

## Comparison of Helmeted Head Impact in Youth and Adult Ice Hockey

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**Abstract** Concussion is a common injury with potentially serious outcomes following head impacts during ice hockey. Currently little is known about the differences resulting in concussion for youth ice hockey players in comparison to adults, which biomechanical factors contribute to brain injury risk, and the implications for helmet development and design. The objective of this research was to describe the mechanisms of injury for both adult and youth ice hockey players. Helmeted 50<sup>th</sup> percentile (Adult: 18+) and 5<sup>th</sup> percentile (Youth: 9-11years) Hybrid III headforms were impacted according to the masses and velocities representative of two common mechanisms of injury in ice hockey: falling and collisions. Five sites were impacted for collisions, and three for falling, with linear and rotational acceleration the measured output. A finite element model of the human brain was used to calculate maximum principal strain values for the impacts. The results indicated that there is a risk of concussion for adults for falls and collisions, whereas the risk for the youths resided in the falling events.

**Keywords** adult, concussion, helmet, ice hockey, youth

### I. INTRODUCTION

Ice hockey is a sport engaged in by all ages that has been reported to have a high incidence of concussion [1-2]. The youth that engage in the sport are at particular risk as they have been identified to be more susceptible to long term effects of this injury affecting their future education and emotional status [3-4]. Much of what is currently known in ice hockey concerning concussion risk and helmet development is based upon adult parameters, parameters of mass and velocity that are not reflective of the way youth play the game. As a result, current helmet technologies are developed to primarily manage energy based upon standards and methods that are reflective of how adults play the game [5]. It is possible that current helmet designs do not adequately reflect the differences between the ways that adults and youth play the game, and as a result youth may not benefit from the improved protection that could exist if the helmets were designed for injury risks specific to their age.

The current Canadian Standards Association (CSA) ice hockey standard is a monorail drop onto an MEP anvil at 4.5 m/s, which is reflective of a fall from an adult height [5]. While there is some accommodation for head mass, as represented by the variable mass used in the standard, the impact parameters result in adult energies for purposes of helmet certification, and the materials used in the helmets are designed to manage these adult impact energies, not those of youth. The result of this standard is that helmets are designed with liners that absorb energy to pass the standard. Once the helmet has been designed to pass the standard, the resulting helmet liner is then re-sized to fit the range of helmet sizes, with the youth players typically wearing the smaller sizes. This results in an energy absorbing liner that was designed to manage the impact energies that reflect a risk of injury for an adult, but fitted to a small helmet worn by youth. As the liner was designed to absorb a higher energy than that likely to occur in youth ice hockey, the liner may not compress as efficiently under a load and result in a decrease in the protective capacity of the helmet in comparison to a liner that was designed specifically for youth impact conditions. In effect an adult helmet re-sized to a youth head size may not protect as well as if the energy absorbing foam was designed for loads incurred in youth sport. While falling injury events are reflected in the current adult ice hockey helmet standard, the most common injury event type for

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this age group has been identified as collisions, such as shoulder to head collisions [6]. These types (falling to ice and collisions) of impact mechanics have been investigated to determine how helmets function to reduce linear and rotational acceleration and maximum principal strain [7-10] as those parameters are commonly used to quantify risk of brain injury [11-13]. However, the impact parameters from these studies have focused solely on the way the adult game of ice hockey is played and very little is known concerning how ice hockey helmets perform to protect young players in the environment in which they play the game. As youth have lower body masses [14], smaller heads, and skate at slower speeds it would be assumed that a helmet designed for an adult environment may not be suitable for a youth ice hockey environment. The purpose of this research is to examine how helmets perform for adult and youth impact parameters.

## II. METHODS

### *Experimental Testing*

#### *Determination of parameters*

The parameters of the reconstructions (velocity, mass, compliance) for the adult (18+) and youth (9-11 yrs) impacts were determined from the literature. For the adult collision helmet impacts, the velocity (6.5 m/s) was determined from the literature describing impact velocities in the game of ice hockey [15-16]. The mass (13.1 kg) and compliance was determined from the literature examining a shoulder to head impact for ice hockey [17]. The adult fall head contact velocities were derived from CSA standard that represents a fall to an ice surface (4.5 m/s), which is within the range of reported falling velocities for adults [18-19]. The youth collision velocities were determined from literature describing the velocities of the game for that age range (3.0 m/s) [20], with the mass (9.6 kg) determined from effective mass calculations derived from [17] and [14]. The youth fall head contact velocities were derived from the literature characterising paediatric brain injury for a heights (1.5 m) and weights typical of the 9-11 year age group [19][21]. From this literature the average falling head contact velocity of 3.8 m/s for the 9-11 years age group was selected.

