Head Acceleration Experienced by Man: Exposure, Tolerance, and Applications

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Abstract

Between 1.6 and 3.8 million sports-related concussions are sustained by persons living in the United States annually. While sports-related concussion was once considered to only result in immediate neurocognitive impairment and symptoms that are transient in nature, recent research has correlated long-term neurodegenerative effects with a history of sports-related concussion. Increased awareness and current media attention have contributed to concussions becoming a primary health concern. Although much research has been performed investigating the biomechanics of concussion, little is understood about the biomechanics that cause concussion in humans. The research presented in this dissertation investigates human tolerance to head acceleration using methods that pair biomechanical data collected from human volunteers with clinical data. Head impact exposure and injury risk are quantified and presented.

In contrast to the publicly available data on the safety of automobiles, consumers have no analytical mechanism to evaluate the protective performance of football helmets. With this in mind, the Summation of Tests for the Analysis of Risk (STAR) evaluation system was developed to evaluate the impact performance of footballs helmets and provide consumers with information about helmet safety. The STAR evaluation system was designed using real world data that relate impact exposure to injury risk.

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Introduction to Concussion, Brain Injury Biomechanics, and Research Objectives

Concussions: A Problem of Increasing Concern

An estimated 1.7 million people sustain a traumatic brain injury (TBI) in the United States (US) each year [1]. Of these injuries: 52,000 result in death, 275,000 result in hospitalization, and 1.365 million are treated and released from the emergency department. The direct medical costs and indirect costs, such as loss of productivity, are estimated to be on the order of $60 billion annually in the US [2]. However, this estimate only takes into account the incidence of TBI that is reported by hospitals. Of the 1.7 million TBIs that are recorded each year, 75% are classified as mild traumatic brain injuries (mTBI), also known as concussions [3]. The incidence of concussion is thought to be vastly underestimated as many or these injuries receive no medical attention [1, 4]. Recent estimates that account for this have suggested that there are as many as 3.8 million concussions associated with sports participation alone in the US annually [4]. While sports-related concussion was once considered to only result in immediate neurocognitive impairment and symptoms that are transient in nature, recent research is correlating long-term neurodegenerative effects with a history of sports-related concussion [5-7]. Furthermore, there is concern that repetitive sub-concussive head impacts in sports may lead to neurocognitive deficits [8-10]. Increased awareness and current media attention to these issues have contributed to concussion injuries becoming a primary health concern.

Concussions are being sustained in every sport and this phenomenon has been well-documented. Table 1 displays concussion incidence rates for 15 sports that were compiled from the National College Athletic Association Injury Surveillance System [11-26]. These numbers are quantified in terms of concussions per 1000 athletic-exposures, where an athletic-exposure is defined as an athlete participating in at least 1 play of 1 game or practice. For all sports, games have a higher incidence rate than practices. Football has the highest incidence of concussion because of a high incidence rate relative to other sports and the largest number of participants. Other studies investigating concussion incidence rates in football have found similarly high rates. Guskiewicz et al. [27] reported concussion incidence rates in terms of concussions per 1000 athletic-exposures as 3.81 for games, 0.47 for practices, and 0.81 overall; which are similar to that reported in Table 1. However, Booher et al. [28] reported concussion incidence rates in terms of concussions per 1000 athletic-exposures as 5.56 for games, 0.25 for practices, and 0.74 overall. Booher et al. (2003) reported higher game injury incidence rates because that study included injuries that did not result in loss of playing time, where Guskiewicz et al. (2003) and Dick et al. (2007) only considered injuries associated with loss of playing time. To further complicate the issue, the incidence of concussion in sports is suspected to be widely under-reported, meaning that actual concussion incidence rates are likely much greater than what has been previously published. It has been suggested that as many as 53% of concussions in football are unreported, and is likely not limited to football [9, 29]. The most common reasons for players not reporting concussions was them not thinking the injury was serious enough to warrant medical attention and that they did not want to be withheld from participation [29]. Due to the popularity and high overall incidence of concussion in football, concussions in football have been put under the national spotlight. However, concussion is problem in all sports.

Table 1: Overall concussion incidence rates for 15 sports reported by the National College Athletic Association Injury Surveillance System [11-26]. An athletic-exposure is defined as a 1 athlete participating at least 1 play of 1 game or practice.

		Concussion Rate per 1000 Athletic-Exposures		
Sports	**Frequency**	**Games**	**Practices**	**Overall**
Men's baseball	210	0.19	0.03	0.07
Men's basketball	387	0.32	0.12	0.16
Women's basketball	475	0.50	0.15	0.22
Women's field hockey	129	0.52	0.09	0.18
Men's football	4404	2.34	0.21	0.37
Women's gymnastics	64	0.40	0.14	0.16
Men's ice hockey	527	1.47	0.10	0.41
Women's ice hockey	79	2.72	0.33	0.91
Men's lacrosse	271	1.08	0.12	0.26
Women's lacrosse	213	0.70	0.15	0.25
Men's soccer	500	1.08	0.08	0.28
Women's soccer	593	1.42	0.12	0.41
Women's softball	228	0.25	0.07	0.14
Women's volleyball	141	0.15	0.06	0.09
Men's wrestling	317	1.27	0.14	0.25

Definition of Concussion

The research presented in this dissertation aims to investigate biomechanics associated with concussion, but first, the definition of a concussion must be discussed. For the purposes of this work, a concussion is defined by a disturbance in brain function that is a result of a direct or indirect force to the head. Concussion is associated with changes in symptoms (such as headache), physical signs (such as unsteadiness), impaired brain function (such as confusion), and/or abnormal behavior. Table 2 is a list of symptoms that are used to help identify concussions [30].

Table 2: Concussion symptoms were evaluated using the Sport Concussion Assessment Tool 2 [30].

Concussive Symptoms Evaluated Using SCAT2	
Headache	“Don’t feel right”
“Pressure in head”	Difficulty concentrating
Neck pain	Difficulty remembering
Nausea or vomiting	Fatigue or low energy
Dizziness	Confusion
Blurred vision	Drowsiness
Balance problems	Trouble falling asleep
Sensitivity to light	More emotional
Sensitivity to noise	Irritability
Feeling slowed down	Sadness
Feeling like “in a fog"	Nervous of anxious

Concussive brain injury is unique, in that the injury has a graded response. Table 3 displays criteria that have been recently used to grade the severity of concussion occurring in sports [31, 32]. These grading scales used in sports are different from the grading scales that have been traditionally used to evaluate concussion severity.

Table 4 displays Ommaya's concussive grading scale, which ranges from transient confusion to death [33]. When neurosurgeons discuss concussion, they are normally referring to severe concussion, such as Ommaya's Grades III-VI, which is indicative of diffuse axonal injury. The form of concussion discussed in this dissertation are much less severe (Table 2 and Table 3), and would be similar to Ommaya's Grades I-II.

Table 3: Grading scales used to classify sports-related concussion [31, 32]. Concussions in this dissertation were graded using the American Academy of Neurology scale.

	Grade 1	**Grade 2**	**Grade 3**
Cantu (1986)	No LOC, post-traumatic amnesia < 30 minutes	LOC < 5 minutes or post-traumatic amnesia of 30 minutes to 24 hours	LOC > 5 minutes or posttraumatic amnesia > 24 hours
American Academy of Neurology (1997)	Transient confusion, no LOC, concussion symptoms < 15 minutes	Transient confusion, no LOC, concussion symptoms > 15 minutes	And LOC (brief or prolonged)

Table 4: Ommaya's concussive grading scale that ranges from transient confusion to death [33].

AIS Level	Concussive Brain Injury Grade	Clinical Descriptions	Pathologic Description	Outcome (1 month)
1	I	Confusion without amnesia (ding, stunned)	Not Known; CT scans usually normal; skull fractures and intracranial bleeds uncommon	Normal except for PCS and occasional vascular complications
2	II	Amensia without coma (type A, slow onset; type B rapid onset)		
3	III	Coma < 6 hours (includes classic cerebral concussion, minor and moderate head injuries)	Increasing intensity and distribution of diffuse axonal injuries and/or intracranial bleeding and other visible brain lesions; CT scans usually abnormal; skull fracture incidence 20-50%	
4	IV	Coma 6-24 hours (severe head injuries)		Morbidity increasing to 35%+ and mortality to 50%+
5	V	Coma >24 hours (severe head injuries)		
6	VI	Coma -> death within 24 hours (fatal head injuries)		

Biomechanical Characterization of Concussion

Historically, the majority of brain injury biomechanics research has defined concussion as a severe and life-threatening injury. Concussive brain injury is unique in that the injury has a graded response that can vary from minor confusion to death [33]. The sports-related concussions investigated in this dissertation are considered mild in severity. Annually in the United States, there are only an estimated 300,000 concussion concussions that result in loss of consciousness, with the remaining injuries resulting in less severe symptoms [4, 34]. However, the varying grades of concussion are likely a scaled result of the varying mechanical stimuli input to the head [33]. Previous experiments involved investigating brain injury mechanisms, and how these mechanisms relate to the kinematics of the head. Kinematic parameters of the head are related to brain injury because they are thought to be indicative of the inertial response of the brain. Ideally, the head kinematics of a human surrogate could be measured in a safety testing scenario and used to predict the tissue level response of the brain in an effort to evaluate injury potential. With this goal in mind, many researchers have studied the relationship between head kinematics and brain injury. Most experiments have investigated linear or rotational kinematics independently, as these inputs have long been thought to result in different injury mechanisms [35]. Explanations of these theories have been previously documented in great detail [36].

The Wayne State Tolerance Curve (WTSC) was developed from a series of tests on dogs and cadavers and related linear acceleration and duration of acceleration to injury tolerance [37]. Injury metric functions such as severity index (SI) and head injury criterion (HIC) were subsequently developed from analyses of the WTSC [38, 39]. These injury metrics were primarily developed to predict skull fracture, although they were thought to correlate with severe brain injury. Notably, only linear acceleration is considered in these injury metrics, and all current safety standards for head injury are based on these works. However, rotational acceleration is believed by many to be a primary mechanism for brain injury [40]. Unlike linear acceleration, there is currently no accepted injury criterion for rotational acceleration. Additionally, previous research investigating rotational kinematics has focused on animal models (primate or rat), in which pure rotational was applied to the head [33, 41-46]. These experiments, including those evaluating linear and rotational acceleration, utilize little data from humans. Cadavers have no physiologic response, and animal data cannot be directly applied to humans. Optimally, these experiments would utilize data derived from humans. However, recording potentially injurious data from humans has been challenging.

The high occurrence of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury (MTBI). Competitive football has been used as an experimental environment for collecting human head acceleration data since the 1970's. Several early studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games [47-49]. Another study instrumented football helmets directly to measure helmet acceleration [50]. While laying the groundwork for future research and providing a proof of concept, these older studies were limited in their ability to measure head acceleration and measured only a single player. Later, Naunheim et al. [51] instrumented the helmets of 1 high school hockey player and 2 high school football players (both linemen) with accelerometers to measure linear head acceleration.

However, there were no documented incidents of mild traumatic brain injury in this limited dataset.

One study has quantified head accelerations experienced by professional football players by recreating concussive impacts in a laboratory setting. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on game video [52-54]. The authors recreated 31 impacts, 25 of which were concussive. From the data collected in the reconstructed impacts, injury risk curves were developed for MTBI [40, 54]. The limitations of this study were that data were collected from ATDs rather than humans, and that the NFL dataset is biased towards concussive impacts. Several other studies have utilized a commercially available football helmet accelerometer system, Head Impact Telemetry System (HITS) (Simbex, Lebanon NH), to measure head accelerations. HITS is a six accelerometer measurement device that is integrated into existing football helmets. The measurement device records resultant linear head acceleration for every head impact a player experiences using a novel algorithm [55]. In addition, HITS reports impact location and estimates of rotational accelerations about the x and y axes of the head.

Duma et al. [56] presented a study to quantify head acceleration in collegiate football players by collecting over 3000 impacts from 38 players using 8 HITS measurement devices, in which one concussive event was measured. A subsequent study expanded this dataset to include over 27,000 impacts (4 concussions) and analyzed risk using a unique statistical analysis [57]. The nominal injury values reported representing 10% risk of concussion were a peak linear acceleration of 165 g and HIC of 400. In separate studies, Schnebel et al. [58] presented data for over 62,000 (6 concussions) head impacts experienced by 56 collegiate and high school football players recorded using HITS. Guskiewicz et al. [59] collected over 104,000 impacts (13 concussions) from 88 collegiate football players. Mihalik et al. [60] analyzed more than 57,000 these impacts from 72 players to look at positional differences in impacts. Broglio et al. [61] recorded over 54,000 head impacts (including 13 concussions) experienced by 78 high school athletes. These studies have provided great insight to the head kinematics associated with head impacts in football, but have largely been descriptive studies with small concussive sample sizes that made it difficult to draw conclusions about injury [56, 58-63].

Research Objectives

The research in this dissertation aims to biomechanically characterize sports-related concussion with respect to linear and rotational acceleration. This was accomplished by instrumenting and observing a population (football players) that is at increased risk of concussion using HIT System technology. Data were collected for several years and large head acceleration datasets were compiled. This dissertation focuses on the analysis, not collection, of these data.

The following research objectives are investigated in this dissertation:

1. To biomechanically characterize sub-concussive and concussive impacts experienced in football with respect to linear acceleration.
2. To develop a football helmet evaluation system based on real-world data that can be consumer tool for assessing helmet impact performance.
3. To develop and validate a custom in-helmet accelerometer array that can compute head acceleration with six degrees of freedom.
4. To biomechanically characterize sub-concussive and concussive impacts experienced in football with respect to rotational acceleration.

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Chapter 2:

Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion

Abstract

In contrast to the publicly available data on the safety of automobiles, consumers have no analytical mechanism to evaluate the protective performance of football helmets. The objective of this paper is to fill this void by introducing a new equation that can be used to evaluate helmet performance by integrating player head impact exposure and risk of concussion. The Summation of Tests for the Analysis of Risk (STAR) equation relates on field impact exposure to a series of 24 drop tests performed at 4 impact locations and 6 impact energy levels. Using 62,974 head acceleration data points collected from football players, true impact exposure for one full season is associated with each of the 24 drop test configurations. Next the risk of concussion for each impact is determined by new injury risk function developed from 32 measured concussions and associated exposure data. Finally, the data from all 24 drop tests is combined into one number using the STAR formula that incorporates the predicted exposure and injury risk for one player for one full season of practices and games. The new STAR evaluation equation will provide consumers with a meaningful metric to assess the relative performance of football helmets.

Introduction

Recent research has suggested that there are as many as 3.8 million sports-related concussions each year in the United States, with participation in football resulting in the highest incidence of injury [1, 2]. Studies showing the potential long-term effects of these injuries have put sports-related concussions under the national spotlight as a primary health concern [3-5]. Furthermore, there is concern that repetitive sub-concussive head impacts in sports may lead to neurocognitive deficits [6-8]. While limiting the number of head impacts in sports is an important component of reducing injury incidence, improving head protection is another essential element of injury mitigation [9]. This paper focuses on a new mechanism to evaluate the protective capabilities of football helmets.

