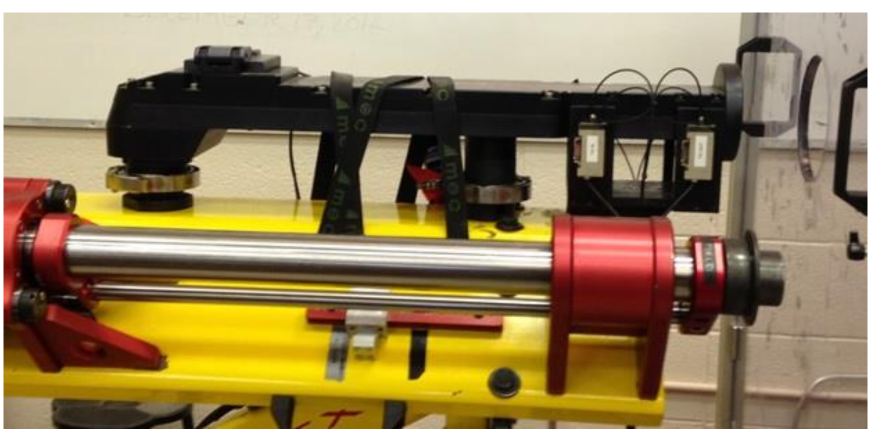

Figure 3. Pneumatic puck launcher.

For collisions and puck impacts the Hybrid III headform and unbiased neckform were attached to a sliding Table (12.78 ± 0.001 kg) to allow for movement post impact. The mass of the sliding table was chosen to represent the mass of the struck player which is typically less than the striking mass of a collision.[76,77] A foam braking system was used to stop the headform and sliding table after impact. A movable locking base on the table is 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 to remain fixed in position during testing.

Procedure

The headform was impacted under helmeted and unhelmeted conditions for four types of impact events. The headform was impacted at three velocities and two locations that were common in ice hockey collisions (Fig. 4).[27] The inbound velocities for falls and collisions were 3, 5 and 7 m/s. Collisions in ice hockey may occur at closing velocities outside this range, however these velocities were chosen as they represent commonly reported low to high skating speeds in ice hockey.[45] The inbound velocities for puck impacts were selected to be 20, 30 and 40 m/s. These velocities represented a range of low to high velocity shots in ice hockey.[15,51,70,84] For the helmeted impact conditions, the headform was equipped with a vinyl nitrile (VN) CSA and HECC certified ice hockey helmet. Ice hockey helmeted lined with VN foam are commonly worn by ice hockey players and used in research.[3,38,67,81] The particular name of the helmet make and model has not been identified and remains confidential, following the supplier's request that this research was intended to examine how an ice hockey helmet may perform under different impact events. A total of 138 impacts were performed in which three trials were conducted for each condition to establish the variance and repeatability of the results. Peak resultant linear and rotational accelerations of the headform were obtained. The resulting linear and rotational accelerations served as input to a finite element model of the head and brain. The model was used to calculate the magnitude of peak MPS in the cerebrum. To insure the integrity of the model's response, hourglass energies and aspect ratios were checked and were found to remain below the recommended values (10% of the total energy and 3:1, respectively).[22]

Figure 4. Impact locations and vectors on the ice hockey helmet as shown by the red arrows.

Finite Element Model

The model used in this study was the University College Dublin Brain Trauma Model (UCDBTM).[22,23] The head geometry of the UCDBTM was derived from computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver.[23] The model of the head and brain consists of 26,000 hexahedral elements representing the scalp, skull, pia, falx, tentorium, CSF, grey and white matter, cerebellum and brain stem.[22,23] The UCDBTM was validated against cadaveric pressure responses conducted by Nahum et al.[42] and brain motion research conducted by Hardy et al.[19] Further validations were conducted by Doorly and Gilchrist[10] and Post et al.[60] using reconstructions of real world traumatic brain injury incidents with results that were in agreement with anatomical tissue thresholds. The material characteristics of the model were taken from Ruan,[71] Willinger et al.,[83] Zhou et al.,[89] and Kleiven and von Holst[34] (Tables 1 and 2). The material behaviour of the brain tissue was modeled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus.[22] The compressive behaviour of the brain is considered elastic. The nature of the shear characteristics of the viscoelastic behavior of the brain was defined using the following equation:

