

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

Marshall Kendall¹, Andrew Post¹, Philippe Rousseau¹, Anna Oeur¹, Michael D. Gilchrist^{2,1} and Blaine Hoshizaki¹

Abstract Ice hockey has been identified as having one of the highest concussion rates. The three most likely causes of concussive injury are; falls to the ice, shoulder to head impacts and punches to the head. The purpose of this study was to examine how these three mechanisms of injury in the sport of ice hockey influence the dynamic response of the head form and the magnitude and distribution of maximum principal strain in the cerebrum. The three impact mechanisms were simulated using a Hybrid III head and neck form attached to a linear impactor, pendulum or monorail system. Three dimensional linear and rotational acceleration data from each impact condition were used to undertake finite element modeling to calculate maximum principal strain in regions of brain tissue. The results indicated that each mechanism incurred a unique peak resultant linear and rotational acceleration response. The maximum principal strain magnitudes were found to be largest in the fall to the ice. The regions of the brain incurring the largest deformation varied per mechanism of injury. This variation of peak magnitude per brain region might explain the differences in symptomology for concussion. Furthering the understanding of these mechanisms would aid in improving the safety of the game.

Keywords Brain deformation, Concussion, Head impact, Ice hockey,

I. INTRODUCTION

The estimated annual direct and indirect cost of traumatic brain injury in Canada is \$3 billion with over 11,000 associated fatalities annually; brain injury is the leading cause of death and disability for Canadian children [1]. According to the International Ice Hockey Federation, Canada has over 577,000 registered hockey players competing at a variety of levels [2]. The British Columbia Injury Research and Prevention Unit [3] reports that in 1999, 3.78% of all sport-related emergency room visits in Canada for 1999 were head injuries in ice hockey. In fact, ice hockey has been identified as the contact sport having the highest rate of concussion per participants, with 1.8 concussions per 1000 player-hours in the NHL alone [4 - 5]. While the game of hockey is a fast moving, high impact sport, the severity and duration of the injury resulting from different hits to the head is unpredictable [5]. Furthermore, the amount of time a player remains out of the game with reoccurring symptoms varies significantly [5]. The variance in symptomology and time out from the game is likely a result of the mechanism in which the concussive injuries occur. Understanding the different mechanisms creating these types of injuries is still not well defined and is the topic of this research. While there are many ways players get hit to the head in ice hockey, the most common impacts causing concussive injuries involve falls with the head hitting the ice or body segments impacting the head [6]. The differences in these impact conditions likely create unique loading conditions in the various regions of the brain tissue, which may be representative of concussive symptomologies [7]. The purpose of this study was to examine how three distinct mechanisms of injury in the sport of ice hockey influence the dynamic response of the head form and the magnitude and distribution of maximum principal strain in the cerebrum.

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II. METHODS

An adult male 50th percentile Hybrid III head and neckform was used for all impact conditions. The head form was instrumented with nine mounted single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) in a 3-2-2-2 accelerometer array [8]. In order to simulate typical ice hockey situations, the head form was equipped with an EPP ice hockey helmet during the impacts. Three experimental conditions were developed to simulate three specific impact types associated with head injuries in ice hockey: shoulder/elbow to an opponents head; head hitting the ice; and punch to an opponent's jaw region. Three impacts were collected for each condition. Inbound velocity for each impact condition was set at 6.5 m/s and was measured using a velocimeter (time gate) or high speed camera for consistency. This velocity was chosen based on previous studies showing skating speed, boxing punch speed and falls to the ice with ranges between 5 and 10 m/s [9 - 12]. All the accelerometer signal data were sampled at 20 kHz and filtered with a 1000 Hz low pass Butterworth filter according to the SAE J211 convention. The signals were passed through a TDAS Pro Lab system (DTS, Calabasas, CA) prior to being processed by TDAS software. Statistical measures used ANOVAs to determine statistical differences in the dynamic response of the head forms and brain deformation metrics for the 3 different impact conditions.