Substantial effort has been invested in researching head acceleration in relation to brain injury [10]. Head acceleration is thought to be indicative of the inertial response of the brain, and therefore is used to predict brain injury. All head injury safety standards for automobiles and helmets (motorcycle, sports, or bicycle) use measured humanoid head acceleration (or a function of head acceleration) during specified testing conditions to determine whether a product is safe to sell to consumers. While the Federal Motor Vehicle Safety Standards (FMVSS) 201 and 208 govern whether an automobile is safe to sell using pass/fail injury criteria, the New Car Assessment Program (NCAP) provides consumers with a quantitative metric of the relative safety between automobile models [11, 12]. NCAP is a valuable tool for consumers who are concerned with safety.

In contrast to the publicly available NCAP safety data on automobiles, consumers have no information on the relative impact performance between different helmets; moreover, there is no quantified metric that provides meaningful interpretation of the test results. Any sports helmet sold to consumers must have National Operating Committee on Standards for Athletic Equipment (NOCSAE) certification. NOCSAE certification involves testing helmets through a series of drop tests, in which every drop test must result in a head form impact response below a specified threshold. The lack of publicly available safety performance data for helmets can be partially attributed to the previously limited knowledge of impact-induced brain injury and the challenge of interpreting the data; however, recent studies have provided a more complete understanding of the head acceleration patterns associated with impacts in football as well as possible concussions.

Since 2003, researchers have been instrumenting football players with the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH) to collect head acceleration data each time a player experiences a head impact [13]. While Duma et al. (2005) were the first to instrument football players with the HIT System, other researchers have adopted this technology to investigate head impacts in football [14-17]. The measurement and analysis of head acceleration data collected from these in-helmet accelerometer arrays have been well-validated and widely accepted [13, 18-20]. By instrumenting and observing a population that is at risk of concussion, researchers have collected data that can be used to provide the foundation for biomechanically characterizing concussion risk. Including all researchers collecting head acceleration data from football players, there exists over 1.5 million head impacts have been recorded to date.

The objective of this study is to develop and introduce the concept of a new evaluation system that can be used to provide quantitative insight into the protective performance of football helmets. This new evaluation system is designed to integrate the overall helmet impact performance into a singular metric that is derived from the true head impact exposure and injury risk based on real-world impact measurements in football players. This new system is analogous to NCAP, in that biomechanical data are interpreted for the public in order to provide consumers with a meaningful metric to use when deciding which product to purchase.

Materials and Methods

<u>The STAR Equation</u>

The Summation of Tests for the Analysis of Risk (STAR) equation is presented as a mechanism to compute a singular metric from a total of 24 drop tests that can be used evaluate the relative performance of football helmets. The STAR equation was developed to relate the true head impact exposure that a football player experiences throughout one full season of participation to the injury risk associated with each head impact (Equation 1). Fundamentally, this equation correlates every head impact that a football player experiences during one season of participation to 24 drop tests (4 locations x 6 drop heights) that can be performed using a NOCSAE-style drop tower and head form [21]. The products of impact exposure and injury risk associated with each impact location and drop height are summated to compute a predicted concussion incidence for one player during one season wearing a specific helmet.

$$\text{STAR} = \sum_{L=1}^{4}\left(\sum_{H=1}^{6} E(L,H)\cdot R(a)\right) \qquad \text{Equation 1}$$

Where: *STAR* is the Summation of Tests for the Analysis of Risk,

L represents one of 4 impact locations (front, rear, side, or top)

H represents one of 6 drop heights (60 in, 48 in, 36 in, 24 in, 12 in, and lowest)

E represents head impact exposure as a function of impact location and drop height

R represents injury risk as a function of peak resultant head acceleration

a represents peak resultant head acceleration resulting from each specific drop height and impact location

The methods presented in this paper detail the derivation of the head impact exposure and injury risk components of the STAR equation. Next, as an example of how the equation should be used, the STAR value of a hypothetical helmet is computed and its interpretation is discussed.

Head Impact Exposure

For the purpose of the STAR equation, head impact exposure is defined as the number of impacts that one football player will experience through one complete season of participation. Crisco et al. (2010) quantified impact exposure in detail for 3 collegiate football teams over the duration of a single season; and reported that players can experience over 1400 impacts at the highest levels over the course of one season [22]. These data were utilized for this study because their analysis investigated exposure on a per player basis. As expected, this study reported that the number of head impacts per player per season increased with the number of games and practices in which they participated [22]. The 90th percentile head impact exposure per player per season was chosen in order to account for a full season of participation at a conservatively high level. Given these data, the STAR equation uses 1000 head impacts as the total exposure that a collegiate player participating in one full season would experience.

Exposure per player per season was further investigated by impact location by utilizing 62,974 head impacts that were recorded between 2009 and 2010 with the HIT System at Virginia Tech [13, 23]. These impacts were analyzed to determine the head impact exposure on an impact location basis. Each recorded head impact was categorized into one of four general impact locations based on the computed azimuth and elevation of the impact vector (Figure 1). Any impact with an elevation greater than 65° was categorized as an impact to the top of the helmet. Impacts with elevations less than 65° and azimuths between 45° and -45° were categorized as impacts to the front of the helmet. Impacts with elevations less than 65° and azimuths between 135° and -135° were categorized as impacts to the rear of the helmet. All other impacts were categorized as impacts to the side of the helmet, given the symmetry of the human head about the sagittal plane. The number of impacts to each generalized location was normalized by the total number of impacts to determine the percentage of impacts to each location. The percentage of impacts to each location was multiplied by the number of impacts that each player experiences throughout a season of full participation to determine the number of impacts to each location per player per season. Using this normalization approach allows for the location exposure to be scaled up or down based on the targeted number of impacts throughout a season.

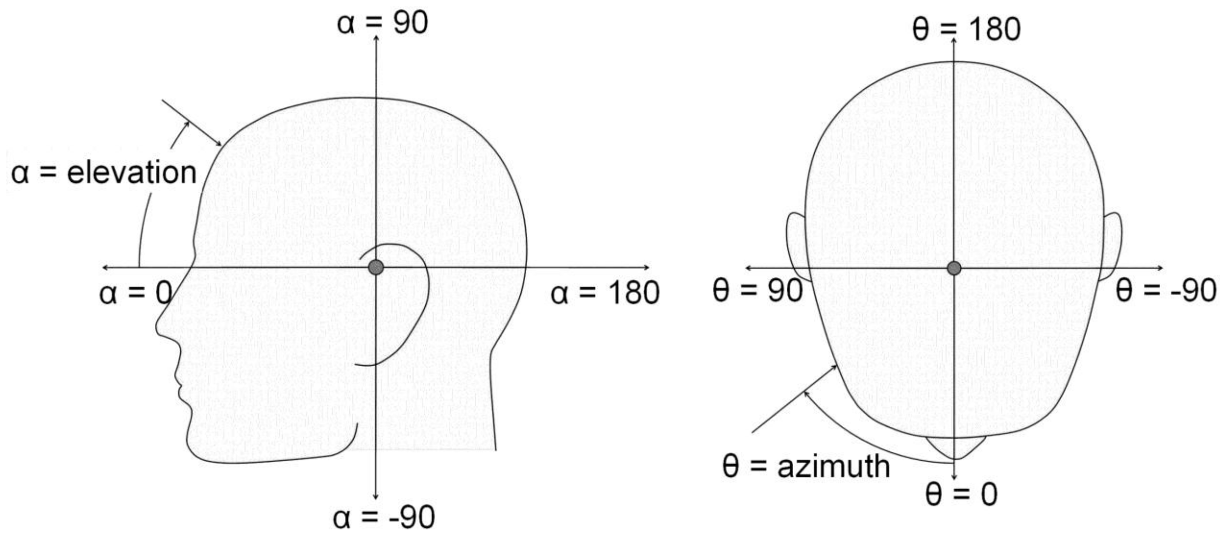

Figure 1: Definition of parameters used to group impact locations. Azimuth and elevation values for the impact vector were computed for each impact based on the head acceleration data measured.

Exposure per player per season for each impact location was further investigated by impact severity. The severity distributions were determined by using 51,008 head impacts that were recorded between 2006 and 2010 with the HIT System from players wearing Riddell VSR4 helmets (Riddell Inc, Elyria, Ohio) [13, 23]. Only one helmet type was used so that potential variations in helmet response between types would be removed for this part of the analysis. These impacts were categorized into the aforementioned impact locations and analyzed to determine head impact severity distributions. A Weibull distribution was fit to the head acceleration data for each impact location. The Weibull probability density function (pdf) takes the form of Equation 2. The pdf represents the probability of a player experiencing any given head acceleration for one impact. The area under the entire pdf curve represents the probability of an impact resulting in any head acceleration, which is 100%. Moreover, the pdf can be integrated over specific bounds to determine the probability of an impact between the upper and lower bounds. The quality of the distribution fit was assessed by comparing an empirical cumulative distribution of the raw data to the computed Weibull cumulative density function (cdf) (Equation 3). The Weibull pdf for each impact location was multiplied by the number of impacts to each impact location so that the area under each curve represented the number of impacts to each location.

$$\mathrm{w\_pdf} = \frac{\alpha(x-\theta)^{\alpha-1}}{\beta^{\alpha}} e^{-\left(\frac{x-\theta}{\beta}\right)^{\alpha}} \qquad \text{Equation 2}$$

$$\mathrm{w\_cdf} = 1-e^{-\left(\frac{x-\theta}{\beta}\right)^{\alpha}} \qquad \text{Equation 3}$$

Where: *w_pdf* is the Weibull probability density function

w_cdf is the Weibull cumulative density function

α is the shape parameter

β is the scale parameter

θ is the location parameter

x is the peak resultant head acceleration

Next, it was important to decouple the specific helmet used to collect the head acceleration data from the overall STAR analysis. In order to do this, the head acceleration distributions were transformed to impact energy distributions, which are then independent of the specific helmet response. This was accomplished by defining the relationship between impact energy and resulting head acceleration for the specific helmet used for data collection. A total of 3 Large

Riddell VSR4 helmets were tested on a medium NOCSAE head form using a NOCSAE-style drop tower. Drop heights ranged from 6 in to 66 in and were incremented by 6 in. For each of the 3 VSR4 helmets, these 11 drop heights were tested at the front, rear, side, and top impact locations that are defined by NOCSAE; resulting in a total of 132 tests. All equipment and instrumentation were calibrated to NOCSAE specification. Acceleration data were sampled at 20,000 Hz and filtered to NOCSAE specification (SAE J211 CFC 1000) for each test. Peak resultant head acceleration and severity index were computed for each test [24]. A 2nd order polynomial regression analysis was performed using a least squares technique to determine the relationship between drop height and the average VSR4 peak head acceleration for each impact location (Equation 4).

$$H = p_1 a^2 + p_2 a + p_3 \qquad \text{Equation 4}$$

Where: p_1, p_2, *and* p_3 are regression coefficients

h is drop height

a is peak resultant head acceleration

Utilizing the polynomial regression models for each impact location, head acceleration distributions were transformed to drop height distributions. Once the distributions were representative of the impact energy to the helmet, not the head acceleration resulting from the response of the helmet to impact, the number of impacts associated with varying severities was determined for each impact location. Drop heights of 12 in, 24 in, 36 in, 48 in, and 60 in were selected because they encompass the impact energies associated with a wide range of impacts seen on the field. The Weibull pdf for each location was integrated over the bounds defined in Table 5 to determine the number of impacts experienced by one player participating in one season for each drop height. Each number of impacts was rounded to the nearest integer, with the exception that any number between 0.0 and 1.0 was rounded up to 1.0. The 'lowest' category was used to describe impacts less with head accelerations less than 19 g. These impacts were separated from the analysis as they are not considered to be relevant given that the lowest reported concussion in the literature is 42 g [15, 23]. Moreover, the drop test height for this value would be only a few inches and that would not be a reasonable or necessary test to perform.

Table 5: Integral bounds used to determine the number of impacts associated with each drop height for each impact location. Note that the lowest bounds include all impacts below 19 g (median impact). This drop height is not evaluated because these impacts are not associated with injury.

Drop Height	**Integral Bounds**	
	Lower	**Upper**
Lowest	0 g	19 g
12 in	19 g	18 in
24 in	18 in	30 in
36 in	30 in	42 in
48 in	42 in	54 in
60 in	60 in	Infinity

Injury Risk

The sub-concussive head acceleration dataset utilized in the risk analysis consisted of 62,974 head impacts that were recorded between 2009 and 2010 with the HIT System at Virginia Tech.

Concussive head acceleration data were compiled from 3 separate studies using identical data collection protocols with the HIT System to create a dataset of 32 clinically diagnosed concussions [15, 23, 25]. These datasets are capable of accurately defining the distributions of head accelerations associated with sub-concussive and concussive impacts. Sub-concussive impacts were fit to a Weibull distribution in the form of Equation 2. Concussive impacts were found to be normally distributed, and therefore were fit to a normal probability density function (Equation 5).

$$n\_pdf = \frac{1}{\sigma\sqrt{2\pi}} e^{\frac{-(x-\mu)^2}{2\sigma^2}} \qquad \text{Equation 5}$$

Where: *n_pdf* is the normal probability density function

μ is the mean

σ is the standard deviation

Published injury incidence rates for game participation were used to determine the proper weighting between sub-concussive and concussive head acceleration distributions. For collegiate athletes, there are 5.56 concussions per 1000 athletic exposures, where an athletic exposure is defined as one athlete participating in at least one play of one game or practice [26]. For athletes in the National Football League (NFL), an injury incidence rate of 0.41 concussions per game was considered for professional athletes [27]. To relate the number of concussions to the number of sub-concussive impacts for both the collegiate and NFL groups, it was assumed that the median player experiences 16.3 impacts per game [22]. For collegiate athletes, 5.56 concussions per 1000 games played with 16.3 impacts per game per player can be expressed as an injury incidence rate of 0.341 concussions per 1000 impacts. For NFL athletes, 0.41 concussions per game with 88 players participating in each game and 16.3 impacts per game per player can be expressed as 0.286 concussions per 1000 impacts. It is important to note that current research suggests that as many as 53% of concussions go unreported [28]. This underreporting rate was applied to both calculated injury incidence rates, resulting in 0.726 concussions per 1000 impacts for collegiate athletes and 0.609 concussions per 1000 impacts for NFL athletes.

Next, estimated injury incidence rates were used to combine the sub-concussive and concussive head acceleration distributions in order to have the proper sub-concussive to concussive impact ratio. The weighting of each distribution based on injury incidence rates allows for an unbiased risk analysis. A logistic regression analysis based on the weighted sub-concussive and concussive head acceleration distributions was used to express risk as a function of head acceleration for both the collegiate and NFL groups (Equation 6). The regression coefficients were determined using a generalized linear model technique.

$$\text{risk}=\frac{1}{1+e^{-(\alpha+\beta x)}} \qquad \text{Equation 6}$$

Where: *risk* is the probability of injury

α and β are regression coefficients

STAR Value Assessment

For illustrative purposes, a STAR value was calculated using hypothetical head acceleration data. Example peak resultant head acceleration values were generated for the 12 in, 24 in, 36 in, 48 in, and 60 in drop tests for each impact location. Equation 6 was used to calculate probability of concussion for each testing configuration based on collegiate injury incidence rates. Risk of concussion was multiplied by the number of impacts that a player experiences for each testing configuration and summed to calculate an overall concussion incidence (Equation 1).

