

Angular Acceleration Responses of American Football, Lacrosse and Ice Hockey Helmets Subject to Low-Energy Impacts

R. Anna Oeur, Katrina Zanetti, T. Blaine Hoshizaki

Abstract Neurological complications associated with sports-related brain trauma have been a recent concern and focus of research. In turn, sensor technologies have been developed to count head impacts during sports. Low-level impacts present a low risk for concussion but are implicated in long-term neurological conditions. The effectiveness of helmets at managing angular acceleration at this energy level is unknown. The objective of this research was to examine the ability of American football, lacrosse and hockey helmets and the bare head to manage angular accelerations for low-level impacts (20g). A 50th percentile Hybrid III headform was impacted using a pendulum system at nine centric and non-centric impact sites. Linear and angular accelerations were recorded using nine single-axis accelerometers arranged in a 3-2-2 array. A consistent trend across centric and non-centric impact sites was observed for peak angular acceleration: football helmets produced the lowest levels, followed by the bare headform, with the lacrosse and hockey helmets producing the highest values. At the crown site, however, the football helmet produced the highest angular acceleration. Peak angular acceleration at low impact energies was influenced by the type of helmet and impact condition. While there is little risk in terms of linear acceleration for low-level impacts, angular accelerations approach a 25% risk of concussion. Helmets must be tested and designed to manage these impacts.

Keywords Head impact count, Hybrid III headform, Pendulum impact system, Sub-concussive impact

I. INTRODUCTION

Concussion is defined as short-lived functional cognitive impairment resulting in symptoms such as headaches, dizziness and nausea [1]. While most concussions resolve, some symptoms do persist [2], often leading to disability and depression decreasing the quality of life for many [3-4]. Long-term neurological complications associated with brain trauma have been linked to neurodegenerative disease called chronic traumatic encephalopathy (CTE) which may be defined by symptoms of dementia, memory loss and emotional instability [5-6]. The exact cause of CTE is currently unknown; however, this disease is commonly found in athletes with a history of repetitive head trauma [7]. This has led researchers to propose that athletes at risk for the disease are likely those who have suffered a history of concussion or those who have sustained repetitive low-level (sub-concussive) impacts [6][8].

Experimental evidence derived from physical and animal studies has led researchers to hypothesize that the spectrum of injuries ranging from concussion to diffuse axonal injury is associated with injurious brain tissue strains from head rotations [9-12]. This has led a number of researchers to characterize the magnitudes of angular acceleration from sub-concussive and

R. A. Oeur is a PhD student (613-562-5800 ext. 7207, anna.oeur@uottawa.ca), K. Zanetti is a MSc student, and T.B. Hoshizaki is a Prof. of Biomechanics in the School of Human Kinetics at University of Ottawa, Canada.

concussive head impacts in sport. Pellman et al. [13] conducted reconstructions of head impacts occurring in professional American football using helmeted anthropometric test dummies. These authors reported average angular accelerations for concussion and no injury to occur at 6432 and 4235 rad/s^2 , respectively [13]. Based on Mathematical Dynamic Models (MADYMO) reconstructions of Australian rules football and rugby, Patton et al. [14] suggested an angular acceleration of 4500 rad/s^2 to predict concussive injuries resulting in prolonged loss of consciousness. Other research used helmets of American football players instrumented with accelerometer arrays that measured head acceleration during practice and games [15]. These researchers reported head impacts resulting in concussion and no injury to occur at average head angular accelerations of 5022 and 1230 rad/s^2 , respectively [15]. In summary, proposed angular acceleration values vary across a range of around 1230 - 4235 rad/s^2 for no injury and at 4500 – 6434 rad/s^2 for concussion. This variation of angular acceleration resulting from impacts resulting in no injury and concussion demonstrates the range of proposed values for low severity injury. A definitive threshold separating impacts causing injury from impacts resulting in no injury is currently not known. While concussive impacts are associated with clinical signs and symptoms, sub-concussive impacts are defined as occurring below this range and do not result in clinically-diagnosed concussion. However, studies have proposed that an accumulation of these low-level impacts do compromise the brain over time by disrupting cellular and metabolic processes [6][8].

