# The Use of Decoupling Structures in Helmet Liners to Reduce Maximum Principal Brain Tissue Strain for Head Impacts

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# Abstract

The primary goal of the American football helmet has been protection of players against skull fractures and other traumatic brain injuries (TBI) [Cantu 2003, Benson 2009]. TBI can result from short, high magnitude linear impact events typical of when the head impacts a hard surface [Gilcrhist 2003, Doorly 2007]. The modern helmet, which has evolved and become well designed to mitigate TBI injuries, does not offer sufficient protection against injury such as concussion, and the incident rate remains high in sport [Broglio 2009, Rowson 2012]. Researchers speculate rotation of the head leads to shear strain on the brain tissue, which may be the underlying mechanism of injury leading to concussive type injuries [Gennarelli 1971, Ommaya 1974, Gennarelli 1982, Prange 2002, Gilcrhist 2003, Aare 2003, Zhang 2004, Takhounts 2008, Greenwald 2008, Meaney 2011]. This has led researchers to investigate new liner materials and technologies to improve helmet performance and include concussive injury risk protection by attempting to address rotational acceleration of the brain [Mills 2003, Benson 2009, Caserta 2011, Caccese 2013]. To improve current football helmet designs, technology must be shown to reduce the motion of the brain, resulting in lower magnitudes of dynamic response thus reducing maximum principal strain and the corresponding risk of injury [Margulies 1992, Zhang 2004, Mills 2003, Kleiven 2007, Yoganandan 2008 Caserta 2011, McAllister 2012, Caccese 2013, Post 2013, Fowler 2015, Post 2015a/b]. Recent research has studied the use of decoupling liner systems in addition to the existing liner technology, to address resultant rotational acceleration. However, none of this previous work has evaluated the results in terms of the relationship between brain motion, tissue strain, and injury risk reduction. This thesis hypothesises the use of decoupling strategies to reduce the dominant coordinate component of acceleration in order to decrease maximum principal strain values. The dominant component of acceleration, defined as the coordinate component with the highest contribution to the resultant acceleration for each impact, is a targetable design parameter for helmet innovation.

The objective of this thesis was to demonstrate the effect liner strategies to reduce the dominant component of rotational acceleration to decrease maximum principal strain in American football helmets.

# Preface

This dissertation was organized into three parts with multiple sections in each.

**Part I:** Introduction, which presents relevant background and literature review.

**Part II:** Research studies, which includes the research question, study overview and four research articles.

**Part III:** Closing, this includes a global discussion and concluding remarks.

The dissertation is comprised of four original research articles, for which I was the lead author responsible for both conducting and completing the following; 1: the conception and design of the studies; 2: data acquisition, analysis and interpretation (including helmet construction, impact testing and finite element analysis); 3: drafting, revising, and submission of the articles for peer review in scholarly journals.

# List of collaborators:

**Karen Taylor** (PhD student) was involved in all aspects of this thesis, from conception to data collection and analysis. The four research studies and final thesis writing was all completed by the student.

**T. Blaine Hoshizaki** was the thesis supervisor and was involved in a supervisory role in all aspects of the thesis.

**D. G. E. Robertson** was involved as a committee member for this thesis.

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# PART I

## Introduction

The incidence of concussive injury in American football remains high despite mandatory requirement for players to wear protective helmets [Guskiewicz 2000, Delaney 2002, Walsh 2011]. American football helmets were introduced in the 1940's to provide players protection against injuries such as skull fractures and hematomas, both of which are classified as traumatic brain injuries (TBI) [Gurdjian 1968, Gennarelli 1971, Hardy 1994, Cantu 2003]. TBI's characteristically result from short duration, high magnitude impact events causing intracranial pressure difference due to relative motion between the skull and brain [Gurdjian 1966, Viano 1989]. These types of impacts and subsequent TBI are correlated with linear acceleration of the brain [Ward 1980, Hardy 2007]. Additionally, the impacts known to cause these types of injuries are analogous to falling events in which the head impacts an unyielding surface [Bailes 2001, Doorly 2007, Post 2012, Hoshizaki 2013, Post 2015, Beckwith 2018]. For these types of events, the bulk of the impact force is the vertical (normal) component of the ground reaction force, which is transferred directly into the head. The effectiveness of the football helmet at attenuating this type of linear acceleration force and decreasing the risk of TBI is evidenced by the reduced incident rate in the game over the last several decades [Bailes 2001b, Pellman 2003, Casson 2010]. Despite the success of the helmet at reducing TBI, American football is one of the sports with the highest rate of recorded concussive injury [McCrea 2003, Duma 2005, Guskiewicz 2007, Broglio 2009, Rowson 2012]. Concussion typically results from a different impact event than those leading to TBI [Broglio 2012]. It is believed to occur from low magnitude, longer duration impacts generating linear, but predominantly rotational acceleration of the brain [Holbourn 1943, Gennarelli 1982, Viano 2005]. These types of impacts are more analogous with the categorization of collision events, which are representative of long duration, low magnitude accelerations generated by a strong rotational component [Hoshizaki 2013].

Ommaya's early primate studies demonstrated the role rotational acceleration plays in the generation of concussion [Ommaya 1974, Johnston 2015]. Brain tissue has a low shear modulus and therefore rotational brain motion and exposure to strain for longer durations can result tissue shearing and subsequent deformation [Ommaya 1974, Donnelly 1997, Kleiven 2007, Takhounts 2008, Yoganandan 2008]. Unterharnscheidt (1970), Gennarelli (1982) and Adams (1982) supported the argument stating shear deformation caused by rotational acceleration as the predominant mechanism of injury leading to concussion [Meaney 2011]. The continued incidence of concussive injury in football and the link between concussion, rotational acceleration and brain tissue strain has led researchers to investigate the use of helmet liner technology alternatives, such as head-helmet decoupling mechanisms [Alghamdi 2001, Caserta 2011]. However, none of this research has yet to show a correlation between the decoupling technology and reduced injury risk mesaures. The gap in current research regarding the correlation between impact event, mechanism of concussive injury, and effective helmet evaluation offers room for improvement in helmet design to increase protection to players and reduce injury risk [Doorly 2006].

To improve helmet design, with the expectation of reducing injury risk, researchers and designers must determine which measures can effectively predict risk, which design strategies to implement, and how to measure helmet performance in terms of appropriate injury risk [Zhang 2001, Pellman 2006]. Studies have been conducted to show how the mechanical properties and design of helmets affect energy attenuation and accelerations of head-forms due to helmet use [Vetter 1987, Moss 2011, Bartsch 2012, Post 2012a/b, Forero Rueda 2010, Tinard 2012, Lloyd 2014, Darling 2016, Sone 2016]. To augment this research, design for improved injury risk prevention should include investigation into strategies aimed not only at reducing the force transferred into the brain but redirecting a portion of the applied force away from the head altogether. This comes from the concept that reducing overall brain motion will result in decreased brain tissue strain, which has been associated with a reduced risk of injury [Margulies 1992, Zhang 2004, Kleiven 2007]. Decoupling technologies provide reduced friction at the head helmet interface in attempt to "redirect" impact force into the tangential force component around the head rather than the normal force. These decoupling systems allow for the helmet to move independently from the head, reducing the motion of the head and limiting the impact force directed into the brain tissues [Aare 2003, Finan 2008].

The governing objective of this thesis was to test the effectiveness of a helmet decoupling liner strategy to evoke impact force redistribution and/or redirection to decrease the maximum principal strain values for concussive events in professional American football.

## Thesis Objectives and Hypotheses

The objects and hypotheses for each of the studies with in this thesis are:

**Study 1**

*Objective:* To examine the performance of a current certified American football helmet using directional motion (X, Y, Z, Resultant and Dominant coordinate components) and the resulting maximum principal strain for 20 different impact conditions.

*Null hypothesis:* There will be no correlation between the components of linear and rotational acceleration to maximum principal strain.

**Study 2**

*Objective:* To demonstrate the influence of impact force redirection and redistribution strategies on maximum principal strain values using finite element modeling.

*Null hypothesis:* There will be no significant reductions in the magnitude of maximum principal strain resulting from the applied redirection or redistribution strategies.

**Study 3**

*Objective:* To demonstrate the effectiveness of a decoupling strategy to reduce maximum principal strain values in an American football helmet.

*Null hypothesis*: There will be no significant differences in the magnitude of maximum principal strain resulting from the applied helmet liner strategies.

**Study 4**

*Objective*: To test the effectiveness of a decoupling strategy to reduce maximum principal strain for concussive impact reconstructions in professional American football.

*Null hypothesis:* There will be no significant differences in the magnitude of maximum principal strain between the prototype helmet and the commercial helmet for each of the reconstructions of the known concussive impact cases.

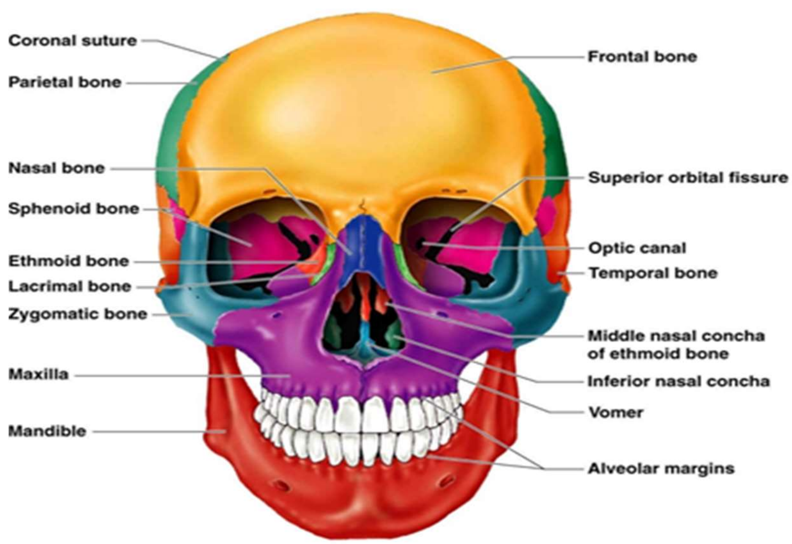
# Chapter 1 - Anatomy

The primary function of the skull is to protect the brain from injury. Upon impact to the head, the characteristics of the skull and the brain determine how impact force is mitigated, the extent of damage caused due to stresses and strains to the brain tissue, and the resulting type of injury. To further protect individuals and reduce brain injury risk, helmets are often worn when the plausibility of head impact is high such as military environments, on high-speed motor vehicles, and sporting events. To improve helmet technology and further reduce impact forces transferred to the brain, an understanding of the anatomy of the human skull and brain are first required. While various characteristics, such as the shape, size, and material properties of both the skull and brain can differ among individuals based on age, health, and biological sex, the gross anatomy is similar for all humans and will be discussed in general in the sections below [Donnelly 1997, Gur 2002, Vappou 2008, Sack 2009].

## 1.1 The Skull

The skull is a complex geometric structure comprised of 22 fused bones [Tortora 1995]. Fourteen form the facial bones and 8 form the cranium, the region of the skull that houses the brain [Gray 1991]. Portions of the cranium, particularly the frontal and occipital bones (base of the skull) are thicker and have large curvatures and even some pointed protuberances (external occipital) [Gray 1991]. As bone is generally more susceptible to fracture by traction than compression these heavier bones may break rather than bend when subject to deforming forces [Moritz 1943]. The bones comprising the vault of the skull, such as the upper temporal and side parietal, are thinner and flatter and are therefore relatively elastic in nature [Moritz 1943]. This indicates that a frontal impact typically results in a fracture that begins at the site of the injury and runs backwards through the base of the skull and an impact against the side or back of the head characteristically results in a fracture, that runs respectively inward through the middle or anteriorly through the posterior fossa [Moritz 1943]. Varying thickness and strength of bone of the skull result in regional differences in the protection and differences in how impact forces are transferred to the brain. Although fractures of the skull do not generally occur during a helmeted impact, the results speak to the potential differences in energy transfer which will be relevant in later sections.

Internally, the surface of the skull contains many protrusions, bony irregularities such as the anterior and middle cranial fossae [Bailes 2001], depressions that accommodate blood vessels, or sinus cavities such as those found in the sphenoid bone [Tortoro 1995]. The bony irregularities can lead to the development of focal injuries if the brain abuts against them. However, a layer of cranial meninges and cerebrospinal fluid (CSF) exits between the brain and skull to provide some protection to the brain from these irregularities on the internal surface. The meninges appear to be much tougher than the interior brain tissue [Fallenstein 1969] while the CSF allows for buoyancy type protection of the brain [Bailes 2001]. This anatomical arrangement allows the brain to have some freedom of movement before it abuts against the cranium [Bailes 2001]. However, even without contact between the brain and skull injury can occur. Movement of the brain inside the skull can cause stress and strains within the brain tissue that result in injury [Gurdjian 1944, Gennarelli 1971 Gurdjian 1978]. To reduce injury risk, any type of movement of the brain within the skull should be minimized.



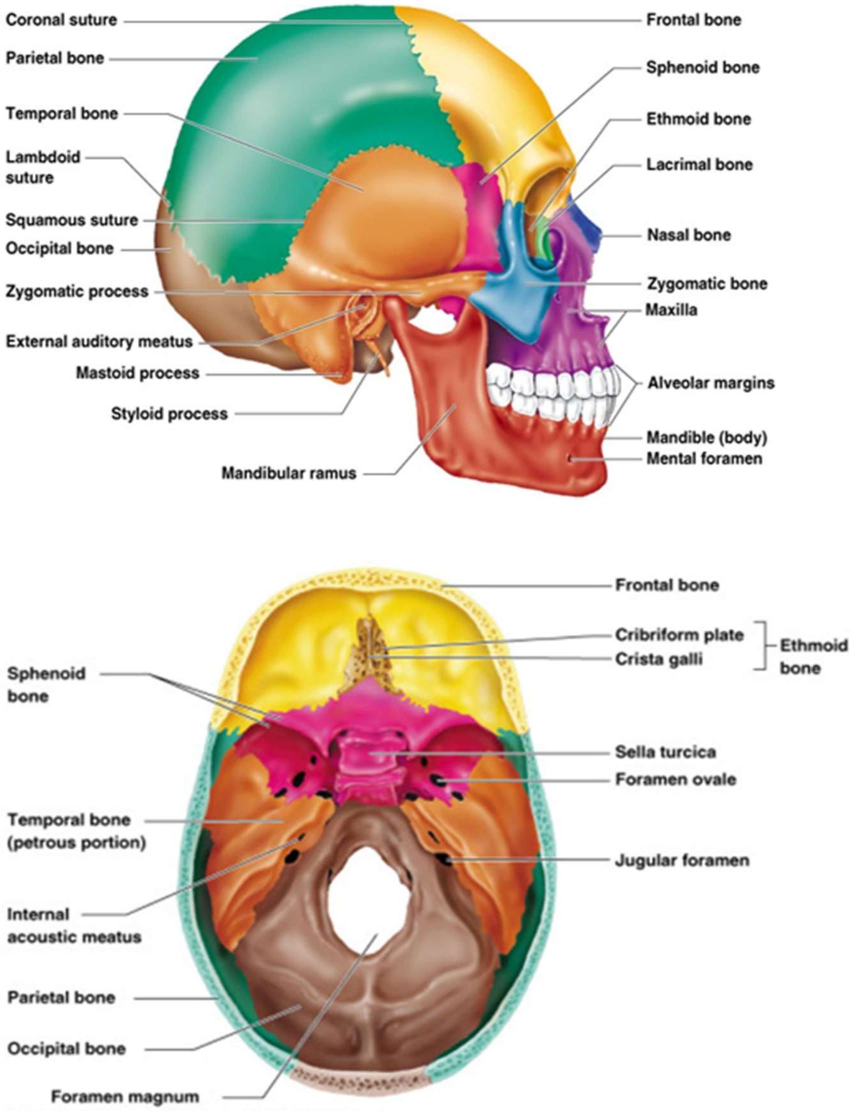



Figure 1: Frontal, Lateral and Internal view of the human skull [Pearson Education 2003]

## 1.2 The Brain

The brain itself is a complicated assembly of gray matter, white matter, blood vessels, membranes, fissures and voids surrounded by, or filled with cerebrospinal fluid [Donnelly 1997]. The outer surface of the cerebrum is comprised of 2-4 mm of grey matter called the cerebral cortex. Deep to the cortex is primarily white matter. Depending on the source cited, the brain, along with the nerves and vasculature, is comprised of three main regions; the brain stem, the cerebellum and the cerebrum [Gray 1991, Tortoro 1995]. Each region contains additional sub-elements and controls different functions of the body [Martini 2017]. The cerebrum can be subdivided into four lobes; the frontal, temporal, parietal and occipital. The regions within these lobes are the; prefrontal cortex, dorsolateral prefrontal area, motor association cortex and primary motor cortex, primary somatosensory cortex, sensory and visual cortex, auditory cortex and auditory association area. Of significance to impact mechanics is the distinction between regions both in function and material properties. Injury to different regions within the brain results in loss of or diminished function controlled by that region [Prange 2002, Elkin 2007, Yoganandan 2008]. Variation in regional material properties influences the type and extent of resulting tissue damage upon impact [Gur 2002, Allen 2005, Elkin 2010, Van Dommelen 2010, Arani 2015].

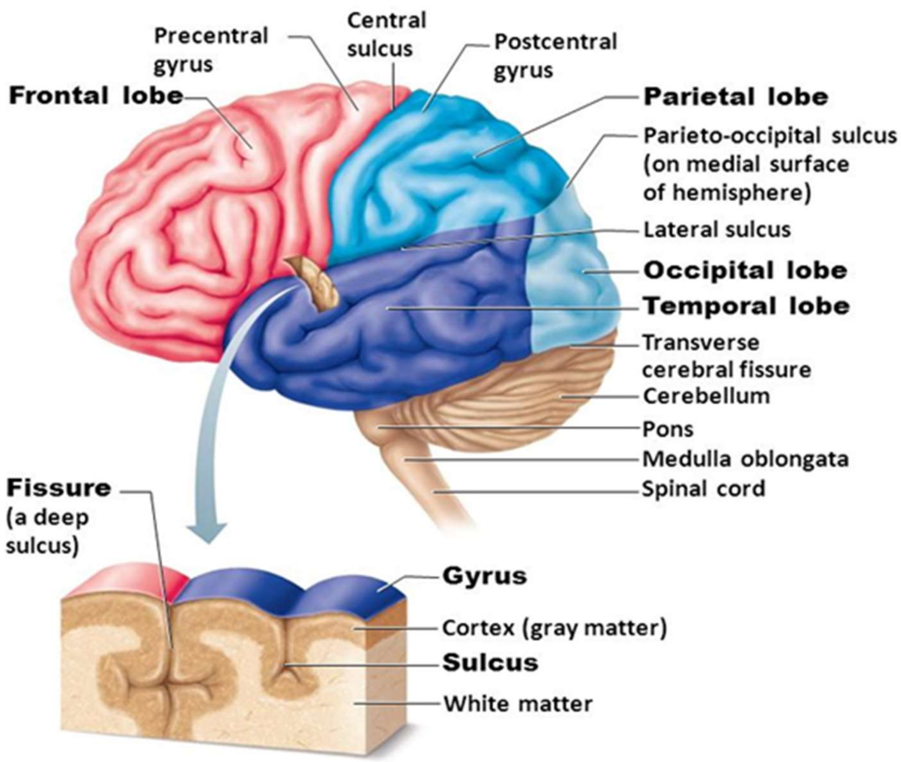



Figure 2: Lobes and Sulci of the cerebrum of the human brain [Martini, 2017]

### 1.3 Mechanical Properties of Brain Tissue

The mechanical response of brain tissue to impact is very complex, primarily resulting from variations in tissue composition. The mechanical properties of the brain influence the resulting deformations and, in turn, injury patterns within the brain [Prange 2002, Christ 2010, Jin 2013]. Generally, researchers agree that the brain as a whole is anisotropic, inhomogeneous and reacts as a soft, non-linear, viscoelastic, incompressible (~78% water, 10% lipids) structure susceptible to tensile and shear forces [Ommaya 1968, Fallenstein 1969, Donnelly 1997, Hrapko 2008]. As mentioned previously, the brain primarily consists of grey and white matter and is surrounded by cerebral spinal fluid. Cerebral spinal fluid (CSF) is an almost incompressible fluid that occupies the subarachnoid space and the ventricular system inside and around the brain and spinal cord. CSF provides a neutral buoyant environment for the brain, offering some mechanical protection through "cushioning" between the brain and skull [Ommaya 1968, Kurtcuoglu 2005]. Grey matter consists mainly of cell bodies, dendrites and unmyelinated axons of neurons and forms the outer layers (2-4mm) of the cerebrum (cerebral cortex) and cerebellum [Budday 2015]. Grey matter has less variation in directional properties than white matter and is considered isotropic, where as white matter is anisotropic [Miller 2002, Prevost 2011, Jin 2013, Budday 2015]. Because of this fibre orientation, grey matter is likely to be more susceptible to tensile stress while white matter is vulnerable to shear stress [Prange 2002, Zhang 2011]. White matter consists mostly of myelinated axons, forms the inner portion of the cerebrum and cerebellum and is responsible for communication between different parts of the brain [Budday 2015, Christ 2010]. The general arrangement of white and grey matter is shown in figure 3.

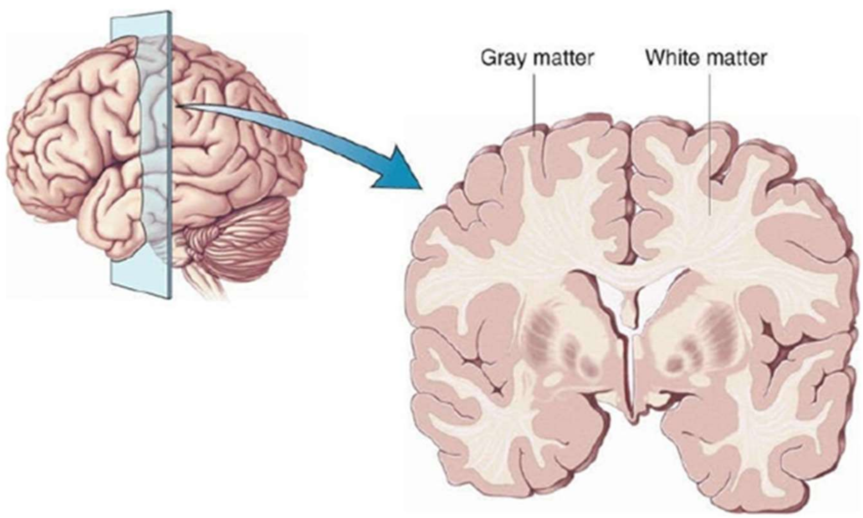



Figure 3: Distribution of gray and white matter within the cerebrum [Martini, 2017]

Brain tissue is viscoelastic, containing both viscous and elastic properties therefore; it possesses characteristics of stress-relaxation, creep, strain-rate sensitivity and hysteresis and is affected by external loading accordingly. Concerning stress-relaxation, the higher the strain rate, the larger peak force/stress is exerted on the tissue and therefore the greater the magnitude of relaxation. Studies have shown that stress relaxation occurs in both white and grey matter [Prange 2002, Prevost 2011]. Budday (2015) noted that white matter had a stress relaxation of 70% while grey matter was 50%. Creep is also a phenomenon discussed in relation to viscoelastic properties and refers to the elongation of tissues when constant force/stress is applied across the tissue. Creep may account for the increased risk of injury for impacts of extended duration, such as impacts with increased compliance. While the force exerted may not be exactly constant for an impact, the peak force acceleration curves have been shown to "flatten" when the duration is extended.

Hysteresis of brain tissue is also observed under various loading rates for animal and human tissue but it not commonly reported on or included in research study data [Galbraith 1993, Horgan 2004]. Strain rate sensitivity refers the different response of tissue to deformation applied at different rates [Prange 2002, Prevost 2011]. Grey and white matter tissue has been shown to have substantially varying responses to strain rate [Budday 2015].

Viscosity refers to the measure of a material's resistance to flow, and, as expected, viscosity of brain tissue varies. Zhang (2011) showed white matter was more viscous than grey matter and the cerebrum more viscous than the cerebellum. However, this may simply be a result of the higher ratio of white matter to grey matter in the cerebrum.

Stiffness is a material property describing the ability of a material to resist deformation and is characterized by the slope of the stress strain curve. The larger the stiffness, the greater force required to cause deformation. Laboratory research often uses stiffness in conjunction with strain rate as the key parameters for detailing the properties of brain tissue. Typically brain tissue is depicted in terms of the elastic (G' – storage modulus) and viscous (G'' – loss modulus) components of shear modulus. There is significant debate surrounding the stiffness of brain tissue with some studies reporting grey matter being substantially stiffer than white matter [Green, 2008, Christ 2010, Zhang 2011], others propose equal stiffness [Feng 2013], while others observe that white matter is stiffer [McCracken 2005, Sack 2009, Van Dommelen 2010, Budday 2015]. Zhang (2011) reported that cerebellum stiffness (greater ratio of white to grey matter) was found to be significantly lower than cerebrum stiffness. Specifically, the average grey matter properties are 1.3 times stiffer than the most compliant region, the corpus callosum. Arani (2015) observed a lower shear modulus in the cortex (grey matter) than inner portion of the brain (white matter) and higher stiffness in white than grey matter. Johnson (2013) demonstrated variations in stiffness within the white matter regions themselves, indicating the corpus callosum and corona radiate, both white matter regions, were significantly stiffer than other white matter regions in healthy volunteers (24–53 yrs).

Clearly, there is controversy surrounding the exact mechanical properties of white matter and grey matter that is still ongoing, even after multiple decades of research. Variation exists in the published information primarily due to the source the data (animal, cadaver, in-vivo, in-vitro) and experimental testing conditions. Interspecies data cannot be directly applied to human response and thus require scaling techniques and cadavers do not provide an accurate physiologic response [Rowson 2012]. In vivo testing can be detrimental to the subject while in vitro testing of brain tissue does not yield realistic results as tissues start to change once removed from the living body [Fallenstein 1969]. Other factors such as age, post-mortem time, temperature, loading configurations (compression, shear, or tension), loading histories (cyclic, stress relaxation, creep) and testing protocol (pre-load, strain rate, and strain load) can also affect the resulting tissue mechanics data [Ommaya 1966, Donnelly 1997, Miller 2002, Prange 2002, Takhounts 2003, Hrapko 2008, van Dommelen 2010, Prevost 2011]. Each of these factors can influence the brain's response to injury and the resulting type and severity of injury that occurs. This variation creates known limitations and can result in difficulty when providing universal protection through helmet design and intervention but should not limit the search for technology advancements as long as the limitations are known and well understood.

## 1.4 Summary

The non-spherical shape and mechanical properties of the human skull create regions of non-uniform protection to the brain. In addition to the shape of the skull, variation in the material properties of the brain influence injury in response to impact. Upon impact to the head, the brain can experience compression, tension and shear forces [Holbourn 1943, Fallenstien 1969, Jin 2013]. The brain is well equipped to handle compressive forces due to its incompressible nature [Ommaya 1968, Fallenstein, 1969]. However, because the brain is inhomogeneous, the different tissue types and fibre orientations will generate different response to stress and strain varying by impact conditions such as impact location and direction. Variation in brain deformation response affects the type and magnitude of injury and must be analyzed during helmet design and evaluation. To account for this, impact testing to evaluate helmet performance for this thesis included various impact velocities, locations, and directions covering the realm of conditions resulting in known concussive injury in professional American football.

# Chapter 2 – Brain Injury

## 2.1 Epidemiology of Brain Injury

Head and brain injury account for 68% of all trauma-related fatalities and 34% of all injuries in the United States [Cantu 2003, Gessel 2007]. The estimated annual cost for hospitalization and rehabilitation of head injured patients is estimated at 60 billion dollars in the U.S. [Stevens 2006, Finkelstein 2006, Faul 2010, Daneshvar 2011]. While head injuries can occur from accidents such as slips, trips, falls and general impacts, head injury in sport is also very common. Each year, an estimated 44 million children and adolescents participate in sport, and 170 million adults participate in physical actively in the US [Danveshvar 2011, NCCDPHP 2006]. In the US, American football has the highest rate of reported brain injury, particularly concussion [Powell 1999, Langlois 2006, Daneshvar 2011, Smith 2018]. The rate of traumatic brain injury has been significantly reduced since the introduction and advancements of the American football helmet [Mueller 1998, Bailes 2001b, Pellman 2003, Casson 2010]. In youth football alone, a 51% reduction in fatal injuries and 65% reduction in cranial fractures had been observed [Hodgson 1985, Viano 2012]. However, the injury rate of concussion per 1000 athletic exposures is 1.55 and 3.02 resulting in 300,000 and 1.2 million high school and collegiate football concussive injuries respectively [Thompson 1987, Turbeville 2003, NCCA 2005, NFSHSA 2005-2006, Ramirez 2006, Daneshvar 2011]. Broglio (2009) reports the incident rate of 7.7% correlating to 130 of the 1700 players experiencing a concussion in professional American football annually. However, the incidence may be higher as 53% of athletes are suspected of not reporting their injuries [Broglio 2010]. The leading mechanism of injury resulting in concussive injury in football is contact with another player (67.7% of all injuries), and contact with the playing surface accounted for an additional 14.9% of injuries [Shankar 2007, Broglio 2009].

It is apparent, that despite the mandatory requirement to wear a certified football helmet, the risk of brain injury remains in all levels of American football. Along with possible changes to rules and the style of play, further developments in helmet technology are necessary in order to see reductions in the rate of brain injury occurring. This requires an understanding of the different types of brain injury.

## 2.2 Brain Injury

A direct or indirect impact to the head resulting in energy transfer to the brain can result in injury [Bailes 2001]. Direct impacts occur when the external force is applied to the head. Indirect head impacts result from an impact to another part of the body where force is transferred to the head, through the neck. Indirect impacts are common in sporting events when a player is impacted by body contact or in motor-vehicle collisions resulting in whiplash type injuries [Holbourn 1943, Moritz 1944, Bandak 1995, Bailes 2001]. In either event, the mechanics involved can be extremely complex and difficult to ascertain. Head impacts can also be classified as penetrating (open) or non-penetrating (closed) injuries and the external force applied can be static or dynamic [Holbourn 1943, Yoganandan 1995]. Static injury or injury sustained at slow rate results in crushing of the skull leading to severe traumatic brain injury or death [Gurdjian 1944, 1975, Yoganandan 1995, Shaw 2002]. For both penetrating and static impacts, the link between impact mechanism and brain damage is apparent and well documented. Accordingly, the primary focus going forward will be impact mechanisms and resulting brain injury due to direct, non-penetrating impacts resulting from a dynamic force.

Research pioneered by Holbourn, Gennarelli and Gurdjian in the 1940's and 1950's has led to an improved level of understanding of head and brain injuries. However, the various mechanisms causing different types of injury are still not yet well defined. Researchers generally agree on one of three outcomes occurs as a result of impact. The first results in a "slapping" of the brain due to bending of the skull leading to translational motion of the brain and collision with the inner surface of the skull. This often results in intracranial hematomas at higher magnitudes and contusions at lower magnitudes [Gennarelli 1971]. Secondly, as the brain and skull do not always move in unison on impact, the brain can lag behind the skull building up intracranial pressure gradients causing local tissue damage at the coup and/or contre-coup sites [Gurdjian 1944, Gurdjian 1978]. Finally, rotation of the brain within the skull is linked with high shear strains and resulting tissue damage [Holbourn 1943, Hardy 1994]. The injuries than can result from head impact are outlined below.

### 2.2.1 Traumatic brain injury (TBI)

Traumatic brain injury (TBI), results from strong and predominantly translational (linear) movement of the brain, often as a result of a high magnitude, short duration (3-10ms) impact [Gennareli 1971, Hardy 1994, Post 2012a]. TBI occurs along a continuum, ranging from subdural hematomas, increasing in severity to contusions, subarachnoid haemorrhages, and epidural hematomas. Contusions result from the brain butting against the irregular surface of the skull and have been shown to occur at maximum principal strain (MPS) values ranging from 0.14-0.53, with von Mises stress (VMS) levels of 5-17 kPa [Doorly 2006]. In a similar motion resulting in contusions, subdural hematomas occur if the force is substantial enough to cause rebounding of the brain within the skull and separation of the dura from tensile forces. Doorly (2007) reported threshold values for subdural hematoma's to be MPS of 0.16 – 0.23 and VMS of 5-8kPa. Epidural hematomas and subarachnoid hemorrhages occur when force exerted to the tissue is high enough that shear strains tear the large blood vessels, including bridging veins, within the brain and small tears within the brain tissue [Gurdjian 1975]. Doorly (2007) reported values of for epidural hematomas of MPS of 0.16 – 0.27 and VMS of 4.8 – 7.9 kPa, and MPS of 0.34 with VMS of 12.2 kPa for subarachnoid hemorrhaging.

### 2.2.2 Diffuse Axonal Injury (DAI)

Diffuse Axonal Injury (DAI), results when high shearing and tensile forces are exerted on the brain severing axonal connections [King 1995]. DAI research has linked the injury to high magnitude, and or rotational events such as punches, and motor-vehicle collisions. It is theorized that shear stress resulting from excessive rotational motion arises from the limited ability of the brain to rotate within the skull, leading to tissue shearing at regions of tissue density change. Published DAI thresholds are 0.041-0.276 for strain and 18,000 rad/$s^2$ for rotational acceleration [Marguilies 1992, King 1995, Ommaya 2002].

### 2.2.3 Mild traumatic brain injury (mTBI)

Mild traumatic brain injury (mTBI), often labeled concussion, is linked to both linear and rotation motion of the brain resulting from low magnitude, long duration impacts [Hardy 1994, Post 2014a]. Recent research has advocated that mTBI occurs along a continuum ranging in severity to include sub-concussive injury, transient concussion and persistent concussion [Hoshizaki 2013, Post 2014b]. Sub-concussive injury can appear asymptomatic or for a very brief period of time. Transient concussion is defined when symptoms resolve within 14 days [McClincy 2006]. Persistent concussive syndrome (PCS), or post-concussion syndrome, is severe injury on the mTBI scale and presents with prolonged symptomology lasting greater than one month [Alves 1993, Post 2014b]. The mTBI continuum leads to threshold values that can range dramatically. Published results indicate a 50% risk of concussive injury due to linear acceleration of 82g, rotational acceleration of ~5900 rad/$s^2$, MPS values of 0.19-0.26, VMS values of 8.4-18 kPa, and strain rates of 48.5-60 $s^{-1}$ [Ommaya 1985, Marguilies 1992, King 2003, Willinger 2003, Zhang 2004, Kleiven 2007].

Concussion, particularly in sport, can result from low magnitude, long duration (10-30ms) impacts resulting in rotational motion of the brain [Holbourn 1943, Gennarelli 1982, Viano 2005, Rousseau 2009, Post 2012a]. Rotational brain motion leads to tissue shearing due to the low shear modulus, inhomogeneity, and anisotropic nature of brain tissue [Ommaya 1974, Prange 2002, Takhounts 2008]. Tissue shearing results in a physiological and metabolic cascade causing detrimental neurological tissue damage and concussive injury [Unterstharnscheidt 1969, Gennarelli 1982, Adams 1982, Meaney 2011]. As a result of the injurious response to shear stress, tissue strain is typically selected as the measure of brain deformation [Margulies 1992, Kleiven 2003, Zhang 2004]. The extent, to which the tissue is sheared, is influenced by the degree of three-dimensional (3D) brain motion due to the geometry and mechanical properties of the brain [Gennarelli 1987, Kleiven 2003, Hrapko 2008, Giordano 2014]. Maximum principal strain (MPS) is a measure of brain tissue stretch, and is in close agreement with anatomic failure testing allowing for comparison to rheological research [Zou 2007, Rousseau 2009, Post 2012a, Hoshizaki 2014]. As MPS is a measure of tissue strain, and tissue strain due to rotational motion of the brain is linked to the cause of concussive injury, it is widely used for the prediction of concussive injury [Viano 2005, Zhang 2006, Casson 2010, McAllister 2011, Patton 2013, Kleiven 2013, Patton 2015, Kleiven 2015]. MPS refers to the largest normal strain observed and will be used as a measure of helmet performance for the purpose of this thesis.

## 2.3 Summary

Providing protection against the risk of various brain injuries through helmet technology is difficult because of the complexity of the brain and the varied mechanical response of brain tissues. Research is required to understand how sport specific impact conditions influence tissue trauma and injury risk to provide detailed information to helmet designers to improve upon current designs for each sport. In addition, modifications to the current American football helmets to address concussive injury risk must be done without compromising the current protective capacity against traumatic brain injury. To address concussive injury risk, this thesis evaluated current helmet performance, then tested and evaluated performance of new design strategies (to be discussed in a later section) to include impacts eliciting linear and rotational response, a 10-30 ms impact duration response, covering a range of impact velocities, locations, and directions which were all plausible in professional American football. Impact conditions are intrinsic to different impact events that influence the resulting brain injury and are discussed in the following section.

## Chapter 3 – Injury Events

Head impact events are best described by the characteristics of the impact such as; mass, velocity, compliance, impact direction, and impact location. Researchers have studied the influence of each of these mechanisms on resulting brain damage [Fahlstedt 1969, Rousseau 2009, Walsh 2011, Ghajari 2011, Karton 2013]. However, the research investigated each independently with limited and narrow variable ranges, which has resulted in insufficient understanding of their effect with regards to brain injury. Until research is published which delineates the influence of end ranges of each characteristic and the interactive effect between them, alternate solutions have been proposed to describe impact events.

### 3.1 Head Impact Events

Impact events can be generalized into four broad categories; falls, punches projectiles and collisions [Hoshizaki 2014]. Each have a unique range for its characteristic mechanisms. For example: the range of mass involved in a punch is less significant than the mass range that can occur for a motor-vehicle collision. In general, the same is true for each variable allowing for the four categories to cover a range of impact scenarios yet be definable under the four groupings. By grouping the mechanism of injury into categories, a link between mechanism and injury emerges. The following mechanisms summarize the range of impact events and the loading conditions used to detail the impact: falls, punches, collisions, and projectiles.

#### 3.1.1 Falls

Falling is one of the most common mechanisms of head and brain injury and is the leading cause of non-fatal hospitalization [O'Riordain 2003, Styrke 2007]. A fall, resulting in a direct impact to the head, commonly involves the head contacting an immovable object such as the ground. The force of the energy transferred to the head from a fall will be dependent on the individual's mass, fall height, impact surface compliance, and location of the impact on the head. Falls are characterized as high mass, high velocity, typically non-compliant impacts, and are associated with focal injuries such as skull fracture and other traumatic brain injuries [Doorly 2007, Hoshizaki 2013]. Manavais (1991) reported skull fracture occurs in 66% of the fatal falls investigated and acute subdural haematoma (ASDH) was found in 78%-85% of the fall cases. Zwimper (1997) and Gennarelli (1982) found similar results indicating ASDH was present in 85% and 72% respectively. Yanagidi (1989) suggested ASDH normally presents as a contre-coup injury and is prevalent when high accelerations occur prior to the impact, such as during a fall, where the head then rapidly decelerates on impact. During deceleration, the cerebral spinal fluid (CSF) moves towards the impact site compromising the dampening protection for the brain on the opposite side that often results in a collision between the brain and skull [Edberg 1963].

O'Riordain (2003) investigated the dynamic response and impact force of four fall cases onto hard surfaces. The first case presented with contralateral contusion and a side hematoma resulting from a linear acceleration of 243-293g and impact force of 7.8-9.3kN. Skull fracture was not observed, which is consistent with published threshold values for both linear acceleration and impact force values for skull fracture (~300g). The other three cases resulted in one traumatic hematoma and two skull fractures, each accompanied with coup haematoma. For each case, the average linear acceleration was above 302g while the impact force was greater than 13kN. Doorly (2006) used Finite element (FE) modelling to reconstruct two reported fall cases. The first case presented as contre-coup subarachnoid haemorrhage from an impact force of 13.5kN, linear acceleration of 366g and rotational acceleration of 36.9 $krad/s^2$ with brain deformation values VMS ~12.75kPa, strain of ~0.315 and strain rate of ~$102s^{-1}$. While the linear acceleration and force values would indicate skull fracture, fracture did not occur. This may be explained by the individual's age (11years old) as the stiffness youth skulls are reported as being 75% that of an adult making them less susceptible to fracture [Yoganandan 1995]. The second case presented as a small hemorrhage and contre-coup contusion from an impact force of 8.07kN, linear acceleration of 236g, and rotational acceleration of 33.8$krad/s^2$ causing deformation values of 8.5kPa, strain of ~0.235, and a strain rate of ~$106s^{-1}$. In this case, the individual sustained less severe TBI (contusion and haemorrhage), which is consistent with thresholds and stress/strain values for contusion.

For the cases above, the individual fell directly on their head without bracing nor were they wearing helmets. If individuals wear a helmet or can brace themselves the risk of injury is reduced. Bracing reduces the velocity of the head upon impact decreasing the force exerted to the head. A helmet increases the contact surface area, spreading the force over a greater area thus reducing the force to the head. The helmet increases impact compliance, increasing the impact duration diminishing the magnitude of the force. Reducing the force transfer explains why helmeted falls in sport, do not often result in skull fracture and ASDH.

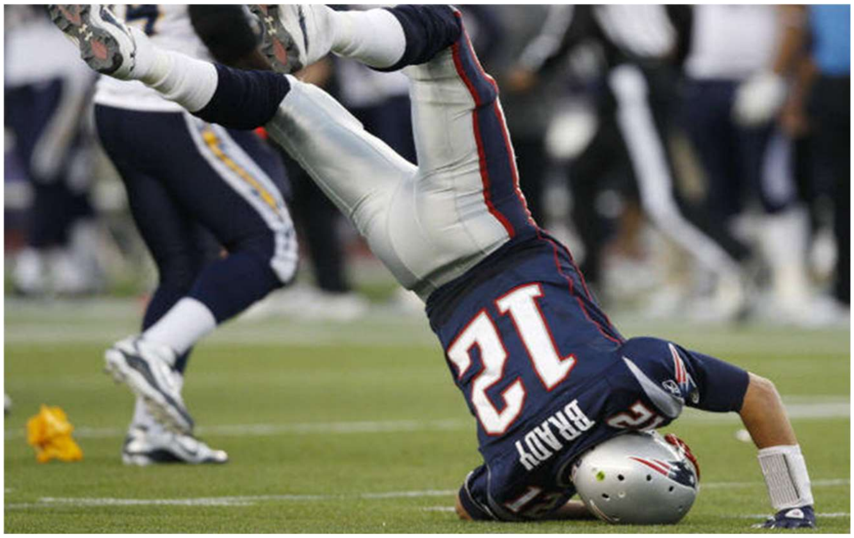


Figure 4: Head to ground impact resulting from a fall in American football [http://spencerspeziale.blogspot.com]

### 3.1.2 Punches

Punches occur during assaults, fights, sporting events, and organized boxing matches. The most common injuries from punches are head and face lacerations (60%) and concussion (16%) [Stojsih 2010, Zaryn 2003]. Punches can induce fatal injuries [Stojsih 2010]. Between 1945 and 1980 over 335 documented boxing fatalities occurred in the United States due to head injuries [Jordan 1990, Jordan 2000]. Approximately 75% of these deaths resulted from subdural hematoma, while 25% were due to traumatic intracranial hemorrhaging [Unterharnscheidt 1970, 1972]. Punch impacts are characterized by high velocity and low mass (arm and hand), when compared to simple falls and collisions [Kendall 2012]. Research involving professional boxers noted the average punch produced during a fight ranged from 9.1 ± 2.1 m/s with a mass 2.9 ± 2 kg [Walilko 2005] and most commonly to the front or side of the head. In a study by Stojsih (2010) during the 2008/2009 world championships, the total number of impacts were recorded and divided as: front impacts (34%), right (21%), left (18%), and top (3%).

Punches in football are rare and when they do occur typically involve the impacted player wearing a helmet. The rate of brain injury resulting from punches in football is negligible [Pelman 2003, Broglio 2009, Ramirez 2006].

