An examination of American football helmets using brain deformation metrics associated with concussion

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Abstract

The sport of American football is associated with a high incidence of concussion, which research has identified may lead to long term neurological damage. As a result, it is important that protective technologies be developed to help mitigate the incidence of this type of brain trauma. This research examines how the design characteristics between different American football helmet models affect the linear and rotational acceleration responses as well as brain deformation metrics using a centric/non-centric impacting protocol. The protocol involved impacting the helmets at nine centric/non-centric sites. Brain deformation metrics were calculated using the University College Dublin Brain Trauma Model. The results revealed that design characteristics do influence the brain deformation metrics associated with incidence of concussion. Further analysis revealed that rotational acceleration was more related to brain deformation metrics than linear acceleration. These results show that when attempting to reduce brain deformation metrics, the development of rotational acceleration diminishing technologies may be beneficial. This research indicates that helmet design may be able to reduce the risk of concussive injury.

Keywords: Concussion, American football, Helmets, Impact Biomechanics, Protective devices

1.0 Introduction

American football is a contact sport where helmets are required to decrease the risk of traumatic brain injury (TBI) [1]. Helmets have been largely successful in managing these types of injuries, however mild traumatic brain injury remains a problem [2]. Until recently, concussion was not considered a serious injury and as a result helmet manufacturers did not consider developing helmets specifically to deal with concussions. However, considering the seriousness of both short and long term effects of concussions [3], more attention has been given to methods of preventing and reducing the incidence of brain trauma using technology [4; 5; 6; 7].

Originally, American football helmets were primarily designed to prevent TBI injuries [1]. Originally, American football used cloth and then leather helmets to help prevent this type of injury, which was leading to deaths within the sport. As technology advanced, the capacity of the American football helmet and other sports helmets to prevent TBI increased, and is now at the point where these injuries are relatively rare. The mechanism of traumatic brain injury has been well documented and has been found to be associated with peak linear acceleration responses to head impacts [8; 9]. As a result, the impact absorbing materials used in American football helmets are designed to mitigate large magnitude linear accelerations induced from impact. The polycarbonate shell was designed to spread the impact force across the energy absorbing liner. The materials used to absorb the impact energy varied from suspension helmets to vinyl nitrile (VN) foams and expanded polypropylene (EPP). Currently, most American football helmets use VN or EPP foams to manage the energy of impact as they are useful for multi impact scenarios. Some helmets use newer novel three dimensional engineered structures (Schutt DNA and Xenith X1) to manage the impact energy which use geometry and material characteristics to mitigate linear acceleration instead of density and thickness as is prevalent in the use of foams. These materials are designed to reduce the magnitude of linear acceleration incurred from impacts in American football.

To evaluate the performance of the helmets, testing methodologies and performance measurement of American football helmets have primarily involved linear acceleration [1]. These test methodologies also use threshold values (300g or 1200 GSI) which are based upon cadaveric skull fracture thresholds [1; 10]. Thus the materials developed for use in American football helmets were designed to mitigate linear acceleration. As a result of the use of helmets the incidence of skull fractures and other TBI such as subdural hematoma are rare in the sport of American football. Concussion has a different mechanism to injury than TBI, and as a result American football helmets are ill equipped to mitigate this type of injury [5; 11; 12; 13; 14]. Unlike TBI, concussion has been shown to be associated with rotational acceleration when both linear (through the centre of gravity, or centric) and non-linear (not through the centre of gravity, or non-centric) impacts are used [5; 6; 15]. While current materials used in American football helmets are good for the reduction of linear acceleration, the effect they have on rotational acceleration has yet to be characterized. To date, there have been no technological developments in American football helmets which are designed to mitigate the severity of injuries known to have a rotational component in their mechanism. To accomplish this development an appropriate test method must be used which measures these accelerations and allows helmets to be compared.

American football helmets have not yet been evaluated using a centric/non-centric protocol and under rotational acceleration conditions. Consequently, it is not known how different American football helmets perform when tested under rotational acceleration. However, neither linear acceleration nor rotational acceleration on their own are suitable metrics for predicting the risk of concussion. The low correlation between peak linear and rotational accelerations and concussion may be explained by the effects that the dynamic response of an impact have on brain tissue [4].