Results

Head Impact Exposure

Overall, impacts to the front of the helmet occurred most frequently, and were followed by impacts to the rear, top, and side of the helmet (Figure 2). Using these percentages, the number of impacts to each impact location for a single player participating in a complete season was computed based on the assumption that a total of 1000 head impact are experienced. This transformation gives that for a single season, a player will experience 347 impacts to the front of the helmet, 319 impacts to the rear of the helmet, 171 impacts to the top of the helmet, and 163 impacts to the sides of the helmet.

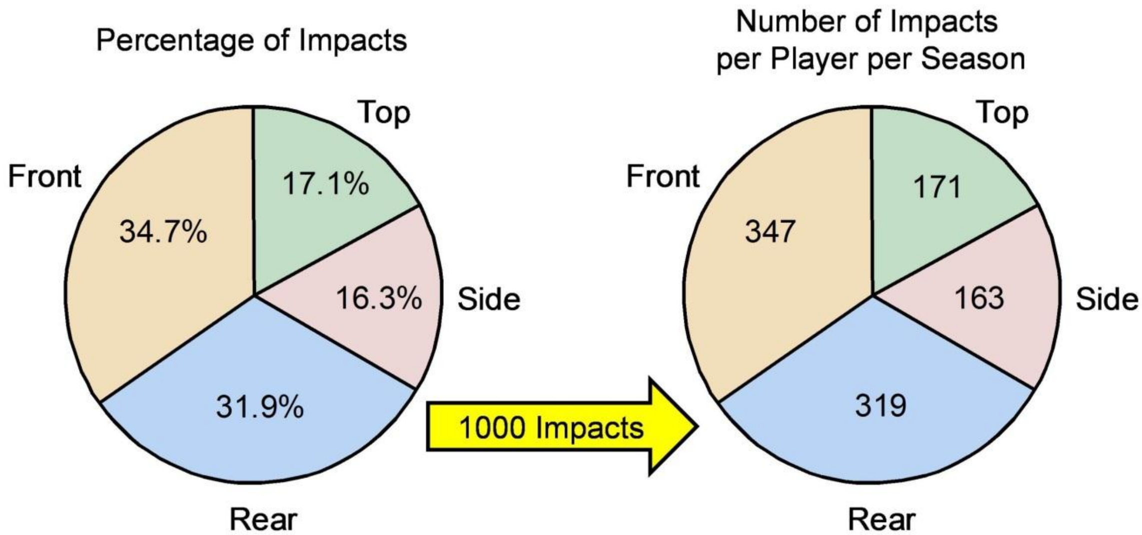

Figure 2: Percentage of impacts associated with each impact location (left). Assuming 1000 impacts per player per season, the number of impacts to each location was determined (right).

Impact severity distributions in terms of peak resultant head acceleration were determined for each impact location. Each distribution was fit to a Weibull probability density function (Equation 2) and Table 6 displays the computed parameters for each impact location. As an example, Figure 3 displays the Weibull probability density function for impacts to the front of the helmet. A comparison of the Weibull cumulative density function to the empirical cumulative density function is also show in Figure 3. The quality of fit was consistently good for all impact locations.

Table 6: Weibull probability density function parameters for impact severity distributions separated by each impact location.

Impact Location	α	β	θ
Front	15.1228	0.8897	10.0
Rear	16.5724	0.9333	10.0
Side	13.3692	0.8837	10.0
Top	21.8923	0.8899	10.0

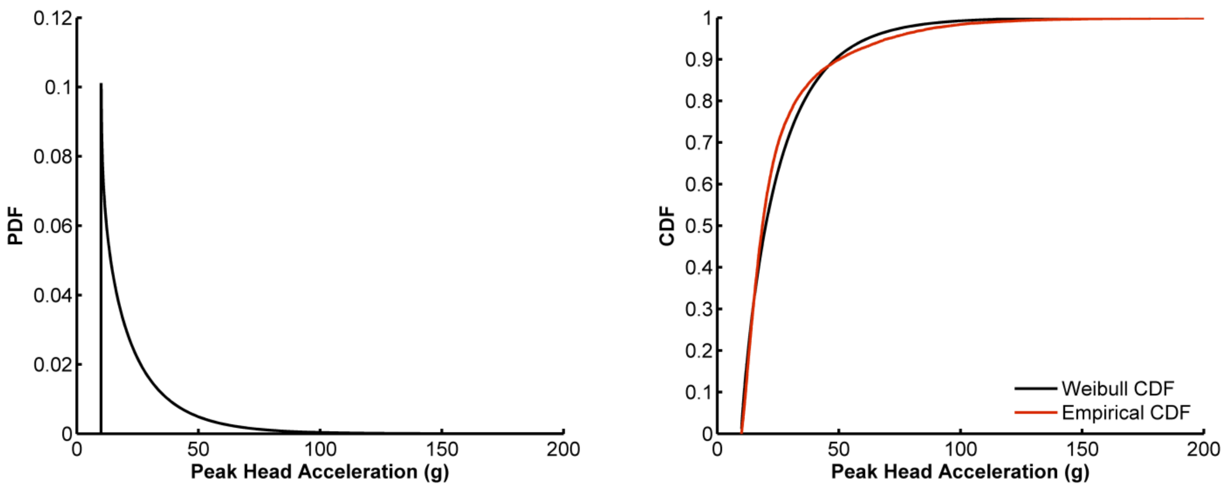

Figure 3: Weibull probability density function fitted to head accelerations resulting from impacts to the front of the helmet (left). Quality of fit can be investigated through comparison of the Weibull cumulative density function to an empirical cumulative density function of the same data (right).

A total of 132 drop tests were performed investigating average head acceleration as a function of drop height and impact location for 3 Riddell VSR4 helmets.

Table 7 displays the regression coefficients and R^2 values for the relationship between head acceleration and drop height for these tests (Equation 4). Figure 4 displays how the number of impacts for each drop height were determined from the Weibull distribution for the front impact location, while Table 8 displays the number of impacts for each drop height for all impact locations.

Table 7: The 2nd order polynomial regression coefficients and R^2 values for drop height and head acceleration relationships (Equation 4).

	p_1	p_2	p_3	R^2
Front	0.009	1.428	15.010	0.999
Rear	-0.026	3.826	5.042	0.992
Side	-0.023	3.241	12.975	0.998
Top	-0.015	2.870	11.403	0.997

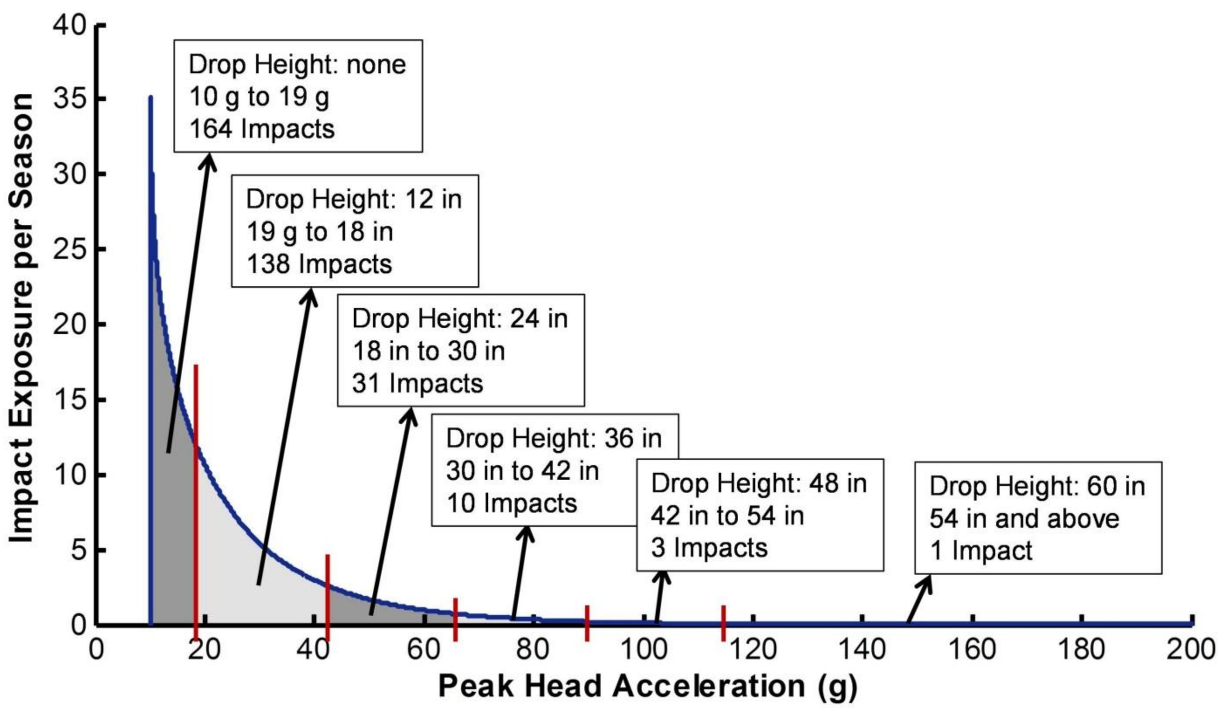

Figure 4: Number of impacts associated with each drop height for the front impact location, with integral bounds displayed for each drop height.

Table 8: Number of impacts per player per season associated with each impact location and drop height used in the STAR evaluation methodology, which is representative of the 90^{th} percentile player. Note: impacts less than 19 g are not considered in the STAR testing protocol.

Drop Height	Front	Rear	Side	Top
Impacts < 19 g	164	139	81	63
12 in	138	165	75	85
24 in	31	11	4	14
36 in	10	2	1	5
48 in	3	1	1	2
60 in	1	1	1	2
Total	347	319	163	171

Injury Risk

The 62,974 sub-concussive impacts had an average head acceleration of 26 g ± 20 g (median of 19 g), while concussive impacts had an average head acceleration of 105 g ± 27 g (median of 103 g). Figure 5 displays the probability density functions for sub-concussive and concussive impacts. Injury risk curves based on the collegiate and NFL injury rates are show in Figure 6 with corresponding parameter values for Equation 6 shown in Table 9.

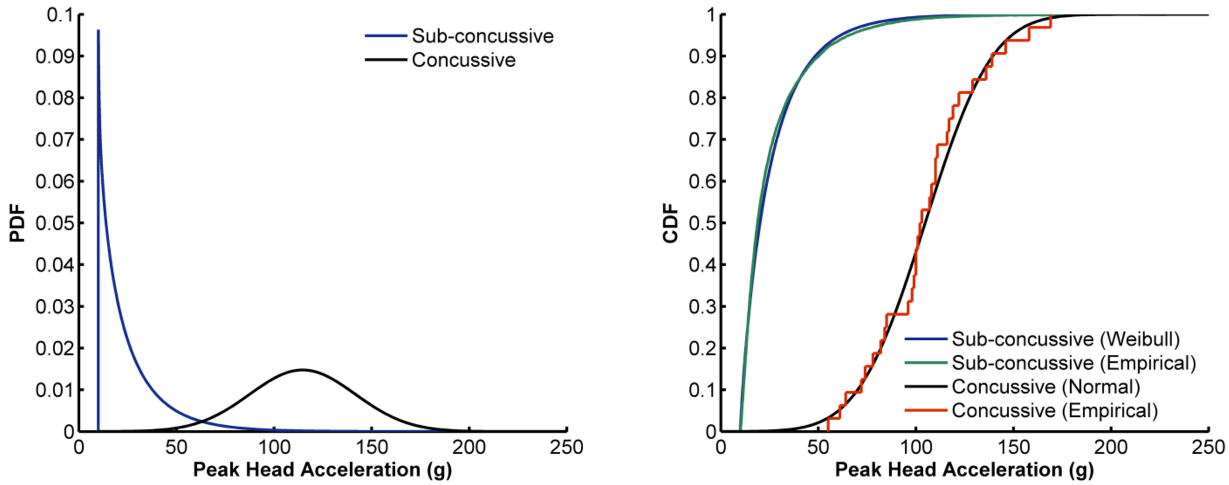

Figure 5: Weibull probability density function for all sub-concussive impacts and normal probability density function for all concussive impacts (left). Comparison of distribution fits for sub-concussive and concussive data to empirical data using cumulative density functions (right).

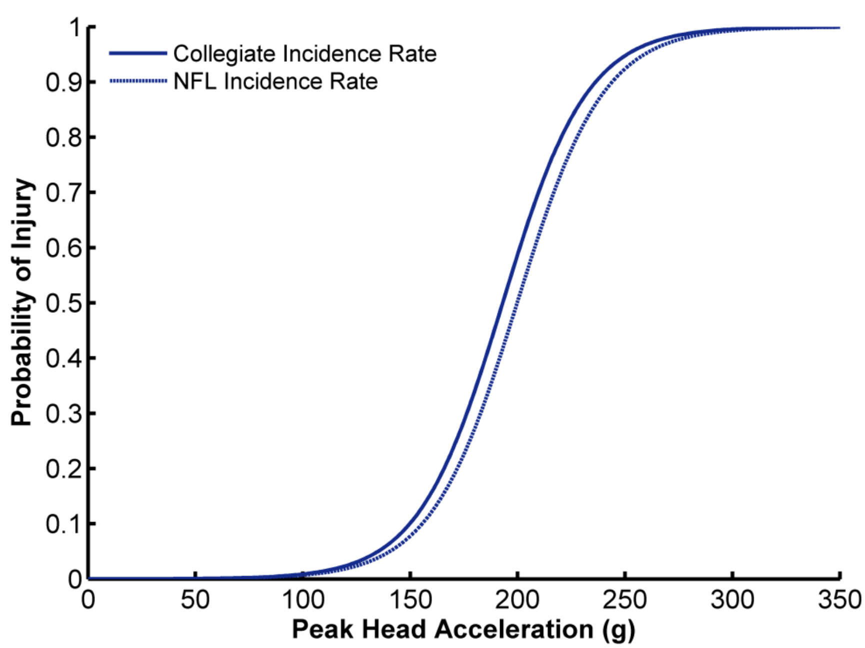

Figure 6: Injury risk curves based on the collegiate and NFL injury rates using resultant peak linear head acceleration.

Table 9: Logistic regression parameters for the college and NFL injury incidence rate-based risk curves (Equation 6).

	α	**β**
Collegiate	-9.805	0.051
Professional	-9.928	0.497

STAR Value Assessment

Table 10 displays the hypothetical head accelerations associated with a hypothetical helmet for each drop height and impact location configuration that result in a STAR value of 1.501. This value can be interpreted on average as: if one player wore the hypothetical helmet for one season of 1000 head impacts in practices and games, then the player would theoretically sustain approximately 1.5 concussions.

Table 10: STAR value assessment of a hypothetical helmet that resulted in the listed head accelerations for each drop height and impact location, and exposure per season is representative of 1000 impacts for the 90th percentile player.