Figure 5: Punch to the head in professional boxing [www.punchinthehead.com]

### 3.1.3 Projectiles

Projectile impacts to the head are characterized by short duration, high velocity events occurring over a small contact area where there is little opportunity to dissipate the imparted energy [Adams 1982, Di Maio 1981]. The extreme example is a gunshot wound where skull fracture or fragmentation of the skull results from the explosive effects of the collision [Moritz 1943]. However, skull fracture can also occur from lower velocity impacts if the impacting object is sharp, such as a knife [Rosenberg 1984]. In either case, the risk of injury is related to the energy delivered by the impacting object as well as the object's shape [Viano 2005]. A bullet exerts a lateral and forward thrust into the brain causing tissue rupture as it passes usually resulting in fatal injury. Gunshot wounds do not cause concussive injury as the bullet mass is insufficient to exert the necessary kinetic energy to the head [Gurdjian 1954, Shaw 2002]. However, if the head is struck by a larger object (greater mass), even traveling at a lower speed, the less severe injuries may ensue though the overall force may be same in both conditions. [Shaw 2002, Hoshizaki 2013]. This indicates that for injuries such as DAI and concussion, sufficient kinetic energy is required to trigger intracranial stresses and strains [Shaw 2002].

Projectile impacts are common in sports such as baseball (balls) and ice hockey (pucks) [Hoshizaki 2013]. These impacts produce short duration acceleration pulses (<5ms) resulting from low mass and very high impact velocities (~ 30 m/s for elite ice hockey puck impacts and 42 m/s for Major League Baseball pitches respectively) [Rousseau 2012]. For puck impacts, Hoshizaki (2013) reported peak acceleration values of 105.6 ± 14.6g and 12187 ± 2104 rad/s$^2$ along with MPS values of 0.141 ± 0.009 for concussive injury. If a sporting helmet were not worn during these cases skull fracture could occur due to high forces applied to the skull over a small contact area. Projectile impacts in football are essentially non-existent [Pellman 2003, Ramirez 2006, Clark 2017].


Figure 6: Projectile ball impact to the helmeted head in professional baseball

### 3.1.4 Collisions

Collisions represent a broad and complex mechanism of injury due to the wide range of impact mass, velocity, compliance (protective equipment), and locations. The incidence of traumatic brain injury due to head impacts has decreased in most sports due to the implementation of helmet use. However, there is evidence suggesting that concussion may be a much more common result from collision impact than once thought [Bailes 2001, Delaney 2002, Gessel 2007, Benson 2009]. In contact sport, individual's exposed to head impacts are often wearing helmets and can be impacted by players wearing compliant protective equipment. Both helmets and protective padding increase the impact compliance and therefore impact duration. Thus, sporting collisions are characterized as mid-long duration (10-40ms), possible high velocities, varying mass, and lower peak accelerations.

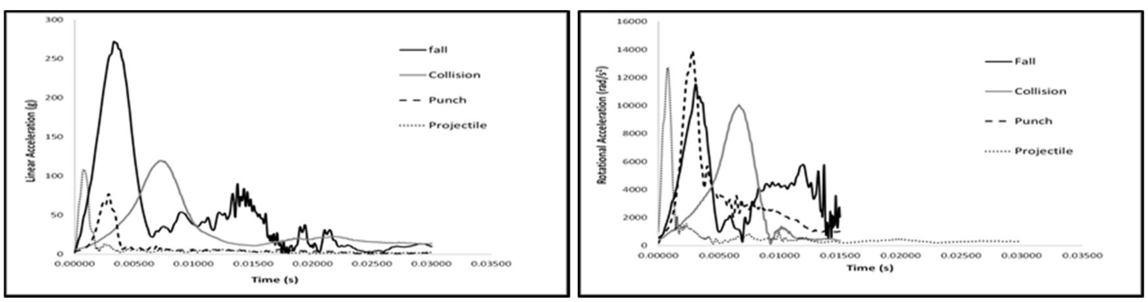
These characteristics are common in corresponding to concussive injury [Hoshizaki 2013]. In a study conducted by Pellman (2006), reviewing 182 cases of reconstructed professional football impacts resulting in concussion; they noted linear acceleration values of 60.7 – 135.4g and 3234 – 11,202 rad/s$^2$.

Figure 7: Helmet-to-Helmet collision in American Football [http://floridapolitics.com/archives/tag/concussions]

### 3.2 Impact Events Leading to Concussive Injury in Football

Although there are numerous ways for players to sustain a concussion, research studies report the majority (76%) in American football occur from collisions with other players, particularly head-to-head [Pellman 2003, Crisco 2004, Broglio 2009, Broglio 2010, Meehan 2010]. In a study utilizing an NFL database of concussive injury; 61% of all reported concussions resulted from head-to-head collision impacts, 16% were from collisions between head-to-shoulder or head-to-arm, and 16% were falls to the ground [Pellman 2003]. Collision events are categorized as resulting in dynamic response curves of low magnitude and long duration, as shown in figure 8 [Hoshizaki 2013]. Additionally, in American football, head collision impact location has been identified as a distinguishing factor in the risk of concussive injury in collision impacts [Crisco 2010]. In American football, both injurious and non-injurious collision impacts are the most common at the front followed by the side, back and top region [Pellman 2003, Greenwald 2008, Broglio 2009, Crisco 2010]. Most concussive injuries in football are from impacts to the front (44.7%) and side (22.3%) of the head [Kerr 2014, Clark 2017]. Additionally, the majority of collision impacts have been shown to occur at oblique angles with 76% of facemask impacts below the centre of gravity of the head and 79% of all helmet shell contacts above the head CG [Pellman 2003a]. In a study by Liao (2016) analyzing collision head impact data from 33 Division I NCAA football players a significant association ($p = 0.017$) was found between head impact location and concussion. The concussive impact group sustained a lower frequency of collision impacts to the front and more to the sides and top of the head. Viano (2005) analyzed reconstructions of concussive collision impact events and then determined tissue strain using FE to show that strain was higher for frontal (0°-45°) and side (45°-90°) impacts compared to rear oblique (90°-135°).

The impact characteristics associated with collision impacts in American football contribute to and are known to be associated risk factors for concussive injury. Current Football helmet design may not be optimal for protection under collision event conditions, particularly head-to-head collisions and will be discussed in the following section. This has resulted in limitations in the protective capacity of the helmet to reduce the risk of concussion for common head-to-head collision impacts in American football and must be address in order to provide adequate protection to players and reduce injury risk.




a: b:

Figure 8: a Linear acceleration for falls, collision (Shoulder), punches and projectiles, b: rotational acceleration for falls, collisions, punches and projectiles

### 3.3 Summary

Head impact events can be classified as falls, collisions, punches, and projectiles [Hoshizaki 2013]. Each can be categorized by their unique set of impact characteristics, which include mass, velocity, location, direction, and compliance [Kendall 2013, Hoshizaki 2013, Post 2012b, Zanetti 2014]. Research has shown most of concussive injuries in American football occur from head to head collisions, which result in lower magnitude, longer duration events often in conjunction with greater rotational acceleration.

To account for concussive injury risk, testing protocols within this thesis will incorporate impacts representative of non-centric and helmet to helmet collision events using maximum principal strain as the evaluation metric.

## Chapter 4 – Helmet Design & Evaluation

The focus of the current football helmet design has been to reduce peak magnitude of resultant acceleration, predominantly linear acceleration to pass helmet performance testing and reduce the rate of traumatic brain injury. As such, the American football helmet has advanced from padded leather to a combined hard outer shell and soft liner, which is well designed to mitigate the typical magnitudes of linear impact forces experienced in football [Meuller 1998, Naunheim 2000, Post 2012a]. However, as evident by the continued high rate of concussion seen in the sport, this current helmet design is not as effective at providing players protection against all injury risk [Bailes 2001, Cantu 2003, McCrea 2004, Zhang 2004, Collins 2007, Crisco 2010, Oukama 2011]. To advance helmet technology and provide innovative technologies aimed at managing brain motion, decreasing tissue strain and further reducing the risk of injury, an understanding of current helmet design is required.

### 4.1 Helmet Design

The shell of modern football helmets is most often fabricated from thermoplastics such as polycarbonate or Acrylonitrile Butadiene Styrene (ABS) [Hoshizaki 2004]. In contrast to crash helmets, that dissipate force through plastic deformation, football helmets must withstand repeated blows without material failure [Oukama 2012, Forero Reuda 2010]. The typical impact energy dissipation by the shell due to deformation is in the range of 10-30% [Hoshizaki 2004, Caserta 2011]. A key function of the shell is to spread impact load over a broader area to reduce peak magnitude of acceleration [Rousseau 2009, Walsh 2011]. In addition to the shell, a thick soft liner is used [Bailes 2001a, Lamb 2009, Post 2012a]. Current liner technologies, typically recoverable foams such as Expanded Polypropylene (EPP) and Vinyl Nitrile (VN) or engineered 3D structures, can mitigate large magnitudes of linear acceleration [Gimble 2008, Lamb 2009, Post 2015]. Material thickness, primarily provided by the helmet liner, helps to reduce peak magnitude of force by increasing the stopping distance "dx". This is more easily seen by the equation $F = mv^2/2d$ where "d" is the distance [Mills 2003, Lamb 2009]. The current offset (helmet liner thickness) for standard American football helmets is 1.5 to 2 inches but is limited by comfort constraints and practicality of use [Newman 1993, Hoshizaki 2004]. Therefore, research efforts looking to reduce acceleration magnitude further, through compliance, have investigated modifying material stiffness or density [Rousseau 2009, Lamb 2009]. Materials that are too soft can "bottom out" becoming fully compressed, limiting the distance allowable before sufficient acceleration force is managed [Post 2011]. This results in delayed but sudden spikes in peak acceleration magnitudes, known to increase the risk of injury [Post 2014a]. Stiff, dense materials reduce the distance for acceleration to occur and lead to rapid elevated peaks of linear and rotational acceleration, indicating the force has exceeded the material's critical threshold before being sufficiently managed [Gilchrist 2003, Song 2005, Post 2011]. Compliance variation, created by varying stiffness and thickness, results in changes to impact duration, which has also been shown to affect maximum principal strain values [Kendall 2016]. High compliance (softer, less dense or thicker materials) elongates impact duration, increasing brain tissue strain [Hoshizaki 2013]. Recent work has suggested impact duration as a factor differentiating between concussive injury and non-injury in collision events such as helmet-to helmet impacts [Rousseau 2015, Kendall 2016]. Given the compliance of liners in typical football helmets, impacts in football occur over 12-15 ms duration for helmet-to-helmet condition, which is an increase from a 3-6ms impact for bare head condition [Pellman 2003, Oukama 2012]. Zhang (2004) indicated if the head is exposed to combined rotational acceleration with impact durations greater than 10 ms, the suggested 50% tolerable concussive injury level was less than the 82 g and 5900 rad/s$^2$ reported for translational acceleration.

Research has shown that helmets are proficient at reducing impact energy below levels known to cause TBI injury, but may have reached their energy attenuating limits possibly due to liner thickness restraints and material property limitations, which remains above those known to cause concussive injury as noted previously [McIntosh 2014]. It is therefore, suggested that alterative means to simple peak resultant acceleration attenuation are required in order to provide protection against lower magnitude impact events resulting in tissue strain causing concussive injury. However, advancements in helmet design must be conducted with the known limitations of shell and liner material properties, (stiffness and thickness) while limiting any increase in impact duration.

## 4.2 Improving Helmet Design to Reduce Dynamic Response and MPS

The continued incidence of concussive injury in football and the link between concussion, rotational acceleration and brain tissue strain have led researchers to investigate the use of helmet liner technologies alternative to traditional systems and which aim to reduce rotational acceleration [Alghamdi 2001, Gimbel 2008, Lamb 2009, Caserta 2011, Johnston 2015]. Aare (2003) was among the first to introduce a novel low-friction helmet consisting of a helmet liner and shell separated by a layer of Teflon allowing free motion between the shell and liner on impact. The helmet was shown to reduce rotational acceleration without increasing linear acceleration [Aare 2003, Finan 2008]. The objective of the design was to mitigate a portion of the normal force typically transferred into head by transferring it to the tangential force around the helmet (redirection through de-coupling). Additionally, Finan (2008) reported findings showing improvements in peak rotational acceleration by reducing the coefficient of friction of two impacting surfaces. While this study reported the friction between the impact surface and helmet, a similar outcome could be predicted between the helmet and head interface. This low friction or "decoupling" concept has been expanded on by commercial brands such as Multi-Directional Impact Protection System (MIPS), Cascade Lacrosse helmets, CCM Rotational Energy Dampening (R.E.D) System, and 6D motorcycle helmets, which have all implemented technologies aimed at providing reduced risk of concussion by attempting to separate the movement of the head and helmet, thus "de-coupling" them. MIPS use a low-friction slip plane layer that allows for 5cm of lateral movement within the helmet. The CCM R.E.D. system uses a series of bladders and impact pods between the liner and shell to help manage a variety of impacts. The Cascade Seven Technology helmet utilizes a liner system that compresses to laterally displace energy from direct high-energy impacts and poron XRD foam to dissipate linear force from low energy impacts. 6D Motorcycle Helmets use an array of 27 elastomeric isolation damper systems, which the company claims provide six degrees of freedom of movement between the head and helmet. While these technologies claim reductions in peak linear and peak rotational acceleration, none have investigated the effect on the corresponding values of brain deformation. Several companies, have in fact, been ordered to cease claims of reduced risk of concussive injury as they can only prove reduced dynamic response, particularly of peak resultant linear acceleration, and not actual reductions in brain deformation (such as maximum principal strain) or correlated reductions in risk of injury. To improve current football helmet designs, technology must be shown to reduce the relative motion of the brain, result in lower magnitudes of dynamic response thus reducing brain tissue strain and the corresponding risk of injury.

### 4.3 Redistribution and Redirection of Impact Force

"Redistribution" and "Redirection" of the impact force are two methods hypothesized to decrease dynamic response and resulting brain tissue strain. Impact force "redistribution" is attained by decreasing the contribution of the dominant coordinate component to the resultant acceleration and forcing re-distribution to one or two of the non-dominant coordinate components. The dominant component of acceleration is defined as the coordinate component with the highest contribution to the resultant acceleration for each impact and is a targetable design parameter for helmet innovation. For example, looking at linear acceleration for a front impact, the dominant component contributing to resultant linear acceleration is the "X" component. Due to the predominant motion of the brain in the frontal plane, minimal linear acceleration occurs in the non-dominant axes' "Y" and "Z" and their contribution to the resultant is small. Using a decoupling mechanism, motion of the brain can be influenced to increase the "Y" and "Z" components upon reduction to the "X" component. This will result in lower magnitude of resultant acceleration. "Redirection" of impact force is analogous to that of a boxer moving away from a glancing blow where a portion of the impact energy is forced into a tangential plane rather than the normal and energy transfer to the brain is reduced. This is also a similar phenomenon to an impact offset from the centre of gravity of the head. Upon collision between two bodies, the force transmitted is related in part to the angle of the transmitted force vector during contact [Rousseau 2009]. Athletes in all sports including football have the capacity to decrease the risk of injury by deflecting a portion of the force transferred to the head. Reid (1975) determined that a football player was able to reduce the force transmitted to the head by moving the head away and deflecting the direct impact. Rousseau (2009) determined a lateral deflection from centre of gravity impact to a bare head-form resulted in reduced dynamic response. The authors showed that impacts to a helmeted head (ice hockey helmet) that was laterally displaced or "deflected" impact typically resulted in lower resultant dynamic response values [Rouseeau 2009]. While neither Reid's or Rousseau's work included maximum principal strain as a measure of injury risk reduction, the work supports the concept of impact force redirection and warrants investigation, in conjunction with the concept of redistribution, as a means of reducing injury risk through helmet liner strategies.

To demonstrate the proposed approach to redistribute or redirect a portion of the impact force, a front centre of gravity impact was taken from a previous pilot study and mathematically altered. In addition to the original impact, seven different scenarios were run to test the redistribution concept and three for redirection. For the redistribution cases, a reduction (10%) was applied to the dominant coordinate component and the magnitude of that reduction was either re-assigned to one of the other non-dominant axes, or evenly between the two axes. This was conducted for linear acceleration components, rotational acceleration components and a combination of both. In the re-direction scenarios, a ten percent reduction was applied to the dominant coordinate component of linear acceleration, rotational acceleration or both. The resulting data (X, Y, Z linear and X, Y, Z rotational curves) were fed into the finite element brain model and reductions in maximum principal strain values (indicator of concussive injury risk) were determined. The results warranted further explanation and are examined in Study two of this thesis. The theoretical demonstration of the effectiveness of energy redistribution and redirection has shown that force re-direction (loss of energy between application to the helmet and transfer to the head) does result in lower values of MPS in the brain. However, somewhat non-intuitively, redistribution of energy does not always produce the same result in all cases, even when the net effect is reduced magnitude of the resultant acceleration vector. Due to the geometry of the head and brain, sensitivity to directional motion may increase MPS values even with a lower resultant value. Studies have shown that the mechanical response of brain tissue is affected to a larger degree due to changes in the magnitude and orientation of the strain [Giordano 2014]. McIntosh (2014) revealed concussive impacts were more likely to occur with higher angular kinematics in the coronal and transverse plane. Weaver (2012) used computational FE modeling of 14 impact conditions varying location and direction to depict differences in strain response of the brain model with different directions of rotation even when the input magnitude was controlled. They reported that increased rotation in the Z-axis (translational plane) may be the most injurious coordinate component of dynamic response for injury to the cerebrum [Weaver 2012, Johnston 2015]. The resilience of the head/brain system varies with direction and redistribution changes the direction of the resultant vector. Forcing motion into one axis may result in increased resultant value. This can occur if the magnitude of the reduction in the dominant axis is greater than the difference between the dominant and the non-dominant axis. This is particularly important at non-centric impact sites when there may be no clear dominant coordinate vector and the magnitudes are similar. Decoupling strategies used to redistribute and redirect applied impact energy offer potential to reduce the risk of concussive injury, but must be well understood as they may also increase it. Further exploration of the complex physics involved and experimental study can lead to design guidelines for the use of engineered helmet liner strategies to reduce brain tissue strain associated with concussive injury risk.

## 4.4 Injury Risk Predictors used in Helmet Performance

Experimental methods for testing the effectiveness of football helmets were developed by Roberts (1964) who gave results in terms of head accelerations, by Kindt (1973) who evaluated helmet protection from signal traces of impacts on helmets, and by Berger (1976) who related their results to the Severity Index (SI). In 1973, the National Operating Committee on Standards for Athletic Equipment (NOCSAE) established protocols for impact performance of football helmets to improve helmet design and further reduce the rate of traumatic brain injuries [Bishop 1984, Pellman 2003, Cantu 2003]. Between 1978 and 1980, standardized helmet use resulted in a 51% reduction in fatal head injuries and a 65% reduction in cranial fractures [Hodgson 1980]. Mertz (1996) reported the introduction of helmeted standards reduced the rate of traumatic head injury from 55 to 12% at a professional level. Hodgson (1985) reported a reduction in SI between pre and post-standardized helmet designs from 1450 to 1064. However, these values are still above the published thresholds of GSI=300, which currently identifies the nominal tolerance of concussive injury risk [Pellman 2003, Duma 2005]. The Gadd Severity Index (GSI = 1200) is currently used to measure performance in a pass/fail capacity and is a linear acceleration and impact duration-based metric [Hoshizaki 2004, Zhang 2004]. While GSI is in common use, it is limited as it does not utilize dynamic response directionality (brain motion) or account for impact location [Hoshizaki 2004]. Additionally, the Wayne State University Tolerance Curve (WSUTC), from which GSI was based, resulted from peak linear acceleration magnitudes determined from animal concussion tests and cadaveric skull fractures (14.1-43.5 J, 2.5-17 kN, or ~275–300 g) [Hodgson 1966, Yoganandan 1995]. Both animal and cadaveric testing requires speculative scaling to live human response and therefore leaves room for error in human injury prediction [Rowson 2012, McIntosh 2014]. When linear acceleration is used, the pass/fail criterion of 275-300g is well above published thresholds of 65.1 and 88.5g corresponding to 50 and 75% likelihood of concussion [McIntosh 2014]. Additionally, there is scrutiny regarding the link between linear acceleration and concussion and research supports the notion that linear acceleration can be a poor predictor of injury risk [Greenwald 2008, Viano 2005, Oukama 2011]. In attempts to improve helmet performance evaluation and injury risk prediction for concussion, researchers have begun to investigate alternative metrics and combinations of metrics, as a single metric such as linear acceleration is inadequate to accurately assess concussive injury risk [Newman 2005, Greenwald 2008, Rowson 2012]. Greenwald (2008) used a professional football concussion database to create a single biomechanical measure, such as linear acceleration, is not sensitive enough for determining concussion risk and composite variables are better for injury prediction [Greenwald 2008, Kendall 2013]. Broglio (2010) showed that combined peak linear acceleration, peak rotational acceleration, and impact location was required to best predict 13 football concussive case studies. The study did not detail the exact correlation and the interactions between the impact risk factors remain unclear [Giza 2001, Shaw 2002, Kleiven 2007]. Other research suggests kinematic measures, such as peak resultant linear and peak resultant 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 [Gurdjian 1966, Forero Rueda 2010]. This is supported using an example, in figure 9, where both theoretical curves (red and blue) have identical resultant linear acceleration peaks of 300 g and therefore would indicate identical risk of injury. Research has shown, that the variance in the characteristics of these impact curves; curve shape, duration, slope etc. would indicate injury risk differences for these two impacts [Post 2012c, Post 2012d]. In fact, two identical resultant acceleration curves can result in different injury risk as shown in figure 10. The hypothetical resultant curves of acceleration are the same yet the direction of the impact, which is not reflected in the resultant, is different. Figure 10a represents an impact to the front of the head (predominant X response; red), while figure 10b is a side impact (+Y response; green). Studies have shown difference in injury risk between frontal and lateral impact and these two hypothetical curves show possibility for error in injury risk prediction [Newman 2000, Yoganandan 2004, Zhang 2004]. These figures help to portray the potential inaccuracy of using exclusively dynamic response as a predictor of injury risk and measurement of helmet performance.

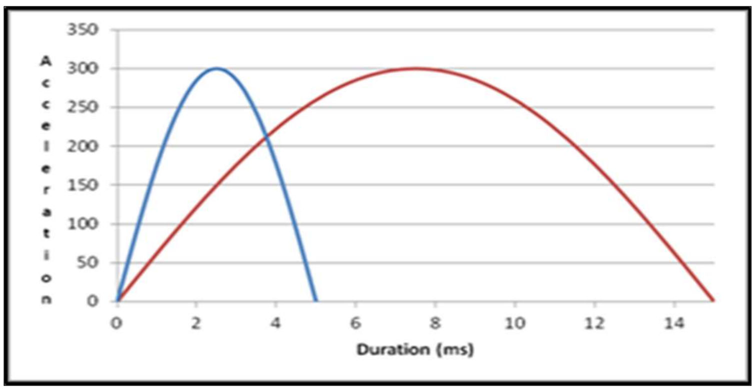


Figure 9: Theoretical acceleration curves with identical peak magnitudes.

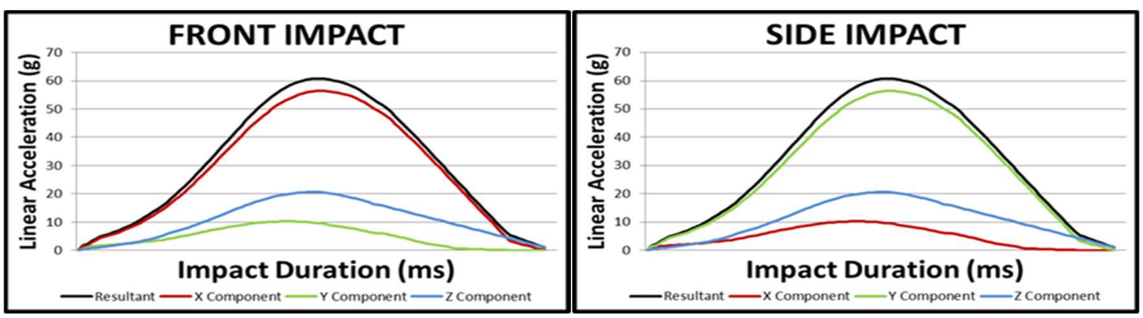


Figure 10: Theoretical acceleration curves with different directional inputs.

Researchers, such as Zhang (2003) and Willinger (2003), have proposed that to study brain injury mechanics, focus should be on analyzing brain tissue reaction and motion response variables rather than input variables, such as linear and rotational acceleration [Kleiven 2007, Takhounts 2008, Rowson 2012]. Finite element modeling (FEM) is a tool used to help establish how linear and rotational accelerations influence the stresses and strains imparted to the brain tissue from football and other sporting helmeted impacts [Hoshizaki 2013]. As such, finite element model outputs provide an opportunity to assess the ability of helmets to manage the risk of brain injury [Hoshizaki 2013, Zhang 2003, Kleiven 2007]. One commonly used output of an FE model is maximum principal strain (MPS). MPS is a measure of brain tissue stretch, and is in close agreement with anatomic failure testing allowing for comparison to rheological research [Zou 2007, Rousseau 2009, Post 2012a, Hoshizaki 2014]. As MPS is a measure of tissue strain, and tissue strain due to rotational motion of the brain is linked to the cause of concussive injury, it is widely used for the prediction of concussive injury [Viano 2005, Zhang 2006, Casson 2010, McAllister 2011, Patton 2013, Kleiven 2013, Patton 2015, Kleiven 2015]. MPS refers to the largest normal strain observed. Bain (2000) studied the optic nerve of an adult guinea pig, reported a strain range 13 - 34% for general brain tissue injury. Zhang (2004) found levels of 14, 19 and 24% corresponding with 25, 50 and 80% probability of concussive injury. Galbraith (1993) conducted experiments on squid giant axons and determined that axons subjected to elongation greater than 20% never fully recovered and that elongation above 25% caused structural failure of the axon resulting in TBI. Clearly great variation exists in the strains observed both though animal/cadaver experiments and finite element models. The possible variation in MPS values across different testing and laboratory environments make MPS a non-optimal tool for a standalone measure of helmet performance. Additionally, while useful as a tool for measuring brain trauma loading, MPS is unable to provide specific information linking impact event characteristics, global kinematics, and resulting injury [Giordano 2014]. It is therefore insufficient, alone, to provide prevalent information to optimize helmet design and technology. To adequately evaluate helmet innovation effectiveness in terms of reducing the risk of injury such as concussion, metrics more closely associated with concussive injury risk should be utilized. It is therefore beneficial to incorporate a method of describing the characteristics of brain motion that can link the mechanism of injury, brain tissue strain and resulting brain trauma.

### 4.5 Coordinate Components of Three-Dimensional Brain Motion

An alternative approach to measure brain motion and evaluate helmet performance in terms of injury protection is proposed here. Current laboratory reconstruction procedures are used to assess three-dimensional dynamics of head motion subjected to an impact [Viano 1989, Walsh 2011]. The resulting brain motion is measured using head based orthogonal coordinate components; X, Y, Z of linear acceleration for the centre of gravity of the brain, which in turn, are used to calculate the X, Y, Z components of rotational acceleration [Viano 1989, Kendall 2013]. Currently, the six linear and rotational acceleration curves are used two-fold; first, to determine resultant linear and rotational acceleration and secondly, as inputs to the finite element model. Since the coordinate components of linear and rotational acceleration are direct measurements of brain motion, it is hypothesized they could be utilized to evaluate helmet effectiveness to reduce global kinematics and in turn decrease maximum principal strain. McAllister (2011) used this approach when using coordinate components of dynamic response, MPS, strain rate, and pre and post imaging to analyze five collegiate and four high school concussed football player. While the research did not link the individual coordinate components to strain, the author stated lower head acceleration values correlated with reductions in brain tissue strain. Duma (2005) was among the first to report acceleration data in terms of coordinate response in a study analyzing 3312 head impacts in collegiate football players. The results did not differentiate between injury and non-injury impacts, nor report maximum principal strain or Z-axis rotational acceleration, and only estimated rotational acceleration. However, the research noted a greater number of larger Y-axis rotational accelerations than X-axis rotational accelerations indicating the importance of component analysis [Duma 2005]. Rowson (2009) improved upon the methodology introduced by Duma and reported data on 1712 collegiate football impacts. That study indicated variation in average magnitudes of the components of both linear and rotational acceleration and noted the largest and lowest peak linear acceleration occurred in the Z-axis and Y-axis respectively. The work also reported larger peak rotational accelerations (>3000 rad/s$^2$) being more common about the Z-Axis, with low values (<1000 rad/s$^2$) about the X and Y Axes. In a study by Kendall (2013) comparing reconstruction methodologies for concussive injury cases, the authors isolated the component coordinates of dynamic response and noted the Z component of rotational acceleration was consistently higher for larger MPS values. MacIntosh (2014) incorporated component analysis of dynamic response as a method for predicting concussive verses non-injury impact events and conducted a study using video analysis of Australian Rules football and MADYMO reconstructions. The results indicated that in addition to the resultant there were significant differences for the magnitudes of Z component of linear acceleration and both the X and Z components of rotational acceleration between concussive and non-injury impacts [McIntosh 2014]. These studies identify the susceptibility of the brain to directional motion and the importance of understanding overall brain motion in relation to brain injury. While these studies utilized coordinate components to analyze data in various means, it is hypothesized they could be beneficial when used in terms of injury risk prediction and helmet protection technology, which warrants further investigation.

To demonstrate the approach to use coordinate components to influence helmet design to modify global brain response, previously collected pilot data was used to compare bare and helmeted impacts, using a NOCSAE certified American football helmet over 20 impact conditions. The results (shown in Appendix 3a/3b) depicted variations in the percent contribution of the coordinate components between the bare and helmeted impacts. This indicated the effect of the current helmet design on three-dimensional brain motion. For example, when viewing the "Side Centre of Gravity (CG)" impact for the bare condition the dominant coordinate component of rotational acceleration is "X" at 97.2% of the peak resultant followed by "Z" (34.3%) and "Y" (4.6%). For the helmeted condition, the dominant coordinate component remains "X" followed by "Z" and "Y" but there is a shift in the percentage of resultant: 83.9%, 63.6% and 22.6% respectively. To compare the relationship between each coordinate component and maximum principal strain bivariate correlations were analyzed (shown in Appendix 4). Different correlations arise when analyzing both the bare and head-form conditions. This demonstrated the influence of the helmet on brain motion and the effect this change in motion can have on maximum principal strain values. Additionally, this supports the notion that helmet technology can be used to manage brain motion to reduce maximum principal strain and the risk of concussive injury.

### 4.6 Summary

It is hypothesized within this thesis, decoupling strategies can influence the motion of the brain, and decrease measures associated with concussive injury risk, such as maximum principal strain. Helmet design, using current decoupling strategies, do not account for all aspects of the helmet/brain system and interactions involved. The existing literature and pilot studies conducted for this thesis support the notion that impact redirection and redistribution are reasonable strategies to reduce motion of the brain and to decrease tissue strain. However, modifying brain motion requires a thorough review because the brain is directionally sensitive to strain and resulting tissue strain values may increase with changes in directional motion. Such understanding is necessary to avoid unintentionally reducing the protection a helmet provides. When properly investigated and understood, decoupling helmet liner strategies should be capable of decreasing brain tissue trauma. This thesis set out to demonstrate a comprehensive approach to the problem of injury risk reduction in American football through helmet innovation using decoupling strategies to reduce maximum principal strain. A helmet developed from the information gained during the comprehensive testing was evaluated using known concussive head-to-head impact events in American football and shown effective in reducing maximum principal strain values.

# PART II

## Introduction

Helmet liners are important for brain injury protection and can be further improved to reduce factors associated with concussive injury, such as maximum principal strain. Liner strategies such as decoupling materials, alone and in conjunction with impact force attenuating structures, should be evaluated for their ability to manage brain motion, and thus reduce the corresponding tissue strain. To accomplish this, a study was conducted to analyze the effect of a standard American football helmet on directional components (X, Y, Z and R) of brain dynamic response and corresponding maximum principal strain for 20 impact conditions and a range of test velocities. Subsequently, a second study was conducted to mathematically demonstrate impact force redirection and redistribution for helmeted head impacts to determine the optimum strategy to reduce MPS under each of the 20 impact conditions from Study One. Study Three evaluated the effectiveness of liner strategies to influence the motion of the brain under the same impact conditions as Study One. The final study produced and assessed the effectiveness of a helmet designed to decouple the head and helmet and reduce the motion of the brain and decrease brain tissue strain. The helmet, designed using the information gained in Study One, Two, and Three was evaluated against certified manufactured helmets for its ability to reduce dynamic response and maximum principal strain in reconstructions of known concussive impacts in American football. While the work within was aimed for use in football helmets, similar future research could lead to improvements in helmets used for various multi-impact sports such as hockey, lacrosse, and boxing. Additionally, the information gained from this thesis can be used to develop more comprehensive helmet testing standards aimed to incorporate further injury risk prevention.

## Overview

**Study One: The relationship between directional components of dynamic response and maximum principal strain for impacts to an American football helmet**

The purpose of this study was to evaluate the influence of the American football helmet on directional components of dynamic response and the magnitude of maximum principal strain. This information can be used in conjunction with the new proposed NOCSAE standard testing to aid designers in developing helmets to further reduce brain tissue strain values.

**Study Two: The influence of head impact force redistribution and redirection on Maximum Principal Strain**

The purpose of this study was to determine the influence on brain tissue strain due to three redistribution and seven redirection strategies by mathematically altering 20 different impact conditions and analyzing the results collapsed across impact condition, impact location, and impact direction. This information can be used to aid designers in developing helmet technology to further reduce brain tissue strain values.

**Study Three: Decoupling liner strategies to redirect impact energy and reduce maximum principal strain using novel design principles in American football helmets.**

The purpose of this study was to demonstrate how the implementation of a novel design principle influenced reductions in maximum principal strain values in comparison to an existing commercially available helmet and helmets designed using current design principals and materials aimed for linear and rotational acceleration management.

**Study Four: The effect of a novel impact management strategy on maximum principal strain for reconstructions of American football concussive events.**

The purpose of this study was to test the effectiveness of a helmet designed to decouple the helmet head interface and reduce maximum principal strain under concussive impact reconstructions in American football. The prototype helmet was evaluated against manufactured helmets for its ability to reduce maximum principal strain.

## Study 1

### The relationship between directional components of dynamic response and maximum principal strain for impacts to an American football helmet

**Abstract**

Impact parameters used to design the American football helmet and the parameters associated with the mechanism of concussive injury are not consistent. Head impacts resulting in concussive injury in football are characterized as low magnitude events creating rotational motion of the head that generates brain tissue strain. The extent of tissue strain influences the resulting severity of injury. Helmet technology aimed to decrease brain tissue strain by reducing the extent of brain motion could help to reduce injury risk. Current helmet performance and evaluation measures, commonly peak resultant of linear and rotational acceleration, do not define brain motion and therefore do not provide sufficient information for this type of improvement. This study was conducted to determine if coordinate components (X, Y, and Z) of linear and rotational acceleration would correlate with maximum principal strain (MPS), a common measure of brain injury risk. Coordinate components define directional motion of the head and offer a more targetable design parameter than peak resultant. In addition to coordinate components, this study introduces the dominant component, defined as the coordinate component with the highest contribution to the resultant acceleration, for additional evaluation.

The results identify the relationship between the X, Y, and Z coordinate components of acceleration and MPS is location and direction dependent, indicating the significance of directional brain motion for helmet testing and design. The study demonstrates the strong relationship between the peak resultant and dominant components of acceleration to MPS. Identifying the relationship between dominant acceleration and MPS reveals the potential use of this targetable design metric to improve future helmet innovation aimed at reducing tissue strain.

**Null Hypothesis**: There will be no difference in the correlation between the components (X, Y, Z, Resultant, and Dominant) of dynamic response (linear and rotational acceleration) to maximum coordinate components.

**Key words**: Dominant coordinate component, brain injury, maximum principal strain

## Introduction

Despite the mandatory use of certified helmets, American football is the sport with the highest incident rate of players with recorded concussive injury.[1-10] Research has reported concussion is associated with rotational motion leading to brain tissue strain and shearing.[11-16] The extent of strain is the result of the degree of three-dimensional (3D) motion of the head, which in turn is influenced by brain geometry and mechanical properties, as well as the impact location and direction.[7,17-22] Additionally, head impact location and direction have been identified as factors distinguishing between concussive and non-concussive injury, in American football.[7,19,23-26] Despite the known link between concussive injury and rotational motion of the head, typical current American football helmets have a consistent, non-location/direction specific liner designed to reduce linear acceleration of the head.[22] This results from the current American football helmet standard, which uses a drop test and resultant linear acceleration as the performance metric due to the link with traumatic brain injury.[27] A new standard has been proposed and comes in to effect May 2019. This standard includes additional impact testing and uses resultant rotational acceleration to help address collision type events.[28] However, neither the resultant of linear or rotational acceleration is capable of properly defining the directional motion of the head. It is proposed that helmet performance evaluation should incorporate the dominant and/or the coordinate components (X, Y, and Z) of acceleration to analyze head motion upon impact. The components of acceleration define directional motion of the head and may offer a more sensitive design parameter than resultant for improvement of helmet development. In addition, the dominant component of acceleration, defined as the coordinate component with the highest contribution to the resultant acceleration for each impact, is location and direction sensitive. This parameter offers valuable directional information and is a targetable design parameter for helmet innovation aimed at reducing brain motion and tissue strain.

While not specific to helmet design innovation, researchers have investigated coordinate components of dynamic response for concussive impact mechanics and the relationship between motion of the head and injury. Duma (2005) was among the first to utilize coordinate components of acceleration response analyzing 3312 head impacts in collegiate football players wearing helmets equipped with the HIT System.[3] The authors noted the acceleration magnitudes around the Y-axis were almost double those around the X-axis, indicating the directional sensitivity of the brain. McIntosh (2014) incorporated component analysis of dynamic response as a novel method for predicting concussive versus non-injury impact events for a study using video analysis and MADYMO reconstructions of Australian Rules football.[29] The results indicated that in addition to the resultant, significant differences in the magnitudes of coordinate components of linear acceleration and rotational acceleration occurred between concussive and non-injury impacts.29 Duma's results, indicating the directional sensitivity of the brain, and McIntosh's data, identifying the relationship between the coordinate components of acceleration and injury, support using coordinate components as evaluation and design metrics for injury risk reduction through helmet innovation.[28,29]

The purpose of this research was to examine the relationship between coordinate components (X, Y, Z, and Dominant) of dynamic response (linear and rotational acceleration) and the magnitude of maximum principal strain (MPS) for helmeted impacts using a current certified American football helmet. Brain tissue strain has been previously used as an indicator of brain tissue injury and can be predicted using kinematic response data and finite element (FE) models.[21,22,30-33] Research has well established the relationship between resultant dynamic response and brain tissue deformation.[33-37] However, improvement to current helmet design requires comprehensive measurement and analysis of changes to three-dimensional head motion and the influence to brain tissue strain values when the helmeted head is impacted at different locations and directions. This study was conducted to compare the correlations of coordinate components of acceleration and resultant to maximum principal strain.

## Methodology

### *Test Procedure*

The dynamic response data for different impact conditions were determined by impacting a helmeted male 50th percentile Hybrid III head-form with a pneumatic linear impactor as shown in figure 1. A total of eighty impact conditions were conducted; 5 locations, 4 directions and 4 velocities. The locations included front, front boss, side, rear boss and rear as outlined in figure 2. The directions included centre of gravity, positive azimuth, negative azimuth, and positive elevation detailed in figure 3. The test velocities were 2.0, 4.0, 6.0 and 8.0 m/s, chosen to represent reported helmet-to-helmet impact velocities in amateur and professional football.[28,38] Each impact condition was conducted three times for reliability purposes for a total of 240 (80 *x* 3) impact trials.

A certified American football helmet comprising of a polycarbonate shell and layered liner system made up of three-dimensional TPU cushioning and vinyl nitrile covered foam. Each helmet including mask (Carbon steel, meeting NOCSAE standard 12-2014) and chin strap had a mass of 2.0 kg (±0.005 kg).[28] The helmets were fitted to the headform using the manufacturer's guidelines. One helmet sample per testing velocity was used for a total of 60 impacts per helmet (3 impacts *x* 5 locations *x* 4 directions). Helmets were replaced after a total of 60 impacts to avoid potential liner degradation and compromised impact performance.[39]

The dominant coordinate component of linear and rotational accelerations was determined from the time history curves. The dominant component of linear (LA) and rotational acceleration (RA) at each impact site are as presented in table 1.

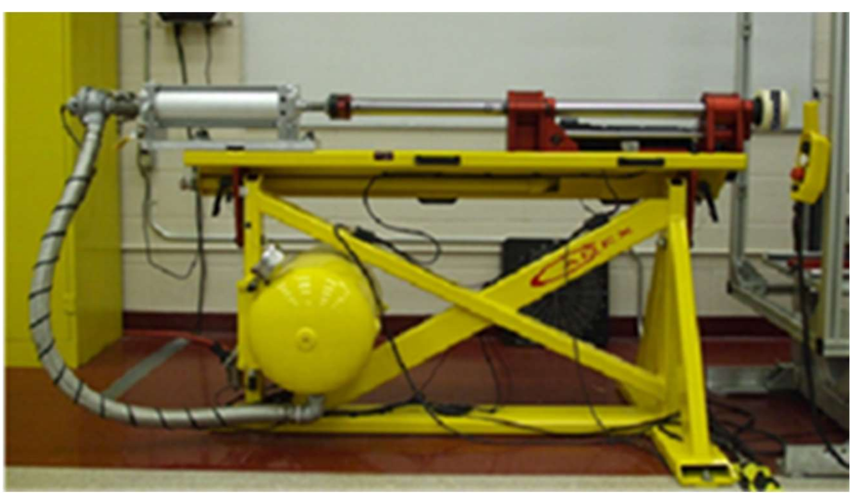
| **Location** | **LA** | **RA** |
|---|---|---|
| Front CG | X | Y |
| Front NA | X | Z |
| Front PA | X | Z |
| Front PE | X | Y |
| Front Boss CG | Y | X |
| Front Boss NA | X | Y |
| Front Boss PA | Y | Z |
| Front Boss PE | Y | Z |
| Side CG | Y | X |
| Side NA | Y | X |
| Side PA | Y | X |
| Side PE | Y | X |
| Rear Boss CG | Y | Z |
| Rear Boss NA | Y | Z |
| Rear Boss PA | X | Y |
| Rear Boss PE | Y | Z |
| Rear CG | X | Y |
| Rear NA | X | Z |
| Rear PA | X | Z |
| Rear PE | X | Y |

Study 1 - Table 1: Dominant coordinate components for twenty impact conditions; LA, Linear acceleration; RA, Rotational acceleration.

The dynamic response data files, each containing the X, Y, Z linear and X, Y, Z rotational acceleration time curves, were input into the finite element brain model to calculate maximum principal strain.

***Linear Impactor***

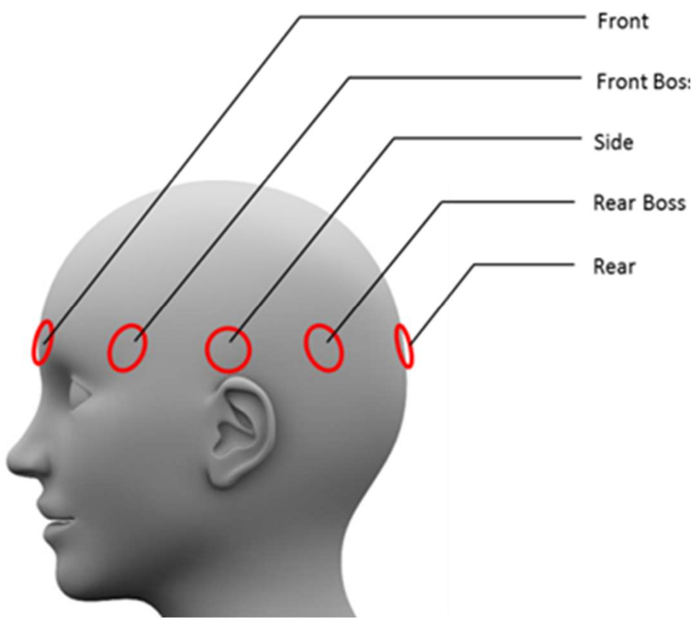
A linear impact system was used to simulate helmet-to-helmet collisions (figure 1). The apparatus consisted of a support frame, a compressed air tank, piston, impactor ram, and a mobile table attached to rails on the support frame. The effective striking mass of an impacting player (13.1 kg) was approximated as the mass of the body parts contacting the struck players head.[34,36,37,40-43] A Modular Elastomer Programmer (MEP) impactor cap attached to the end of the impactor ram to elicit a dynamic response curve (shape and duration) similar to those reported in helmet-to-helmet collisions.[44-47] Additionally, the MEP impactor decreases testing variation seen when using actual helmet-to-helmet impacts and as such is used in helmet testing and design research.[28,36,37,48] The impacting ram (13.2kg) was propelled horizontally using pressurized air to impact the head-form and unbiased neck fixed to the sliding table.[49] The 50th percentile adult male Hybrid III head-form was instrumented according to Padgaonkar's (1975) orthogonal 3-2-2-2 linear accelerometer array protocol.[50] Upon impact, the assembly (head-form, unbiased neck, and table) was free to move along the rails until reaching stop pads at the end of the support table (0.65 m).