Impact conditions

Shoulder to Head

A linear impactor was used to simulate the shoulder impact condition (Fig 1). The linear impactor was built with a stationary steel frame secured to a cement floor supporting a $1.28 \pm 0.01\text{m}$ long cylindrical, free-moving impactor arm ($13.1 \pm 0.1\text{kg}$). A Hybrid III 50th percentile adult male head form and neck form was attached to a $12.78 \pm 0.01\text{kg}$ sliding table to allow for post impact movement. The inbound velocity of the impacts was 6.5 m/s, which are within the possible impact velocities during the game of ice hockey [9 -10]. The impacts were to the side (centre of gravity) of the head form as this is a common impact site for shoulder to head impacts. The tip of the impactor was fitted with a hemispherical nylon pad covering a modular elastomer programmer (MEP) 60 Shore Type A (0.05m thick) disc covered with a Pro shoulder pad with a plastic cap. The MEP pad was chosen to simulate the rigidity of the shoulder of the impacting player.



Fig. 1. Linear impactor system used to simulate the shoulder impact condition.

Head to ice

A monorail drop system with MEP anvil was used to simulate the head impacting the ice condition (Fig 2). The helmeted Hybrid III head and neck form (mass $6.08\text{kg} \pm 0.01\text{kg}$) were attached to the monorail drop system and impacted with an inbound velocity of 6.5 m/s. The location of the impact for this reconstruction was to the rear of the head form to simulate a player falling backwards onto the ice.

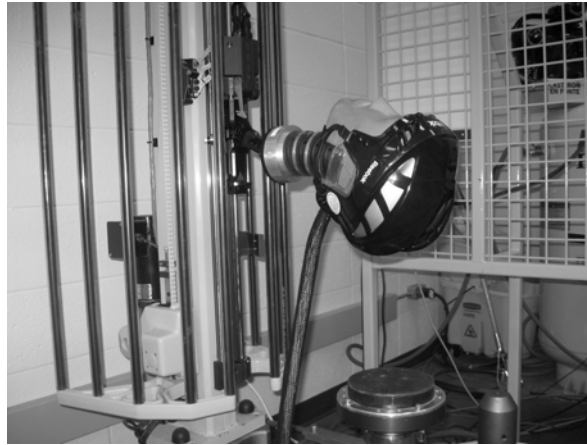


Fig. 2. Monorail drop system used to simulate head impacting the ice (MEP anvil)

Punch to the head

The third impact condition was designed to simulate a punch to the head (Fig 3). This condition was reconstructed using a pendulum system with a 2 kg striker, to represent the lower mass condition of a punch. The inbound velocity was set at 6.5 m/s and impact location was to the left jaw region of the head form. The velocity of impact was measured using a High Speed Imaging PCI-512 Fasctam running at 2 kHz. The Hybrid III head and neck form was rigidly fixed to the linear impactor table in order to prevent it from sliding during the impacts.



Fig. 3. Pendulum impact system used to simulate the punch impact condition.

Finite Element Analysis

The resulting linear and rotational accelerations were used as input for the University College Dublin Brain Trauma Model (UCDBTM) [13 -14]. Maximum principal strain (MPS) values were obtained from the brain model and averaged over ten regions within the brain, which represent functional areas associated with the symptomology of concussion (Fig 4) [15]. This study uses these areas to track the brain deformation fields in the brain tissue and is not intended to be representative of concussive symptoms, as this study does not involve reconstructions of actual injury cases. These nine regions have particular roles in motor and physiological human response which, if damaged, may present particular symptoms affecting their functionality (Table 1). Reconstructive research has shown that maximum principal strain may be a good predictor for concussion and thus, was used as the brain deformation metric in this study [16 - 18]. Segmenting the brain in this manner allows for identification of changes in regions of the brain according to the type of injury reconstructed.