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

where G_∞, is the long term shear modulus, G_0, is the short term shear modulus and β is the decay factor.[22] To simulate the skull brain interface, the cerebral spinal fluid (CSF) was modeled using solid elements with low shear modulus and a high bulk which allows it to behave like water. The contact interaction at the skull brain interface was assigned no separation and used a friction coefficient of 0.2.[40] The brain shear is modeled as hyperelastic and is represented by the following equation:

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

where C_{10} is the mechanical energy absorbed by the material when the first strain invariant changes by a unit step input and C_{01} is the energy absorbed when the second strain invariant changes by a unit step[37,39] and t is the time in seconds.

Table 1: Material properties of UCDBTM

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

Grey Matter	Hyperelastic	0.49	1060
White Matter	Hyperelastic	0.49	1060

Table 2: Material characteristics of the brain tissue for the UCDBTM

Material	G_0	G_∞	Decay Constant (s^{-1})	Bulk Modulus (Gpa)
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

Analysis

Peak resultant linear and rotational acceleration, MPS and impact duration were used to assess the performance of ice hockey helmets for four impact events. To assess the protective capacity of an ice hockey helmet, helmeted and unhelmeted impacts were compared in order to determine if the helmet was able to absorb impact energy, as measured by engineering parameters for a wide range of loading conditions representative of ice hockey head impacts. In order to make comparisons, the data was separated by impact events (falls, puck impacts, stiff elbow collisions and shoulder collisions), location and velocity conditions. Independent sample t tests were then conducted. For all comparisons the probability of making a type 1 error was set at a = 0.05. All data analyses were performed using the statistical software package of SPSS 19.0 for Windows.

Results

The protective capacities of the ice hockey helmet tested to reduce linear and rotational acceleration and MPS for different impact events are presented in Figs. 5, 6, and 7 and Table 3. Comparison of helmeted and unhelmeted impacts showed helmeted conditions produced significantly lower linear and rotational acceleration and MPS values for all fall and stiff elbow impact conditions ($p<0.01$). Helmeted shoulder impacts resulted in significantly lower peak linear acceleration compared to unhelmeted shoulder impacts for all conditions ($p<0.05$). Peak rotational acceleration and MPS produced by helmeted shoulder impacts were significantly lower compared to unhelmeted shoulder impacts for Site 2 ($p<0.05$) but not significantly different for Site 1. Puck impacts to a helmeted headform produced significantly lower peak linear and rotational accelerations compared to impacts to an unhelmeted headform ($p<0.01$) except for rotational accelerations at 40.0 m/s. MPS was significantly lower for helmeted puck impacts compared to unhelmeted puck impacts ($p<0.05$) except for Site 1 at 30 m/s and Site 2 at 40 m/s.