Impact Location	**Drop Height**	**Peak G**	**Risk of Injury**	**Exposure per Season**	**Incidence per Season**
Front	Impacts < 19 g	-----	0.0000	164	0.00
Front	12 in	50	0.0007	138	0.10
Front	24 in	70	0.0019	31	0.06
Front	36 in	90	0.0053	10	0.04
Front	48 in	120	0.0239	3	0.07
Front	60 in	150	0.1011	1	0.10
Side	Impacts < 19 g	-----	0.0000	81	0.00
Side	12 in	55	0.0009	75	0.07
Side	24 in	90	0.0053	4	0.02
Side	36 in	105	0.0113	1	0.01
Side	48 in	125	0.0306	1	0.03
Side	60 in	150	0.1011	1	0.10
Rear	Impacts < 19 g	-----	0.0000	139	0.00
Rear	12 in	65	0.0015	165	0.25
Rear	24 in	90	0.0053	11	0.06
Rear	36 in	120	0.0239	2	0.05
Rear	48 in	135	0.0499	1	0.05
Rear	60 in	155	0.1267	1	0.13
Top	Impacts < 19 g	-----	0.0000	63	0.00
Top	12 in	50	0.0007	85	0.06
Top	24 in	80	0.0032	14	0.04
Top	36 in	100	0.0088	5	0.04
Top	48 in	120	0.0239	2	0.05
Top	60 in	145	0.0809	2	0.16
				STAR Value:	**1.501**

Discussion

It is important to note that no helmet will ever be perfect and that there will always be a risk of head injury in any sporting activity, regardless of the effectiveness of the protective equipment. There are many variables that affect the risk of concussion in sports, with player history and genetic differences likely domination the variation. Having noted this, the primary purpose of this study was to introduce the STAR equation and methodology, which can be used to evaluate football helmet performance. The STAR evaluation system combines true head impact exposure with injury risk to predict concussion incidence for a specific helmet throughout the course of a football season. Both the impact exposure and injury risk components of the STAR evaluation system are based on real-world data collected from human athletes; and as a result, they reflect the impacts that the average player actually experiences on the field. The methods used to incorporate head impact exposure and injury risk are parametric analyses that determine complete distributions, and are similar to previous work using a more limited dataset [29].

The STAR evaluation system is not intended to replace or criticize the role of NOCSAE in ensuring the safety of athletic equipment. In fact, our data illustrate that the NOCSAE-style drop test and head form configuration do an excellent job of replicating the type of impacts football players experience on the field. Figure 7 displays a comparison of typical NOCSAE head form impact response to football head impact acceleration corridors for the front and side impact locations. This figure illustrates that the NOCSAE head form accurately models head acceleration pulse shape and duration for impacts in football. For this reason, the NOCSAE-style drop tests are utilized as the core of the STAR system.

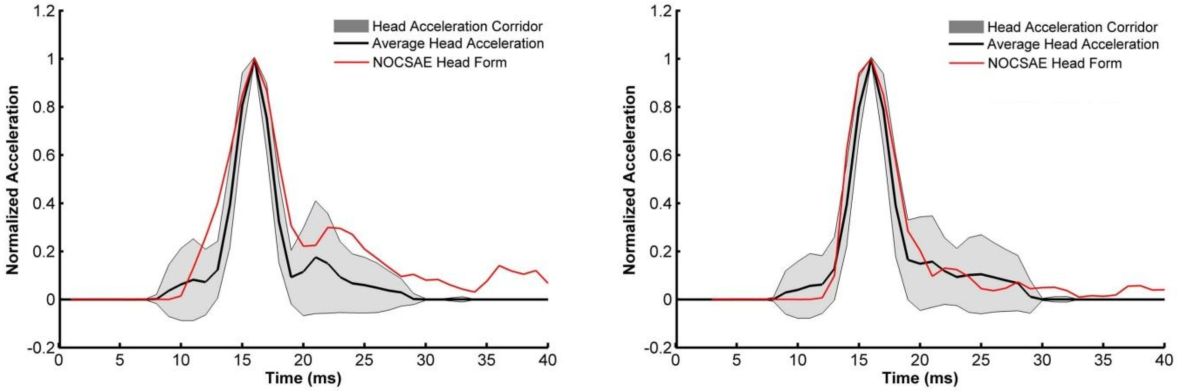

Figure 7: Comparison of typical normalized NOCSAE head form impact response to football head impact acceleration corridors for the front (left) and side (right) impact locations. All acceleration curves were normalized to their peak values.

For the STAR evaluation system, a single collegiate player participating in every game and practice throughout a football season was assumed to experience 1000 head impacts. The head impact exposure per player per season used in this study is consistent with other published studies. Guskiewicz et al. (2007) reported that the average collegiate football player experiences 950 impacts per season. In addition, Schnebel et al. (2007) found the average collegiate player to experience 1353 impacts per season [15, 17]. However, it should be noted that these numbers may not be directly applicable to high school athletes. Broglio et al. (2009) measured an average of 565 impacts per player in high school athletes [14]. Differences among these studies can be attributed to the number of athletic exposures each player was exposed to, as well as the relative playing time per event. In any case, the exposure number is a scalable constant and can be adjusted for any level (Table 11).

Table 11: Exposure per season for each drop configuration is presented for 50th, 75th, and 90th percentile impact exposures.

Impact Location	Drop Height	Exposure Per Season		
		50th Percentile	**75th Percentile**	**90th Percentile**
Front	Impacts < 19 g	71	110	164
Front	12 in	61	94	138
Front	24 in	14	21	31
Front	36 in	4	7	10
Front	48 in	1	2	3
Front	60 in	1	1	1
Side	Impacts < 19 g	33	53	81
Side	12 in	33	51	75
Side	24 in	2	3	4
Side	36 in	1	1	1
Side	48 in	1	1	1
Side	60 in	1	1	1
Rear	Impacts < 19 g	59	93	139
Rear	12 in	73	112	165
Rear	24 in	5	8	11
Rear	36 in	1	1	2
Rear	48 in	1	1	1
Rear	60 in	1	1	1
Top	Impacts < 19 g	28	44	63
Top	12 in	37	58	85
Top	24 in	6	9	14
Top	36 in	2	3	5
Top	48 in	1	1	2
Top	60 in	1	1	2
	Total Impacts:	**438**	**677**	**1000**

Furthermore, the head impact exposure per player per impact location is nearly identical to other published studies. Table 12 compares the impact location weightings used in the STAR evaluation system to that of another study that utilized the same data collection methodologies [30]. Moreover, these results are consistent with a study of 3 collegiate teams [22].

Table 12: Comparison of the distribution of impact locations based on data collected from instrumented collegiate football players.

Impact Location	**Percentage of All Impacts**	
	VT Data	**Mihalik et al. (2007)**
Front	34.7%	35.9%
Rear	31.9%	30.9%
Side	16.3%	14.4%
Top	17.1%	18.8%

In order to determine the proper weighting between sub-concussive and concussive head acceleration distributions, published injury incidence rates were considered. Most injury incidence rates are reported as the number of concussions per 1000 athletic exposures, where an athletic exposure is defined as one athlete participating in at least one play of one game or practice. Table 13 is a summary of published injury incidences rates for concussion in football [26, 31, 32]. Booher et al. (2003) reported higher injury incidence rates because that study included injuries that did not result in loss of playing time, where Guskiewicz et al. (2003) and Dick et al. (2007) only considered injuries associated with loss of playing time. In addition, Pellman et al. (2004) reported an injury incidence rate of 0.41 concussions per game in the National Football League [27]. To side with conservatism, only game data were considered and injury incidence rates of 5.56 concussions per 1000 athletic exposures and 0.41 concussions per game were analyzed in this study. Furthermore, an underreporting rate was applied to the injury incidence rates, as previous studies have suggested that underreporting is a prevalent issue with the diagnosis of concussion [28, 33-35].

Table 13: Published injury incidence rates expressed as concussions per 1000 athletic exposures in collegiate football [26, 31, 32].

Study	**Concussions / 1000 A-E**		
	Game	**Practice**	**Total**
Booher et al.	5.56	0.25	0.74
Guskiewicz et al.	3.81	0.47	0.81
Dick et al.	2.34	0.21	0.37

Injury risk curves were generated from a logistic regression analysis of properly weighted sub-concussive and concussive data distributions. Logistic risk curves were deemed appropriate due to the censored nature of the data, as the exact head accelerations associated with the onset of injury cannot be identified. While there are advocates of non-parametric approaches [36], such approaches are only representative of the data that are experimentally recorded [37]. Moreover, Kent and Funk (2004) have shown that parametric injury risk functions typically fall within the 95% confidence intervals of non-parametric injury risk functions for any given dataset [37]. They noted the largest discrepancy between the two approaches at the risk function tails (around 0% and 100% risk) where experimental data was limited. The real-world data that the injury risk curves are based on are unique in that the distributions are well-defined throughout the entire continuum of head accelerations experienced by football players. This likely provides a better representation of actual risk around the tails of the risk curves than most biomechanical experiments used to characterize risk. The parametric approach utilized in this study was chosen so that the risk curve could be representative of the entire population, rather than only the data collected in the experiment.

The analysis used in this study is unique because it weights sub-concussive and concussive acceleration distributions based on calculated injury incidence rates so that the injury risk curves are unbiased. Both the collegiate and NFL injury incidence rates produced very similar risk curves. Figure 8 compares the injury risk curves determined in this study to that of previously published studies. Pellman et al. (2003) created injury risk curves based on reconstructed concussive impacts from NFL football games [38]. Hybrid III crash test dummies were used to reconstruct 31 impacts based on game video. The risk curves from that experiment are limited because they are biased to concussive impacts. Their analysis was based on 25 concussive and 33 sub-concussive data points, and therefore overestimates risk of concussion. Interestingly, the average linear acceleration associated with concussion reported by Pellman et al. (2003) was 98 g ± 27 g, which is nearly identical to the average concussive linear acceleration collected from human volunteers (105 g ± 27 g). Although acceleration magnitude is accurately captured using the Hybrid III, it is likely that acceleration duration is less accurate due to the stiffness of the Hybrid III neck [39]. Funk et al. (2007) quantified risk using unique statistical analysis of data collected from human volunteers [29]. However, that risk curve was based on only 4 concussive data points, and was thought to underestimate risk. For head accelerations that are associated with less than 10% risk, the risk curve from Funk et al. (2007) closely matches the risk curves generated from this study. While the head accelerations associated with concussion are well defined in this study, injury risk associated with these head accelerations is low (< 10%) because impacts of these severities occur frequently in collegiate football without concussion. Since true head impact exposure was paired with the concussive impact distribution, these risk curves are thought to be accurate.

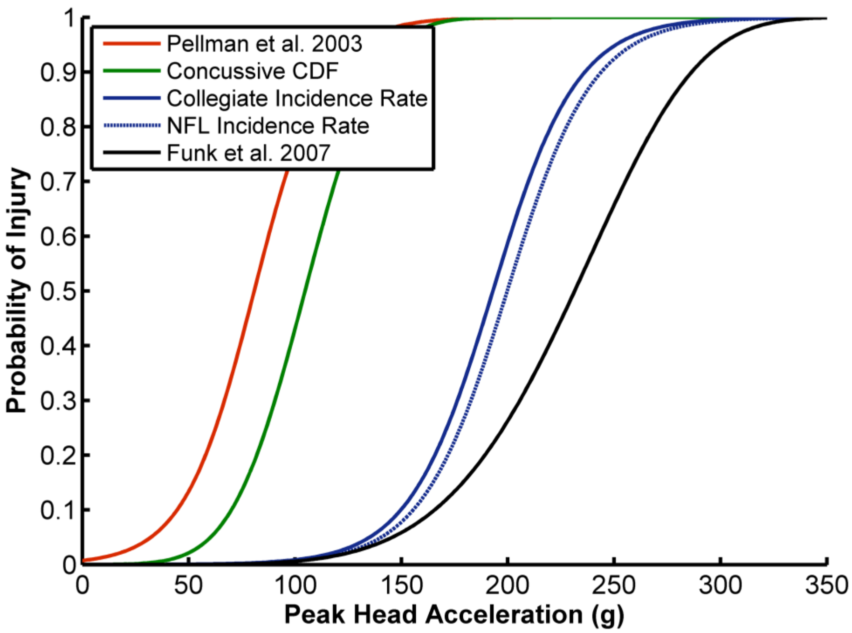

Figure 8: Comparison of injury risk curves generated in this study to that of previously published research. The cumulative density function of the normally distributed concussive dataset (Concussive CDF) is displayed to illustrate the head accelerations associated with only concussive impacts.

The injury risk curves utilized in the STAR equation currently only incorporate linear acceleration. Although rotational acceleration is not considered in this analysis, work is underway investigating human tolerance to rotational kinematics [16]. Furthermore, rotational acceleration cannot be applied to this testing methodology until protocols for testing rotational kinematics are developed. It has also been suggested that a composite injury metric consisting of many biomechanical parameters may have the best predictive capabilities for concussion [40].

The objective of these biomechanical parameters is to predict the tissue-level response of the brain to impact. While researchers are currently investigating the physiologic response of the brain tissue to mechanical insult, linking the tissue-level response of the brain to the kinematics of the skull remains a challenge [41, 42]. Finite element models will likely be an important factor in understanding this relationship [43]. Once a better understanding of these relationships is gathered, more complex methods can be used to predict brain injury in the future.

The STAR equation is determined through relating impact exposure to drop test performance for a specific helmet. A helmet that has lower accelerations associated with each drop test will therefore predict a lower incidence of concussions. Traditionally, it has been challenging to account for all the scenarios in sports that can cause injury in a laboratory testing [44-46]. This is a result of not knowing the true exposure of impact, as well as the vast amount of possible scenarios in sports. The STAR evaluation system generalizes all possible head impacts in football into 24 NOCSAE-style drop tests, which consists of 4 impact locations and 6 impact severities. This system can be a valuable tool in educating consumers on helmet performance, much like the current NCAP rating system aids consumers intending to purchase automobiles. In the future, the underlying principles of the STAR evaluation system can be applied to other levels of football, as well as other sports, once the exposure data are known.

Acknowledgments

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Chapter 3:

A Six Degree of Freedom Head Acceleration Measurement Device for Use in Football

Abstract

The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury. The goal of this study was to validate a six degree of freedom (6DOF) measurement device with 12 single-axis accelerometers that uses a novel algorithm to compute linear and angular head accelerations for each axis of the head. The 6DOF device can be integrated into existing football helmets and is capable of wireless data transmission. A football helmet equipped with the 6DOF device was fitted to a Hybrid III head instrumented with a 9 accelerometer array. The helmet was impacted using a pneumatic linear impactor. Hybrid III head accelerations were compared to that of the 6DOF device. For all impacts, peak Hybrid III head accelerations ranged from 24 g to 176 g and 1,506 rad/s^2 to 14,431 rad/s^2. Average errors for peak linear and angular head acceleration were 1% ± 18% and 3% ± 24%, respectively. The average RMS error of the temporal response for each impact was 12.5 g and 907 rad/s^2.