The disparity between linear and rotational acceleration loading curves and concussive injury can be researched using finite element models of the human brain [16;

13; 17]. Employing finite element models to calculate brain tissue deformation allows for these interactions between linear and rotational acceleration to be evaluated within the context of brain tissue characteristics. Researchers have proposed a series of thresholds using brain tissue deformation metrics based on reconstructed concussive events using football impacts and other real world accidents [16; 13; 17]. This research supports the notion that using values of brain tissue deformation is important when assessing whether a helmet will manage the risk of injury and there is a likelihood of sustaining a concussion [4]. Finite element modelling has been used to evaluate energy absorbing liners and shell geometries in the past [18; 19; 20]. However, it is currently unknown whether different models of the American football helmet using different energy absorbing liners can influence the amount of brain deformation incurred through a centric/non-centric impact. This information would be useful to future design considerations that will seek to reduce those brain deformation metrics that are known to be associated with risk of concussion.

The purpose of this research is to examine three distinct models of the American football helmet under a centric/non-centric impact protocol and thus to determine if different helmet designs (geometry, material) influence the levels of stress and strain in those regions of the brain that are associated with concussion.

2.0 Methodology

Three American football helmet models were impacted using a pneumatic linear actuator. The impacting device consisted of a frame housing an air tank and piston impacting arm. The impacting arm had a mass of 13.1 ± 0.1 kg and was propelled at the helmeted Hybrid III headform at a velocity of 7.5 m/s [21]. The tip of the impacting arm consisted of a hemispherical nylon striker cap which included a vinyl nitrile 602 foam layer. The football helmets were affixed to a Hybrid III headform which, in turn, was attached to a sliding table (12.78 ± 0.01 kg) designed to simulate the kinematics of a player's movement after an impact to the head. The sliding table was built to allow the Hybrid III headform be adjusted in five degrees of freedom and locked into position for the duration of the impacts.

A 50th percentile Hybrid III male headform (4.54 ± 0.01 kg) was instrumented with Endevco 7264C-2KTZ-2-300 uniaxial accelerometers according to Padgaonkar's [22] 3-2-2-2 array for the collection of three-dimensional kinematics. The accelerometers were sampled at 20 kHz and filtered using a 2nd order 1000 Hz low pass Butterworth filter. The signals were collected by Diversified Technical Systems TDAS Pro lab module and processed using TDAS software.

2.1 Helmets

Three commercially available helmets were tested using a centric/non-centric impacting protocol. Each helmet was fitted according to manufacturer's specifications on the Hybrid III headform. Descriptions of the physical characteristics of the helmets are found in Table 1. One helmet had a vinyl nitrile (VN) foam liner in certain locations in the helmet. The other two helmets had three dimensional structure liners in the form of either a thermoplastic polyurethane(TPU) structure liner or a structure and fluid venting system liner. The 3D structure liner is a structure which relies on aspects of its design to reduce the impact forces rather than material density, as is common with conventional foams (VN). These design characteristics include various factors such as wall stiffness, geometry, material characteristics, and fluid venting.

2.2 Procedure

Nine different impact sites (Figures 1 and 2), representing both centric and non-centric loading conditions, were used to evaluate the performance of the three helmets models [15]. Three of each helmet types were tested under these conditions to provide data to analyse statistical variability without repeatedly impacting each helmet at the same location, which would have induced progressive degradation. Overall, 81 impact tests were conducted. The velocity used for all of the impacts was 7.5 m/s, which was typical of collisions in American football [21]. The resulting linear and rotational acceleration loading curves were then applied to the University College Dublin Brain Trauma Model in order to analyse the resulting brain deformations. These were measured using the common metrics of maximum principal strain (MPS) and von Mises stress (VMS). All results were compared statistically using an ANOVA with a 95% confidence interval.