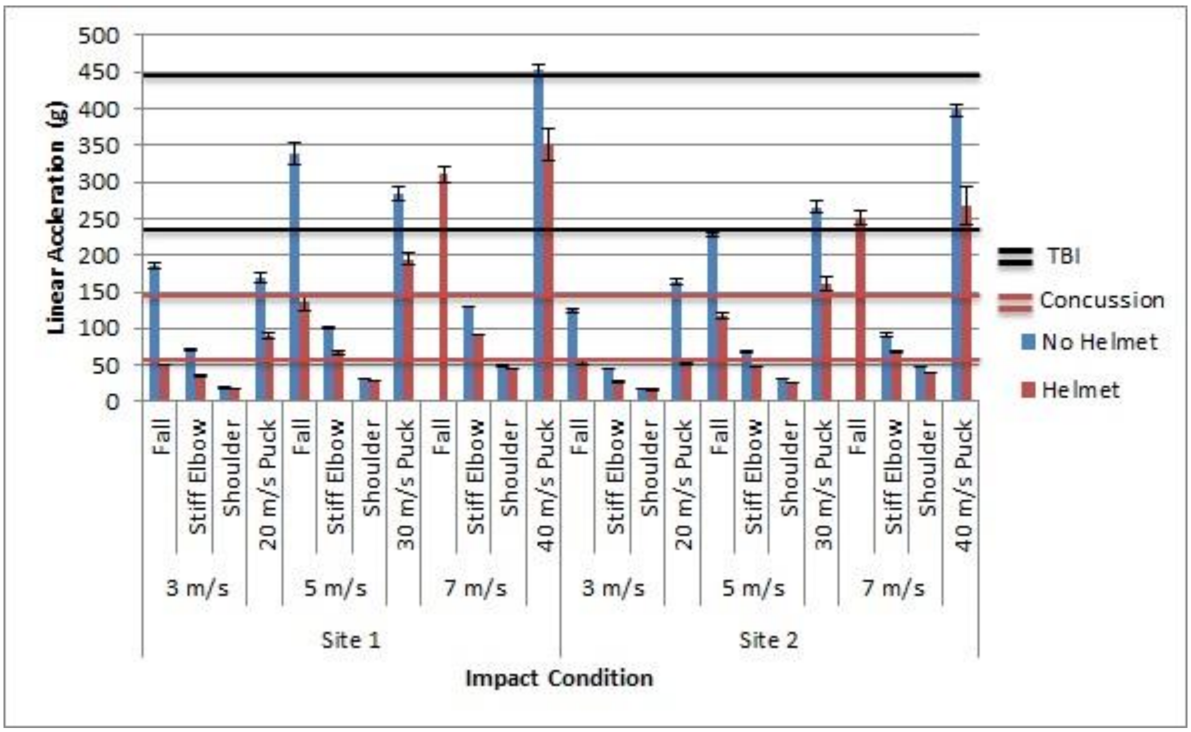

Figure 5. The protective capacity of the ice hockey helmet tested for four impact events as measured by linear acceleration. The error bars refer to 61 standard deviation. The two black lines represent reported range of traumatic brain injury (TBI)[8,61,75,86] and the two red lines represent the range of reported concussions.[43,78]