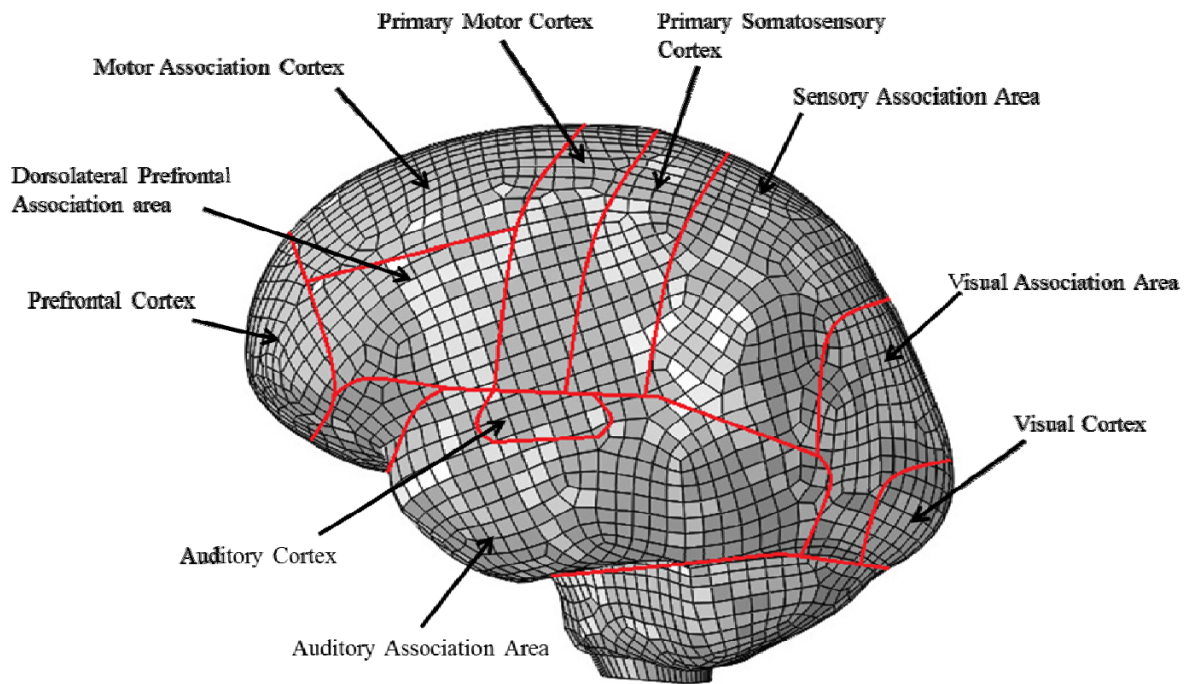


Fig. 4. Image of the UCDBTM segmented into regions for analysis

The University College Dublin Brain Trauma Model was developed by Horgan and Gilchrist [13 - 14]. The geometry of the UCDBTM was resolved from computed tomography (CT) and magnetic resonance imaging (MRI) scans of a male cadaver. The version of the model used for this research was composed of the following parts: scalp, three-layered skull (cortical and trabecular bone), dura, cerebrospinal fluid (CSF), pia, falx, tentorium, cerebral hemispheres, cerebellum and brainstem. Overall, this version of the UCDBTM has approximately 26 000 elements. The model was validated according to cadaveric pressure and brain motion data [19 -20] as well as reconstructions of traumatic brain injuries [21].

The material properties (Tables 2 and 3) of the cerebrum, cerebellum and brain stem were derived from work done by Zhang et al [22]. The remaining parts of the model (scalp, cortical and trabecular bone, and intracranial membranes) were derived from research conducted by Ruan [23], Willinger et al [24], Zhou et al [25] and Kleiven and von Holst [26]. Work done by Zhou et al [25] and Shuck and Advani [27] were used in the characterization of the shear modulus of viscoelastic brain tissue:

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

where G_{∞} is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor [13]. The brain skull interface was modeled by modeling the CSF as brick elements with a low shear modulus which allows it to behave like water. The contact definition at the brain skull interface was assigned as no separation and used a friction coefficient of 0.2 [34].

