

## Dynamic Head Response and Maximum Principal Tissue Strain for Helmeted (American Football) and Non-helmeted Impacts.

Karen Taylor, Thomas B. Hoshizaki, T. Blaine Hoshizaki, and Michael D. Gilchrist

**Abstract** The football helmet is credited with reducing the incidence of traumatic brain injury by protecting against linear impacts. Despite this, concussion rates remain high. It is proposed that linear acceleration, used as the sole metric to measure helmet performance, may be limited for predicting concussive injury. The purpose of this study was to compare bare and helmeted headforms impacts at five impact locations and four impact directions using peak linear acceleration, peak rotational acceleration, and maximum principal strain.

The results demonstrated the individual metrics; peak linear acceleration, peak rotational acceleration and maximum principal strain did not significantly vary over impact condition when viewing the bare and helmeted headforms separately. When reviewing the metrics individually in terms of risk prediction, reductions observed for peak linear (76%) and peak rotational acceleration (79%) for the helmeted conditions reflected substantial decreases in the risk of concussion (25-50% risk). Reductions for maximum principal strain (44%) were less compared to the dynamic head response results and indicate high risk of concussion (>80% risk). The reported differences in injury risk between dynamic head response values and brain tissue strains challenges the use of dynamic head response variables to measure the ability of helmets to protect against concussion.

**Keywords** dynamic response, helmet performance, maximum principal strain

### I. INTRODUCTION

American football reports the highest rate of head injury, including concussion, in North America [1-4]. The implementation of stringent rules, game changes, and helmet protection has attempted to reduce quantity and severity of head injuries [5]. In particular, the hard-shelled football helmet, implemented in the National Football League in the 1970s, has been credited with drastically reducing the incidence of traumatic brain injuries (TBIs) [5-7]. However, helmets are not designed to reduce rotational acceleration, often associated with concussive type injuries [8-9]. Concussion is a common but puzzling injury resulting from mechanical force causing pathophysiological damage to the brain and is diagnosed using symptoms [10-13]. Concerns relating to long term detrimental physical and mental effects of concussive injury have led to increased apprehension surrounding the effectiveness of current American football helmets [14-15].

Despite decades of research linking rotational acceleration and concussive injury, most helmet testing standards, including the National Operating Committee on Standards for Athletic Equipment (NOCSAE) test for American football, continue to require helmets to pass linear impact testing [16-18]. The few research studies that have investigated football helmet performance are conflicting regarding the correlation between linear acceleration and concussive injury [19-21]. Pellman (2003) used laboratory reconstructions to analyze open-field impacts in professional football resulting in concussions, noting the strongest correlation between linear acceleration and concussion with a mean threshold of injury to be 98g and a minimum of 75-76g to cause injury [6]. However, Greenwald (2008) investigated 17 concussive impacts taken from high-school and collegiate level players and noted that a combination of linear acceleration, rotational acceleration as well as impact location was required to predict the risk of concussion [22]. In agreement, Broglio (2010) reported that a combination of linear acceleration, rotational acceleration, and impact location was best to differentiate between 13 concussive impacts and non-injury events [23]. Guskiewicz (2007) studied collegiate football players reporting

Karen Taylor is a PhD student in Biomechanics in the Human Kinetics Department at the University of Ottawa, Canada (ktaylor078@uottawa.ca). Thomas B. Hoshizaki is an undergraduate student in Kinesiology in the School of Kinesiology at Lakehead University, Thunder Bay, Canada. T. Blaine Hoshizaki is Prof. of Biomechanics in the Human Kinetics Department at the University of Ottawa, Canada. Michael D. Gilchrist is Prof. of Engineering at the School of Mechanical & Materials Engineering at the University College Dublin, Ireland.

mean linear and rotational acceleration values of 103g and 5312 rads/s<sup>2</sup>, respectively, for 13 concussed players. Most notably they showed no correlation between the magnitudes of acceleration and the severity of injury related symptoms and were the first to postulate the difficulty in linking dynamic response and concussion for football impacts [24]. Current published values of 66, 82, and 106g as well as 4600, 5900 and 7900 rads/s<sup>2</sup> correspond to 25, 50 and 80% risk of concussive injury for linear and rotational acceleration respectively [25].