Study 1 - Figure 1: Linear Impactor

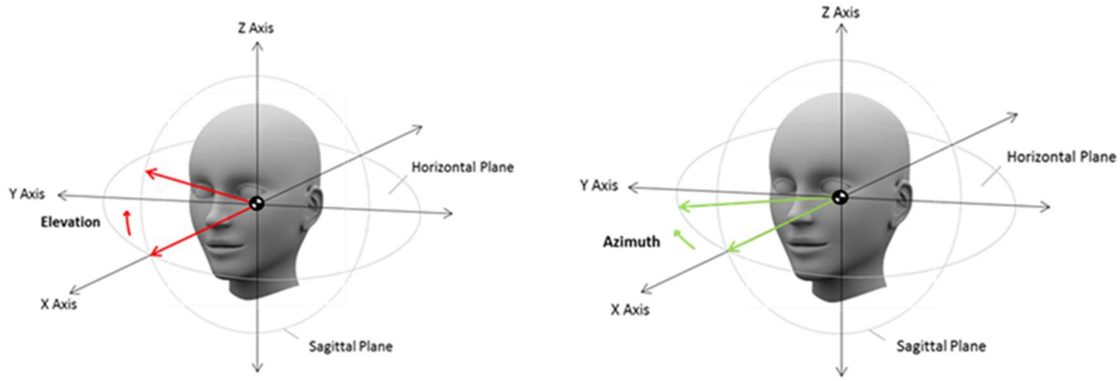
### *Impact Locations*

In American football, impact location is shown to be distributed all over the helmet.[7,9,26,51] In 2013 rules were implemented making crown impacts illegal.[26] While they do still occur, it is a less common occurrence and will not be included in this study. All impact positions and angles were measured from a standardized head-form position based on a front centric impact location. Azimuth (θ) is defined as the angle between the impact location and the positive X axis in the X-Y (transverse) plane. Elevation (α) is defined as the angle between the impact location and the X-Y (transverse) plane. Impact site accuracy was ensured using pre-marked locations on the head-form. Impact conditions were defined by the relative orientation of the impactor to the Hybrid III structure. The five impact sites and four impact directions are defined in Figures 2 and 3 below. Typical set up for a Front PE (positive elevation) and Rear Boss CG (centre of gravity) helmeted head-form are shown in figure 4.




Study 1 - Figure 2: Impact Locations

| Direction | Definition |
| --- | --- |
| Center of Gravity | Perpendicular to the head-form, no vertical or horizontal rotation applied |
| Positive Elevation | A 15° elevation of the Hybrid III head and neck-form |
| Positive Azimuth | A 45° rotation of the Hybrid III head and neck-form |
| Negative Azimuth | A -45° rotation of the Hybrid III head and neck-form |




Study 1 - Figure 3: Impact Directions

a) Helmeted Front Positive Elevation b) Helmeted Rear Boss Centre of Gravity
Study 1 - Figure 4: Helmeted impact set up

***Collection System***

The velocity of the impact ram was captured by an electronic time gate (width 0.2525 ± 0.0001 m) and recorded using National Instruments VI-logger software. An array of nine mounted single-axis Endevco 7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) were used and signals were passed through a TDAS Pro Lab system (DTS, Calabasas, CA) before being processed by TDAS software. The impact data were sampled at 20 kHz, filtered with a CFC 1000 filter (low pass cut off of 1650 Hz) and recorded using DTS TDAS PRO software.

***Computational Modeling***

Finite element models of the brain represent a tool for evaluating changes to impact location can influence brain response through comparisons of the computed brain tissue strains.[31] The University College Dublin Brain Trauma Model (UCDBTM) was used to obtain maximum principal strain for comparison measure of brain tissue deformation between impact conditions (location and direction).[13,52] Computed Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced contour photographs were used to determine the geometry from a single male cadaver.[53] The model is constructed of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brainstem. A linearly viscoelastic material model combined with a large deformation theory was chosen to model the brain tissue. The material properties can be found in Tables 2 and 3. The behaviour of the tissue was modeled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus.[53] The shear characteristics of the viscoelastic behaviour were expressed through the formula:

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

Where G is the long-term shear modulus, $G_0$ is the short-term shear modulus and β is the decay factor.[53]

| Material | Young's Modulus (MPa) | Poisson's Ratio | Density (kg/m$^2$) |
|---|---|---|---|
| Scalp | 16.7 | 0.42 | 1000 |
| Cortical bone | 15,000 | 0.22 | 2000 |
| Trabecular bone | 1000 | 0.24 | 1300 |
| Dura | 31.5 | 0.45 | 1130 |
| Pia | 11.5 | 0.45 | 1130 |
| Falx | 31.5 | 0.45 | 1140 |
| Tentorium | 31.5 | 0.45 | 1140 |
| CSF | 0.015 | 0.50 | 1000 |
| Grey matter | Viscoelastic | 0.49 | 1060 |
| White matter | Viscoelastic | 0.49 | 1060 |

Study 1 - Table 2: Material properties used in the University of College Dublin Brain Trauma Model.[53]

| | Shear Modulus (kPa) | | Decay Constant (s$^{-1}$) | Bulk Modulus (GPa) |
|---|---|---|---|---|
| | $G_0$ | $G_\infty$ | | |
| Grey matter | 10 | 2 | 80 | 2.19 |
| White matter | 12.5 | 2.5 | 80 | 2.19 |

| | | | | |
|---|---|---|---|---|
| Brain stem | 12.5 | 4.5 | 80 | 2.19 |
| White matter | 10 | 2 | 80 | 2.19 |

Study 1 - Table 3: Material properties of the different parts of the brain used in the University College Dublin Brain Trauma Model.[54]

The CSF layer was modelled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF, and brain was used. The model is comprised of 7318 hexahedral elements representing the brain, and 2874 hexahedral elements representing the CSF layer. This version of the UCDBTM does not include elements representing the cerebral blood vessels. The model was completed using comparisons with intracranial impact cadaveric pressure data from Nahum (1977) and Trosseille (1992) and brain motion from Hardy (2001).[55-57]

***Statistical Analysis***

The coefficient of determination ($r^2$) was calculated between the peak resultant (R), peak Dominant (D), the three-dimensional components (X, Y, Z) of dynamic response, and maximum principal using SPSS statistics software (v17.0, IBM, Armonk, NY).

The analysis was conducted using the full data set, grouped by location (front, front boss, side, rear boss and rear), and by condition (location and direction).

**Results**

The impacts conducted generated a wide span of linear and rotational kinematics for the various impact conditions. The linear acceleration magnitudes ranged as follows; the peak resultant ranged from 13 to 282g, peak X coordinate component from 4 to 174g, peak Y coordinate component from 1 to 227g, peak Z coordinate component from 3 to 66g, and dominant component ranged from 11 to 227g (table 4). The rotational acceleration magnitudes varied as follows; the peak resultant ranged from 603 to 28,270rad/s$^2$, peak X coordinate component from 94 to 12,618rad/s$^2$, peak Y coordinate component from 224 to 16,170rad/s$^2$, peak Z coordinate component from 123 to 15,565rad/s$^2$, and dominant component ranged from 552 to 20,005rad/s$^2$ (table 4). The maximum principal strain generated from the kinematic data ranging from 0.07 to 0.88 (table 4).

| Impact Location | Impact Speed | Linear Acceleration | | | | | Rotational Acceleration | | | | | MPS |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | R (g) | X (g) | Y (g) | Z (g) | D (g) | R (rads/ $s^2$) | X (rads/ $s^2$) | Y (rads/ $s^2$) | Z (rads/ $s^2$) | D (rads/ $s^2$) | |
| Front CG | 2.0 m/s | 19.9 | 19.6 | 1.4 | 10.8 | 19.6 | 602.7 | 341.0 | 561.3 | 270.3 | 561.3 | 0.07 |
| | 4.0 m/s | 56.8 | 56.3 | 5.9 | 9.6 | 56.3 | 3248.9 | 537.7 | 3224.2 | 831.1 | 3224.2 | 0.26 |
| | 6.0 m/s | 96.4 | 95.2 | 7.3 | 36.5 | 95.2 | 5681.4 | 1432.1 | 5661.5 | 887.3 | 5661.5 | 0.37 |
| | 8.0 m/s | 120.2 | 118.0 | 10.8 | 47.7 | 118.0 | 8656.5 | 2975.5 | 8611.0 | 1825.5 | 8611.0 | 0.49 |
| Front NA | 2.0 m/s | 18.2 | 14.6 | 11.0 | 7.6 | 14.6 | 1319.1 | 615.8 | 496.4 | 1096.8 | 1096.8 | 0.10 |
| | 4.0 m/s | 52.3 | 38.5 | 35.8 | 9.9 | 40.7 | 3510.3 | 1938.1 | 2580.5 | 2852.2 | 2892.2 | 0.29 |
| | 6.0 m/s | 94.3 | 65.1 | 70.0 | 28.5 | 72.4 | 7518.0 | 4618.9 | 3434.9 | 5587.1 | 5587.1 | 0.48 |
| | 8.0 m/s | 119.5 | 89.7 | 79.5 | 41.6 | 89.7 | 8055.6 | 4976.4 | 4717.6 | 5141.2 | 5406.8 | 0.57 |
| Front PA | 2.0 m/s | 21.6 | 15.9 | 14.7 | 6.5 | 15.9 | 1337.3 | 865.7 | 597.0 | 775.5 | 865.7 | 0.12 |
| | 4.0 m/s | 52.3 | 29.4 | 43.4 | 10.8 | 43.4 | 3359.3 | 2146.0 | 2002.1 | 2746.4 | 2746.4 | 0.25 |
| | 6.0 m/s | 99.5 | 68.6 | 70.7 | 30.3 | 73.1 | 6352.4 | 3640.3 | 2171.7 | 5118.3 | 5118.3 | 0.39 |
| | 8.0 m/s | 118.5 | 78.3 | 88.2 | 32.6 | 88.2 | 10291.7 | 7047.9 | 3847.4 | 6872.2 | 7144.5 | 0.62 |
| Front PE | 2.0 m/s | 20.3 | 19.9 | 2.0 | 9.6 | 19.9 | 636.0 | 283.0 | 585.7 | 372.3 | 585.7 | 0.08 |
| | 4.0 m/s | 43.8 | 42.2 | 4.7 | 19.2 | 42.2 | 4286.4 | 959.7 | 4275.9 | 694.9 | 4275.9 | 0.36 |
| | 6.0 m/s | 126.1 | 115.4 | 8.4 | 53.4 | 115.4 | 11526.5 | 2770.7 | 11385.0 | 1703.7 | 11385.0 | 0.58 |
| | 8.0 m/s | 121.2 | 111.1 | 14.5 | 66.5 | 111.1 | 16233.5 | 4237.5 | 16170.1 | 1978.6 | 16170.1 | 0.78 |
| Front Boss CG | 2.0 m/s | 18.6 | 12.9 | 13.6 | 5.8 | 13.9 | 1056.0 | 700.3 | 577.2 | 539.7 | 700.3 | 0.10 |
| | 4.0 m/s | 57.6 | 37.6 | 44.2 | 9.3 | 44.2 | 3203.8 | 2448.6 | 2116.3 | 2252.8 | 2448.6 | 0.29 |
| | 6.0 m/s | 94.4 | 56.0 | 77.3 | 29.0 | 77.3 | 7320.7 | 5223.3 | 3613.9 | 4416.3 | 5517.7 | 0.47 |
| | 8.0 m/s | 119.9 | 67.6 | 101.4 | 40.8 | 101.4 | 7762.9 | 6229.9 | 3596.9 | 4911.1 | 6229.9 | 0.54 |
| Front Boss NA | 2.0 m/s | 20.2 | 19.9 | 3.8 | 10.2 | 19.9 | 896.6 | 442.7 | 857.5 | 308.9 | 857.5 | 0.09 |
| | 4.0 m/s | 56.8 | 55.8 | 9.3 | 18.3 | 55.8 | 3047.2 | 1019.6 | 2964.4 | 1061.4 | 2964.4 | 0.26 |
| | 6.0 m/s | 78.6 | 76.7 | 25.7 | 47.5 | 76.7 | 6254.0 | 2815.8 | 6185.0 | 3580.1 | 6185.0 | 0.35 |
| | 8.0 m/s | 96.1 | 92.1 | 30.8 | 41.3 | 92.1 | 6259.0 | 2705.4 | 5913.5 | 4403.9 | 5913.5 | 0.38 |
| Front Boss PA | 2.0 m/s | 19.3 | 3.8 | 19.2 | 3.2 | 19.2 | 1234.9 | 1062.5 | 284.5 | 648.3 | 1062.5 | 0.11 |
| | 4.0 m/s | 44.3 | 5.0 | 44.3 | 9.8 | 44.3 | 3227.2 | 2354.5 | 340.2 | 2344.9 | 2373.0 | 0.23 |
| | 6.0 m/s | 102.6 | 12.3 | 102.4 | 24.0 | 102.4 | 7685.1 | 3481.2 | 1130.4 | 7197.1 | 7197.1 | 0.37 |
| | 8.0 m/s | 173.4 | 17.3 | 172.7 | 26.6 | 172.7 | 11480.8 | 8501.4 | 1475.5 | 8254.3 | 9389.5 | 0.56 |
| Front Boss PE | 2.0 m/s | 15.8 | 9.8 | 12.4 | 4.5 | 12.4 | 736.8 | 471.7 | 274.0 | 552.4 | 552.4 | 0.07 |
| | 4.0 m/s | 61.8 | 38.6 | 48.4 | 13.6 | 48.4 | 3374.0 | 2167.4 | 2296.9 | 2488.7 | 2488.7 | 0.26 |
| | 6.0 m/s | 107.0 | 50.9 | 94.2 | 32.0 | 94.2 | 9489.5 | 7221.1 | 2021.8 | 5903.3 | 7221.1 | 0.55 |
| | 8.0 m/s | 119.1 | 59.3 | 104.7 | 40.4 | 104.7 | 10748.6 | 7530.8 | 3743.6 | 8034.7 | 8281.2 | 0.54 |

| | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Side CG | 2.0 m/s | 17.8 | 3.9 | 17.4 | 3.8 | 17.4 | 1136.6 | 1102.4 | 223.8 | 671.5 | 1102.4 | 0.12 |
| | 4.0 m/s | 45.8 | 10.5 | 44.8 | 13.4 | 44.8 | 2743.1 | 2541.0 | 418.2 | 1185.6 | 2541.0 | 0.24 |
| | 6.0 m/s | 91.7 | 18.4 | 89.7 | 25.0 | 89.7 | 4434.2 | 4265.8 | 1071.9 | 1530.4 | 4265.8 | 0.40 |
| | 8.0 m/s | 219.2 | 27.5 | 218.5 | 31.5 | 218.5 | 12158.4 | 11528.3 | 1463.0 | 6107.5 | 11528.3 | 0.59 |
| Side NA | 2.0 m/s | 17.0 | 9.8 | 14.3 | 5.7 | 14.3 | 1161.5 | 810.3 | 875.4 | 484.6 | 911.7 | 0.11 |
| | 4.0 m/s | 52.4 | 26.0 | 45.8 | 9.3 | 45.8 | 2984.4 | 2404.7 | 1856.5 | 2203.4 | 2404.7 | 0.28 |
| | 6.0 m/s | 84.5 | 46.1 | 71.2 | 19.4 | 71.2 | 4942.4 | 4152.6 | 1981.5 | 2523.2 | 4152.6 | 0.36 |
| | 8.0 m/s | 122.0 | 47.7 | 112.6 | 28.0 | 112.6 | 7203.1 | 5266.7 | 2888.1 | 5595.4 | 6212.8 | 0.48 |
| Side PA | 2.0 m/s | 13.1 | 6.9 | 11.2 | 2.9 | 11.2 | 1376.7 | 972.9 | 739.2 | 857.1 | 972.9 | 0.14 |
| | 4.0 m/s | 33.0 | 13.7 | 30.4 | 9.6 | 30.4 | 3238.5 | 2588.6 | 1532.1 | 2527.1 | 2649.3 | 0.29 |
| | 6.0 m/s | 74.8 | 26.3 | 70.3 | 15.5 | 70.3 | 7761.7 | 4297.4 | 2846.3 | 5864.2 | 5864.2 | 0.46 |
| | 8.0 m/s | 177.5 | 66.2 | 164.5 | 23.0 | 164.5 | 20032.0 | 11101.2 | 6308.2 | 15564.7 | 15564.7 | 0.82 |
| Side PE | 2.0 m/s | 18.2 | 5.8 | 17.8 | 3.1 | 17.8 | 1488.4 | 1096.8 | 160.1 | 1003.9 | 1096.8 | 0.13 |
| | 4.0 m/s | 50.2 | 9.4 | 49.6 | 13.3 | 49.6 | 2527.7 | 2463.9 | 594.9 | 1338.3 | 2463.9 | 0.26 |
| | 6.0 m/s | 89.0 | 19.9 | 87.0 | 24.4 | 87.0 | 4738.1 | 4499.1 | 1064.0 | 1742.0 | 4499.1 | 0.39 |
| | 8.0 m/s | 227.5 | 24.4 | 227.3 | 32.4 | 227.3 | 13040.6 | 12480.2 | 1531.9 | 5641.2 | 12480.2 | 0.60 |
| Rear Boss Cg | 2.0 m/s | 18.0 | 12.0 | 13.5 | 3.0 | 13.5 | 1596.8 | 1031.3 | 957.4 | 990.7 | 1042.5 | 0.13 |
| | 4.0 m/s | 45.7 | 34.5 | 30.5 | 10.3 | 30.5 | 3854.6 | 2284.2 | 2661.3 | 3111.2 | 3111.2 | 0.28 |
| | 6.0 m/s | 88.8 | 60.1 | 65.8 | 17.5 | 65.8 | 7223.4 | 3745.6 | 4346.8 | 4407.9 | 4610.3 | 0.39 |
| | 8.0 m/s | 196.2 | 129.7 | 146.6 | 31.0 | 146.6 | 19578.6 | 9161.8 | 11287.2 | 13256.2 | 13256.2 | 0.75 |
| Rear Boss NA | 2.0 m/s | 16.9 | 4.4 | 16.3 | 3.3 | 16.3 | 1500.7 | 831.8 | 236.2 | 1384.8 | 1384.8 | 0.13 |
| | 4.0 m/s | 33.8 | 8.4 | 33.5 | 8.8 | 33.5 | 3387.6 | 2248.3 | 938.7 | 2853.9 | 2853.9 | 0.31 |
| | 6.0 m/s | 64.7 | 18.4 | 64.0 | 11.5 | 64.0 | 7747.9 | 4333.1 | 1177.6 | 6345.5 | 6345.5 | 0.47 |
| | 8.0 m/s | 183.6 | 35.3 | 179.6 | 18.9 | 179.6 | 22916.5 | 13282.1 | 4099.5 | 18299.0 | 18299.0 | 0.77 |
| Rear Boss PA | 2.0 m/s | 16.7 | 14.0 | 9.5 | 2.6 | 14.0 | 1381.3 | 612.0 | 743.8 | 1026.1 | 1026.1 | 0.10 |
| | 4.0 m/s | 63.3 | 62.4 | 10.7 | 13.2 | 62.4 | 3761.2 | 1058.2 | 3599.4 | 862.0 | 3599.4 | 0.23 |
| | 6.0 m/s | 67.0 | 61.6 | 27.5 | 14.3 | 61.6 | 6032.3 | 1692.3 | 4681.1 | 3813.4 | 4681.1 | 0.35 |
| | 8.0 m/s | 107.7 | 97.5 | 45.6 | 24.9 | 97.5 | 7319.0 | 2286.1 | 6654.8 | 6039.6 | 6654.8 | 0.44 |
| Rear Boss PE | 2.0 m/s | 14.5 | 9.7 | 10.9 | 3.3 | 10.9 | 1131.9 | 583.6 | 651.5 | 831.6 | 831.6 | 0.10 |
| | 4.0 m/s | 34.6 | 20.5 | 28.4 | 6.9 | 28.4 | 3288.3 | 2147.4 | 2015.1 | 2558.3 | 2558.3 | 0.24 |
| | 6.0 m/s | 117.1 | 77.7 | 87.1 | 17.1 | 87.1 | 11554.6 | 5412.5 | 6826.9 | 7591.0 | 7041.0 | 0.51 |
| | 8.0 m/s | 281.6 | 174.5 | 214.8 | 52.5 | 214.8 | 28269.6 | 12618.4 | 15629.1 | 20005.3 | 20005.3 | 0.88 |
| Rear CG | 2.0 m/s | 21.8 | 21.8 | 0.9 | 4.7 | 21.8 | 1114.0 | 93.6 | 1113.2 | 199.0 | 1113.2 | 0.10 |
| | 4.0 m/s | 68.5 | 68.0 | 3.7 | 15.0 | 68.0 | 4636.6 | 536.6 | 4629.9 | 421.3 | 4629.9 | 0.26 |

| | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | 6.0 m/s | 87.6 | 87.3 | 7.4 | 22.3 | 87.3 | 7095.2 | 792.0 | 7015.2 | 1091.5 | 7015.2 | 0.35 |
| | 8.0 m/s | 166.6 | 166.5 | 8.0 | 39.9 | 166.5 | 8869.6 | 1392.2 | 8863.0 | 1178.4 | 8863.0 | 0.48 |
| Rear NA | 2.0 m/s | 17.1 | 14.5 | 9.4 | 3.2 | 14.5 | 1285.2 | 583.5 | 725.8 | 915.5 | 915.5 | 0.11 |
| | 4.0 m/s | 52.6 | 45.7 | 26.1 | 13.0 | 45.7 | 4217.4 | 1612.4 | 2738.8 | 3084.4 | 3084.4 | 0.26 |
| | 6.0 m/s | 74.3 | 69.6 | 25.9 | 15.0 | 69.6 | 6159.4 | 1944.5 | 4792.6 | 4003.0 | 4792.6 | 0.39 |
| | 8.0 m/s | 149.4 | 135.0 | 64.4 | 21.9 | 135.0 | 13306.6 | 2523.2 | 9085.3 | 9825.3 | 9689.6 | 0.77 |
| Rear PA | 2.0 m/s | 16.7 | 14.0 | 9.5 | 2.6 | 14.0 | 1381.3 | 512.0 | 743.8 | 1026.1 | 1026.1 | 0.10 |
| | 4.0 m/s | 51.3 | 43.3 | 27.4 | 10.3 | 43.3 | 4308.8 | 1978.3 | 2672.7 | 2782.9 | 2782.9 | 0.23 |
| | 6.0 m/s | 65.8 | 53.4 | 39.1 | 16.0 | 53.4 | 5720.1 | 2170.2 | 3517.2 | 4380.8 | 4380.8 | 0.36 |
| | 8.0 m/s | 146.1 | 126.6 | 73.1 | 22.9 | 126.6 | 12912.0 | 2621.5 | 8456.9 | 9725.8 | 9725.8 | 0.77 |
| Rear PE | 2.0 m/s | 15.3 | 15.2 | 1.0 | 3.9 | 15.2 | 1023.6 | 149.7 | 1021.5 | 123.0 | 1021.5 | 0.10 |
| | 4.0 m/s | 44.2 | 44.2 | 2.1 | 7.6 | 44.2 | 3469.7 | 643.1 | 3458.2 | 578.4 | 3458.2 | 0.22 |
| | 6.0 m/s | 69.6 | 68.7 | 5.3 | 14.6 | 68.7 | 5521.0 | 839.3 | 5501.8 | 695.1 | 5501.8 | 0.35 |
| | 8.0 m/s | 171.0 | 168.3 | 5.2 | 38.3 | 168.3 | 14293.3 | 1232.2 | 14276.5 | 1032.2 | 14276.5 | 0.55 |

Study 1 - Table 4: Peak head kinematics and maximum principal strain.

Abbreviations: R, resultant; D, dominant; CG, centre of gravity; NA, negative azimuth; PA, positive azimuth; PE, positive elevation; MPS, maximum principal strain.

***Relationship ($r^2$) between components of dynamic response and MPS across all impacts***

The values for coefficient of determination between the coordinate components of acceleration and MPS are represented in table 5 for all impacts, grouped and analysed together. The resultant and dominant component of both linear and rotational acceleration had high correlation to MPS. The X, Y, and Z components of linear and rotational acceleration had moderate correlation to MPS.

| | $r^2$ Values | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | Linear Acceleration | | | | | Rotational Acceleration | | | | |
| | X | Y | Z | D | R | X | Y | Z | D | R |
| All Impacts | 0.48 | 0.50 | 0.58 | 0.77 | 0.83 | 0.61 | 0.51 | 0.63 | 0.85 | 0.87 |

Study 1 - Table 5: $r^2$ values depicting the correlation between the coordinate components (X, Y, Z, and Dominant) and peak resultant of linear and rotational acceleration with MPS collapsed across all impact conditions.

***Relationship ($r^2$) between components of dynamic response and MPS by impact locations***

The values for coefficient of determination between the coordinate components of acceleration and MPS analysed by grouping the data by impact location are presented in table 6. At each location the dominant and resultant components of both linear and rotational acceleration had high correlation to MPS.

At the front location: the X, and Z components of linear acceleration had high correlation, the X, Y and Z components of rotational acceleration had moderate correlation and the Y component of linear acceleration had low correlation to MPS.

At the front boss location: The Y component of linear and the X and Z component of rotational acceleration had high correlation, the Z component of linear acceleration had moderate correlation and the X component of linear and Y component of rotational acceleration had low correlation to MPS.

At the side location: the Y and Z components of linear and the X and Z components of rotational acceleration had high correlation, the X of linear and the Y of rotational acceleration had moderate correlation to MPS.

At the rear boss location: the Y and Z components of linear and the X and Z components of rotational acceleration had high correlation, the X component of linear and the Y component of rotational acceleration had moderate correlation to MPS.

At the rear location: the X component of linear had high correlation, the Y and Z of linear and the X, Y, and Z components of rotational acceleration had moderate correlation to MPS.

| | $r^2$ Values | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | Linear Acceleration | | | | | Rotational Acceleration | | | | |
| | X | Y | Z | D | R | X | Y | Z | D | R |
| Front | 0.77 | 0.19 | 0.81 | 0.80 | 0.87 | 0.68 | 0.67 | 0.32 | 0.83 | 0.95 |
| Front Boss | 0.31 | 0.71 | 0.65 | 0.82 | 0.90 | 0.89 | 0.28 | 0.81 | 0.88 | 0.93 |
| Side | 0.66 | 0.78 | 0.70 | 0.78 | 0.81 | 0.95 | 0.63 | 0.80 | 0.93 | 0.93 |
| Rear Boss | 0.59 | 0.92 | 0.77 | 0.93 | 0.92 | 0.89 | 0.69 | 0.93 | 0.93 | 0.94 |
| Rear | 0.74 | 0.55 | 0.55 | 0.74 | 0.82 | 0.61 | 0.69 | 0.63 | 0.76 | 0.92 |

Study 1 - Table 6: $r^2$ values depicting the correlation between the coordinate components (X, Y, Z, and Dominant) and peak resultant of linear acceleration with MPS by impact location.

***Relationship ($r^2$) between components of dynamic response and MPS at each impact condition***

When analyzed by impact condition, all components (X, Y, Z, D and R) of linear and rotational acceleration had high correlation to MPS as presented in table 7.

| | | $r^2$ Values | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | Linear Acceleration | | | | | Rotational Acceleration | | | | |
| | | X | Y | Z | D | R | X | Y | Z | D | R |
| Front | CG | 0.99 | 0.99 | 0.79 | 0.99 | 0.99 | 0.82 | 0.98 | 0.89 | 0.98 | 0.98 |
| | NA | 0.98 | 0.99 | 0.88 | 0.99 | 0.99 | 0.97 | 0.97 | 0.98 | 0.95 | 0.98 |
| | PA | 0.89 | 0.94 | 0.84 | 0.93 | 0.93 | 0.99 | 0.95 | 0.93 | 0.98 | 0.99 |
| | PE | 0.87 | 0.93 | 0.94 | 0.87 | 0.86 | 0.94 | 0.97 | 0.97 | 0.97 | 0.97 |
| Front Boss | CG | 0.99 | 0.99 | 0.90 | 0.99 | 0.99 | 0.98 | 0.96 | 0.94 | 0.98 | 0.97 |
| | NA | 0.99 | 0.86 | 0.81 | 0.99 | 0.98 | 0.85 | 0.93 | 0.99 | 0.93 | 0.94 |
| | PA | 0.96 | 0.84 | 0.94 | 0.93 | 0.93 | 0.93 | 0.92 | 0.85 | 0.95 | 0.97 |
| | PE | 0.92 | 0.99 | 0.94 | 0.99 | 0.98 | 0.98 | 0.94 | 0.90 | 0.97 | 0.97 |
| Side | CG | 0.99 | 0.94 | 0.97 | 0.94 | 0.94 | 0.90 | 0.98 | 0.92 | 0.90 | 0.90 |
| | NA | 0.90 | 0.98 | 0.91 | 0.98 | 0.99 | 0.98 | 0.97 | 0.80 | 0.97 | 0.98 |
| | PA | 0.97 | 0.99 | 0.97 | 0.99 | 0.98 | 0.97 | 0.99 | 0.91 | 0.97 | 0.97 |
| | PE | 0.94 | 0.94 | 0.96 | 0.94 | 0.95 | 0.92 | 0.98 | 0.98 | 0.92 | 0.91 |
| Rear Boss | CG | 0.99 | 0.99 | 0.98 | 0.98 | 0.99 | 0.99 | 0.99 | 0.84 | 0.98 | 0.98 |
| | NA | 0.97 | 0.92 | 0.99 | 0.92 | 0.92 | 0.92 | 0.90 | 0.98 | 0.92 | 0.92 |
| | PA | 0.89 | 0.87 | 0.92 | 0.89 | 0.92 | 0.97 | 0.98 | 0.92 | 0.98 | 0.99 |
| | PE | 0.98 | 0.97 | 0.93 | 0.97 | 0.97 | 0.98 | 0.98 | 0.85 | 0.95 | 0.97 |
| Rear | CG | 0.94 | 0.93 | 0.96 | 0.94 | 0.94 | 0.98 | 0.99 | 0.96 | 0.99 | 0.99 |
| | NA | 0.99 | 0.96 | 0.88 | 0.99 | 0.99 | 0.85 | 0.99 | 0.88 | 0.99 | 0.99 |
| | PA | 0.99 | 0.99 | 0.91 | 0.99 | 0.99 | 0.76 | 0.99 | 0.99 | 0.99 | 0.99 |
| | PE | 0.95 | 0.81 | 0.93 | 0.95 | 0.95 | 0.96 | 0.95 | 0.99 | 0.95 | 0.95 |

Study 1 - Table 7: $r^2$ values between the coordinate components (X, Y, Z, and Dominant) and peak resultant of linear acceleration with MPS for each impact condition. [CG, centre of gravity; NA, negative azimuth; PA, positive azimuth; PE, positive elevation]

**Discussion**

Research has identified directional susceptibility of head motion to injury, yet this has yet to be fully considered in helmet innovation in relation to injury protection.[12,18,19,22,58] The purpose of this study was to determine the correlation between coordinate components of acceleration (X, Y, Z and Dominant), which can define head motion, and MPS, a common measure used to depict injury severity. Current and proposed American football standards use peak resultant linear and/or rotational acceleration as the pass/fail performance metrics, which cannot fully account for the influence of impact location and direction on resulting head motion. The information from this study, in conjunction with current standards, can be used to guide helmet development towards improved performance.

To understand the influence of impact location and direction on tissue strain, it was important to examine the results of this study as a whole data set and with impact location and condition independently. The relationship between the coordinate component of linear and rotational acceleration and MPS changes with impact location and condition. A moderate correlation existed between the X, Y, and Z coordinate components and MPS when all the impact locations and conditions are collapsed. This is because the influence of location and direction is removed and a wash out effect is created. When the impact dataset was analyzed by location, a strong correlation between one or two components of acceleration correlated strongly to MPS, highlighting the relationship between impact location and severity of injury. However, the most influential component varied according to impact location and was not consistent for linear and rotational acceleration as shown in table 6. This research supports the notion that location specific protection could decrease risk of head injury.[16,31,34,51,59,60] When analyzing impact condition (location and direction) independently, all coordinate components of linear and rotational acceleration correlated highly to MPS. This occurs as the magnitude of each component increases with velocity, resulting in increased MPS values and a positive linear relationship. The components of acceleration (X, Y, and Z) characterize the directional motion of the head and define the dominant component of acceleration for each impact condition. However, the relationship to MPS varies dependent on the method of analysis (collapsed, by location, or by condition). This creates the need to search for alternative measures capable of defining directional motion yet insensitive to analysis method, such as the dominant component of acceleration.

The results demonstrated a high correlation between both the resultant and dominant components of linear and rotational acceleration and MPS. The relationship strength between resultant and dominant component of acceleration to MPS remained strong regardless of whether the data analysed was collapsed (including all impacts at different locations and directions), by impact location, or by impact condition. The link between resultant linear and rotational acceleration and MPS is well documented and consistent with previous research.[33-37] However, the correlation between the dominant component of acceleration and MPS is novel information. The dominant component accounts for impact direction and location and therefore provides a more tangible performance evaluation and design metric than resultant. This new information offers an effective metric for helmet innovation aimed at reducing maximum principal strain.

***Limitations***

The results of this study are presented within the context of certain limitations. The impact data collected within the laboratory use a non-biofidelic physical headform that may not result in identical responses to those of the human head, neck musculature, body mass, and overall human response to injury. However, the use of a controlled laboratory setting for helmet impact performance testing controls the wide variance reflected in on-field impact scenarios.[61] This study was conducted using a model of American football helmet. Experimental data obtained using other football helmet models and helmets from other sports would further add to the context of this study. The finite element model has limitations which are well documented.[36,37,53,54,62,63] However, the FE model is an approximation intended to simulate human responses to injurious loading and is a useful tool to compare between different impacts giving relative rather than absolute values.[36]

**Conclusion**

This paper identified the correlation of not only the resultant, which is consistent with previous research, but also the dominant coordinate component of linear and rotational acceleration with maximum principal strain regardless of whether the data was analyzed using all impact together, by location or condition. This identifies the potential usefulness of dominant coordinate components of acceleration for helmet design, innovation and performance evaluation aimed at reducing maximum principal strain. Dominant coordinate components of acceleration provide a novel tool and useful measure that accounts for impact direction and location, which resultant acceleration does not. Additionally, the paper identifies the importance of impact location in relation to injury risk, which again is consistent with previous research, but can be addressed through helmet innovation using the dominant coordinate component of acceleration to inform innovation.

**Data Availabilit**y: The raw/processed data required to reproduce these findings is shared within the context of this manuscript.

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**References:**


1. **Bailes JE,** Hudson V. Classification of Sport-Related Head Trauma: A Spectrum of Mild to Severe Injury. J Athletic Train. 2001; 36(3):236-243.
2. **McCrea M,** Hammeke T, Olsen G, et al. Unreported concussion in high school football players: implications for prevention. Clin J Sports Med. 2004; 14(1):13-17
3. **Duma SM,** Manoogian SJ, Bussone WR. Analysis of real-time head accelerations in collegiate football players. Clin. J. Sports Med. 200; 15(1):3-8.
4. **Collins CL,** Comstock RD, Dick RW, et al. Concussions among United States high school and collegiate athletes. J Athletic Train. 2007; 42(4):495-503.
5. **Gessel LM,** Fields SK, Collins CL, et al. concussions among United States high school and collegiate athletes. J Athl Train. 2007; 42(2):495-503.
6. **Hootman JM,** Dick R, Agel J. Epidemiology of Collegiate Injuries for 15 Sports: Summary and Recommendations for Injury Prevention Initiatives. J Athl. Train. 2007; 42(2):311-319.
7. **Crisco JJ,** Fiore R, Beckwith JG, et al. Frequency and location of head impact exposures in individual collegiate football players. J Athletic Train. 2010; 45(6): 549-559.
8. **Casson IR,** Viano DC, Powell JW, et al. Twelve years of National Football League concussion data. Sports Health. 2010; 2(6): 471–483.
9. **Rowson S**, Duma SM, Beckwith JG, et al. Rotational head kinematics in football impacts: An injury risk function for concussion. Ann. Biomed. Eng. 2012; 40(1): 1–13.
10. **Fowler JN,** Pallanta AA, Swanik CB, et al. The use of shear thickening nanocomposites in impact resistant materials. J. Biomed. Eng. 2015; 137:054505-1-5.
11. **Ommaya AK,** Gennarelli TA. Cerebral concussion and traumatic unconsciousness. Brain. 1974; 97:633 – 654.
12. **Prange MT,** Margulies SS. Regional, Directional, and Age-Dependent Properties of the Brain Undergoing Large Deformation. J. Biomed. Eng. 2002. 124:244 – 252.
13. **Gilchrist, M.D.,** 2003. Modeling and Accident Reconstruction of Head Impact Injuries. Engineering Materials 245/245, 417-429.
14. **Zhang L,** Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury." J. Biomed. Eng. 2004; 126: 226 – 236.
15. **Takhounts EG,** Ridella SA. Investigation of traumatic brain injuries using the next generation of simulate injury monitor (SIMon) finite element head model. Stapp Car Crash Journal. 2008; 52: 1-31.
16. **Meaney DF,** Smith DH. Biomechanics of Concussion. Clin Sports Med 2011; 30:19-31. DOI: 10.1016/j/csm.2010.08.009.
17. **Gennarelli TA,** Thibault LE, Tomei G. Directional Dependence of Axonal Brain Injury Due to Centroidal and Non-centroidal Acceleration. 31st Stapp Car Crash Conference. 1987. SAE paper no. 8721`97. Society of Automotive Engineers.
18. **Kleiven S.** Influence of Impact Direction on Human Head in Prediction of Subdural Hematoma. J. Neurotrauma 2003; 20(4):365-379.
19. **Pellman EJ,** Viano DC, Tucker AM, et al. Concussion in professional football: reconstruction of game impacts and injuries. J. Neurosurgery. 2003; 53(4):799-814.
20. **Hrapko M,** van Dommelen JA, Peters GW. Et al. The Influence of Test Conditions on Characterization of the Mechanical Properties of Brain Tissue. J. Biomed. Eng Engineering 2008; 130(3): 031003-1-10.

21. **Giordano C,** Cloots RJH, van Dommelen JAW. Et al. The Influence of Anisotropy on Brain Injury Prediction. J. Biomech. 2014; 47:1052-1059.
22. **Post A,** Hoshizaki TB. Rotational acceleration, brain tissue strain, and the relationship to concussion. J. Biomech. Eng. 2015; 137:1-8
23. **Greenwald RM,** Gwin JT, Chu JJ, et al. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 2008; 62(4):789 – 797.
24. **Broglio SP,** et al. Head impacts during high school football: A biomechanical assessment. J Athletic Train. 2009; 44(4):342-349.
25. **Kerr ZY,** Collins CL, Mihalik JP. Et al. Impact Locations and Concussion Outcomes in High School Football Player-to-Player Collisions. Pediatrics 2014; 134(3):489-496.
26. **Clark MD,** Breton MA, Marshall SW. et al. Descriptive Characteristics of Concussion in National Football League Games. American Journal of Sports Medicine. 2017; 45(4):929-936.
27. **National Operating Committee on Standards for Athletic Equipment** (2004) Standard Drop Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear: NOCSAE Standard 001-04m05. Available at www.nocsae.org.
28. **National Operating Committee on Standards for Athletic Equipment** (2014) Standard Performance Specification for Newly Manufactured Football Helmets: NOCSAE DOC (ND) 002-14. Available at www.nocsae.org
29. **McIntosh AS,** Patton DA, Frechede B. et al. The Biomechanics of Concussion in Un-helmeted Football Players in Australia: A Case-Control Study. Biomedical Journal 2014; 4:1–9.
30. **McAllister TW,** Ford JC, Songbai J, et al. Maximum principal strain and strain rate associated with concussion diagnosis correlates with changes in corpus callosum white matter indices. Ann. Biomed. Eng. 2012;40(1):127-140.
31. **Elkin BS,** Gabler LF, Panzar MB. Et al. Brain tissue strains vary with head impact location: A possible explanation for increased concussion risk in struck versus striking football players. Clinical Biomechanics. 2018. [In Press]
32. **Yoganandan N,** Li J, Zhang J, et al. Influence of angular acceleration-deceleration pulse shapes on regional brain strains. J. Biomech. 2008; 41:2253-2262.
33. **Zhang L.** Computational Biomechanics of Traumatic Brain Injury, An Investigation of Head Impact Response and American football head injury. Doctor of Philosophy – Dissertation. Graduate School of Wayne State University, Detroit, Michigan. 2001
34. **Viano DC,** Casson IR, Pellman EJ, et al. Concussion in professional football comparison with boxing head impacts. Neurosurgery. 2005; 57(6): 1154-1172.
35. **Kleiven S.** Predictors for traumatic brain injuries evaluated through accident reconstructions. Stapp Car Crash Journal. 2007; 51: 81-114.
36. **Post A,** Oeur A, Hoshizaki B, et al. An examination of American football helmets using brain deformation metrics associated with concussion. Mater. Des. 2013; 45:653-662.
37. **Post A**, Oeur A, Hoshizaki TB, et al. The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research. International Research Council on Biomechanics of Injury Conference (IRCOBI), 2012 Dublin, Ireland.
38. **Post A,** Oeur A, Hoshizaki TB. The influence of velocity on the performance range of American football helmets. Proceeding of the 1st International Conference on Helmet Performance and Design. London, UK. HPD-2013-10

39. **Cournoyer J,** Post A, Rousseau P, Hoshizaki TB. The Ability of American Football Helmets to Manage Linear Acceleration with Repeated High-Energy Impacts. J. Athletic Train. 2016; 51(3):258-263.
40. **Newman JA.** Biomechanics of human trauma: Head protection. Accidental Injury Biomechanics and Prevention. New York, Springer-Verlag 1993; 292–310.
41. **Pellman EJ**, Viano DC, Withnall C, et al. Concussion in professional football: Helmet testing to assess impact performance – Part 11. J. Neurosurgery. 2006; 58:78 – 96.
42. **Gharjari M,** Peldschus S, Galvanetto U. et al. Evaluation of the Effective Mass of the Body for Helmets Impacts. International Journal of Crashworthiness. 2011; 16(6):621–631.
43. **Rousseau P,** Hoshizaki TB. Defining the effective impact mass of elbow and shoulder strikes in ice hockey. Sports Biomechanics. 2015; 1-11.
44. **Bartsch A**, Benzel E, Miele V, et al. Impact test comparisons of 20th and 21st century American football helmets. J. Neurosurgery. 2012; 116: 222-233.
45. **Post A,** Walsh ES, Hoshizaki TB. Et al. Analysis of Loading Curve Characteristics on the Production of Brain Deformation Metrics. Journal of Sports Engineering and Technology. 2012. DOI: 10.1177/1754337112449420.
46. **Schnebel B,** Gwin JT, Anderson S. et al. In Vivo Study of Head Impacts in Football: A Comparison of National Collegiate Athletic Association Division I Versus High School Impacts. Neurosurgery 2007; 60:490-496.
47. **Viano DC,** Casson I, Pellman EJ.Concussion in Professional Football: Biomechanics of the Struck Player – Part 14. Neurosurgery 2007; 16(2):313–327.
48. **ASTM F1446-13,** American Society for Testing and Materials International, 2013.
49. **Walsh ES.** Kendall M, Post A. et al. Comparative analysis of Hybrid III neckform and an unbiased neckform. Sports Engineering 2018 [In Review].
50. **Padgaonkar AJ,** Kreiger KW, King AI. Measurements of angular accelerations of a rigid body using linear accelerometers. J. Appl. Mech. 1975; 42: 552-556.
51. **Pellman EJ,** Viano DV, Tucker AM. Et. al. Concussion in Professional Football: Location and Direction of Helmet Impacts – Part 2. Neurosurgery 2003; 53:1328-1341.
52. **Doorly MC**, Gilchrist MD. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. Computer Methods in Biomechanics and Biomedical Engineering. 2006; 9(6): 371 – 377.
53. **Horgan TJ**, Gilchrist MD. The creation of 3D finite element models for simulating head impact biomechanics. Int. J. Crashworthiness. 2003; 8(4): 353 – 366.
54. **Horgan TJ,** Gilchrist MD. Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. Int. J. Crashworthiness. 2004; 9(4):401-418
55. **Nahum, A.M.,** Smith, R., Ward, C.C., 1977. Intracranial pressure dynamics during head impact. 21st Annual Stapp Car Crash Conference. Society of Automotive Engineers.
56. **Hardy WN,** Foster CD, Mason MJ, et al. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar x-ray. Stapp Car Crash Journal; The Stapp Association, Ann Arbor, Michigan. 2001.
57. **Trosseille X,** Tarrie´re C, Lavaste F, et al. Development of a F.E.M. of the human head according to a specific test protocol. In: 36th Stapp Car Crash Conference, Seattle, WA, USA, SAE paper 922527. 1992

58. **Weaver AA,** Danelson KA, Stitzel JD. Modeling Brain Injury Response for Rotational Velocities of Varying Directions and Magnitudes. J. Biomed. Eng. 2012; 40(9):2005-2018.
59. **Walsh ES,** Rousseau P, Hoshizaki TB. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. Sports Eng. 2011; 13(3):135-143.
60. **Liao S,** Lynall RC, Mihalik JP. The effect of head impact location on day of diagnosed concussion in college football. Med. Sci. Sports Exercise. 2016; 48(7):1239-1243
61. **Krzeminski DE**, Fernando BMD, Curtzwiler G, et al. Repetitive impact exposure and characterization of stress-whitening of an American football helmet outer shell material. Polym. Test. 2016; 55:190-203.
62. **Post A,** Oeur A, Walsh E. et al. A centric/non-centric impact protocol and finite element model methodology for the evaluation of American Football Helmets to Evaluate Risk of Concussion. Computer Methods in Biomechanics and Biomedical Engineering 2013;1-16.
63. **Oeur RA,** Karton C, Post A. et al. A Comparison of Head Dynamic Response and Brain Tissue Stress and Strain Using Accident Reconstruction for Concussion, Concussion with Persistent Postconcussive Symptoms and Subdural Hematoma. J. Neurosurgery. 2015; 123(2):415-422.