In an attempt to make contact sports safer, sensor technologies have been developed to count the number of impacts to the head in helmeted and non-helmeted sports. These sensors are intended for prophylactic use, serving to monitor and inform parents and coaches of impact dosage and are designed and certified to count impacts to the head that exceed a linear acceleration of 20g [16-17]. Since the purpose of these sensors was not to capture concussive level impacts (approximately 60g [18]), they had to be able to register abnormal head accelerations that are higher than those occurring from common movements in sports such as running and jumping (10-15g [19]). Therefore, a 20g linear acceleration criterion was proposed by Sports Legacy Institute [16-17]. Although this level of peak linear acceleration presents a low risk for concussive injury, angular acceleration also creates brain trauma. Angular acceleration has been implicated with concussion and diffuse axonal injury (DAI) [12][20-21]. With sensors counting impacts based on linear acceleration, a component of brain trauma influenced by angular acceleration is neglected. Therefore, the relative risk of brain injury as measured by angular acceleration for these 20g impacts is unknown. Additionally, it is not known if helmets under these conditions provide any level of protection against angular acceleration. Therefore, the objective of this research was to examine angular accelerations of American football, lacrosse and hockey helmets in comparison to a bare head condition for low-level impacts occurring at 20g.

II. METHODS

Experimental Testing

Impacts to a Hybrid III headform were performed according to four different head conditions and nine prescribed locations on the headform using a pendulum system (Fig. 1). The pendulum system consisted of a metal frame, a low-friction sliding table to which the Hybrid III head and neckform is attached and a vinyl nitrile (VN) impactor cap attached to the end of the pendulum frame that was used to impact the Hybrid III headform.

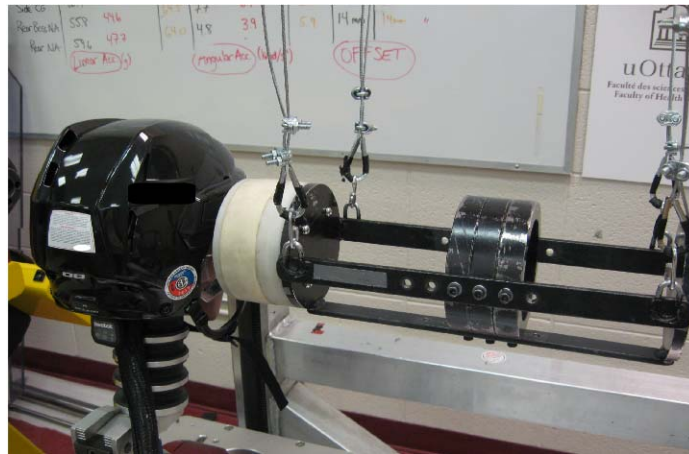


Figure 1. Set-up of impact to hockey-helmeted headform by the pendulum system.

The 50th percentile Hybrid III headform was outfitted with Endevco 7264C-2KTZ-2-300 uniaxial accelerometers in a 3-2-2-2 array for measuring three-dimensional kinematics [22]. Signals were collected at 20 kHz using Diversified Technical Systems TDAS Pro lab module software and filtered with an 800Hz filter using Bioproc 3 software [23].

Impact certified American football, hockey and lacrosse helmets (Table 1) were each fitted on the Hybrid III headform and impacted at nine sites. The American football and lacrosse helmets were certified according to the National Operating Committee on Standards for Athletic Equipment (NOCSAE) for football [24] and lacrosse [25]. The hockey helmet was certified according to the Canadian Standards Association for ice hockey helmets [26]. These helmet models were chosen to represent conventional helmets that are used in each sport and were not selected based on particularly good or poor ratings during testing corresponding to the respective standards. The rationale for these different types of helmets was to evaluate head impact counts that are reflective of the typical trauma sustained by players of each sport.