Kinematic measures, such as peak linear and peak rotational acceleration, are unable to fully describe the mechanism of injury or quantify the strain within the brain and brain tissue resulting in brain trauma [21][26-27]. Brain tissue strain, which leads to a metabolic cascade disrupting neural function, has been strongly linked to concussive injury [11-12]. In particular, maximum principal strain (MPS), a measurement of brain tissue stretch, is in close comparison to anatomic failure testing allowing for some comparison to rheological research [27][51]. MPS is determined using finite element modelling (FEM) and data from components of kinematic response curves (X, Y, Z) of linear and rotational acceleration [28][51]. FEM analysis allows for interpretation of how dynamic response of the head can influence strain to tissues throughout the brain resulting in injury [21][29]. Zhang (2004) utilised previously published professional football data as input to a finite element model showing highest correlation between shear strain and concussion followed by a combined metric using both resultant linear and rotational acceleration [6][25]. Kleiven (2007) attempted to improve upon these findings using a more extensively validated model to analyse concussive events from collegiate level football indicating high correlation between maximum tissue strain and rotational acceleration but not linear acceleration [30]. While FEM is a tool used to correlate neural tissue loads to actual injury, previous experimental research was conducted to determine brain tissue tolerances and influence of strain to resulting injury [21]. Margulies (1992) proposed human tolerances ranging as low as 5-10% resulted in Diffuse Axonal Injury (DAI), which is often considered a more severe injury than concussion [32]. Galbraith conducted experiments on brain tissue of squid giant axons and determined tissue elongation greater than 20% never fully recovered and that elongation above 25% caused structural failure resulting in injury [33]. Current published values determined from reconstructions of concussive football impact resulted in MPS values of 0.14, 0.19, and 0.24 corresponding to 25, 50 and 80% risk of concussive injury [25][30][34].

Research has yet to fully determine the influence of the helmet by comparing bare headform to helmeted impacts on dynamic response of the brain and resulting brain deformation over multiple impact locations and angles. Studies have shown concussive injury events result from a combined set of unique impact conditions that define how energy is transferred to the head and brain [27][35]. A key component to understanding concussive injury biomechanics, and thus helmet protection, is detailing the relationship between impact characteristics, head dynamic response and brain tissue deformation. Studies have shown injury can occur at almost any location on the head and from any direction, yet helmet testing standards are limited to single or minimal number of impact areas, typically through centre of gravity [36-37]. Rousseau reviewed the effect of impact location for hockey helmets and established that location affected the ability of the helmet to decrease both linear and rotational acceleration [38]. Gurdjian noted that two blows delivered in the same location but with varying lines of action produce markedly different effects on the brain [26]. Gennarelli (1979) demonstrated impact direction influences region and type of resulting injury [39]. Walsh (2011) identified certain impact location and angle conditions were capable of producing higher linear and rotational accelerations than others but did not include brain deformation metrics or comparison between bare and helmeted headform [40]. Gurdjian (1966) and Meaney (2011) identified that, regardless of the magnitude of acceleration, the brain responds differently when force is applied in lateral planes in comparison to the sagittal plane [26][41]. Without a full analysis of helmet performance using dynamic response and brain deformation over the entire helmeted surface it is not possible to reliably improve current helmet technology in order to reduce the risk of concussive injury. Therefore, the purpose of this study is to analyse bare and helmeted impacts using dynamic response and brain deformation to determine helmet performance across location and angle and to determine the usefulness of dynamic response as an evaluation tool for concussive injury risk.

## II. METHODS

A linear impactor system consisting of a weighted, pneumatically driven impactor arm, a Hybrid III head and hybrid neck-form on a sliding table and a computerised collection system was used to produce measureable three-dimensional bare headform and helmeted headform impacts. The linear impactor is used to represent various impact events including those such as helmet-to-helmet collisions seen in American football.

Helmet-to-helmet (including helmet-to-facemask) collisions represent 59% of all impacts in professional football resulting in injury [6].