2.3 Finite element model

The finite element model used for this research was the University College Dublin Brain Trauma Model (UCDBTM) [23; 24]. The geometry of the model was developed from computed tomography (CT), magnetic resonance imaging (MRI) and sliced colour photographs of a male cadaver to represent the human head and brain. The three-dimensional finite element model includes the scalp, three-layered skull (cortical and trabecular bone), dura, cerebrospinal fluid (CSF), pia, falx, tentorium, cerebral hemispheres, cerebellum and brain stem. In total there are approximately 26 000 elements in the model. The baseline UCDBTM was validated against cadaveric pressure responses conducted by Nahum et al [25] and brain motion research conducted by Hardy et al. [26]. Further validations were conducted by Doorly and Gilchrist [27] on reconstructions of real world traumatic brain injury incidents with agreeable results. The material characteristics of the model were taken from Ruan [28], Willinger et al. [29], Zhou et al. [30], and Kleiven and von Holst [31] (Tables 2 and 3). A linearly viscoelastic constitutive model combined with large deformation theory was used to model the brain tissue [23]. The material was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [23]. The compressive nature of the brain was considered elastic. The nature of the shear characteristics of the viscoelastic behaviour of the brain was defined as:

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

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor [23]. To simulate the brain skull interface, the CSF was modelled using solid elements with a high bulk modulus and low shear modulus so that it would behave in a similar fashion to a fluid. The contact interaction at this interface specified no separation and used a friction coefficient of 0.2 [32].

2.3.1 Brain regions

While the UCDBTM differentiates between grey and white matter and the cerebrum, cerebellum and brain stem, the brain was further separated into regions associated with the symptoms of concussion so as to analyse the differences between the American football helmet models (figure 3) [33]. The prefrontal cortex is known to be involved in decision making and social behaviour [34]. The dorsolateral prefrontal area is involved with memory and organization. The motor association cortex and primary motor cortex are involved in the planning and execution of movement patterns [35]. The primary somatosensory cortex is used in sense of touch and proprioception. Sensory integration and perception of the environment are located at the sensory and visual association areas [36]. Visual information is processed at the visual cortex [37], while sound processing occurs at the auditory cortex [38]. Finally, the auditory association area performs functions associated with recognition of sounds and sound recognition. Any injury to these (10) areas of the cerebrum may be represented by losses of functioning as represented by these distinct regions. It is recognized that there are other ways to represent these regions when it comes to concussion and functionality [6; 13; 17]. The regions chosen here are comprised of both grey and white matter. This method is proposed as a way to compare relative changes in levels of stress and strain in these regions which are directly linked to concussive symptoms. The brain stem was not analysed since it does not exist in the model, although it would be suitable for further analyses as it can be associated with the loss of consciousness, a key symptom of concussion.

4.0 Results

The results from the impact protocol are shown in Tables 4, 5 and 6. The results from the auditory association area were omitted due to unresolvable errors in the model related to the layer of elements present between the tentorium and the brain tissue in that region.

4.1 Influence of helmet models on brain tissue response

When examining the influence of helmet model and liner material on the resulting magnitudes of kinematic and brain deformation response for site 1, no differences were found for linear acceleration. The rotational acceleration magnitudes were different between helmet A and helmet B but not helmet C ($p<0.05$). For the brain deformation responses, the only region of the brain where significant differences were found was where the peak magnitude VMS occurred (visual association area) where helmet A had a significantly higher response than helmet B, but not helmet C ($p<0.05$). For the MPS response, helmet A yielded larger magnitudes than helmet B in the dorsolateral prefrontal area.

At site 2, the linear response between the helmet models were not significantly different. The rotational acceleration magnitudes were different between helmet A and helmet C but not helmet B ($p<0.05$). The regional brain deformation responses showed no difference between the helmets at this site for VMS ($p<0.05$). There were no differences between the helmets across the cerebrum for MPS.

At site 3, the linear and rotational acceleration response was not different between helmets. The VMS magnitudes showed that there were no differences between brain regions with this metric ($p<0.05$). The MPS response for this site yielded larger magnitudes for helmet A when compared to helmet B for the prefrontal cortex and visual cortex. Helmet C also yielded larger magnitudes in MPS than helmet B.