Table 1. Nine brain regions used in this study and their functionality

Region of the Brain	Functionality	Reference
Prefrontal cortex	decision-making and social behaviour	Yang et al. [28] Yang et al. [28]
Dorsolateral prefrontal area	memory and organization	
Motor association cortex and Primary motor cortex	planning and execution of movement patterns	Luppino & Rizzolatti [29]
Primary somatosensory cortex	sense of touch and proprioception	Price [30]

Visual cortex	Visual information	Braddick et al. [31]
Sensory and Visual association areas	Sensory integrations and perception of the environment	Price [30]
Auditory cortex	sound processing	Purves et al. [15]
Auditory association area	functions associated with recognition of sounds	Creutzfeldt et al. [32 – 33]

Table 2. Properties of the materials used in the finite element model

Material	Poisson's Ratio	Density (kg/m ³)	Young's Modulus (Mpa)
Grey Matter	0.49	1060	30
White Matter	0.49	1060	37.5
Trabecular Bone	0.24	1300	1000
Dura	0.45	1130	31.5
Pia	0.45	1130	11.5
Falx	0.45	1140	31.5
Tentorium	0.45	1140	31.5
CSF	0.5	1000	-

Table 3. Characteristics of the brain tissue

Material	Shear Modulus (kPa)			Bulk Modulus (s ⁻¹)
	G ₀	G _∞	Decay Constant (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

III. RESULTS

The dynamic response and average magnitude of brain deformation results are shown in Table 4. The regions of the brain where the largest maximum principal strain magnitudes were incurred are shown in Table 5. It is important to note that data for the cerebrum areas were restricted to regions not in contact with the tentorium as errors are associated with the contact between the tentorium and the surrounding brain tissue [35]. The results showed that the head impact to the ice produced the highest peak linear accelerations and highest average MPS (264 g and 0.468 respectively). The shoulder impacts produced the next highest results (112 g and 0.370 MPS). The punch condition yielded the lowest average peak linear acceleration and MPS (87.9 g and 0.266 MPS). The highest average peak rotational accelerations occurred with the punch condition (14001 rad/s²).

Table 4. Average peak resultant acceleration and average maximal principal strain (standard deviations in parentheses)

Mechanism	Peak resultant acceleration		Brain deformation
	Linear (g)	Rotational (rad/s ²)	Average MPS
Ice	264.4 (33.8)	11204 (1867)	0.424 (0.019)
Shoulder	112.5 (8.6)	9659 (728.5)	0.305 (0.011)
Punch	87.9 (9.8)	14001 (1003)	0.229 (0.003)

Table 5. Average Maximal Principal Strain values for 9 regions of the brain (*brain locations denoted below, standard deviations in parentheses)

	1	2	3	4	5	6	7	8	9
Ice	0.468 (0.043)	0.456 (0.009)	0.454 (0.009)	0.458 (0.010)	0.426 (0.024)	0.356 (0.036)	0.279 (0.031)	0.426 (0.017)	0.250 (0.015)
Shoulder	0.190 (0.006)	0.370 (0.025)	0.267 (0.008)	0.370 (0.025)	0.347 (0.016)	0.308 (0.011)	0.307 (0.006)	0.209 (0.010)	0.109 (0.006)
Punch	0.161 (0.001)	0.248 (0.005)	0.168 (0.005)	0.212 (0.007)	0.266 (0.004)	0.255 (0.012)	0.231 (0.002)	0.220 (0.003)	0.086 (0.003)

*1- Prefrontal Cortex, 2- Dorsolateral prefrontal cortex, 3- Motor Association Cortex, 4- Primary motor Cortex, 5- Primary somatosensory cortex, 6- Sensory association area, 7- Auditory cortex, 8- Visual association area, 9- Visual cortex

Significant differences were found between peak linear accelerations for the ice/punch and ice/shoulder conditions (p<0.01). For peak rotational acceleration, only significant differences between the shoulder and punch condition (p<0.05) were found. Significant differences in average maximal principal strain values between all three impact conditions were found (p<0.01). The region of the brain producing the highest MPS values for the ice impact condition was the prefrontal cortex. The shoulder had the highest values in the dorsal-lateral prefrontal cortex and primary motor cortex, while the punch condition showed highest MPS values in the primary somatosensory cortex. Significant differences in MPS values were found between the three conditions for each of the regions depicting the highest MPS values (p<0.05) except in the prefrontal cortex region between the shoulder and punch conditions.

