P. Rousseau[1], *T.B. Hoshizaki*[1], *and M.D. Gilchrist*[1,2]

For ASTM F-08: Protective Capacity of Ice Hockey Player Helmet Against Puck Impacts

---

ABSTRACT: Many studies have assessed the ability of hockey helmets to protect against falls and collisions, yet none have addressed the injury risk associated with puck impacts. Thus, the purpose of this study was to document the capacity of a typical vinyl nitrile ice hockey helmet to reduce head accelerations and brain deformation caused by a puck impact. A bare and a helmeted Hybrid III male 50th percentile headform were struck with a puck three times to the forehead at 17, 23, 29, and 35 m/s using a pneumatic puck launcher. Linear and rotational accelerations were captured using accelerometers fitted in the headform and used as input in the University College Dublin Brain Trauma Model to obtain brain deformation. The helmet reduced peak resultant linear acceleration, peak resultant rotational acceleration, and maximum principal strain but a comparison with published brain injury risk curves shows that it did not reduce the concussion risk below 50 % for impacts at or above 23 m/s. Thus, a vinyl nitrile ice hockey helmets can protect players from direct puck impacts in amateur and youth leagues but may not be adequate in competitive elite leagues, where the puck can be shot at velocities well above 23 m/s. Furthermore, competitive adult male ice hockey players struck to the helmet by a puck may need to consider changing their helmet as it was shown that direct impacts above 29 m/s decreased the helmet's ability to reduce head peak linear acceleration in subsequent impacts.

KEYWORDS: Finite element modeling, Head injury, Helmet, Ice hockey, Injury reconstruction, Puck impact

[1] Neurotrauma Impact Science Laboratory, University of Ottawa, Ottawa ON, Canada
[2] School of Mechanical and Materials Engineering, University College Dublin, Dublin, Ireland

Introduction

Ice hockey was identified as the contact sport with the highest rate of concussion per participant [1]. Increased media scrutiny has brought attention to the concussion risks at the professional level, yet it is important to recognize that young players are also at risk with a recent study reporting a 25 % concussion rate at the junior level (16-21 year old) [2].

Hockey players are typically injured to the head following body checking, collision with the boards or the net, contact with the ice, or contact with the puck [3-4]. Perhaps due to a lower incidence when compared to the other mechanisms, little research has been done to assess the capacity of ice hockey helmets to protect the head against direct puck impacts. While helmets are known to reduce head accelerations and brain tissue deformation in player-to-player collisions [5], the protective ability of helmets cannot be generalized to puck impact due to its distinct characteristics. Ice hockey pucks are low compliance discs composed of vulcanized rubber, weigh between 160 and 170 g, and travel at very high velocities.

While the official National Hockey League record for a slap shot was set at 48 m/s (174 km/h), skilled amateur players can shoot the puck at velocities above 30 m/s (108 km/h) [6-9]. More specifically, young players aged 11 and 12 (pee-wee) were reported to be able to shoot the puck at a maximum velocity of 19.2 ± 2.9 m/s while teenage players aged 15 and 16 years old (midget) could shoot the puck at a velocity of 26.1 ± 1.6 m/s [8]. Women's recreational players (23.0 ± 4.7 y.o.) were reported to be able to shoot a puck at a velocity of 13.3 ± 2.0 m/s while women having competed in the Canadian university league (19.1 ± 1.7 y.o.) could shoot the puck at 18.8 ± 2.6 m/s [9]. Men's recreational players (25.4 ± 7.3 y.o.) were reported to be able to shoot a puck at a velocity of 23.3 ± 3.9 m/s while competitive men, competing at the Canadian university level (22.8 ± 1.6 y.o.) were reported to be able to shoot at a velocity of 30.0 ± 2.6 m/s [9].

During a direct impact to the head, momentum and energy are transferred creating forces which modify the head's motion and deform the brain. The motion of the head, quantified using linear and angular acceleration, can be obtained by using accelerometers inserted in an anthropomorphic headform [10]. While peak acceleration may not fully describe the complex nature of head injury, it's linear and angular components were found to be sufficiently associated to the onset of head injury to be used as injury criteria [11-17]. Nevertheless, acceleration only describes the rapid change in motion of the head and does not provide accurate insight into the movement of the brain.