# Study 2

## The influence of impact force redistribution and redirection strategies on maximum principal strain

**Abstract**

Sporting helmets, with linear attenuating strategies, are proficient at reducing the risk of traumatic brain injury. However, the continued high incidence of concussion in American football, has led researchers to investigate novel helmet liner strategies. These strategies typically supplement existing technologies by adding or integrating head-helmet decoupling mechanisms. Decoupling strategies aim to redirect or redistribute impact force around the head, reducing impact energy transferred to the brain. This results in decreased brain tissue strain, which is beneficial in injury risk reduction due to the link between tissue strain and concussive injury. While offering the potential to advance helmet innovation and injury risk reduction, these strategies require further investigation to fully understand the influence on brain tissue strain and injury risk prior to implementation.

The purpose of this study was to demonstrate the effect of redirection and redistribution helmet liner strategies on brain tissue strain resulting from impacts to the head. The kinematic response data from head impacts, using 20 different impact conditions, was mathematically modified and used as input parameters to determine the effect on maximum principal strain (MPS) values, calculated using finite element modeling. The results showed liner strategies that reduce the dominant coordinate component, the component that contributes the greatest to the resultant, of rotational acceleration can reduce maximum principal strain in American football helmets. The study demonstrated redirection and/or redistribution strategies, if applied correctly, influence brain motion, reduce brain tissue strain, and decrease injury risk when implemented in an American football helmet.

**Null Hypothesis:** No significant differences will result between the maximum principal strain of the original impact to the maximum principal strain using the 10 strategies.

**Key words:** Helmet technology, impact redirection, impact redistribution, dominant coordinate component, brain injury, maximum principal strain

## Introduction

American football helmets are proficient at reducing impact induced head accelerations below levels known to cause traumatic brain injury (TBI).[1,2,3] However, current helmet designs may have reached their force attenuating limits as a result of liner thickness and material property constraints.[4] This can result in the magnitude of acceleration remaining above those known to cause concussive injury and may, in part, explain why the incident rate of concussion remains high in American football.[5-9] The force attenuating limitations of current helmet designs have led to alternative helmet liner methods aimed to reduce overall magnitude of impact force transferred to brain tissue.[10] Impact force redirection, using liner strategies to decouple the head and helmet interface, is a comparatively novel approach aimed at reducing force transferred to the brain by reducing brain motion. Decoupling to reduce dynamic response of the brain, primarily rotational acceleration, was studied by Aare (2003) who used finite element modeling to study how a Teflon layer created motion between the helmet shell and liner on impact.[11] The objective was to mitigate of a portion of the normal force, typically transferred into the head, shifting it to tangential force around the helmet (redirection through de-coupling). However, this was computational research and not physically tested. Finan (2008) reported reductions in peak rotational acceleration by decreasing the coefficient of friction between an impact surface and helmet.[12] The results support that similar results could be expected between the helmet-head interface.[12] The low friction or "decoupling" concept has been expanded by commercial helmet brands such as Multi-directional Impact Protection System (MIPS), CCM Rotational Energy Dampening (R.E.D) System, and 6D motorcycle helmets. These companies' implement technologies aimed at reducing rotational acceleration through decoupling. However, to the author's knowledge, none have investigated the influence of decoupling on three-dimensional brain motion or corresponding changes in brain tissue strain.

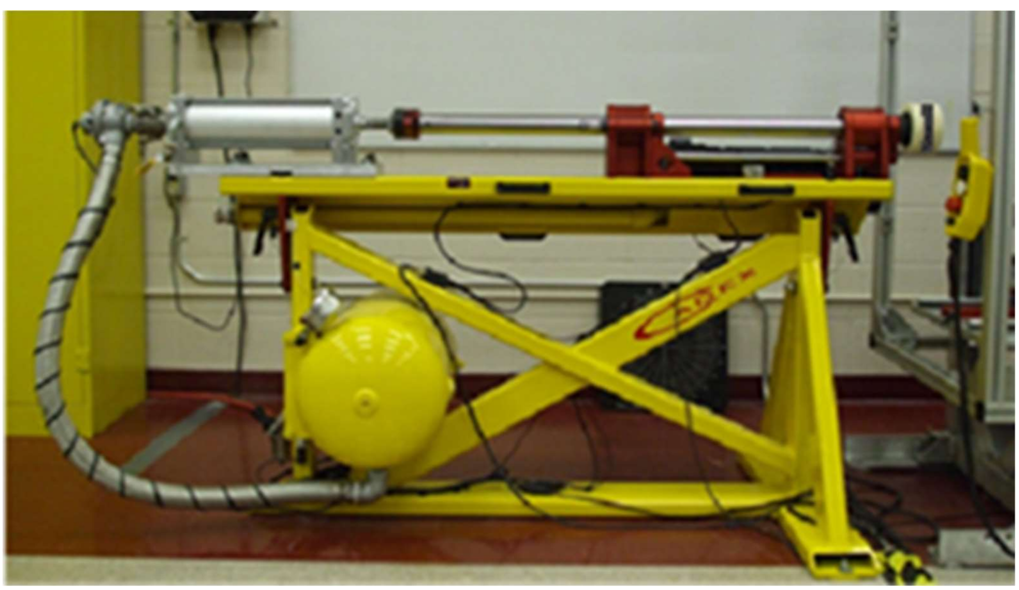
This study used a novel method to demonstrate the effect of impact force redirection and introduce the concept of impact force redistribution to reduce brain tissue strain, using helmeted head impacts with an American football helmet. Redirection of impact force is analogous to a boxer moving away from a glancing blow where a portion of the impact force is directed into a tangential plane (around the head), rather than the normal plane, and force transferred to the brain is reduced. Reid (1975) determined that a football player was able to reduce the force transmitted to the head by deflecting an impact and moving out of the direct line of contact.[13] To emulate impact deflection without the player having to avoid or move away from the impact, it is proposed that decoupling the head-helmet interface would accomplish similar results. In addition to redirection, redistribution of the impact force can be attained by lowering the contribution of the dominant coordinate component and forcing re-distribution to one or two of the non-dominant coordinate components. For the context of this study, the dominant coordinate component will be defined as the component that makes the highest contribution to the resultant. Helmet liner strategies, used to redirect and redistribute applied impact force, offer potential to reduce the risk of concussive injury. However, the effect on brain tissue strain must be well understood to offer effective protection and reduced injury risk.[14]

Studies have identified the susceptibility of the brain and resulting injury risk due to changes in brain motion.[15-17] Kendall (2013) compared various sport concussive injury cases, noting larger brain strain values resulted from higher magnitudes of rotational acceleration in the Z directional component.[18] McIntosh (2014) studied Australian Rules football using video analysis and MADYMO reconstructions noting significant differences for the magnitudes of Z component linear acceleration and both the X and Z components of rotational acceleration between concussive and non-injury impacts.[8] Brain motion, which is defined by the X, Y, Z components of acceleration in the studies noted above, is primarily influenced by impact location and direction.[15-17,19-24] In football, most concussive injuries result from player-to-player collisions with impacts to the front (44.7%) and side (22.3%) of the head.[25] This is relevant to injury prevention, as higher injury risk results from impacts to the side of the head than the front.[15-17,19,26] Pellman (2003) studied concussive impacts in the National Football League (NFL) and showed concussion impacts typically occurred in a downward direction and that impact vector affected the direction of brain motion altering concussion tolerance.[27] These studies help to identify the susceptibility of the brain to directional motion and the necessity of understanding changes to brain motion in relation to brain injury to provide adequate protection. However, helmet innovation, particularly technology aimed at impact force redirection to reduce injury risk, must understand the effect of changes to brain motion and the influence on brain tissue strain to improve injury risk levels.[28,29] This study determined the influence of three redistribution and seven redirection strategies on brain tissue strain for 20 different impact conditions; 5 impact locations and 4 impact directions.

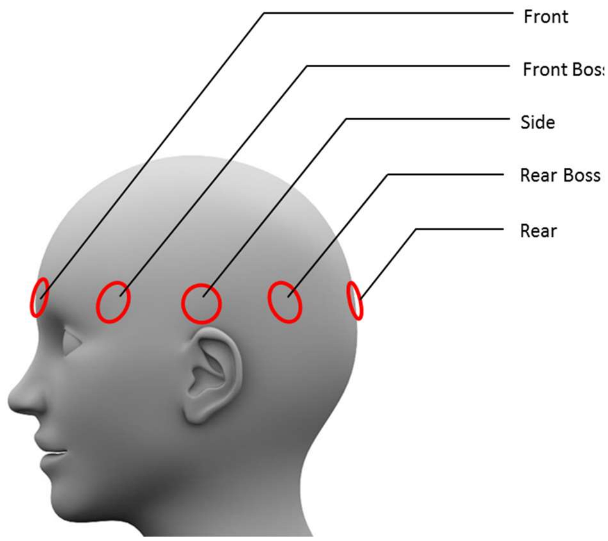
## Methodology

### *Test Procedure*

The study used previously collected impact dynamic response data from research investigating the effect of impact condition on brain deformation. A pneumatic linear impactor, seen in figure 1, was used to impact a helmeted 50th percentile Hybrid III head-form. The helmet used was a standard certified American football helmet comprised of a polycarbonate shell and a layered liner system made up of three-dimensional TPU cushioning and vinyl nitrile covered foam. The full helmet including mask and chin strap had a mass of 2.0 kg (±0.005 kg). Data were collected for twenty impact conditions and at each impact condition, three impact trials were conducted for repeatability. A summary of the 20 impact conditions (5 locations and 4 directions) is presented in figure 2 and 3.

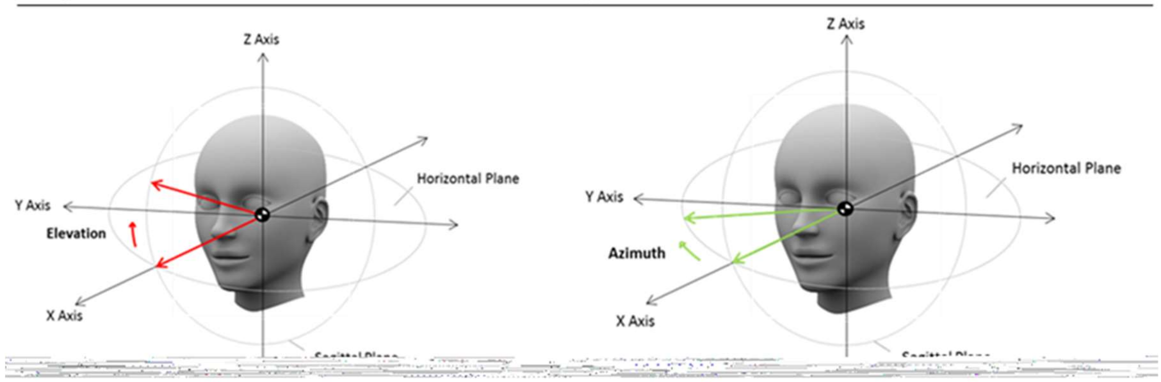


Study 2 - Figure 1: Linear Impactor




Study 2 - Figure 2: Impact Location

| Direction | Definition |
|---|---|
| Center of Gravity | Perpendicular to the head-form, no vertical or horizontal rotation applied |
| Positive Elevation | A 15° elevation of the Hybrid III head and neck-form |
| Positive Azimuth | A 45° rotation of the Hybrid III head and neck-form |
| Negative Azimuth | A -45° rotation of the Hybrid III head and neck-form |




Study 2 - Figure 3: Impact Orientation

The dynamic response data for each original impact (strategy 1) were mathematically altered to represent the effect of 10 different liner strategies, three redirection strategies (strategy 2, 6 and 10) and seven redistribution strategies (strategy 3, 4, 5, 7, 8, 9, and 11). For the purpose of this study, the dominant component of acceleration is defined as the coordinate component that makes the highest contribution to the resultant at each impact as presented in table 1. To modify for the first **redirection** strategy (strategy 2), a twenty percent (20%) reduction was applied to the dominant coordinate component of linear acceleration at each time segment (0.5ms). For the second redirection strategy (strategy 6), a twenty percent (20%) reduction was applied to the dominant coordinate component of rotational acceleration at each time segment (0.5ms). For the third redirection strategy (strategy 10), a twenty percent (20%) reduction was applied to the dominant coordinate component of both linear and rotational acceleration at each time segment (0.5ms). Twenty percent reduction was chosen as this is a value associated with known reductions between helmets designed with linear acceleration strategies and then augmented with a technology to address rotational acceleration.

| Location | LA | RA |
|---|---|---|
| Front CG | X | Y |
| Front NA | X | Z |
| Front PA | X | Z |
| Front PE | X | Y |
| Front Boss CG | Y | X |
| Front Boss NA | X | Y |
| Front Boss PA | Y | Z |
| Front Boss PE | Y | Z |
| Side CG | Y | X |
| Side NA | Y | X |
| Side PA | Y | X |
| Side PE | Y | X |
| Rear Boss CG | Y | Z |
| Rear Boss NA | Y | Z |
| Rear Boss PA | X | Y |
| Rear Boss PE | Y | Z |
| Rear CG | X | Y |
| Rear NA | X | Z |
| Rear PA | X | Z |
| Rear PE | X | Y |

Study 2 - Table 1: Dominant coordinate components for twenty impact conditions

For the first **redistribution** strategy (strategy 3), a twenty percent reduction (20%) was applied to the dominant coordinate component of linear acceleration at each time segment (0.5ms). In addition, the magnitude of the reduction was divided and equally reassigned to both non-dominant axes of linear acceleration. For redistribution strategy 4, a reduction (20%) was applied to the dominant coordinate component of linear acceleration and the magnitude of that reduction was reassigned to the first non-dominant axis of linear acceleration at each time segment (0.5ms). For strategy 5, a reduction (20%) was applied to the dominant coordinate component of linear acceleration and the magnitude of that reduction was divided and reassigned to the second non-dominant axis of linear acceleration at each time segment (0.5ms). Strategy 7, 8, and 9 were conducted the same as strategies 3, 4 and 5 but using the components of rotational acceleration. For strategy 11, a reduction (20%) in the dominant linear and rotational components was applied and divided equally to the two non-dominant axes of linear and rotational acceleration accordingly at each time segment (0.5ms). A summary is shown in table 2 and the applied mathematics for the front centre of gravity trial are attached in Appendix A as an example. The resulting X, Y, Z linear and X, Y, Z rotational acceleration time curves, a total of 660 data files (20 impact conditions *x* 3 trials *x* 1+10 strategies), were input into the finite element brain model and maximum principal strain values were determined.

| Strategy | Concept | Description |
|---|---|---|
| 1 | Original | Original Impact |
| 2 | Redirect | Reduce Dominant Linear Acceleration |
| 3 | Redistribution | Reduce Dominant Linear & Increase Both Non-Dominant Equally |
| 4 | Redistribution | Reduce Dominant Linear & Increase First Non-Dominant |
| 5 | Redistribution | Reduce Dominant Linear & Increase Second Non-Dominant |
| 6 | Redirect | Reduce Dominant Rotational Acceleration |
| 7 | Redistribution | Reduce Dominant Rotational & Increase Both Non-Dominant Equally |
| 8 | Redistribution | Reduce Dominant Rotational and Increase First Non-Dominant |
| 9 | Redistribution | Reduce Dominant Rotational and Increase Second Non-Dominant |
| 10 | Redirect | Reduce Dominant Linear and Rotational Acceleration |
| 11 | Redistribution | Reduce Dominant Linear and Rotational Acceleration and Increase both Non-Dominant Linear and Rotational Acceleration Equally |

Study 2 - Table 2: Description of the redirection and redistribution strategies

***Computational Modeling***

Finite element models of the brain represent a tool for evaluating changes to impact location can influence brain response through comparisons of the computed brain tissue strains.[30] To determine the maximum principal strain (MPS) values for the corresponding trials of each redistribution and redirection strategy, the University College Dublin Brain Trauma Model (UCDBTM) was used.[31,32] Newer and more complex finite element models have been developed and used for head and brain injury research than the UCDBTM.[22,33,34] However, the UCDBTM has been used to process the largest dataset of brain injured, and non-injured cases, and as a result provides a better reference than other models in context to the magnitudes and results for the current research.[35-40]

Computed Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced contour photographs were used to determine the geometry from a single male cadaver.[31] The UCDBTM is constructed of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brainstem. A linearly viscoelastic material model combined with a large deformation theory was chosen to model the brain tissue. The material properties can be found in Tables 3 and 4. The behaviour of the tissue was modeled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus.[31] The shear characteristics of the viscoelastic behaviour were expressed through the formula:

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

Where G is the long-term shear modulus, $G_0$ is the short-term shear modulus and β is the decay factor.[31]

| Material | Young's Modulus (MPa) | Poisson's Ratio | Density (kg/m$^2$) |
|---|---|---|---|
| Scalp | 16.7 | 0.42 | 1000 |
| Cortical bone | 15,000 | 0.22 | 2000 |
| Trabecular bone | 1000 | 0.24 | 1300 |
| Dura | 31.5 | 0.45 | 1130 |
| Pia | 11.5 | 0.45 | 1130 |
| Falx | 31.5 | 0.45 | 1140 |
| Tentorium | 31.5 | 0.45 | 1140 |
| CSF | 0.015 | 0.50 | 1000 |
| Grey matter | Viscoelastic | 0.49 | 1060 |
| White matter | Viscoelastic | 0.49 | 1060 |

Study 2 - Table 3: Material properties used in the University of College Dublin Brain Trauma Model[31]

| | Shear Modulus (kPa) | | Decay Constant ($s^{-1}$) | Bulk Modulus (GPa) |
|---|---|---|---|---|
| | $G_0$ | $G_\infty$ | | |
| Grey matter | 10 | 2 | 80 | 2.19 |
| White matter | 12.5 | 2.5 | 80 | 2.19 |
| Brain stem | 12.5 | 4.5 | 80 | 2.19 |
| White matter | 10 | 2 | 80 | 2.19 |

Study 2 - Table 4: Material properties of the different parts of the brain used in the University College Dublin Brain Trauma Model[32]

The CSF layer was modelled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF, and brain. The model is comprised of 7318 hexahedral elements for the brain, and 2874 hexahedral elements for the CSF layer. This version of the UCDBTM does not include elements representing the cerebral blood vessels. The model was completed using comparisons of intracranial impact cadaveric pressure data from Nahum (1977) and Trosseille (1992) and brain motion from Hardy (2001).[41-43]

***Statistical Analysis***

To determine significant difference between maximum principal strain values generated from the original impact and each of the redirection or redistribution strategies, the mean MPS value of all impacts per strategy was calculated and statistical analysis was run using and using an ANOVA and post hoc Tukey test and adjusting for multiple comparisons ($\alpha<0.05$).

To determine significant difference between maximum principal strain values generated from the original impact and each of the redirection or redistribution strategies at **each location**, the mean MPS value at each location (front, front boss, side, rear boss, and rear) per strategy was calculated and statistical analysis was run using and using an ANOVA and post hoc Tukey test and adjusting for multiple comparisons ($\alpha<0.05$).

To determine significant difference between maximum principal strain values generated from the original impact and each of the redirection or redistribution strategies **by impact condition**, the mean MPS value for each condition (location and direction; centre of gravity, negative azimuth, positive azimuth, and positive elevation) per strategy was calculated and statistical analysis was run using and using an ANOVA and post hoc Tukey test and adjusting for multiple comparisons ($\alpha<0.05$).

## Results

### *Effect of redistribution and redirection strategies on MPS for all impact conditions*

The purpose of this section of the study was to determine if a universal helmet liner strategy designed to redirect or redistribute impact force would significantly reduce maximum principal strain when compared to the original data. The results are summarized in table 5 with significant reductions shaded in green. A significant reduction in MPS occurred from the original (MPS = 0.257) and strategy 6 (MPS = 0.227, p=0.001). Significant reduction in MPS was found between the original (MPS = 0.257) and strategy 10 (MPS = 0.223, p=0.001).

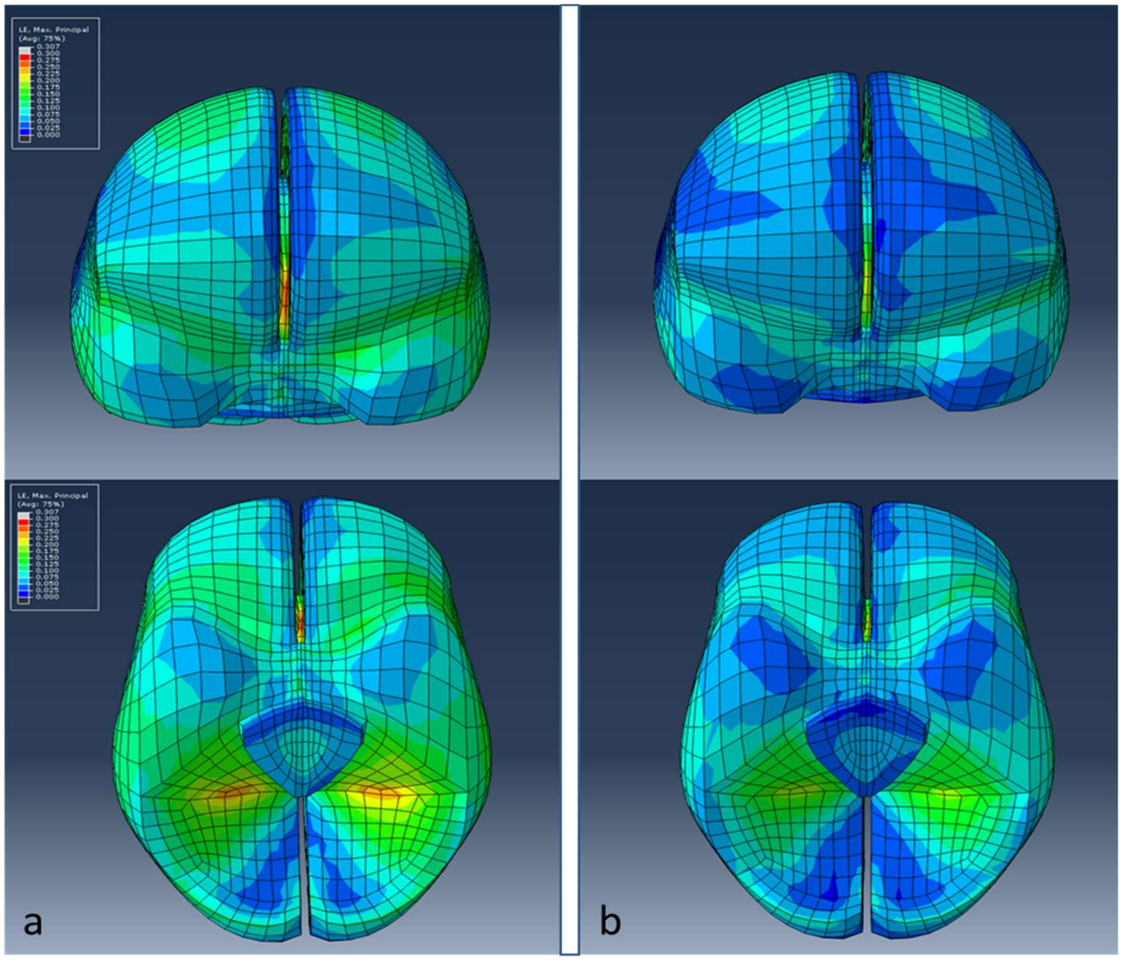
### Effect of redistribution and redirection strategies on MPS for each impact location

The data were analyzed to determine if a redirection or redistribution helmet liner strategy would significantly reduce maximum principal strain values in comparison to the original data set at each impact location (front, front boss, side, rear boss, rear). The results are summarized in table 6 with significant reductions shaded in green. A significant reduction in MPS was found between the original design (0.246) and strategy 6 (0.2182, p=0.003) and strategy 10 (0.216, p= 0.001) for the front boss location. At the side location, significant difference resulted between the original (0.249) and strategy 6 (0.209, p=0.007), strategy 8 (0.215, p=0.042), and strategy 10 (0.210, p=0.010). Additionally, significant difference was found at the rear location between the original (0.242) and strategy 10 (0.211, p=0.034). At the front and rear boss locations, no significant differences in MPS occurred.

### *Effect of redistribution and redirection strategies on MPS for each impact condition*

The data were analyzed to determine if impact direction would significantly lower maximum principal strain values when implementing redirection or redistribution strategies in comparison to the original data set at each of the twenty impact conditions. The results are summarized in table 7 with significant reductions and increases shaded in green and red respectively. A significant reduction in MPS occurred between the original impact and various strategies at all impact locations except for Front CG, Front NA, Front PA, Front Boss PE, Rear Boss NA, Rear Boss PE. An example of the FE output for the Front CG location is shown in figure 4 showing the reduction in MPS between the original (strategy 1) and strategy 10. Significant increase in the MPS value was observed at; the

Front Boss PA location for strategy 8 (p=0.001), the Rear Boss CG (p=0.001), Rear Boss NA (p=0.029) and Rear PA (p=0.002) locations for strategy 9 as indicated by the red shading in table 7.




a: Front CG Strategy 1          b: Front CG Strategy 10

Study 2 - Figure 4: The finite element model output for a Front Centre of Gravity (CG) impact for Strategy 1 (a) and strategy 10 (b)

| Strategy | MPS VALUES | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | **Original** | **Linear Acceleration Strategies** | | | | **Rotational Acceleration Strategies** | | | | **Mixed** | |
| | **1** (SD) | **2** (SD) | **3** (SD) | **4** (SD) | **5** (SD) | **6** (SD) | **7** (SD) | **8** (SD) | **9** (SD) | **10** (SD) | **11** (SD) |
| All Impacts | 0.256 (0.033) | 0.252 (0.034) | 0.249 (0.036) | 0.251 (0.035) | 0.250 (0.036) | 0.226 (0.029) | 0.237 (0.032) | 0.238 (0.034) | 0.243 (0.037) | 0.223 (0.028) | 0.237 (0.032) |

Study 2 - Table 5: MPS values including standard deviation (SD) for each impact strategy (Strategy 2- 11). Green indicates significant reduction from the original (Strategy 1).

| Strategy | MPS VALUES | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | **Original** | **Linear Acceleration Strategies** | | | **Rotational Acceleration Strategies** | | | | | **Mixed** | |
| | **1** (SD) | **2** (SD) | **3** (SD) | **4** (SD) | **5** (SD) | **6** (SD) | **7** (SD) | **8** (SD) | **9** (SD) | **10** (SD) | **11** (SD) |
| Front | 0.282 (0.050) | 0.278 (0.053) | 0.278 (0.058) | 0.279 (0.055) | 0.274 (0.057) | 0.245 (0.031) | 0.254 (0.029) | 0.260 (0.033) | 0.258 (0.027) | 0.238 (0.033) | 0.252 (0.035) |
| Front Boss | 0.246 (0.015) | 0.237 (0.012) | 0.235 (0.013) | 0.238 (0.014) | 0.233 (0.014) | 0.218 (0.015) | 0.233 (0.019) | 0.239 (0.023) | 0.228 (0.021) | 0.216 (0.016) | 0.235 (0.015) |
| Side | 0.249 (0.028) | 0.247 (0.030) | 0.246 (0.026) | 0.248 (0.027) | 0.247 (0.028) | 0.209 (0.019) | 0.222 (0.023) | 0.215 (0.023) | 0.230 (0.025) | 0.210 (0.019) | 0.225 (0.029) |
| Rear Boss | 0.265 (0.031) | 0.262 (0.033) | 0.255 (0.034) | 0.258 (0.036) | 0.262 (0.037) | 0.248 (0.033) | 0.260 (0.040) | 0.257 (0.040) | 0.271 (0.055) | 0.242 (0.039) | 0.255 (0.041) |
| Rear | 0.242 (0.018) | 0.237 (0.015) | 0.232 (0.014) | 0.234 (0.015) | 0.235 (0.013) | 0.214 (0.022) | 0.219 (0.027) | 0.220 (0.029) | 0.234 (0.032) | 0.211 (0.016) | 0.221 (0.028) |

Study 2 - Table 6: MPS values including standard deviation (SD) for each strategy (Strategy 2- 11) at each impact location. Green indicates significant reduction from the original (Strategy 1).

| Strategy | MPS VALUES | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | Original | Linear Acceleration Strategies | | | | Rotational Acceleration Strategies | | | | Mixed | |
| | **1** (SD) | **2** (SD) | **3** (SD) | **4** (SD) | **5** (SD) | **6** (SD) | **7** (SD) | **8** (SD) | **9** (SD) | **10** (SD) | **11** (SD) |
| Front CG | 0.254 (0.014) | 0.251 (0.017) | 0.256 (0.016) | 0.250 (0.017) | 0.247 (0.020) | 0.214 (0.010) | 0.223 (0.018) | 0.216 (0.014) | 0.232 (0.015) | 0.202 (0.007) | 0.215 (0.014) |
| Front NA | 0.274 (0.006) | 0.265 (0.019) | 0.254 (0.017) | 0.269 (0.003) | 0.254 (0.019) | 0.262 (0.009) | 0.265 (0.020) | 0.278 (0.022) | 0.265 (0.001) | 0.250 (0.007) | 0.279 (0.007) |
| Front PA | 0.240 (0.008) | 0.236 (0.007) | 0.233 (0.007) | 0.233 (0.010) | 0.234 (0.009) | 0.224 (0.012) | 0.244 (0.005) | 0.254 (0.005) | 0.239 (0.006) | 0.219 (0.012) | 0.228 (0.013) |
| Front PE | 0.359 (0.030) | 0.360 (0.030) | 0.369 (0.025) | 0.363 (0.031) | 0.363 (0.029) | 0.282 (0.023) | 0.284 (0.027) | 0.292 (0.019) | 0.294 (0.015) | 0.278 (0.019) | 0.287 (0.022) |
| Front Boss CG | 0.236 (0.008) | 0.225 (0.008) | 0.221 (0.012) | 0.224 (0.009) | 0.221 (0.005) | 0.211 (0.006) | 0.223 (0.004) | 0.220 (0.006) | 0.226 (0.003) | 0.206 (0.012) | 0.220 (0.010) |
| Front Boss NA | 0.263 (0.007) | 0.248 (0.011) | 0.247 (0.008) | 0.253 (0.012) | 0.246 (0.014) | 0.206 (0.006) | 0.215 (0.016) | 0.218 (0.017) | 0.198 (0.011) | 0.203 (0.017) | 0.228 (0.009) |
| Front Boss PA | 0.236 (0.004) | 0.234 (0.001) | 0.229 (0.005) | 0.235 (0.001) | 0.223 (0.005) | 0.225 (0.002) | 0.242 (0.003) | 0.256 (0.004) | 0.239 (0.006) | 0.224 (0.004) | 0.243 (0.004) |
| Front Boss PE | 0.250 (0.020) | 0.239 (0.016) | 0.241 (0.010) | 0.241 (0.011) | 0.240 (0.015) | 0.231 (0.021) | 0.252 (0.020) | 0.261 (0.016) | 0.247 (0.017) | 0.229 (0.015) | 0.248 (0.018) |
| Side CG | 0.244 (0.009) | 0.246 (0.007) | 0.242 (0.004) | 0.244 (0.008) | 0.244 (0.007) | 0.199 (0.006) | 0.211 (0.014) | 0.197 (0.007) | 0.227 (0.020) | 0.207 (0.006) | 0.209 (0.015) |
| Side NA | 0.214 (0.003) | 0.206 (0.002) | 0.212 (0.006) | 0.215 (0.003) | 0.211 (0.007) | 0.194 (0.009) | 0.202 (0.005) | 0.199 (0.004) | 0.203 (0.006) | 0.187 (0.005) | 0.202 (0.003) |
| Side PA | 0.288 (0.003) | 0.285 (0.010) | 0.280 (0.012) | 0.284 (0.016) | 0.283 (0.007) | 0.237 (0.014) | 0.254 (0.006) | 0.249 (0.008) | 0.264 (0.007) | 0.237 (0.007) | 0.267 (0.017) |
| Side PE | 0.251 (0.020) | 0.253 (0.001) | 0.251 (0.003) | 0.251 (0.002) | 0.251 (0.008) | 0.205 (0.003) | 0.215 (0.010) | 0.214 (0.004) | 0.226 (0.006) | 0.209 (0.002) | 0.221 (0.006) |
| Rear Boss CG | 0.281 (0.004) | 0.284 (0.010) | 0.276 (0.001) | 0.278 (0.003) | 0.283 (0.006) | 0.269 (0.003) | 0.285 (0.004) | 0.286 (0.008) | 0.306 (0.006) | 0.265 (0.011) | 0.281 (0.004) |

| | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Rear Boss NA | 0.303 (0.004) | 0.299 (0.009) | 0.296 (0.007) | 0.298 (0.010) | 0.302 (0.004) | 0.285 (0.003) | 0.304 (0.006) | 0.297 (0.020) | 0.320 (0.005) | 0.285 (0.004) | 0.299 (0.006) |
| Rear Boss PA | 0.233 (0.003) | 0.224 (0.003) | 0.217 (0.008) | 0.209 (0.011) | 0.210 (0.004) | 0.205 (0.001) | 0.205 (0.007) | 0.206 (0.007) | 0.193 (0.006) | 0.190 (0.011) | 0.198 (0.011) |
| Rear Boss PE | 0.241 (0.014) | 0.240 (0.012) | 0.232 (0.002) | 0.247 (0.004) | 0.253 (0.012) | 0.231 (0.008) | 0.246 (0.007) | 0.238 (0.002) | 0.253 (0.008) | 0.227 (0.004) | 0.242 (0.009) |
| Rear CG | 0.258 (0.009) | 0.241 (0.004) | 0.234 (0.009) | 0.238 (0.013) | 0.230 (0.008) | 0.212 (0.013) | 0.206 (0.010) | 0.210 (0.009) | 0.221 (0.010) | 0.192 (0.009) | 0.193 (0.018) |
| Rear NA | 0.255 (0.011) | 0.254 (0.014) | 0.249 (0.017) | 0.252 (0.012) | 0.254 (0.009) | 0.241 (0.004) | 0.247 (0.002) | 0.245 (0.007) | 0.271 (0.007) | 0.229 (0.002) | 0.257 (0.011) |
| Rear PA | 0.232 (0.006) | 0.233 (0.004) | 0.228 (0.002) | 0.229 (0.003) | 0.232 (0.002) | 0.218 (0.004) | 0.239 (0.005) | 0.244 (0.020) | 0.252 (0.012) | 0.219 (0.004) | 0.233 (0.004) |
| Rear PE | 0.221 (0.002) | 0.219 (0.007) | 0.218 (0.002) | 0.218 (0.003) | 0.224 (0.002) | 0.184 (0.004) | 0.186 (0.004) | 0.180 (0.005) | 0.192 (0.006) | 0.204 (0.007) | 0.201 (0.003) |

Study 2 - Table 7: MPS values including standard deviation (SD) for each strategy (Strategy 2- 11) at each impact condition (location and direction. Green indicates significant reduction and red indicates significant increase from the original (Strategy 1).

## Discussion

### *Effect of redistribution and redirection strategies on MPS when collapsed across all impact conditions*

The data were analyzed first to determine the influence of each liner strategy on maximum principal strain values by looking at the data set as a whole, all impacts combined. This is analogous to a universal helmet liner that is consistent throughout the entirety of the helmet. The results demonstrated strategy 6 (reduce dominant rotational acceleration) and strategy 10 (reduce dominant linear and rotational acceleration) resulted in a 12 and 13 percent reduction in maximum principal strain respectively. Strategy 2 (reduce dominant linear acceleration), did not result in a significant reduction in MPS, indicating the reduction benefit of strategy 10 (reduction of the dominant coordinate components of linear and rotational acceleration) comes primarily from reducing rotational acceleration. The results are consistent with past research demonstrating the correlation between rotational acceleration and brain tissue deformation and noting the lack of correlation between linear acceleration and tissue deformation.[16,19,44-47] The results from this section of the study indicate reductions in brain tissue strain can be accomplished by implementing a universal liner strategy, designed to redirect rotational acceleration, in the American football helmet.

### *Effect of redistribution and redirection strategies on MPS for each impact location*

The results of this study, when analyzed by impact location, indicate reductions in maximum principal strain can be made at the front boss, side and rear impact locations by using redirection and or redistribution liner strategies. Strategy 6 (reduce dominant rotational acceleration) resulted in significant reduction at the front boss and side location, strategy 8 (reduce dominant rotational and increase first non-dominant) generated reduction at the side location, and strategy 10 (reduce dominant linear and rotational acceleration) showed reductions at the front boss, side and rear location. This information can be used to improve helmet design for American football where increased injury risk at specific impact locations is well known, front being the most common followed by the back, side and top region.[7,25,27,48-51] Liao (2016) analyzed head impact data from 33 Division I NCAA football players a significant association ($p = 0.017$) was found between head impact location and concussion.[53] The concussive impact group sustained a lower frequency of impacts to the front and more to the sides and top of the head. Increasing protection in impact location/regions that are more susceptible to injury may offer reduced risk of concussive injury. The results of this section of the study indicate under the current conditions, site-specific improvements can be made at three impact locations; the front boss, side and rear locations through redirection and/or redistribution strategies. Similar analysis should be conducted under different impact conditions and for various sporting helmets to aid helmet designers to incorporate location specific redirection and redistribution strategies to help reduce brain tissue strain.

***Effect of redistribution and redirection strategies on MPS for each impact condition***

The results of this section of the study, when analyzed to include impact direction, demonstrate that both reductions and increases in maximum principal strain can occur and are dependent on the impact condition and liner strategy. The increase in MPS under certain loading conditions may be due to the directional sensitivity of the brain which is well documented.[8,21,22,53,54] but is unclear within this study, warranting further investigation. In three of the four occurrences where increased MPS occurred (Front Boss PA, strategy 8; rear boss CG, strategy 9; and rear boss NA, strategy 9), the dominant component of rotational acceleration was about the Z axis. This component was reduced and redistributed to the component about the X axis. The fourth case of increased MPS was the Rear PA, strategy 9 where, again, the dominant coordinate component of rotational acceleration was about the Z axis but the redistribution was made to the Y component of rotational acceleration. In these four impact conditions, the dominant coordinate component vector is a similar magnitude to one or both of the non-dominant components creating unclear impact force dominance. This data supports previous research indicating the directional sensitivity of the brain but requires further study to fully understand the implications of redirection and redistribution at impact sites where there is no clear directional dominance.[8,18] Although MPS correlates with peak resultant, redistribution of rotational acceleration away from the peak resultant, towards complementary axes may still increase MPS due to the directional sensitivity of the brain. This is an important result, as technologies that consider only peak resultant may fail to recognize increases in MPS.

In American football, the numerous combinations of impact location and direction vary significantly making it difficult to protect against all possible impact scenarios.[30,48,55,56] It is therefore, important to understand that under certain impact conditions, helmet liner strategies implemented to reduce injury risk may in fact increase risk. Redirection and redistribution strategies offer a potential to reduce brain tissue strain and decrease injury risk but warrant comprehensive investigation to determine the influence on maximum principal strain before being implemented.

**Limitations**

The results of this study are presented within the context of certain limitations. The impacts to the helmeted hybrid III headform are conducted 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 helmet liner strategies and the resulting mechanics. This study was conducted using on model of American football helmet. Experimental data obtained using other football helmet models and helmets from other sports would further add to the context of this study.

While the finite element 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, including impacts for performance evaluation of American football helmets, giving relative values rather than an absolute.[8,37,43,57] Due to the directional nature of the induced strain by varying impact location and direction, the results may vary if a directional anisotropic model is implemented.

**Conclusion**

This study utilized a methodology to analyze impact force redirection and redistribution strategies that can be used by American football helmet designers. The results of the study demonstrated redirection and redistribution strategies that reduce the dominant coordinate component of rotational acceleration can be used to reduce maximum principal strain in American football helmets.

**Data Availability**: The raw/processed data required to reproduce these findings can be requested through contact with the corresponding author.

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**Conflict of interest:** The authors declare that there are no conflicts of interest.

## REFERENCES


1. **Pellman EJ**, Viano DC, Withnall C, et al. Concussion in professional football: Helmet testing to assess impact performance – Part 11. J. Neurosurgery. 2006; 58:78 – 96.
2. **Viano** DC, Halstead D. Change in Size and Impact Performance of Football helmets from the 1979's to 2010. Annals of Biomed. Eng. 2012; 40(1):175-184. DOI: 10.1007/s10439-011-0395-1
3. **Post A,** Oeur A, Walsh E. et al. A centric/non-centric impact protocol and finite element model methodology for the evaluation of American Football Helmets to Evaluate Risk of Concussion. Computer Methods in Biomechanics and Biomedical Engineering 2013;17(16):1-16.
4. **Langlois JA**, Rutland-Brown W, Wald MM. The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview. J. Head Trauma Rehab. 2006; **21**(5):375-378.
5. **McCrea M,** Hammeke T, Olsen G, et al. Unreported concussion in high school football players: implications for prevention. Clin J Sports Med. 2004; 14(1):13-17
6. **Collins CL,** Comstock RD, Dick RW, et al. Concussions among United States high school and collegiate athletes. J Athletic Train. 2007; 42(4):495-503.
7. **Crisco JJ,** Fiore R, Beckwith JG, et al. Frequency and location of head impact exposures in individual collegiate football players. J Athletic Train. 2010; 45(6): 549-559.
8. **McIntosh AS,** Patton DA, Frechede B. et al. The Biomechanics of Concussion in Un-helmeted Football Players in Australia: A Case-Control Study. Biomedical Journal 2014; 4:1–9.
9. **Cournoyer J,** Post A, Rousseau P, Hoshizaki TB. The Ability of American Football Helmets to Manage Linear Acceleration with Repeated High-Energy Impacts. J. Athletic Train. 2016; 51(3):258-263.
10. **Fowler JN,** Pallanta AA, Swanik CB. et al. The use of shear thickening nanocomposites in impact resistant materials. J. Biomed. Eng. 2015; 137: 1-5.
11. **Aare M,** Kleiven S, Halldin P. Injury Criteria for Oblique Helmet Impacts. International Research Council on Biomechanics of Injury Conference (IRCOBI), Dublin, Ireland. 2003:349-350.
12. **Finan JD,** Nightingale RW, Myers BS. The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts. Traffic Injury Prevention. 2008; **9**(5):483–488. DOI: 10.1080/15389580802272427
13. **Reid SE,** Herbert ME, Louis MW. et al. Physiologic Response to Impact. J Trauma. 1975; 15(2):150-152.
14. **Margulies SS,** Thibault LE. A proposed tolerance criteria for diffuse axonal injury. J. Biomech. 1992; 25:917- 923.
15. **Gennarelli TA,** Thibault LE, Adams JH. et al. Diffuse Axonal Injury and Traumatic Coma in the Primate. Annals of Neurology. 1982; 12(6):564-574. DOI: 10.1002/ana.410120611
16. **Gennarelli TA,** Thibault LE, Tomei G. Directional Dependence of Axonal Brain Injury Due to Centroidal and Non-centroidal Acceleration. 31st Stapp Car Crash Conference. 1987. SAE paper no. 8721`97. Society of Automotive Engineers.