A Hybrid III 50<sup>th</sup> headform was used for the impact testing to represent an adult head size and geometry. The Hybrid III 5<sup>th</sup> headform was chosen for the impact testing under the youth parameters as its size and geometry was similar to that of youth aged 9-11 years [14]. As the Hybrid III neck is known to create a biased response for impact testing, the headforms were attached to the testing equipment via an unbiased neckform, which was constructed of alternating symmetrical aluminum and rubber discs [22].

#### *Collisions*

The adult collision impacts were conducted using a linear impactor system (Fig 1). The linear impactor consisted of a steel frame that was permanently affixed to the concrete floor of the laboratory. The parts of the linear impactor included a cylindrical impacting rod (13.1 kg  $\pm$  0.01) attached to the frame with ball bushings and a compressed air canister that released air to launch the rod to simulate a collision impact. In the case of collisions, both the adult and youth impacts had the headform attached via the unbiased neckform to a 12.78 kg  $\pm$  0.01 kg sliding table that allowed post impact movement [12]. The mass of the table was based on the literature on collisions on American football [12]. The table and hydraulic frame also allowed for positioning of the headform in 5 degrees of freedom to adjust the headform for the correct impact sites. The striker on the end of the linear impactor rod was comprised of vinyl nitrile (VN) foam and matched the compliance of a padded shoulder impact to the head in ice hockey as described in [17]. The velocity of impact was measured using a photoelectric timegate 0.02 m before the impact. For the helmet impacts, a medium VN hockey helmet was affixed to a 50<sup>th</sup> Hybrid III headform and impacted in five locations (Fig 2), three times per location.



Fig 1. Linear impactor used for the adult collisions with the impact cap that simulated the compliance of a shoulder.

The youth collisions were conducted using a pendulum system because the lower mass was not attainable using the linear impactor system. The pendulum was a steel cylinder (9.6 kg) that was attached to the ceiling using a four wire system that allowed for a stable and predictable delivery of the impact to the headform. The same compliant shoulder impact cap was used for the youth helmet impacts as for the adult impacts. The velocity of impact for the pendulum was measured using a PCI 512 high speed camera (Photron, San Diego, USA) sampling at 1000 fps. A small VN helmet (same thickness of VN as the adult helmet) of the same model as that used for the adult helmet collisions was affixed to a 5<sup>th</sup> Hybrid III headform and impacted in the same 5 locations on the helmet (Fig 2), three times per location.

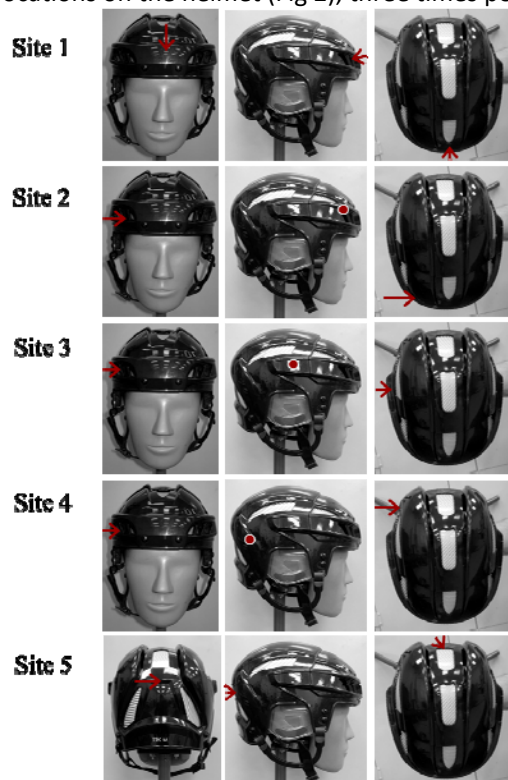


Fig 2. Impact locations on the helmet for the collision impacts

### Falls

The adult and youth fall helmet impacts were conducted using a 4.7m long monorail. The headform (50<sup>th</sup> for adult, 5<sup>th</sup> for youth) was affixed to the monorail carriage via an unbiased neckform and dropped at the height that would produce the desired impact velocity. The velocity was measured 0.02 m prior to impact by photoelectric timegate. The anvil for the monorail impacts was ice that was frozen at -25 degrees C for 48 hours before impacting, and was re-conditioned for 15 minutes between impacts to prevent melting of the surface

(Fig 3). Sites 1, 3, and 5 (rear, through the centre of gravity) were impacted for the falls (Fig 2). Sites 2 and 4 were not impacted as the headform could not be adjusted to impact these sites on the monorail. Three impacts were conducted per site.