Introduction

Each year there are an estimated 1.6 million to 3.8 million sports related concussions in the United States (Langlois et al., 2006). About 300,000 of these concussions involve loss of consciousness, with football having the largest occurrence (Thurman et al., 1998). The high incidence of concussions in football provides a unique opportunity to collect biomechanical impact data from humans to characterize mild traumatic brain injury (MTBI). Competitive football has been used as an experimental environment for collecting human head acceleration data since the 1970's. Several studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games (Moon et al., 1971; Reid et al., 1974; Reid et al., 1971). While laying the groundwork for future research and providing a proof of concept, these studies measured only a single player and were limited in their ability to measure head acceleration accurately.

One study has quantified head accelerations by recreating concussive impacts sustained by football players. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on game video (Newman et al., 2000; Newman et al., 1999; Pellman et al., 2003). This study was biased towards concussive events and was limited to impacts that could be clearly identified on video. With such a labor intensive testing methodology, it was not realistic to recreate each and every impact that football players experienced. More recently, two studies quantified head accelerations during impact by instrumenting helmets worn by collegiate football players using a six accelerometer measurement device integrated into football helmets (Duma et al., 2005; Funk et al., 2007). These six accelerometer measurement devices utilized the same technology described in this paper but with fewer accelerometers and with the sensing axes of the accelerometers oriented differently with respect to the surface of the head. Resultant linear and peak angular head acceleration for every head impact instrumented players experienced was computed from the measured accelerations, producing a large and unbiased dataset of over 27,000 events including 4 cases of diagnosed concussion.

The 6 accelerometer measurement device used by Duma et al. (2005) is part of the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). The commercially available football HIT System measurement device consists of 6 non-orthogonally mounted single-axis accelerometers which are oriented normal to the head. Each time an impact occurs, data are transmitted wirelessly from the measurement device to the computer, which processes and displays data in real-time. The HIT System utilizes a novel algorithm for determining impact magnitude and location (Crisco et al., 2004). This algorithm is capable of determining the temporal response of the resultant linear acceleration of the head. In addition to resultant linear acceleration, the algorithm estimates peak x and y axis rotational accelerations based on an assumed pivot point in the neck. While this device can be used to collect valuable human head acceleration data from impacts, it cannot completely characterize the head kinematics of impacts because it computes only peak angular accelerations rather than the temporal angular acceleration response for each axis of the head. The temporal response of acceleration is desired with applications investigating the tissue-level response of the brain using computational models.

The goal of this study was to validate a six degree of freedom (6DOF) measurement device that completely characterizes the head kinematics resulting from impacts in football. This device records all head impacts sustained by football players during competitive play and computes the resulting linear and angular acceleration about each axis of the head. Such data are ideal for the validation of computational injury models and investigating the tissue-level response of the brain following impact.

Methods

The 6DOF measurement device was designed to be integrated into Riddell Revolution (Elyria, OH) football helmets (Figure 9). The measurement device utilizes 12 single-axis, 250-g iMEMS accelerometers (ADXL193, Analog Devices, Norwood, MA). The 12 accelerometers are grouped in orthogonally oriented pairs at 6 different locations within fabric padding. All accelerometers are orientated so that their sensing axes are tangential to the skull. The fabric pad system serves as an elastomeric spring and damper system that regulates the contact pressure and maintains accelerometer orientation relative to and in contact with the head during an impact event, and has a maximum deflection of 1 inch. When the helmet shifts position on the head due to impact, the fabric pad either compresses or expands to remain in contact with the player's head ensuring head acceleration, not helmet acceleration, is measured (Manoogian et al., 2006). The fabric pad is tethered to the helmet, so that in the event of helmet rotation relative to the head, the pad does not rotate with the helmet. Data acquisition is triggered when any accelerometer exceeds 10 g. Data are collected for 40 ms at 1000 Hz, of which 8 ms are pre-trigger and 32 ms are post-trigger. After each impact is recorded, the data are sent to a computer via a 903-927 MHz FHSS wireless transceiver. Following data transmission, each impact is processed for linear and angular acceleration about all three axes of the head center of gravity (CG), in addition to impact location.

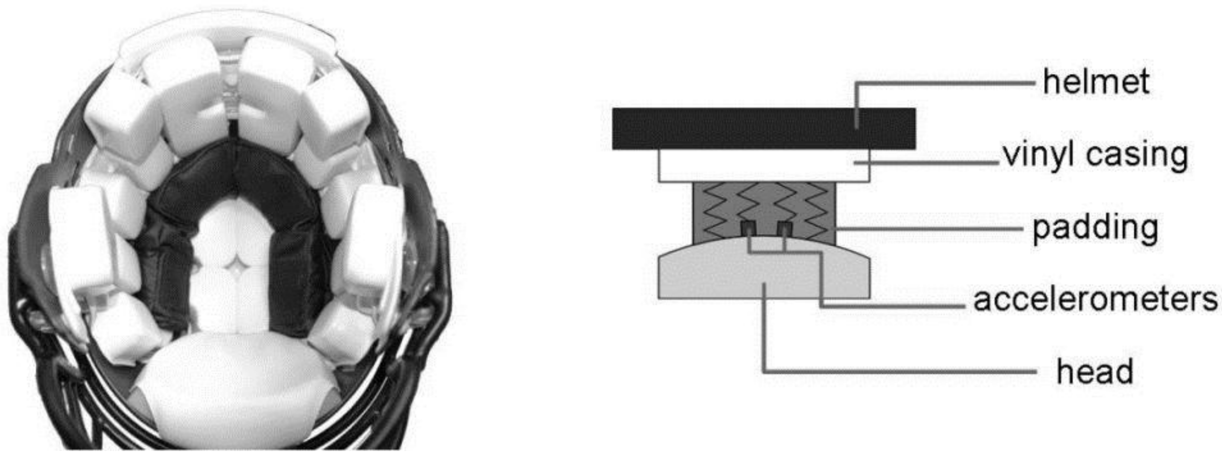

Figure 9: 6DOF measurement device installed in a Riddell Revolution helmet (left) and a schematic of the measurement device in the helmet (right). The black padding of the measurement device contrasts from the white padding of the helmet.

Linear and angular acceleration are determined by iteratively optimizing the equations of motion for the head during impact (Chu et al., 2006). Assuming rigid body dynamics, the acceleration of any point on the surface of the head relative to the head CG can be expressed by Equation 1; $\|a_i\|$ is the acceleration magnitude at each individual accelerometer, $\overrightarrow{r_{ai}}$ is the orientation of the sensing axis of each accelerometer, $\vec{H}$ is the head CG linear acceleration, $\vec{\alpha}$ is angular acceleration about the head CG, $\vec{r_i}$ is the accelerometer location relative to the head CG, and $\overrightarrow{\omega_i}$ is the angular velocity of the head.

$$\|a_i\| = \overrightarrow{r_{ai}} \bullet \vec{H} + \overrightarrow{r_{ai}} \bullet \left(\vec{\alpha} \times \vec{r_i}\right) + \overrightarrow{r_{ai}} \bullet \left(\overrightarrow{\omega_i} \times \left(\overrightarrow{\omega_i} \times \vec{r_i}\right)\right) \qquad (1)$$

By orienting the 6DOF measurement device's accelerometers tangentially to the skull, the centripetal acceleration term of Equation 1 becomes negligible, simplifying to Equation 2.

$$\|a_i\| = \overrightarrow{r_{ai}} \bullet \overrightarrow{H} + \overrightarrow{r_{ai}} \bullet \left(\vec{\alpha} \times \vec{r_i}\right) \qquad (2)$$

The linear and rotational acceleration vectors are estimated by iteratively solving for the vector components using a simulated annealing optimization algorithm. The cost function for the optimization algorithm is the minimization of the sum of square error between each measured acceleration and the estimated acceleration at each accelerometer location based on the estimated head linear and rotation vectors.

The location and orientation of each accelerometer relative to the head CG were determined by digitizing their positions (Microscribe G2, Amherst, VA) within a football helmet relative to the head CG of a Hybrid III 50th percentile male head. For field use, this digitization can be repeated for all available helmet sizes, providing a scalable solution for all players; however, it is still assumed that the location of the head CG of each player is represented by a Hybrid III head. The location of the Hybrid III head CG has been previously defined and is based on the data from cadaver heads (Foster et al., 1977; Hubbard & McLeod, 1974; Walker Jr et al., 1973).

A total of 114 impact tests were conducted to assess the accuracy of the 6DOF measurement device using an instrumented 50th percentile male Hybrid III head and neck assembly. The Hybrid III head was equipped with 9 accelerometers (7264-2000B, Endevco, San Juan Capistrano, CA) in a 3-2-2-2 orientation; which allowed linear and angular acceleration to be calculated (Padgaonkar et al., 1975). Hybrid III data were sampled at 10,000 Hz and filtered in accordance with SAE J211 using Channel Frequency Class (CFC) 1000. The head and neck were mounted on a custom linear slide table built to National Operating Committee on Standards for Athletic Equipment (NOCSAE) specification. The linear slide table was permitted 5 degrees of freedom, allowing repeatable adjustment of the head and neck orientation. All impacts were performed using a pneumatic linear impactor that was built to NOCSAE specification (NOCSAE, 2006). At the end of the 15 kg impactor arm, a disc of high-density vinyl nitrile foam attached to a hemispherical nylon shell were used to create an impacting surface that replicated the characteristics of a typical football helmet (Figure 10) (Newman et al., 2005).

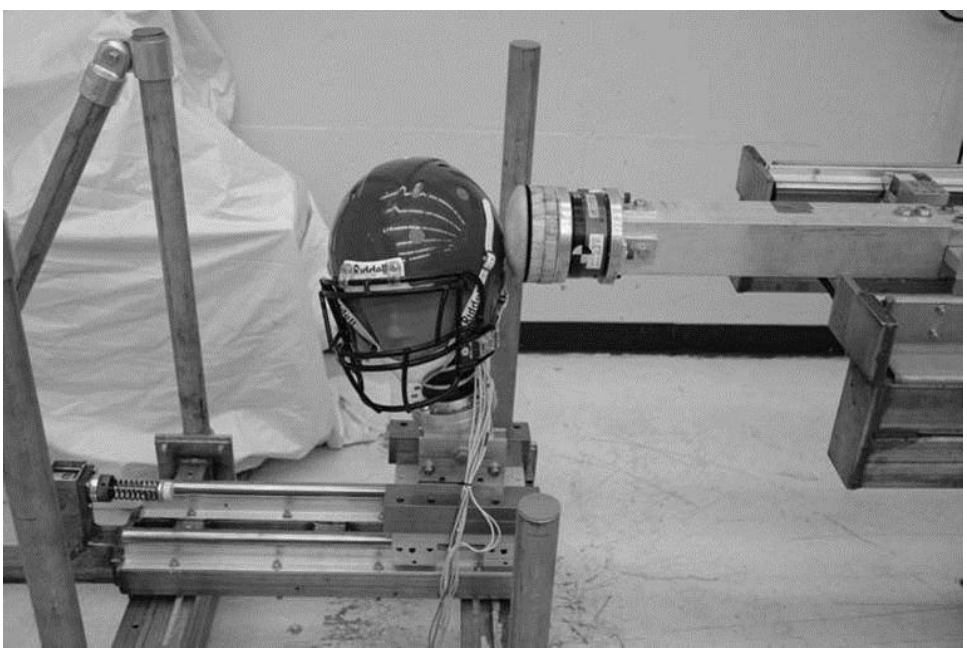

Figure 10: Pneumatic linear impactor and helmeted Hybrid III head mounted on the linear slide table.

A 6DOF measurement device was installed in a medium Riddell Revolution helmet, which was fitted on the Hybrid III head. A skull cap (89% nylon, 11% spandex) was put on the Hybrid III head to reduce the friction between the head-to-helmet interface to simulate a worst case scenario. The same helmet was used and repositioned for every test. The medium-sized jaw pads of the helmet were replaced with large jaw pads to better fit the narrow face of the Hybrid III head. A custom helmet positioning tool was used to ensure that the helmet's fit on the head was consistent and in accordance with NOCSAE standards. This plexiglass tool uses landmarks on the helmet and head to align the center of the facemask with the center of the nose and the top of the facemask with the brow of the Hybrid III face. A standard four-point chin strap was used to secure the helmet on the head. An air pump was used to inflate the padding of the helmet with accordance to the manufacturer's specification.

The helmeted Hybrid III head was struck with the pneumatic linear impactor with several combinations of impact energies and locations. The impact energies ranged from 67.5 J to 607.5 J and were chosen to replicate on-field impact data as determined by the NFL through impact reconstructions (Pellman et al., 2003). To account for the various ways a helmet can be struck, 5 impact locations were chosen based on NFL video analysis (Pellman et al., 2003b). Impact locations ranged from the front to the backside of the helmet (Figure 11).

Table 14 defines each impact location in terms of azimuth and elevation. Azimuth refers to the angle that the impact location makes with the sagittal plane of the Hybrid III head. Elevation refers to the angle that the impact location makes with the transverse plane of the Hybrid III head. A negative elevation is interpreted as having the Hybrid III head tilted away from the impactor. The number of tests in each configuration can be seen in Table 15.

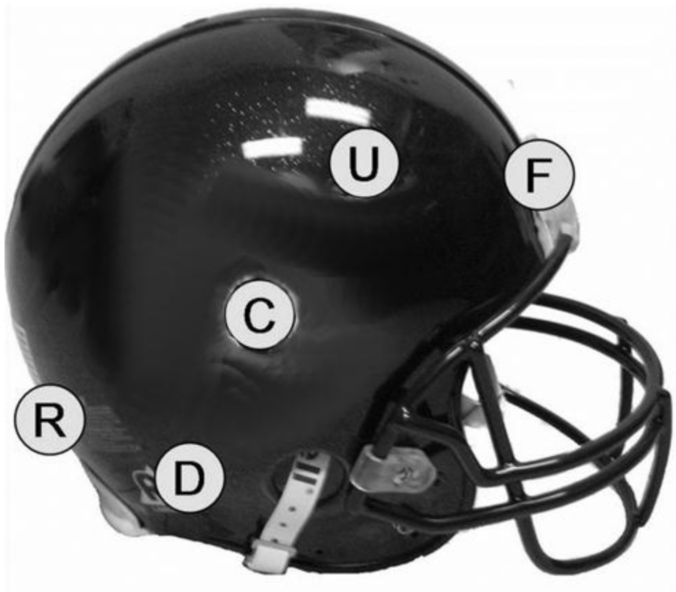

Figure 11: 5 locations on the helmet were impacted at a range of impact energies.

Table 14: Definitions of the 5 impact locations tested.

Location	Elevation	Azimuth
C	-11°	112.5°
D	-11°	157.5°
F	25°	0°
R	0°	180°
U	20°	67.5°

Table 15: Impact testing test matrix.