17. **Hodgson VR,** Thomas LM, Khalil TB. The Role of Impact Location in Reversible Cerebral Concussion. 27th Stapp Car Crash Conf. SAE paper no. 831618. Society of Automotive Engineers.1983; 225–240.
18. **Kendell M,** Post A, Zanetti K. et al. Comparison of Dynamic Versus Static Head Impact Reconstruction Methodology by Means of Dynamic Impact Response and Brain Deformation Metrics. International Research Council on Biomechanics of Injury Conference (IRCOBI). 2013. Gothenberg, Sweeden.
19. **Zhang L.** Computational Biomechanics of Traumatic Brain Injury, An Investigation of Head Impact Response and American football head injury. Doctor of Philosophy – Dissertation. 2001. Graduate School of Wayne State University, Detroit, Michigan.
20. **Bailes JE,** Hudson V. Classification of Sport-Related Head Trauma: A Spectrum of Mild to Severe Injury. J Athletic Train. 2001; 36(3):236-243.
21. **Kleiven S.** Influence of Impact Direction on Human Head in Prediction of Subdural Hematoma. J. Neurotrauma 2003; 20(4):365-379.
22. **Giordano C,** Cloots RJH, van Dommelen JAW. Et al. The Influence of Anisotropy on Brain Injury Prediction. J. Biomech. 2014; 47:1052-1059.
23. **Ahmadisoleymani SS,** Yang J. American Football Helmet for Preventing Concussion, A Literature Review. Procedia Manufacturing. 2015; 3:3796-3803. DOI: 10.1016/j.promfg.2015.07.882
24. **Breedlove KM,** Breedlove E, Nauman E. et al. The Ability of an Aftermarket Helmet Add-on Device to Reduce Impact-Force Accelerations During Drop Tests. J. Athletic Train. 2017; 52(9):802-808. DOI: 10.4085/1062-6050-52.6.01
25. **Kerr ZY,** Collins CL, Mihalik JP. Et al. Impact Locations and Concussion Outcomes in High School Football Player-to-Player Collisions. Pediatrics 2014; 134(3):489-496.
26. **Tarriere C.** Relationship Between Experimental Measuring Techniques and Real World Accidents. Head Injury Symposium. 1987. New Orleans, AAAM Report.
27. **Pellman EJ,** Viano DV, Tucker AM. Et. al. Concussion in Professional Football: Location and Direction of Helmet Impacts – Part 2. Neurosurgery 2003; 53:1328-1341.
28. **Meaney DF,** Smith DH. Biomechanics of Concussion. Clin Sports Med 2011; 30:19-31. DOI: 10.1016/j/csm.2010.08.009.
29. **Clark MD,** Breton MA, Marshall SW. et al. Descriptive Characteristics of Concussion in National Football League Games. American Journal of Sports Medicine. 2017; 45(4):929-936.
30. **Elkin BS,** Gabler LF, Panzar MB. Et al. Brain tissue strains vary with head impact location: A possible explanation for increased concussion risk in struck versus striking football players. Clinical Biomechanics. 2018. [In Press]
31. **Horgan TJ**, Gilchrist MD. The creation of 3D finite element models for simulating head impact biomechanics. Int. J. Crashworthiness. 2003; 8(4): 353 – 366.
32. **Horgan TJ.** A Finite Element Model of The Human Head for Use in the Study of Pedestrian Accidents. Doctor of Philosophy Thesis. Department of Mechanical Engineering, Faculty of Engineering and Architecture. University of College Dublin, February 2005.
33. **Sahoo D,** Deck C, Willinger R. Brain injury tolerance limit based on computation of axonal strain. Acc Anal Prev. 2016; 92:53-70.

34. **Takhounts EG,** Craig MJ, Moorhouse K, et al. Development of brain injury criteria (BrIC). Stapp Car Crash J 2013; 57:243-266.
35. **Hoshizaki TB,** Post A, Kendall M, et al. The development of a threshold curve for the understanding of concussion in sport. Trauma 2017; 19(3):196-206.
36. **Kendall M.** Comparison and Characterization of Different Concussive Brain Injury Events. PhD thesis, University of Ottawa, Canada, 2016.
37. **Oeur RA,** Karton C, Post A. et al. A Comparison of Head Dynamic Response and Brain Tissue Stress and Strain Using Accident Reconstruction for Concussion, Concussion with Persistent Postconcussive Symptoms and Subdural Hematoma. J. Neurosurg. 2015; 123(2):415-422.
38. **Post A,** Hoshizaki TB. Rotational acceleration, brain tissue strain, and the relationship to concussion. J. Biomech. Eng. 2015; 137:1-8.
39. **Post A,** Clark JM, Hoshizaki TB, et. al. The effect of acceleration signal processing for head impact numeric simulations. Sport Eng. 2016; 20(2):111-119.
40. **Post A,** Kendall M, Cournoyer J, et al. Falls resulting in mild traumatic brain injury and focal traumatic brain injury: A biomechanical analysis. Int J Crash: 2017 Doi: 10.1080/13588265.2017.1316048.
41. **Nahum, A.M.,** Smith, R., Ward, C.C., 1977. Intracranial pressure dynamics during head impact. 21st Annual Stapp Car Crash Conference. Society of Automotive Engineers.
42. **Hardy WN,** Foster CD, Mason MJ, et al. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar x-ray. Stapp Car Crash Journal; The Stapp Association, Ann Arbor, Michigan. 2001.
43. **Trosseille X,** Tarrie´re C, Lavaste F. et al. Development of a F.E.M. of the human head according to a specific test protocol. 36th Stapp Car Crash Conference. 1992. Seattle, WA, USA, SAE paper 922527.
44. **Gennarelli TA**, Ommaya AK, Thibault LE. Comparison of translational and rotational head motions in experimental cerebral concussion. 15th Stapp Car Crash Conference Proceedings, 17-19 November 1971, Coronado, Calif. 797-803. New York: Society of Automotive Engineers.
45. **Forero Rueda** MA, Cui L, Gilchrist MD. Finite Element Modeling of Equestrian Helmet Impacts Exposes the Need to Address Rotational Kinematics in Future Helmet Designs. Computer Methods in Biomechanics and Biomedical Engineering. 2010; 14(12):1021-1031. DOI: 10.1080/10255842.2010.504922
46. **Walsh ES,** Rousseau P, Hoshizaki TB. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. Sports Eng. 2011; 13(3):135-143.
47. **Post A,** Hoshizaki TB, Gilchrist MD, Brien S, Cusimano MD, Marshall S: Traumatic brain injuries: The influence of the direction of impact. Neurosurg 2015; 76(1):81-91.
48. **Pellman EJ,** Viano DC, Tucker AM, et al. Concussion in professional football: reconstruction of game impacts and injuries. J. Neurosurgery. 2003; 53(4):799-814.
49. **Viano DC,** Casson IR, Pellman EJ, et al. Concussion in professional football comparison with boxing head impacts. Neurosurgery. 2005; 57(6): 1154-1172.

50. **Greenwald RM,** Gwin JT, Chu JJ, et al. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 2008; 62(4):789–797.
51. **Rowson S,** et al. Linear and Angular Head Acceleration Measurements in Collegiate Football. J. Biomech. Eng. 2009; 131(6):7. DOI: 10.1115/1.3130454
52. **Liao S,** Lynall RC, Mihalik JP. The effect of head impact location on day of diagnosed concussion in college football. Med. Sci. Sports Exercise. 2016; 48(7):1239-1243
53. **Duma SM,** Manoogian SJ, Bussone WR. Analysis of real-time head accelerations in collegiate football players. Clin. J. Sports Med. 200; 15(1):3-8.
54. **Post A**, Oeur A, Hoshizaki TB, et al. The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research. International Research Council on Biomechanics of Injury Conference (IRCOBI), 2012 Dublin, Ireland.
55. **Broglio SP,** Surma T, Ashton-Miller JA. High school and collegiate football athlete concussions: a biomechanical review. Annals of Biomed. Eng. 2012; 40(1): 37-46. DOI: 10.1007/s10439-011-0396-0
56. **Beckwith JG,** Zhao W, Songbai J. et al. Estimated Brain Tissue Response Following Impacts Associated with and without Diagnosed Concussion. Annals of Biomedical Engineering. 2018. [Published online February 2018]
57. **Darling T,** Muthuswamy J, Rajan SD. Finite Element Modeling of Human Brain Response to Football Impacts. Computer Methods in Biomechanics and Biomedical Engineering. 2016; 19(13):1432-1442. DOI: 10.1080/10255842.2016.1149574

**Appendix A**

An example of the mathematics applied for the Front Centre of Gravity impact. The dominant coordinate component of linear acceleration is aX. The dominant coordinate component of rotational acceleration is αY.

| | | |
|---|---|---|
| Strategy 1 | Original | $aX_1 = aX_1$ |
| | | $aY_1 = aY_1$ |
| | | $aZ_1 = aZ_1$ |
| | | $aR_1 = (aX_1 + aY_1 + aZ_1)^{1/2}$ |
| | | $\alpha X_1 = \alpha X_1$ |
| | | $\alpha Y_1 = \alpha Y_1$ |
| | | $\alpha Z_1 = \alpha Z_1$ |
| | | $\alpha R_1 = (\alpha X_1 + \alpha Y_1 + \alpha Z_1)^{1/2}$ |
| Strategy 2 | Re-Direction (Linear Acceleration) | $aX_2 = aX_1 * 0.8$ |
| | | $aY_2 = aY_1$ |
| | | $aZ_2 = aZ_1$ |
| | | $aR_2 = (aX_2 + aY_2 + aZ_2)^{1/2}$ |
| | | $\alpha X_2 = \alpha X_1$ |
| | | $\alpha Y_2 = \alpha Y_1$ |
| | | $\alpha Z_2 = \alpha Z_1$ |
| | | $\alpha R_2 = (\alpha X_2 + \alpha Y_2 + \alpha Z_2)^{1/2}$ |
| Strategy 3 | Re-Distribution (Linear Acceleration) | $aX_3 = aX_1 * 0.8$ |
| | | $aY_3 = \text{IF } (aY_1 < 0, (aY_3 – (\text{ABS}((aX_1 – aX_3)/2))), (aY_3 + (\text{ABS}((aX_1 – aX_3)/2))))$ |
| | | $aZ_3 = \text{IF } (aZ_1 < 0, (aZ_3 – (\text{ABS}((aX_1 – aX_3)/2))), (aZ_3 + (\text{ABS}((aX_1 – aX_3)/2))))$ |
| | | $aR_3 = (aX_3 + aY_3 + aZ_3)^{1/2}$ |
| | | $\alpha X_3 = \alpha X_1$ |
| | | $\alpha Y_3 = \alpha Y_1$ |
| | | $\alpha Z_3 = \alpha Z_1$ |
| | | $\alpha R_3 = (\alpha X_3 + \alpha Y_3 + \alpha Z_3)^{1/2}$ |
| Strategy 4 | Re-Distribution (Linear Acceleration) | $aX_4 = aX_1 * 0.8$ |
| | | $aY_4 = \text{IF } (aY_1 < 0, (aY_4 – (\text{ABS}(aX_1 – aX_3))), (aY_4 + (\text{ABS}(aX_1 – aX_3))))$ |
| | | $aZ_4 = aZ_1$ |
| | | $aR_4 = (aX_4 + aY_4 + aZ_4)^{1/2}$ |
| | | $\alpha X_4 = \alpha X_1$ |
| | | $\alpha Y_4 = \alpha Y_1$ |
| | | $\alpha Z_4 = \alpha Z_1$ |
| | | $\alpha R_4 = (\alpha X_4 + \alpha Y_4 + \alpha Z_4)^{1/2}$ |
| Strategy 5 | Re-Distribution (Linear Acceleration) | $aX_5 = aX_1 * 0.8$ |
| | | $aY_5 = aY_1$ |
| | | $aZ_5 = \text{IF } (aZ_1 < 0, (aZ_5 – (\text{ABS}(aX_1 – aX_5))), (aZ_5 + (\text{ABS}((aX_1 – aX_5))))$ |
| | | $aR_5 = (aX_5 + aY_5 + aZ_5)^{1/2}$ |
| | | $\alpha X_5 = \alpha X_1$ |
| | | $\alpha Y_5 = \alpha Y_1$ |
| | | $\alpha Z_5 = \alpha Z_1$ |
| | | $\alpha R_5 = (\alpha X_5 + \alpha Y_5 + \alpha Z_5)^{1/2}$ |
| Strategy 6 | Re-Direction (Rotational Acceleration) | $aX_6 = aX_1$ |
| | | $aY_6 = aY_1$ |
| | | $aZ_6 = aZ_1$ |
| | | $aR_6 = (aX_6 + aY_6 + aZ_6)^{1/2}$ |

| | | |
|---|---|---|
| | | $\alpha X_6 = \alpha X_1$ |
| | | $\alpha Y_6 = \alpha Y_1 * 0.8$ |
| | | $\alpha Z_6 = \alpha Z_1$ |
| | | $\alpha R_6 = (\alpha X_6 + \alpha Y_6 + \alpha Z_6)^{1/2}$ |
| | Re-Distribution (Rotational Acceleration) | $aX_7 = aX_1$ |
| | | $aY_7 = aY_1$ |
| | | $aZ_7 = aZ_1$ |
| | | $aR_7 = (aX_7 + aY_7 + aZ_7)^{1/2}$ |
| | | $\alpha X_7 = IF\ (\alpha X_1 < 0, (\alpha X_7 - (ABS((\alpha Y_1 - \alpha Y_7)/2))), (\alpha X_7 + (ABS((\alpha Y_1 - \alpha Y_7)/2))))$ |
| | | $\alpha Y_7 = \alpha Y_1 * 0.8$ |
| | | $\alpha Z_7 = IF\ (\alpha Z_1 < 0, (\alpha Z_7 - (ABS((\alpha Y_1 - \alpha Y_7)/2))), (\alpha Z_7 + (ABS((\alpha Y_1 - \alpha Y_7)/2))))$ |
| | | $\alpha R_7 = (\alpha X_7 + \alpha Y_7 + \alpha Z_7)^{1/2}$ |
| Strategy 8 | Re-Distribution (Rotational Acceleration) | $aX_8 = aX_1$ |
| | | $aY_8 = aY_1$ |
| | | $aZ_8 = aZ_1$ |
| | | $aR_8 = (aX_8 + aY_8 + aZ_8)^{1/2}$ |
| | | $\alpha X_8 = IF\ (\alpha X_1 < 0, (\alpha X_8 - (ABS(\alpha Y_1 - \alpha Y_8))), (\alpha X_8 + (ABS(\alpha Y_1 - \alpha Y_8))))$ |
| | | $\alpha Y_8 = \alpha Y_1 * 0.8$ |
| | | $\alpha Z_8 = \alpha Z_1$ |
| | | $\alpha R_8 = (\alpha X_8 + \alpha Y_8 + \alpha Z_8)^{1/2}$ |
| Strategy 9 | Re-Distribution (Rotational Acceleration) | $aX_9 = aX_1$ |
| | | $aY_9 = aY_1$ |
| | | $aZ_9 = aZ_1$ |
| | | $aR_9 = (aX_9 + aY_9 + aZ_9)^{1/2}$ |
| | | $\alpha X_9 = \alpha X_1$ |
| | | $\alpha Y_9 = \alpha Y_1 * 0.8$ |
| | | $\alpha Z_9 = IF\ (\alpha Z_1 < 0, (\alpha Z_9 - (ABS(\alpha Y_1 - \alpha Y_9))), (\alpha Z_9 + (ABS(\alpha Y_1 - \alpha Y_9))))$ |
| | | $\alpha R_9 = (\alpha X_9 + \alpha Y_9 + \alpha Z_9)^{1/2}$ |
| Strategy 10 | Re-Direction (Linear and Rotational Acceleration) | $aX_{10} = aX_1 * 0.8$ |
| | | $aY_{10} = aY_1$ |
| | | $aZ_{10} = aZ_1$ |
| | | $aR_{10} = (aX_{10} + aY_{10} + aZ_{10})^{1/2}$ |
| | | $\alpha X_{10} = \alpha X_1$ |
| | | $\alpha Y_{10} = \alpha Y_1 * 0.8$ |
| | | $\alpha Z_{10} = \alpha Z_1$ |
| | | $\alpha R_{10} = (\alpha X_{10} + \alpha Y_{10} + \alpha Z_{10})^{1/2}$ |
| Strategy 11 | Re-Distribution (Linear and Rotational Acceleration) | $aX_{10} = aX_1 * 0.8$ |
| | | $aY_{10} = IF\ (aY_1 < 0, (aY_{10} - (ABS((aX_1 - aX_{10})/2))), (aY_{10} + (ABS((aX_1 - aX_{10})/2))))$ |
| | | $aZ_{10} = IF\ (aZ_1 < 0, (aZ_{10} - (ABS((aX_1 - aX_{10})/2))), (aZ_{10} + (ABS((aX_1 - aX_{10})/2))))$ |
| | | $aR_{10} = (aX_{10} + aY_{10} + aZ_{10})^{1/2}$ |
| | | $\alpha X_{10} = IF\ (\alpha X_1 < 0, (\alpha X_{10} - (ABS((\alpha Y_1 - \alpha Y_{10})/2))), (\alpha X_{10} + (ABS((\alpha Y_1 - \alpha Y_{10})/2))))$ |
| | | $\alpha Y_{10} = * 0.8$ |
| | | $\alpha Z_{10} = IF\ (\alpha Z_1 < 0, (\alpha Z_{10} - (ABS((\alpha Y_1 - \alpha Y_{10})/2))), (\alpha Z_{10} + (ABS((\alpha Y_1 - \alpha Y_{10})/2))))$ |
| | | $\alpha R_{10} = (\alpha X_{10} + \alpha Y_{10} + \alpha Z_{10})^{1/2}$ |

## Study 3

### Influence of liner technology on maximum principal strain in American football helmets

**Abstract**

Impact attenuating liner technologies used in American football helmets are proficient at mitigating linear acceleration but are not designed explicitly to manage factors associated with concussive injury risk. The material properties of brain tissue create vulnerability to rotational head motion. This leads to shear strain in the brain tissue and is associated with the mechanism of concussive injury risk. The correlation between brain tissue strain and concussion has led researchers to study methods of reducing rotational head and brain motion rather than simply attenuating linear acceleration. This has led to the concept of "decoupling" liner strategies that are designed to separate the head-helmet interface. Decoupling strategies allow for the helmet to move independently from the head, to a degree, resulting in reduced head motion and a decrease in brain tissue strain.

This study used a linear impactor to evaluate liner strategies in American football helmets in their effectiveness to reduce maximum principal strain. In comparison to a certified commercially available helmet, a helmet built using novel design principles to create decoupling at the head-helmet interface and redirect impact energy resulted in a 19% and 17% reduction in the peak resultant and peak dominant component of rotational acceleration and a 16% reduction in maximum principal strain. The results demonstrate a decoupling liner technology reduced rotational acceleration and decreased tissue strain which can help improve impact management in American football helmets.

**Null Hypothesis**: No significant differences in maximum principal strain between the four prototype helmets (systems) and the original manufactures helmet.

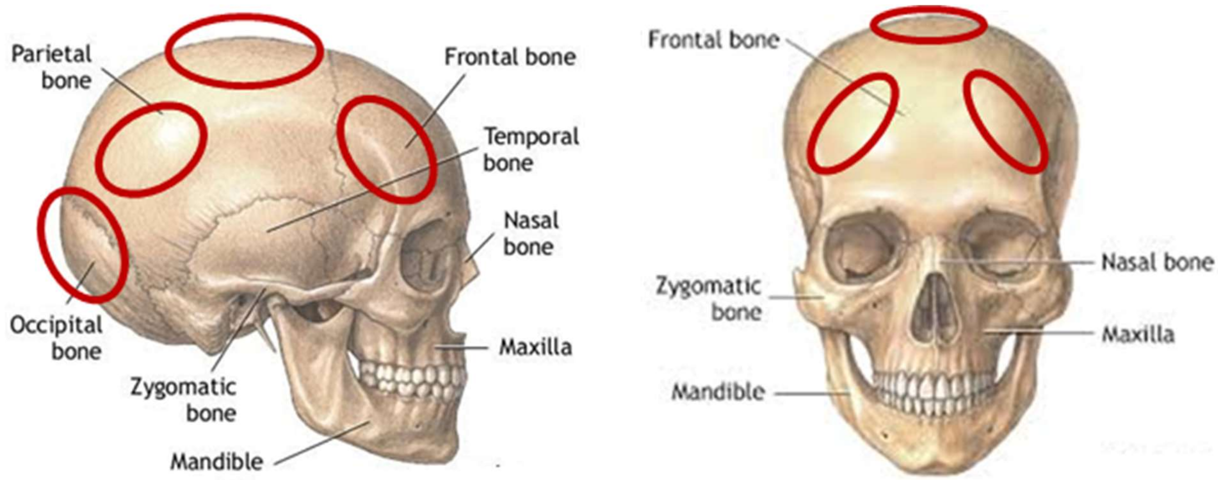
**Key words:** Brain trauma, dominant coordinate component, decoupling, helmet design

**Introduction:**

Despite the use of helmets in American football, the rate of concussion remains high indicating improvements to existing helmet designs are required to further reduce the risk of injury to players [McCrea 2004, Collins 2007, Crisco 2010, McIntosh 2014, Cournoyer 2016]. The liner of a current American football helmet provides most of the impact force mitigation and therefore offers opportunity for improved design innovation [Spyrou 2000, Rousseau 2009, Caserta 2011, Post 2013]. Current liner technologies, typically recoverable foams such as Expanded Polypropylene (EPP) and Vinyl Nitrile (VN) or engineered 3D structures, are design to mitigate large magnitudes of linear acceleration which is managed by material thickness and compliance [Gimble 2008, Lamb 2009, Post 2015]. Material thickness, primarily provided by the helmet liner, helps to reduce peak magnitude of force using the stopping distance "d". This is more easily seen by the equation $F = mv^2/2d$ where "d" is the distance [Mills 2003, Lamb 2009]. The current offset (helmet liner thickness) for standard American football helmets is 1.5 to 2 inches but is limited by comfort constraints and practicality of use [Newman 1993, Hoshizaki 2004]. Therefore, research efforts looking to reduce acceleration magnitude further through liner compliance have investigated modifying material stiffness or density [Rousseau 2009, Lamb 2009]. Materials that are too soft can "bottom out" becoming fully compressed, limiting the distance allowable before sufficient acceleration force is managed [Post 2011]. This results in delayed but sudden spikes in peak acceleration magnitudes, known to increase the risk of injury [Post 2014a]. Stiff, dense materials reduce the distance for acceleration to occur and lead to rapid elevated peaks of linear and rotational acceleration, indicating the force has exceeded the material's critical threshold before being sufficiently managed [Gilchrist 2003, Song 2005, Post 2011]. Compliance variation, created by varying stiffness and thickness, results in changes to impact duration, which has also been shown to effect maximum principal strain MPS) values [Kendall 2016]. High compliance (softer, less dense or thicker materials) elongates impact duration, increasing brain tissue strain [Hoshizaki 2013].

The material constraints (thickness and compliance) of current liner technologies and the link between rotational acceleration, tissue strain and concussive injury have led researchers to investigate different liner technologies, in particular, decoupling strategies. Strategies to "decouple" the helmet from the head allow separation and promote the helmet to move independently from the head [Post 2014]. It has been proposed that decoupling strategies "redirect" or "redistribute impact force from the normal component into the tangential component, reducing the transfer of force into the brain [Aare 2003, Finan 2008]. Halldin (2000) subjected spherical helmets to near-tangential impacts from a pendulum test to show that a decoupling mechanism could reduce rotational acceleration up to 30%. Johnston (2015) conducted preliminary work using a material-by-design approach to use liner technology and a rotation specific impact rig to reduce Z-axis rotational acceleration for side impacts. These studies provided an understanding of the influence of the helmet on brain motion and demonstrated that helmet technology can be used to manage brain motion, reduce brain tissue strain, and decrease injury risk [Halldin 2000, Aare 2003, Finan 2008, Johnston 2015].

Conventional helmet liner designs and recent decoupling designs cradle and cover the entire surface of the head. This is done primarily for comfort and load distribution. However, to optimize decoupling performance the strategy proposed in this study is accomplished by following a different design principle. The principle is to create some void space in the helmet liner to prevent coupling of the head and helmet at high ridge points of the human head. The skull is ellipsoidal in shape and does not often fit symmetrically within a round football helmet creating non-uniform contact with between the head and helmet at the high point regions of the head. Allowing for voids in the liner at these regions removes the high point contact areas between the head and helmet interface that could hinder release and couple the head and helmet. Typically, this occurs at; the crown of the head, the occiput, and the four corners of the head (the two high temporal regions, and the two rear corners where the parietal, occipital, and temporal lobes meet). Therefore, placement of oil filled bladders inside the helmet should articulate with the head in common high risk impact areas while avoiding high points to create release and separation of the head and helmet. The concept of increased coupling due to the high point regions of the head creating greater contact area with the helmet is a hypothesized theory derived from laboratory impact experience which requires future investigation for verification. This can be shown in figure 1. The purpose of this paper is to demonstrate the effectiveness of different liner strategies to reduce maximum principal strain, two strategies designed predominatly to reduce linear accleraiton, two to address linear and rotational acceleration, and a novel design to decouple the head and helmet.




Study 3 - Figure 1: Regions of high points on the head where contact occurs with the helmet and requires decoupling.

**Methodology**

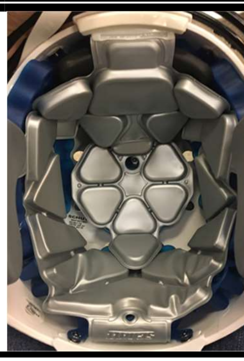
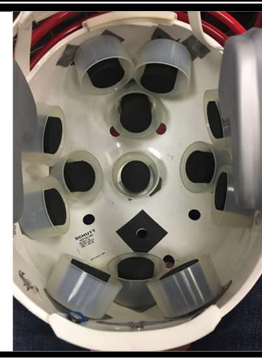
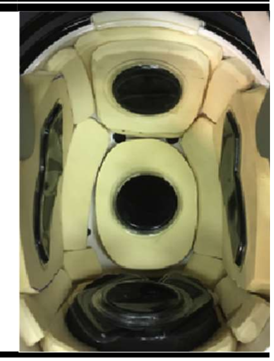
**Test procedure**

This study investigated five different energy management strategies to reduce maximum principal strain in comparison to a commercial helmet. The construction of each helmet was completed in the University of Ottawa Neurotrauma Science Impact Laboratory (NISL), shown in table 1. The effect of helmet design on maximum principal strain was evaluated by impacting a helmeted 50th percentile Hybrid III head-form at 6.0 m/s with a pneumatic linear impactor, shown in figure 2. The test velocity was 6.0 m/s, chosen as it falls within the range of helmet-to-helmet impact velocities occurring in amateur and professional football and is the impact velocity used in helmet standard testing [Pellman 2003, Bartsch 2012, Jadischke 2013, Post 2013, NOCSAE 2004, NOCSAE 2014]. For each helmet utilizing a different liner strategy, impacts were conducted at twenty conditions (5 locations and 4 directions). For each helmet, at each impact condition, three trials were collected for reliability purposes. This resulted in a total of 60 impact trials per helmet and a total of 300 (20 impact conditions *x* 3 trials *x* 5 helmets) dynamic response data files. The dynamic response data files, each containing the X, Y, Z linear and X, Y, Z rotational acceleration time curves, were input into the finite element brain model and maximum principal strain values were determined.

*Liner Structures*

The elements of the five helmets are detailed below.

- **Vinyl Nitrile** foam is common foam utilized within existing American football helmet design and is used primarily for linear acceleration attenuation [Viano 2012, Krzeminski 2016]. Layers of VN 600 and/or VN 1000 foam were cut and placed in the helmets to match the offset of the commercial helmet.
- **3D Engineered Structures - TPU pod and VN600 foam insert structures.** Made of Thermoplastic Urethane (TPU), the pod structure is designed to compress upon impact to control the rate of linear acceleration [Lamb 2009]. Inside the TPU pod is an engineered Vinyl Nitrile (VN 600) foam inserts help to mitigate high-energy impacts and allow for a greater functional range to mitigate linear impact energy [Gimble 2008, Lamb 2009, Post 2012a].
- **The bladder** is fluid (oil) filled with an ethylene-vinyl acetate (EVA) foam insert designed to be used alone to decouple the head-helmet or in conjunction with the TPU structure to decouple the pod from the head-helmet. The derivation of the bladder design has come from iterations of decoupling technologies developed and tested within the Neurotrauma Impact Science Laboratory at the University of Ottawa.
- **Decoupling design principle.** Instead of maintaining the same array of pod or bladder placements, the bladders were placed at typical impact locations and avoiding the high points on the human head. This placement allows for decoupling to occur and to avoid where there would naturally be the greatest contact between the head and helmet.

| Helmet | Manufacture | Helmet One | Helmet Two | Helmet Three | Helmet Four | Helmet Five |
|---|---|---|---|---|---|---|
| **Picture** |  |  |  |  |  |  |
| **Design Elements** | Linear Acceleration | Linear Acceleration | Linear Acceleration | Linear & Rotational Acceleration | Linear & Rotational Acceleration | Decoupling - Overall Impact force attenuation |
| **Liner Material** | VN Foam & TPU Cushioning Layer | VN foam (VN 600) | 3D Structures (TPU Cup and Foam Insert) | VN Foam & Bladder (VN 600) & (TPU with EVA foam & Oil) | 3D Structures & Bladder (TPU Cup & Foam Insert) & (TPU with EVA foam &Oil ) | VN Foam & Bladder (VN 600) & (EVA Oil Filled) |
| **Shell Material** | Polycarbonate | Polycarbonate | Polycarbonate | Polycarbonate | Polycarbonate | Polycarbonate |
| **Mass** | 2.0 (±0.005) | 1.59 (±0.005) | 1.67 (±0.005) | 1.78 (±0.005) | 1.75 (±0.005) | 1.63 (±0.005) |

Study 3 - Table 1: Helmet design parameters for the commercial helmet and five prototype helmets

### *Linear Impactor*

The linear impactor system consists of a weighted, pneumatically driven impactor arm, a Hybrid III head and unbiased neck-form on a sliding table in conjunction with a computerized collection system used to simulate helmet-to-helmet collisions [Pellman 2006, Viano 2012]. The apparatus consists of a support frame, a compressed air tank, piston and impactor arm (13.1 kg ±0.1 kg), and a mobile table (12.8 kg ± 0.01 kg) with attached rails. The impacting arm accelerates horizontally using pressurized air to impact the hybrid III head-form and fixed to the sliding table. The 50th percentile adult male Hybrid III head-form (mass 4.54 ± 0.01 kg) instrumented according to Padgaonkar's (1975) orthogonal 3-2-2-2 linear accelerometer array protocol fixed to a 50$^{th}$-percentile unbiased neck-form (mass 1.54 ± 0.01 kg). Upon impact to the head-form, the assembly (head-form, neck, and table) was free to move down the rails until reaching stop pads at the end of the support table (0.65 m). A Modular Elastomer Programmer (MEP) impactor cap attached to the end of the impactor arm to elicit a dynamic response curve (shape and duration) like those reported in helmet-to-helmet collisions [Bartsch 2012, Post 2012c, Schnebel 2007, Viano 2007]. Additionally, the MEP impactor decreases testing variation for reconstructions of helmet-to-helmet impacts, helmet testing, and design research [Kindt 1971, Post 2012a/b, ASTM 2013, NOCSAE 2014].

Study 3 - Figure 2: Linear Impactor

### *Impact Locations*

The linear impactor was used to produce measurable three-dimensional helmeted head-form impacts at 5 locations and 4 impact directions shown in figure 3. All impact positions and angles were measured from a standardized head-form position based on a front centric impact location. Azimuth (θ) is defined as the angle between the impact location and the positive X axis in the X-Y (transverse) plane. Elevation (α) is defined as the angle between the impact location and the X-Y (transverse) plane. Azimuth and elevation are depicted in figure 4. Impact site accuracy was ensured using pre-marked locations on the head-form.

***Collection system***

Within the hybrid III headform, an array of nine mounted single-axis Endevco 7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) were used and signals were passed through a TDAS Pro Lab system (DTS, Calabasas, CA) before being processed by TDAS software. The impact data were sampled at 20 kHz, processed using a1650 Hz Butterworth low-pass filter and recorded using DTS TDAS PRO software. The output from the DTS software is the x, y and z time history curves of linear and rotational acceleration as well as magnitudes of the peak resultant linear and rotational acceleration.

***Computational Modeling***

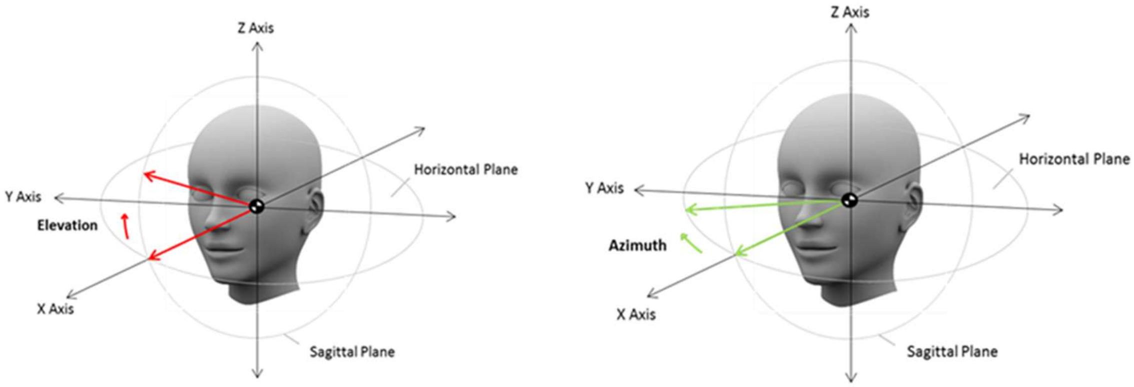
The University College Dublin Brain Trauma Model (UCDBTM) was used to determine maximum principal strain [Gilchrist 2003, Doorly 2006]. Computed Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced contour photographs were used to determine the geometry from a male cadaver [Horgan 2003]. The model is constructed of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brainstem. A linearly viscoelastic material model combined with a large deformation theory was chosen to model the brain tissue. The material properties for which, can be found in Tables 2 and 3. The compressive behaviour of the brain is considered elastic. The behaviour of this tissue was modeled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [Horgan 2003]. The shear characteristics of the viscoelastic behaviour are expressed through the formula:

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

Where G is the long-term shear modulus, $G_0$ is the short-term shear modulus and β is the decay factor [Horgan 2003].


Study 3 - Figure 3: Impact location and direction [CG, centre of gravity; NA, negative azimuth; PA, positive azimuth; PE, positive elevation]

| **Direction** | **Definition** |
|---|---|
| Center of Gravity | Perpendicular to the head-form, no vertical or horizontal rotation applied |
| Positive Elevation | A 15° elevation of the Hybrid III head and neck-form |
| Positive Azimuth | A 45°  rotation of the Hybrid III head and neck-form |
| Negative Azimuth | A -45° rotation of the Hybrid III head and neck-form |




Study 3 - Figure 4: Impact Orientation

| Material | Young's Modulus (MPa) | Poisson's Ratio | Density (kg/m$^2$) |
|---|---|---|---|
| Scalp | 16.7 | 0.42 | 1000 |
| Cortical bone | 15,000 | 0.22 | 2000 |
| Trabecular bone | 1000 | 0.24 | 1300 |
| Dura | 31.5 | 0.45 | 1130 |
| Pia | 11.5 | 0.45 | 1130 |
| Falx | 31.5 | 0.45 | 1140 |
| Tentorium | 31.5 | 0.45 | 1140 |
| CSF | 0.015 | 0.50 | 1000 |
| Grey matter | Viscoelastic | 0.49 | 1060 |
| White matter | Viscoelastic | 0.49 | 1060 |

Study 3 - Table 2: Material properties used in the University of College Dublin Brain Trauma Model [Horgan 2003]

| | Shear Modulus (kPa) | | Decay Constant (s$^{-1}$) | Bulk Modulus (GPa) |
|---|---|---|---|---|
| | $G_0$ | $G_\infty$ | | |
| Grey matter | 10 | 2 | 80 | 2.19 |
| White matter | 12.5 | 2.5 | 80 | 2.19 |
| Brain stem | 12.5 | 4.5 | 80 | 2.19 |
| White matter | 10 | 2 | 80 | 2.19 |

Study 3 - Table 3: Material properties of the different parts of the brain used in the University College Dublin Brain Trauma Model [Horgan 2006]

The CSF layer was modelled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF, and brain was used. The model is comprised of 7318 hexahedral elements representing the brain, and 2874 hexahedral elements representing the CSF layer. This version of the UCDBTM does not include elements representing the cerebral blood vessels.

The model is one of a few partially validated models sourced for this type of research [Horgan 2003, Horgan 2004]. 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). Validation was conducted using real world head injury events that were found to be consistent with anatomical research [Horgan 2003, Horgan 2004, Horgan 2006, Doorly 2006, Post 2015]. This model has been used in several recent papers in the use for head impacts, head injury, helmet design, and concussion prediction [Doorly 2006, Forero Reuda 2010, Oeur 2015, Post 2012d, Post 2013].

***Statistical Analysis***

To determine significant difference between maximum principal strain values generated from impacts to the original (commercial helmet) and each of the helmets implemented different liner technologies, the mean MPS value of all impacts per helmet was calculated and statistical analysis was run using and using an ANOVA and post hoc Tukey test and adjusting for multiple comparisons ($\alpha<0.05$).

**Results**

**Effect of different helmet liner strategies MPS collapsed across all impact conditions**

The dominant coordinate component for the commercially available helmet at each impact condition is shown in table 4. The data were analyzed to determine the effectiveness of each liner strategy, in comparison to a commercially available helmet, to reduce peak linear acceleration, peak dominant linear acceleration, peak resultant rotational acceleration, peak dominant rotational acceleration, and maximum principal strain (table 5).

| Location | LA | RA |
|---|---|---|
| Front CG | X | Y |
| Front NA | X | Z |
| Front PA | X | Z |
| Front PE | X | Y |
| Front Boss CG | Y | X |
| Front Boss NA | X | Y |
| Front Boss PA | Y | Z |
| Front Boss PE | Y | Z |
| Side CG | Y | X |
| Side NA | Y | X |
| Side PA | Y | X |
| Side PE | Y | X |
| Rear Boss CG | Y | Z |
| Rear Boss NA | Y | Z |
| Rear Boss PA | X | Y |
| Rear Boss PE | Y | Z |
| Rear CG | X | Y |
| Rear NA | X | Z |
| Rear PA | X | Z |
| Rear PE | X | Y |

Study 3 - Table 4: Dominant coordinate components for each impact condition [CG, center of gravity; NA, negative azimuth; PA, positive azimuth; PE, positive elevation]

| Helmet Type | Linear Acceleration (g) | | Rotational Acceleration ($rad/s^2$) | | MPS (SD) |
|---|---|---|---|---|---|
| | **R** (SD) | **D** (SD) | **R** (SD) | **D** (SD) | |
| Commercial | 88.7 (17.5) | 57.4 (27.7) | 7037.9 (1993.1) | 3919.1 (2160.4) | 0.417 (0.07) |
| Helmet 1 | 85.1 (19.3) | 55.8 (28.9) | 5915.0 (2533.4) | 3477.7 (2121.2) | 0.361 (0.11) |
| Helmet 2 | 75.6 (14.2) | 48.9 (22.3) | 6204.5 (1938.5) | 3637.2 (2067.2) | 0.359 (0.08) |
| Helmet 3 | 110.3 (39.8) | 73.5 (51.6) | 8181.1 (4188.6) | 4825.4 (3973.2) | 0.399 (0.13) |
| Helmet 4 | 85.2 (18.1) | 55.5 (26.9) | 7154.6 (2416.7) | 4073.5 (2142.0) | 0.385 (0.10) |
| Helmet 5 | 81.5 (15.1) | 52.1 (23.3) | 5692.9 (1778.0) | 3343.9 (1912.1) | 0.351 (0.07) |

Study 3 - Table 5: Linear and Rotational acceleration data and generated MPS for helmets tested helmet (R = Resultant acceleration (g's), D = Dominant acceleration (g's), MPS = maximum principal strain) with standard deviations (SD). Green highlight is for $p \leq 0.05$ reduction, Red highlight is for $p \leq 0.05$ increase

Helmet 1 resulted in a significant reduction in dominant rotational acceleration (14%) and reduction in MPS (13%). Helmet 2 resulted in a reduction in resultant linear acceleration (15%), a reduction in the dominant component of rotational acceleration (17%), and a reduction in MPS

(14%). Helmet 3 resulted in an increase in resultant linear acceleration (24%), an increase in dominant component of linear acceleration (28%), and no change in MPS. Helmet 4 resulted in no change to any variables. Helmet 5 resulted in a reduction in resultant rotational acceleration (19%), a reduction in dominant rotational acceleration (17%), and a reduction in MPS (16%).

Large standard deviations were expected when averaging the data resulting from all impact conditions due to the known variability resulting from impact location and direction.

**Effect of different helmet liner strategies on MPS by location**

The data were analyzed to determine the effectiveness of each liner strategy, in comparison to a commercially available helmet, to reduce peak linear acceleration, peak dominant linear acceleration, peak resultant rotational acceleration, peak dominant rotational acceleration, and maximum principal strain at each impact location; front, front boss, side, rear boss and rear. The results are shown in table 6.

**Front Location**

Helmet 1 resulted in; a reduction in resultant linear acceleration (30%), reduction in dominant linear acceleration (24%), a reduction in resultant rotational acceleration (61%) and dominant rotational acceleration (62%), and a reduction in MPS (50%). Helmet 2 resulted in a reduction in resultant linear acceleration (36%), a reduction in dominant linear acceleration (35%), a reduction in resultant rotational acceleration (46%), a reduction in dominant rotational acceleration (59%), and a reduction in MPS (41%). Helmet 3 resulted in, a reduction in resultant linear acceleration (39%), a reduction in dominant linear acceleration (37%), a reduction in resultant rotational acceleration (61%), a reduction in dominant rotational acceleration (58%), and a reduction in MPS (47%). Helmet 4 resulted in; a reduction in resultant linear acceleration (33%), a reduction in resultant rotational acceleration (40%), a reduction in dominant rotational acceleration (34%), and a reduction in MPS (36%). Helmet 5 resulted in; a reduction in resultant linear acceleration (16%), a reduction in resultant rotational acceleration (51%), a reduction in dominant rotational acceleration (52%), and a reduction in MPS (39%).

To summarize, Helmet 1 through 5 resulted in significant reductions in all variables with the exception of Helmet 4 and Helmet 5 where no significant reductions in the dominant coordinate component of linear acceleration were observed.