Site 4 had no significant difference between the helmets for linear or rotational acceleration ($p<0.05$). The VMS response showed a difference between helmet C and helmet B but not helmet A at the visual cortex region. The MPS response at this site showed no significant differences between the helmets.

Site 5 showed a significant difference between helmets A and C as well as B and C for linear acceleration, with helmet B yielding the lowest acceleration (66.4 g) ($p<0.05$). Rotational acceleration response was not significantly different between the helmets. For VMS, helmet C yielded a significantly lower result (5.56 kPa) than helmet B, but not helmet A. For the MPS response at this site, helmet B yielded larger magnitudes than helmet A at the prefrontal cortex. At the visual cortex helmets A and B produced lower results than helmet C ($p<0.05$).

For site 6, linear acceleration was significantly different between helmet A (49.9 g) and helmet C (57.2 g). There was no significant difference in the rotational acceleration responses between helmet models ($p<0.05$). The VMS response indicated that helmet A was significantly different from helmet C and C from B at the visual cortex region of the brain. The MPS response yielded lower magnitudes for helmet C when compared to helmet B, but similar results to helmet A at the dorsolateral prefrontal area ($p<0.05$). Helmets A and B were similar with helmet C yielding significantly lower magnitudes MPS at the visual cortex, sensory association area and visual association area.

Site 7 yielded significant differences in linear and rotational acceleration between helmet A and helmet C ($p<0.05$). The VMS showed differences between helmets A and C as well as B and C for the dorsolateral prefrontal, motor associated cortex, primary motor cortex, auditory cortex, sensory association area and visual association area. For MPS response, helmet A produced significantly lower magnitudes than helmet C but not helmet B at the motor association cortex, primary motor cortex, auditory cortex, sensory association area and visual association area ($p<0.05$). Helmet C produced larger magnitudes for all of these regions.

Site 8 produced significantly different results between helmets A and B, although there were no significant differences between helmets for rotational acceleration ($p<0.05$). The VMS response from the simulation produced significantly different results between helmet A (6.45 kPa) and helmets B (7.09 kPa) and C (7.15 kPa) at the prefrontal cortex region of the brain. There were also significant differences found at the visual cortex region between helmets A and B ($p<0.05$). The MPS response yielded significantly lower results for helmet A over helmet B; both helmets produced similar results to helmet C for the visual cortex region.

Site 9 yielded significantly different results between helmet B and helmet C for linear acceleration. This impact site also produced significant differences in rotational acceleration between helmets A and C as well as between B and C ($p<0.05$). The VMS response from the simulations yielded significant differences between helmets A and C and helmets B and C for the dorsolateral prefrontal area, motor associated cortex, primary motor cortex, prefrontal cortex, auditory cortex, primary somatosensory area, sensory association area and the visual association area ($p<0.05$). For the MPS response, helmet A and helmet B were equivalent but produced lower magnitudes than helmet C for the dorsolateral prefrontal area, motor association cortex, auditory cortex, primary somatosensory cortex, sensory association area and the visual association area. At the primary motor cortex, helmet B produced the lowest magnitudes (0.171) and helmet C the largest (0.311) ($p<0.05$). At the prefrontal cortex, helmet C produced larger magnitudes MPS than helmets A and B. Helmet A produced significantly lower magnitudes of MPS (0.161) at this region than helmet B (0.206) ($p<0.05$).

4.2 Correlations

Pearson correlations were run to examine the relationship of peak linear and rotational accelerations on von Mises stress and maximum principal strain in the different regions of the brain (Tables 7 and 8).

