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Finite Element Modeling of Brain Injury for Performance Evaluation of Football Helmets

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Abstract

Finite Element Modeling of Brain Injury for Performance Evaluation of Football Helmets

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Football helmets worn today are primarily designed to prevent skull fracture and irreversible brain injury. In that limited respect, today's helmets are a success with the frequency of reported football-related skull fracture a rare occurrence. In recent decades, however, concussion, mild traumatic brain injury (mTBI), and chronic traumatic encephalopathy (CTE) are being reported with increasing incidence. Helmet manufacturing companies are limited in their performance analysis of helmets, with no standard helmet testing procedures incorporating anatomical data or physiological weaknesses of the brain. Finite Element Modeling of the brain has allowed for analysis of the mechanics of brain injury and can provide injury metrics based on simulated brain injury. In this study the background on current helmet testing procedures is provided as well as a finite element study of brain injury to determine helmet performance. In addition, this modeling will be used to identify what types of impacts are most dangerous.

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Chapter 1. INTRODUCTION

Concussions have rapidly become a public health priority [1, 2, 3]. In 2003, the United States Centers for Disease Control and Prevention issued a report to Congress stating that concussions in sports had reached ‘epidemic’ levels in the United States [4, 5]. Reports of the current annual incidence of sports-related concussions in the U.S. are around 3.8 million [1, 6, 7, 8]. Football players comprise the largest population of reported incidents [8]. Guskiewicz et al. [9] reported that 5% of high school and college football players endure concussions; however, there is speculation that a significant number of concussions in high school remain unreported [6]. In an anonymous survey, 15% of high school football players reported they had experienced concussions [6, 10]. Recurrence of concussion is also of concern, studies show the risk of concussion in football is three to six times higher in players who have had a previous concussion [11].

The typical duration of these impacts is relatively short, lasting approximately 15 milliseconds, however they can produce high levels of acceleration where the brain can impact the inner table of the skull or the dividing membranes (meninges). Coup-contrecoup injuries can entail these rapid movements, leading to TBIs. Acceleration injuries can also lead to axon-shearing injuries, referred to as diffuse axonal injuries (DAIs). These types of injuries commonly occur along the midline structures of the brain, including the corpus callosum, thalamus, and brainstem. More commonly, concussive impacts lead to a temporary metabolic derangement of axons without overt structural damage. Routine imaging studies, such as computed tomography (CT) or magnetic resonance imaging (MRI), are rarely abnormal. High-fidelity imaging techniques, such as functional or resting-state functional MRI, diffusion tensor imaging, or MRI spectroscopy, are necessary to elucidate subtle findings on routine concussion patients, and even these studies are rarely positive.

Recently, the alarm over repeated, low-velocity head impacts has raised concerns for delayed neurologic sequelae. Chronic traumatic encephalopathy has been popularized in the lay press and movies (e.g., “Concussion” featuring Will Smith) as being a direct consequence of repeated head trauma, most notably in the setting of contact football. Chronic traumatic encephalopathy can be a progressive neurodegenerative disease with sequelae including depression and other mental illness. The disease was previously called dementia pugilistica, or "punch-drunk," as it was initially found in those with a history of boxing. The specific cause is likely repeated head injury or recurrent concussions; however, the exact pathophysiology and genetic predispositions are still being explored, and many players with repeated concussions have failed to demonstrate chronic traumatic encephalopathy or related symptoms.

Several high-profile injury cases have led to class-action and individual lawsuits from the high school to professional levels, increasing media scrutiny. U.S. Congressional hearings related to concussion mitigation have promoted positive change [4]. Between 2009 and 2015, all 50 states and the District of Columbia passed laws to address TBI [12]. In addition, rule changes and improvement in tackling techniques have positively influenced the safety of football. Legal proceedings have also targeted helmet manufacturers in recent years, as claims for concussion reduction or mitigation were unsubstantiated. Historically, only the National Operating Committee on Standards for Athletic Equipment (NOCSAE) regulated the football helmet industry. More recently, additional testing bodies have come forth with various testing paradigms meant to investigate helmet performance and improve the testing metrics. Both the National Football League (NFL) and Virginia Tech University have developed independent testing standards that exceed NOCSAE predicate testing by including the rotational testing paradigms that are thought to more closely replicate hitting scenarios that lead to concussive events.

1.1 MODELLING OF HEAD IMPACT FORCES AND CONSEQUENCES

The development of safer products for sports participation requires an understanding of the underlying pathology and necessitates non-human, non-cadaveric methods by which physical responses can be tested. The entire rigid body kinematics of the head can be described using three linear and three rotational coordinates and subsequently used in brain injury analysis.

Previous studies that attempted to understand the human head’s tolerance to external loads involved measuring peak linear acceleration and resultant linear acceleration. New studies have suggested that angular velocity and angular accelerations may be a better indicator of brain injury than linear acceleration, because brain tissue’s bulk shear modulus is a million times weaker than its bulk elastic modulus [13, 14]. This is due to patterns in the brain myelination and the directionality of fiber/tract pathways. These forces are thought to be causal in mTBI, notably in the setting of concussive events.

1.1.1 *MODELING THE KINEMATICS OF THE HUMAN HEAD*

TBI involves physical damage and/or metabolic derangements that impact or alter function. Concussions considered to be mTBI are caused by “blunt” forces applied to the player’s head [15]. Several factors are thought to contribute to the risk of concussion including age, sex, history of prior concussion, genetic predisposition, etc. [13]. Recovery from these events is complex, no two brain injuries are alike and the symptoms of injuries can be dramatically different at onset and over time.

To better understand the relationship between imposed external stresses and TBI, researchers have developed mathematical models using finite element modeling (FEM) to estimate the resulting deformation of brain tissue as a surrogate marker of injury. Modeling allows evaluation of variables that may increase or mitigate forces transmitted to the skull, brain, or other structures.

Modern finite element head models (FEHMs) simulate individual brain components and the complex contact relationships between them. These models have been used to develop better brain injury metrics and provided more information on the mechanics of brain injury.

1.1.2 *HEAD INJURY PREDICTION*

The first head injury prediction models emphasized the quantification of the mechanics of skull fracture and irreversible brain injury. This was done through animal and cadaver studies, where principal accelerations were measured and compared with brain injury histology. The introduction of FEMs has advanced head injury prediction methods, leading to the helmet impact testing standards and brain injury assessment methods currently in use. The earliest—the Wayne State Tolerance Curve (WSTC)—still serves as the basis for many injury metrics, and the newer finite element brain injury criteria incorporate additional kinematic pathways to better characterize TBI.

1.1.3 *Wayne State Tolerance Curve*

Much of the research in the 1950s sought to define the relationship between head impact loads and TBI [16, 17, 18]. These studies were motivated by brain injury–related deaths seen in automobile accidents, auto-racing, hockey, American football, and other sports. In 1960, Lissner et al. [19] correlated peak acceleration to skull fracture over impact durations of 1–6 ms. In that work, embalmed cadaver heads were dropped onto unyielding, flat surfaces, with a strike to the forehead. In 1966, Gurdjian et al. [17] used comparative cadaver and animal impact studies coupled with subconcussive volunteer restraint system sled test data to extend the impulse duration beyond 6 ms. This research defined a relationship between acceleration level, impulse duration, and head injury known as the Wayne State Tolerance Curve (WSTC). Values above the curve are hypothesized to result in irreversible brain damage. Ono et al. extended the WSTC's impulse duration beyond 10 ms using an asymptotic approach based on data derived from non-injurious impacts using human volunteers [20]. Figure 1 shows the WSTC in its final form. The WSTC demonstrated that high linear acceleration levels could be withstood for short durations, while relatively low acceleration levels could be withstood for longer periods of time without resulting in skull fracture [16, 20].

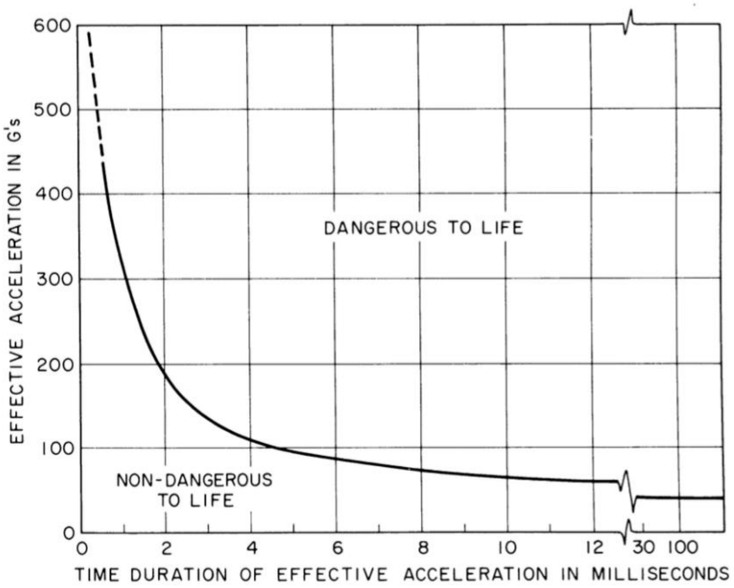

Figure 1: Wayne State Tolerance Curve (WSTC) demonstrating tolerance of the human head to peak linear excitation amplitudes as a function of impact duration [14].

1.1.4 *Injury Metrics Derived from the WSTC*

Since its inception, the WSTC has been criticized for a variety of reasons including a paucity of experimental data, questionable instrumentation, minimal documentation regarding the scaling of animal data, uncertainty in the definition of "effective" acceleration, and considering only a single (linear) acceleration component [21]. Despite these criticisms, the WSTC has served as the basis for many injury metrics still in use today. In 1966, Gadd formed a linear approximation to a log-log plot of WSTC data and used it to form a weighted index criterion to predict irreversible brain injury [22]. The result came to be known as the Gadd Severity Index (GSI).

$$\boldsymbol{GSI = \int_0^T a(t)^{2.5}\, dt} \tag{1.1}$$

GSI values greater than 1000 were thought to be dangerous to life [22]. In 1971, Versace [21] criticized the formulation, stating that the exponent could take on a wide range of legitimate values based on the logic used for its derivation.

Versace proposed a new formulation based on the average acceleration pulse. It was adapted by the National Highway Transportation and Safety Administration in 1972 to form the Head Injury Criterion (HIC15), with an injury threshold set at HIC15 = 1000 [21]. Whereas the WSTC is based only on frontal axis linear acceleration, the HIC15 is based on the resultant linear acceleration of the head during impact.

$$\boldsymbol{HIC = \max_{t_1,t_2}\left\{(t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\, dt\right]^{2.5}\right\}} \tag{1.2}$$

The HIC15 is limited to a time duration $t \in [t_1, t_2]$, where t is measured in seconds and $a(t)$ is expressed in g's (acceleration of gravity). The simulated impact data in Figure 2 illustrates this technique. The left plot shows a family of $HIC(t, \delta)$ curves, where $\delta \overset{\text{def}}{=} t_2 - t_1$. The δ is then chosen that maximizes the measured HIC15 value. The peak value of this curve is shown as

$\max_{t,\delta} HIC(t, \delta) \cong HIC(56, 45) = 727$. The corresponding area under the acceleration curve is shown in the right plot.