Fig 3. Helmeted Hybrid III headform monorail impact onto the ice anvil for the rear location.

### *Signal processing*

The headforms were outfitted with nine Endevco 7264C-KTZ-2-300 accelerometers (Meggitt, Irvine, USA) setup into a 3-2-2-2 array for measurement of linear and rotational accelerations [23]. The signals were sampled at 20 kHz and collected by Diversified Technical Systems (DTS) Pro lab module and processed using TDAS software (Diversified Technical Systems, Seal Beach, USA). The signals were filtered using a CFC 1000 filter.

### **Computational Modelling**

#### *The University College Dublin Brain Trauma Model*

To determine the maximum principal strain magnitudes for the ice hockey helmet impacts the University College Dublin Brain Trauma Model (UCDBTM) was used. The UCDBTM is a finite element model and is one of the few used for brain injury research [24-25]. The geometry of the model was determined from medical imaging of the head of a male cadaver, and had the following sections: scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum, and the brain stem [25]. In total, the UCDBTM has approximately 26,000 hexahedral elements.

The material properties of the model were developed from cadaveric anatomical testing and tissue sample analyses [26-30] (Tables 1 and 2). The tissues of the brain were modelled using a linearly viscoelastic model combined with large deformation theory. The behaviour of the brain tissues were characterised as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [24]. The compression of the brain tissue was defined as elastic. The shear characteristic of the viscoelastic brain was expressed:

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

with  $G_{\infty}$  representing the long term shear modulus,  $G_0$  the short term modulus and  $\beta$  is the decay factor. A Mooney-Rivlin hyperelastic material model was used for the brain to maintain these properties in conjunction with a viscoelastic material property in ABAQUS (Dassault Systems, Waltham, USA), giving the material a decay factor of  $\beta = 145 \text{ s}^{-1}$  [24]. The hyperelastic law was given by:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)} \quad (2)$$

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 and  $t$  is the time in seconds [31-32]. The brain skull interaction was defined as sliding with no separation between the pia and the CSF. The CSF was modelled by employing solid elements with a bulk modulus of water with a low shear modulus [24-25]. The coefficient of friction for the sliding interface was 0.2 [32].

Validation of the model was accomplished by comparing the UCDBTM's responses to those of cadaveric testing conducted by [34] and [35]. Further examinations of the models' responses were conducted using brain injury reconstructions from real-life incidents by [18][36-38] which achieved good agreement with the magnitudes of strain and stress in the literature.

TABLE I  
MATERIAL PROPERTIES FOR UCDBTM

Material	Young's modulus (Mpa)	Poisson's ratio	Density (kg/m <sup>3</sup> )
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx	31.5	0.45	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 II  
MATERIAL PROPERTIES OF THE BRAIN TISSUE USED IN THE UCDBTM

	Shear modulus (kPa)		Decay constant (s <sup>-1</sup> )	Bulk modulus (GPa)
	$G_n$	$G_{sc}$		
Grey matter	10	2	80	2.19
White matter	12.5	2.5	80	2.19
Brain stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

### Scaling of the UCDBTM

The UCDBTM is an adult-sized brain model that is commonly used for brain injury research using the dynamic response time histories for a 50<sup>th</sup> Hybrid III headform. In this research the full sized version of the UCDBTM was used for determining the adult collision and falling maximum principal strain values for the helmeted headform impacts. Currently there is little agreement concerning the parameters that should be used for the tissue definitions of a youth head model, and some suggest that the brain tissues reach adult characteristics by the ages of 9-11 years old [39-41]. Given these considerations, the UCDBTM was scaled to a similar brain size to that which has been reported for the age group identified in this research (9-11 years old) based on Magnetic Resonance Imaging (MRI) research of the brain [42]. This resulted in the full size UCDBTM being scaled to 95% of its adult dimensions uniformly in all axes. Scaling was chosen as opposed to warping the mesh to reduce the likelihood of element errors during the analysis. Even so an aspect ratio and hourglass energy check was conducted on the resulting scaled model to confirm the model's integrity. From a laboratory comparison this reduction in size (by 5%) of the model tended to reduce the magnitude of MPS response by 4%.

### Statistics

Comparisons were conducted examining linear and rotational acceleration and maximum principal strain (MPS) responses for impact type (collision and fall) and age group (youth and adult) by means of an ANOVA. A Tukey post-hoc test was conducted when significant main effects were identified. The confidence interval was set to 95%.