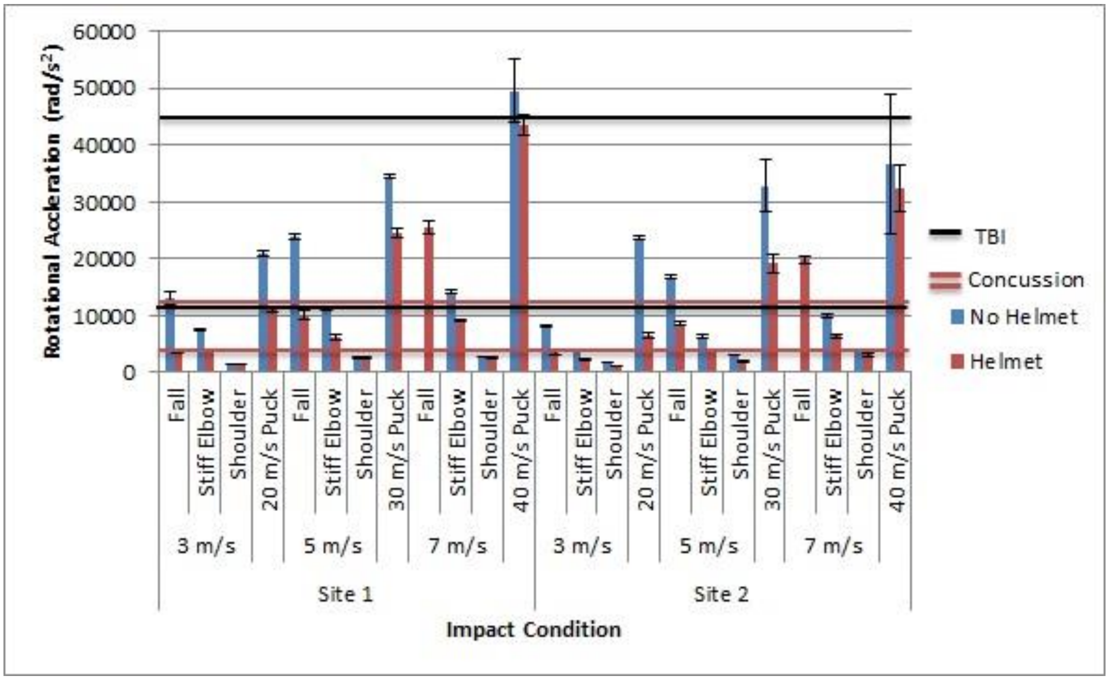

Figure 6. The protective capacity of the ice hockey helmet tested for four impact events as measured by rotational acceleration. The error bars refer to 61 standard deviation. The two black lines represent reported range of traumatic brain injury (TBI)[8,61,75,86] and the two red lines represent the range of reported concussions.[43,64,78]

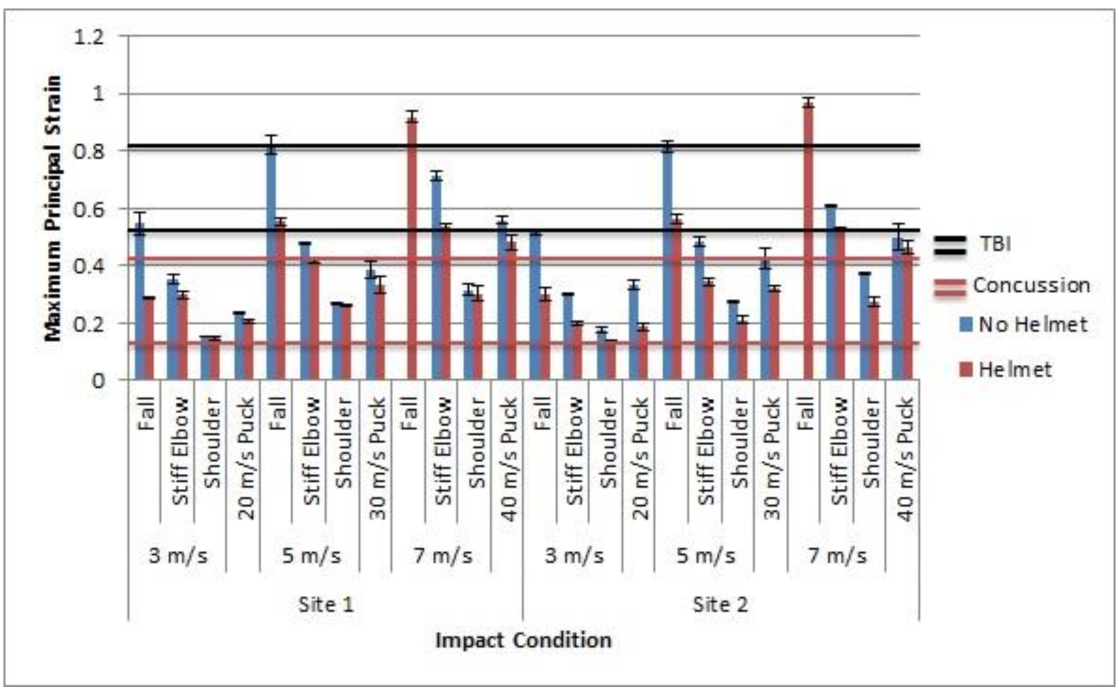

Figure 7. The protective capacity of the ice hockey helmet tested for four impact events as measured by MPS. The error bars refer to 61 standard deviation. The two black lines represent reported range of traumatic brain injury (TBI)[8,9,60] and the two red lines represent the range of reported concussions.[32,64,88]

Table 3. The protective capacity of the ice hockey helmet tested for four impact events.