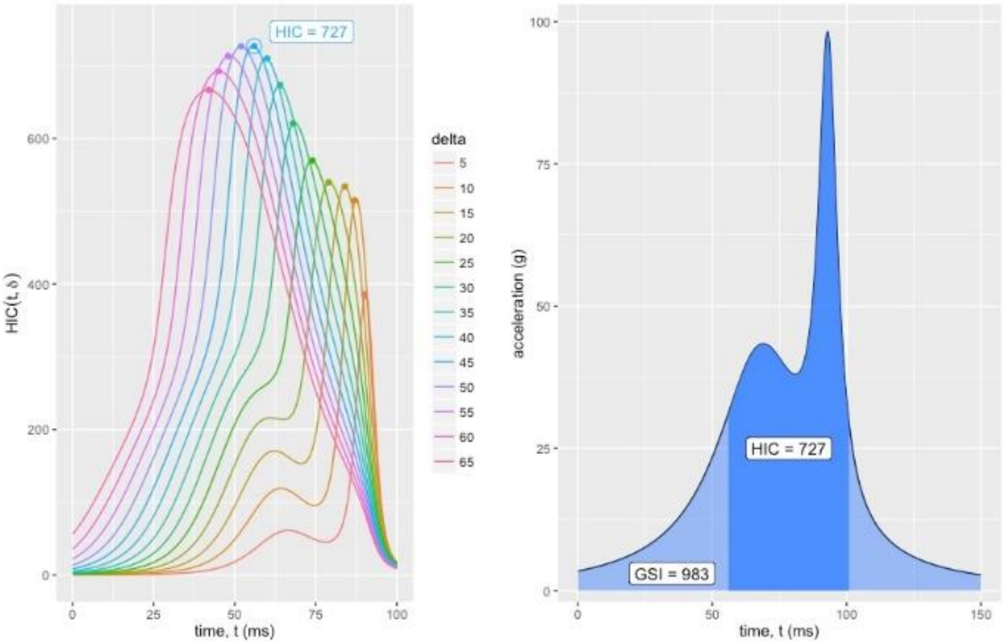

Figure 2: Simulated impact test data illustrating the estimation of the associated HIC15 and GSI values.

In practice, the HIC15 is used to assess safety related to vehicles, personal protective gear, and sports equipment. A maximum time duration of $\delta \in [3, 36]$· with $\delta \triangleq 36$ used for vehicles and $\delta \triangleq 15$ for helmets was set for these assessments [23, 24]. For a HIC15 value of 1000, there is an 18% probability of a severe head injury, a 55% probability of a serious injury, and a 90% probability of a moderate head injury to the average adult.

1.1.5 *Brain Injury Criteria (BrIC)*

Despite efforts to reduce the occurrence and severity of TBI, the weighted risks of the most severe traumatic injuries, according to the Abbreviated Injury Scale (AIS), have been increasing over the last decade [25]. One hypothesis for the increase is that the data collected via anthropomorphic test devices in crash tests is insufficient in capturing and understanding the kinematic pathways of TBI. In 2013, Takhounts et al [25] developed the Brain Injury Criteria (BrIC) as a novel approach to better characterize TBI based on this hypothesis. The BrIC was developed using two brain tissue damage metrics obtained in a FEHM (SIMon): cumulative strain damage metric (CSDM15) and maximum principal stress (MPS) [25, 26, 27]. Each of these were found to be highly correlated with rotational velocity, not linear acceleration. The BrIC is defined as a function of rotational velocity kinematics of the head during impact.

$$\boldsymbol{BrIC} = \sqrt{\left(\frac{\dot{\boldsymbol{\theta}}_{\boldsymbol{x}}}{\dot{\boldsymbol{\theta}}_{\boldsymbol{xc}}}\right)^{\mathbf{2}} + \left(\frac{\dot{\boldsymbol{\theta}}_{\boldsymbol{y}}}{\dot{\boldsymbol{\theta}}_{\boldsymbol{yc}}}\right)^{\mathbf{2}} + \left(\frac{\dot{\boldsymbol{\theta}}_{\boldsymbol{z}}}{\dot{\boldsymbol{\theta}}_{\boldsymbol{zc}}}\right)^{\mathbf{2}}} \tag{1.3}$$

$\dot{\theta}_{xc}$, $\dot{\theta}_{yc}$, and $\dot{\theta}_{zc}$ are the critical angular velocities about the x, y, and z-axis of the human head (vectors shown in 3). Table 1 shows the maximum critical angular velocity values based on CSDM15 and MPS probability risk curves, taken at the 50% probability of achieving an AIS 4+ severity level.

Table 1:Critical maximum angular velocities as a function of brain tissue damage metrics

Critical maximum angular velocity	$\dot{\boldsymbol{\theta}}_{\boldsymbol{xc}}$	$\dot{\boldsymbol{\theta}}_{\boldsymbol{yc}}$	$\dot{\boldsymbol{\theta}}_{\boldsymbol{zc}}$
CSDM15	66.20	59.10	44.25
MPS	66.30	53.80	41.50

Assuming animal subjects experience anatomic brain injuries similarly to humans, the AIS 4+ risk injury curve for CSDM15 and MPS was applied as a baseline for a human model. The AIS 4+ curve was then scaled to produce the remaining curves (AIS 1+, 2+, 3+, and 5+). The injury risk curves for various AIS severity levels based on MPS and CSDM15 injury metrics are shown in the right panels of Figure 3. The BrIC equivalent is also shown for these metrics in the left panels [28].

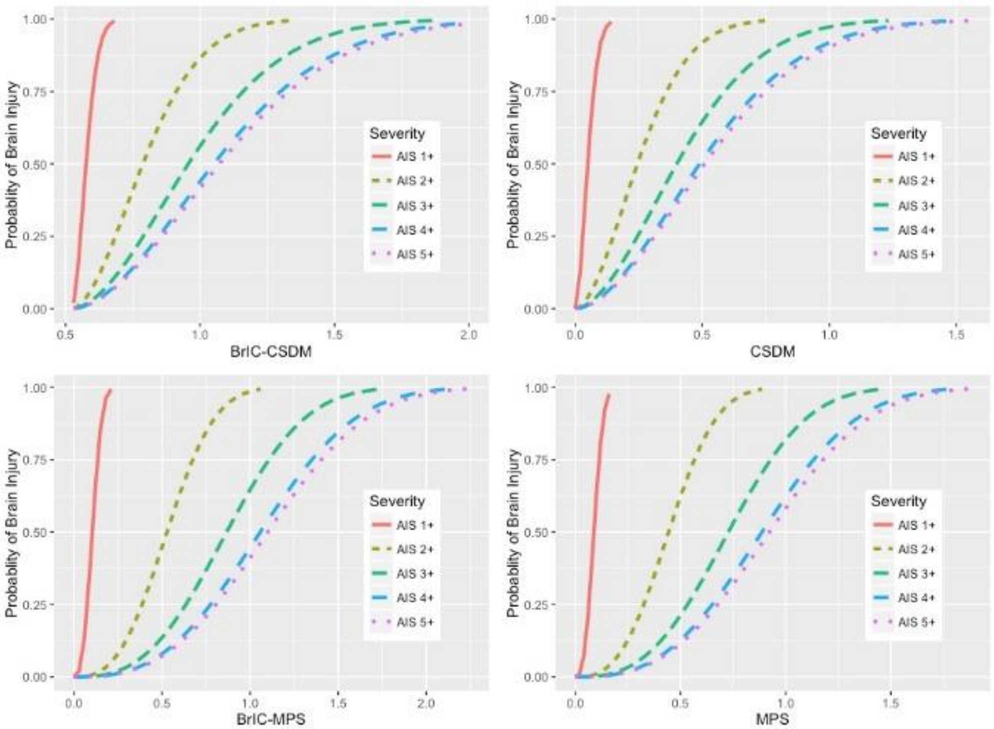

Figure 3: CSDM15 and MPS risk curves for various levels of severity and the corresponding BrIC curves [30].

Takhounts et al. also included simulations based on data collected during on-field tests involving college football players that experienced concussions, assumed to be DAI level AIS 2+ injuries, and found a critical angular velocity that compared well with a prior study [29] . The BrIC may provide an improvement to the existing testing criteria used for the development of football helmets. Takhounts et al. also suggests that it is important to include both the rotational velocity and direction as contributing factors to helmet performance [25].

1.1.6 *Measurement of Head Acceleration on the Field*

Collecting real data from impacts is necessary in validating models and developing test procedures. Data is collected through a variety of methods, such as high-speed video and sensors.

It is important to note the predictive value of sensors is still highly questioned in the sideline and clinical setting. Sensors can provide raw data regarding the mechanics of a hit to a player, but individual variations may or may not result in a concussive event with the same applied force in two matched control subjects. Additionally, the accuracy of sensors has been an outstanding question and led the NFL to end a pilot study using sensors during game-day conditions in 2014.

Several systems have been developed to collect accelerations experienced by the player on the field. These systems include helmet mounted, skin mounted, and mouthguard equipped sensors. Wu et al. [30] demonstrated that of these three types of sensors mouthguards were the most accurate followed by skin mounted sensors and then helmet mounted sensors. The later two have been shown to overpredict head accelerations due to their indirect connection to the head [30, 31, 32, 33]. However, all of these sensors suffer from false readings caused by a tossed mouthguard or helmet. In fact this has led to some devices trying to determine whether the sensor is properly in place [34].

1.2 KINEMATICS DATA, MODELING, AND SENSORS TO IMPROVE HELMET PERFORMANCE STANDARDS

Quantitative metrics provide regulating bodies and consumers means to directly compare the performance of helmet technology as it relates to brain injury risk and improved sports safety. Many current helmet performance standards are based on the WSTC because performance can be compared and evaluated within defined injury limits and evaluation criteria. Helmet standards have evolved to reflect the best data and testing methods of the time; however, the fidelity of these testing and certification paradigms comes into question when the injury criteria does not accurately predict injury.

1.2.1 *ASTM Helmet Testing Standards*

The American Society for Testing and Materials (ASTM) developed a testing standard for American football helmets. These procedures use a drop test, shown in Figure 4, to produce a consistent, fixed-motion impact. Other relevant procedures define proper helmet fit, test for extreme environmental conditions, and average between multiple helmets.

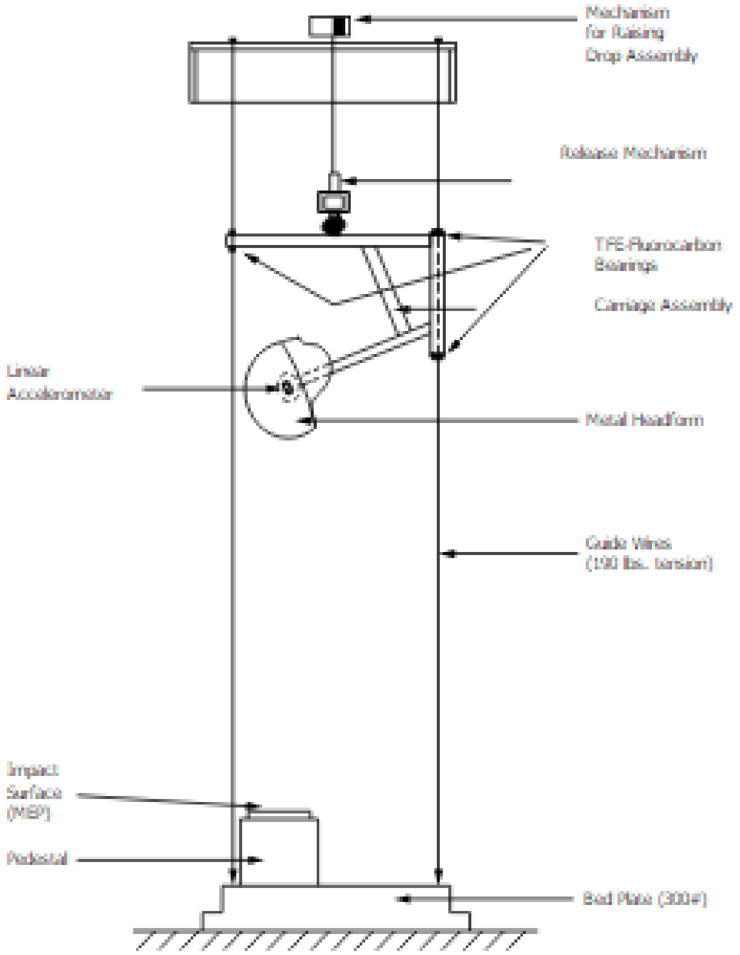

Figure 4: Typical ASTM guided drop tower and helmet sled [35].