IV. DISCUSSION

This study was designed in part to examine the three dimensional dynamic impact responses of three different types of head impacts commonly associated with ice hockey head injuries. The second part of this study examined how the different injury mechanisms influenced the regions of the brain tissue as shown by the UCDBTM. To reconstruct the three different mechanisms of injury, impact velocity was constant but impact mass and location were not matched. All three impact conditions produced linear and rotational accelerations above the 50% risk (82g ; 5900 rad/s²) for concussions [17, 18]. Effective mass and impact location were two variables that varied throughout the study. The reason for this is that the mass involved in a punch is much lower than the mass encompassing an impact with the shoulder. As a result, the peak resultant linear acceleration values for the punch condition were lower than the shoulder and fall to the ice conditions. The high peak resultant linear acceleration values found for the fall to the ice are likely a result of the helmet being

impacted at conditions beyond its functional range. Interestingly, the rotational acceleration values for the punch condition were much higher than for the other two injury mechanisms. The punch impact location, which was to the jaw region (non-centric), likely contributed to the high rotational acceleration and low linear acceleration. With respect to MPS, again all three-impact conditions produced values which were higher than those associated with 50% risk (MPS 0.19 – 0.26) for concussion [17, 18]. The average maximum principal strain showed that the fall to the ice incurred the largest brain deformation, followed by the shoulder to head impact and then the punch. The fall to the ice had very high linear and rotational acceleration magnitudes, which helped create large deformation in the brain tissue.

The region of largest maximum principal strain was shown to vary depending on the type of reconstruction. The fall to the ice had the largest deformation at the prefrontal cortex, the shoulder to the head at the dorsolateral prefrontal cortex and primary motor cortex and the punch to the head at the primary somatosensory cortex. These variations in location of MPS is a result of the unique acceleration loading curve time histories [36] created from the combination of impact location, compliance and mass. Current research using head impact reconstructions have not yet linked particular symptoms of concussions with damage to a specific area of the brain. It has been well reported that symptoms present immediately after a concussive event are dependant of the biomechanical parameters of the collision, as well as the specifically affected brain structures [37 - 41]. Research mapping the brain has shown that each region of the brain has a particular function (Table 1). The UCDBTM is able to differentiate between grey and white matter and the cerebrum, cerebellum and brain stem. The brain was further separated into regions associated with the symptomology of concussion for the analysis of the American football helmet models [42]. The purpose of this study was not to associate specific symptoms for each impact type to the brain region affected. The results, however, do show the possibility of varying impact conditions which can produce a concussion in the sport. Thus, causing damaging brain deformations in different regions of the brain could be represented by differences in symptomology. Future research should reconstruct actual concussive events within the sport of ice hockey to further identify which conditions produce the largest brain deformations associated with concussion and how they relate to symptomology.

The research presented should be considered according to limitations inherent to its methodologies. The Hybrid III head form is commonly used as a physical model in impact reconstruction due to its reliability. The head form being constructed of steel, evidently, does not provide a biofidelic response to impacts. Also, the Hybrid III is primarily used for impacts in the antero-posterior direction and thus impacts to other regions of the head form may create unknown error. The neck form stiffness has also been deemed a limitation as it is shown to influence brain deformation metrics [43]. For this study, the neck form used was consistent during all three impact conditions. While other surrogate head and neck forms are available for impact reconstructions (such as, world-SID), none has been validated for rotational components of impacts. The UCDBTM is one of few partially validated models available for this type of research. However, the finite element model makes assumptions surrounding the characteristics of the brain tissue and the interactions between different parts of the brain and skull. As a result, the comparisons made with the UCDBTM are meant to be representative of how the brain may deform under the loading scenarios and may not represent the exact motion of the brain resulting from impacts. Furthermore, brain deformation metric analysis was limited to the cerebrum since brain stem has not yet been validated for finite element modelling.