**Experimental Testing**

The effect of impact location and angle was evaluated by impacting a bare 50<sup>th</sup> percentile Hybrid III headform with a linear impactor and then repeating the impacts to the helmeted headform. For bare and helmeted headform, three impact trials were conducted at each impact condition. All impact positions and angles were measured from a standardised headform position based on a front centric impact location. Impact site accuracy was ensured through the use of pre-marked locations on the headform. Impact conditions were defined by the relative orientation of the impactor to the Hybrid III structure. Rotational and translational adjustments were made at the interface between the Hybrid III structure and the translating table locking device. The five impact sites and four impact angles are defined in Tables 1 and 2 below. Typical set up depicting a Front PE bare headform, Front PE helmeted headform, are shown in Figure 1.

Table 1  
Headform impact site definitions

Site	Definition
Front	Anterior intersection of the mid-sagittal and absolute transverse plane
Front Boss	Midpoint between the anterior mid-sagittal and right coronal planes in transverse plane
Side	Right intersection of the coronal and absolute transverse planes
Rear Boss	Midpoint between the posterior mid-sagittal and right coronal planes in transverse plane
Rear Boss	Posterior intersection of the mid-sagittal and absolute transverse planes

Table 2  
Headform impact direction definitions

Angle	Definition
Center of Gravity	Perpendicular to the headform, no vertical or horizontal rotation applied
Positive Elevation	A 45° elevation of the Hybrid III head and neck-form relative to the impactor
Positive Azimuth	A 45° rotation of the Hybrid III head and neck-form in the transverse plane
Negative Azimuth	A -45° rotation of the Hybrid III head and neck-form in the transverse plane



Fig.1a) Bare Front PE Fig.1b) Helmeted Front PE Fig.1c) Bare Rear Boss CG Fig.1d) Helmeted Rear Boss CG  
Fig.1: Bare and Helmeted Impact Set ups – Example for Front Positive Elevation (PE) and Rear Boss Center of Gravity (CG)

**Testing Equipment**

The impactor arm and striker were accelerated pneumatically to a velocity of 5.5 m/s initiated by an electronic trigger. The chosen velocity is that of a low to medium impact seen in amateur football and low impact velocities seen in professional football eliciting concussive events [6][15][42-43]. Additionally, this is the maximum used within the NOCSAE helmet testing [44]. The apparatus consists of a support frame, a compressed air tank, piston and impactor arm (13.1 kg ±0.1kg), and a mobile table (12.8 kg ± 0.01kg) with attached rails. Attached to the end of the impacting arm was a nylon impactor cap with a 1 inch thick Modified Elastomer Polymer (MEP) insert. The MEP impactor cap elicits a dynamic response curve (shape and duration) similar to that seen in helmet-to-helmet collisions [15][18][36][45][56]. The impacting arm was propelled horizontally using pressurised air to impact the Hybrid III headform and fixed to the sliding table mentioned above. The 50<sup>th</sup> percentile adult male Hybrid III headform (mass 4.54 ± 0.01 kg) instrumented according to

Padgaonkar's orthogonal 3-2-2-2 linear accelerometer array protocol was fixed to a 50<sup>th</sup> percentile Hybrid III neck-form (mass  $1.54 \pm 0.01$  kg). Impactor parameters of mass, velocity and impactor compliance were unchanged throughout the testing procedure. The headform coordinate system was defined with a left-hand rule and positive axes directed anteriorly, toward the right ear and caudally for x, y and z, respectively. The set up allowed for adjustment in five degrees of freedom providing variable positioning. Once adjustments have been made and the positioning recorded, the levers and turns are locked in to place prior to impact. Upon impact to the headform, the entire assembly (headform, neck and table) was free to move down the rails until reaching the stop pads at the end of the support table (0.65 m).

### **Collection System**

The velocity of the impactor arm was captured by an electronic time gate (width  $0.2525 \pm 0.0001$  m) and recorded using National Instruments VI-logger software. The nine mounted single-axis Endevco 7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA, USA) were sampled at 20 kHz and filtered using the SAE J211 class 1000 protocol. The accelerometer signals were passed through a TDAS Pro Lab system (DTS, Calabasas, CA, USA) before being processed by TDAS software. The impact data were sampled at 20 kHz, filtered at 1000 Hz and recorded using DTS TDAS PRO software. The output from the DTS software is the x, y and z linear and angular acceleration loading curves as well as magnitudes of the peak linear and angular acceleration.