Helmet measurements are taken from a series of drop tests over six locations: front, front boss, side, rear boss, rear, and crown. Each location is impacted three times at 5.47 m/s. The two locations with the highest mean maximum linear acceleration are impacted three additional times, resulting in a total of 24 impacts. The ASTM football helmet test grades helmets on pass/fail criteria. The average maximum acceleration is not to exceed 275 g, with no single impact having a maximum acceleration that exceeds 300 g. These ASTM football helmet tests are based on measured linear acceleration values correlated with the WSTC [35].

1.2.2 *NOCSAE Certification*

All football helmets sold the United States must pass a series of impact attenuation tests as defined by the NOCSAE. The NOCSAE test methods are adapted from the ASTM standards, using injury criterion based on the research initiated in 1971 at Wayne State University. NOCSAE defines all aspects of the testing standard, including the impact equipment, headform specification, helmet fit with the headform, allowable ambient temperature ranges, instrumentation, impact surface characteristics, number of impacts and locations, drop heights and velocities, and severity index thresholds.

The GSI is used as a performance metric, with the maximum allowable tolerance specified by NOCSAE set to eliminate skull fracture in on-field play. In 1970, NOCSAE implemented a GSI <1500 standard for all football helmets and saw a 50% reduction in fatalities in the following year. In 1996, NOCSAE lowered the GSI threshold to 1200 to better align with the auto safety regulation requirement, which has a threshold of 1000 [36, 37].

NOCSAE helmet tests involve drops performed over six standardized locations around the helmet (front, front boss, side, rear boss, rear, and top) as well as one random location chosen by the testing operator. Drops are performed over a range of impact velocities (3.46–5.46 m/s). As part of the certification process for a new helmet, the GSI for any test performed at the 3.46 m/s level cannot exceed 300, while the GSI for all tests cannot exceed 1200 [36, 37].

Additionally, NOCSAE performs tests on the facemask at two impact velocities (4.23 m/s and 5.46 m/s) over two locations: front and lower bar. At the 4.23 m/s level, there may be no contact of the mask with the headform, chin, or nose. At the 5.46 m/s level, there may be no weld breaks. All facemask impact tests are limited to a GSI of 1200 [36, 37].

To consider rotational acceleration, NOCSAE has developed a test procedure to work in tandem with current linear impact testing. This new test procedure accounts for rotational acceleration by utilizing a linearly mobile sled with a hybrid III neck. This allows for headform rotation after impact and more accurately simulates true kinematic response. A proposed NOCSAE rotation test platform is shown in Figure 5. This new test will meet the same requirements as the NOCSAE linear impact, where the peak severity index of any impact shall not exceed 1200 GSI with impacts at 3.456 m/s not exceeding 300 GSI. To account for acceleration, no test shall show rotational acceleration greater than 6,000 rad/s^2. This new testing procedure will be enacted in June 2018 [38, 39].

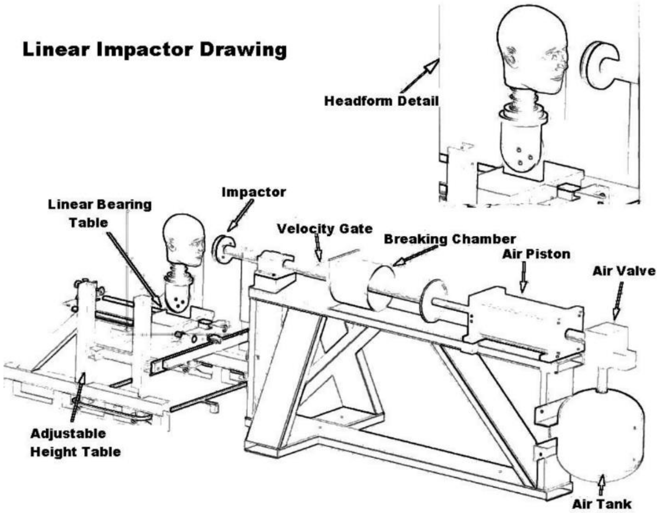

Figure 5: Sketch and labeling of proposed NOCSAE rotation test set-up (NOCSAE, 2016b).

1.2.3 *Virginia Tech STAR Evaluation System*

Rowson and Duma (2011) introduced a metric for evaluating the protective performance of football helmets, known as the Summation of Tests for the Analysis of Risk (STAR) that they developed at Virginia Tech University. The metric is based on a series of 24 NOCSAE-style laboratory drop tests configured over six drop heights and four impact locations. Each test results in a risk incidence, and the sum over all tests forms an overall risk incidence for a given helmet.

$$\boldsymbol{STAR} = \sum_{l=1}^{4} \sum_{h=1}^{6} \boldsymbol{E}(\boldsymbol{l}, \boldsymbol{h}) \cdot \boldsymbol{R}(\boldsymbol{a}) \tag{1.4}$$

Where E is a measure of head impact exposure as a function of impact location l and drop height h, and R is the injury risk as a function of measured peak resultant linear acceleration a. Hypothetically, the lower the STAR value, the lower the risk of concussion. To aid consumers, Rowson and Duma mapped the STAR values for variety of helmets to a symbolic rating system (1–5 stars), where higher star ratings suggest higher protection from concussion. The STAR rating is based primarily on the STAR value, where the two are inversely related (low STAR value leads to high STAR rating).

1.2.3.1 Quantifying exposure

To quantify exposure, Rowson and Duma required an estimate of the number of hits a typical player would experience in a given season, as well as the location and severity of those hits. The total number of hits a collegiate player was expected to receive during a single season was established as $E_{total} = 1000$, based on the 90th percentile of data that quantified hits per player for three college football teams over the course of a single season. Using over 62,000 on-field impacts recorded between 2009 and 2010 with HITS [40, 41, 42], E_{total} was then distributed over four distinct helmet impact locations—front, rear, side, and top—as shown in Figure 6and Table 2. The peak linear accelerations experienced by the players were statistically mapped to impact energies imparted on the helmets. The resulting impact energy distribution was discretized and correlated with six drop test levels (60, 48, 36, 24, 12 inches and *lowest*), where *lowest* is defined to be any height producing less than 19 g in the drop tests and is thought to be irrelevant to concussion [7].

Table 2: Values to describe impact locations shown in Figure 6.

location	**elevation**	**azimuth**
Front	$\alpha \leq 65°$	$-45° < \theta < 45°$
Rear	$\alpha \leq 65°$	$-135° < \theta < 135°$
Side	$\alpha \leq 65°$	$\pm45° \leq \theta \leq \pm135°$
Top	$\alpha > 65°$	$0° \leq \theta \leq 360°$

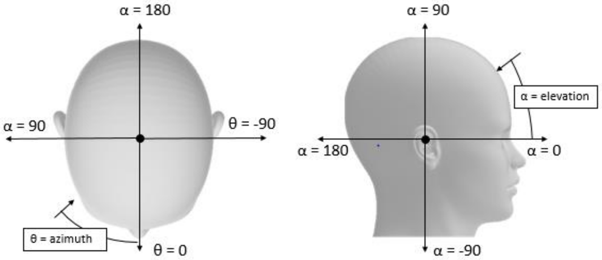

Figure 6: Elevational and azimuthal ranges for discrete helmet locations in STAR equation.

1.2.3.2 Quantifying risk

Rowson and Duma [7] estimated the probability of injury, $R(a)$, using both subconcussive (52g +- 21g) and concussive (104g +- 30g) impact data gathered from a variety of sources. Injury rates from both collegiate and professional football levels were used in the analysis and weighted based on estimated levels of concussion underreporting in the literature. Combined subconcussive and concussive head acceleration distributions were used in a logistic regression model, producing a probability of risk to head injury.

1.2.3.3 Interpretation, praise, and criticism of STAR system

Rowson and Duma filled a void by developing the STAR system; at the time, consumers had no real means of evaluating the protective performance of a given helmet. This was in stark contrast to the automotive industry, where automobile safety records based on a variety of standardized crash tests are published regularly. The STAR system received praise for its basis on data collection from real impacts, rather than cadaver and animal data.

When the STAR system was introduced in 2011, only one helmet received a 5-star rating. Today, there are 16 helmets from several manufacturers with a 5-star rating; however, there has not been a validated commensurate decrease in the rate of concussion overall. Figure 7 shows a relatively small decrease in the number of reported concussions in the NFL from 2012 through 2014, but 2015 marked the highest incidence of reported concussions, although this is likely due to increased player reporting. A prospective study in the *Journal of Neurosurgery* [43] found that using an old-style helmet (Riddell VSR4) compared with a newer-style helmet (Riddell Revolution) resulted in a 5% decrease in force, which accounted for a 46% decrease in concussion among collegiate players followed over the course of one football season. Although this represents Class III data, it is encouraging that improved technology can lead to concussion reduction.

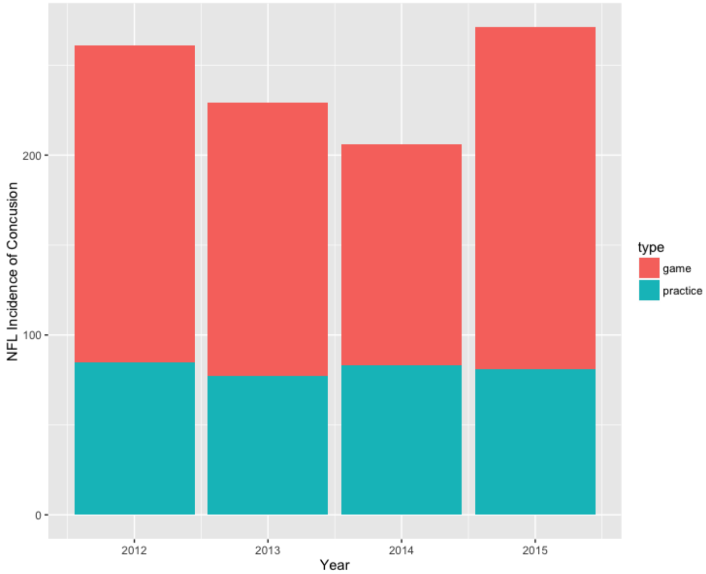

Figure 7: Incidence of reported concussion in the NFL over the years 2012-2015.

The current STAR system disregards all rotational kinematics, now hypothesized to be a primary causative component in most concussions. The STAR rating system is being revised to include rotational kinematics specifics of the new protocol, but the testing paradigms are not available for review at the time of this publication. In response to a competing rating system with potential regulatory applications, NOCSAE issued a public statement in 2013,

"The Virginia Tech Helmet Ratings system approaches the very broad and complex issue of concussion protection from a narrow vantage point of linear accelerations only and does not address other biomechanical variables such as rotational accelerations, particularly where rotational accelerations precede the linear acceleration in a hit. … Helmets were not tested under game conditions. For example, air bladder fitting and protection systems were not inflated to achieve fit, even though the NOCSAE standards require that manufacturer fitting instructions be followed." [44]. Additionally, the STAR ranking and rating only apply to adult large helmets used by collegiate players, which naturally excludes a large population of players.

1.2.4 *National Football League Testing*

The NFL has become a leader in helmeting and had developed its own proprietary method of testing and safety ranking. In 2011, the NFL established the Head, Neck, and Spine committee, under which an engineering subcommittee was tasked with understanding the current state of helmet testing and certification. Ultimately, the NFL developed its own metric for testing and utilized an independent company, Biokinetics, Inc. of Ottawa, Canada, to perform testing on all helmets worn by players in the League.

To supplement the NOCSAE standard, the NFL partnered with biomechanical experts in the NFL Players Association (NFLPA) in 2015 and 2016 to develop a new helmet performance metric based on both translational and rotational response to impact. A survey was conducted to collect a list comprising 99% of all helmets worn by NFL players. A ranked list of the performance of these helmets is disseminated to players, coaches, trainers, equipment mangers, and team doctors each season to help make informed decisions.