Cerebral tissue deformation can be estimated using advanced finite element models. These models decompose the brain into small elements, each having the material properties of the tissue it represents. Thus, internal stresses and subtle internal motions of the modelled structure can be analyzed to predict injury [18]. The mechanical (structural) and physiological (functional) tolerance levels of brain tissue are characterised using the relationship between stress and strain. For example, an artery subject to tensile stress greater than its tensile strength will tear (mechanical failure), causing intracranial haematoma [19]. Physiological failures occur at lower levels and may temporarily disturb nerve activity such as in concussion [20]. Maximum principal strain (MPS) is a measure of tissue deformation which has been correlated to loss of nerve functionality [20-22] and concussion in sports [14,17]. Accordingly, MPS should be considered when assessing head injury risk.

Ice hockey helmets designed to protect against high mass, low velocity collisions and falls and may not adequately protect a player's head against a low mass, low compliance puck travelling at high velocity. Thus, the purpose of this study was to document the capacity of an ice hockey helmet to reduce head accelerations and brain deformation following a puck impact.

Methodology

A Hybrid III 50th percentile headform was struck by a puck (0.166 kg) propelled using a pneumatic puck launcher (See Fig. 1). The bare headform was struck three times at each of the following five velocities: 17 m/s, 23 m/s, 29 m/s, and 35 m/s. The velocities were chosen to approximate the shooting ability of players aged 12 and under, collegiate women, males between the age of 13 and 18, collegiate men and elite players, respectively. The headform was first impacted unhelmeted to provide a comparison point to determine the ability of the hockey helmet to reduce head acceleration and brain tissue deformation.

The headform was then equipped with a CSA certified medium-sized helmet and impacted at the same four velocities. The helmet's shell consisted of two acrylonitrile butadiene styrene (ABS) pieces which allowed the helmet to be adjusted to fit the Hybrid III headform. At the impact site, the vinyl nitrile (VN) foam liner had a thickness of 21 mm. A new helmet was used at each velocity to limit the effect of material degradation. The results were analysed using a detailed finite element (FE) model of an adult male head. Brain tissue deformation was assessed using peak maximum principal strain (MPS) which is often used to represent brain deformation and has been identified as a possible indicator of concussion [17,23-25].

Laboratory set-up

The headform was attached to a Hybrid III neckform and positioned in front of the pneumatic puck launcher. The neckform was pitched forward 7.5° and the headform was tilted backward 5° to mimic the natural head position [26]. All impacts were located 30 mm above the anterior intersection of the mid-sagittal and absolute transverse planes.

The Hybrid III headform was equipped with nine single-axis Endevco accelerometers (Endevco, San Juan Capistrano, CA) adjusted orthogonally following a 3-2-2-2 array [10]. Head acceleration was sampled at 20 kHz and filtered using the SAE J211 class 1000 protocol [27]. Acceleration signals were processed using a TDAS Pro Lab system (DTS, Seal Beach CA). The average time between impacts was 5.00 ± 0.75 min.

Finite Element Analysis

The resulting linear and angular accelerations were used as input for the University College Dublin Brain Trauma Model [28-29]. The model included the scalp, three-layered skull (cortical and trabecular bone), dura, cerebrospinal fluid (CSF), pia, falx, tentorium, cerebral hemispheres, cerebellum and brain stem. To improve brain deformation distribution, separate representations of the grey, white, and ventricular matter were implemented. Geometry was determined using Computer Tomography (CT) data available through the US National Library of Medicine Visible Human Database and was not meant to represent a 50th percentile male head.

Shell elements were used to model the scalp, falx, and tentorium; brick elements were used to model the cortical bone, trabecular bone, CSF, cerebrum, cerebellum, and brain stem; membrane elements were used to model the dura and pia mater. In total, the model was built using over 26 000 elements. The model was validated against the pressure response of Nahum's [30] cadaveric impact test and the displacement response of Hardy's [31] high-speed x-ray cadaver impact test. The model was found to correlate well with experimental data for head injury using a force free boundary condition at the foramen magnum [32]. Changing this boundary condition may improve results; however, there is no evidence to suggest that this would be the case.

Various constitutive models have been used to describe the mechanical behaviour of brain tissue, including using an integral model in combination with hyperelasticity. In this present analysis, the neural tissue was simply characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus, while the compressive behaviour was considered as elastic. The CSF layer had a 1.3 mm depth and was modeled using solid elements with a low shear modulus and a high bulk modulus so that it would behave like a fluid. The boundary condition used at the brain/skull/CSF interfaces was a sliding boundary condition [33-34]. The remaining parts of the model were taken from the literature [28-29]. All material properties can be found in Tables 1 and 2.

Statistical Analysis

Independent-samples t-tests were conducted to compare each dependent variable (peak linear acceleration, peak angular acceleration, and MPS) in the bare and helmeted conditions. Statistical significance was set at an alpha of 0.05.