**Front Boss Location**

Helmet 1 resulted in a reduction in resultant rotational acceleration (41%) and a reduction in dominant rotational acceleration (42%). Helmet 2 resulted in a reduction in resultant rotational acceleration (36%) and dominant rotational acceleration (57%). Helmet 3 resulted in a reduction in dominant rotational acceleration (27%). Helmet 4 resulted in a reduction in dominant rotational acceleration (38%). Helmet 5 resulted in; a reduction in resultant rotational acceleration (39%), a reduction in dominant rotational acceleration (55%), and a reduction in MPS (23%).

To summarize, Helmets 1 through 4 resulted in no changes to resultant or dominant linear acceleration or MPS. Helmets 1 and 2 resulted in significant reductions to resultant and dominant rotational acceleration. Helmet 3 and 4 resulted in significant reduction to dominant rotational acceleration. Helmet 5 resulted in significant reduction in resultant and dominant rotational acceleration and MPS.

**Side Location**

Helmet 3 resulted in an increase in resultant linear acceleration (28%) and an increase in resultant rotational acceleration (42%). Helmet 4 resulted in an increase in resultant rotational acceleration (51%) and dominant rotational acceleration (59%). Helmet 5 resulted in no significant change in resultant and dominant linear acceleration, resultant and dominant rotational acceleration, or MPS.

To summarize, Helmet 1, Helmet 2, and Helmet 5 resulted in no changes in resultant or dominant linear acceleration, resultant or dominant rotational acceleration, or MPS. Despite some increases to some of the acceleration variables (noted above) with Helmet 3 and Helmet 4, Helmets 1-5 resulted in no significant difference in MPS.

**Rear Boss Location**

Helmet 3 resulted in; an increase in resultant linear acceleration (52%), an increase in dominant linear acceleration (24%), an increase in resultant rotational acceleration (55%), and an increase in dominant rotational acceleration (31%).

To summarize, helmets 1, 2, 4, and 5 resulted in no significant changes in resultant or dominant linear acceleration, resultant or dominant rotational acceleration. Despite the increases to some of the acceleration variables (noted above) with Helmet 3, Helmets 1-5 resulted in no significant difference in MPS.

**Rear Location**

Helmet 3 resulted in; an increase in resultant linear acceleration (114%), an increase in dominant linear acceleration (1175), an increase in resultant rotational acceleration (94%), an increase in dominant rotational acceleration (103%), and a 27% increase in MPS. Helmet 4 resulted in; an increase in resultant linear acceleration (32%), an increase in dominant linear acceleration (30%), an increase in resultant rotational acceleration (33%).

To summarize, Helmets 1 and Helmet 2 resulted in no significant changes in resultant or dominant linear acceleration, resultant or dominant rotational acceleration. Despite the increases to some of the acceleration variables (noted above) with Helmet 4 and 5, Helmets 1, 2, 4, and 5 resulted in no significant difference in MPS. Helmet 3 had increases in all acceleration variables and MPS.

**Discussion**

Dynamic response and brain tissue strain have been shown to be important factors influencing the risk of concussive injury [Gennarelli 1982, Zhang 2001, Kleiven 2007]. Developing technologies to manage linear and rotational acceleration, establishing methods to detail the influence to brain motion, and determining the effect to changes in MPS are important elements to address concussive injury risk through helmet technology. The data presented in this paper demonstrated that variations in liner materials (both structure and positioning) influenced changes to components of kinematic response and maximum principal strain.

When **all impact conditions** are considered, Helmet 1, 2, and 5 significantly reduced maximum principal strain in comparison to the commercially available helmet. In the incidence of each of these helmets, a reduction in the peak dominant coordinate component of rotational acceleration was observed. This was not the case for other variables of dynamic response, such as resultant and dominant linear acceleration or resultant rotational acceleration. The results support previous literature indicating that a helmet designed and tested to reduce linear acceleration is not as effective at reducing rotational acceleration and brain tissue strain [Delaney 2002]. However, the results also indicate the importance of the dominant component of rotational acceleration in relation to MPS, where previous research has focused on the resultant [Zhang 2001, Pellman 2003, Post 2015, Beckwith 2018, Elkin 2018]. The resultant component of acceleration does not account for directionality and motion of the head in relation to MPS.

| Location | Helmet | Linear Acceleration (g) | | Rotational Acceleration (rad/$s^2$) | | MPS (SD) |
|---|---|---|---|---|---|---|
| | | ***R*** (SD) | **D** (SD) | **R** (SD) | ***D*** (SD) | |
| **Front** | Commercial | 104.1 (15.0) | 86.1 (22.5) | 7769.6 (2447.4) | 5663.2 (3751.0) | 0.456 (0.09) |
| | Helmet 1 | 72.5 (16.6) | 65.2 (20.4) | 3023.2 (512.1) | 2176.5 (740.0) | 0.226 (0.03) |
| | Helmet 2 | 67.1 (7.5) | 55.9 (13.2) | 4193.6 (1486.8) | 2392.9 (1438.7) | 0.269 (0.05) |
| | Helmet 3 | 63.1 (13.3) | 54.1 (18.8) | 3035.7 (1253.4) | 2388.1 (730.2) | 0.242 (0.08) |
| | Helmet 4 | 69.4 (8.5) | 61.2 (12.7) | 4699.8 (1352.1) | 3757.1 (591.0) | 0.292 (0.08) |
| | Helmet 5 | 87.4 (12.7) | 75.2 (18.6) | 3805.8 (1134.1) | 2732.6 (1273.1) | 0.276 (0.09) |
| **Front Boss** | Commercial | 95.6 (12.1) | 74.9 (31.3) | 7687.3 (1319.4) | 5274.2 (1590.1) | 0.432 (0.09) |
| | Helmet 1 | 83.9 (21.0) | 66.8 (32.6) | 4344.2 (2270.3) | 3046.6 (2340.0) | 0.319 (0.11) |
| | Helmet 2 | 75.5 (12.9) | 57.5 (24.5) | 4957.9 (1309.2) | 3915.1 (1764.3) | 0.339 (0.08) |
| | Helmet 3 | 92.3 (32.4) | 76.3 (43.2) | 5640.5 (2999.5) | 3884.8 (2999.5) | 0.341 (0.15) |
| | Helmet 4 | 84.9 (14.9) | 65.5 (30.2) | 5732.4 (1453.9) | 4112.8 (1452.4) | 0.375 (0.10) |
| | Helmet 5 | 84.3 (9.1) | 62.3 (22.7) | 4700.3 (546.9) | 3498.6 (910.3) | 0.332 (0.05) |
| **Side** | Commercial | 85.0 (9.7) | 79.6 (12.2) | 5469.1 (1481.6) | 4303.8 (449.4) | 0.403 (0.05) |
| | Helmet 1 | 86.8 (17.2) | 84.0 (19.6) | 6575.2 (1151.5) | 5432.5 (1177.6) | 0.439 (0.07) |
| | Helmet 2 | 83.4 (20.4) | 79.9 (24.0) | 6672.4 (964.3) | 5709.4 (1765.5) | 0.424 (0.07) |
| | Helmet 3 | 108.6 (23.2) | 104.4 (26.9) | 7754.4 (1721.2) | 6674.4 (1724.0) | 0.452 (0.05) |
| | Helmet 4 | 92.3 (20.3) | 88.2 (24.6) | 8259.1 (1936.9) | 6843.6 (2845.3) | 0.431 (0.06) |
| | Helmet 5 | 92.7 (20.1) | 87.5 (24.6) | 5722.0 (1560.7) | 5017.3 (1293.7) | 0.372 (0.06) |
| **Rear Boss** | Commercial | 84.4 (23.0) | 61.1 (22.7) | 8139.6 (2225.7) | 5539.4 (1666.4) | 0.430 (0.07) |
| | Helmet 1 | 92.9 (19.5) | 63.3 (27.3) | 8799.0 (2036.9) | 5477.2 (1681.5) | 0.419 (0.06) |
| | Helmet 2 | 68.5 (8.6) | 45.7 (20.7) | 7723.1 (1102.1) | 4866.7 (1606.3) | 0.383 (0.04) |
| | Helmet 3 | 128.4 (24.7) | 93.9 (43.2) | 12589.7 (2476.9) | 7465.6 (2761.6) | 0.495 (0.07) |
| | Helmet 4 | 81.2 (17.4) | 56.8 (20.1) | 8934.0 (2595.9) | 5930.4 (3184.7) | 0.448 (0.10) |
| | Helmet 5 | 72.8 (10.4) | 52.6 (22.8) | 7583.8 (1664.4) | 4653.9 (1875.5) | 0.412 (0.04) |
| **Rear** | Commercial | 74.3 (9.0) | 69.8 (12.9) | 6123.9 (680.3) | 5206.7 (1333.5) | 0.364 (0.02) |
| | Helmet 1 | 89.6 (18.3) | 84.5 (22.5) | 6833.7 (1116.8) | 5746.3 (1514.2) | 0.400 (0.08) |
| | Helmet 2 | 83.5 (10.3) | 76.5 (7.7) | 7475.8 (1860.7) | 5834.1 (869.7) | 0.380 (0.06) |
| | Helmet 3 | 158.9 (19.8) | 151.4 (26.6) | 11885.1 (1463.6) | 10559.9 (2370.7) | 0.463 (0.04) |
| | Helmet 4 | 98.2 (14.6) | 90.8 (16.5) | 8147.7 (1501.3) | 7020.0 (1686.4) | 0.381 (0.05) |
| | Helmet 5 | 70.3 (8.3) | 65.2 (12.0) | 6652.6 (546.7) | 5449.8 (833.6) | 0.362 (0.04) |

Study 3 - Table 6: Linear and Rotational acceleration data and generated MPS data by impact location for helmets tested helmet [R = Resultant acceleration (g's), D = Dominant acceleration (g's), MPS = maximum principal strain, with standard deviations (SD)]. Green highlight is for p ≤ 0.05 reduction, Red highlight is for p ≤ 0.05 increase

Helmet 1, Helmet 2 and the commercial helmet were constructed with the materials aimed to lower linear acceleration. Therefore, significant improvements in linear acceleration between Helmet 1 and Helmet 2 with the commercial helmet were not expected. The reduction in dominant rotational acceleration of these two helmets in comparison to the commercial helmet may be due to a reduced coupling offered by the low surface friction of pure VN in Helmet 1 and the reduced surface area contact in Helmet 2. This indentifies that reduction in MPS values can be optimized with reduced contact between the head and helmet allowing for separate movement of the helmet and also for a low friction layer where contact does occur to prevent coupling.

It is hypothesized that the commercial helmet may couple to the head more than the prototype helmets. With the addition of the facemask at the front and front boss regions, the helmet may become coupled to the head and the impactor cap may essentially grab the facemask and create a greater force transfer into the head of the player. Additional methods to reduce this coupling appear to reduce MPS values. Pellman (2003) reported 29.3% of severe impacts in American football occurred to facemask or front the helmet making this finding important. Improvements to current helmets can be made in locations with high risk of impact by reducing the coupling of the head and helmet as much as possible, particularly at the front and front boss locations.

None of the prototype helmets significantly reduced MPS at the side, rear boss, or rear impact locations. It is suggested that for the side, rear boss and rear impact locations adequate decoupling did not occur with Helmet 5. This is likley due to the placement of the pods in Helmet 5 in relation to the geometry of the head. It is possible that at high point locations of the head, particularly the crown and occiput, were not effectively managed by the helmet [Aare 2003].

**Limitations**

The impacts within the laboratory make use of a non-biofidelic physical headform and linear impacting system. This set up yields 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. However, the use of a controlled laboratory setting is suitable for helmet impact performance testing and evaluation [Krzeminski 2016]. The test conditions used in this study were limited to specific impact sites and one test velocity. Additional information to test the helmets under impact conditions derived from actual concussive events in American football is essential for future development decoupling helmet liner strategies.

The finite element model used to calculate maximum principal strain 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. However, it is a useful tool to compare between different impacts giving relative values rather than an absolute such as the case comparing the response of two helmets for the same impact conditions in this study.

**Conclusion**

In comparison to a commercial American football helmet, the helmet designed using principles to create decoupling and redirect impact energy resulted in a 16% reduction in maximum principal strain when analysing all impact locations together, a 39% reduction at the front location and a 23% reduction in at the front boss location. These results support the hypothesis that a novel liner technology could effectively manage rotational acceleration and tissue strain for impact in American football helmets.

**References**


64. **Aare, M.,** Kleiven, S., Halldin, P., 2003. Injury Criteria for Oblique Helmet Impacts. International Research Council on Biomechanics of Injury Conference (IRCOBI), Dublin, Ireland, 349-350.
65. **ASTM F1446-13,** American Society for Testing and Materials International, 2013.
66. **Bailes, J.E.,** Hudson, V., 2001. Classification of Sport-Related Head Trauma: A Spectrum of Mild to Severe Injury. Journal of Athletic Training 36(3), 236-243.
67. **Bartsch, A.,** Benzel, E., Miele, V., Prakash, V., 2012. Impact test comparisons of 20th and 21st century American football helmets. Journal of Neurosurgery 116, 222-233.
68. **Bensen, B.W.,** McIntosh, A. S., Maddocks, D., Herring, S. A., Raftery, M., Dvorak. J., 2013. What are the Most Effective Risk-Reduction Strategies in Sport Concussion? British Journal of Sports Medicine, 47(5), 321-326.
69. **Cantu, R.C.,** Mueller, F.O. 2003. Brain Injury Related Fatalities in American Football, 1945-1999. Neurosurgery – Special Edition, 52(4), 846-853.
70. **Caserta, G.D.,** Iannucci, L., Galvanetto. U. 2011. Shock Absorption Performance of a Motorcycle Helmet with Honeycomb Reinforced Liner. Composite Structures. 93: 2748 – 2759. **Darling, T.,** Muthuswamy, J., Rajan. S.D. 2016. Finite Element Modeling of Human Brain Response to Football Impacts. Computer Methods in Biomechanics and Biomedical Engineering. 19(13):1432-1442.
71. **Delaney, J.S.,** Lacroix, V. J., Leclerc, S., Johnston. K. M. 2002. Concussions among University Football and Soccer Players. Journal of Sport Medicine 12(6):331-338.
72. **Doorly, M.C.**, Gilchrist, M.D., 2006. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. Computer Methods in Biomechanics and Biomedical Engineering 9(6), 371 – 377.
73. **Finan, J.D.,** Nightingale, R.W., Myers, B.S. 2008. The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts. Traffic Injury Prevention. 9: 483 – 488.
74. **Forero Rueda, M.A**., Cui, L., Gilchrist, M.D., 2010. Finite Element Modeling of Equestrian Helmet Impacts Exposes the Need to Address Rotational Kinematics in Future Helmet Designs. Computer Methods in Biomechanics and Biomedical Engineering 10(5), 1-11.
75. **Fowler, J.N.,** Pallanta, A.A., Swanik, C.B., Wagner, N.J., 2015. The Use of Shear Thickening Nanocomposites in Impact Resistant Materials. Journal of Biomedical Engineering 137, 054505-1-5.
76. **Gennarelli, T.A.,** Ommaya, A. K., Thibault, L. E. 1971. Comparison of translational and rotational head motions in experimental cerebral concussion. In 15th Stapp Car Crash Conference Proceedings, 17-19, Coronado, Calif., pg. 797-803. New York: Society of Automotive Engineers.
77. **Gilchrist, M. D.,** Mills, N. J. 1994. Modelling of the Impact Response of Motorcycle Helmets. International Journal of Engineering. 15(3): 201-218**.**
78. **Gilchrist, M.D.,** 2003. Modeling and Accident Reconstruction of Head Impact Injuries. Engineering Materials 245/245, 417-429.
79. **Gimbel, G.M.,** Hoshizaki, T. B. 2008. Compressive Properties of Helmet Materials Subjected to Dynamic Impact Loading of Various Energies. European Journal of Sport Science. 8(6): 341-349.

80. **Gurdjian, E.S.,** Hodgson, V. R., Thomas, L. M., Patrick. L.M. 1968. Significance of Relative Movements of Scalp, Skull, and Intracranial Contents During Impact Injury of the Head. Journal of Neurosurgery. 29(1): 70-72.
81. **Guskiewicz, K.M.,** Weaver, N.L., Padua, D. A., Garrett, W. E. 2000. Epidemiology of Concussion in Collegiate and High School Football Players. The American Journal of Sports Medicine Vol. 28(5): 643 – 650.
82. **Guskiewicz, K.M.,** Mihalik, J. P., Shankar, V., Marshall, S. W., Cro-well, D. H., Oliaro, S. M., Ciocca, M.F., Hooker. D. N. 2007. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical out-come after concussion. Neurosurgery. 61: 1244–1253.
83. **Halldin, P.H.,** Aare, M., von Holst. H. 2000. An Experimental Safety Helmet with Improved Rotational Protection. Crash Prevention and Injury Control.
84. **Hansen, K.**, Dau, N., Fiest, F., Deck, C., Willinger, R., Mady, S., Bottlang. M. 2013. Angular Impact Mitigation System for Bicycle Helmets to Reduce Head Acceleration and Risk of Traumatic Brain Injury. Accid Anal Prev. 59:109-117.
85. **Hardy, W.N.,** Khalil, T. B., King. A. I. 1994. Literature Review of Head Injury Biomechanics. International Journal of Impact Engineering. 15(4):561-586.
86. **Hardy, W.N.,** Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., Tashman. S. 2001. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar x-ray. Stapp Car Crash Journal; The Stapp Association, Ann Arbor, Michigan.
87. **Horgan, T.J.**, Gilchrist, M.D., 2003. The creation of 3D finite element models for simulating head impact biomechanics. International Journal of Crashworthiness 8(4), 353 – 366.
88. **Horgan, T.J.,** Gilchrist, M.D., 2004. Influence of FE Model Variability in Predicting Brain Motion and Intracranial Pressure Changes in Head Impact Simulations. International Journal of Crashworthiness 9(4), 401-418.
89. **Horgan, T.J.,** 2006. A Finite Element Model of The Human Head for Use in the Study of Pedestrian Accidents. Doctor of Philosophy Thesis. Department of Mechanical Engineering, Faculty of Engineering and Architecture. University of College Dublin, February 2005.
90. **Hoshizaki, T.B.,** Post, A., Oeur, R. A., Brien. S. E. 2014. Current and Future Concepts in Helmet and Sports Injury Prevention. Neurosurgery. 75(4): 5136-5148.
91. **Jadischke, R.,** Viano, D.C., Dau, N., King, A.I., McCarthy, J., 2013. On the Accuracy of the Head Impact Telemetry System Used in Football Helmets. Journal of Biomechanics DOI: 10.1016/j.jbiomech. 2013.05.030
92. **Johnston, J.M.,** Ning, H., Kim, J. E., Kim, Y. H., Soni, B., Reynolds, R., Cooper, L., Andrews, J. B., Vaidya. U. 2015. Simulation, Fabrication and Impact Testing of a Novel Football Padding System That Decreases Rotational Acceleration. Sports Engineering. International Sports Engineering Association. 18:11-20.
93. **Kindt, G.W.,** Scheider, R. C., Robinson. J. L. 1971. A Method of Comparing the Impact Absorption Characteristics of Football Helmets. Medicine and Science in Sports 3(4): 203-209.
94. **Kleiven, S.** 2007. Predictors for Traumatic Brain Injuries Evaluated Through Accident Reconstructions. Stapp Car Crash Journal. 51: 81-114.
95. **Krzeminski, D.E**., Fernando, B.M.D., Curtzwiler, G., Gould, T.E., Piland, S.G. 2016. Repetitive Impact Exposure and Characterization of Stress-Whitening of an American Football Helmet Outer Shell Material. Polymer Testing 55, 190-203.

96. **Langlois, J.A.**, Rutland-Brown, W., Wald, M.M., 2006. The Epidemiology and Impact of Traumatic Brain Injury: A Brief Overview. Journal of Head Trauma Rehabilitation 21(5), 375-378.
97. **Lamb, L.,** Hoshizaki, T. B. 2009. Deformation Mechanisms and Impact Attenuation Characteristics of Thin-Walled Collapsible Air Chambers Used in Head Protection. Journal of Mechanical Engineering in Medicine. 223(8), 1021-1031.
98. **Margulies, S.S.,** Thibault, L.E., 1992. A proposed tolerance criteria for diffuse axonal injury. Journal of Biomechanics 25, 917- 923.
99. **Nahum, A.M.,** Smith, R., Ward, C.C., 1977. Intracranial pressure dynamics during head impact. 21st Annual Stapp Car Crash Conference. Society of Automotive Engineers.
100. **National Operating Committee on Standards for Athletic Equipment** (2004) Standard Drop Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear: NOCSAE Standard 001-04m05. Available at www.nocsae.org.
101. **National Operating Committee on Standards for Athletic Equipment** (2014) Standard Performance Specification for Newly Manufactured Football Helmets: NOCSAE DOC (ND) 002-14. Available at www.nocsae.org
102. **Oeur, R.A.,** Karton, C., Post, A., Rousseau, P., Hoshizaki, T.B., Marshall, S., Brien, S., Smith, A., Cusimano, M.D., Gilchrist, M.D., 2015. A Comparison of Head Dynamic Response and Brain Tissue Stress and Strain Using Accident Reconstruction for Concussion, Concussion with Persistent Postconcussive Symptoms and Subdural Hematoma. Journal of Neurosurgery. DOI:10.3171/2014.10.JNS14440.
103. **Oukama, R.,** Pearsall, D. J. 2011. Evaluation of a Flexible Force Sensor for Measurement of Helmet Foam Impact Performance. Journal of Biomechanics. 44(5): 904-909.
104. **Padgaonkar, A.J.,** Kreiger, K.W., King, A.I., 1975. Measurements of angular accelerations of a rigid body using linear accelerometers. Journal of Applied Mechanics 42, 552-556.
105. **Pellman, E.J.,** Viano, D.V., Casson, I.R., Waeckerle, J.F., 2003. Concussion in Professional Football: Reconstruction of Game Impacts and Injuries. Neurosurgery 53, 799-814.
106. **Pellman, E.J.**, Viano, D.C., Withnall, C., Shewchenko, N., Bir, C.A., Halstead, P.D., 2006. Concussion in professional football: Helmet testing to assess impact performance – Part 11. Journal of Neurosurgery 58, 78 – 96.
107. **Post, A.,** Oeur, A., Hoshizaki, B., Gilchrist, M.D., 2012a. An Examination of American Football Helmets Using Brain Deformation Metrics Associated with Concussion. Journal of Materials and Design. DOI:10/1016/ j.matdes. 2012.09.017
108. **Post, A.**, Oeur, A., Hoshizaki, T.B., Gilchrist, M.D., 2012b. The Influence of Centric and Non-Centric Impacts to American Football Helmets on the Correlation Between Commonly Used Metrics in Brain Injury Research. International Research Council on Biomechanics of Injury Conference (IRCOBI), Dublin, Ireland.
109. **Post, A.,** Walsh, E.S., Hoshizaki, T.B., Gilchrist, M.D., 2012c. Analysis of Loading Curve Characteristics on the Production of Brain Deformation Metrics. Journal of Sports Engineering and Technology. DOI: 10.1177/1754337112449420.
110. **Post, A.,** Hoshizaki, T. B., Gilchrist. M. D., 2012d. Finite Element Analysis of the Effect of Loading Curve Shape on Brain Injury Predictors. Journal of Biomechanics. 45:679-683.
111. **Post, A.,** Oeur, A., Walsh, E., Hoshizaki, T.B., Gilchrist, M.D., 2013. A centric/non-centric impact protocol and finite element model methodology for the evaluation of American

Football Helmets to Evaluate Risk of Concussion. Computer Methods in Biomechanics and Biomedical Engineering 1-16.

112. **Post, A.,** Hoshizaki, T.B., 2015. Rotational Acceleration, Brain Tissue Strain, and the Relationship to Concussion. Journal of Biomechanical Engineering 137, 1-8.
113. **Rousseau, P.,** Post, A., Hoshizaki. T. B. 2009. The Effects of Impact Management Materials in Ice Hockey Helmets on Head Injury Criteria. J sport Eng Tech. 233:159-165.
114. **Rowson, S.,** Duma, S., Stemper, B., Shah, A., Mihalik, J., Harezlak, J., Riggen, L., Giza, C., Difiori, J., Brooks, A., Guskiewicz, K., Campbell, D., Mcginty, G., Svoboda, S., Cameron, K., Mcallister, T., Mccrea, M. 2018. Correlation of Concussion Symptom Profile with Head Impact Biomechanics: A Case for Individual-Specific Injury Tolerance. Journal of Neurotrauma, 35(4): 681-690.
115. **Schnebel, B.,** Gwin, J.T., Anderson, S., Gatlin, R., 2007. In Vivo Study of Head Impacts in Football: A Comparison of National Collegiate Athletic Association Division I Versus High School Impacts. Neurosurgery 60, 490-496.
116. **Spyrou, E.,** Pearsall, D. J., Hoshizaki. T. B. 2000. Effect of Local Shell Geometry and Material Properties on Impact Attenuation of Ice Hockey Helmets. Sports Engineering. 3:25-35.
117. **Trosseille, X.,** Tarrie´re, C., Lavaste, F., et al. (1992) Development of a F.E.M. of the human head according to a specific test protocol. In: 36th Stapp Car Crash Conference, Seattle, WA, USA, SAE paper 922527.
118. **Unterharnscheidt, F.J.** 1971. Translational versus rotational acceleration: animal experiments with measured inputs.Proceedingsofthe15th Stapp Car Crash Conference SAE 710880.
119. **Vanden Bosche, K.,** Mosleh, Y., Depreitere, B., Vander Sloten, J., Verpoest, I., Ivens J. 2017. Anisotropic Polyethersulfone Foam for Bicycle Helmet Liners to Reduce Rotational Acceleration During Oblique Impact. Journal of Engineering in Medicine. 231(9):851-861, 2017.
120. **Viano, D.C.,** Casson, I., Pellman, E.J., 2007. Concussion in Professional Football: Biomechanics of the Struck Player – Part 14. Neurosurgery 16(2) 313 – 327.
121. **Viano, D.C.,** Halstead, D., 2012. Change in Size and Impact Performance of Football helmets from the 1979's to 2010. Annals of Biomedical Engineering 40(1), 175-184.
122. **Walsh, E.S.,** Rousseau, P., Hoshizaki, T.B., 2011. The Influence of Impact Location and Angle on the Dynamic Impact Response of a Hybrid III Headform. Journal of Sports Engineering and Technology. doi.10.1007/s12283-011-0060-9
123. **Zhang, L.,** Yang, K.H., King, A.I., 2004. A Proposed Injury Threshold for Mild Traumatic Brain Injury."  Journal of Biomedical Engineering 126, 226 – 236.

# Study 4

## The effect of a novel impact management strategy on maximum principal strain for reconstructions of American football concussive events

**Abstract**

The purpose of this study was to test the effectiveness of a helmet designed to redirect motion of the brain and reduce maximum principal strain by decoupling the head-helmet interface in reconstructions of head-to-head concussive impacts in American football. The primary goal of the American football helmet has been protection of players against skull fractures and other traumatic brain injuries. The modern-day helmet has evolved and become well designed to mitigate TBI injuries but does not offer optimized protection against concussive injury. Studies to determine the influence of decoupling strategies on changes to brain motion and the resulting influence on maximum principal strain, a metric associated with concussive injury, have not yet been examined under concussive impact conditions. In this study, nineteen helmet-to-helmet concussive events from professional American football were reconstructed using a pneumatically driven linear impactor to determine the components (resultant and dominant) of linear and rotational acceleration. The University of College Dublin Brain Trauma Model was used to determine maximum principal strain values. A prototype helmet with a decoupling liner strategy was shown to significantly reduce maximum principal strain (MPS) in 7 of the head to head concussive impact reconstructions. In each case, the prototype helmet significantly reduced the dominant coordinate component of rotational acceleration. The results of this study demonstrate that targeting reductions in dominant coordinate component of rotational acceleration through helmet innovation resulted in significant decreases in maximum principal strain for reconstructions of concussive events in American football.

**Null Hypothesis:** No significant differences in maximum principal strain between the original and prototype helmets for each concussive event.

**Key words:** Brain injury, impact management, helmet decoupling technology, dominant coordinate component of acceleration

## Introduction

The hard-shelled American football helmet has evolved over the last several decades using a "common sense intuition" design approach to mitigate linear acceleration through plastic deformation of the shell and compression of the liner material.[1-4] The modern day helmet, designed to mitigate traumatic brain injuries associated with linear acceleration, does not offer effective protection against concussive injury.[5-10] This is important as American football is the sport which accounts for the largest reported rate of concussive injury.[11-13]

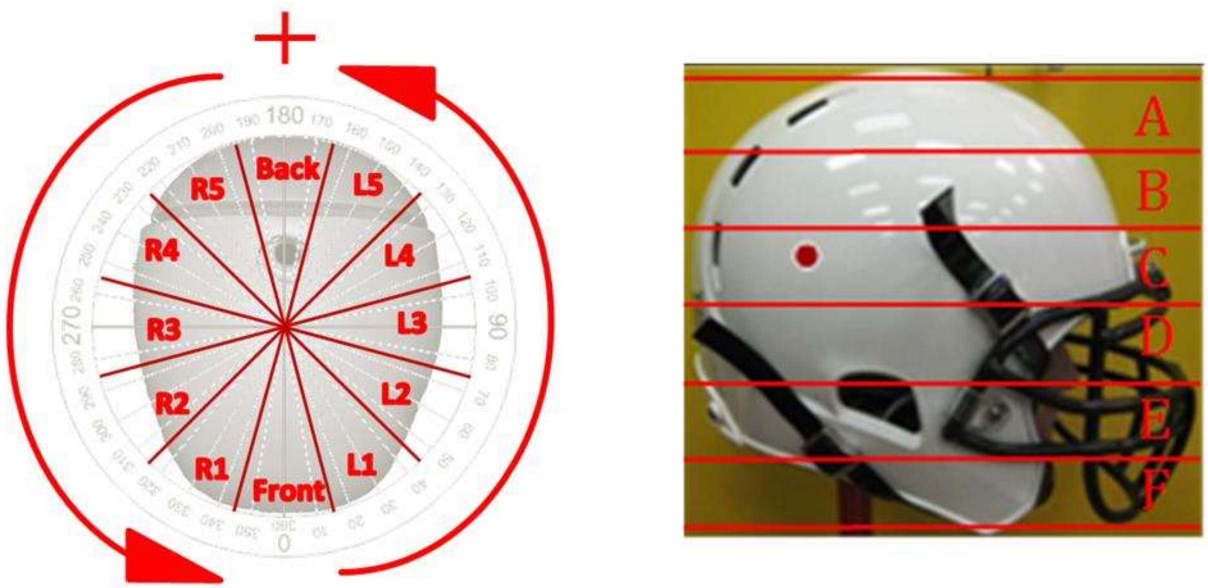
Researchers report the underlying event mechanism leading to concussion is rotational acceleration of the brain, resulting in shear strain on the brain tissue, in conjunction with lower magnitude, longer duration impacts.[14-22] Despite the implementation of rules and fines in 1996 against intentional head impacts, the most common head impact event resulting in concussion in American football is the head-to-head collision.[3,23-27] In football, the compliance of the helmet results in typically longer duration impacts with lower magnitude of brain accelerations for helmet to helmet hits than impacts to hard surfaces such as the ground.[26,28,29] Additionally, the majority of all head-to-head impacts are not through the centre of gravity of the head resulting in a larger rotational component to the impact.[3,30,31] Therefore, the most common impact event in football, the head to head collision, results in a high risk of concussive injury, yet the current helmet is not designed specifically to protect against it.[32,33] The discrepancy between the impact event mechanism of concussive injury and the design and testing of current helmets may explain why the rate of concussion remains high.[24,34-36] Research investigating helmet design improvement has begun to look at using different liner technologies and strategies that specifically address rotational acceleration of the brain to improve helmet performance and include concussive injury risk protection.[28,37-39] As a result, recent research has examined the use of decoupling/force redirecting liner systems. Placement of decoupling strategies within the helmet is integral to the success of the decoupling mechanisms and is governed by two design principles to prevent coupling of the helmet and the head. The first consists of a low friction or slip layer between the head and helmet. The second principal is to reduce the contact area between the head and helmet with the creation of void space at the high points of the human head. This is typically at the four corners of the head, particularly the high temporal region and the rear corners around where the parietal, occipital and temporal lobes meet in addition to the crown of the head. Implementation and placement of decoupling strategies allows for the helmet and head to move separately redirecting a portion of the impact force around the head, reducing the dominant coordinate component of acceleration which results in a decreased maximum principal strain (MPS) value. The dominant component of acceleration is defined as the component that makes the highest contribution to the resultant and provides a tangible performance evaluation and helmet design metric that accounts for impact location and direction, which resultant acceleration does not. The strong link identified between strain, resulting from head motion during an impact event, and tissue damage renders MPS a common measure of injury.[40-42]

Laboratory reconstructions have been utilized in the past when investigating head impact events including those from American football.[3,43-45] However, these studies have primarily focused on the impact characteristics of concussive injury impact events in American football and not on helmet performance. This has led to limited information of how the football helmet performs in real life impacts and the corresponding effect to brain tissue strain.[44] To effectively assess helmet performance in addressing concussive injury risk, evaluation must be conducted under the event conditions known to cause concussion using laboratory reconstructions. The purpose of this study was to test the effectiveness of the prototype helmet, designed using a decoupling strategy and the principles outlined above to reduce maximum principal strain. In addition to MPS, to help detail the performance of the prototype helmet, analyses were also run using the components of linear and rotational acceleration (resultant and dominant). The prototype helmet was evaluated against a similar model of manufactured helmet worn during recorded concussive events occurring in the professional American football for its ability to reduce maximum principal strain.

## Methodology

### *Test Procedure*

Reconstructions of head-to-head impacts for reported concussion during the 2009-2014 seasons from professional football were conducted. A total of 46 reported head-to-head concussive events were obtained from the National Football League (NFL) game centre and usable video footage was recorded. Nineteen impacts were excluded as they were greater than 9.0 m/s which is the range where current helmet performance is limited.[46,47] Eight cases were excluded as they were directly to the facemask or chin pad which would not have resulted in a difference in helmet performance. The remaining 19 were reconstructed for this study. Kinovea 0.8.2 video analysis software was used to analyze high definition video of each concussive event to establish the laboratory parameters required to conduct the reconstruction, such as impact location, direction and velocity as contained in table 1 with location grid shown in figure 1. The Kinovea software allows for measurements of distance given that there are known markers, such as the field distance or yard markers, on the surface to use as reference.[48] Impact velocity was determined by applying a perspective grid based on known points and distances on the football field, which established the displacement of the player.[48] The impact velocity associated with collision is calculated by measuring the distance between two players 5 frames prior to the impact that relays to 0.2 second, given a frame rate of 25 f/s.[48] Displacement over time was calculated to determine the contact velocity.[48] The velocity of the impacts used for reconstruction ranged from 5.7 to 8.3 m/s.[3,49,50] An analysis of measurement error of this methodology was conducted and determined to be approximately 5% for velocity and approximately 10 degrees for impact angle.[48,51] For each concussive event case, three impact trials were conducted for repeatability using the commercial helmet (similar to that worn by the concussed player) and then repeated using the prototype helmet.




Study 4 - Figure 1: Impact location grid

| Case # | Velocity (m/s) | Impact Location | Commercial Helmet Worn |
|---|---|---|---|
| 1 | 5.7 | R3 / R4-D | Xenith X1 |
| 2 | 5.8 | L3-B | Riddell 360 |
| 3 | 6.5 | L2-B | Riddell VSR4 |
| 4 | 6.9 | L3-B | Riddell Rev IQ |
| 5 | 6.9 | R3-A | Riddell Rev IQ |
| 6 | 7.2 | R2-D | Riddell Rev IQ |
| 7 | 7.2 | Crown | Riddell Rev Speed |
| 8 | 7.4 | L2-F | Riddell Rev Speed |
| 9 | 7.5 | FR-A | Riddell Rev IQ |
| 10 | 7.6 | R4-D | Riddell VSR4 |
| 11 | 7.8 | L4-D | Riddell Rev Speed |
| 12 | 7.9 | L3-C | Riddell Rev IQ |
| 13 | 8.1 | FR-B | Riddell 360 |
| 14 | 8.3 | R2-C | Riddell VSR4 |
| 15 | 8.3 | Crown | Schutt Air XP Pro |
| 16 | 8.5 | L5-B | Schutt Air XP Pro |
| 17 | 8.5 | L5-D | Schutt Air XP Pro |
| 18 | 8.6 | FR-A | Schutt Air XP Pro |
| 19 | 9.02 | R2-D | Schutt Air XP Pro |

Study 4 - Table 1: Reconstruction case of concussive impact events

***Test Equipment***

To represent the head-to-head collisions, a linear impactor was used to impact the Hybrid III headform, similar to previous reported sport impact reconstructions.[3,35,52] The linear impactor consists of a weighted, pneumatically driven impactor arm, a Hybrid III head and unbiased neck-form on a sliding table in conjunction with a computerized collection system. Head-to-head impacts, as seen in figure 2, employ an additional Hybrid III head and neck attached to the impacting end of the cylindrical arm via an L-frame and angled wedge to represent the striking player. Four different angled wedges (15°, 30°, 45° and 60°) were used to obtain the desired neck angle of the striking player. A heavier free-moving cylindrical impacting arm (16.1 ± 0.1 kg) and a stiff Hybrid III neck form were used for these collisions to represent the higher mass and minimal neck bending of a striking player.[53] for a total striking mass of 24.0 kg. The apparatus consists of a support frame, a compressed air tank, the piston and impactor arm, and a mobile table (12.8 kg ± 0.01 kg) with attached rails. The impacting arm was propelled horizontally using pressurized air to impact the hybrid III head-form and fixed to the sliding table. The 50th percentile adult male Hybrid III head-form (mass 4.54 ± 0.01 kg) instrumented according to Padgaonkar's (1975) orthogonal 3-2-2-2 linear accelerometer array protocol was fixed to a 50th-percentile unbiased neck-form (mass 1.54 ± 0.01 kg).[54] The set up allowed for adjustment in five degrees of freedom providing variable positioning.


Study 4 - Figure 2: Head to head impact set up for Case 1

***Prototype Helmet***

This research presents a novel design principal, which differs from conventional helmet liner design that typically cradle and couple the head primarily for comfort and load distribution and from more recent decoupling designs that cover the entire surface of the head. Decoupling the head helmet interface is best accomplished by following a different design principle. The principle is to create void space in the helmet liner to avoid coupling of the head and helmet at high ridge points of the human head. The skull is generally ellipsoidal in shape and does not often fit symmetrically within a round football helmet creating non-uniform contact with between the head and helmet at the high point regions of the head. Allowing for voids in the liner at these regions removes the high point contact areas between the head and helmet interface which could hinder the release or couple the head and helmet. Typically, this is seen at the four corners of the head, particularly the crown of the head, the high temporal regions, and the rear corners where the parietal, occipital and temporal lobes meet. Therefore, placement of oil filled bladders inside the helmet should articulate with the head in common high risk impact areas while avoiding high points to create release and separation of the head and helmet. For the protypye helmet, the ethylene-vinyl acetate EVA and oil filled bladders were placed at typical impact locations on two layers of vinyl nitrile (VN) 600 and avoiding the high points on the human head. Compared to the commercially available helmets, the prototype helmets had similar impact liner thickness a polycarbonate shell. The full helmet including mask (Carbon steel, meeting NOCSAE standard 12-2014) and chin strap had a mass of 1.63 kg (±0.005 kg).

***Computational Analysis***

To compare the performance of the prototype helmet to the commercially available helmets in terms of brain deformation, the University College Dublin Brain Trauma Model (UCDBTM) was used.[55,56] The geometry used for the model was developed with the use of CT and MRI imaging of a male cadaver head. The version used for this study was composed of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and the brain stem.[55,56]

For each component, the material properties were defined from previous anatomical research on cadavers and are summarized in tables 2 and 3.[57-61] The tissue behaviour was characterized as viscoelastic in shear and elastic in compression. To model the CSF, solid elements were used with low shear moduli and a sliding boundary condition to define the interactions between the CSF, brain and skull.

A linear viscoelastic model with large deformation theory was used to represent the behavior as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus.[55] The compressive nature of the brain was represented as elastic. The shear characteristics of the viscoelastic behaviour were expressed through the formula:

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

Where G is the long-term shear modulus, $G_0$ is the short-term shear modulus and β is the decay factor.[55]

| Material | Young's Modulus (MPa) | Poisson's Ratio | Density (kg/m$^2$) |
|---|---|---|---|
| Scalp | 16.7 | 0.42 | 1000 |
| Cortical bone | 15,000 | 0.22 | 2000 |
| Trabecular bone | 1000 | 0.24 | 1300 |
| Dura | 31.5 | 0.45 | 1130 |
| Pia | 11.5 | 0.45 | 1130 |
| Falx | 31.5 | 0.45 | 1140 |
| Tentorium | 31.5 | 0.45 | 1140 |
| CSF | 15,000 | 0.50 | 1000 |
| Grey matter | Viscoelastic | 0.49 | 1060 |
| White matter | Viscoelastic | 0.49 | 1060 |

Study 4 - Table 2: Material properties used in the University of College Dublin Brain Trauma Model [55]

| | Shear Modulus (kPa) | | Decay Constant ($s^{-1}$) | Bulk Modulus (GPa) |
|---|---|---|---|---|
| | $G_0$ | $G_\infty$ | | |
| Grey matter | 10 | 2 | 80 | 2.19 |
| White matter | 12.5 | 2.5 | 80 | 2.19 |
| Brain stem | 12.5 | 4.5 | 80 | 2.19 |
| White matter | 10 | 2 | 80 | 2.19 |

Study 4 - Table 3: Material properties of the different parts of the brain used in the University College Dublin Brain Trauma Model [56]

The CSF layer was modelled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF, and brain. The model is comprised of 7318 hexahedral elements representing the brain, and 2874 hexahedral elements representing the CSF layer. This version of the UCDBTM does not include elements representing the cerebral blood vessels.

Validation of the model was completed using comparisons with intracranial impact cadaveric pressure data from Nahum (1977) and Trosseille (1992) and brain motion from Hardy (2001).[62-63] Further validations were conducted using brain injury reconstructions from real-life incidents which achieved good agreement with the magnitudes of strain and stress in the literature.[51,65,66] This model has been used in several recent papers in the use for head impacts, head injury, helmet design, and concussion prediction.[30,46,65-68]

***Statistical Analysis***

Comparison of the performance of the two helmets (commercial helmets and prototype helmet) was conducted using maximum principal strain as the primary dependent variable. Secondary analyses were conducted using peak resultant and peak dominant component of linear and rotational acceleration.

An independent sample t-test was used to determine the difference in helmet performance (commercial helmets vs prototype helmet) between the averages of the 19 cases for all measures (MPS, resultant, and dominant component of linear and rotational acceleration).

Further analyses were conducted using independent sample t-tests to determine differences for each case (n=19) between the commercial helmet worn during the impact event and the prototype helmet for each dependent variable.

**Results**

The MPS values and the peak resultant and dominant coordinate component of linear acceleration and the MPS values were averaged for the nineteen head-to-head professional American football concussive impact reconstruction cases (3 impact trial per case) and are listed in table 4. The MPS values and the averaged peak resultant and dominant components of rotational acceleration of the nineteen head-to-head professional American football concussive impact reconstructions are listed in table 5.

No significant difference in MPS (p=0.10) and; the averaged peak resultant linear acceleration (p=0.89), peak dominant linear acceleration (p=0.87), peak resultant rotational acceleration (p=0.84), and peak dominant rotational acceleration (p=0.60) were observed between commercial helmets versus the prototype helmet.

| Linear Acceleration | | | | | | |
|---|---|---|---|---|---|---|
| Average of 19 cases | Commercial Helmets | | | Prototype Helmet | | |
| | MPS (SD) | Resultant g (SD) | Dominant g (SD) | MPS (SD) | Resultant g (SD) | Dominant g (SD) |
| | 0.330 (0.015) | 68.8 (5.3) | 62.1 (6.5) | 0.310 (0.012) | 67.9 (2.6) | 62.4 (2.9) |

Study 4 - Table 4: Average linear acceleration and MPS data for the Commercial and Prototype helmets for 19 Reconstruction cases of concussive impacts in professional American football.

| Rotational Acceleration | | | | | | |
|---|---|---|---|---|---|---|
| Average of 19 cases | Commercial Helmets | | | Prototype Helmet | | |
| | MPS (SD) | Resultant $rad/s^2$ (SD) | Dominant $rad/s^2$ (SD) | MPS (SD) | Resultant $rad/s^2$ (SD) | Dominant $rad/s^2$ (SD) |
| | 0.330 (0.015) | 4601.8 (352.9) | 3876.5 (370.9) | 0.310 (0.012) | 4575.5 (316.2) | 4013.6 (392.3) |

Study 4 - Table 5: Average rotational acceleration and MPS data for the Commercial and Prototype helmets for 19 Reconstruction cases of concussive impacts in professional American football.