The complete protocol used for testing can be found in an online memorandum published by the NFL and NFLPA [45]. A helmet is mounted on a headform/neck combination that is mounted on a sled. This system increases accuracy by taking into account neck stiffness and rigid body motion. A linear impactor was used to perform impact tests over eight locations shown in Figure 8, each with three impact velocities (5.5, 7.4, and 9.3 m/s), resulting in 24 impacts per helmet. Impact locations were selected based on the most commonly observed points of contact in reviews of NFL game films. Impact velocities were chosen to emulate conditions associated with concussions and severe impacts sustained during open-field collisions.

Figure 8: Impact locations and individual metrics measured for NFL-NFLPA performance metric [69]. a, peak resultant linear acceleration (g); H, HIC15(δ=15 ms); $\ddot{\theta}$, peak resultant rotational acceleration (rad/s^2); $\dot{\theta}$, peak resultant rotational velocity (rad/s).

For each metric, an average value over the 24 impact tests was calculated to form the variable set $\left\{\bar{H},\bar{\ddot{\theta}},\bar{\dot{\theta}}\right\}_j$, where $j = 1, \dots, J$ denotes the jth helmet in the study out of J helmets. For the 2015 and 2016 studies, $J = 15$ and $J = 23$, respectively. Note that the set does not include the linear acceleration component because it is included in the calculation of the HIC15. Each of these variables is averaged over the collection of helmets used in the study to form $\bar{\bar{H}} = \sum_{j=1}^{J} \bar{H}_j$, $\bar{\bar{\ddot{\theta}}} =$

$\sum_{j=1}^{J}\bar{\ddot{\theta}}_j$, and $\bar{\bar{\dot{\theta}}} = \sum_{j=1}^{J}\bar{\dot{\theta}}_j$. The test variables for helmet j are then normalized over the collection to form the relative combined performance index, P_j.

$$\boldsymbol{P_j} = \frac{\bar{\boldsymbol{H}}_j}{\bar{\bar{\boldsymbol{H}}}} + \frac{\bar{\ddot{\boldsymbol{\theta}}}_j}{\bar{\bar{\ddot{\boldsymbol{\theta}}}}} + \frac{\bar{\dot{\boldsymbol{\theta}}}_j}{\bar{\bar{\dot{\boldsymbol{\theta}}}}} \tag{1.5}$$

The combined index relates the performance of an individual helmet as the sum of translational acceleration, rotational acceleration, and rotational velocity components, each normalized by the group average to form a relative measure. The better the helmet performs, the lower the value of P_j.

It is important to note that the NFL-NFLPA helmet performance metric is only meaningful for professional players. The relative ranking of the helmets based on this index is for informational purposes only and is not a mandatory prescription. All helmets included in the study are NOCSAE certified, so any individual helmet in the study may be worn by an NFL player, regardless of the performance index and associated ranking. Players—not the club or the League—decide which helmet they will wear.

The NFL-NFLPA helmet performance metric is intended to provide supplementary information. Players and equipment managers also have access to the Virginia Tech STAR rankings, which are based on a different set of performance criteria. Figure 9 displays a 2017 NFL-NFLPA ranking of most of the helmets used in the NFL today, with the top performing helmets on the far left along with the corresponding Virginia Tech STAR values. Eleven helmets given a ranking by the NFL-NFLPA system were not ranked by the STAR system and were assigned an arbitrary STAR value of 1 to provide visual separation in the plot. The dashed line in the plot represents a linear model regression of the STAR value as a function of NFL-NFLPA ranking. In the regression, data points where $\text{star}_{\text{value}} > 0.75$ were excluded to eliminate the helmets assigned an artificial unit STAR rating, as was the Riddell VSR-4 point, which was visually identified as an outlier. The regression line is gently positively sloped.

$$star_{value} = 0.244 + 0.002 * nfl_{ranking},$$

This indicates a weak agreement between the two metrics. The regression has a multiple R-squared value of 0.077 and an adjusted R-squared value of 0.028, both of which support the conclusion that the NFL-NFLPA ranking is not a good predictor of the STAR value.

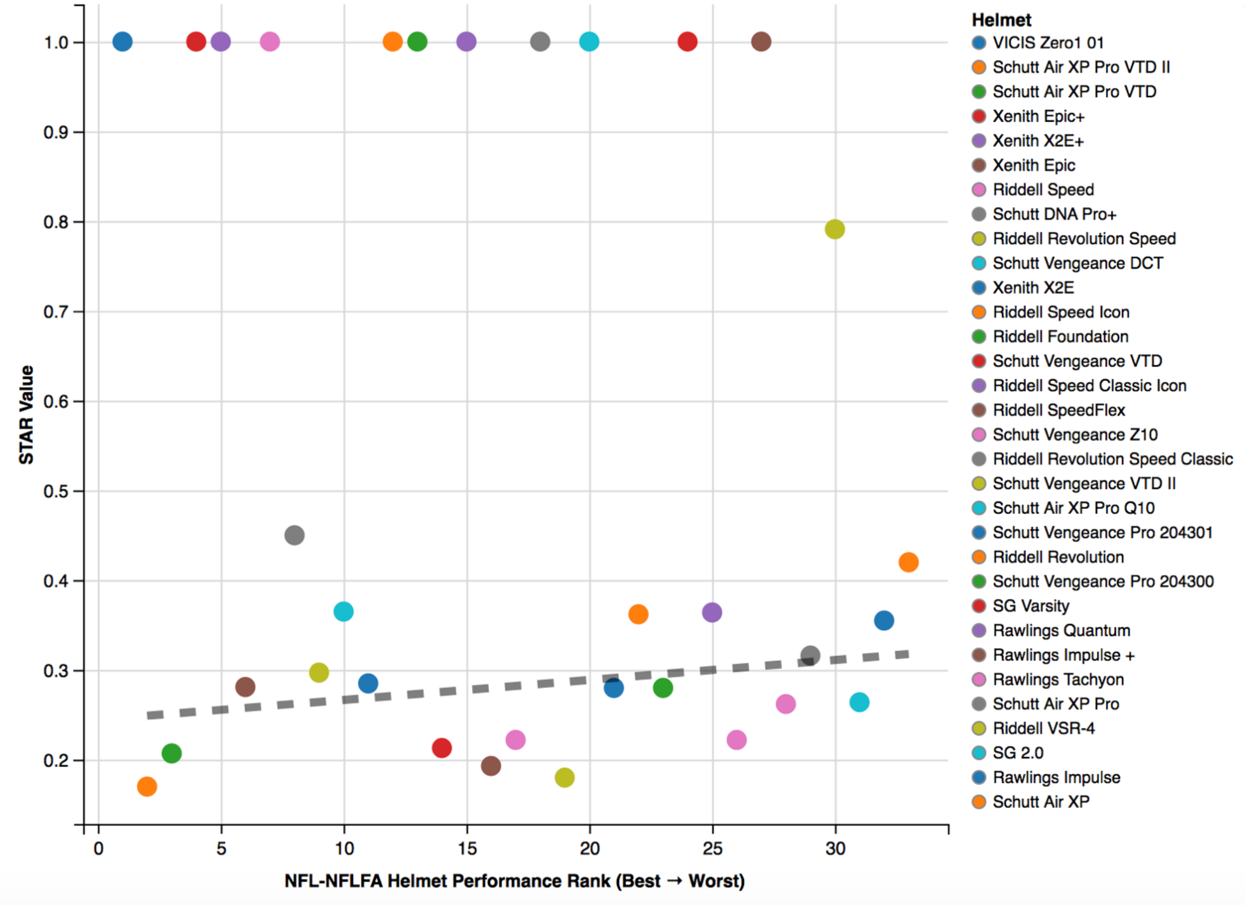

Figure 9: Comparison of helmet performance rankings between the NFL-NFLPA and STAR system. Helmets that were tested by the NFL-NFLPA but not by Virginia Tech are arbitrarily assigned a STAR value of 1 to provide visual separation in the plot and are not included in the linear regression model used to fit the data.

The NFL helmet test has been criticized for its general location testing and evaluations at high-impact velocities. These impact locations are tested with direct impact of a ram, but real-world impacts often involve complex angles and shear forces. Robert Cantu, vice-president of the board of directors for NOCSAE, stated that if helmet performance is determined at only these high velocities then helmets will stiffen to perform better at large impacts, compromising performance at lower impacts (such as subconcussive impacts), which are more common and may also be harmful to the athlete [46].

Research suggests that using linear acceleration as a brain injury metric, such as HIC15, may not be correlated to brain injury [25, 27, 28, 47, 48]. The NFL helmet test addresses the FEHM criticism by Tinard et al. by using a deformable neck that allows the headform to move after the impact. The use of a sled allows for rigid body motion of the system after impact, but no verification has been done to determine whether this motion is consistent with full human body kinematics [49].

1.2.4.1 Dummy-to-dummy impact re-creation

To re-create and simulate all impact kinematics, a helmet test that uses crash test dummies has been developed. These models are fitted with gear, use adjusted weights, and are positioned to re-create impacts that caused head injury. This allows for direct measurement of head kinematics during the impact, giving insight into how specific impacts may cause brain injury. These studies use high-speed video of impacts to determine precise model configuration. The models are then tested with different helmets to determine performance. By comparing helmets in realistic tests, full helmet performance can be assessed. This helmet testing method is new, and few results have been published yet. Little is known about the accuracy of the testing, but criticisms for its poor repeatability and many complex variables have come from several sources [50].

1.3 CRITICISM OF HELMET TESTING PRACTICES AND STANDARDS

Football helmets worn today are designed to reduce the risk of skull fracture and TBI [51]. High linear acceleration was originally thought to be the primary kinematic mechanism of TBI [18, 52, 53]. This led equipment manufacturers to refine their helmets to mitigate linear acceleration [53]. Helmet testing standards and practices were simultaneously developed, relying on predefined tolerance levels of linear acceleration response. These standards have been highly successful in reducing the incidence of catastrophic brain injuries [6, 16, 53] , however, they have done little to mitigate the risk of concussions [54].

More recently, it has been suggested that rotational acceleration may play a substantive and causative role in concussion [47, 48, 53, 55, 56, 57, 58, 59, 60, 61]. Kleiven has argued that rotational acceleration is a more likely link to the underlying mechanism of concussion because the bulk modulus of brain tissue is a million times larger than the shear modulus [62]. Gabler et al. also found a strong correlation between brain tissue strain, estimated via FEM simulations of head impacts, and metrics based on rotational velocity, supported by the findings of Takhounts and colleagues [25, 27, 63]. Despite these findings, there is currently no head injury safety standard that considers rotational response attenuation to impact [6]. Rotational response, however, is proposed to be included in future helmet performance test standards, however even these proposed metrics fail to include the directional sensitivity of brain injury. For example, NOCSAE has proposed a peak resultant rotational acceleration limit of $6{,}000\ \mathrm{rad/s^2}$ for any pneumatic ram test [36, 37], but does not take into account individual acceleration directions or durations. Rowson and Duma have stated that the STAR system will be enhanced in future iterations to include rotational kinematics in their helmet performance assessments [27].

Additional support for including rotational injury metrics in helmet performance testing comes from brain injury simulation analyses using different FEHMs. While brain injury has been shown to correlate well with CSDM, HIC15 shows low correlation to brain injury, suggesting that a linear acceleration based injury metric may be an insufficient metric to assess helmet performance [25, 27, 28, 64, 65].

Another criticism of current testing standards and performance metrics is in the aggregation of measured impact response variables. The NFL-NFLPA and STAR metrics are based on weighted averages over tests conducted across multiple impact sites. Theoretically, a given helmet may perform very poorly at one impact site but perform well, relative to other helmets being tested, at other sites. The poor result may be masked by totaling the results to form a single performance index. On the playing field one hit to the underperforming site can result in dangerous accelerations, despite the seemingly high-rating helmet. Although indices are a convenient means of summarizing information, they may mask potential and critical failures by aggregating the gathered response statistics. This suggestion is in part supported by the BrIC finding that the critical angular velocities have a directional sensitivity. Likewise, the NFL recently released information regarding the most common impact locations that lead to concussive events. Impacts to the temporal region accounted for 44% of the concussive blows. Testing and helmet design need to account for these physiological determinates of concussion and potentially weight testing in favor of common impact sites and those sites with a higher likelihood of injury [66].