Results

Peak linear acceleration, peak rotational acceleration, and MPS obtained for each puck impact are reported in table 3. Peak linear acceleration, peak angular acceleration, and MPS were all significantly reduced ($p < 0.01$) at the four tested velocities. On average, the helmet was able to reduce peak linear acceleration, peak rotational acceleration, and MPS by 64, 54, and 53 %, respectively. Furthermore, for impacts at 17 and 23 m/s, the helmet managed to maintain peak linear acceleration below a 50 % risk of concussion. This was not the case for peak angular acceleration and MPS which were above a 50 % risk for velocities of 23 m/s and above. Interestingly, the second and third impacts generated higher peak linear acceleration than the first impact at the 29 m/s and 35 m/s (Figure 2) indicating a possible deterioration of the helmet's material. This trend was not apparent for peak angular acceleration and MPS.

Discussion

This study was designed to document the ability of ice hockey helmets to reduce head accelerations and computed brain tissue strain generated by a direct puck impact to the front of the head. The helmet was able to keep peak linear accelerations below 82 g which was associated with a 50 % risk of concussion for velocities below 23 m/s. This was calculated using a logistic regression analysis of 21 reconstructed head impacts which occurred in the National Football League [17] and is in agreement with research performed on animals [36-37]. While the proposed risk may be specific to head-to-head collisions in football, there is currently no data available which would indicate that the risk associated with a puck impact to the head would be different. The helmet was only able, however, to keep peak angular acceleration below a 50 % risk of concussion for impacts at 17 m/s. At 23 m/s, the peaks exceeded a magnitude of 5.9 krad/sec^2 which was obtained in the same study by Zhang and colleagues [17]. This could indicate that the helmet was not as effective at reducing angular acceleration.

Furthermore, the MPS computed for impacts at 23 m/s were within a 0.10 to 0.26 range associated with concussion [14,17,20-22,38-39]. The range was based on NFL game impacts reconstructions which reported peak MPS between 0.19 and 0.26 as having a 50% probability of causing concussion [14,17] and anatomical studies which studied the effects of strain on neuronal conductivity. By using squid giant axons, Galbraith et al. [20] were able to study the physiological response associated with an applied load without the interference of surrounding structures or physiological interactions. It was observed that following an elongation of 0.12 in 14 milliseconds the axon was unable to elicit an action potential response when stimulated. This loss of functionality was only transient and the action potential reappeared after a three min period of rest. Axons subjected to strains above 0.20 were found to never fully regain their resting potential while axons subjected to strains above 0.25 failed structurally. Axonal trauma was also studied using guinea pig optic nerve to determine the magnitude of strain necessary to cause functional and structural failure [21]. This would identify 23 m/s as the velocity at which a vinyl nitrile helmet begins to fail to protect the player against concussion.

The performance of the helmet decreased following the first impact at 34 m/s and above. This indicates that an adult male playing in a competitive league should change his helmet after being hit once by a puck. This is even more important for professional players where both the second and third impact generated peak acceleration above 250 *g* which is an average of the values found in the literature for frontal bone fractures [40].

Limitations

While the results of this study indicate that helmets may not offer adequate protection against a direct puck impact in competitive leagues, further investigation is required prior to making recommendations to the manufacturing companies. First, only one impact location was selected for this study; it is possible that the helmet could perform differently at other impact sites. Due to the narrow offset of hockey helmets, the side, rear and top of the head are protected by a thinner liner (14 mm as opposed to 21 mm on the forehead). A thinner liner has a lower capacity to absorb the impact and may offer less protection at these sites. Furthermore, it was demonstrated that the brain is sensitive to impact direction [41-43]; thus helmet performance should also be tested for direct impacts to other regions of the head.

Second, glancing impacts, where the puck hits the helmet at an angle and is deflected, were not considered in this study. Non-centric impacts were shown to increase the dynamic response of the head in certain conditions [44]; however this was performed using a high mass impacting arm which simulated head-to-head collisions. Due to its low mass, the puck is deflected when striking a helmet and does not transfer all its energy as a high mass head-to-head impact would;

this principle was used to guide the design of hockey goalie helmet shells. While it is reasonable to hypothesize that glancing impacts will generate a lower dynamic response of the head, its effects on brain tissue deformation are unknown and it would be interesting to study these types of impacts.