**For each case**, the MPS, the peak resultant linear acceleration, and peak dominant coordinate component of linear acceleration values are listed in table 6. The peak resultant rotational acceleration, peak dominant coordinate component of rotational acceleration, and MPS values of the nineteen head-to-head professional American football concussive impact reconstructions are listed in table 7.

There was a significant reduction **in MPS** using the prototype helmet in seven of the nineteen concussive impact reconstruction events (case 3 p=0.00, case 6 p=0.00, case 8 p= 0.05, case 9 p=0.01, case 13 p=0.02, case 14 p=0.00, and case 19 p=0.03). One event case resulted in a significant increase in MPS for the prototype helmet (case 5 p=0.03). Eleven concussive reconstruction events resulted in no significant differences in the magnitude of MPS between the commercial helmet and the prototype helmet (case 1 p=0.93, case 2 p=0.08, case 4 p=0.07, case 7 p=0.35, case 10 p=0.09, case 11 p=0.67, case 12 p=0.27, case 15 p=0.18, case 16 p=0.68, case 17 p=0.94 and case 18 p=0.54).

There was a significant reduction in **resultant linear acceleration** when using the prototype helmet in two of the concussive impact reconstruction events (case 3 p=0.02, case 7 p=0.00). Three event cases resulted in a significant increase in resultant linear acceleration for the prototype helmet (case 11 p=0.04, case 16 p=0.01, case 17 p=0.02). Fourteen concussive reconstruction events resulted in no significant differences in the magnitude of resultant linear acceleration between the commercial helmet and the prototype helmet (case 1 p=0.64, case 2 p=0.10, case 4 p=0.39, case 5 p=0.11, case 6 p=0.26, case 8 p=0.57, case 9 p=0.47, case 12 p=0.38, case 13 p=0.89, case 14 p=0.24, case 15 p=0.80, case 18 p=0.18, case 19 p=0.48).

There was a significant reduction in **dominant linear acceleration** when using the prototype helmet in one of the concussive impact reconstruction events (case 7 p=0.001). Two event cases resulted in a significant increase in dominant linear acceleration for the prototype helmet (case 2 p=0.03, case 10 p=0.04). Sixteen concussive reconstruction events resulted in no significant differences in the magnitude of dominant linear acceleration between the commercial helmet and the prototype helmet (case 1 p=0.75, case 3 p=0.64, case 4 p=0.41, case 5 p=0.10, case 6 p=0.19, case 8 p=0.62, case 9 p=0.09, case 11 p=0.06, case 12 p=0.37, case 13 p=0.30, case 14 p=0.55, case 15 p=0.27, case 16 p=0.40, case 17 p=0.18, case 18 p=0.26, case 19 p=0.37).

There was a significant reduction in **resultant rotational acceleration** when using the prototype helmet in seven of the concussive impact reconstruction events (case 2 p= 0.040, case 3 p=0.01, case 6 p=0.04, case 7 p=0.01, case 13 p=0.01, case 14 p=0.01, case 19 p=0.02). Five event case resulted in a significant increase in resultant rotational acceleration for the prototype helmet (case 5 p=0.03, case 10 p=0.02, case 11 p=0.03, case 16 p=0.02, case 17 p=0.04). Seven concussive reconstruction events resulted in no significant differences in the magnitude of resultant rotational acceleration between the commercial helmet and the prototype helmet (case 1 p=0.07, case 4 p=0.65, case 8 p=0.44, case 9 p=0.19, case 12 p=0.44, case 15 p=0.26, case 18 p=0.53).

There was a significant reduction in **dominant rotational acceleration** when using the prototype helmet in seven of the concussive impact reconstruction events (case 3 p=0.008, case 6 p=0.02, case 8 p=0.03, case 9 p=0.03, case 13 p=0.00, case 14 p=0.00, case 19 p=0.00). Three event cases resulted in a significant increase in dominant rotational acceleration for the prototype helmet (case 5 p=0.033, case 10 p=0.01, case 11 p=0.04). Nine concussive reconstruction events resulted in no significant differences in the magnitude of dominant rotational acceleration between the commercial helmet and the prototype helmet (case 1 p=0.21, case 2 p=0.07, case 4 p=0.09, case 7 p=0.56, case 12 p=0.12, case 15 p=0.14, case 16 p=0.09, case 17 p=0.38, case 18 p=0.15).

| Linear Acceleration | | | | | | |
|---|---|---|---|---|---|---|
| Case | Commercial Helmets | | | Prototype Helmet | | |
| | MPS (SD) | Resultant (R) g (SD) | Dominant (D) g (SD) | MPS (SD) | Resultant (R) g (SD) | Dominant (D) g (SD) |
| 1 | 0.299 (0.003) | 50.4 (0.8) | 49.2 (1.3) | 0.300 (0.018) | 50.0 (1.1) | 49.6 (1.4) |
| 2 | 0.255 (0.009) | 41.6 (3.8) | 38.6 (1.1) | 0.237 (0.010) | 46.7 (1.5) | 43.3 (2.2) |
| 3 | 0.391 (0.002) | 65.0 (5.9) | 55.5 (21.6) | 0.350 (0.009) | 51.3 (0.5) | 49.2 (0.3) |
| 4 | 0.336 (0.005) | 62.7 (3.2) | 58.7 (2.5) | 0.347 (0.005) | 64.5 (0.4) | 60.2 (1.2) |
| 5 | 0.313 (0.005) | 63.2 (2.5) | 63.0 (2.5) | 0.348 (0.016) | 56.9 (4.6) | 56.6 (4.5) |
| 6 | 0.363 (0.011) | 58.3 (3.6) | 57.8 (3.7) | 0.314 (0.005) | 54.0 (4.4) | 52.8 (4.1) |
| 7 | 0.233 (0.028) | 53.0 (1.3) | 41.6 (2.0) | 0.214 (0.010) | 36.0 (0.6) | 30.4 (0.9) |
| 8 | 0.244 (0.037) | 42.8 (3.7) | 41.6 (4.0) | 0.183 (0.009) | 40.9 (3.7) | 39.8 (4.0) |
| 9 | 0.331 (0.021) | 69.8 (3.8) | 56.8 (3.2) | 0.268 (0.009) | 67.8 (2.2) | 61.8 (2.1) |
| 10 | 0.377 (0.013) | 49.8 (5.2) | 46.2 (4.9) | 0.393 (0.001) | 58.5 (1.9) | 55.6 (1.7) |
| 11 | 0.384 (0.008) | 63.4 (3.7) | 56.1 (4.9) | 0.395 (0.023) | 70.2 (1.1) | 65.0 (3.3) |
| 12 | 0.380 (0.021) | 84.6 (11.9) | 83.2 (10.9) | 0.361 (0.015) | 77.9 (1.2) | 76.9 (1.1) |
| 13 | 0.341 (0.024) | 72.8 (11.7) | 54.7 (7.7) | 0.284 (0.011) | 73.9 (3.9) | 60.5 (3.3) |
| 14 | 0.325 (0.009) | 94.5 (10.9) | 84.4 (10.0) | 0.282 (0.009) | 103.9 (4.6) | 88.7 (5.4) |
| 15 | 0.294 (0.025) | 80.7(5.0) | 72.6 (6.0) | 0.261 (0.024) | 81.7 (3.7) | 77.1 (1.2) |
| 16 | 0.463 (0.029) | 84.7 (2.3) | 75.9 (12.6) | 0.450 (0.044) | 94.6 (3.2) | 82.9 (3.0) |
| 17 | 0.434 (0.013) | 71.4 (2.1) | 67.4 (3.9) | 0.433 (0.014) | 78.8 (2.6) | 74.1 (6.0) |
| 18 | 0.157 (0.009) | 91.9 (12.4) | 77.2 (14.6) | 0.165 (0.015) | 80.0 (2.9) | 66.3 (3.3) |
| 19 | 0.354 (0.020) | 106.3 (7.6) | 99.3 (6.9) | 0.310 (0.011) | 102.3 (5.0) | 94.0 (6.1) |

$p \leq 0.05$ Significant Reduction from Commercial Helmet

$p \leq 0.05$ Significant Increase from Commercial Helmet

Study 4 - Table 6: Linear acceleration and MPS data for the Commercial and Prototype helmets for 19 Reconstruction cases of concussive impacts in professional American football.

| Rotational Acceleration | | | | | | |
|---|---|---|---|---|---|---|
| **Case** | **Commercial Helmets** | | | **Prototype Helmet** | | |
| | **MPS / (SD)** | **Resultant (R) rad/s² (SD)** | **Dominant (D) rad/s² (SD)** | **MPS / (SD)** | **Resultant (R) rad/s² (SD)** | **Dominant (D) rad/s² (SD)** |
| 1 | 0.299 (0.003) | 4136.9 (130.2) | 3054.7 (141.8) | 0.300 (0.018) | 4380.8 (114.2) | 3413.5 (100.5) |
| 2 | 0.255 (0.009) | 2817.9 (105.5) | 2732.7 (120.3) | 0.237 (0.010) | 2435.5 (195.0) | 2385.5 (213.1) |
| 3 | 0.391 (0.002) | 6469.1 (359.4) | 5295.8 (256.6) | 0.350 (0.009) | 5344.5 (114.6) | 4518.1 (104.0) |
| 4 | 0.336 (0.005) | 4072.1 (318.9) | 3785.9 (270.1) | 0.347 (0.005) | 4235.6 (480.4) | 4399.2 (882.9) |
| 5 | 0.313 (0.005) | 3421.5 (538.4) | 3245.6 (611.6) | 0.348 (0.016) | 4981.7 (615.3) | 4853.0 (624.4) |
| 6 | 0.363 (0.011) | 5175.4 (337.3) | 4354.4 (212.5) | 0.314 (0.005) | 4495.0 (204.5) | 3445.2 (290.6) |
| 7 | 0.233 (0.028) | 2844.8 (145.7) | 2409.6 (292.5) | 0.214 (0.010) | 2320.4 (113.4) | 2297.2 (93.3) |
| 8 | 0.244 (0.037) | 2990.1 (415.8) | 2628.8 (433.7) | 0.183 (0.009) | 3234.0 (267.5) | 2842.3 (606.7) |
| 9 | 0.331 (0.021) | 4727.1 (50.6) | 4328.1 (116.3) | 0.268 (0.009) | 4783.4 (36.1) | 3789.6 (270.5) |
| 10 | 0.377 (0.013) | 4369.6 (566.6) | 3708.8 (437.2) | 0.393 (0.001) | 5656.2 (151.9) | 5045.8 (170.3) |
| 11 | 0.384 (0.008) | 5258.2 (375.6) | 3983.3 (339.0) | 0.395 (0.023) | 6839.3 (210.9) | 5864.0 (415.5) |
| 12 | 0.380 (0.021) | 5362.3 (213.2) | 4641.8 (467.2) | 0.361 (0.015) | 5106.1 (477.6) | 4276.1 (247.7) |
| 13 | 0.341 (0.024) | 4439.3 (258.4) | 3351.8 (252.9) | 0.284 (0.011) | 3646.2 (159.2) | 3299.3 (454.4) |
| 14 | 0.325 (0.009) | 6155.1 (195.9) | 4578.1 (233.3) | 0.282 (0.009) | 3933.8 (276.7) | 3634.0 (182.0) |
| 15 | 0.294 (0.025) | 4233.4 (980.8) | 3701.8 (661.8) | 0.261 (0.024) | 3445.2 (339.0) | 3002.5 (260.2) |
| 16 | 0.463 (0.029) | 6015.4 (708.4) | 4956.3 (544.6) | 0.450 (0.044) | 7916.4 (391.6) | 6246.1 (866.6) |
| 17 | 0.434 (0.013) | 5314.3 (596.3) | 4362.1 (647.6) | 0.433 (0.014) | 7023.7 (733.6) | 6281.9 (1129.2) |
| 18 | 0.157 (0.009) | 3202.6 (116.7) | 2746.4 (646.6) | 0.165 (0.015) | 2931.5 (594.0) | 2200.4 (219.6) |
| 19 | 0.354 (0.020) | 6429.9 (290.9) | 5938.5 (117.1) | 0.310 (0.011) | 4226.2 (532.6) | 4070.6 (463.1) |

$p \leq 0.05$ Significant Reduction from Commercial Helmet

$p \leq 0.05$ Significant Increase from Commercial Helmet

Study 4 - Table 7: Rotational acceleration and MPS data for the Commercial and Prototype helmets for 19 Reconstruction cases of concussive impacts in professional American football.

## Discussion

The purpose of this study was to determine the ability of a prototype helmet to reduce MPS in comparison to the commercially worn helmets in 19 reconstructions of concussive head-to-head impact events in professional American football. Examination of the performance of the prototype helmet was also conducted using the components of linear and rotational acceleration (resultant and dominant**).** When the data was analyzed for the average of the 19 reconstruction cases (tables 4 and 5), no significant differences in MPS or the components of linear and rotational acceleration (resultant and dominant**)** were observed between the commercial helmets and the prototype helmet. This is likely a result of the variation of impact event characteristics (location, velocity) in concussive events in American football. The variation in input measures results in a corresponding large standard deviation of dynamic response and corresponding maximum principal strains results that result in statistically non-significant results. To determine how the prototype helmet performed in comparison to the manufacturer's helmet, a case by case analysis was performed to identify the conditions upon which the liner strategies may be effective at reducing MPS. This information may lead to further understanding of decoupling liner strategies and ways to improve impact management through helmet technology.

When analyzed on a case by case basis, the results indicate significant reductions in maximum principal strain occurred in seven cases (cases 3, 6, 8, 9, 13, 14, and 19) using the prototype helmet. For each of these cases a significant reduction in the dominant coordinate component of rotational acceleration also occurred. In case 5, where an increase in MPS was observed using the prototype helmet, an increase in the dominant coordinate component of rotational acceleration also occurred. In the other eleven cases (case 1, 2 4, 7, 10, 11, 12, 15- 18), no significant changes to MPS or dominant component of rotational acceleration occurred. Changes to the coordinate components of linear acceleration (resultant and dominant) and the other component of rotational acceleration (resultant) did not always result in corresponding changes in maximum principal strain. This becomes relevant given that current helmet designs primarily target reductions in peak resultant linear acceleration and more recently peak rotational acceleration. The results demonstrate changes to the resultant and/or coordinate components of linear or rotational acceleration are less likely to influence similar changes in MPS than the dominant component of acceleration. The results confirm design innovations that reduce dominant coordinate component of rotational acceleration offer potential to improve upon current helmet designs and reduce the risk of injury.

Several factors have been shown to influence the magnitude of an impact event and the resulting tissue strain value, such as impact location, impact duration and impact velocity.[18,50,69-73] Increased velocity has been demonstrated to increase the magnitude of the dynamic response and resulting MPS.[50,74] However, in this current study both significant reductions and increases in MPS were observed at the lower, mid and higher range of the velocities indicating velocity did not appear to affect the results. Liao (2016) analyzed head impact data from 33 Division I NCAA football players indicating significant association ($P = 0.017$) between head impact location and concussion. From the 19 concussive events included in this study, 14 were to the side of the head, 3 to the front and 2 to the crown.[75] The results of this study did not appear to be influenced by impact location as significant reductions in MPS. A strong correlation has been shown to exist between increased impact duration, higher brain tissue strains and corresponding risks of concussion.[1,29,73,76-80] For most of the impacts using the prototype helmet, impact duration decreased in comparison to the commercial helmet impacts. However, in these cases, significant increases and decreases in MPS were observed indicating impact duration did was not a factor in changes to MPS under these conditions.

**Conclusion**

The results of this study demonstrate the influence of a decoupling technology on maximum principal strain values for 19 reconstructed concussive events in professional American football. In each of the seven cases where MPS was significantly reduced a corresponding significant decrease in the dominant coordinate component of rotational acceleration was observed. Further reduction in MPS is anticipated through helmet design optimization improving the clinical significance of this decoupling strategy. The clinical significance of the reduction in MPS demonstrated was not studied within the paper, however, should be investigated in future research. The prototype helmet significantly increased the MPS magnitude in one case. For this case, the prototype resulted in a significant increase in the dominant coordinate component of rotational acceleration. There were eleven concussive impact reconstructions that resulted in no significant differences in MPS between the commercial and prototype helmets. For each of these cases no significant differences occurred in the dominant coordinate component of rotational acceleration.

There were no similar trends occurring between MPS and the components (resultant, dominant) of linear acceleration or the resultant component of rotational acceleration. The results of this study demonstrated targeting reductions in dominant coordinate component of rotational acceleration through helmet innovation, for various sports, can create significant decreases in maximum principal strain for reconstructions of concussive events.

***Limitations***

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. However, the use of a controlled laboratory setting for helmet impact performance testing is suitable because on-field impact scenarios and environmental conditions are highly variable.[73]

The Hybrid III headform is an anthropometric test device that has been used to simulate on-field impacts in sport (American football for this study). However, these devices have their limitations in that they do not fully replicated individual characteristics known to influence impact response such as impact awareness, the diversity of individual football player body differences, neck strength, etc.[81] However, the steel frame securely and safely houses the accelerometers and improved synthetic covering material are approaching human cadaver response but are rugged and reusable and result in reliable data for comparison.[3,82] Currently, the Hybrid III headform is the standard for head impact reconstructions measuring linear and rotational acceleration.

The finite element model used to calculate maximum principal strain 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. However, it is a useful tool to compare between different impacts giving relative values rather than an absolute such as the case comparing the response of the two helmets for the concussive impact events in this study.

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**References:**


1. **Vetter L,** Vanderby R. The Influence of materials and structure on performance of a football helmet. Polym. Eng. Sci. 1987; 27(15):1113-1120.
2. **Wennberg RA,** Tator CH. National Hockey League reported concussions, 1986–87 to 2001–02. Can J Neurol Sci. 2003; 30(3): 206–209.
3. **Pellman EJ,** Viano DC, Tucker AM, et al. Concussion in professional football: reconstruction of game impacts and injuries. J. Neurosurgery. 2003; 53(4):799-814.
4. **Casson IR,** Viano DC, Powell JW, et al. Twelve years of National Football League concussion data. Sports Health. 2010; 2(6): 471–483.
5. **McCrea M,** Hammeke T, Olsen G, et al. Unreported concussion in high school football players: implications for prevention. Clin J Sports Med. 2004; 14(1):13-17.
6. **Langlois JA**, Rutland-Brown W, Wald MM. The epidemiology and impact of traumatic brain injury: A brief overview. J. Head Trauma Rehab. 2006; 21(5):375-378.
7. **Collins CL,** Comstock RD, Dick RW, et al. Concussions among United States high school and collegiate athletes. J Athletic Train. 2007; 42(4):495-503.
8. **Broglio SP,** et al. Head impacts during high school football: A biomechanical assessment. J Athletic Train. 2009; 44(4):342-349.
9. **Crisco JJ,** Fiore R, Beckwith JG, et al. Frequency and location of head impact exposures in individual collegiate football players. J Athletic Train. 2010; 45(6): 549-559.
10. **Rowson S**, Duma SM, Beckwith JG, et al. Rotational head kinematics in football impacts: An injury risk function for concussion. Ann. Biomed. Eng. 2012; 40(1): 1 – 13.
11. **Delaney JS,** Lacroix VJ, Leclerc S, et al. Concussions among university football and soccer players. Clin. J. Sports Med. 2002; 12:331-338.
12. **Gessel LM,** Fields SK, Collins CL, et al. concussions among United States high school and collegiate athletes. J Athl Train. 2007; 42(2):495-503.
13. **Bailes JE,** Petragalia AL, Omalu BI, et al. Role of subconcussion in repetitive mild brain injury. J. Neurosurgery. 2013; 119: 1235-1245
14. **Unterharnscheidt F,** Higgins LS. Traumatic lesions of brain and spinal cord due to non-deforming angular acceleration of the head. Tex Rep Biol Med. 1969; 27(1):127–66.
15. **Ommaya AK,** Gennarelli TA. Cerebral concussion and traumatic unconsciousness. Brain. 1974; 97:633 – 654.
16. **Gennarelli TA,** Thibault LE, Adams JH, et al. Diffuse axonal injury and traumatic coma in the primate. Ann. Neurol. 1982; 12: 564-574.
17. **Adams JH,** Graham DI, Murray LS. Diffuse axonal injury due to non-missile head injury in humans: An analysis of 45 cases. Ann Neurol 1982; 12(6):557–63.
18. **Donnelly BR,** Medige J. Shear properties of human brain tissue. J. Biomed. Eng. 1997; 119:423-432.
19. **Kleiven S.** Finite element modeling of the human head. Department of Aeronautics, Royal Institute of Technology. S-100 44 Stockholm, Sweden. PhD Thesis. Report 2022-9, ISSN 0290-4646. 2002
20. **Takhounts EG,** Ridella SA. Investigation of traumatic brain injuries using the next generation of simulate injury monitor (SIMon) finite element head model. Stapp Car Crash Journal. 2008; 52: 1-31.
21. **Yoganandan N,** Li J, Zhang J, et al. Influence of angular acceleration-deceleration pulse shapes on regional brain strains. J. Biomech. 2008; 41:2253-2262.

22. **Greenwald RM,** Gwin JT, Chu JJ, et al. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 2008; 62(4):789 – 797.
23. **Delaney, J.S.,** Puni, V., Rouah, F. Mechanisms for concussion in university football, ice hockey and soccer: A Pilot Study. Clin. J. Sport Med. 2006; 16(2): 162 – 165.
24. **Cantu RC,** Mueller FO. Brain injury related fatalities in American football, 1945-1999 Neurosurgery – Special Edition. 2003; 52(4):846-853
25. **Meehan W,** D,Hemecourt P, Comstock RD. High school concussion in the 2008-2009 academic year. Am. J. Sports Med. 2010; 30(12):2405-2409.
26. **Bartsch A**, Benzel E, Miele V, et al. Impact test comparisons of 20th and 21st century American football helmets. J. Neurosurgery. 2012; 116: 222-233.
27. **Hoshizaki TB,** Karton C, Oeur RA, et. al. An examination of the current NOCSAE system and a new pneumatic ram method for evaluating American football helmet performance to reduce risk of concussion. J Sports engineering and Technology. 2016; 231(2):83-90
28. **Mills NJ,** Fitzgerald C, Gilchrist A, et al. Polymer foams for personal protection: cushions, shoes and helmets. Composite Science and Technology. 2003; 63:2390-2400.
29. **Gilchrist MD.** Modeling and accident reconstruction of head impact injuries. Adv. Eng. Mater. 2003; 245/245:417-429.
30. **Walsh ES,** Post A, Rousseau P, Kendall M, et. al. Dynamic impact response characteristics of a helmeted Hybrid III headform using a centric and non-centric impact protocol. J Sports Engineering and Technology. 2012; 226(3-4): 220-225.
31. **Post A**, Oeur A, Hoshizaki TB, et al. The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research. International Research Council on Biomechanics of Injury Conference (IRCOBI), 2012 Dublin, Ireland.
32. **Holbourn AHS**. The mechanics of brain injuries. *The Lancet.* Research Physicist, University Laboratory of Physiology and Department of Surgery, Oxford. 1943: 147 – 149.
33. **Viano DC,** Casson IR, Pellman EJ, et al. Concussion in professional football comparison with boxing head impacts. Neurosurgery. 2005; 57(6): 1154-1172.
34. **Zhang L,** Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury." J. Biomed. Eng. 2004; 126: 226 – 236.
35. **Walsh ES,** Rousseau P, Hoshizaki TB. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. Sports Eng. 2011; 13(3):135-143.
36. **Oukama R,** Pearsall DJ. Evaluation of a flexible force sensor for measurement of helmet foam impact performance. J. Biomech. 2011; 44(5): 904-909.
37. **Caserta GD,** Iannucci L, Galvanetto U. Shock absorption performance of a motorcycle helmet with honeycomb reinforced liner. Compos. Struct. 2011; 93: 2748 – 2759.
38. **Caccese V,** Ferguson JR, Edgecomb MA, Optimal design of honeycomb material used to mitigate head impact. Compos. Struct. 2013; 100: 404-412.
39. **Fowler JN,** Pallanta AA, Swanik CB, et al. The use of shear thickening nanocomposites in impact resistant materials. J. Biomed. Eng. 2015; 137:054505-1-5.
40. **Bailes JE,** Hudson V. Classification of sport-related head trauma: A spectrum of mild to severe injury. J Athl. Train. 2001; 36(3):236-243.
41. **Jadischke R,** Viano DC, Dau N, et al. On the accuracy of the head impact telemetry (HIT) system used in football helmets. J. Biomech. 2013; 46(13):2310-2315.

42. **Morrison B III**, Cater H L, Benham CD, Sundstrom LE. An in vitro model of traumatic brain injury utilizing two-dimensional stretch of organotypic hippocampal slice cultures. J. Neurosci. Methods. 2006. 150(2): 192-201
43. **Gurdjian ES,** Gurdjian ES. Cerebral contusions: Re-evaluation of the mechanism of their development. J. Trauma. 1976; 16(1):35-51.
44. **Viano DC,** King AJ, Melvin JW, et al. Injury biomechanics research: An essential element in the prevention of trauma. J. Biomech. 1989; 22(5):403-417.
45. **Newman JA,** Beusenberg MC, Shewchenko N, et al. Verification of biomechanical methods employed in a comprehensive study of mild traumatic brain injury and the effectiveness of American football helmets. J. Biomech. 2005; 38:1469-1481.
46. **Post A,** Oeur A, Hoshizaki TB. The influence of velocity on the performance range of American football helmets. Proceeding of the 1st International Conference on Helmet Performance and Design. London, UK. HPD-2013-10
47. **Post A,** Hoshizaki TB. Rotational acceleration, brain tissue strain, and the relationship to concussion. J. Biomech. Eng. 2015; 137:1-8
48. **Post A,** Koncan D, Kendall M, et al. Analysis of speed accuracy using video analysis software. Sports Eng. 2018; ISSN 1369-7072. DOI 10.1007/s12283-018-0263-4.
49. **Pellman EJ**, Viano DC, Withnall C, et al. Concussion in professional football: Helmet testing to assess impact performance – Part 11. J. Neurosurgery. 2006; 58:78 – 96.
50. **Post A,** Oeur A, Hoshizaki B, et al. An examination of American football helmets using brain deformation metrics associated with concussion. Mater. Des. 2013; 45:653-662.
51. **Rousseau P.** An analysis of dynamic concussion metrics associated with elite ice hockey elbow-to-head and shoulder-to-head collisions. PhD Thesis, University of Ottawa. 2014.
52. **Post A.** Oeur A, Hoshizaki TB, et al. Examination of the relationship between peak linear and angular accelerations to brain deformation in hockey helmet impacts. Comput. Methods Biomech. Biomed. Eng. 2011; 1-9.
53. **Pellman EJ,** et al. Concussion in professional football: epidemiological feature of game injuries and review of literature – Part 3. Neurosurgery. 2004; 54(1):81-94.
54. **Padgaonkar AJ,** Kreiger KW, King AI. Measurements of angular accelerations of a rigid body using linear accelerometers. J. Appl. Mech. 1975; 42: 552-556.
55. **Horgan TJ**, Gilchrist MD. The creation of 3D finite element models for simulating head impact biomechanics. Int. J. Crashworthiness. 2003; 8(4): 353 – 366.
56. **Horgan TJ,** Gilchrist MD. Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. Int. J. Crashworthiness. 2004; 9(4):401-418.
57. **Ruan JS,** Khahil T, King AI. Dynamic response of the human head to impact by three-dimensional Finite Element Analysis. J Biomech. Eng. 1994; 116:44-50.
58. **Zhou C,** Khalil T, King A. A new model comparing impact responses of the homogeneous and inhomogeneous human brain. In: Proceedings 39th Stapp Car Crash Conference; 1995; 121–37.
59. **Willinger R**, Taleb L, Pradoura P. Head biomechanics from the finite element model to the physical model. Proceedings from IRCOBP Brunnen, Switzerland 1995 : 245-260.
60. **Zhang L**. Computational biomechanics of traumatic brain injury, An investigation of head impact response and American football head injury. Doctor of Philosophy – Dissertation. Graduate School of Wayne State University, Detroit, Michigan. 2001.

61. **Kleiven S.** Predictors for traumatic brain injuries evaluated through accident reconstructions. Stapp Car Crash Journal. 2007; 51: 81-114.
62. .**Nahum AM,** Smith R, Ward CC. Intracranial pressure dynamics during head impact. 21st Annual Stapp Car Crash Conference. Society of Automotive Engineers. 1977.
63. **Hardy WN,** Foster CD, Mason MJ, et al. Investigation of head injury mechanisms using neutral density technology and high-speed biplanar x-ray. Stapp Car Crash Journal; The Stapp Association, Ann Arbor, Michigan. 2001.
64. **Trosseille X,** Tarrie´re C, Lavaste F, et al. Development of a F.E.M. of the human head according to a specific test protocol. In: 36th Stapp Car Crash Conference, Seattle, WA, USA, SAE paper 922527. 1992
65. **Doorly MC**, Gilchrist MD. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. Computer Methods in Biomechanics and Biomedical Engineering. 2006; 9(6): 371 – 377.
66. **Doorly MC,** Investigations into head injury criteria using numerical reconstructions of real life accident aases". PhD Thesis, University College Dublin, Dublin, Ireland. 2007.
67. **Forero Rueda MA**, Cui L, Gilchrist MD. Finite element modeling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. Comput. Methods Biomech. Biomed. Eng. 2010; 10(5), 1-11.
68. **Oeur RA,** Karton C, Post A, et al. A Comparison of head dynamic response and brain tissue stress and strain using accident reconstruction for concussion, Concussion with Persistent Postconcussive Symptoms and Subdural Hematoma. J. Neurosurgery. 2015; 123(2):415-422.
69. **Mills NJ,** Gilchrist A. Finite-element analysis of bicycle helmet oblique impacts". Int. J. Impact Eng. 2008; 35:1087-1101
70. **Broglio SP,** Surma T, Ashton-Miller JA. High school and collegiate football athlete concussions: a biomechanical review. Ann Biomed Eng. 2012; 10(1):37-46.
71. **Hoshizaki TB**, Post A, Oeur RA, et al. Current and future concepts in helmet and sports injury prevention" Neurosurgery. 2014; 75: Supplement**.**
72. **Post A,** Hoshizaki TB, Gilchrist MD, et al. Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. J. Biomech. 2017; 61:183-192.
73. **Krzeminski DE**, Fernando BMD, Curtzwiler G, et al. Repetitive impact exposure and characterization of stress-whitening of an American football helmet outer shell material. Polym. Test. 2016; 55:190-203.
74. **Rousseau P,** Post A, Hoshizaki TB. The effects of impact management materials in ice hockey helmets on head injury criteria. J sport Eng Tech. 2009; 233:159-165.
75. **Liao S,** Lynall RC, Mihalik JP. The effect of head impact location on day of diagnosed concussion in college football. Med. Sci. Sports Exercise. 2016; 48(7):1239-1243
76. **Gurdjian ES,** Hodgson VR, Thomas LM. Studies on mechanical impedance of the human skull:preliminary report. J. Biomech. 1970; 3: 239 – 247.
77. **Versace JA.** 1971 A review of the severity index. SAE Technical Paper 710881, doi:10.4271/710881.
78. **Rousseau P,** Hoshizaki TB. Defining the effective impact mass of elbow and shoulder strikes in ice hockey. Sports Biomechanics. 2015;1-11.

79. **Kendall, M.** Comparison and characterization of different concussive brain injury events. Doctorate in Philosophy Thesis. Department of Human Kinetics, Faculty of Health Sciences. University of Ottawa, Ottawa, Canada. 2016
80. **LaPlaca MC,** Cullen DC, McLoughlin JJ, Cargill RS II. 2005. High rate shear strain of three-dimensional neural cell cultures: A new in vitro traumatic brain injury model. J. Biomech. 38(5):1093–1105.
81. **McAllister TW,** Ford JC, Songbai J, et al. Maximum Principal Strain and Strain Rate Associated with Concussion Diagnosis Correlates with Changes in Corpus Callosum White Matter Indices. Ann. Biomed. Eng. 2012;40(1):127-140.
82. **Lamy M,** Baumgartner D, Yoganandan N, Stemper BD, Willinger R. 2013 Experimentally validated three-dimensional finite element model of the rat for mild traumatic brain injury. Med. Biol. Eng. Comput. 51(3):353–365

# PART III
## Global discussion and conclusion

The research for this thesis involved four studies, the first study conducted performance testing of a current commercially available American football helmet to determine the influence of the helmet on direction components of dynamic response and the effect to the magnitude of maximum principal strain. The second study implemented a method to mathematically modify twenty impacts to demonstrate the concept of 3 impact force redirection and 7 redistribution strategies and their influence on maximum principal strain. The third study investigated how novel design principles and decoupling strategies could influence reductions in maximum principal strain values in comparison to an existing commercially available helmet. Finally, the fourth study tested the effectiveness of a helmet designed to decouple the helmet head interface and reduce maximum principal strain under concussive impact reconstructions.

The **first study** utilized a comprehensive testing protocol of 20 impact conditions (5 impact sites and 4 directions) and 4 impact velocities to evaluate the influence of the American football helmet on directional components of dynamic response and the resulting magnitude of maximum principal strain. The results indicate that the resultant and dominant components of linear and rotational acceleration correlate highly with maximum principal strain values regardless of how the data is analyzed. However, the correlation between the X, Y, and Z components of dynamic response vary dependent on how the data is analyzed. When all impact conditions are collapsed there is a low to medium correlation to MPS for each component as a washout effect occurs due to variation in impact location. The correlations shift to high or low when viewing the data by location which indicates the strong directional importance if head motion at each impact location. When viewing the data by impact condition all components are highly correlated to MPS due to the positive correlation with velocity for each impact condition. In conclusion, the first study established a novel method for helmet performance evaluation using the coordinate components of dynamic response to indicate overall brain motion in relation to maximum principal strain. This study, with the use of coordinate components of acceleration and a comprehensive impact testing protocol, indicated the potential benefits for helmet design to incorporate the dominant coordinate component of acceleration as a targetable measure, which is more feasible than using the resultant, for maximum principal strain reduction. Additionally, the study identifies impact location specific protection could be addressed using the associated directional coordinate accelerations as metrics to reduce maximum principal strain. Finally, the study identifies the overall significance of investigating directional brain motion in terms of helmet protection aimed at reducing maximum principal strain values.

The **second study** mathematically modified 20 impact conditions and utilized finite element modeling to demonstrate the concept of impact force redirection and redistribution and the resulting effect to maximum principal tissue strain. The results indicate reductions in maximum principal strain can be achieved using impact force redirection strategies that reduce the dominant coordinate component of rotational acceleration. The results are consistent with past research indicating linear acceleration does not always correlate well to tissue deformation but good correlation exists between rotational acceleration and brain tissue deformation. When analyzed by impact location, the results indicate that reductions in maximum principal strain using rotational redirection and or redistribution liner strategies can be made at the front boss, side and rear locations. When analyzed by impact condition, the results show both reductions and increases in maximum principal strain occur, dependent on impact condition and the implemented redirection or redistribution liner strategies. The reason for the increased MPS under certain loading conditions is unclear within this study and warrants further investigation. This research has identified redirection and redistribution strategies offer a potential solution to reduce maximum principal strain. However, it is important to understand the process of force redirection and redistribution under the sport specific loading conditions to achieve optimum results as incorrect use can lead to increased rather than reduced risk. This study identified the importance of investigating the influence of redirection or redistribution strategies on maximum principal strain before being incorporated.

The **third study** was conducted to demonstrate a design principal aimed to decouple the head and helmet interface could improve helmet performance to lower the maximum principal strain values using a comprehensive testing protocol. The helmet incorporating decoupling design strategies significantly reduced maximum principle strain in comparison to the commercial. When analyzed by impact location, the prototype helmet resulted in significant reductions in maximum principal strain at the front and front boss locations. The results of this study indicate that reductions in risk of injury can be made through decoupling strategies aimed at redirection impact energy in American football. The information gained from this study can be applied to other sporting helmets if tested and designed using a similar protocol.

The **fourth study** tested the effectiveness of a helmet designed to decouple the helmet head interface and reduce maximum principal strain under concussive impact reconstructions. The helmet was evaluated against manufactured helmets for its ability to reduce maximum principal strain in 19 reconstructions of concussive head-to-head impacts in American football. The results of this study indicate reductions in maximum principal strain occurred using the prototype helmet when significant reductions in the dominant coordinate component of rotational acceleration occurred. Additionally, increases in MPS occurred when increases in the dominant coordinate component of rotational acceleration were observed. Reductions or increases to coordinate components of linear acceleration did not always result in corresponding changes in maximum principal strain. In conclusion, for each case where MPS was significantly reduced or increased, a corresponding significant reduction or increase in the dominant coordinate component of rotational acceleration was observed. The results of this study indicate targeting reductions in dominant coordinate component of rotational acceleration through helmet innovation can result in significant decreases maximum principal strain for reconstructions of concussive events. The results indicate decoupling strategies can increase both acceleration and MPS if not implemented properly. This information is important in the advancement of helmet technology and reducing brain trauma loading. This information can be used to optimize helmet performance and reduce MPS values by reducing dominant coordinate component of rotational acceleration for all impact scenarios. This information should be utilized in helmet design for other various sports if investigated an implemented appropriated for the impact conditions common to that sport.

## Summary

All four studies completed for this thesis dealt with aspects of helmet performance in American football with the goal of reducing maximum principal strain. The first study evaluated the performance of a current American football helmet to determine the influence on three-dimensional brain motion and the effect the maximum principal strain. The second study demonstrated the theory of redistribution and redirection impact force strategies using mathematics to modify twenty impacts and determine the influence of each strategy on maximum principal strain values. The third study demonstrated the use of design principals and a decoupling strategy to improve helmet performance to reduce maximum principal strain values. The fourth study tested a helmet designed to decouple the head-helmet interface and reduce maximum principal strain for reconstructions of concussive impacts in professional football.

This research has provided information regarding testing methodologies and design principles to reduce maximum principal strain using decoupling liner strategies in American football helmets. More specifically, this research has identified the importance of directional motion of the brain in relation to maximum principal strain. Additionally, the work has shown that targeting reductions in dominant coordinate components of rotational acceleration though decoupling helmet liner strategies can further improve helmet design to reduce MPS. The procedures, testing protocols, design principles and strategies implemented within this thesis can be used to inform future protective equipment, particularly helmet testing and design, to advance helmet technology and innovation in various sports.

Future work could optimize the decoupling innovation strategy to increase performance and ensure reductions in the dominant coordinate component of rotational acceleration under all impact conditions. Studies investigating the performance of novel technologies for impact velocities greater than 8.0 m/s would also be useful, as concussive injury has been observed at higher ranges. It would therefore be prudent to include them in further testing to ensure adequate protection to players. However, the results of this study clearly indicate the link between dominant coordinate components of rotational acceleration and maximum principle strain. This information is important in the advancement of helmet technology and reducing brain trauma loading.

## References


1. **Aare M,** Kleiven S, Halldin P. (2003). Injury criteria for oblique helmet impacts. international research council on biomechanics of injury conference (IRCOBI), Dublin, Ireland: 349-350.
2. **Adams DB.** (1982) Wound ballistics: a review. Military Medicine. 147: 831-834.
3. **Alghamadi AAA.** (2001). Collapsible impact energy absorbers: An overview. Thin Walled Structures. 39:189 – 213.
4. **Alves W,** Macciocchi SN, Barth JT. (1993). Postconcussive symptoms after uncomplicated mild head injury. Journal of Head Trauma Rehabilitation. 8: 48- 59.
5. **Arani A**, Murphy MC, Glaser KJ, Manduca A, Lake DS, Kruse SA, Jack CR, Ehman RL, Huston J. (2015). Measuring the effects of aging and sex on regional brain stiffness with MR elastography in healthy older adults. NeuroImage. 111: 59 – 64.
6. **Bailes JE,** Hudson V. (2001). Classification of sport-related head trauma: A spectrum of mild to severe injury. Journal of Athletic Training. 36(3):236-243.
7. **Bailes JE**, Cantu RC. (2001b). Head injury in athletes. Journal of Neurosurgery. 48(1) January 2001.
8. **Bain AC,** Meaney DF. (2000). Tissue level Thresholds for Axonal Damage in an Experimental Model of Central Nervous System White Matter Injury. J Biomech Eng. 122(6):615-622.
9. **Bandak FA.** (1995). On the Mechanics of Impact Neurotrauma: A Review and Critical Synthesis. Journal of Neurotrauma, 12(4):635-49.
10. **Bartsch A,** Benzel E, Miele V, Prakash V. (2012). Impact test comparisons of 20$^{th}$ and 21$^{st}$ century American football helmets. Journal of Neurosurgery. 116:222-233.
11. **Beckwith JG,** Zhao W, Songbai J. et al. (2018). Estimated brain tissue response following impacts associated with and without diagnosed concussion. Annals of Biomedical Engineering. [Published online February 2018].
12. **Berger RE**. (1976). Considerations in developing test methods for protective headgear product safety engineering section. Centre for Consumer Project Technology, U.S. Department of Commerce.
13. **Bishop PJ.** Norman RW, Kozey JW. (1984). An evaluation of football helmets under impact conditions. The American Journal of Sports Medicine. 12(3): 233-236.
14. **Broglio SP,** Sosnoff JJ, Shin SH, He Z, Alcaraz C, Zimmerman J. (2009) Head impacts during high school football: A Biomechanical Assessment. Journal of Athletic Training. 44 (4): 343-349
15. **Broglio SP,** Schnebel B, Sosnoff JJ, Shin S, Feng X, He X, Zimmerman J. (2010). Biomechanical properties of concussions in high school football. Medicine & Science in Sports & Exercise. 2064 – 2071
16. **Broglio SP,** Surma T, Ashton-Miller JA. (2012). High school and collegiate football athlete concussions: a biomechanical review. Annals of Biomedical Engineering. 10(1):37-46.
17. **Budday S**, Nay R, Wyrobek T, Ovaert TC, Kuhl E. (2015). Mechanical properties of grey and white matter brain tissue by indentation. Submitted to Acta Biomateriala (Sept 2014).
18. **Caccese V,** Ferguson JR, Edgecomb MA, (2013). Optimal design of honeycomb material used to mitigate head impact. Composite Structures 100:404-412.
19. **Cantu RC,** Mueller FO. (2003) Brain injury related fatalities in American football, 1945-1999. Neurosurgery – Special Edition. 52(4):846-853.
20. **Caserta GD,** Iannucci L, Galvanetto U. (2011). Shock absorption performance of a motorcycle helmet with honeycomb reinforced liner. Composite Structures 93:2748 – 2759.