The direction of head impact has been demonstrated to influence injury head motions occurring in the coronal plane associated with worse outcomes [12][27]. This has been attributed to the interaction between the geometry and densities of the inner structures of the skull and brain and how they can injure the tissue [12][28]. In sport, research has documented that the most frequent impact location associated with concussive injuries in American football is the front of the head [13][29], and the side of the head in Australian rules football and rugby [30]. Since the impact location has been demonstrated to be an important factor influencing the risk of injury, this led to research conducted by Walsh et al. [31] to evaluate the risk of impacts to the head occurring in different locations and angles about a Hybrid III headform. These impact conditions were then developed into an impact protocol to evaluate head protection [32]. This served as the motivation for using the nine impact sites in the present study, as they cover a number of impact locations and angles around the head [31][33]. A fourth condition, the bare headform without any helmet, was also impacted. Six of the nine impact locations were centric, directed through the centre of gravity of the headform (Table 2), and three were non-centric, not directed through the headform's centre of gravity (Table 3) [31]. For each head condition, the headform was impacted three times per impact site; a total of 27 impacts were performed for each of the four head conditions. The angular accelerations were recorded for trials with peak resultant headform linear accelerations between 20 ± 2 g.

TABLE 1. Description of helmets used

Helmet Type	Picture	Offset at various locations (cm)	Description of Materials
American Football		Crown: 3.2 Front: 3.2 Side: 3.2 Rear: 2.92 Average Offset: 3.1	Liner: Vinyl nitrile
Ice hockey		Crown: 1.8 Front: 1.3 Side: 1.3 Rear: 1.3 Average Offset: 1.4	Liner: Vinyl nitrile
Lacrosse		Crown: 1.6 Front: 1.5 Side: 1.6 Rear: 1.4 Average Offset: 1.5	Liner: Expanded Polypropylene

TABLE 2. Centric impact locations on Hybrid III headform (impact vector indicated by arrow)

Name and description of impact site	Diagram of impact site
Crown At the apex of the headform where the frontal and median planes intersect.	
Front Positive Elevation 15° On the anterior intersection of the median and horizontal plane at a point 30mm above the reference plane, where the impact vector is elevated 15° from the horizontal.	

Front Boss Centre Gravity

Midpoint between the anterior median and right frontal planes in the horizontal plane at a point 30mm above the reference plane, where the impact vector is applied perpendicular (90°) to the headform surface.

**Side Centre Gravity**

At the right intersection of the frontal and horizontal planes at a point that is 30mm above the reference plane, where the impact vector is applied perpendicular (90°) to the headform surface.

**Rear Boss Centre Gravity**

Midpoint between the posterior median and right coronal planes in the horizontal plane at a point that is 30mm above the reference plane, where the impact vector is applied perpendicular (90°) to the headform surface.

**Rear Centre Gravity**

At the posterior intersection of the median and horizontal planes at a point that is 30 mm above the reference plane, where the impact vector is applied perpendicular (90°) to the headform surface.



TABLE 3. Non-centric impact locations on Hybrid III headform (impact vector indicated by arrow)

Name and description of impact site	Diagram of impact site
Front Boss Positive Azimuth Midpoint between the anterior median and right frontal planes in the horizontal plane at a point 30mm above the reference plane with a 45° positive azimuth rotation of head and neckform in the horizontal plane.	

Rear Boss Negative Azimuth

Midpoint between the posterior median and right frontal planes in the horizontal plane at a point 30mm above the reference plane with a -45° negative azimuth rotation in the horizontal plane.



Rear Negative Azimuth

At the posterior intersection of the median and horizontal plane at a point 30mm above the reference plane with a -45° negative azimuth rotation in the horizontal plane.



For each impact site, a one-way ANOVA with the repeated factor of head condition (4 levels: football, hockey, lacrosse, bare head) was used to analyze the dependent variable of peak resultant angular acceleration. When significance was found, post-hoc analyses were performed using Tukey's HSD test. The probability of making a type I error in all tests was $p < 0.05$. All analyses were performed using the statistical software package SPSS 16.0 for Windows.

III. RESULTS

The angular acceleration results of the impact protocol are shown in Figure 2. Linear acceleration of the head across the four head conditions was consistent with an average peak resultant linear acceleration between 21.1 and 21.5g. Peak resultant angular acceleration was dependent on the head condition and the impact site.

3.1 Angular acceleration results by head condition

When all centric impact sites were collapsed, peak angular acceleration results showed that the lacrosse helmet condition resulted in the highest peak resultant angular acceleration (2368.7 rad/s^2) followed by the hockey helmet (2315.6 rad/s^2), bare headform condition (1414.9 rad/s^2) and the American football helmet (1279.3 rad/s^2). The lacrosse and hockey helmets were statistically different from the bare headform and football helmet conditions ($p=0.003$).