Chapter 2. SIMULATING BRAIN RESPONSE IN IMPACT TO DETERMINE HELMET PERFORMANCE

A major challenge of football helmet development is correlating current helmet testing metrics with actual brain injury that might be sustained during play. Finite element modeling (FEM) allows for testing of different impacts to determine the severity of brain injury in a computational model. In this chapter, a brain simulation was done to give a simulated injury helmet performance value in standard helmet impacts. This gives an injury metric that bases helmet performance directly on injury rather than the indirect correlations made with linear and rotational kinematics.

2.1 METHODS

2.1.1 *Helmets and Impact Locations*

Testing was conducted on the top-performing helmets from each current major helmet manufacturer according to the 2017 NFL and NFLPA helmet performance ranking: the VICIS ZERO1, Riddell Speed, Schutt Air XP Pro VTDII, and Xenith Epic+ [67]. Four impact locations were chosen based on the NFL concussive impacts video analysis from the 2015-2016 seasons. The four impact locations chosen—the side (C), front (F), rear (R), and lower side (UT)—represent >74% of all concussive impacts in the NFL (Figure 10) [68]. The protocol for helmet impact and impact locations were taken from the NFL and NFLPA helmet testing procedures to ensure consistent methodologies.

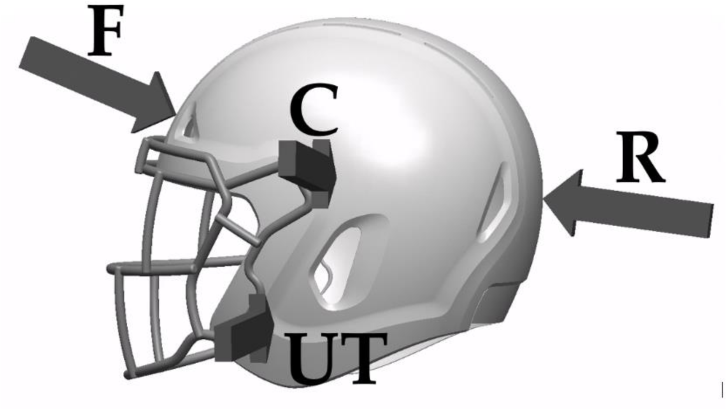

Figure 10: Impact locations on a standard football helmet. The front (F), rear (R), side (C), and lower side (UT) were chosen based on impact analysis from NFL video review (F, R, C) and highest CSDM15 values following impact in ex vivo testing (UT).

2.1.2 *Test Conditions*

Each helmet was fitted onto a calibrated hybrid III headform and fitted according to the manufacturer's fitting instructions. This involved using the appropriate-sized helmet for the hybrid III headform, proper chinstrap tension, and the use of the positioning indicator from the National Operating Committee on Standards for Athletic Equipment (NOCSAE) nose gauge guide to set proper brow height. Resultant head accelerations and rotational velocities were measured by transducers in the head. The neck was mounted onto a linear bearing table, which allowed the head/neck system to move during the impact. This motion was designed to simulate the kinematic effects of the player's body.

The four impact locations were each tested at 5.5 m/s, 7.4 m/s, and 9.3 m/s, a method that is consistent with previous methodologies [69]. Three trials at each location and velocity were performed.

2.1.3 *Testing Equipment*

Helmeted impacts were performed using a Biokinetics pneumatic linear impactor, which uses compressed air to propel the impacting ram to a specified initial impact velocity (±0.9%). The ram is released from the drive system before contact to ensure a free flight impact at the specified energy. The ram is made from aluminum with a mass of 14 Kg (31 lb) to simulate the effective mass of a football player during an on-field impact. The impacting end of the ram has a 127-mm (5-inch) diameter plate of aluminum to mount simulated impact surfaces. This test used a simulated helmet impact surface consisting of 41 mm of Der-Tex VN600 impact foam and a 127-mm-radius spherical nylon cap with a 127-mm base to closely model a "helmet-to-helmet" blow. The velocity of the impactor was measured by a twin beam trip setup, where the second beam was tripped just before impact to ensure accurate velocity measurement and impact speed.

2.1.4 *In Test Procedure*

To ensure equal comparison, all helmets were fitted with a 212E facemask (2 top bars, 1 center bar, 2 vertical middle bars, with eye guard protective bars). Each helmet was fitted to the headform and impacted in order from low- to high-velocity impacts. Three minutes of rest were allowed between impacts to ensure proper helmet recovery and stabilization of the air pressure in the impactor. In between each impact, the helmet fit was checked and reset.

2.1.5 *FEM Brain Simulation*

To determine the effect of helmet performance on brain injury prevention, the SIMon FEM was used. The SIMon FEM consists of 42,500 nodes and 45,875 elements. Brain anatomy in the model included the skull, cerebrum, cerebellum, brainstem, ventricular system, pia, and parasagittal blood vessels, as well as the dural structures, the falx and tentorium [27]. To run the brain injury simulation and simulate brain impact response, the FEM was input into LS Dyna (Livermore Software Technology Corporation, Livermore, California, USA), an explicit FEM solver for high-speed impacts. A script was then written to derive the cumulative strain damage measure, CSDM15, or brain volume percentage that exceeded 15% strain, as well as the CSDM25, or brain volume percentage that exceeded 25% strain.

2.1.6 *Injury risk*

The risk of injury was calculated by plotting the resultant CSDM25 values obtained from SIMon onto abbreviated injury scale (AIS) curves as described by Takhounts et al. [25] The AIS is an anatomically based global severity scoring system. It ranges from one (minor injury) to six (catastrophic and untreatable). Concussion is considered an AIS 2 injury and DAI an AIS 4 injury; these are of the most interest for football-related injuries [70]. Data are presented as mean ± standard deviation.

2.1.7 *Statistical Analysis*

Each helmet underwent repeated impacts (n=3) for each location and velocity to obtain raw data on resultant force and acceleration. Results were compared using ANOVA statistics and were performed using MATLAB (Mathworks, Natick, Massachusetts, USA).

2.2 RESULTS

The average CSDM15 across all helmets was significantly lower for front and rear impacts, which is consistent with previously reported NFL data. Upper side impacts (C), which make up approximately 40% of concussions sustained in the NFL, showed the largest difference between helmets, especially at the higher impact velocities. The average CSDM15 for impacts of 7.4 m/s were 0.36 ± 0.03, 0.43 ± 0.02, 0.54 ± 0.01, and 0.59 ± 0.06 for the VICIS ZERO1, Riddell Speed, Schutt Air XP Pro VTDII, and Xenith Epic+ helmets, respectively (p=0.0007). A similar result was seen at the highest impact velocity, 9.3 m/s, with an average CSDM15 of 0.56 ± 0.001, 0.63 ± 0.01, 0.76 ± 0.01, and 0.87 ± 0.01 for the VICIS ZERO1, Riddell Speed, Schutt Air XP Pro VTDII, and Xenith Epic+ helmets, respectively (p<0.0001). Lower side impacts (UT) resulted in the highest CSDM15 among all impact locations when controlling for impact velocity (Figure 11). The CSDM25 showed similar trends (Figure 12).

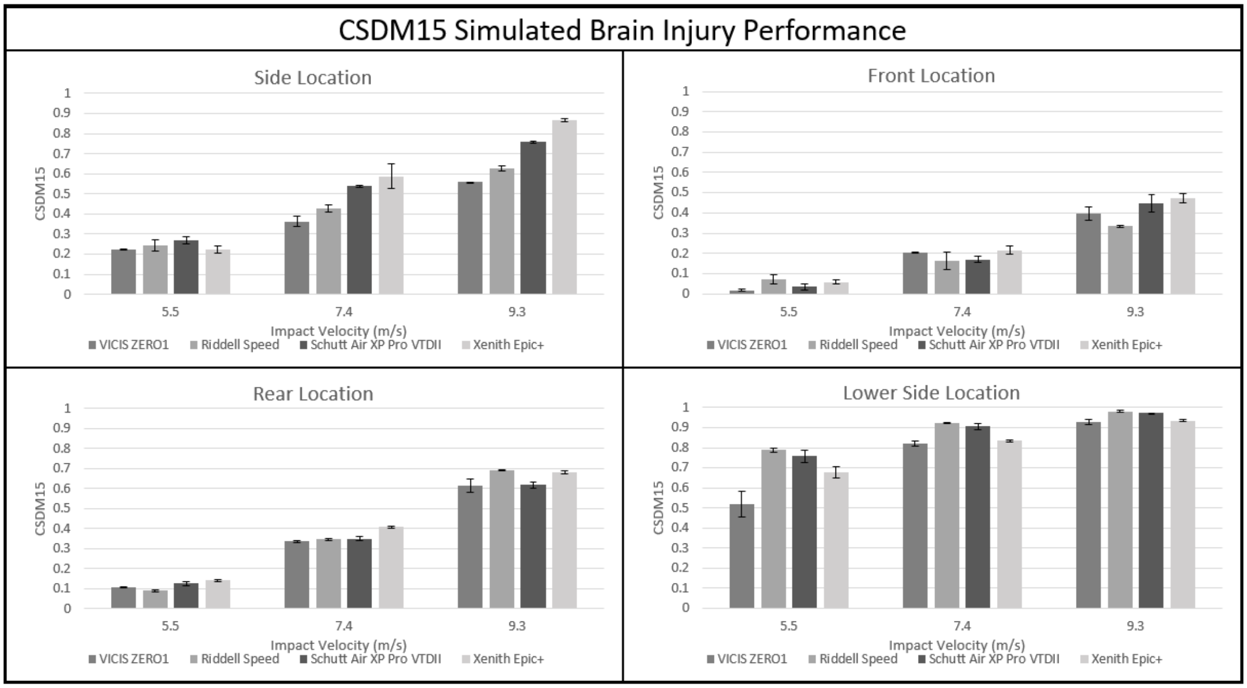

Figure 11: Average CSDM15 values are shown for each impact location. Helmets were tested at three impact velocities to simulate low-, medium-, and high-velocity impacts. Side and lower side impacts resulted in a significantly larger CSDM15 than front or rear impacts.