		Location					
		C	**F**	**R**	**D**	**U**	**Total**
Energy (J)	**67.5**	4	0	0	4	0	**8**
	187.5	4	4	4	4	0	**16**
	270.0	4	4	4	8	4	**24**
	367.5	4	4	4	4	0	**16**
	480.0	10	4	4	12	4	**34**
	607.5	8	4	0	4	0	**16**
	Total	**34**	**20**	**16**	**36**	**8**	**114**

Regression analyses were applied to the data to determine how strongly the 6DOF measurement device correlated with the Hybrid III head form. Subsequently, a power-law regression using a least squares fitting technique was performed to map the 6DOF accelerations to that of the Hybrid III in order to get the highest correlation coefficient. These mapping functions between Hybrid III and 6DOF were desired because the input to many finite element head models used in the injury biomechanics field is kinematic data from the Hybrid III head (Takhounts et al., 2008).

Acceleration error is presented in two ways: percent error for linear and angular peak resultant accelerations, and root mean square (RMS) error for the temporal response of linear and angular resultant acceleration. In both sets of error calculations, the Hybrid III head acceleration is considered the true value of acceleration. The Hybrid III head represents the 'gold standard' for modeling dynamic cranial response of the human head, and is commonly used in automotive safety testing and sports injury biomechanics testing. Equation 3 was used to calculate relative error; where *RE* is relative error, *6DOF* is the 6DOF measurement device peak acceleration predicted by the mapping functions, and *HIII* is the Hybrid III head peak acceleration. Relative error of the linear and angular peak accelerations are presented as an average percent error ± 1 standard deviation. RMS error was calculated using Equation 4; where n is the number of discrete data points, i represents individual discrete data points, *6DOF* is the 6DOF measurement device acceleration predicted by the mapping functions, and *HIII* is the Hybrid III head acceleration. For the RMS calculation, Hybrid III data was downsampled to 1000 Hz. RMS error was calculated for the temporal response of linear and angular resultant acceleration over the acceleration pulse of interest, which equated to the first 25 ms of each impact. The remaining 15 ms of data for each impact have no relevance to the impact. This was consistent for all impacts because of the controlled impact conditions.

$$RE = \frac{6DOF - HIII}{HIII} \quad (3)$$

$$RMS = \sqrt{\frac{\sum_{i}^{n}\left(6DOF_i - HIII_i\right)^2}{n}} \quad (4)$$

Results

The linear and angular accelerations computed from the 3-2-2-2 array in the Hybrid III head were compared to the accelerations computed from the 6DOF measurement device. The power regression revealed strong correlations between Hybrid III and 6DOF measurement device linear (R^2 = 0.88) and angular (R^2 = 0.85) peak resultant accelerations (Figure 12). Equation 5 displays Hybrid III resultant linear acceleration as a function of 6DOF resultant linear acceleration; where A_{6DOF} is 6DOF resultant linear acceleration and A_{Hybrid} is Hybrid III resultant linear acceleration. Equation 6 displays Hybrid III resultant angular acceleration as a function of 6DOF resultant angular acceleration; where α_{6DOF} is 6DOF resultant angular acceleration and α_{Hybrid} is Hybrid III resultant angular acceleration. The average relative error for peak resultant linear acceleration was 1% ± 18%. The average relative error for peak resultant angular acceleration was 3% ± 24%.

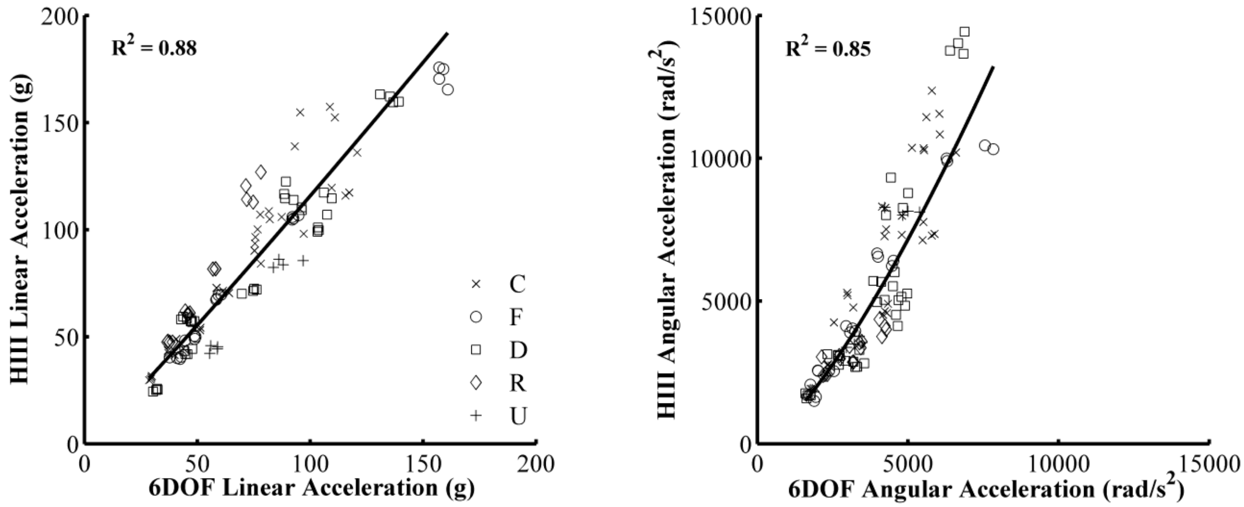

Figure 12: Non-linear relationships between 6DOF measurement device and Hybrid III head peak resultant accelerations.

Linear: $$A_{Hybrid} = 0.8695 * A_{6DOF}^{\ 1.0622} \quad (5)$$

Angular: $$\alpha_{Hybrid} = 0.0638 * \alpha_{6DOF}^{\ 1.3652} \quad (6)$$

Figure 13 compares the temporal acceleration responses for the Hybrid III and 6DOF measurement device for an impact energy of 480 J at location D. For all impacts, peak Hybrid III head accelerations ranged from 24 g to 176 g and 1,506 rad/s^2 to 14,431 rad/s^2. The average RMS errors for the resultant temporal response of all impacts were 12.5 ± 8.32 g and 907 ± 685 rad/s^2. RMS error for the temporal response was not greater for any specific axis of the head for either linear or angular acceleration. Furthermore, RMS error was not significantly influenced by impact location. RMS error increased linearly with impact energy, which is expected.

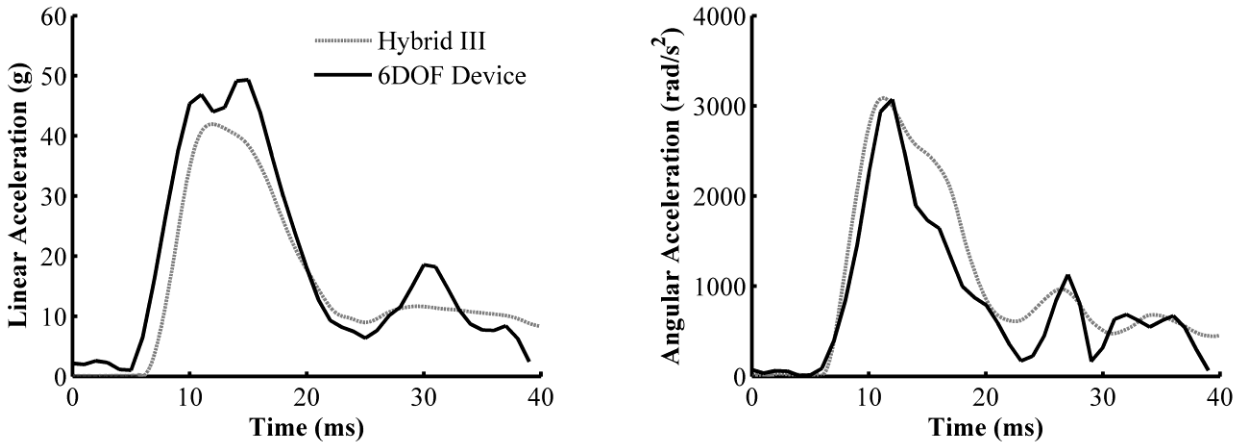

Figure 13: Typical temporal response of the Hybrid III and 6DOF measurement device resultant accelerations.

In addition to impact location, direction of force was also varied. Figure 14 displays the relationships between linear and angular acceleration for each impact location. These relationships describe the eccentricity of each impact location, as the slope of each line is indicative of the distance that the force vector was from the head CG. A steeper slope indicates the force was directed further away from the CG.

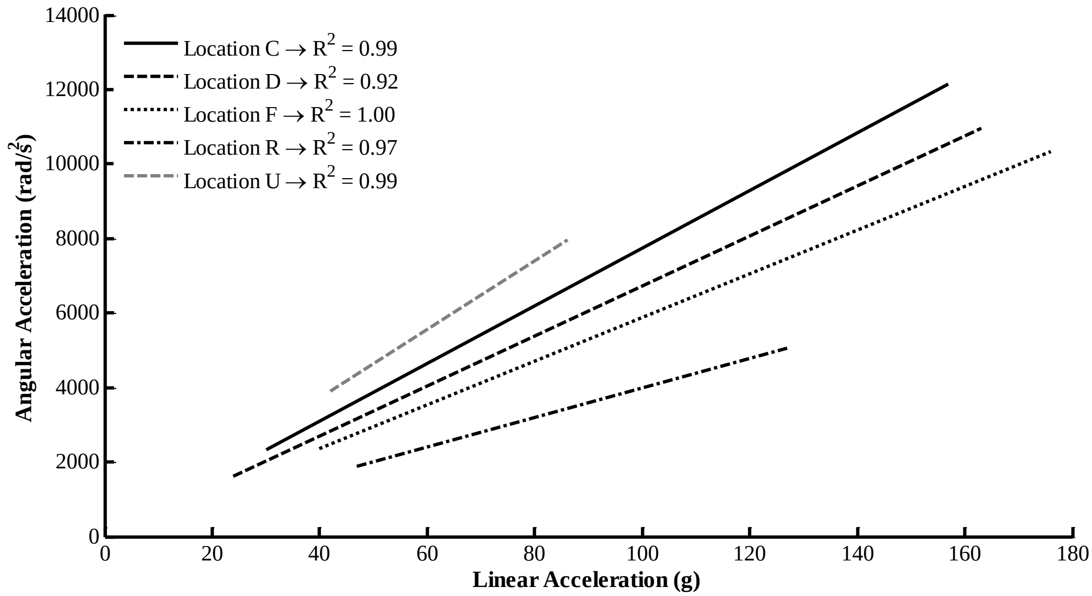

Figure 14: The relationships between linear and angular for the Hybrid III head for each impact location, as determined by linear regressions. These varying relationships are indicative of how principle direction of force varied with each impact.

Discussion

A non-linear relationship between 6DOF measurement device acceleration and Hybrid III head acceleration was observed with angular acceleration. It is thought that the reason for this non-linear trend is relative movement between the Hybrid III head and the accelerometers of the 6DOF measurement device; resulting in the 6DOF measurement device to under-predict angular acceleration at high severities. Although the helmet was properly fitted to the Hybrid III head, the helmet could move relative to the head due to the low friction interface between the Hybrid III head and nylon skull cap. This type of helmet movement relative to the head could be seen in on-field impacts in the game of football. The linear acceleration trend was practically linear, but a power regression was performed to transform the linear acceleration data in a method consistent with that used for angular acceleration. Furthermore, the power regression analysis resulted in high R^2 values between 6DOF measurement device and Hybrid III head for both linear and angular acceleration. The resulting equations allow the 6DOF measurement device to be mapped to Hybrid III measurements, which is preferred for computational modeling applications that have traditionally utilized Hybrid III data. 6DOF measurement device data undergoing the power regression transformation are representative of the accelerations of the Hybrid III head with the stated error in this study.

Similar testing has been used to validate instrumented boxing head gear (IBH) for use during boxing competition (Beckwith et al., 2007). In this study, IBH acceleration was shown to correlate strongly with Hybrid III head acceleration for linear (R^2 = 0.91) and angular (R^2 = 0.91) acceleration. The 6DOF measurement device utilizes a similar algorithm (Chu et al., 2006) and sensor orientation as the IBH. Average RMS errors for the IBH were 5.9 ± 2.6 g and 595 ± 405 rad/s^2, compared to 12.5 ± 8.32 g and 907 ± 685 rad/s^2 for the 6DOF measurement device. Difference in accuracy between the two systems can be attributed to several factors. The boxing head gear had a tighter fit than a football helmet on the Hybrid III head. In addition, the 6DOF measurement device was validated over a larger range of impact energies. The maximum accelerations in the IBH validation testing were 77.3 g and 6,433 rad/s^2, compared to 176 g and 14,431 rad/s^2 in the 6DOF measurement device validation testing. When comparing similar acceleration ranges, average RMS error for the 6DOF measurement device is similar to that of Beckwith et al. 2007. For example, average 6DOF RMS errors are only 3.7 ± 4.3 g and 252 ± 267 rad/s^2 for impact with peak linear acceleration less than 80 g.

Frontal facemask impacts were not included in this analysis due to limitations of the linear impactor. The linear impactor was designed to mimic reconstructed head-to-head football collisions; however, the existing impactor surface was optimized for helmet shell contact and initially used for testing helmets without facemasks (Newman et al., 2005). Preliminary trials revealed an unrealistic interaction between the impactor surface and the facemask. Severe facemask bending atypical of on-field occurrence would occur even at low impact energies. In addition, at impact energies exceeding 480 J, the impactor surface would penetrate the facemask and contact the surface of the Hybrid III head. Substantial stroking was also observed on the chin strap following each of these impacts. This is where the chin strap slides through its grips, allowing the helmet's position relative to the head to change. For these reasons, frontal facemask testing was not completed. In addition, inertially induced head accelerations were not examined.

Possible sources of observed error include the helmet changing position relative to the head during an impact and non-ideal orientation of the accelerometers with respect to the head. While it was not possible to measure changes in accelerometer location and orientation in this experiment, the error levels in the 6DOF accelerations are similar to that of other measurement devices and techniques. The NFL reconstruction was reported to have error as high as 17% for peak linear acceleration and 25% for peak angular acceleration (Newman et al., 2005). In addition, chest bands, which are used to measure chest deflection, can have error as high as 10% (Rath et al., 2005). Relative error for the 6DOF measurement device is 1% ± 18% for peak linear acceleration and 3% ± 24% for peak angular acceleration. Considering the vast amounts of data that can be collected with the 6DOF measurement device on human volunteers and the error levels of other biomechanical experiments, the error inherent in the 6DOF measurement device can be considered acceptable. Furthermore, this study has provided quantification of uncertainty error that can be used for the design of experiments. Although some error is present, the data provides a distribution that can be utilized to help optimize design of experiments and bootstrapping techniques.

Measuring head accelerations experienced by human volunteers at potentially injurious severities has been traditionally challenging. Football provides a unique opportunity to collect such data using HIT System technology. Additionally, the 6DOF measurement device used in this study provides the temporal response of linear and angular acceleration about each axis of the head due to impact, which improves applicability for validation of computational injury models, as most models are time sensitive. This technology provides the opportunity to collect a large and unbiased dataset, since every head impact that an instrumented football player experiences would be recorded including both non-injurious and concussive impacts, which can be applied to a wide array of research studies that will ultimately lead to a better understanding of the mechanisms of concussion.