Impact Event	Velocity (m/s)	Site	P Value Linear Acceleration	P Value Rotational Acceleration	P Value Maximum Principal Strain
Fall	3	1	0.001	0.001	0.001
		2	0.001	0.001	0.001
	5	1	0.001	0.001	0.001
		2	0.001	0.001	0.001
Stiff Elbow	3	1	0.001	0.001	0.008
		2	0.001	0.001	0.001
	5	1	0.001	0.001	0.001
		2	0.003	0.001	0.001
	7	1	0.001	0.001	0.001
		2	0.001	0.001	0.001
Shoulder	3	1	0.014	0.446	0.180
		2	0.003	0.001	0.001
	5	1	0.001	0.394	0.187
		2	0.001	0.003	0.002
	7	1	0.005	0.555	0.575
		2	0.001	0.012	0.001
Puck	20	1	0.001	0.001	0.001
		2	0.001	0.001	0.001
	30	1	0.001	0.001	0.098
		2	0.001	0.008	0.010
	40	1	0.002	0.148	0.010
		2	0.001	0.601	0.317

Tables 4, 5, and 6 present the duration of acceleration curves for different impact events. Comparison of helmeted and unhelmeted impacts showed helmeted conditions produced significantly longer impact durations for falls, stiff elbow collisions and puck impacts ($p<0.05$). No significant difference was found for impact duration between helmeted and unhelmeted shoulder collisions.

Table 4. The duration of acceleration curves for falls.

Site	Velocity (m/s)	No Helmet/Helmet	Duration (ms)
1	3	No Helmet	4.9 ± 0.2
		Helmet	9.8 ± 0.1
	5	No Helmet	4.1 ± 0.1
		Helmet	8.0 ± 0.1
	7	No Helmet	-
		Helmet	6.0 ± 0.1
2	3	No Helmet	7.2 ± 0.3
		Helmet	14.5 ± 0.5
	5	No Helmet	6.1 ± 0.1
		Helmet	10.0 ± 0.2
	7	No Helmet	-
		Helmet	8.6 ± 0.3

Table 5. The duration of acceleration curves for collisions.

Site	Velocity (m/s)	Injury Event	No Helmet/Helmet	Duration (ms)
1	3	Stiff Elbow	No Helmet	6.9 ± 0.2
			Helmet	9.9 ± 0.1
		Shoulder	No Helmet	24.8 ± 0.1
			Helmet	24.8 ± 0.2
	5	Stiff Elbow	No Helmet	6.9 ± 0.2
			Helmet	9.9 ± 0.1
		Shoulder	No Helmet	20.0 ± 0.1
			Helmet	20.0 ± 0.1
	7	Stiff Elbow	No Helmet	7.0 ± 0.2
			Helmet	9.9 ± 0.1
		Shoulder	No Helmet	19.8 ± 0.3
			Helmet	19.8 ± 0.2
2	3	Stiff Elbow	No Helmet	9.9 ± 0.1
			Helmet	15.1 ± 0.3
		Shoulder	No Helmet	25.0 ± 0.1
			Helmet	24.8 ± 0.1
	5	Stiff Elbow	No Helmet	9.9 ± 0.1
			Helmet	14.9 ±0.1
		Shoulder	No Helmet	20.1 ± 0.2
			Helmet	19.9 ± 0.1
	7	Stiff Elbow	No Helmet	9.9 ± 0.2
			Helmet	14.9 ± 0.2
		Shoulder	No Helmet	19.8 ± 0.1
			Helmet	19.8 ± 0.1

Table 6. The duration of acceleration curves for puck impacts.