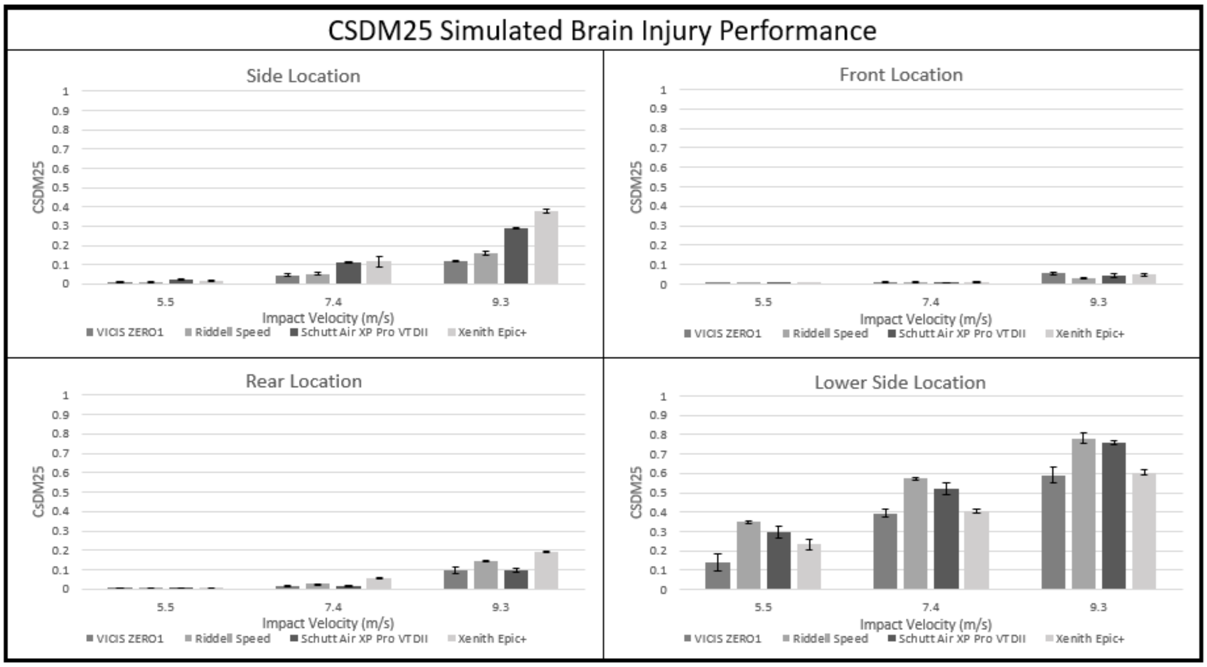

Figure 12: Average CSDM25 values are shown for each impact location. Helmets were tested at three impact velocities to simulate low-, medium-, and high-velocity impacts. Side and lower side impacts resulted in a significantly larger CSDM25 than front or rear impacts, similar to that seen for CSDM15.

To compute the risk of injury, the CSDM25 values computed from SIMon were plotted along previously calculated AIS injury curves [25, 71]. The risk for AIS2 injury (e.g., concussions) was <1% for all helmets tested for side impacts at an impact velocity of 5.4 m/s. The Riddell Speed and VICIS ZERO1 helmets had a ~1% risk of AIS2 injury for side impacts at 7.4 m/s while the Schutt Air XP Pro VTDII and Xenith Epic+ had a 5.8% and 7.3% risk, respectively ($p=0.0440$). At 9.3 m/s, there was a 6.5%, 15.5%, 55.4%, and 82.3% risk of AIS2 injury for side impacts for the VICIS ZERO1, Riddell Speed, Schutt Air XP Pro VTDII, and Xenith Epic+ helmets, respectively ($p<0.0001$). Similar trends were seen for other AIS injury levels (Table 3). Lower side impacts (UT), which had higher CSDM15 and CSDM25 values as upper side (C) impacts, also had a higher risk of injury. The risk of AIS2 injury for these impacts at 5.4 m/s were 12.7%, 74.5%, 58.1%, and 36.6% for the VICIS ZERO1, Riddell Speed, Schutt Air XP Pro VTDII, and Xenith

Epic+ helmets, respectively (p=0.0002). Risk of AIS2 injury was higher for subsequent tests at 7.4 and 9.3 m/s, consistent with the CSDM15 and CSDM25 data (Table 4). Front and rear impacts had modest injury risks, which were significantly lower than side or lower side impacts.

Table 3: Risk of injury as calculated by AIS injury curves associated with upper side (C) impacts at three velocities

AIS Injury Level	**Speed**	**Helmet**				**P-value**
		VICIS ZERO1	**Riddell Speed**	**Schutt Air XP Pro VTDII**	**Xenith Epic+**	
AIS 1	5.5 m/s	0.021 ± 0.012	0.021 ± 0.019	0.095 ± 0.032	0.028 ± 0.013	0.006
	7.4 m/s	0.359 ± 0.080	0.456 ± 0.098	0.990 ± 0.000	0.958 ± 0.055	<0.0001
	9.3 m/s	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.99
AIS 2	5.5 m/s	0.003 ± 0.001	0.003 ± 0.001	0.005 ± 0.001	0.003 ± 0.001	0.017
	7.4 m/s	0.010 ± 0.002	0.012 ± 0.002	0.058 ± 0.005	0.073 ± 0.054	0.044
	9.3 m/s	0.065 ± 0.006	0.155 ± 0.031	0.554 ± 0.021	0.823 ± 0.033	<0.0001
AIS 3	5.5 m/s	0.003 ± 0.001	0.002 ± 0.001	0.005 ± 0.001	0.003 ± 0.001	0.02
	7.4 m/s	0.007 ± 0.001	0.008 ± 0.001	0.020 ± 0.001	0.024 ± 0.014	0.041
	9.3 m/s	0.022 ± 0.001	0.045 ± 0.009	0.192 ± 0.010	0.365 ± 0.029	<0.0001
AIS 4	5.5 m/s	0.003 ± 0.001	0.003 ± 0.001	0.006 ± 0.001	0.003 ± 0.001	0.018
	7.4 m/s	0.009 ± 0.001	0.010 ± 0.001	0.019 ± 0.001	0.021 ± 0.008	0.013
	9.3 m/s	0.020 ± 0.001	0.033 ± 0.005	0.119 ± 0.006	0.231 ± 0.020	<0.0001
AIS 5	5.5 m/s	0.001 ± 0.000	0.001 ± 0.001	0.002 ± 0.000	0.001 ± 0.000	0.019
	7.4 m/s	0.004 ± 0.000	0.004 ± 0.000	0.010 ± 0.001	0.012 ± 0.007	0.041
	9.3 m/s	0.011 ± 0.001	0.023 ± 0.005	0.103 ± 0.005	0.208 ± 0.019	<0.0001

Table 4: Risk of injury as calculated by AIS injury curves associated with lower side (UT) affects at three velocities

AIS Injury Level	**Speed**	**Helmet**				**P-value**
		VICIS ZERO1	**Riddell Speed**	**Schutt Air XP Pro VTDII**	**Xenith Epic+**	
AIS 1	5.5 m/s	0.920 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.441
	7.4 m/s	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.99
	9.3 m/s	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.99
AIS 2	5.5 m/s	0.127 ± 0.095	0.745 ± 0.032	0.581 ± 0.116	0.366 ± 0.110	<0.0001
	7.4 m/s	0.853 ± 0.047	0.990 ± 0.000	0.982 ± 0.015	0.877 ± 0.020	<0.0001
	9.3 m/s	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.990 ± 0.000	0.99
AIS 3	5.5 m/s	0.038 ± 0.025	0.302 ± 0.022	0.209 ± 0.057	0.115 ±0.042	<0.0001
	7.4 m/s	0.397 ± 0.051	0.756 ± 0.014	0.661 ± 0.070	0.420 ± 0.023	<0.0001
	9.3 m/s	0.780 ± 0.072	0.967 ± 0.022	0.951 ± 0.007	0.808 ± 0.024	<0.0001
AIS 4	5.5 m/s	0.029 ± 0.015	0.188 ± 0.014	0.129 ± 0.035	0.073 ± 0.025	<0.0001
	7.4 m/s	0.253 ± 0.036	0.553 ± 0.015	0.463 ± 0.062	0.269 ± 0.016	<0.0001
	9.3 m/s	0.583 ± 0.083	0.855 ± 0.034	0.831 ± 0.016	0.610 ± 0.028	<0.0001
AIS 5	5.5 m/s	0.019 ± 0.013	0.168 ± 0.014	0.113 ± 0.033	0.060 ± 0.023	<0.0001
	7.4 m/s	0.229 ± 0.034	0.514 ± 0.014	0.427 ± 0.059	0.244 ± 0.015	<0.0001
	9.3 m/s	0.543 ± 0.081	0.818 ± 0.037	0.792 ± 0.018	0.569 ± 0.027	<0.0001

Representative images of SIMon FEM of each helmet undergoing impact at the side location at 9.3 m/s are shown to illustrate the most severe impact at the location most likely to cause concussion. Strain was decreased in the cortex, subcortical white matter, brainstem, and cerebellum in the VICIS ZERO1 helmet compared with the three other helmets (Figure 13).

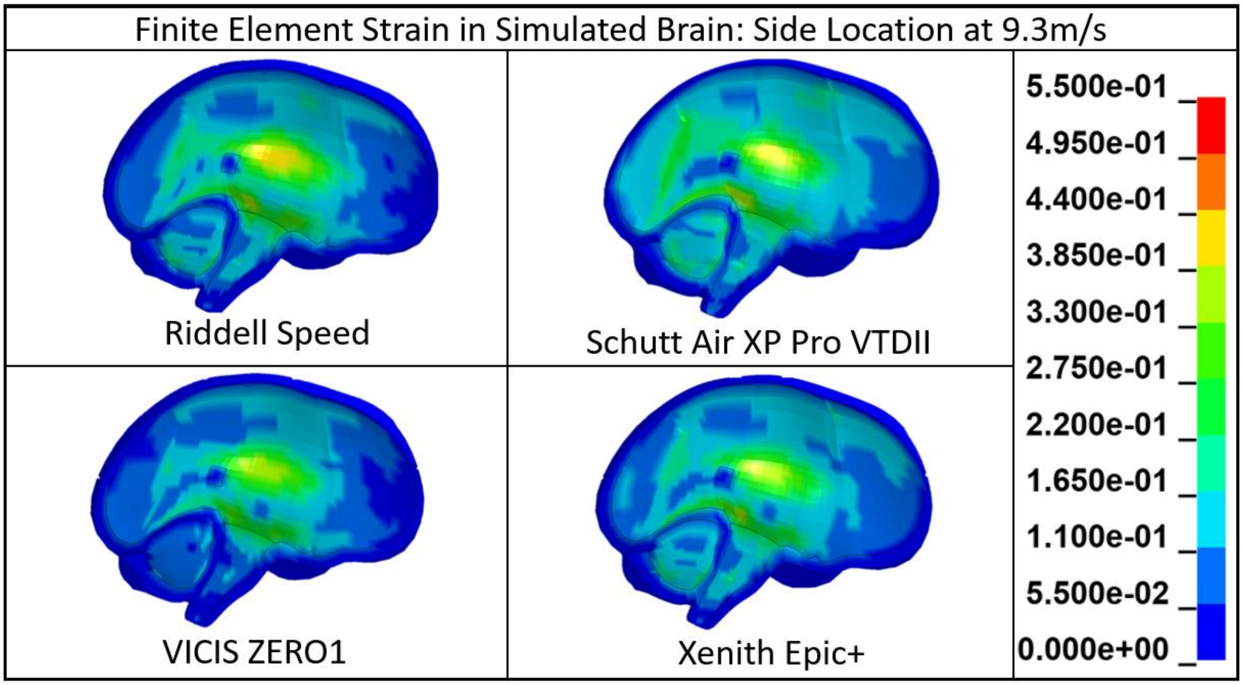

Figure 13: Representative sagittal images of plastic strain; the SIMon output are shown for all tested helmets undergoing a side (C) impact at 9.3 m/s.

2.3 DISCUSSION

Sports-related concussions can lead to significant long-term morbidity. Although multiple avenues for prevention have been implemented, it is clear that improvements in helmet technology are needed to make American football safer for participants at all levels of play. Football helmets were first mandated in 1905 after 19 players died as a result of cranial trauma in a single season; the earliest helmets were made from soft leather. Helmets underwent continued improvements, with the first plastic helmet being introduced in 1939. Other improvements, including the addition of a facemask and chinstrap, were later added. Modern helmets are composed of multiple layers, in some cases including deforming shells, and can be custom tailored to head size [72, 73].

The importance of continued helmet development and testing standards is critical for player safety as participation in American football has an inherent risk of concussive and sub-concussive impacts to the head. As the science of concussion and the pathophysiologic consequences of repeated mild head trauma are further elucidated, it goes without question that improving helmet technologies to reduce head impact forces can only be beneficial [74, 75].