## III. RESULTS

The results of the impacts to the helmeted adult and youth for falls and collision type impacts are presented in Tables 3 and 4. A graphical comparison detailing the described risk of concussive injury from the literature can

be found in Figs. 4 and 5. Significant main effects were found for all comparisons ( $p < 0.05$ ). The youth and adult collisions had significantly lower magnitudes of response in comparison to the falls. In comparison between the adult and youth responses, no significance was found between rotational acceleration and MPS for the fall impacts between these two groups ( $p > 0.05$ ). In addition, the rotational accelerations were not found to be significantly different between the adult and youth collision impacts ( $p > 0.05$ ). All other comparisons between the adult and youth responses were significantly different ( $p < 0.05$ ).

TABLE III  
COMPARISON OF RESPONSES BETWEEN THE ADULT AND YOUTH FALL IMPACTS

	Velocity (m/s)	Peak resultant acceleration		MPS
		Linear (g)	Rotational (rad/s <sup>2</sup> )	
Youth	3.8	83.3 (8.0)	6198.7 (2665.4)	0.435 (0.12)
Adult	4.47	131.4 (18.2)	5868.6 (2114.3)	0.404 (0.13)

TABLE IV  
COMPARISON OF RESPONSES BETWEEN THE ADULT AND YOUTH COLLISION IMPACTS

	Velocity (m/s)	Peak resultant acceleration		MPS
		Linear (g)	Rotational (rad/s <sup>2</sup> )	
Youth	3.0	14.7 (0.7)	1269.4 (257.1)	0.143 (0.02)
Adult	6.5	32.5 (5.3)	2661.4 (653.5)	0.256 (0.04)

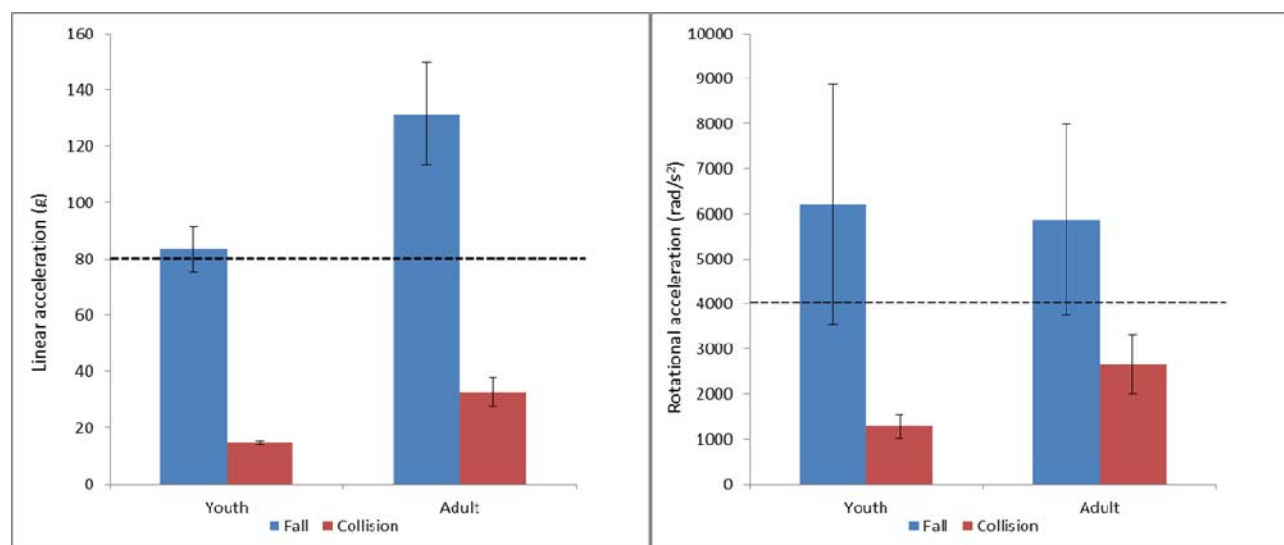


Fig 4. Linear (left) and rotational (right) acceleration comparisons for the fall and collision impacts. Black dotted line denotes magnitudes above which brain injury has been reported to occur in the literature for adults [11-12][43-46]