Third, the 50^{th} percentile adult male Hybrid III head and neck were designed and validated for indirect head impacts following car crash [45-46]. Although widely used, there is an inherent limitation when using rigid physical models to represent human head response. The headform is made of steel, and is therefore not biofidelic and can only approximate the dynamic properties and impact response of a real human head [47-48]. The hybrid III neckform is also made of stiff materials and was validated against inertial loading rather than direct impact [45]. Neckform compliance was shown to influence the dynamic response of the head for front impacts [49]; yet this was performed using a high mass impacting arm which simulated head-to-head collisions. The momentum transfer of a puck is different due to its low mass and the neck may not have influenced the results. Nonetheless, this study was not meant to replicate human head injury, but rather create a controlled environment in which the comparison of the dynamic response dependent variables can be made.

Fourth, Comparisons between the MPS computed with the UCDBTM cannot be directly compared to strain values obtained using a different model or animal data. The data is a direct result of the model's representation of the head and material composition which differs between models. Finite element modeling requires many assumptions, and comparison with experimental data is crucial for their validation [18]. Correlation with experimental data is difficult due to the scarce experimental data. A model is considered validated if its response is reasonably correlated to the response measured experimentally. It is therefore not proper to assume that the model has been validated for a different situation or parameter [18]. For this reason, the MPS computed in this study were put into context using a wide range representing concussion risk.

Conclusion

In conclusion, a typical vinyl nitrile ice hockey player helmets was shown to maintain peak linear acceleration, peak angular acceleration, and MPS below values associated with a 50 % risk of concussion for a direct puck impact to the forehead at a velocity of 17 m/s. This was not the case for velocities of 23 m/s and above. Furthermore, at 29 m/s and 35 m/s, it was shown that the ability to reduce peak linear acceleration was reduced after a single impact to the forehead, indicating that the helmet may need to be replaced. This research needs to be expanded prior to making recommendations to leagues and manufacturing companies.

Acknowledgments

The authors confirm that there were no financial or personal relationships with other people or organizations that could inappropriately influence this study. The authors would also like to thank NSERC, CIHR, and Reebok-CCM for funding the laboratory.

References

[1] Koh, J. O., Cassidy, J. D. and Watkinson, E. J., “Incidence of Concussion in Contact Sports: A Systematic Review of the Evidence,” Brain Injury, Vol. 17, 2003, pp. 901-917.

[2] Echlin, P. S., Tator, C. H., Cusimano, M. D., Cantu, R. C., Taunton JE, Upshur, R. E. G., Hall, C. R., Johnson, A. M., Forwell, L. A. and Skopelja, E. N., “A Prospective Study of Physician-Observed Concussions During Junior Ice Hockey: Implications for Incidence Rates,” Neurosurg. Focus, Vol. 29, 2010, pp. 1-10.

[3] Emery, C. A. and Meeuwisse, W. H., “Injury Rates, Risk Factors, and Mechanisms of Injury in Minor Hockey,” Am. J. Sports Med., Vol. 34, 2006, pp. 1960-1969.

[4] Gerberich, S. G., Fink, R., Madden, M., Priest, J. D., Aamoth, G. and Murray, K., “An Epidemiological Study of High School Ice Hockey Injuries,” Child. Nerv. Syst., Vol. 3, 1987, pp. 59-64.

[5] Rousseau, P., Post, A. and Hoshizaki, T. B., “A Comparison of Peak Linear and Angular Headform Accelerations Using Ice Hockey Helmets,” J. ASTM Intl., Vol. 6, 2006, Paper ID JAI101877.

[6] Gilenstam, K., Henriksson-Larsén, K., Thorsen, K., “Influence of Stick Stiffness and Puck Weight on Puck Velocity During Slap Shots in Women’s Ice Hockey,” Sports Eng., Vol. 11, 2009, pp. 103-107.

[7] Pearsall, D. J., Montgomery, N., Rothsching, N. and Turcotte, R. A., “The Influence of Stick Stiffness on the Performance of Ice Hockey Slap Shots,” Sports Eng., Vol. 2, 1999, pp. 3-11.

[8] Roy, B. and Doré, R., “Kinematics of the Slap Shot in Ice Hockey as Executed by Players of Different Age Classifications,” *5th International Congress of Biomechanics*, Jyväskylä, Finland, 1975, International Congress of Biomechanics.

[9] Wu, T. C., Pearsall, D., Hodges, A., Turcotte, R., Lefebvre, R., Montgomery, D. and Bateni, H., “The Performance of the Ice Hockey Slap and Wrist Shots: The Effects of Stick Construction and Player Skill,” Sports Eng., Vol. 6, 2003, pp. 31-40.

[10] Padgaonkar, A. J., Krieger, K. W., and King, A. I., “Measurement of angular acceleration of a rigid body using linear accelerometers,” J. Appl. Mech., Vol. 42, 1975, pp. 552-556.