One of the major challenges in helmet design and testing is identifying the forces responsible for causing concussion and working to minimize them without causing a subsequent increase in other injury-causing forces. Early helmet design focused on decreasing the energy transferred as a result of linear velocity; however, although these forces may be responsible for skull fractures resulting in intracranial hematomas, it is more likely that rotational velocity and rotational acceleration are the causative forces most responsible for central nervous system injury and resultant mTBI [76, 77]. Brain injury as a result of rotational forces has been analyzed by *post-hoc* video and wearable sensor analysis of football collisions [29, 40]. However, because of the nature of injury testing, rigorously controlled human testing can be tedious and impractical. FEM of the brain and the resultant injury risk curves represent a computational alternative to *in vivo* experimentation. Previous studies have used FEM to study the brain's response to cranial impacts. Using an FEM model, Post et al. showed that impact conditions, such as a centric and noncentric impacts, greatly affect parts of the brain associated with concussion. In addition, significant strain values are seen in cortical sulci of FEM models [78], similar to the location of the pathognomonic lesions found in postmortem brain specimens of former players with CTE [79].

Although multiple brain FEMs exist, we chose to use the SIMon FEM for its ease of availability and widespread use in other studies of TBI [28]. The topology of the model was derived from computed tomography scans and represents the mass of a 50th-percentile male [27]. The output, CSDM, is the computational equivalent of tensile strain experienced by the brain. CSDM15 represents the volume fraction of the brain that has exceeded tensile strains over 15%

and has been shown to be the approximate level for which axonal swelling and loss of axonal transport occur [59, 80]. Similarly, CSDM25 represents the volume fraction of the brain that has exceeded 25% strain. In validating FEM data to previous cadaveric studies, a CSDM25 level of 55% correlated with a 50% probability of DAI [28, 81, 82, 83].

This is the first study to compare top-performing football helmets using a FEM. Our data show that there are statistically significant differences in CSDM15 and CSDM25 values between four of the top-performing NFL football helmets. In addition, by calculating the risk of AIS injury, we further show there is a significant difference among helmets in preventing neurological injury in a computational model. Other studies have attempted to compare helmet performance in actual game settings. Level III evidence from retrospective studies suggests that newer helmet designs may significantly decrease the rate of concussion [43, 84]. Although limited in scope and methodology, these studies suggest improved technology can provide a measured improvement in concussion risk and should be further validated with improved research methodologies.

Based on these results, we argue that FEM should be considered as part of the testing standard for new football helmets. FEM allows for analysis of the brains response to cranial impact and can provide useful information regarding the risk of traumatic brain injury. Four of the top-performing helmets worn by NFL athletes were tested, and a clear difference in brain deformation in response to cranial impacts was found. Multiple testing groups certify American football helmets using various testing methods. The American Society for Testing and Materials (ASTM) certifies helmets based on a series of metrics including response to drop tests, proper fit, and response to environmental conditions. A second certifying body, the National Operating Committee on Standards for Athletic Equipment (NOCSAE), uses a series of drop tests and measures the Gadd Severity Index to quantify the response to a linear impact. Further testing to be enacted in June 2018 will include rotational acceleration testing.

The Summation of Tests for the Analysis of Risk (STAR), developed by Rowson and Duma, was the first testing metric designed to inform consumers about helmet safety performance using a 1 (worst)- to 5 (best)-star rating system by subjecting helmets to a series of drop tests, similar to the NOCSAE standard. Criticism of this metric, as with others, is focused on its lack of testing rotational forces that are thought to be the primary cause of concussion, although future testing will include rotational acceleration testing [7]. Perhaps the most comprehensive metric was developed by the NFL and the NFL Players Association. In 2011, the NFL established a working group to study helmet testing standards, and ultimately, a new metric was developed with all testing performed by Biokinetics Inc. Impacts are performed at three velocities at eight different helmet locations. The resulting output is a summation of the Head Injury Criterion 15 (accounts for linear acceleration), rotational acceleration, and rotational velocity and is the only current metric to include rotational components [85]. Testing of helmets in this study was performed in strict accordance with the NFL testing protocols.

Limitations of this study include generalizing CSDM15 and CSDM25 values to actual concussion risk as we utilized a previously validated algorithm to define the risk of AIS injury. These remain computational models as opposed to prospective human trials. In addition, TBI represents a spectrum of injuries that are likely athlete specific, and thus a threshold of brain deformation may cause different clinical symptoms between different athletes based on susceptibility to concussion and history of previous brain injury.

In conclusion, we present the results of four top-performing football helmets worn in the NFL. FEM allows for a novel analysis of brain deformation in response to cranial impacts and should be considered as part of standard testing of football helmets to optimize the safety of football athletes.

Chapter 3. USING SIMULATED BRAIN INJURY TO DETERMINE MOST DANGEROUS IMPACTS

Studies have shown that rotational forces are more likely the causative forces responsible for mTBI over linear forces. Furthermore, these studies have also demonstrated that the brain is more susceptible to injury based on the axis of rotation, with the neck axis (z-axis) being the most dangerous type of kinematic motion [25, 77, 86]. These findings further support why the Lower Side impact location from the last study resulted in the highest levels of simulated brain injury, with its non-centric impact location causes high levels of rotational motion about the neck axis. To explore this further, the most common NFL concussive location, was modified in varying levels of impact centricity to determine if these non-centric impacts may be the cause of the high levels of concussions for this impact site.

3.1 METHODS:

Impact tests were conducted on five popular helmets used in the National Football League (NFL): the Riddell Speed, Riddell Speedflex, Schutt Air XP Pro, Vicis Zero1, and Xenith Epic+. To evaluate the severity of non-centric impacts, variations to the most common concussion causing impact location, the side of the head [68, 87, 88], were tested. The NFL and NFL Players Associations (NFLPA) testing setup for the side impact location was used with the impact site translated horizontally in 25mm increments across ± 125mm on each side of the neck axis (Figure 14).

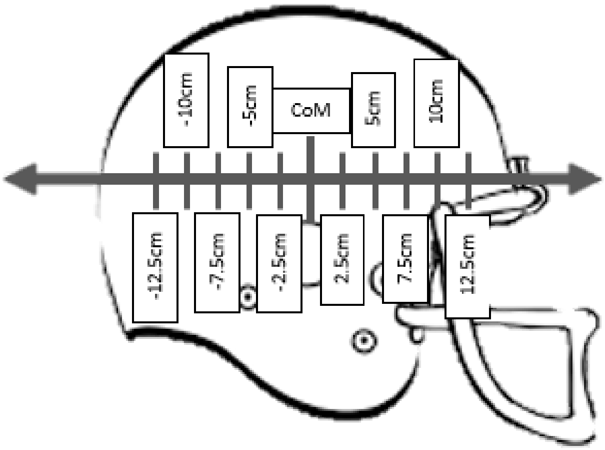

Figure 14: Impact Locations on a generic football helmet. Based on the NFL-NFLPA side impact location. Impacts were chosen based on 2016 NFL high speed video analysis of concussive impacts.

3.1.1 *Test Conditions:*

Each helmet was fitted per the manufacturers fitting instructions to a hybrid III 50th percentile male headform. Proper helmet size, chinstrap tension, and brow positioning were ensured by using the manufacturers fitting instructions. All helmets were outfitted with a 212E facemask (2 top horizontal bars, 1 center horizontal bar, 2 vertical middle bars, with eye guard protective bars). The headform was outfitted with an accelerometer and gyrometer system to measure the full kinematics of the system. To simulate the motion of the body after impact, the headform was mounted onto a hybrid III 50th percentile male neckform and a linear bearing table.

Each of the 11 impact locations were tested at 5.5m/s. Impact velocities above this caused extreme rotations of the neck resulting in wear to the equipment. Three trials at each impact location were performed for statistical analysis.

3.1.2 *Testing Equipment*

Impacts were performed using a Biokinetics pneumatic linear impactor, which uses compressed air to propel an impacting ram to a specified initial impact velocity (±0.9%). The impacting ram was designed to simulate a helmet to helmet blow matching the energy and impact mechanics of a player. The ram has a mass of 14Kg (31lb) with a 127mm (5”) diameter impacting surface. The simulated helmet impact surface consists of a 41mm (1.61”) thick, 127mm (5”)-diameter Der-Tex VN600 impact foam and a 127mm (5”)-radius spherical nylon cap with a 190.5mm (7.5”)-diameter base. The velocity of the impactor was measured at impact using a twin beam trip setup.

3.1.3 *Test Procedure*

All impacts were completed in the same session. Three minutes rest was allowed between impacts to ensure proper helmet recovery and to reset the helmet on the headform after the previous impact.

3.1.4 *FEA Brain Simulation*

The Simulated Injury Monitor (SIMon) FEM was used to simulate brain response during impact. This finite element head model includes the skull, cerebrum, cerebellum, brainstem, ventricular system, pia, parasagittal blood vessels, as well as the dural structures; the falx and tentorium [27]. SIMon was developed to simulate brain response in car crash studies and has proven to better predict brain injury than other kinematic-based metric [89]. The injury metrics used in SIMon are based on the cumulative strain damage measure, CSDM15, or the brain volume percentage that exceeded 15% strain, CSDM25, the volume percentage that exceeded 25% strain, is also used.

3.1.5 *Injury risk*

To quantify the risk of injury, CSDM25 values were plotted onto abbreviated injury scale (AIS) curves as described by Takhounts et al. [25]. AIS is an anatomically based global severity scoring system. It ranges from one (minor injury) to six (catastrophic and untreatable). Concussion is considered an AIS 2 injury, while diffuse axonal injury (DAI) an AIS 4 injury [70].

3.1.6 *Statistical Analysis*

Statistical analysis was performed using MATLAB for each impact location for the numerous injury criteria analyzed (peak linear acceleration, peak rotational acceleration, peak rotational velocity, HIC15, CSDM15, and CSDM25. To determine statistical significance between values obtained for each helmet, a one-way ANOVA was performed for each impact location and injury metric.

3.2 RESULTS:

Traditional injury metrics including peak linear acceleration (PLA), head injury criterion (HIC15), peak rotational acceleration (PRA), and peak rotational velocity (PRV) were measured to compare helmets and impact locations. PLA and HIC15 showed significant difference between centric and non-centric impacts, with centric impacts having a much higher value ($p<0.0001$). Rotational metrics did not show a statistical difference with regards to the centricity of impact. The average PLA (g) for all impacts were 55.40 ± 0.53, 46.42 ± 0.38, 72.89 ± 0.95, 39.36 ± 3.25, and 68.641 ± 1.54 ($p<0.0001$) for the Riddell Speed, Riddell Speedflex, Schutt Air XP Pro, Vicis Zero1, and the Xenith Epic+ respectively, with similar trends shown in HIC15 ($p<0.0001$) (Figure 15). The PRA was highly variable, especially for impacts to the facemask while PRV (deg/s) was more consistent with average values of 28.48 ± 0.23, 31.24 ± 0.71, 33.68

± 0.28, 27.87 ± 1.35, and 29.04 ± 1.15 (p<0.0001) for the Riddell Speed, Riddell Speedflex, Schutt Air XP Pro, Vicis Zero1, and the Xenith Epic+ respectively.