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Chapter 4:

Rotational Kinematics Associated with Concussion in Humans

Abstract

Given that recent research has correlated long-term neurodegenerative effects with a history of sports-related concussion, these injuries are of increasing concern. Kinematic parameters of the head are used to predict brain injury because they are thought to be indicative of the inertial response of the brain, but no injury criterion has been widely accepted for rotational kinematics. The objective of this study was to characterize the rotational kinematics of the head associated with sub-concussive and concussive impacts using a large head acceleration dataset collected from human volunteers. Between 2007 and 2009, the helmets of 335 collegiate football players were instrumented with accelerometer arrays that measured head acceleration for every head impact each player experienced. This biomechanical data was paired with clinical data identifying concussive impacts. The rotational kinematics of 300,977 sub-concussive and 57 concussive head impacts were quantified and analyzed in this study. The average sub-concussive impact had a rotational acceleration of 1230 rad/s^2 and a rotational velocity of 5.5 rad/s, while the average concussive impact had a rotational acceleration of 5022 rad/s^2 and a rotational velocity of 22.3 rad/s. The distributions of sub-concussive and concussive impacts were characterized by peak rotational acceleration. Each distribution was weighted to reflect current injury incidence rates. Injury risk curves were developed using an unbiased logistic regression analysis. Nominal injury values related to rotational accelerations and rotational velocities are reported. An increased understanding of the biomechanics associated with

concussion can provide insight to the injury mechanisms, human tolerance to mechanical stimuli, and injury prevention techniques.

Introduction

There are an estimated 1.6 to 3.8 million sports-related concussions occurring annually in the United States [1]. While sports-related concussion was once considered to only result in immediate neurocognitive impairment and symptoms that are transient in nature, recent research has correlated long-term neurodegenerative effects with a history of sports-related concussion [2-4]. Increased awareness and current media attention have contributed to concussions becoming a primary health concern. This paper focuses on the biomechanics of the head associated with sports-related concussion, which is a milder form of brain injury than what has been traditionally investigated. An increased understanding of the biomechanics associated with concussion can provide insight to the injury mechanisms, human tolerance to mechanical stimuli, and injury prevention techniques.

Historically, the majority of brain injury biomechanics research has defined concussion as a severe and life-threatening injury. Concussive brain injury is unique in that the injury has a graded response that can vary from minor confusion to death [5]. The sports-related concussions investigated in this study are considered mild in severity. Annually in the United States, there are only an estimated 300,000 concussion concussions that result in loss of consciousness, with the remaining injuries resulting in less severe symptoms [1, 6]. However, the varying grades of concussion are likely a scaled result of the varying mechanical stimuli input to the head [5]. Previous experiments involved investigating brain injury mechanisms, and how these mechanisms relate to the kinematics of the head. Kinematic parameters of the head are related to brain injury because they are thought to be indicative of the inertial response of the brain. Ideally, the head kinematics of a human surrogate could be measured in a safety testing scenario and used to predict the tissue level response of the brain in an effort to evaluate injury potential. With this goal in mind, many researchers have studied the relationship between head kinematics and brain injury. Most experiments have investigated linear or rotational kinematics independently, as these inputs have long been thought to result in different injury mechanisms [7]. Explanations of these theories have been previously documented in great detail [8].

The Wayne State Tolerance Curve (WTSC) was developed from a series of tests on dogs and cadavers and related linear acceleration and duration of acceleration to injury tolerance [9]. Injury metric functions such as severity index (SI) and head injury criterion (HIC) were subsequently developed from analyses of the WTSC [10, 11]. These injury metrics were primarily developed to predict skull fracture, although they were thought to correlate with severe brain injury. Notably, only linear acceleration is considered in these injury metrics, and all current safety standards for head injury are based on these works. However, rotational acceleration is believed by many to be a primary mechanism for brain injury [12]. Unlike linear acceleration, there is currently no accepted injury criterion for rotational acceleration. Additionally, previous research investigating rotational kinematics has focused on animal models (primate or rat), in which pure rotational was applied to the head [5, 13-18]. These experiments, including those evaluating linear and rotational acceleration, utilize little data from humans. Cadavers have no physiologic response, and animal data cannot be directly applied to humans.

Optimally, these experiments would utilize data derived from humans. However, recording potentially injurious data from humans has been challenging.

Of all sports, football has the greatest incidence of concussion due to its large number of participants and its violent nature [19]. The high incidence of concussion in football has provided scientists with a unique opportunity to collect biomechanical data that can characterize this injury. With this in mind, a series of studies reconstructed concussive impacts experienced by players in the National Football League (NFL) using Hybrid III anthropometric test devices (ATD) [20-22]. Using game film, 31 impacts were reconstructed and the resulting head kinematics were analyzed. From these analyses, separate injury risk curves for concussion were developed for linear and rotational kinematics. The limitations of this study were that data were collected from ATDs rather than humans, and that the NFL dataset is biased towards concussive impacts.

More recently, researchers have instrumented and observed a population that is at high risk for concussion (football players) to collect head impact data at potentially injurious severities from human volunteers in a natural and ethically sound manner [23]. In these studies, the helmets of football players were instrumented with commercially available accelerometer arrays, known as the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). Each time an instrumented player's helmet was impacted, head acceleration data was recorded and stored. This method of data collection allows biomechanical data measured in humans to be paired with clinical data assessing injury. These studies have provided great insight to the head kinematics associated with head impacts in football, but have largely been descriptive studies with small concussive sample sizes that made it difficult to draw conclusions about injury [24-30].

The objective of this study was to characterize the rotational kinematics of the head associated with sub-concussive and concussive impacts using a large head acceleration dataset collected from human volunteers. While descriptive statistics for sub-concussive and concussive impacts are provided, injury risk is also assessed through an unbiased risk analysis. Data presented in this study provides valuable insight to the concussive tolerance of humans to rotational acceleration.

Methods

<u>Data Collection</u>

Between 2007 and 2009, the helmets of 335 collegiate football players were instrumented with accelerometer arrays that measured head acceleration for every head impact each player experienced. Players were recruited from three Division 1 National College Athletic Association (NCAA) football teams (Brown, Dartmouth, and Virginia Tech), and all participants gave informed consent approved by each school's Institutional Review Board (IRB). Two accelerometer arrays were utilized in this study: the commercially available HIT System and a custom 6 degree of freedom (6DOF) measurement device.

A total of 314 players were instrumented with the HIT System for every game and practice they participated in while included in this study. The HIT System consists of 6 accelerometers that are spring-mounted so that they remain in contact with the head at all times, ensuring that head acceleration is measured rather than helmet acceleration [31]. When an accelerometer exceeded a specified threshold (14.4 g) during play, data acquisition was automatically triggered and data were collected for 40 ms (including 8 ms of pre-trigger data) at 1000 Hz. Once data collection was complete, data were wirelessly transmitted to a computer on the sideline. Resultant linear head acceleration at the center of gravity (CG) of the head was computed using a novel algorithm [32]. The HIT System has been well-validated [26, 32], and has been widely adopted by other researchers studying concussion in athletes [23]. This study utilized data collection protocols that are described in greater detail by previous studies [26, 33].

In addition, the helmets of 21 Virginia Tech football players were instrumented with a custom 6DOF head acceleration measurement device [29]. This measurement device consists of 12 accelerometers that are spring-mounted so that they remain in contact with the head, and collects data in the same manner as the HIT System. Any time an accelerometer exceeded 14.4 g during play, data acquisition was automatically triggered and data were collected for 40 ms (including 8 ms of pre-trigger data) at 1000 Hz. Data were also wirelessly transmitted to the HIT System computer on the sideline after data collection was complete. Linear and rotational acceleration about each axis of the head is computed using a novel algorithm [34, 35]. While an overview is presented here, a detailed technical comparison of the HIT System and 6DOF measurement device has previously been reported [35].

Measured impacts were categorized as either being sub-concussive or concussive. For the purposes of this study, concussive impacts were defined as any impact that caused a disturbance in brain function. Symptoms leading to the clinical diagnosis of concussion included headache, nausea, vomiting, dizziness/balance problems, fatigue, trouble sleeping, drowsiness, sensitivity to light or noise, blurred vision, difficulty remembering, and/or difficulty concentrating [36]. Concussive impacts were diagnosed by each team's trained medical staff. All other head impacts recorded were considered sub-concussive. To increase the sample size of the concussive dataset, the concussive impacts measured in this study were compiled with concussive data collected from published studies that utilized identical data collection methods [24, 27].

Data Analyses

Traditionally with the HIT System, peak rotational acceleration has been estimated from the linear acceleration vector and an assumed point of rotation 10 cm inferior to the head CG. In this study, peak rotational acceleration was estimated using Equation 1; where: α is peak rotational acceleration, m is the mass of the head, ax is peak linear acceleration along the anterior-posterior axis of the head, ay is peak linear acceleration along the medial-lateral axis of the head, I is the moment of inertia of the head, and d is the perpendicular distance from the head CG to the force vector. The mass and moment of inertia of the head were assumed to be constant for all impacts and based on cadaver data. The mass of the head was set to equal 4.375 kg and the moment of inertia was set equal to 233.2 kg-cm^2 [37]. The perpendicular distance from the head CG to the force vector was determined based on an analysis of recorded 6DOF acceleration data and laboratory validation experiments. A least squares technique was used to solve for d, which was determined to be 3.45 cm. Peak rotational accelerations were determined for all recorded HIT System impacts using Equation 1.

$$\alpha = \frac{m\sqrt{ax^2+ay^2}}{I}d \qquad \text{Equation 1}$$

Each recorded head impact was categorized into one of four general impact locations: front, rear, side (left and right), and top [33]. Impacts to the left and right locations were assumed symmetric, and thought to invoke coronal plane rotation. Impacts to the front and back of the helmet were grouped to together and thought to invoke sagittal plane rotation. Impacts to the top of the helmet have been shown to primarily cause linear events, as the head loaded is in line with the cervical spine. For this reason, impacts to the top of the helmet were removed from this analysis and reported separately.

The data collected in this study were used to define the overall distribution of sub-concussive and concussive impacts with relation to rotational acceleration. Sub-concussive impacts recorded using the HIT System and 6DOF measurement device were fit to Weibull distributions. The Weibull probability density function (pdf) takes the form of Equation 2, while the Weibull cumulative density function (cdf) takes the form of Equation 3. For these equations: x is the peak resultant rotational acceleration, α is the shape parameter, and β is the scale parameter. Weibull distribution parameters were estimated using a maximum likelihood technique.

$$w_{pdf} = \frac{\alpha(x)^{\alpha-1}}{\beta^{\alpha}} e^{-\left(\frac{x}{\beta}\right)^{\alpha}} \qquad \text{Equation 2}$$

$$w_{cdf} = 1 - e^{-\left(\frac{x}{\beta}\right)^{\alpha}} \qquad \text{Equation 3}$$

Concussive impacts collected with the HIT System were fit to a Rician distribution, which is a form of a normal distribution that is non-negative. The Rician pdf takes the form of Equation 4, while the Rician cdf takes the form of Equation 5. For these equations: x is the peak resultant rotational acceleration, v is the location parameter, σ is the scale parameter, I_o is the modified Bessel function of the first kind, and Q_1 is the Marcum Q-function. Rician distribution parameters were estimated using a maximum likelihood technique.

$$r_{pdf} = \frac{x}{\sigma^2} e^{\left(\frac{-(x^2+v^2)}{2\sigma^2}\right)} I_0\left(\frac{xv}{\sigma^2}\right) \qquad \text{Equation 4}$$

$$r_{cdf} = 1 - Q_1\left(\frac{v}{\sigma}, \frac{x}{\sigma}\right) \qquad \text{Equation 5}$$

Then, the relationship between resultant rotational acceleration and resultant rotational velocity was determined. For this sub-analysis, only impacts with peak linear accelerations greater than 40 g in the 6DOF dataset were considered. To determine resultant rotational velocity, rotational acceleration about each individual axis of the head was numerically integrated with respect to time throughout the entire acceleration trace. Resultant rotational velocity was then calculated. Each impact was visually inspected so that the rotational acceleration pulse of interest could be examined and peak values identified. Once peak rotational acceleration and peak change in rotational velocity were determined for each impact, a linear regression analysis was performed using a least squares technique. The regression model was constrained so that a rotational acceleration of 0 rad/s^2 resulted in a rotational velocity of 0 rad/s. Equation 6 displays the regression model, where: ω is resultant rotational velocity, $\dot{\omega}$ is resultant rotational acceleration, and m is the inverse slope parameter. Equation 6 was used to estimate resultant rotational velocities associated with the peak rotational accelerations in the HIT System dataset.

$$\omega = \frac{\dot{\omega}}{m} \qquad \text{Equation 6}$$

An injury risk function for resultant rotational acceleration was developed. To do this, published injury incidence rates for game participation were used to determine the proper weighting between sub-concussive and concussive head acceleration distributions. For collegiate athletes, there are 5.56 concussions per 1000 athletic exposures, where an athletic exposure is defined as one athlete participating in at least one play of one game or practice [38]. To relate the number of concussions to the number of sub-concussive impacts, it was assumed that the median player experiences 16.3 impacts per game [33]. For collegiate athletes, 5.56 concussions per 1000 games played with 16.3 impacts per game per player can be expressed as an injury incidence rate of 0.341 concussions per 1000 impacts. It is important to note that current research suggests that as many as 53% of concussions go unreported [39]. This underreporting rate was applied to the calculated injury incidence rate, resulting in 0.726 concussions per 1000 impacts for collegiate athletes.

Next, estimated injury incidence rates were used to combine the sub-concussive and concussive head acceleration distributions in order to have the proper sub-concussive to concussive impact ratio. The weighting of each distribution based on injury incidence rates allows for an unbiased risk analysis. A logistic regression analysis based on the weighted sub-concussive and concussive head acceleration distributions was used to express risk as a function of rotational head acceleration. Equation 7 displays the risk function, where: α and β are regression coefficients. The regression coefficients were determined using a generalized linear model technique.

$$risk = \frac{1}{1+e^{-(\alpha+\beta x)}} \qquad \text{Equation 7}$$

Results

A total of 300,977 head impacts were recorded and analyzed in this study. Of these impacts, 286,636 head impacts were recorded using the HIT System and 14,341 head impacts were recorded using the 6DOF measurement device. A total of 57 concussions were compiled for this analysis. There were 193,465 sub-concussive (67.5%) and 33 concussive (57.9%) impacts that were primarily composed of sagittal plane rotation (impacts to the front and back of the helmet). There were 49,645 sub-concussive (17.3%) and 7 concussive (12.3%) impacts that were primarily composed of coronal plane rotation (impacts to the sides of the helmet). There were 43,526 (15.2%) sub-concussive and 17 (29.8%) concussive impacts to the top of the helmet recorded with the HIT System, which were separated from this analysis.