[11] Gong, S.W., Lee, H. P., and Lu, C., "Computational simulation of the human head response to non-contact impact," Comput. Struct., Vol. 86, 2008, pp. 758-770.

[12] Gurdjian, E. S., and Lissner, H. R., "Mechanism of head injury as studied by the cathode ray oscilloscope, preliminary report," J. Neurosurg., Vol. 1, 1944, pp. 393-399.

[13] Gurdjian ES, Lissner HR, Latimer FR, Haddad BF, and Webster JE (1953), Quantitative determination of acceleration and intracranial pressure in experimental head injury – Preliminary report, Neurology, 3, 417-423

[14] Kleiven, S., "Predictors for traumatic brain injuries evaluated through accident reconstructions," Stapp Car Crash J., Vol. 51, 2007, pp. 81-114.

[15] Kopecky, J. A., and Ripperger, E. A., "Closed brain injuries: an engineering analysis," J. Biomech., Vol. 2, 1969, pp. 29-34.

[16] Thomas, L. M., Roberts, V. L., and Gurdjian, E. S., "Impact-induced pressure gradients along three orthogonal axes in the human skull," J. Neurosurg., Vol. 26, 1967, pp. 316-349.

[17] Zhang, L., Yang, K. H., and King, A. I., "A proposed injury threshold for mild traumatic brain injury," J. Biomech. Eng., Vol. 126, 2004, pp. 226-236.

[18] Ward, C. C., and Nagendra, G. K., "Mathematical models: Animal and human models," *The Biomechanics of Trauma*, A. M. Nahum and J. Melvin, Eds., Appleton-Century-Croft, Norwalk, CT, 1985, pp. 77-100.

[19] Goldsmith, W., and Plunkett, J., "A biomechanical analysis of the causes of traumatic brain injury in infants and children," Am. J. Forensic Med. Pathol., Vol. 25, 2004, pp. 89-100.

[20] Galbraith, J. A., Thibault, L. E, and Matteson, D. R., “Mechanical and electrical responses of the squid giant axon to simple elongation,” J. Biomech. Eng., Vol. 115, 1993, pp. 13-22.

[21] Bain, A. C., and Meaney, D. F., “Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury,” J. Biomech. Eng., Vol. 122, 2000, pp. 615-622.

[22] Yu, Z., Elkin, B. S., and Morrison III, B., “Modeling traumatic brain injury in vitro: functional changes in the absence of cell death,” *Biomedical Science & Engineering Conference*, Oak Ridge, TN, 2009, pages 1-4, Institute of Electrical and Electronics Engineers.

[23] Kleiven, S., “Finite Element Modeling of the Human Head,” Doctoral dissertation, Royal Institute of Technology, Stockholm, Sweden, 2002.

[24] Willinger, R. and Baumgartner, D., “Human Head Tolerance Limits to Specific Injury Mechanisms,” Int. J. Crashworthiness, Vol. 8, 2003, pp. 605-617.

[25] Zhou, C., Khalil, T. B. and King, A. I., “A New Model Comparing Impact Responses of the Homogeneous and Inhomogeneous Human Brain,” *39th Stapp Car Crash Conference*, Coronado CA, 1995, SAE International, Warrendale, PA.

[26] Madsen, D. P., Sampson, W. J. and Townsend, G. C., “Craniofacial Reference Plane Variation and Natural Head Position,” Euro. J. Orthod., Vol. 30, 2008, pp. 532–540.

[27] SAE Standard J211/1: Instrumentation for Impact Test – part 1 – Electronic Instrumentation, Surface Vehicle Recommended Practice, Society of Automotive Engineers International, Warrendale, PA, 2007.

[28] Horgan, T. J. and Gilchrist, M. D., "The Creation of Three-Dimensional Finite Element Models for Simulating Head Impact Biomechanics," Int. J. Crashworthiness, Vol. 8, 2003, pp. 353-366.

[29] Horgan, T. J. and Gilchrist, M. D., "Influence of FE Model Variability in Predicting Brain Motion and Intracranial Pressure Changes in Head Impact Simulations," Int. J. Crashworthiness, Vol. 9, 2004, pp. 401-418.

[30] Nahum, A. M., Smith, R. W. and Ward, C. C., "Intracranial Pressure Dynamics During Head Impact," *21st Stapp Car Crash Conference*, New Orleans, LA, 1977, SAE International, Warrendale, PA.

[31] Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H. and King, A. I., "Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-Ray," Stapp Car Crash J., Vol. 45, 2001, pp. 337-368.

[32] Horgan, T., "A Finite Element Model of the Human Head for Use in the Study of Pedestrian Accidents," Doctoral dissertation, University College Dublin, Ireland, 2005.