### **Computational Modeling**

In order to determine MPS, the University College Dublin Brain Trauma Model (UCDBTM) was used. The model used male cadaveric geometry obtained from medical imaging and includes a scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), cerebrum, cerebellum and brain stem [46]. The behaviour of the brain tissue was described as viscoelastic in shear with deviatoric stress rate dependent on shear relaxation modulus. The current finite element model, UCDBTM, used for this study is one of few partially validated models sourced for this type of research [58]. Validation of the model was completed using comparisons with intracranial impact cadaveric pressure data from Naham (1977) and Trosseille (1992) and brain motion from Hardy (2001)[47-48][61]. Additional validation was conducted using real world head injury events assessing the viability of the model for injury assessment [46][59-60]. This model has been used in several recent papers in the use for head impacts, head injury, helmet design, and concussion prediction [18] [62-68].

## **III. RESULTS**

The performance of the football helmet was evaluated over 20 impact conditions using dynamic head response and brain tissue deformation. The results presented in Figures 2, 3 and 4 depict peak linear acceleration, peak rotational acceleration and peak MPS respectively for the bare versus helmeted impacts over the 20 impact conditions. The values for linear acceleration ranged from 158g-368g for the bare headform impacts and 39g-92g for the helmeted headform impacts. The percent reductions due to the helmet over all 20 impact locations ranged from 59%-84%. The values for rotational acceleration ranged from 11,868 rads/s<sup>2</sup>-38232 rads/s<sup>2</sup> for the bare headform impacts and 3445 rads/s<sup>2</sup>-5959 rads/s<sup>2</sup> for the helmeted headform impacts. This resulted in a percent reduction due to the helmet ranging from 62%-88%. With regards to maximum principal strain, the values for the impacts to the bare headform impacts ranged from 0.41-0.78, while the helmeted headform impacts resulted in a range of 0.18-0.39. This corresponded to a percent reduction due to the helmet ranging from 25%-66%. To summarise, the average percent reduction due to the helmet, in comparison to the bare headform, for peak magnitude linear and rotational acceleration was 76 and 79% respectively. In contrast, the average percent reduction of MPS was 44%. Figure 5 indicates the percent reduction between the bare and helmeted headform at each impact condition for linear acceleration, rotational acceleration, and MPS. Analysing these results for all impacts, using an ANOVA and post hoc Tukey's or Dunnett's post hoc and adjusted for multiple comparisons ( $P < 0.05$ ), no significant differences occur in the reduction of linear and rotational acceleration when comparing bare to helmeted impacts. However, using the same analysis method significant differences were observed between the reductions in linear acceleration and MPS as well as between rotational acceleration and MPS.

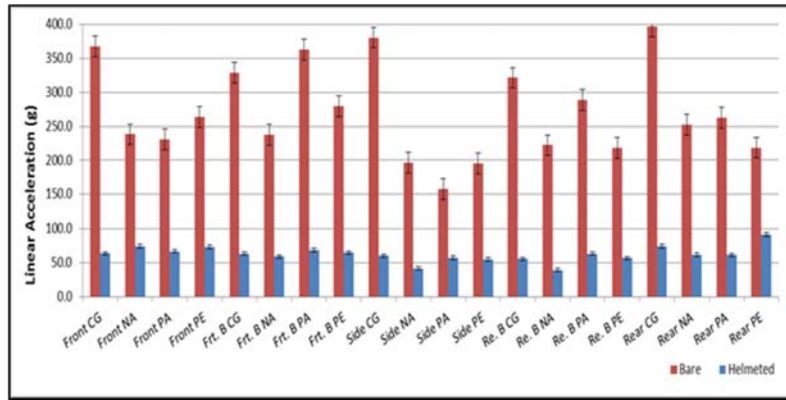


Fig. 2. Linear acceleration for bare vs helmeted impacts over 20 conditions (showing error bars with SD) (CG – centre of gravity, NA – negative azimuth, PA – positive azimuth, PE - positive elevation).