V. CONCLUSIONS

The purpose of this study was to simulate three different types of head impacts observed in ice hockey, that often result in injury. The three mechanisms of injury represented in this research have been shown to produce differing magnitudes of linear and rotational acceleration as well as brain deformation. The punch in particular produces high rotational acceleration with a low linear acceleration response. Also, it was shown that these mechanisms produce large deformations in different regions of the brain tissue all of which are above the 50% risk of concussive injury. This variance might help explain why there are many different symptomologies to concussion. The conditions surrounding the mechanisms of injury may create loading to the brain tissue which would incur damage to one region and not another. This research demonstrates that it is important to understand the differences between the three types of brain injury mechanisms in ice hockey. Further research looking specifically at the effects of different parameters such as; velocity, mass and impact location on brain

deformation metrics could provide additional understanding as to how different regions of the brain are injured. This understanding could aid in improving the safety of the game and equipment development to better protect players against these types of impact conditions.

VI. REFERENCES

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APPENDIX

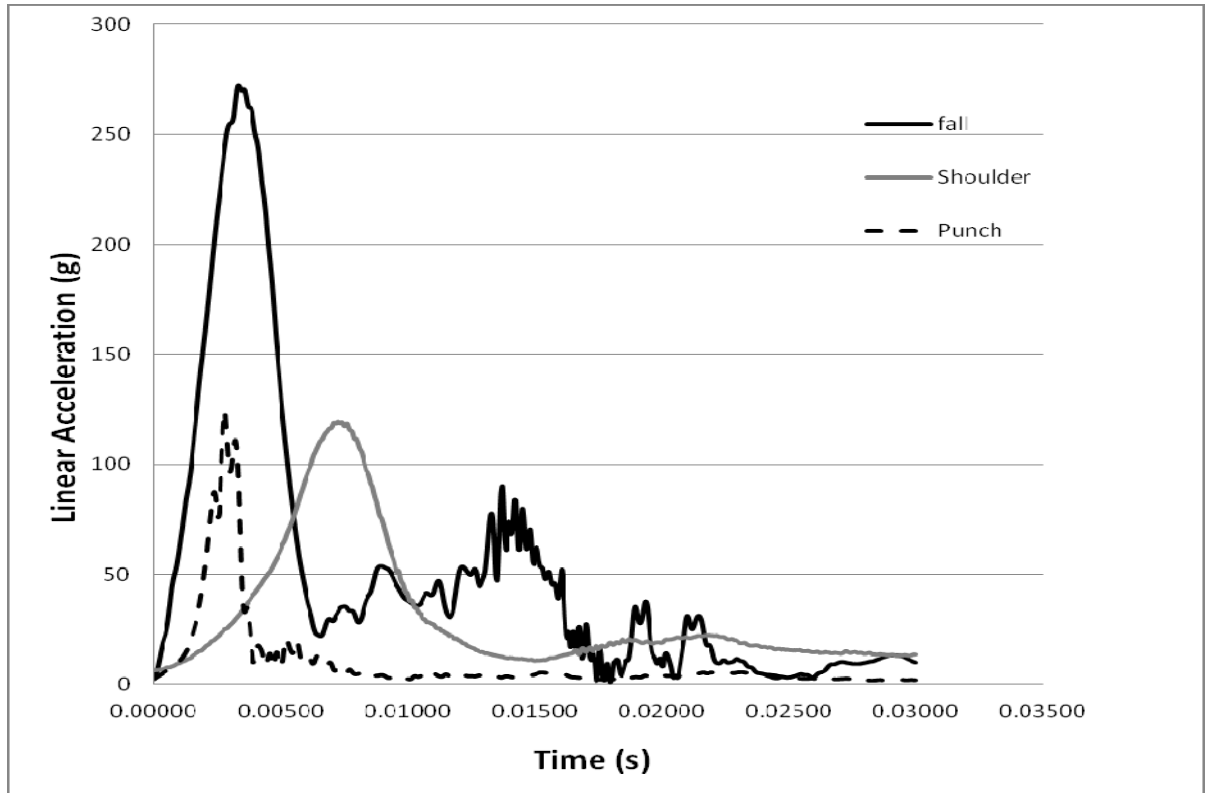


Fig. 4. Resultant linear acceleration time history curves for the three impact conditions.

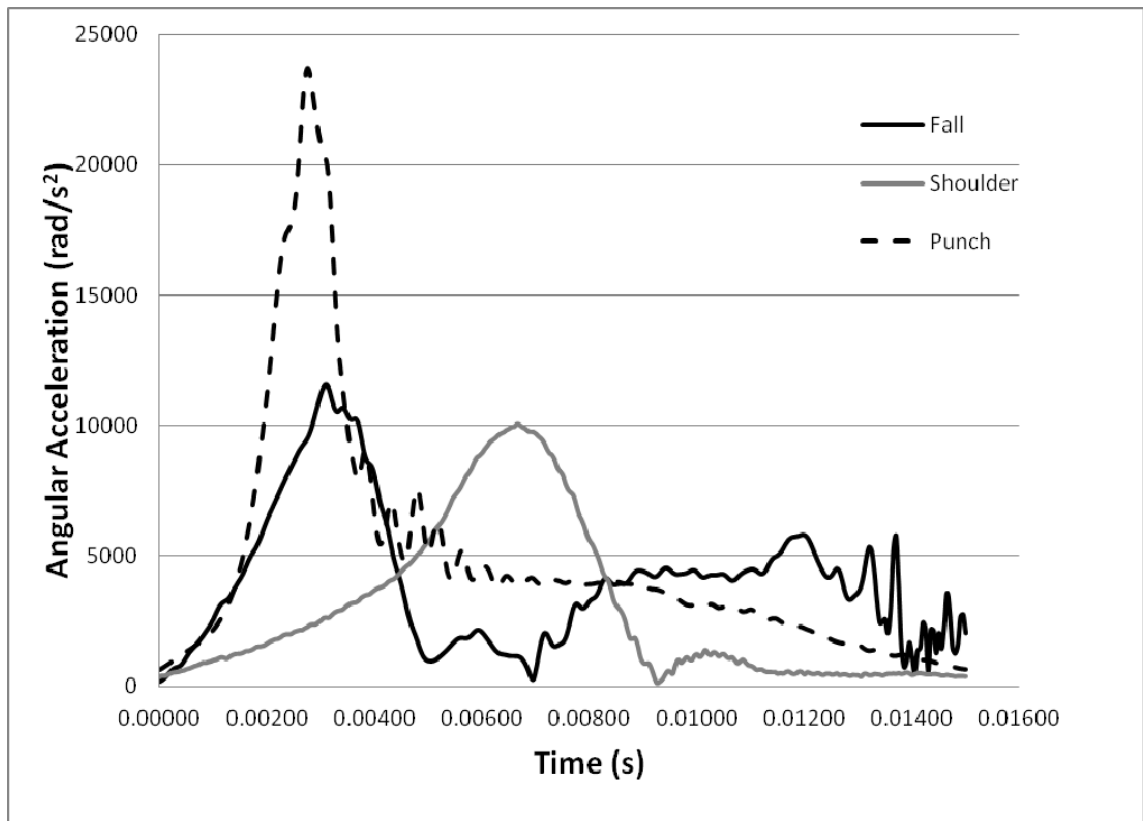


Fig. 5. Resultant angular acceleration time history curves for the three impact conditions.