Similarly, when all non-centric impact sites were collapsed, average peak angular acceleration results were the highest for the lacrosse helmet (3031.0 rad/s^2), followed by the hockey helmet condition (2986.2 rad/s^2), and the third highest values were found for the bare headform condition (1814.0 rad/s^2). The football helmet resulted in the lowest average angular acceleration of the head (1486.7 rad/s^2). The lacrosse and hockey helmet results significantly differed from the bare headform and football helmet results ($p < 0.0001$).

3.2 Angular acceleration results by impact site

Non-centric impact sites:

The effect of impact site on the head was also examined. At the front boss positive azimuth site,

the lacrosse and hockey helmets had the highest values of angular acceleration compared to the bare head condition and the football helmet ($p < 0.05$), whereas the football helmet produced the lowest values of angular acceleration ($p < 0.05$). Similar results were found at the rear boss negative azimuth site: the lacrosse and hockey helmet conditions produced the highest angular accelerations; the bare headform and football helmeted impacts showed significantly lower values ($p < 0.005$) although they were not significantly different from one another ($p = 0.509$). Similarly, at rear negative azimuth, the hockey helmet resulted in the highest angular acceleration values, followed by the lacrosse helmet and bare headform, while the football helmet resulted in the lowest angular acceleration of the head. At this impact site, statistical significance was found between each of the four head conditions ($p \leq 0.008$).

Centric impact sites:

The front positive elevation 15° site resulted in the lacrosse and hockey helmets having the highest values of angular acceleration compared to the bare head condition and the football helmet ($p < 0.05$). The football helmet produced the lowest values of angular acceleration ($p < 0.05$). At the crown site, a different pattern of angular acceleration results was observed. The highest angular acceleration was seen for the football helmet, followed by the lacrosse helmet and hockey helmet; the lowest values were seen for the bare headform condition. All differences were statistically significant ($p \leq 0.001$). Angular acceleration at the front boss CG site showed a similar pattern across the four head conditions to many of the non-centric impacts. The highest values were seen for the lacrosse helmet condition, followed by the hockey helmet and bare headform condition, and the lowest values were seen for the football helmet. Results for each condition significantly differed from one another ($p \leq 0.013$). The rear boss CG and side CG sites had an increase in angular acceleration in the following order: football helmet, bare headform, lacrosse helmet and hockey helmet. At these two sites, each comparison differed significantly from one another ($p \leq 0.006$). Rear CG results also showed a similar pattern, with the football helmet resulting in the lowest angular acceleration values, followed by the bare headform, and then the lacrosse and hockey helmets. At this site, only the lacrosse and hockey helmet comparison was not significantly different ($p = 0.963$).