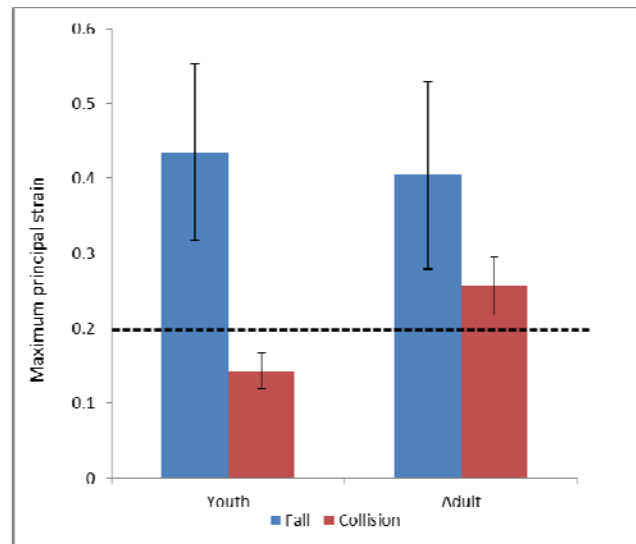


Fig 5. Maximum principal strain comparisons for the fall and collision impacts. Black dotted line denotes magnitudes above which brain injury has been reported to occur in the literature for adults [11-13][47].

#### IV. DISCUSSION

Significant differences were found between the adult and youth for the helmeted headform impact conditions. When examining just the adult impact condition, there was a risk of brain injury reflected by the linear and rotational acceleration and MPS magnitudes for the falling impacts, whereas risk was identified through only MPS for the collision impacts (Figs 2 and 3). This may be a reflection of the longer duration of the linear and rotational accelerations of the collision events, which were typically around 20-25 ms. These long duration acceleration pulses result in strains in the brain tissues at lower magnitudes of peak acceleration. This phenomena has been described previously by [48] through impacts using primate subjects, demonstrating that concussion often occurred through longer duration acceleration pulses. Research conducted by [49] and [50] have also identified this relationship, as well as [51] using finite element and computational methods. The youth impacts demonstrated that for the parameters that describe the typical falling and collision events that the risk of brain injury would be primarily from falls to the ice. While the collision events were long duration accelerations to the brain (20 ms), the low magnitudes (14.7 g and 1.3 krad/s<sup>2</sup>) were not sufficient to create a damaging strain environment (0.14 MPS) within the model. This may be a reflection of the lower velocities and masses that are involved in the collision events for youth ice hockey players [14]. It is likely that the conditions of the impact event do not create enough of a transfer of energy to the helmet and head to create a high risk of brain injury. This result coincides with research involving concussion in youth ice hockey that identified falls to the ice as being the primary cause of brain injury [2]. As this is a frequent event for youth playing ice hockey, future helmet designs should reflect fall to ice impact events under youth impact conditions to aid in the improvement of helmet designs. The improvement in impact absorbing capacity could be gained from adjusting the density of the impact absorbing liner to perform within the youth ranges of energy of impact. Making adjustments to the shell may also be beneficial; however improvements of the liner would likely be the easiest route with the largest effect. In addition the current ice hockey helmet standard would need modification to account for the energies that the youth players are encountering when a brain injury is likely to occur. For adults, ice hockey helmet development and standards may consider focusing on collision events in addition to falls, as collisions have been reported as the event most likely to result in concussion for that age group [6]. This improved protection for adults may reside in a stiffer liner for falls, and possible a secondary rotational acceleration damping technology to help mitigate injury from shoulder contacts.

This research should be interpreted with certain limitations. The tolerance literatures used as points of comparison in this research are from human adult reconstructions. It is possible that the tolerances to brain injury for youth may differ somewhat from the adult magnitudes. To date there is no human youth concussion

threshold data available with which to compare this data to, which is why adult data was used to help put the results in context. The headforms are commonly used for impact research, and while they have been described to produce results in the range of cadaveric testing may not provide completely biofidelic responses [52]. The impact parameters were determined through measures from literature sources and are representative of the impacts for the target age groups but do not represent actual brain injury reconstruction. The results are specific to those assumptions and representations from the literature of impact mass, velocity, compliance, and location. In addition, the UCDBTM and scaled version are both dependent on the material properties and conditions that define the interactions of the represented tissues and all results from these models should be considered in light of those limitations.

## V. CONCLUSIONS

This research identifies adult ice hockey players are likely at risk of concussion from both collisions and falls, whereas youth are at risk primarily from falls. These results suggest that to improve protection from brain injuries for youth that the helmet liner density should be adjusted to maximize the energy absorbing capacity of the helmet for falls. For adults, it is likely that improvements of helmet design needs to focus on both falls and collision conditions for improved protection for concussion. New helmet designs may need to focus on dual energy absorbing systems, one for linear falls, and the other for rotational impacts that are common to shoulder to head contacts. In addition, to improve protection new brain injury tolerances specific to youth must be researched to determine appropriate thresholds to target to optimize helmet designs and standard development.

## VI. ACKNOWLEDGEMENT

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