[33] Doorly, M. C. and Gilchrist, M. D., "The Analysis of Traumatic Brain Injury Due to Head Impacts Arising from Falls Using Accident Reconstruction," Comput. Methods Biomech. Biomed. Eng., Vol. 9, 2006, pp. 371-377.

[34] Miller, R., Margulies, S., Leoni, M., Nonaka, M., Chen, X., Smith, D. H. and Meaney, D. H., "Finite Element Modeling Approaches for Predicting Injury in an Experimental Model of Severe Diffuse Axonal Injury," *42nd Stapp Car Crash Conference*, Charlottesville, VA, pages 155-166, 1998, SAE International, Warrendale, PA.

[35] Doorly, M. C., "Investigations into Head Injury Criteria Using Numerical Reconstruction of Real Life Accident Cases," Doctoral dissertation, University College Dublin, 2007.

[36] Gurdjian, E. S., Roberts, V. L., and Thomas, L. M., "Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury," J. Trauma, Vol. 6, 1966, pp. 600-604.

[37] Ono, K., Kikuchi, A., and Nakamura, M., "Human head tolerance to sagittal impact reliable estimation deduced from experimental head injury using subhuman primates and human cadaver skulls," *24th Stapp Car Crash Conference*, Troy, MI, pages 103-160, 1980, SAE International, Warrendale, PA.

[38] Cater, H. L., Sundstrom, L. E., and Morrison III, B., "Temporal development of hippocampal cell death is dependent on tissue strain but not strain rate," J. Biomech., Vol. 39, 2006, pp. 2810-2818.

[39] Elkin, B. S., and Morrison III, B., "Region-specific tolerance criteria for the living brain," Stapp Car Crash J., Vol. 51, 2007, pp. 127-138.

[40] Yoganandan, N., and Pintar, F.A., "Biomechanics of temporo-parietal skull fracture," Clin. Biomech., Vol. 19, 2004, pp. 225-239.

[41] Kleiven, S. "Influence of impact direction on the human head in prediction of subdural hematoma," J. Neurotrauma, Vol. 20, 2003, pp. 365-379.

[42] Margulies, S. S., Thibault, L. E., and Gennarelli, T. A., "Physical model simulations of brain injury in the primate," J. Biomech., Vol. 23, 1990, pp. 823-836.

[43] Pudenz, R. H., and Shelden, C. H., "The lucite calvarium – A method for direct observation of the brain. II. Cranial trauma and brain movement," J. Neurosurg., Vol. 3, 1946, pp. 487-505.

[44] Walsh, E. S., Rousseau, P., and Hoshizaki, T. B., "The influence of impact location and angle on the dynamic impact response of a hybrid III headform," Sports Eng., Vol. 13, 2011, pp. 135-143.

[45] Deng, Y. C., "Anthropomorphic dummy neck modeling and injury considerations," Accident Anal. Prev., Vol. 21, 1989, pp. 85–100.

[46] Mertz, H. J., "Biofidelity of the Hybrid III head," *29th Stapp Car Crash Conference*, Washington, DC, pages 111-119, 1985, SAE International, Warrendale, PA.

[47] Hubbard, R. P., and McLeod, D. G., "Definition and development of a crash dummy head," *Hybrid III: The First Human-like Crash Test Dummy*, S. H. Backaitis, and H. J. Mertz, Eds., Society of Automotive Engineers, Warrendale, PA, 1974, pp. 95-110.

[48] Seemann, M. R., Muzzy, W. H., and Lustick L. S., "Comparison of human and Hybrid III head and neck response," *30th Stapp Car Crash Conference*, San Diego, CA, pages 291-311, 1986, SAE International, Warrendale, PA.

[49] Rousseau, P., Hoshizaki, T. B., "The influence of deflection and neck compliance on the impact dynamics of a Hybrid III headform," J. Sports Eng. Technol., Vol. 223, 2009, pp. 89-98.