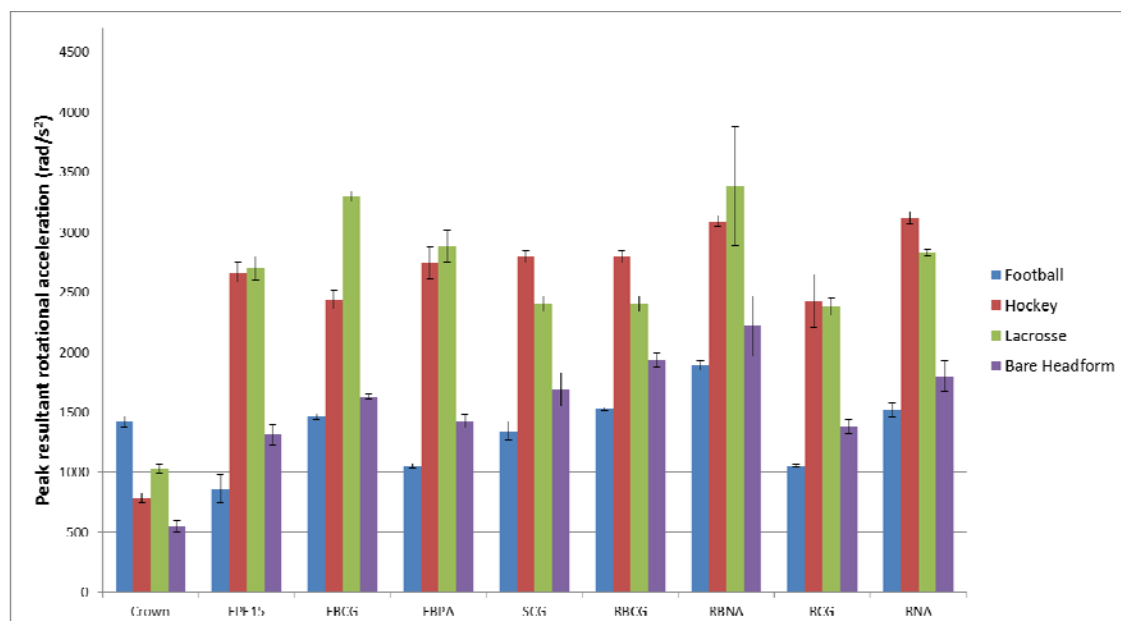


FIGURE 2. Peak angular acceleration results for impacts across all four head conditions. (Averaged across three trials; standard deviations shown as error bars.)

IV. DISCUSSION

A comparison of how different types of helmets (including a bare head condition) manage angular acceleration for sub-concussive impacts ($21.1 \pm 1.2g$) was studied. For non-centric impact sites the football helmet managed peak angular acceleration most effectively. The lacrosse helmet resulted in the highest angular acceleration values for all non-centric impacts except the rear negative azimuth location, where the hockey helmet had the highest values. Similarly at centric sites, angular acceleration was best managed by the football helmet, except at the crown, where the bare head condition had the lowest angular acceleration. For this site, wearing a helmet increases the magnitude of angular acceleration experienced by the head. For centric sites, the hockey helmet produced the highest angular accelerations with the exception of the front boss CG site where the lacrosse helmet resulted in the highest values. For low-level head impacts wearing hockey and lacrosse helmets increased angular acceleration when compared to the no helmet condition. However, a football helmet affords protection as compared to not wearing a helmet in all conditions except at the crown.

The ability of helmets to manage angular acceleration for low-level impacts is likely related to characteristics of each helmet type such as helmet geometry, liner type and thickness (Table 1). Football helmets have the greatest liner offset (3 cm) as compared to hockey and lacrosse helmets (1.4-1.5 cm). The football and hockey helmet liners were composed of vinyl nitrile (VN) foams while the lacrosse helmet was composed of expanded polypropylene (EPP).

The thicker VN liner of the football helmet may have been able to compress under impact in addition to having enough compliance to manage the shear-induced angular component. Additionally, the bare headform produced lower angular acceleration when compared to the thinner liners of hockey and lacrosse helmets. This phenomenon was observed for all impact conditions with the exception of the crown impact site. These unexpected results had the football helmet producing the highest values of angular acceleration and the no helmet condition with the lowest values. The crown impact site is a common location in helmet certification testing [24-26]; however, the impact energy levels used to certify these helmets are much higher often producing values around 100-200g up to maximum allowable value around 275g [33]. These helmets are certified using a guided drop onto a stationary anvil measuring protection based on linear acceleration criteria. The helmets used in this study may have performed poorly for low-level pendulum impacts as compared to the bare headform condition simply because they have not been optimized for angular acceleration for low-level impacts.

The angular acceleration results were also dependent upon the impact site. All impact sites and head conditions produced angular acceleration values between $549 - 3384 \text{ rad/s}^2$, which agrees with values reported by Rowson et al. for subconcussive impacts occurring around 1230 rad/s^2 and Pellman et al. 2003 for impacts resulting in no injury (4235 rad/s^2) in American football. While the levels of angular acceleration reported in this study are lower than those reported to cause concussion ($4500-6432 \text{ rad/s}^2$), they demonstrate that wearing head protection can increase the magnitudes of angular acceleration experienced by the head when levels of linear acceleration is maintained at 20g. Although these impacts are not at a level sufficient to produce a high risk for concussion, these levels of acceleration produce sub-concussive impacts [8][15].

In addition to peak resultant dynamic response variables, other characteristics of the impact may contribute to creating risk for head injury. At the sub-concussive level, risk of brain trauma may be introduced by angular acceleration as demonstrated in this study or by other characteristics of the impact such as the duration of the acceleration pulse [20]. Figure 3 demonstrates the duration of the angular acceleration pulse for each head condition at the front boss CG site. The bare head condition has the shortest duration of angular acceleration when compared to the helmeted conditions with the lacrosse helmet producing the longest duration. Adding compliance with a helmet may create different conditions for head injury risk as the head and brain tissues are subject to angular

accelerations for longer durations. Future research should encompass examining the risk of brain injury using tissue stress and strain measures from finite element analysis for these low level head impacts.