Site	Velocity	No Helmet/Helmet	Duration (ms)
1	20	No Helmet	1.3 ± 0.1
		Helmet	2.1 ± 0.1
	30	No Helmet	1.3 ± 0.1
		Helmet	1.7 ± 0.2
	40	No Helmet	1.3 ± 0.1
		Helmet	1.5 ± 0.1
2	20	No Helmet	1.2 ± 0.1
		Helmet	2.4 ± 0.1
	30	No Helmet	1.3 ± 0.1
		Helmet	1.7 ± 0.2
	40	No Helmet	1.3 ± 0.1
		Helmet	1.5 ± 0.1

Discussion

The purpose of this study was to measure the protective capacity of ice hockey helmets for different impact events. A comparison of no-helmet to helmet for all impact events found that significant differences existed for most conditions; however, the decrease of engineering parameters to below reported levels for concussion and TBI varied. The ice hockey helmet tested reduced these engineering parameters below reported ranges for both concussion and TBI for falls up to 5 m/s.[8,32,43,60,61,62,64,75,86,88] At 7 m/s helmeted falls resulted in engineering parameters within reported ranges TBI[8,9,60,61,75,86] suggesting the ice hockey helmet is being impacted beyond its functional range.[31] Ice hockey helmets are designed to pass certification standards replicating the injurious impact events examined by Gurdjian et al.[18] which involved cadaver head drops to a rigid surface. The results of this study support the capacity of the tested ice hockey helmet to reduce engineering parameters from falls to hard surfaces such as ice and, as a result, TBI has largely disappeared from the sport.[24]

Overall, the ice hockey helmet tested decreased linear and rotational acceleration to levels below the reported ranges of concussion and TBI for puck impacts at 20 m/s.[9,43,60,61,62,64,75,86,88] When higher velocity (30 m/s) puck impacts were assessed, ice hockey helmets decreased linear acceleration below reported ranges of TBI.[9,61,75,86] Despite linear acceleration for 40 m/s impacts and rotational acceleration for 30 m/s impacts being significantly lower for helmeted conditions compared to unhelmeted conditions, helmeted puck impacts remained within the range of TBI.[9,61,75,86] In addition it should be noted that there is a large amount of variance in rotational acceleration for puck impacts at 40 m/s, which may have affected the statistical results comparing helmeted and unhelmeted impacts. This variation could have been a result of the physical geometry of the helmet deflecting the puck slightly differently for the same nominally identical impact conditions, resulting in more pronounced differences in the rotational acceleration responses. Nevertheless, it can be observed that the error bars remain within or above the reported range of TBI for both helmeted and unhelmeted impacts(Fig. 6).[9,61,75,86] As a result, the tested helmet may not provide adequate protection from TBIs for high velocity puck impacts. These results are consistent with previous research, as ice hockey helmets have been found to reduce engineering parameters for low velocity puck impacts but not for high velocities.[49,66] However when MPS was assessed, the ice hockey helmet that was tested was unable to reduce MPS to below the levels of concussion and TBI reported in the literature [32,60,62,88] except in the site 1 at 40 m/s and site 2 at 20 m/s conditions. As such, the tested ice hockey helmet adequately reduced this engineering parameter in a limited number of conditions.

The ice hockey helmet reduced peak linear accelerations better than peak rotational accelerations for collision impacts.[67,79] Unhelmeted and helmeted shoulder impacts produced peak linear and rotational accelerations below reported ranges of concussion.[43,64,88] However when examining the results for MPS,[60,62,64,88] collisions represented values within a range of concussion and the ice hockey helmet tested was unable to reduce MPS values below this range.

For shoulder collisions neither the magnitude nor duration of the loading curve was affected by the helmet. The inability of ice hockey helmets to reduce engineering parameters for shoulder impacts found in this study may reflect the high compliance of the human shoulder surrogate used in this study. For high compliance impacts such as shoulder impacts, most of the energy is absorbed by the shoulder and the helmet absorbs relatively little energy. This results in the helmet having a limited capacity to reduce engineering parameters. Whereas an ice hockey helmet was able to absorb energy for the impact for stiff elbow collisions, as the helmet lowered the magnitude and increased the duration of the loading curve. However, longer duration impacts have been suggested to cause high brain stress and strain.[16,82] Such a phenomenon has been noted in the reconstruction of real world concussions, as ice hockey collisions produce lower peak linear and rotational accelerations and longer impluse durations compared to American football helmet-to-helmet collisions but result in similar brain stresses and strains.[32,64,88] In this study helmeted collisions in ice hockey resulted in longer duration impacts with similar MPS values when compared to unhelmeted collisions.