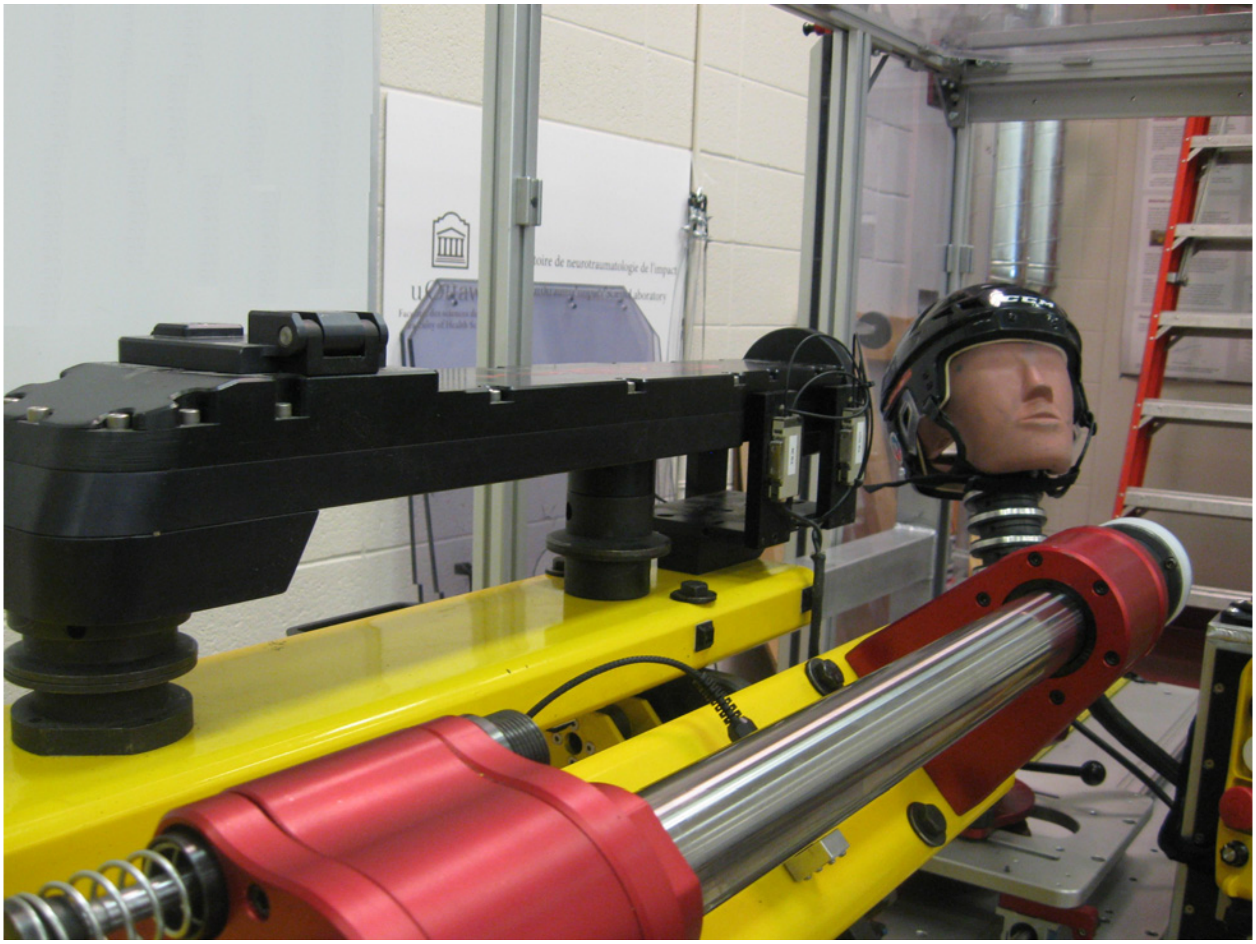

Figure 1 – Oblique view of the pneumatic puck launcher apparatus.