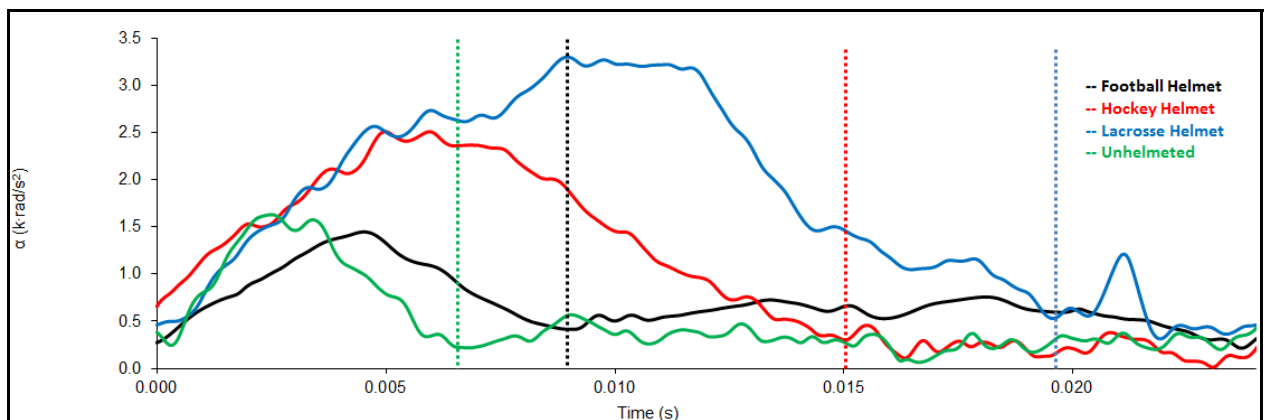


FIGURE 3. Duration of angular acceleration pulse for the four head conditions at front boss centre of gravity location.

The use of the Hybrid III head and neckform in the experimental set-up to examine angular acceleration is associated with some limitations. Previous literature has demonstrated such limitations imposed by the HIII anthropometric test device as a surrogate for a human head and neck. Such limitations include limited biofidelity of the HIII neck and neck joint [15]. Specifically, the stiffness of the dummy's neck has been questioned in terms of its ability to reflect an athlete's neck stiffness in real life. In vivo, an athlete can often brace himself or herself for impact, tensing muscles in the neck which increases the effective mass of the head [34]. Thus, the stiffness of the HIII likely should be higher in order to be more biofidelic.

The 50th percentile Hybrid III neck has been validated according to neck flexion and extension tests. Specific test criteria used for validation include velocity, pendulum deceleration, maximum occipital moment, and time at maximum occipital moment. Since validation of the neck is performed at higher energy levels than the 20g low-level impacts in this study, using the neckform for the level of impacts in this study likely did not influence the angular response values. Further, although the HIII was developed originally by the automotive industry for use in crash testing [34], it has been widely used to investigate head injuries in sport and is still a valuable set-up for measuring dynamic response data as a more feasible alternative to obtaining this data from human subjects [13][36-37].

V. CONCLUSIONS

This research examined peak resultant angular acceleration of football, lacrosse and hockey helmets as compared to a no helmet condition for low energy impacts. In all sites except the crown, the football helmet was able to decrease the magnitude of angular acceleration below those reported for the no helmet condition. At the crown site, the football helmet had the highest level of angular acceleration which may be the result of a combination of helmet shell geometry, liner type and thickness. Hockey and lacrosse helmets consistently produced the highest levels of angular acceleration, with magnitudes above 3000 rad/s² at the FBCG, RBNA and RNA sites for 20g impacts. Wearing a helmet increases the compliance of the impact system by increasing the duration of the acceleration experienced by the head. The thick liners characteristic of football helmets produce lower levels of angular acceleration as compared to not wearing a helmet for low level impacts; however, the relatively thinner liners of hockey and lacrosse helmets increase the values of angular

acceleration. With linear acceleration maintained at 20g, the type of helmet influences the level of angular acceleration experienced by the head.

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