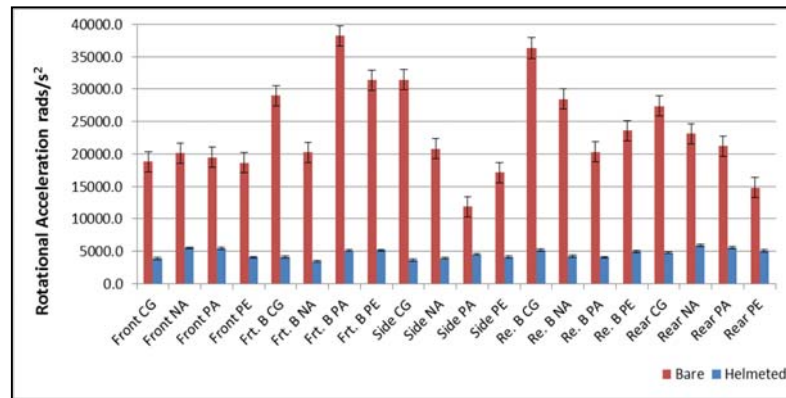


Fig. 3. Rotational acceleration for bare vs helmeted impacts over 20 conditions (showing error bars with SD) (CG – centre of gravity, NA – negative azimuth, PA – positive azimuth, PE - positive elevation).

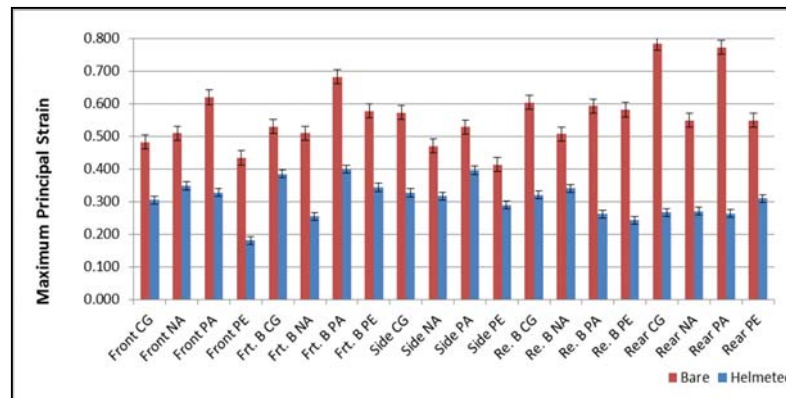


Fig. 4. Maximum principal strain for bare vs helmeted impacts over 20 conditions (showing error bars with SD) (CG – centre of gravity, NA – negative azimuth, PA – positive azimuth, PE - positive elevation).

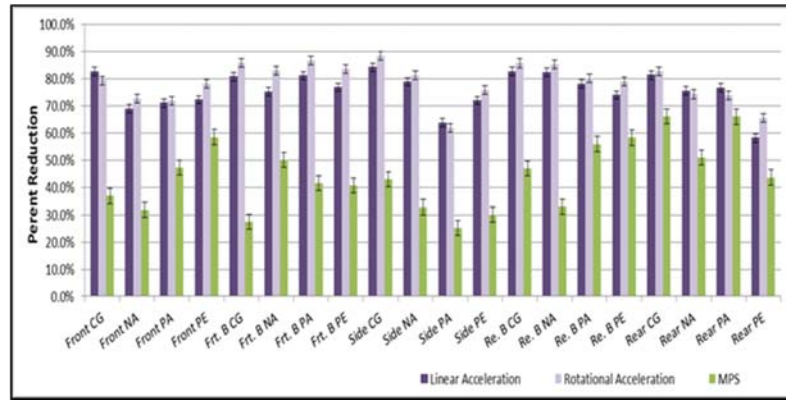


Fig. 5. Percent reduction of linear acceleration, rotational acceleration and MPS (CG – centre of gravity, NA – negative azimuth, PA – positive azimuth, PE - positive elevation).