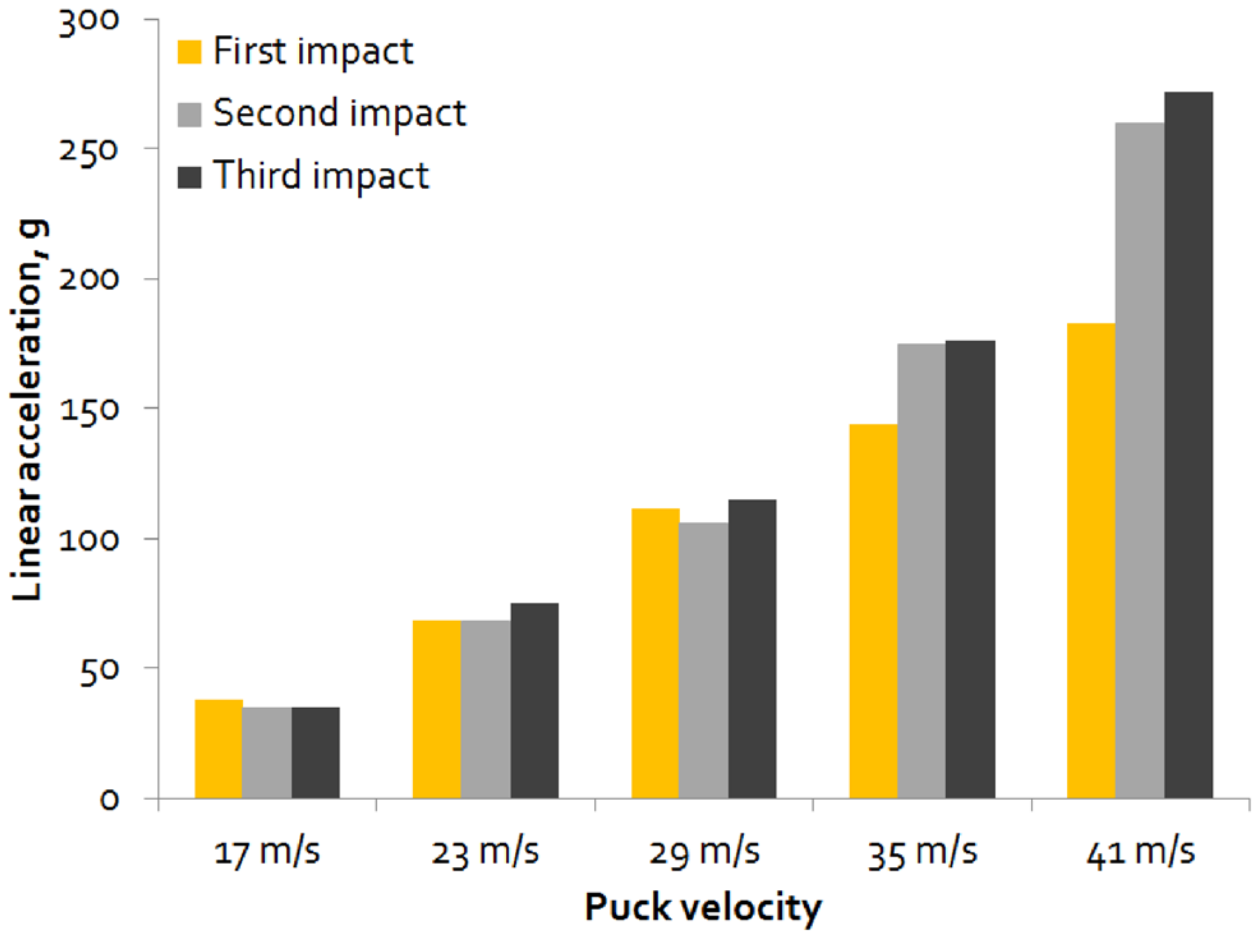

Figure 2 – Peak linear acceleration for three consecutive puck impacts to a vinyl nitrile ice hockey player helmet at five velocities.

Tables

Table 1 - Material properties for the human head finite element.

Material	Density, kg/m^3	Young's modulus, MPa	Poisson's ratio
Facial bone	2100	5 000	0.23
Scalp	1 000	16.7	0.42
Cortical bone	2 000	15 000	0.22
Trabecular bone	1 300	1 000	0.24
Dura	1 130	31.5	0.45
Pia	1 130	11.5	0.45
Falx and tentorium	1 130	31.5	0.45

Table 2 - Material properties used for the neural tissue.

Material	Density, kg/m^3	Shear modulus, kPa		Decay constant, MPa	Bulk modulus, GPa
		G_0	G_∞		
Brain white matter	1 060	12.5	2.5	80	2.19
Brain grey matter	1 060	10	2	80	2.19
Brain stem	1 060	22.5	4.5	80	2.19
Cerebellum	1 060	10	2	80	2.19

Table 3 – Mean and standard deviation of head accelerations and brain tissue strain generated by puck impacts to a bare and helmeted Hybrid III 50th percentile headform at five velocities.

		Head accelerations		Maximum principal strain
		Linear, *g*	Rotational, krad/sec^2	
17 m/s	Bare headform	138 ± 13	13.3 ± 0.3	0.29 ± 0.08
	Helmeted	36 ± 2	3.3 ± 0.6	0.08 ± 0.002
23 m/s	Bare headform	205 ± 8	19.5 ± 1.3	0.36 ± 0.01
	Helmeted	71 ± 4	7.9 ± 0.4	0.17 ± 0.01
29 m/s	Bare headform	294 ± 17	22.9 ± 0.6	0.52 ± 0.05
	Helmeted	111 ± 5	10.8 ± 1.0	0.19 ± 0.01
35 m/s	Bare headform	420 ± 16	31.1 ± 2.0	0.57 ± 0.07
	Helmeted	165 ± 18	15.4 ± 1.3	0.32 ± 0.01