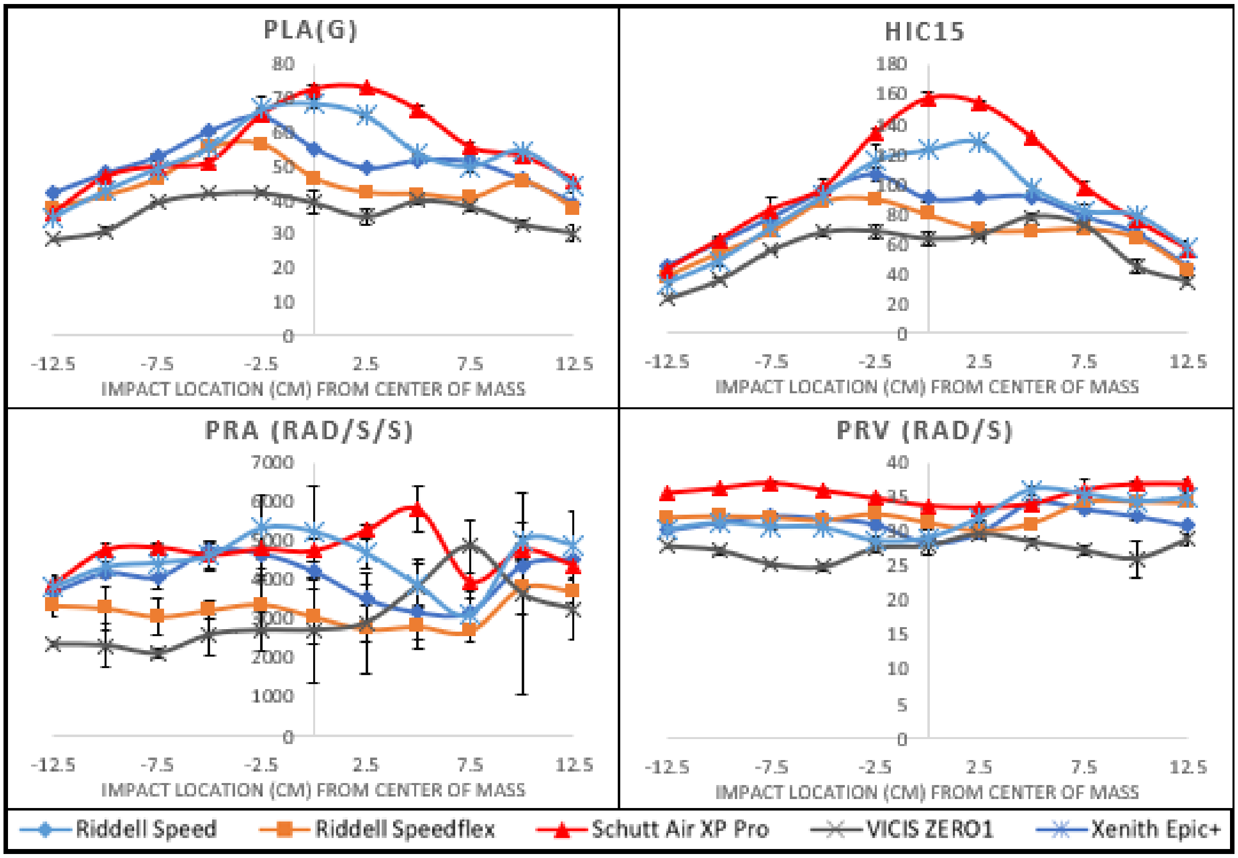

Figure 15: Traditional metrics for all helmets in impacts across the side of the helmet.

The main performance metric used in SIMon, CSDM15, showed significantly higher values on highly non-centric impacts across all helmets. These non-centric impacts also showed the largest variability in CSDM15 values from helmet to helmet. The average CSDM15 for side impacts at -7.5cm showed the largest separation between helmets and were 0.40 ± 0.02, 0.32 ± 0.02, 0.54 ± 0.01, 0.14 ± 0.01, and 0.36 ± 0.01 (p<0.0001) for the Riddell Speed, Riddell

Speedflex, Schutt Air XP Pro, Vicis Zero1, and the Xenith Epic+ respectively. CSDM25 ($p<0.0001$), AIS2+ ($p<0.0001$) and AIS4+ ($p<0.0001$) metrics showed similar trends (Figure 16).

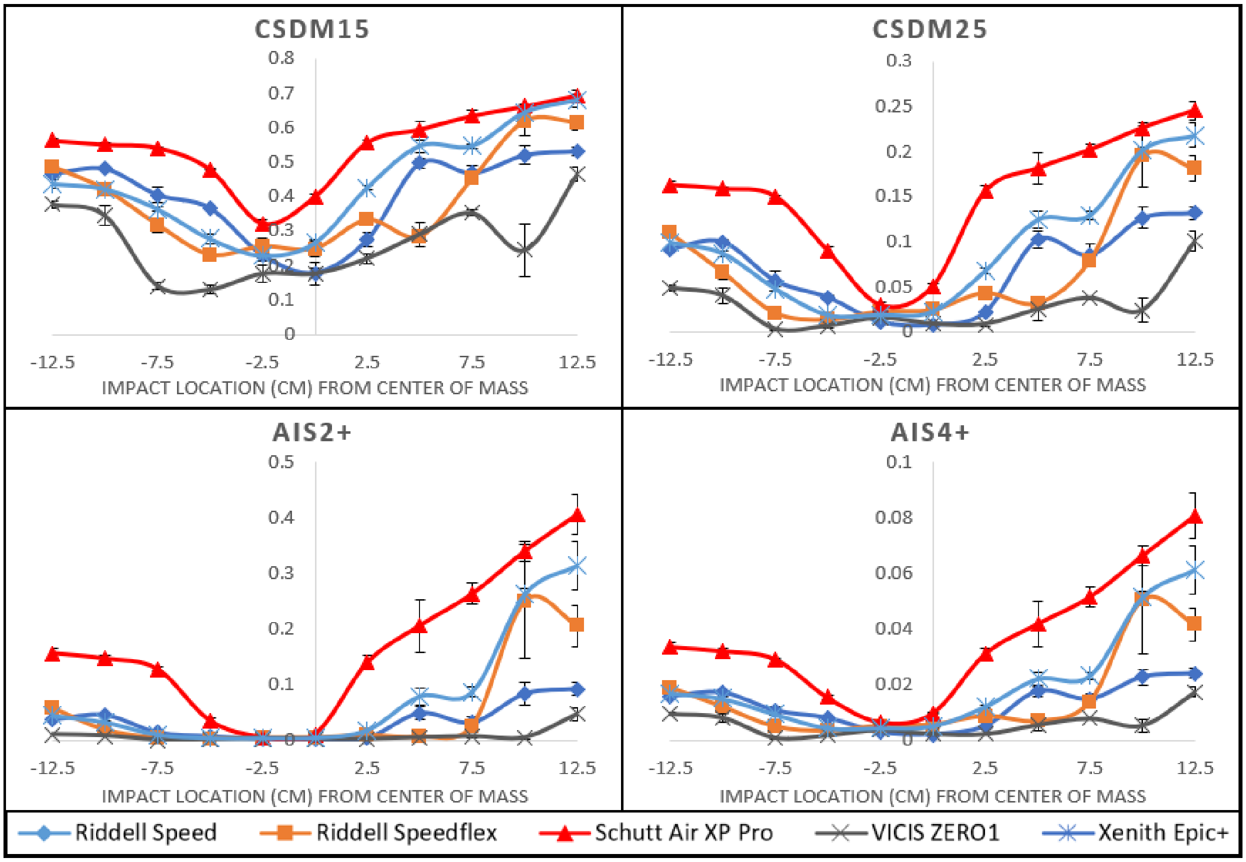

Figure 16: Finite Element damage metrics and abbreviated injury scale curves for impacts across the side of each helmet.

3.3 DISCUSSION:

The National Football League published high speed video analysis of all concussions in the NFL for the 2015-2016 seasons, showing that 59% of all NFL concussions were the result of impacts to the side of the helmet [88]. However, neither this analysis nor other studies using video analysis or lab testing has reported on the difference between centric and non-centric side impacts. The data from this study shows that centric impacts are less likely to cause brain injury than non-centric impacts. As such, we suggest that helmet testing standards include both a centric and non-centric side impact test.

Non-centric impacts have a load line that pass that creates torque and subsequent rotation of the head. Helmet performance can be evaluated by how it transfers these non-linear forces to the head. A helmet that absorbs these rotational forces rather than imparting them will reduce resultant head rotation forces and subsequent brain injury. This is especially important in impacts that cause rotation about the neck axis, as the low moment of inertia about this axis allows for rapid head rotation, increasing the chance of injury [25, 77].

Traditional injury metrics that are based on linear kinematics, PLA, and HIC15 that clinically correlate to skull fractures and intracranial hematomas [90] showed a significant difference between centric and non-centric impacts, with higher values in centric impacts. While the PRA and PRV had little to no correlation with impact centricity, FEM brain simulations showed that non-centric impacts were capable of significantly more injury. Furthermore, while it has been reported that the 5.5m/s impacts performed lie in the sub-concussive impact region [69], our brain FEM data shows that depending on impact location and centricity, the injury risk for a concussion at this impact energy can be significant.

Simulated brain injury data shows significant differences in each helmet's ability to reduce brain strain. These differences can be explained in part by each helmet's ability to reduce linear and rotational acceleration and velocity. However, the comparison between the performance metrics and the risk of injury AIS2 and AIS4 suggests that the magnitude of impact forces is not the only factor contributing to injury. Each helmet tested had different designs, materials, and impact mitigation structures that contribute differently in how they reduce forces and injury risk.

There was significant variation in PRA following impacts to the facemask, +5cm - +12.5cm. This may have been caused by the overlapping facemask bars, allowing for gripping points during an impact. An impact bar may or may not catch the impactor, resulting in large variance in helmet performance. Each manufacturer attached facemasks differently, leading to different facemask shapes despite the same style being used. It is likely that facemask designs not used in this study would also show significant variations in helmet performance at their impact location. Therefore, facemask impacts should not be considered for determining overall helmet performance.

Limitations to this study are in generalizing CSDM15 and CSDM25 simulated brain injury metrics to actual concussion risks. These remain computational models and not human trials. Additionally, the impacts used for this study are an approximation of on-field impacts. Here the rigid linear impactor was guided and not allowed to deflect away from the impact - resulting in a more extreme impact such as that by a shoulder pad and helmet collision. Lastly, TBI is a spectrum, with brain injury thresholds potentially differing between athletes.

In conclusion, we present an analysis of simulated brain injury performance following centric and non-centric impacts and we report on helmet performance of the most common helmets used in the NFL. Our data suggests that non-centric impacts lead to a higher computational risk of brain injury and thus should be included when testing for helmet performance.

Chapter 4. CONCLUDING REMARKS

A finite element head model was applied to testing impacts to simulate brain injury and to provide a helmet performance value that includes physiological weaknesses in the brain. The initial study tested the most common concussive level impact locations to show helmet performance based on simulated brain injury for these locations. This data demonstrated the high dependence of simulated brain injury on impact location. Analyzing this further to identify why certain locations produce higher levels of injury, the data supported that noncentric impacts showed higher levels of injury. These impacts were then explored in depth to determine how much more simulated injury they can cause as well as how helmet design can reduce them.

General conclusions from this research are as follows.

- Simulated brain injury is highly dependent upon impact location and moderately linear with impact energy.
- Simulated brain injury metrics can provide a statistically different separation in helmet performance between helmets in overall and individual impacts. This continues further that helmet design can reduce simulated brain injury.
- Non-centric impacts shown lower levels of traditional injury metrics than centric impacts in terms of peak linear acceleration and HIC15. However, non-centric impacts show significantly more simulated injury than centric impacts. This shows that traditional injury metrics are not good predictors of simulated brain injury and may not correlated well with actual brain injury.
- Non-centric impacts show higher levels of simulated brain injury due to non-centric impacts causing increased z-axis rotations which are more damaging to the brain.
- Simulated brain injury provides a separation in helmet performance in non-centric impacts. This Demonstrates that helmet design can reduce simulated brain injury.

Directions for future work are as follows.

- The use of a FEHM to simulate brain injury for helmet performance evaluation is impractical due to computational run times. Future work would implement a machine learning program to follow the acceleration profiles from head axis and correlate them to past simulated brain injury values. This would provide a more efficient simulation of brain injury as well as providing further insight into what types of kinematic motion is most dangerous to the brain: individual axis motion, vibration, jerk, etc.
- Future helmet performance evaluation should include simulated brain injury as a performance criterion. Its approximation of brain injury makes it an ideal metric and addresses the physiological weaknesses of the brain.
- Future helmet performance evaluation should include non-centric impacts. These impacts show higher levels of simulated brain injury and therefore should be included in helmet performance providing helmet manufacturers have a set cost function to minimize to develop better helmets.

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