21. **Casson IR,** Viano DC, Powell JW, Pellman EJ. (2010). Twelve years of National Football League concussion data. Sports Health. 2(6):471–483.
22. **Christ AF,** Franze K, Gautier H, Moshayedi P, Fawcett J, Franklin RJM, Karadottie RT, Guck J (2010). Mechanical difference between white and gray matter in the rat cerebellum measured by scanning force microscopy. Journal of Biomechanics. 43:2986-2992.
23. **Clark MD,** Breton MA, Marshall SW, Guskiewicz KM. (2017). Descriptive characteristics of concussion in National Football League games. American Journal of Sports Medicine. 45(4):929-936.
24. **Collins CL,** Comstock RD, Dick RW. Fields SK, Gessel LM. (2007). Concussions among United States high school and collegiate athletes. Journal of Athletic Training. 42(4):495-503
25. **Crisco JJ,** Chu JJ, Greenwald RM. (2004). An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single axis accelerometers. Journal of Biomedical Engineering. 126:849–854.
26. **Crisco JJ,** Fiore R, Beckwith JG, Chu J.J. Brolinson PG, Duma S, Greenwald RM (2010) Frequency and Location of Head Impact Exposures in Individual Collegiate Football Players. Journal of Athletic Training. 45(6):549-559
27. **Daneshvar DH**, Nowinski CJ, McKee AC, Cantu RC. (2011) The Epidemiology of Sport Related Concussion. Clin Sports Med. 30:1-17.
28. **Darling T.,** Muthuswamy J., Rajan SD. (2016) Finite Element Modeling of Human Brain Response to Football Impacts. Computer Methods in Biomechanics and Biomedical Engineering. 19(13):1432-1442.
29. **Delaney JS,** Lacroix VJ, Leclerc S, Johnston KM. (2002) Concussions Among University Football and Soccer Players. 12:331-338.
30. **Donnelly BR,** Medige J. (1997) Shear Propertoes of Human Brain Tissue. Journal of Biomechanical Engineering. 119:423-432.
31. **Doorly MC**, Gilchriste MD. (2006). The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. Computer Methods in Biomechanics and Biomedical Engineering. 9(6):371 – 377.
32. **Doorly MC.** (2007). Investigations into head injury criteria using numerical reconstruction of real life accident cases [PhD thesis]. University College Dublin.
33. **Duma SM,** Manoogian SJ, Bussone, WR, et. al. (2005) Analysis of real-time head accelerations in collegiate football players. Clinical Journal of Sports Medicine. 15(1):3-8.
34. **Edberg S,** Rieker J, Angrist A. (1963) Study of impact pressure and acceleration in plastic skull models. Labs Invest. 12:1305 – 1311.
35. **Elkin BS,** Morrison B. (2007). Region-specific tolerance criteria for the living brain. Strapp Car Crash Journal. 51.
36. **Elkin BS**, Ilankova A, Morrison B. (2010). Dynamic, regional mechanical properties of the porcine brain: indentation in the coronal plane. Journal of Biomechanical Engineering. 133:1-7.
37. **Fallenstein JW,** Hulce VD, Melvin JW. (1969). Dynamic mechanical properties of human brain tissue. Journal of Biomechanics. 2:217 – 226.
38. **Fahlstedt M,** Baeck K, Halldin P, Vander Sloten J, Goffin J, Depreitere B, Kleiven S. (2012). Influence of impact velocity and angle in a detailed reconstruction of a bicycle accident; IRCOBI Conference 2012, IRC-12-84:787 – 799.

39. **Faul M,** Xu L, Wald MM, et al. (2010). Traumatic brain injury in the United States: emergency department visits, hospitalizations and deaths 2002-0226. Altalnta (GA): Centres for Disease Control and Prevention, National Centre for Injury Prevention and Control
40. **Feng Y,** Clayton EH, Chang Y, Okamoto RJ, Bayly PV. (2013) Viscoelastic properties of the ferret brain measured in vivo at multiple frequencies by magnetic resonance elastography. Journal of Biomechanics. 46:863-870.
41. **Finan JD,** Nightingale RW, Myers BS. (2008). The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts. Traffic Injury Prevention. 9:483–488.
42. **Finklestein E**, Corso P, Miller T. (2006). The incidence and economic burden of injuries in the Unites States. New York: Oxford University Press.
43. **Forero Rueda MA**, Cui L, Gilchrist MD. (2010). Finite element modeling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. Computer Methods in Biomechanics and Biomedical Engineering. 10(5):1-11.
44. **Galbraith J,** Thibault L, Matterson D. (1993). Mechanical and electrical responses of the squid giant axon to simple elongation. Journal of Biomedical Engineering. 115:13–22.
45. **Gadd CW.** (1966). Use of a weighted impulse criterion for estimating injury hazard. In Proceedings of the 10th STAPP Car Crash Conference, SAE paper No. 660793.
46. **Gennarelli TA**, Ommaya AK, Thibault LE. (1971). Comparison of translational and rotational head motions in experimental cerebral concussion. In 15th Stapp Car Crash Conference Proceedings, 17-19 November 1971, Coronado, Calif., pp. 797-803. New York: Society of Automotive Engineers.
47. **Gennarelli TA,** Thibault LE, Adams JH, Graham DI, Thompson CJ, Marcincin RP. (1982) Diffuse axonal injury and traumatic coma in the primate. Annals of Neurology. 12:564-574.
48. **Gennarelli TA,** Thibault LE, Tomei G, (1987). Directional Dependence of Axonal Brain Injury Due to Centroidal and Non-centroidal Acceleration. 31st Stapp Car Crash Conference. SAE paper no. 872197. Society of Automotive Engineers.
49. **Gessel LM,** Fields SK, Collins CL, Dick RW, Comstock RD. (2007). Concussions among United States high school collegiate athletes. Journal of Athletic Training. 42(4):495–503.
50. **Gilchrist MD.** (2003). Modelling and accident reconstruction of head impact injuries. Engineering Materials. 245/245:417-429
51. **Gimbel GM,** Hoshizaki TB. (2008). Compressive properties of helmet materials subjected to dynamic impact loading of various energies. European Journal of Sport Science. DOI: org/10.1080/17461390802438763.
52. **Giordano C,** Cloots RJH, van Dommelen JAW, Kleiven S. (2014). The Influence of Anisotropy on Brain Injury Prediction. Journal of. Biomech, 47: 1052-1059
53. **Greenwald RM,** Gwin JT, Chu JJ, Crisco JJ. (2008). Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery 62(4):789–797.
54. **Gurdjian ES** Lissner HR. (1944). Mechanisms of head injury as studied by the cathode Ray oscilloscope: Preliminary report. Journal of Neurosurgery and Psychiatry. 1:393-399.
55. **Gurdjian ES,** Lissner HR, Webster JE, Latimer FR, Haddad BF. (1954). Studies on experimental concussion: relation of physiologic effect to time duration of intracranial pressure increase at impact. Neurology. 4:674–681.
56. **Gurdjian ES,** Roberts VL, Thomas LM. (1966). Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. The Journal of Trauma. 6(5):600-606.

57. **Gurdjian ES,** Hodgson VR, Thomas LM, Patrick LM. (1968). Significance of relative movements of scalp, skull, and intracranial contents during impact injury of the head.
58. **Gurdjian ES,** Gurdjian ES. (1975). Re-evaluation of the biomechanics of blunt impact injury of the head. Surgery, Gynecology and Obstetrics. 140(6):845-850.
59. **Gurdjian ES.** (1978). Movement of the brain and brain stem from impact induced linear and angular acceleration. Trans. Am Neurological Association. 99:248-249.
60. **Guskiewicz KM,** Weaver NL, Padua DA, Garrett WE. (2000). Epidemiology of concussion in collegiate and high school football players. The American Journal of Sports Medicine 28(5):643-650.
61. **Guskiewicz KM,** Mihalik JP, et al. (2007). Measurement of head impacts in collegiate football players: Relationship between head impact biomechanics and acute clinical outcome after concussion. Neurosurgery. 62(6):1244-1252.
62. **Gray H.** (1991). Anatomy descriptive and surgical. St Louis, Mosby Year Book J.W. Parker and Son © 1858.
63. **Green MA,** Bilston LE, Sinkus R. (2008). In vivo brain viscoelastic properties measured by magnetic resonance elastography. NMR in Biomedicine. 21(7):755–764.
64. **Gur RC,** Gunning-Dixon, FM, Turetsky BI, Bilker WB, Gur RE. (2002). Brain region and sex differences in age association with brain volume. American Journal of Geriatric Psychiatry. 10:72–80.
65. **Hardy WN,** Khalil TB, King AI. (1994) Literature review of head injury biomechanics. International Journal of Impact Engineering. 15(4):561-586.
66. **Hardy WN,** Mason MJ, Foster CD, Shah CD, Kopacz JM, Yang KH, King AI, Bishop J, Bey M, Anderst A, Tashman S. (2007). A study of the response of the human cadaver head to impact. Strapp Car Crash Journal, 51:7-70.
67. **Hodgson MS,** Gurdjian ES, Thomas LM. (1966). Experimental skull deformation and brain displacement demonstrated by flash X-Ray technique. Department of Engineering Mechanics, College of Engineering, and Department of Neurosurgery, College of Medicine, Wayne State University, Detroit, Michigan.
68. **Hodgson VR.** (1980). Reducing serious injury in sports. Intrascholastic Athletic Association. 7(2).
69. **Hodgson VR,** Thomas LM, Khalil TB. (1985). The role of impact location in reversible cerebral concussion. 27th Stapp Car Crash Conf. SAE paper no. 831618. Society of Automotive Engineers, 225–240.
70. **Holbourn AHS** (1943). The mechanics of brain injuries. The Lancet. Research Physicist, University Laboratory of Physiology and Department of Surgery, Oxford. 147–149.
71. **Hoshizaki TB,** Brien SE. (2004). The science and design of head protection in sport. Journal of Neurosurgery. 55(4): 956-976. DOI: 10.1227/01. NEU. 0000137275.502.46.OB.
72. **Hoshizaki TB,** Post A, Kendall M, Karton C, Brien S. (2013). The relationship between head impact characteristics and brain trauma. Neurology and Neurophysiology. 5(1):1-8.
73. **Hoshizaki TB,** Post A, Oeur RA, Brien SE. (2014). Current and future concepts in helmet and sports injury prevention. Neurosurgery. 75(4):5136-5148.
74. **Horgan TJ,** Gilchrist MD. (2004). Influence of FE model variability in predicting brain motion and intracranial pressure changes in head impact simulations. International Journal of Crashworthiness. 9(4):401-418.

75. **Hrapko M,** Van Dommelen JA, Peters GW, Wismans JS. (2008). The influence of test conditions on characterization of the mechanical properties of brain tissue. Journal of Biomechanical Engineering. 130(3):1-10.
76. **Jin X,** Zhu F, Mao H, Shen M., Yang K. (2013). A comprehensive experimental study on material properties of human brain tissue. Journal of Biomechanics. 46.2795 – 2801.
77. **Johnson C,** McGarry MDJ, Gharibans AA, Weaver JB, Paulsen KD, Wang H, Olivero WC, Sutton BP, Georgiadia JG. (2013). Local mechanical properties of white matter structures in the human brain. NeuroImage. 79:145 – 152.
78. **Johnston JM,** Ning H, Kim JE, Kim YH, Soni B, Reynolds R, Cooper L, Andrews JB, Vaidya U. (2015) Simulation, fabrication and impact testing of a novel football padding system that decreases rotational acceleration. Sports Engineering. International Sports Engineering Association. 18:11-20.
79. **Karton C**, Hoshizaki TB, Gilchrist M.D. (2013). The influence of inbound mass on the dynamic impact response of the Hybrid III headform and tissue deformation response characteristics. Journal of ASTM international.
80. **Kendall M**, Post A, Rousseau P, Oeur A, Gilchrist M. (2012). 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. Proceedings of the IRCOBI conference in Dublin, Ireland.
81. **Kendell M,** Post A, Zanetti K, Hoshizaki TB, Gilchrist MD. (2013). Comparison of dynamic versus static head impact reconstruction methodology by means of dynamic impact response and brain deformation metrics. International Research Council on Biomechanics of Injury Conference (IRCOBI), Gothenberg, Sweeden.
82. **Kendall M.** (2016) Comparison and characterization of different concussive brain injury events. Doctorate in Philosophy Thesis. Department of Human Kinetics, Faculty of Health Sciences. University of Ottawa, Ottawa, Canada.
83. **Kerr ZY,** Collins CL, Mihalik JP, Marshal SW, Guskiewicz KM, Comstock RD. (2014). Impact locations and concussion outcomes in high school football player-to-player collisions. Pediatrics. 134(3):489-496.
84. **Kindt GW,** Scheider RC, Robinson JL. (1973). A method of comparing the impact absorption characteristics of football helmets. Medicine and Science in Sports 3(4):203-209.
85. **King AI,** Ruan JS, Zhou C, Hardy WN, Khalil TB. (1995). Recent advances in biomechanics of brain injury: a review. Journal of Neurotrauma. 12(4):651 – 658.
86. K**ing AI,** Yang KH, Zhang L, Hardy W, Viano DC. (2003). Is head injury caused by linear or angular acceleration? In: Proceedings of the International Research Conference on the Biomechanics of Impact (IRCOBI).
87. **Kleiven S, (**2003). Influence of impact direction on human head in prediction of subdural hematoma. Journal of Neurotrauma 20(4):365-379.
88. **Kleiven S** (2007). Predictors for traumatic brain injuries evaluated through accident reconstructions. Stapp Car Crash Journal. 51:81-114.
89. **Kurtcuoglu V,** Poulikakos D, Ventikos Y. (2005). Computational modeling of the mechanical behaviour of the cerebrospinal fluid system. Journal of Biomechanical Engineering. 127:264-269.
90. **Lamb L,** Hoshizaki TB. (2009). Deformation mechanisms and impact attenuation characteristics of thin-walled collapsible air chambers used in head protection. Journal of Mechanical Engineering in Medicine. DOI:10.1243/09544119JEIM573.

91. **Langlois JA**, Rutland-Brown W, Wald MM. (2006). The epidemiology and impact of traumatic brain injury: A brief overview. Journal of Head Trauma Rehabilitation. 21(5):375-378
92. **Liao S,** Lynall RC, Mihalik JP. (2016). The effect of head impact location on day of diagnosed concussion in college football. Medicine & Science in Sports & Exercise. 48(7):1239-1243.
93. **Lloyd J,** Conidi F. (2014). Comparison of common football helmets in preventing concussion, hemorrhage and skull fracture using a modified drop test. Neurology 82:P5.320 (Abstract)
94. **Manavais J,** Blumbergs PC, Scott G, North JB. (1991). Brain injury patterns in falls causing death. In: Proceedings of the IRCOBI Conference. 77–88.
95. **Margulies SS,** Thibault LE, (1992). A proposed tolerance criteria for diffuse axonal injury. Journal of Biomechanics 25:917-923.
96. **Martini F,** Ober W, Nath J. (2017). Visual anatomy & physiology OLP with eText, Global Edition. 2nd Edition. Electronic Product ISBN12 :9781292069203
97. **McAllister TW,** Ford JC, Songbai J, Beckwith JG, Flashman LA, Paulsen K, Greenwald RM. (2011). Maximum principal strain and strain rate associated with concussion diagnosis correlates with changes in corpus callosum white matter indices. Annals of Biomedical Engineering. 40(1):127-140.
98. **McCracken PJ,** Manduca A, Felmlee J, Ehman RL. (2005). Mechanical transient based magnetic resonance elastography. Magnetic Resonance Medicine. 53(3):628–639.
99. **McClincy MP,** Lovell MR, Pardini J, Collins MW, Spore MK (2006). Recovery from sports concussion in high school and collegiate athletes. Brain Injury. 20:33-39.
100. **McCrea M,** Hammeke T, Plsen G, Leo P, Guskiewicz KM. (2004). Unreported concussion in high school football players: implications for prevention. Clinical Journal of Sports Medicine. 14(1):13-17.
101. **McIntosh AS,** Patton DA, Frechede B, Pierre PA, Ferry E, Barthels T. (2014). The Biomechanics of Concussion in Unhelmeted Football Players in Australia: A Case-Control Study. Biomedical Journal. 4:1–9. e005078. doi:10.1136/bmjopen-2014-005078
102. **Meaney DF,** Smith DH. (2011). Biomechanics of concussion. Clinical Sports Medicine. DOI: 10.1016/j/csm.2010.08.009.
103. **Meehan WD,** Hemecourt P, Comstock RD. (2010). High school concussion in the 2008-2009 academic year. American Journal of Sports Medicine. 30(12):2405-2409.
104. **Miller K,** Chinzei K. (2002). Mechanical properties of brain tissue in tension. Journal of Biomechanics. 35:483-490.
105. **Mills NJ,** Fitzgerald C, Gilchrist A, Verdejo R. (2003). Polymer foams for personal protection: cushions, shoes and helmets. Composite Science and Technology. 63:2390-2400.
106. **Moritz AR.** (1943). Mechanisms of head injury. Annals of Surgery. 117(4):562 – 575.
107. **Moss WC,** King MJ. (2011). Impact response of US Army and National Football League helmet pad systems. Livermore, CA: Lawrence Livermore National Laboratory.
108. **Mueller EA,** Moore MM, Kerr DCR, Sexton G, Camicioli RM, Howieson DB. (1998). Brain volume preserved in healthy elderly through the eleventh decade. Neurology. 51:1555–1562.
109. **National Centre for Chronic Disease Prevention & Health Promotion.** (2006) Behavioooral Risk Factor Survellance System : Exercise. CDC
110. **National Collegiate Athletic Association.** 1981-82-2004-05 NCAA Sports Sponsorship and Participation Rates Report. Indianapolis, IN : The National collegiate Athletic Association ; 2005. Available at : http://www.ncaa.org/library/research/participation_rates/1982-2005/1982_2005_particiation_rates.pdf.

111. **National Federation of State High School Associations**. 2005-06 High School Athletics Participation Summary. Available at: http://www.nfhs.org/core/contentmanager/uploads/2005_06NFHS participationsurvey.pdf.
112. **National Operating Committee on Standards for Athletic Equipment** (2004) Standard Drop Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear: NOCSAE Standard 001-04m05. Available at www.nocsae.org.
113. **National Operating Committee on Standards for Athletic Equipment** (2014) Standard Performance Specification for Newly Manufactured Football Helmets: NOCSAE DOC (ND) 002-14. Available at www.nocsae.org
114. **Naunheim RS**, Standeven J, Richter C, Lewis LM (2000) Comparison of impact data in hockey, football, and soccer. J Trauma 48:938–941.
115. **Newman JA.** (1993). Biomechanics of human trauma: Head protection. Accidental Injury Biomechanics and Prevention. New York, Springer-Verlag 292–310.
116. **Newman JA**, Shewchenko N, Welbourne E. (2000). A proposed new biomechanical head injury assessment function—the maximum power index. In: Proceedings of the Stapp Car Crash Journal, 44 PaperNo.2000–01-SC16, pp 215–248.Paper No. 2000–01-SC16.
117. **Newman JA,** Beusenber MC, Shewchenko N, Withnall C, Fournier E. (2005). Verification of biomechanical methods employed in a comprehensive study of mild traumatic brain injury and the effectiveness of American football helmets. Journal of Biomechanics 38: 1469-1481.
118. **Ommaya AK.** (1966). Experimental head injury in the monkey. Head Injury. Lippincott Philadelphia. 160-175
119. **Ommaya AK,** Faas F, Yarnell P. (1968). Whiplash injury and Brain damage: An experimental study. JAMA 204(4):285-289
120. **Ommaya AK,** Gennarelli TA. (1974). Cerebral Concussion and Traumatic Unconsciousness. Brain. 97:633–654.
121. **Ommaya AK.** (1985). Biomechanics of head injury; experimental aspects. In The Biomechanics of Trauma, Appleton-Century-Crofts, Norwalk. 245-279.
122. **Ommaya A K,** Goldsmith W, Thibault L. (2002). Biomechanics and neuropathology of adult and paediatric head injury. British Journal of Neurosurgery. 16(3):220–242.
123. **O'Riordain K,** Thomas PM, Phillips JP, Gilchrist MD. (2003). Reconstruction of real work head injury accidents resulting from falls using multibody dynamics. Clinical Biomechanics. 18:590–600.
124. **Oukama R,** Pearsall DJ. (2011). Evaluation of a flexible force sensor for measurement of helmet foam impact performance. Journal of Biomechanics. Doi:10.1016/j.jbiomech.2010.11.035
125. **Oukama R,** Pearsall DJ. (2012). Impact performance of ice hockey helmets: head acceleration versus focal force dispersion. Journal of Sports Engineering and Technology – Special Issue 226(314):185-192.
126. **Pellman EJ,** Viano DV, Casson IR, Waeckerle JF. (2003). Concussion in professional football: reconstruction of game impacts and injuries. Neurosurgery 53:799-814.
127. **Pellman EJ,** Viano DC, Withnall C, Shewchenko N, Bir CA, Halstead PD. (2006). Concussion in professional football: Helmet testing to assess impact performance – Part 11. Journal of Neurosurgery. 58:78–96.
128. **Post A,** Oeur A, Hoshizaki B, Gilchrist MD. (2012a). An examination of American football helmets using brain deformation metrics associated with concussion. Journal of Materials and Design.DOI:10/1016/ j.matdes. 2012.09.017

129. **Post A**, Oeur A, Hoshizaki TB, Gilchrist MD. (2012b). The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research. International Research Council on Biomechanics of Injury Conference (IRCOBI), Dublin, Ireland.
130. **Post A,** Walsh ES, Hoshizaki TB, Gilchrist MD. (2012c). Analysis of loading curve characteristics on the production of brain deformation metrics. Journal of Sports Engineering and Technology. DOI: 10.1177/1754337112449420.
131. **Post A,** Hoshizaki TB, Gilchrist MD. (2012d). Finite element analysis of the effect of loading curve shape on brain injury predictors. Journal of Biomechanics. 45: 679-683
132. **Post A,** Hoshizaki TB, Gilchrist MD, Brien S, Cusimano MD, Marshall S. (2014a). The influence of acceleration loading curve characteristics on traumatic brain injury. Journal of Biomechanics. 47:1074-1081.
133. **Post A,** Kendall M, Koncan D, Cournoyer J, Hoshizaki TB, Gilchrist MD, Brien S, Cusimano MD, Marshall S. (2014b.) Characterization of persistent concussive syndrome using injury reconstruction and finite element modelling. Journal of Mechanical Behaviour and Biomedical Materials.
134. **Post A,** Hoshizaki TB. (2015). Rotational acceleration, brain tissue strain, and the relationship to concussion. Journal of Biomechanical Engineering. 137:1-8.
135. **Powell JW,** Barber-Foss KD. (1999). Traumatic brain injury in high school athletes. JAMA 282(10):958-963.
136. **Prevost** TP, Guang J, de Moya MA, Alam HB, Subra S, Simona S. (2011). Dynamic mechanical response of brain tissue in indentation in vivo, in situ, and in vitro. Acta Biomaterialia. 7(12):4090-4101.
137. **Prange MT,** Margulies SS. (2002). Regional, directional, and age-dependent properties of the brain undergoing large deformation. Journal of Biomedical Engineering. 124:244-252.
138. **Ramirez M.** Schaffer KB, Shen H, Kashani S, Kraus JF. (2006) Injuries to high school football athletes in California. Am J Sports Med. 34:1147-1158
139. **Reid SE,** Herbert ME, Louis MW, Reid SE Jr. (1975) Physiologic response to impact. The Journal of Trauma. 15(2):150-152.
140. **Roberts JB**, Hugh EM. (1964). A comparison of methods for the evaluation of protective headgear. Aerospace Medicine 35:1044-1047.
141. **Rosenberg ML**, Gelle RJ, Holinger PC, Zahn MA, Conn JM, Faiman NN, Karison TA. (1984). Violence-homicide, assault, suicide. Position paper for the Carter Centre of Emory University Health Policy Consultation, Atlanta, GA.
142. **Rousseau P,** Post A, Hoshizaki TB. (2009). The effects of impact management materials in ice hockey helmets on head injury criteria. Journal of Sports Engineering and Technology. DOI: 10.1243/ 17543371JSET36.
143. **Rousseau P,** Hoshizaki B, Gilchrist MD. (2012). A reconstruction of a puck impact to the head resulting in concussion in ice hockey. Paper presented at: Symposium on the Mechanics of Concussion in Sports, ASTM International; Atlanta, Georgia.
144. **Rowson S,** et al. (2009). Linear and angular head acceleration measurements in collegiate football. Journal of Biomechanical Engineering, 131(6):7. DOI: 10.1115/1.3130454
145. **Rowson S**, Duma SM, Beckwith JG, Chu JJ, Greenwald RM, Crisco JJ, Gunnar Brolinson P, Duhaime AC, McAllister TW, Maerlender AC. (2012). Rotational head kinematics in football Impacts: An injury risk function for concussion. Annals of Biomedical Engineering. 40(1):1–13.

146. **Shankar PR,** Fields SK, Collins CL, Dick RW, Comstock RD. (2007). Epidemiology of high school and collegiate football injuries in the United States, 2005-2006.
147. **Sack I,** Beierback B, Wuerfel J, Klatt D, Hamhaber U, Papazoglou S, Martus P, Braun J. (2009). The impact of aging and gender on brain viscoelasticity. NeuroImage. 46:652-657.
148. **Shaw NA.** (2002). The neurophysiology of concussion. Progress in Neurobiology. 67:281-344.
149. **Smith PJ,** Hollins AM, Sawyer JR, Spence DD, Outlaw S, Kelly DM. (2018). Characterization of American football injuries in children and adolescents. Journal of Pediatric Orthopedics. 38(2):57-60.
150. **Sone JY,** Kondziolka D, Huang JH, Samadani U. (2016). Helmet efficacy against concussion and traumatic brain injury: A review. Journal of Neurosurgery. doi/abs/10.3171/2016.2.JNS151972
151. **Song B,** Chen WW, Dou S, Winfree N A, Kang JH. (2005). Strain-rate effects on elastic and early cell-collapse responses of a polystyrene foam. International Journal of Impact Engineering. 31:509-521.
152. **Stevens JA,** Corso PS, Finkelstein EA, Miller TR. (2006). The costs of fatal and non-fatal falls among older adults. Injury Prevention. 12:290-295.
153. **Stojsih S.** (2010). The biomechanics of amateur boxers master of science thesis. Wayne State University. Detroit Michigan.
154. **Styrke J,** Stålnacke BM, Sojka P, Björnstig U. (2007). Traumatic brain injuries in a well-defined population: epidemiological aspects and severity. Journal of Neurotrauma. 24:1425-1436.
155. **Takhounts EG,** Ridella SA. (2008). Investigation of traumatic brain injuries using the next generation of simulate injury monitor (SIMon) finite element head model. Stapp Car Crash Journal. 52:1-31.
156. **Thompson N,** Halpern B, Curl WW. et at. (1987). High school football injuries: evaluation. Americal Journal of Sports Medicine 15:117-124.
157. **Tinard V,** Deck C. (2012). New methodology for improvement of helmet performances during impacts with regards to biomechanical criteria. Materials and Design. 37:447-459.
158. **Tortora GJ.** (1995). Principles of anatomy and physiology. Harper Collins College Div. Addison-Wesley, 7th Revised edition (March 1995).
159. **Van Dommelen** JAW, Van der Sande TPJ, Hrapko M, Peters GWM. (2010). Mechanical properties of brain tissue by indentation: Interregional variation. Journal of the Mechanical Behaviour of Biomedical Materials. 3:158-166.
160. **Unterharnscheidt F**. (1970). About boxing: review of historical and medical aspects. Tex Rep Biol Med. 28:421–495.
161. **Unterharnscheidt FJ.** (1972). Head injury after boxing. Scandinavian Journal of Rehabilitation Medicine. 4: 77–84.
162. **Vappou J,** Breton E, Choquet P, Willinger R, Constantinesco A. (2008). Assessment of in vivo and post mortem mechanical brain tissue using magnetic resonance elastography. Journal of Biomechanics. 41:2954-2959.
163. **Vetter L,** Vanderby R. (1987). The influence of materials and structure on performance of a football helmet. Polym. Eng. Sci. 27(15):1113-1120.
164. **Viano DC,** King AJ, Melvin JW, Weber K (1989). Injury biomechanics research: An essential element in the prevention of rrauma. Journal of Biomechanics 22(5):403-417.

165. **Viano DC,** Casson IR, Pellman EJ, Bir CA, Zhang L, Sherman DC, Boitano MA. (2005). Concussion in professional football comparison with boxing head impacts. Neurosurgery. 57(6):1154-1172.
166. **Viano DC,** Halstead D. (2012). Change in size and impact performance of football helmets from the 1979's to 2010. Annals of Biomedical Engineering 40(1):175-184.
167. **Walilko TJ,** Viano DC. (2005). Biomechanics of the head for Olympic boxer punches to the face. British Journal of Sports Medicine. 39:710-719.
168. **Walsh ES,** Rousseau P, Hoshizaki TB. (2011). The influence of impact location and angle on the dynamic impact response of a Hybrid III headform. Journal of Sports Engineering and Technology. doi.10.1007/s12283-011-0060-9
169. **Ward C,** Chan M, Nahum A. (1980). Intracranial pressure- A brain injury criterion. Proc. 24th Stapp Car Crash Conference. Society of Automotive Engineers; Warrendale, PA p. 163-185.
170. **Weaver AA,** Danelson KA, Stitzel JD. (2012). Modeling brain injury response for rotational velocities of varying directions and magnitudes. Journal of Biomedical Engineering. 40(9):2005-2018.
171. **Willinger R,** Baumgartner D. (2003). Human head tolerance limits to specific injury mechanics. International Journal of Crashworthiness. 8(6):605-617.
172. **Yoganandan N**, Pontar FA, Sances AJ, Walsh DJ, Ewing CL, Daniel JT, Snyder RG. (1995). Biomechanics of skull fracture. Journal of Neurotrauma. 12(4):659–668.
173. **Yoganandan N,** Pintar FA. (2004). Biomechanics of temporo-parietal skull fracture. Clinical Biomechanics. 19:225–239.
174. **Yoganandan N,** Li J, Zhang J, Pintar F, Gennarelli TA. (2008). Influence of angular acceleration-deceleration pulse shapes on regional brain strains. Journal of Biomechanics. 41:2253-2262.
175. **Zanetti K.** (2014). The association between peak resultant linear acceleration and brain tissue deformation in American football-related helmeted head impacts. Masters of Science in Human Kinetics – Thesis, Faculty of Health Sciences, School of Human Kinetics, University of Ottawa. Ottawa, Canada.
176. **Zaryn TR,** Finch CF. (2003). A 16 year study of injuries to professional boxers in the state of Victoria, Australia. British Journal of Sports Medicine. 37:321-324.
177. **Zhang L**. (2001). Computational biomechanics of traumatic brain injury, An investigation of head impact response and American football head injury. Doctor of Philosophy – Dissertation. Graduate School of Wayne State University, Detroit, Michigan.
178. **Zhang L**. Yang KH, King AI, Viano KC. (2003). A new biomechanical predictor for mild traumatic brain injury – A preliminary finding. Summer Bioengineering Conference, June 25-29, Sonesta Beach Resort, Key Biscayne, Florida, USA.
179. **Zhang L,** Yang KH, King AI. (2004). A proposed injury threshold for mild traumatic brain injury. Journal of Biomedical Engineering. 126:226 – 236. DOI: 10.1115/1.1681446.
180. **Zhang J,** Yoganandan N, Pintar F, Gennarelli T, (2006). Brain strains in vehicle impact tests. Annu. Proc. Assoc. Adv. Automot. Med. 50:1–12.
181. **Zhang J,** Green MA, Sinkus R, Bilston LE. (2011). Viscoelastic properties of human cerebellum using magnetic resonance elastography. Journal of Biomechanics. 44(10):1909 – 1913.

182. **Zou H,** Schmiedeler JP, Hardy WN**.** (2007). Separating brain motion into rigid body displacements and deformation under low-severity impacts. Journal of Biomechanics. 40: 1183-1191
183. **Zwimper TJ**, Brown J, Sullivan I, Moulton RJ. (1997). Head injuries due to falls caused by seizures: a group at high risk for traumatic intracranial haematomas. Journal of Neurosurgery. 86:433–437.

## Appendix 1: Nomenclature

| | |
|---|---|
| σ | stress |
| ε | strain |
| τ | tensile stress |
| a | acceleration |
| A | area |
| ABS | Acrylonitrile Butadiene Styrene |
| ASDH | Acute SubDural Hematoma |
| CDC | Centre for Disease Control |
| CoG | Centre of gravity |
| CSDM | cumulative strain damage measure |
| CSF | cerebral spinal fluid |
| CT | computed tomography |
| DAI | Diffuse axonal injury |
| EPP | Expanded Polypropylene |
| EPS | Expanded Polystyrene |
| F | force |
| FEM | Finite Element Modelling |
| g | (9.81 $m/s^2$) linear acceleration |
| G' | storage modulus |
| G'' | loss modulus |
| GSI | Gadd severity index |
| HIC | Head injury criteria |
| HIP | Head impact power |
| Hz | Hertz |
| ICP | Intracranial pressure |
| kN | kilonewtons |
| kPa | kilopascals |
| m | mass |
| ms | milli-second |
| mTBI | mild traumatic brain injury |
| MPS | maximum principal strain |
| MRI | magnetic resonance imaging |
| NOCSAE | National Operating Committee on Standards for Athletic Equipment |
| PC | Polycarbonate |
| PE | Polyethylene |
| TBI | traumatic brain injury |
| $Rad/s^2$ | radians per second squared – rotational acceleration measure |
| VMS | von Mises stress |
| VN | vinyl nitrile |

## Appendix 2: Published Thresholds for Injury

| Threshold Value | Dependent Variable | Injury | Reference |
|---|---|---|---|
| 0.05-0.06 | Strain | Skin Lesion | Sharkey 2011 |
| 4000N | Force | Skin Lesion | Sharkey 2011 |
| 66, 82, 106 g | Linear Accel. | Concussion (25/50/80% risk) | Zhang 2004 |
| 4600, 5900, 7900 rad/s^2 | Rotational Accel. | Concussion (25/50/80% risk) | Zhang 2004 |
| 4,500 rad/s^2 | Rotational Accel | Concussion (primates) | Ommaya 1985 |
| 16,000 rad/s^2 | Rotational Accel | Concussion (primates) | Marquilies (1992) |
| 10,000 rad/s^2 | Rotational Accel. | Concussion (rats) | Davidsson (2009) |
| 0.21 | MPS | Concussion (50% risk) | Kleiven 2007 |
| 0.26 | MPS | Concussion (50% risk) | Kleiven 2007 |
| 0.19 | MPS | Concussion (50% risk) | Zhang 2004 |
| 48.5 s-1 | Strain rate | Concussion (50% risk) | Kleiven 2007 |
| 60 s-1 | Strain rate | Concussion (50% risk) | King 2003 |
| 8.4 kPa | Von Mises Stress | Concussion (50% risk) | Kleiven 2007 |
| 7.8 kPa | Von Mises Stress | Concussion (50% risk) | Zhang 2004 |
| 18 kPa | Von Mises Stress | Concussion (50% risk) | Willinger 2003 |
| 0.05 – 01 | Critical strain | DAI | Marguilies 1992 |
| 0.041 – 0.276 | Strain | DAI | King 1995 |
| 18,000 rad/s^2 | Rotational Accel. | DAI | Ommaya 2002 |
| 200g (short dur.) | Linear Accel. | Contusion | Auer 2001 |
| 0.14 – 0.53 | MPS | Contusion | Doorly 2006/2007 |
| 5-17 kPa | Von Mises Stress | Contusion | Doorly 2006/2007 |
| 147 – 348 s-1 | Strain Rate | Contusion | Doorly 2006/2007 |
| 200g (short dur.) | Linear Accel. | Subdural hematoma | Gennarelli 1982 |
| 0.16 – 0.23 | MPS | Subdural hematoma | Doorly 2006/2007 |
| 5 – 8 kPa | Von Mises Stress | Subdural hematoma | Doorly 2006/2007 |
| 119 – 265 s-1 | Strain Rate | Subdural hematoma | Doorly 2006/2007 |
| 0.34 | MPS | Subarachnoid Hemorrhage | Doorly 2006/2007 |
| 12.2 kPa | Von Mises Stress | Subarachnoid Hemorrhage | Doorly 2006/2007 |
| 197 s-1 | Strain Rate | Subarachnoid Hemorrhage | Doorly 2006/2007 |
| 0.16 – 0.23 | MPS | Epidural hematoma | Doorly 2006/2007 |
| 4.8 – 7.9 kPa | Von Mises Stress | Epidural hematoma | Doorly 2006/2007 |
| 129 – 141 s-1 | Strain Rate | Epidural hematoma | Doorly 2006/2007 |
| 8.8 – 14.1 KN | Force | Skull Fracture | Yoganandan 1995 |
| 14.1 – 43.5J | Fracture Energy | Skull Fracture | Yoganandan 1995 |
| 5.8 – 17 kN | Force | Skull Fracture | Allsop 1991 |
| 2000 – 2450 N | Force | Skull Fracture | Nahum 1980 |

## Appendix 3a: Percent Coordinate Components of Peak Acceleration

| By Location | Linear Acceleration | | | | | |
|---|---|---|---|---|---|---|
| | Bare Headform | | | Helmeted Headform | | |
| | % X of R | % Y of R | % Z of R | % X of R | % Y of R | % Z of R |
| Front CG | 98.8% | 13.6% | 6.4% | 99.9% | 6.5% | 27.8% |
| Front NA | 85.3% | 55.5% | 8.7% | 69.4% | 71.8% | 15.5% |
| Front PA | 89.5% | 0.0% | 9.4% | 74.3% | 69.4% | 18.2% |
| Front PE | 96.8% | 12.0% | 32.1% | 95.6% | 8.8% | 36.7% |
| Front Boss CG | 62.4% | 78.3% | 6.7% | 63.1% | 78.2% | 21.2% |
| Front Boss NA | 96.3% | 24.6% | 11.1% | 95.0% | 32.1% | 30.9% |
| Front Boss PA | 18.2% | 98.7% | 4.6% | 17.5% | 98.8% | 20.2% |
| Front Boss PE | 56.3% | 82.7% | 23.4% | 52.3% | 86.6% | 17.4% |
| Side CG | 5.5% | 99.8% | 7.8% | 7.9% | 99.9% | 22.9% |
| Side NA | 51.1% | 86.4% | 15.6% | 50.7% | 89.7% | 33.4% |
| Side PA | 47.3% | 91.1% | 11.3% | 47.8% | 87.9% | 17.7% |
| Side PE | 18.2% | 94.0% | 40.0% | 30.2% | 93.1% | 44.0% |
| Rear Boss CG | 5.5% | 74.2% | 15.0% | 72.5% | 71.1% | 31.5% |
| Rear Boss NA | 25.2% | 97.0% | 10.1% | 16.1% | 99.4% | 30.4% |
| Rear Boss PA | 96.4% | 27.0% | 18.4% | 96.3% | 20.9% | 33.1% |
| Rear Boss PE | 65.1% | 77.2% | 26.5% | 48.1% | 86.8% | 28.4% |
| Rear CG | 99.9% | 6.7% | 14.3% | 98.3% | 10.4% | 32.2% |
| Rear NA | 88.6% | 48.3% | 16.2% | 79.0% | 61.1% | 30.3% |
| Rear PA | 92.1% | 40.9% | 17.8% | 89.8% | 55.3% | 27.7% |
| Rear PE | 92.2% | 10.7% | 43.7% | 86.2% | 8.9% | 56.5% |

## Appendix 3b: Percent Coordinate Components of Peak Acceleration

| By Location | Rotational Acceleration | | | | | |
|---|---|---|---|---|---|---|
| | Bare Headform | | | Helmeted Headform | | |
| | % X of R | % Y of R | % Z of R | % X of R | % Y of R | % Z of R |
| Front CG | 22.8% | 99.4% | 21.0% | 19.6% | 99.9% | 16.7% |
| Front NA | 58.3% | 76.1% | 45.6% | 51.5% | 51.7% | 82.0% |
| Front PA | 51.8% | 87.1% | 34.6% | 56.0% | 58.5% | 68.6% |
| Front PE | 42.4% | 84.6% | 32.8% | 44.4% | 96.9% | 24.7% |
| Front Boss CG | 65.5% | 36.5% | 74.0% | 80.2% | 59.5% | 38.8% |
| Front Boss NA | 28.5% | 97.8% | 33.3% | 42.8% | 87.6% | 56.3% |
| Front Boss PA | 61.5% | 6.5% | 79.2% | 71.6% | 16.8% | 72.5% |
| Front Boss PE | 69.3% | 31.5% | 71.5% | 73.5% | 31.6% | 68.3% |
| Side CG | 97.2% | 4.6% | 34.3% | 83.9% | 22.6% | 63.6% |
| Side NA | 75.2% | 36.8% | 64.2% | 73.7% | 51.6% | 72.5% |
| Side PA | 59.1% | 22.6% | 79.5% | 91.2% | 40.4% | 28.7% |
| Side PE | 95.2% | 32.3% | 44.8% | 96.0% | 41.3% | 40.5% |
| Rear Boss CG | 55.0% | 47.7% | 73.9% | 57.2% | 63.8% | 78.2% |
| Rear Boss NA | 66.9% | 18.9% | 75.1% | 72.9% | 20.2% | 68.9% |
| Rear Boss PA | 40.5% | 91.0% | 25.0% | 37.4% | 94.4% | 27.0% |
| Rear Boss PE | 60.8% | 41.5% | 69.5% | 74.1% | 29.0% | 70.1% |
| Rear CG | 18.1% | 99.2% | 18.3% | 37.3% | 99.6% | 34.0% |
| Rear NA | 44.4% | 68.0% | 67.4% | 48.9% | 59.7% | 77.4% |
| Rear PA | 40.8% | 72.8% | 67.0% | 50.7% | 66.4% | 77.1% |
| Rear PE | 19.4% | 95.5% | 35.1% | 25.4% | 97.7% | 38.0% |

# Appendix 4: Correlation of Coordinate Components to MPS

| | | | RLINEAR | XLINEAR | YLINEAR | ZLINEAR | RROTATIONAL | XROTATIONAL | YROTATIONAL | ZROTATIONAL | MPS |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Bare Front | MPS | Pearson Correlation | -0.264 | -0.301 | 0.526 | -.741** | 0.012 | 0.2 | 0.121 | 0.032 | 1 |
| | | Sig. (2-tailed) | 0.406 | 0.342 | 0.079 | 0.006 | 0.972 | 0.533 | 0.709 | 0.922 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Helmeted Front | MPS | Pearson Correlation | -0.282 | -.778** | .630* | -.856** | 0.528 | 0.306 | -.670* | 0.564 | 1 |
| | | Sig. (2-tailed) | 0.375 | 0.003 | 0.028 | 0 | 0.078 | 0.334 | 0.017 | 0.056 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Bare Front Boss | MPS | Pearson Correlation | .680* | -.973** | .790** | -0.141 | .902** | .713** | -.885** | .831** | 1 |
| | | Sig. (2-tailed) | 0.015 | 0 | 0.002 | 0.663 | 0 | 0.009 | 0 | 0.001 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Helmeted Front Boss | MPS | Pearson Correlation | .751** | -.813** | .900** | -.638* | .668* | .840** | -.705* | 0.387 | 1 |
| | | Sig. (2-tailed) | 0.005 | 0.001 | 0 | 0.026 | 0.018 | 0.001 | 0.01 | 0.214 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Bare Side | MPS | Pearson Correlation | .897** | -0.431 | .870** | -0.04 | .976** | .946** | -0.165 | 0.434 | 1 |
| | | Sig. (2-tailed) | 0 | 0.162 | 0 | 0.901 | 0 | 0 | 0.608 | 0.159 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Helmeted Side | MPS | Pearson Correlation | 0.17 | 0.346 | 0.036 | -.655* | 0.523 | 0.298 | 0.005 | -0.337 | 1 |
| | | Sig. (2-tailed) | 0.597 | 0.271 | 0.913 | 0.021 | 0.081 | 0.346 | 0.987 | 0.284 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Bare Rear Boss | MPS | Pearson Correlation | .656* | .820** | -0.275 | .828** | 0.037 | -0.29 | .809** | -0.209 | 1 |
| | | Sig. (2-tailed) | 0.021 | 0.001 | 0.387 | 0.001 | 0.91 | 0.361 | 0.001 | 0.515 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Helmeted Rear Boss | MPS | Pearson Correlation | -.737** | -0.515 | 0.137 | -0.531 | -0.039 | 0.104 | -0.217 | 0.301 | 1 |
| | | Sig. (2-tailed) | 0.006 | 0.086 | 0.672 | 0.076 | 0.904 | 0.748 | 0.498 | 0.341 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Bare Rear | MPS | Pearson Correlation | -0.056 | -0.069 | 0.231 | -0.064 | 0.011 | 0.191 | -0.123 | 0.275 | 1 |
| | | Sig. (2-tailed) | 0.863 | 0.832 | 0.471 | 0.844 | 0.973 | 0.552 | 0.703 | 0.388 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |
| Helmeted Rear | MPS | Pearson Correlation | 0.231 | 0.343 | -0.324 | 0.225 | -0.054 | -0.114 | 0.479 | -0.267 | 1 |
| | | Sig. (2-tailed) | 0.47 | 0.274 | 0.304 | 0.482 | 0.869 | 0.724 | 0.115 | 0.401 | |
| | | N | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 | 12 |