Using an ANOVA and Dunnett’s post hoc to analyse helmet performance in terms of impact location, the results show no significant differences in the reduction in linear acceleration (Front – 75%, Front Boss – 79%, Side 77%, Rear Boss – 80%, and Rear 74%) and rotational acceleration (Front – 75%, Front Boss – 85%, Side – 80%, Rear Boss – 83%, and Rear 75%) over the five impact sites. However, significant differences in the reduction of MPS due to the helmet were indicated between the side (33%) and rear (58%) locations. No other significant differences in MPS reductions were shown between the other impact sites (Front – 43%, Front Boss – 40%, and Rear Boss – 49%). Figure 6 shows the impact magnitudes of the bare versus helmeted impacts averaged over location.

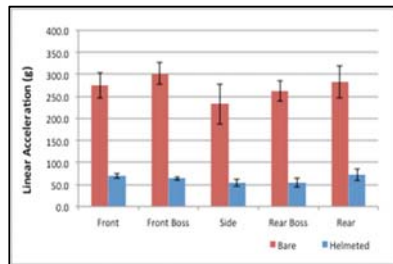


Fig. 6a. Linear acceleration

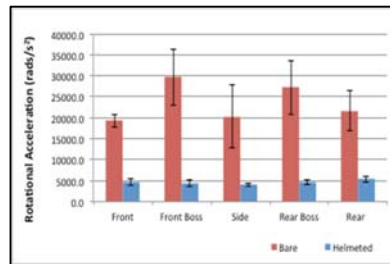


Fig. 6b. Rotational acceleration

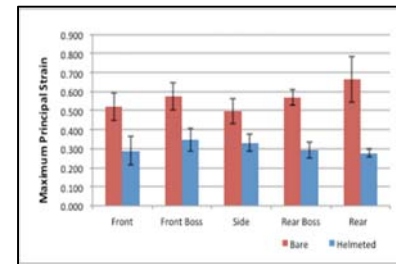


Fig. 6c. Maximum principal strain

Fig. 6: Bare versus helmeted impacts by location (showing error bars with SD).

Examining helmet performance in terms of centric versus non-centric impacts using ANOVA and Dunnett’s post hoc resulted in significant differences ( $P < 0.05$ ) in the percent reduction in linear acceleration between centric (centre of gravity – 82%) and the non-centric impact (positive elevation - 71%). No significant differences ( $P < 0.05$ ) were observed in the reductions of either rotational acceleration or MPS between centric and non-centric impacts. Figure 7 shows the impact magnitudes of the bare versus helmeted impacts averaged by centric and non-centric impact angles (centric – centre of gravity “CG”, non-centric – negative azimuth “NA”, positive azimuth “PA”, and positive elevation “PE”).

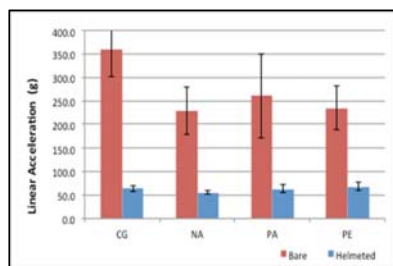


Fig. 7a. Linear acceleration

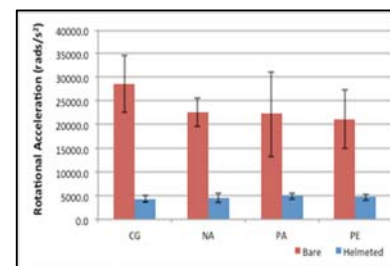


Fig. 7b. Rotational acceleration

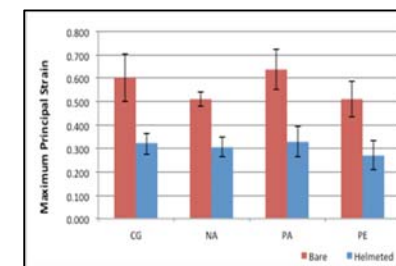


Fig. 7c. Maximum principal strain

Fig. 7: Bare versus helmeted impacts by angle/direction (showing error bars with SD).

In addition, for all bare headform impacts, using Pearson Correlation, moderate correlation exists between linear and rotational acceleration, low correlation between linear acceleration and MPS, and no correlation between rotational acceleration and MPS. However, for the helmeted impacts, no correlation occurred between linear and rotational acceleration, linear acceleration and MPS, or rotational acceleration and MPS.