As a result, this study is indicative of the protective capabilities of the tested ice hockey helmet against falls and elbow impacts. This would suggest that the ice hockey helmet is currently designed to manage only a small range of the loading conditions seen in ice hockey. An opportunity exists to improve helmet designs so that ice hockey helmets can manage a larger range of loading conditions as represented by the impact events examined in this study. As a result, future helmet designs and test methods should account for these differences when evaluating the performance of ice hockey helmets.

Limitations

The present research and results should be considered in light of its limitations. The Hybrid III headform is not a human head and may not imitate the dynamic properties of a human head.[7,72] Despite such limitations, the Hybrid III headform is widely accepted and used as a human head surrogate. While cadaveric head drops have shown a high degree variability,[46,48] the Hybrid III headform produces highly repeatable and correlated data compared to cadaveric head impacts.[30] Additional, the unbiased neckform used in this study may affect the response of the head as the neck might not imitate the exact response of the human neck. The unbiased neckform has not yet been validated for centric and non-centric impacts. The unbiased neckform was modelled after the Hybrid III neckform. However, unlike the Hybrid III neckform, the unbiased neckform is designed to respond the same in all directions to eliminate any potential bias in the response to an impact. The headform is attached to a low resistant sliding table to allow for movement of the headform post impact. The effect of such a table on the impact mechanics has not yet been well defined. A single model of an ice hockey helmet was tested. Differences in shell and liner design between helmet models have been found to cause differences in performance.[49,67,68] However, performance differences are small in magnitude.[49,67,68] One single ice hockey helmet model which represents a commonly worn design was tested,[3,38,67,81] as a representation of how ice hockey helmets may commonly perform in real world situations. The shoulder pad striker used in this study may not be an exact match of a human shoulder. The development of this striker was based on volunteer ice hockey players impacting a free-hanging Hybrid III headform,[65] which may not represent the exact response of the head in real world shoulder-to-head collisions. As a result, comparisons made with this shoulder pad striker are meant to be representative of how the head may respond due to shoulder collisions in ice hockey. The UCDBTM is a partially validated model in which the brain stem has not yet been validated for finite element modeling. As such, brain deformation metric analysis was limited to the cerebrum. Further, the response of the UCDBTM is dependent on the material properties that each respective part of the head/brain complex has been assigned. As a result the response of the UCDBTM is meant to be representative of how the brain may deform under the loading scenarios and may not represent the exact motion of the brain. Finite element analysis was limited to measuring MPS, which represents a single mechanism of brain injury and does not include other TBIs such as contusions or skull fractures which may be more likely with puck impacts.

The pneumatic linear impactor was not used to simulate falls to the ice as it would represent different impact mechanics than experienced in falls. The pneumatic linear impactor is designed to propel a defined impact mass towards the headform and push the headform out of the way as the impacting arm moves along its original path. This may not represented the way in which the head impacts an immovable surface with infinite mass during a fall against the ground. Such impact mechanics are better represented using a monorail drop rig as performed in this study.

Conclusion

This study examined the protective capacities of ice hockey helmets for different impact events. When comparing helmeted and unhelmeted impacts it was found that the helmet tested was effective at reducing linear and rotational acceleration for falls up to 5 m/s and elbow collisions to below the reported ranges of concussion and TBI, but was less so for shoulder collisions and puck impacts. However, when MPS was assessed, the ice hockey helmet only reduced values to below the reported ranges of concussion and TBI for falls up to 5 m/s. This result demonstrates that the ice hockey helmet is not designed to manage impact conditions other than those which represent a drop to a rigid surface. This suggests that an opportunity exists to improve the protective capacity to account for a wider range of impact conditions seen in ice hockey.

Conflict of interest

The University of Ottawa holds research agreements for testing and development of ice hockey helmets with the helmet manufacturer who supplied the helmets for this study.

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