5.0 Discussion

5.1 Helmet influences on injury metrics

Upon analysis of the different models of the American football helmet it became evident that differences in geometry and energy absorbing characteristics in the liner does influence the resulting kinematic and brain deformation metric responses. Significant differences were discovered between the three helmets tested at many of the impact sites. As has been shown by previous research, differences in geometry and energy absorbing characteristics of the liner influenced the shape and duration of the linear and rotational acceleration loading curves and in so doing changed the magnitude and location of brain deformation responses [6; 40; 41]. When linear and angular accelerations are used as the comparative measure to judge performance (Table 4), helmet A produced lower values of linear acceleration, while helmets A and C tended to produce the lowest magnitudes of rotational acceleration, with those of helmet B being slightly higher. However, when using maximum principal strain as a measure of performance (Table 5), helmet B followed by helmet C performed the best (lowest) results. When analyzing the results using von Mises stress (Table 6), helmet C followed by helmet B consistently produced lower peak results than helmet A (i.e., C was best and A was worst). Considering helmet A was the best performer for linear acceleration and not the best performer on the other performance metrics this shows that low linear acceleration does not necessarily produce lower rotations or brain deformation metrics. It is likely that the rotational acceleration incurred by the headform and resulting high brain deformations are a result of the location of impact and 'coupling' between the impactor and the helmet. This coupling effectively 'grabs' the helmet and causes a more aggressive rotation of the head underneath. To reduce headform rotations which are correlated to brain deformations through the use of helmets a couple options may be possible. It is likely that reducing the amount of coupling to the helmet shell or controlling the transfer of rotations using innovative liner technologies would be beneficial.

5.2 Helmet material characteristics

As helmet A has a liner constructed of vinyl nitrile material (i.e., isotropic and homogeneous) and helmets B and C of three dimensional engineered materials (i.e., anisotropic and inhomogeneous), these results would indicate that when brain deformation metrics are used as indicators of performance, designing protection using engineered materials may be more effective. Vinyl nitrile is only best at attenuating linear accelerations [1]. The testing and product certification methodology under which football helmets are designed only involves measuring linear acceleration: as a result the material would be expected to perform well under those conditions. The effect of rotational acceleration on VN foam has not been characterized and compared to that of engineered materials. However, the behaviour of VN foam under the action of rotational acceleration has been compared to that of expanded polypropylene (EPP) in hockey helmets and it was found to be superior [6]. Its better ability to dampen rotational energy is believed to be because of an ability to dissipate more energy through shearing and torsion of the material than the EPP although neither material has been designed for this purpose [6]. The increased performance of the engineered materials over the VN foam is due to the many characteristics that can be optimized to dissipate both linear and rotational impact energy. While VN foam is limited in its energy absorbing ability by liner thickness and density, the engineered materials can be modified in terms of constituent materials, stiffness, geometry, and other characteristics, all of which can be optimized specifically to dissipate rotational energy. As a result, both the linear (centric) and angular (noncentric) acceleration loading curves generated through impacts to helmets using this type of material produce lower brain deformation values, and consequently less severe levels of injury than can a conventional isotropic homogeneous material such as VN foam, which only tends to be superior under linear (centric) impacts.

5.3 Risk of concussion measures

The peak brain deformations (i.e., MPS and VMS) were also analysed according to regions of the brain associated with the symptoms of concussion [2]. The differences in helmet design were seen to influence the regions of peak brain deformation in the

UCDBTM at specific impact sites. For example the noncentric impact loading conditions of impact site 7, when applied to each helmet model, produced large maximum principal strains in distinct regions of the brain tissue. Many of the eight other impact conditions also led to similar locations of peak brain deformations. These results suggest that it may be possible to affect not only the magnitude of the deformation but also the region of the brain in which the peak is incurred using helmet design.

When comparing the results of the centric/non-centric impact conditions to threshold values for concussion found in the literature through anatomical and reconstructive research (Tables 9 – 11), the brain deformation values reveal a high likelihood of many of the impacts producing a concussive injury. The present results are significant because the VMS and MPS values suggest a high risk of concussion when compared against the available brain injury literature, while the peak linear and rotational acceleration responses are well below most suggested tolerance levels (Tables 9-11). This indicates that the current method of evaluating the performance of helmets, which only uses linear acceleration and linear dominant impacts [1], may not consider enough information to be effective in reducing concussive injury. These results support the hypothesis that using linear and rotational acceleration as well as brain tissue responses may be the best way to evaluate a helmet's ability to mitigate the risk of concussion [4].