#### IV. DISCUSSION

Evaluating the performance of a football helmet by comparing bare to helmeted impacts, the reductions in peak linear acceleration, peak rotational acceleration and MPS are consistent across impact location and direction for each metric independently. Secondly, while the reductions are similar for the dynamic response metrics; peak linear acceleration (average reduction = 76%) and peak rotational acceleration (average reduction = 79%), the average reductions observed for MPS were less (average reduction = 44%).

This study reports that when viewing each metric separately, the helmet protects to a similar degree regardless of the location of the impact to the head. While this contradicts some previous bare head impact literature, which reports vulnerability of the head at different impact sites and directions, this study compared bare to helmeted impacts where the particular helmet tested was rounded with a smooth surface, which may mask the variation in the geometry of the human head [49-51]. Helmets with geometric structures have been shown to influence impact response by modifying the elastic response of the shell [52]. Some external geometric configurations on the shell can act as levers, which increase rotational torque to the head while geometry that can reduce the angle of inclination can diminish the rotational component of an impact [52-53]. An additional explanation for consistent helmet performance observed in this study is the uniformity of liner material and thickness (~1.5"-2.0") in the helmet tested. There is a limit to the helmet liner thickness due to comfort constraints and practicality of use. This becomes important when evaluating helmets of mixed liner materials or variable liner thicknesses as variation in helmet performance may occur [54]. Therefore, it is possible that while no variation in dynamic response or maximum principal strain was observed using the football helmet in this study, inconsistency in helmet performance may be observed when using helmets with different geometric features, non-uniform liner materials or thicknesses.

The second finding of the study is the dissimilarity in the average reduction between dynamic response and brain deformation, which becomes important when examining prediction of injury risk. It is important to note that while the impact characteristics used in this study are within ranges known to cause concussive injury in football they do not represent actual concussive injury reconstructions [6][15][36][42][45]. Based on published risk of injury criteria, under these conditions the helmet was able to reduce both peak linear and peak rotational acceleration magnitudes below 50% risk of injury with some locations lower than 25% risk of injury [25][30][34]. In comparison, for the same impacts to the helmeted headform, all values of maximum principal strain remained above 80% risk of injury for concussion [25][30][34]. Therefore, in terms of injury risk and for the impact conditions used in this study, the helmet performed well at reducing linear and rotational acceleration but not at reducing MPS. This suggests that under these impact conditions, the combined mechanical properties of the helmet may be capable of absorbing enough energy to sufficiently reduce the peak magnitudes of acceleration of the brain. However, the reduction in peak acceleration observed may not translate into a sufficient enough reduction in overall motion of the brain and typically occurs with an increased impact duration. Both of which can alter the shape of the dynamic response curve and therefore may explain why the corresponding MPS values to remain high [54-55].

The difference between percent reduction of dynamic head response and MPS supports the notion that current testing protocols utilising linear acceleration as an indicator of helmet performance may not be adequate. Recently, researchers have proposed the inclusion of a rotational component into helmet testing protocol, which may increase the robustness of helmet testing [56]. However, this research suggests that including both dynamic response variables still may not adequately measure helmet performance.

#### V. CONCLUSIONS

The football helmet performed consistently in terms of percent reduction of dynamic response (peak linear and peak rotational acceleration) and brain deformation (MPS) over different impact conditions. Under the specified impact conditions, the football helmet works well at reducing dynamic response, peak linear acceleration and peak rotational acceleration. The football helmet does not work as well at managing MPS. These findings indicate that different metrics may result in varying outcomes suggesting that the use of a single metric may not be advisable to evaluate helmet performance or predict injury.

The reconstructions within the laboratory make use of a non-biofidelic physical headform. These models are impacted to yield linear and angular acceleration loading curves, which may not result in exactly similar responses to those of the human head, neck musculature, body mass, and overall human response to injury. Although it has been well documented that effective mass plays a significant role in impact mechanics, this study was not conducted as an exact reconstruction of impacts experienced in football but as a comparison of bare headform to helmeted head mechanics. While the model has its limitations, such as assumptions surrounding the characteristics of brain tissue and the interactions between different parts of the brain and skull, it is an approximation intended to simulate human responses to injurious loading. Therefore, it is a useful tool to compare between different impacts giving relative values rather than an absolute [18].

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