The methodology used in this research demonstrated that the centric and non-centric testing protocol was sensitive and robust enough to distinguish differences in performance between the helmets which would have not been evident had only linear acceleration been used. This difference in performance may be a reflection of differences in the geometry of each model as well as the influence of the energy absorbing liner. The results clearly indicate that the testing methodology does indeed show that these helmets perform differently from each other. The impact sites used in this method were based upon risks of injury using reconstructive research and until now have not been examined for their appropriateness for helmet testing [15]. The present results show that these impact sites do serve to test the performance of the helmet for centric and non-centric conditions in a sensitive and robust manner.

5.4 Analysis of correlation results

The Pearson correlations were examined to see how linear and rotational accelerations may be linked to induced brain deformations, as measured by von Mises stress and maximum principal strain. Peak linear and peak rotational accelerations were not significantly related to each other, reflecting the importance of measuring both kinematic response measures when examining the performance of protective technologies. Also, peak linear acceleration was not correlated with either VMS or MPS values in any region of the brain, whereas peak rotational acceleration had moderately significant ($p>0.05$) correlations with VMS and MPS in almost every region of the brain tissue. These results are significant in that they show that if the goal of helmet design is to reduce the risk of concussive injury measured by brain deformations, then efforts to reduce the amount of rotational acceleration would be more effective than developing improvements in linear acceleration. These results also show the importance of developing a robust and reliable testing protocol to effectively assess the performance of American football helmets using both linear and rotational acceleration along with brain deformation measurements.

6.0 Conclusions

The data presented in this paper demonstrate that current American football helmets influence kinematic and brain deformation responses when evaluated using a centric/non-centric impacting protocol. The Pearson correlations presented in this study reveal a significant relationship between rotational acceleration and levels of brain tissue deformation. If brain deformation metrics are important in guiding the design of helmets, then developing technologies which manage rotational acceleration as well as linear acceleration will be of value.

The results indicate that future innovations to designs of American football helmets could reduce the magnitude of dynamic response and brain deformation metrics resulting from impacts. More specifically, it may be of value to develop technologies which cause a controlled decoupling between the impact source, helmet, and head to reduce rotational acceleration. Furthermore, linear accelerations were not significantly correlated with those brain deformation metrics believed to be associated with concussion, which may explain why current helmets do little to manage this injury.

The use of physical and finite element models impart certain limitations to this type of research. The use of a Hybrid III headform allows for easily controllable impacting conditions but does not permit biofidelity of the system. As a result, since the headform is steel and the neckform is stiff, they are unlikely to produce dynamic responses that are identical to those incurred by a human head. Also, this physical model system was designed for antero-posterior impacts and not for centric/non-centric impacts and thus may bias the results. The UCDBTM is a representation of the human head and brain and provides estimations of brain tissue damage. Its material properties and boundary conditions which govern its reaction to the dynamic responses supplied by the Hybrid III are approximations made from cadaveric studies and thus may not fully simulate the living human. Adjustments to the constitutive material data will produce different results than those that were found in this research. The protocol used does not represent actual injury scenarios but rather possible impacts which could occur in American football. Changes such as velocity and mass of impactor would influence the results and are the focus of ongoing research.

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Figure Captions:

Figure 1. Impact sites 1-5 (i.e., centric) on the American football helmets

Figure 2. Impact sites 6-9 (i.e., noncentric) on the American football helmets

Figure 3. Depiction of the regions of the brain analyzed by the UCDBTM.

Table Captions:

Table 1. Helmet characteristics

Table 2. Material properties of the UCDBTM

Table 3. Material characteristics of the different areas of brain tissue in the UCDBTM

Table 4.The peak linear and rotational accelerations for impacts to the American football helmets.

Table 5.The UCDBTM simulation maximum principal strain results for the impacts to the American football helmets.

Table 6.The UCDBTM simulation von Mises stress results for the impacts to the American football helmets.

Table 7.Pearson correlation results for dynamic response to maximum principal strain for the American football helmet impacts (95% confidence interval).

Table 8.Pearson correlation results for dynamic response to von Mises stress for the American football helmet impacts (95% confidence interval).

Table 9.Proposed concussion thresholds for dynamic response from reconstructive research.

Table 10.Proposed concussion thresholds for reconstructive research using finite element modelling.

Table 11.Proposed concussion thresholds from anatomical research.