The sub-concussive impact distribution recorded with the 6DOF measurement device was right skewed with a 25th percentile rotational acceleration of 531 rad/s^2, median rotational acceleration of 872 rad/s^2, and 75th percentile rotational acceleration of 1447 rad/s^2 (average rotational acceleration of 1158 ± 972 rad/s^2). The sub-concussive impact distribution recorded with the HIT System was right skewed with a 25th percentile rotational acceleration of 682 rad/s^2, median rotational acceleration of 981 rad/s^2, and 75th percentile rotational acceleration of 1506 rad/s^2 (average rotational acceleration of 1230 ± 915 rad/s^2). Concussive impacts recorded with the HIT System were normally distributed with a 25th percentile rotational acceleration of 4026 rad/s^2, median rotational acceleration of 4948 rad/s^2, and 75th percentile rotational acceleration of

6209 rad/s^2 (average rotational acceleration of 5022 ± 1791 rad/s^2). No concussive impacts were recorded with the 6DOF measurement device. Figure 15 displays the probability density functions and cumulative density functions for all sub-concussive and concussive impacts with relation to rotational acceleration. Figure 16 displays that the empirical cumulative density functions closely match the fitted cumulative distributions for each dataset. Table 16 displays the parameter estimates for each distribution fit (Equations 2-5).

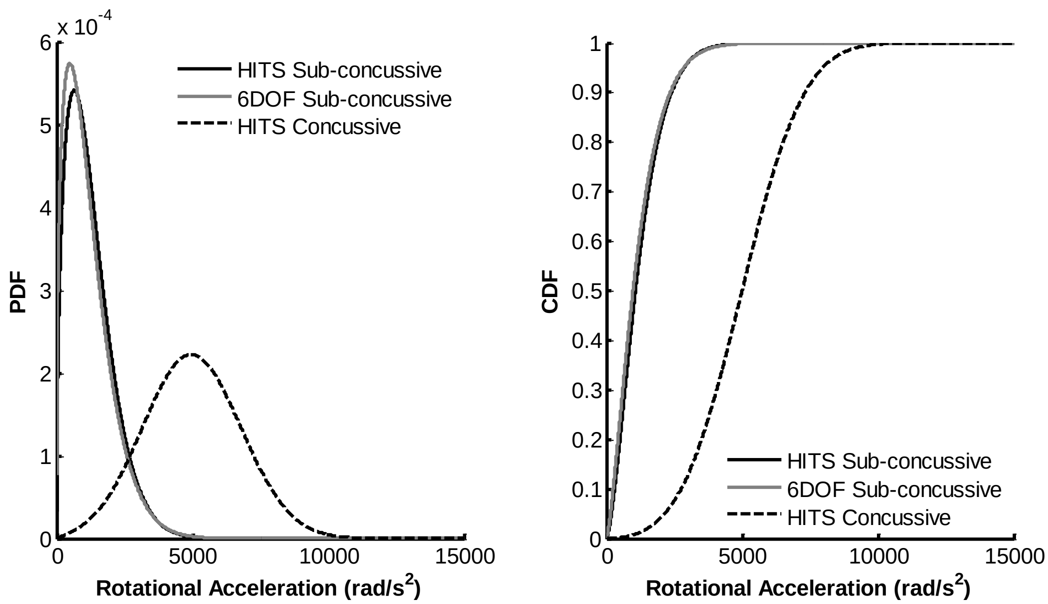

Figure 15: Weibull distributions were fitted to resultant rotational head acceleration for sub-concussive impacts recorded with the HIT System and 6DOF measurement device. A Rician distribution was fitted to resultant rotational head accelerations for concussive impacts recorded with the HIT System. Probability density functions (left) and cumulative density functions (right) are displayed for each distribution fit.

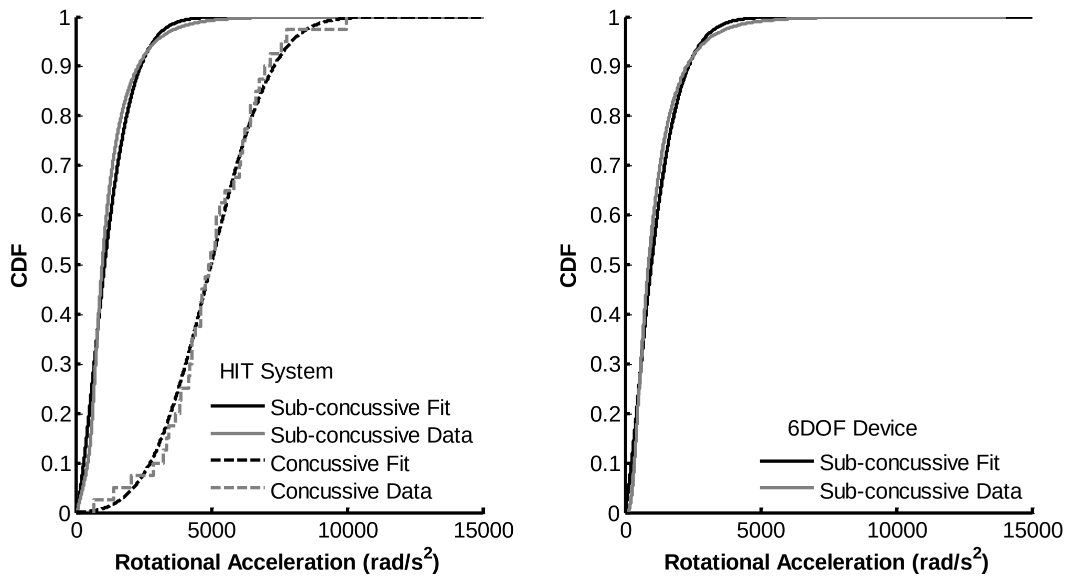

Figure 16: Comparison of the empirical cumulative density functions to the fitted cumulative density functions suggest good fits for both the HIT System datasets (left) and 6DOF measurement device dataset (right).

Table 16: Distribution fitting parameter estimates for Weibull (Equations 2 and 3) and Rician (Equations 4 and 5) distributions.

	Weibull		Rician	
	α	β	σ	ν
Sub-concussive HITS	1369.8	1.4875	-----	-----
Sub-concussive 6DOF	1277.6	1.3670	-----	-----
Concussive HITS	-----	-----	1863.2	4626.2

A total of 1285 impacts were recorded with the 6DOF measurement device that had peak linear accelerations greater than 40 g and were used to quantify the relationship between rotational acceleration and rotational velocity. Peak rotational acceleration and peak rotational velocity correlated strongly (R^2 = 0.94) in the 6DOF dataset, proving to be a linear relationship (Figure 17). The inverse slope parameter (m) in Equation 6 was determined to be 225.5 with nominal units of s^{-1}. Table 17 displays the rotational velocities associated with descriptive rotational accelerations of note.

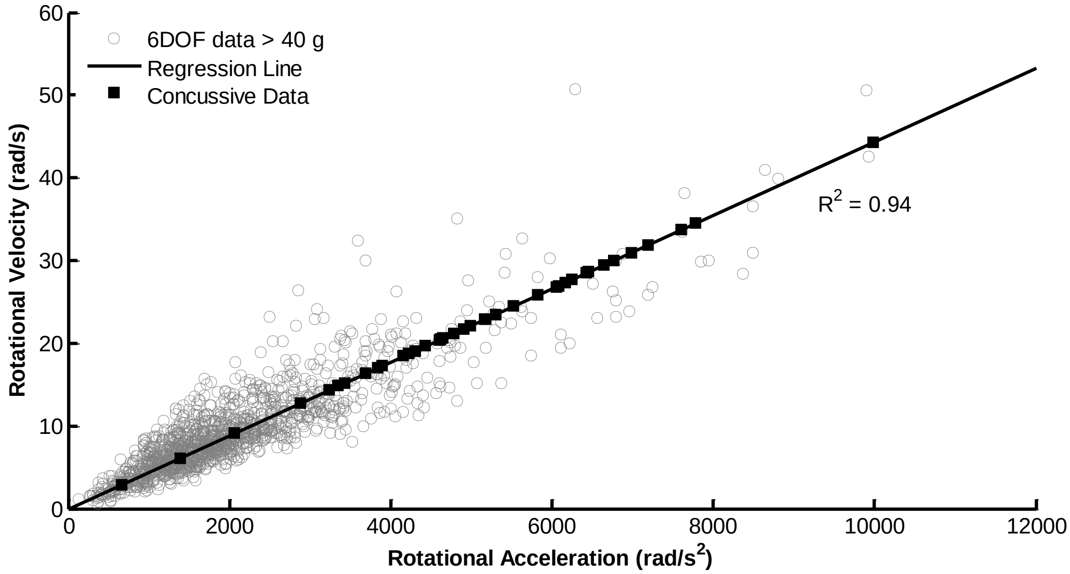

Figure 17: Linear regression relating rotational acceleration to rotational velocity for 1285 impacts recorded using the 6DOF measurement device that had peak linear accelerations greater than 40 g. Using this model, rotational velocities were estimated for concussive impacts recorded using the HIT System.

Table 17: Descriptive statistics of rotational accelerations distributions with associated rotational velocities.

	Descriptive Statistics							
	25th Percentile		Median		75th Percentile		Average	
	α	ω	α	ω	α	ω	α	ω
Sub-concussive HITS	682	3.0	981	4.4	1506	6.7	1230	5.5
Sub-concussive 6DOF	531	2.4	872	3.9	1447	6.4	1158	5.1
Concussive HITS	4026	17.9	4948	21.9	6209	27.5	5022	22.3

α is rotational acceleration with units rad/s^2; ω is rotational velocity with units rad/s

Figure 18 displays the probability of concussion as a function of peak rotational acceleration. The risk function (Equation 7) parameter estimates were determined to be -12.531 for α and 0.002 for β. Table 18 displays rotational accelerations and rotational velocities for nominal injury risk values.

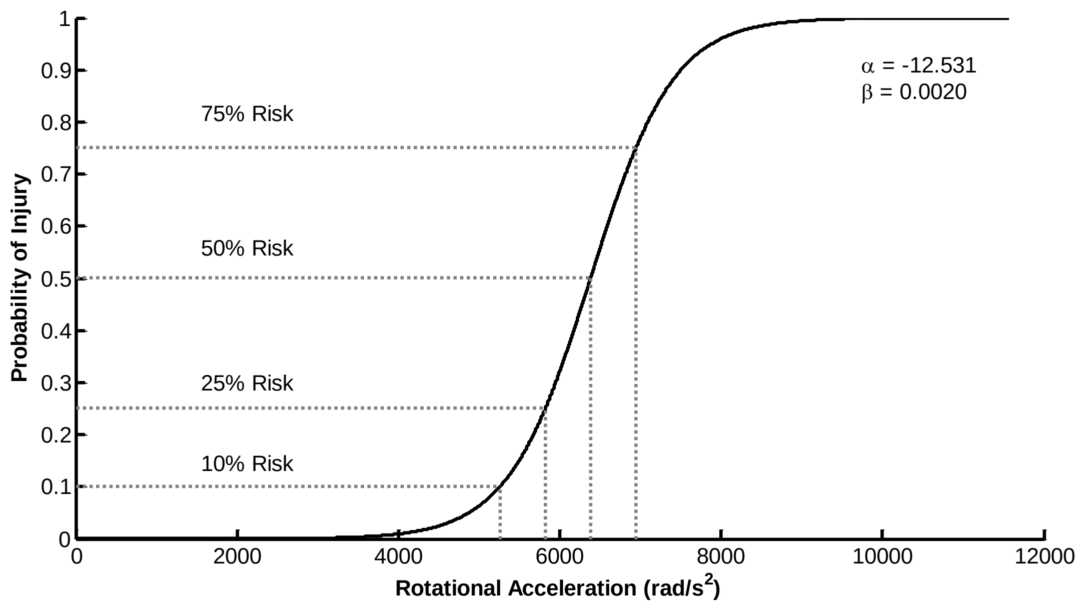

Figure 18: Injury risk as a function of peak resultant rotational acceleration. Parameter estimates for Equation 7 and nominal injury risk values are superimposed on the plot.

Table 18: Rotational accelerations and rotational velocities associated with nominal injury risk values.

Nominal Injury Risk	Rotational Acceleration (rad/s^2)	Rotational Velocity (rad/s)
10%	5260	23.3
25%	5821	25.8
50%	6383	28.3
75%	6945	30.8
90%	7483	33.2

Rotational accelerations of lesser magnitudes were observed with impacts to the top of the helmet. Sub-concussive impacts to the top of the helmet recorded with the 6DOF measurement device were right skewed with a 25^{th} percentile rotational acceleration of 346 rad/s^2, median rotational acceleration of 595 rad/s^2, and 75^{th} percentile rotational acceleration of 1057 rad/s^2 (average rotational acceleration of 845 ± 798 rad/s^2). Sub-concussive impacts to the top of the helmet recorded with the HITS System were right skewed with a 25^{th} percentile rotational acceleration of 266 rad/s^2, median rotational acceleration of 446 rad/s^2, and 75^{th} percentile rotational acceleration of 768 rad/s^2 (average rotational acceleration of 615 ± 565 rad/s^2). Concussive impacts to the top of the helmet recorded with the HIT System had a 25^{th} percentile rotational acceleration of 617 rad/s^2, median rotational acceleration of 1822 rad/s^2, and 75^{th} percentile rotational acceleration of 3673 rad/s^2 (average rotational acceleration of 2192 ± 1790 rad/s^2).

Discussion

The rotational acceleration distributions for the 6DOF measurement device and the HIT System were in good agreement. The small differences between the distributions can be attributed to the effect of varying head impact exposures for different football positions among instrumented players. The 6DOF dataset was collected from lineman, because these subjects wear larger helmets that could accommodate the 6DOF measurement device. The HIT System dataset was collected from lineman and skill players. Recent research has shown that lineman sustain impacts more frequently at lower magnitudes relative to skill players. The minimal difference in distributions between the two datasets suggests that the HIT System was capable of accurately quantifying the head impact exposure of rotational acceleration experienced by the instrumented football players.

While rotational acceleration could be reasonably calculated with the HIT System, a rotational acceleration without a rotational velocity is difficult to interpret with relation to injury. A rotational velocity associated with a rotational acceleration provides information about the temporal component of the acceleration pulse. Rotational head accelerations of great magnitudes can be tolerable over very short durations; however, as duration increases, tolerance decreases. Moreover, rotational velocity was of particular interest in this study because it has been shown to have a stronger correlation with relative brain motion than any other kinematic parameter [40, 41]. Computational studies have also found rotational velocity to be a predictor of the strain response when modeling real-world head impacts that were experimentally recorded from football players [42]. The relationship between rotational acceleration and rotational velocity in the 6DOF dataset was strong and linear. The strong correlation between the two parameters suggests that head acceleration pulses as a result of head impacts in football are similar in duration and acceleration shape. The linear regression model was used to determine the average rotational velocity associated with peak rotational acceleration at sub-concussive and concussive severities.

Injury risk was assessed as a function of rotational acceleration using an unbiased analysis of a large dataset. Sub-concussive and concussive acceleration distributions were weighted to reflect an accurate ratio between sub-concussive and concussive impacts. The distribution weighting techniques utilized published concussion incidence rates and considered the under-reporting of concussions, which is a problem of increasing concern [23, 39, 43, 44]. Pellman et al. [22] generated injury risk curves for concussion from reconstructed NFL impacts using Hybrid III ATDs. In that study, the average concussive impact (n = 25) had a rotational acceleration of 6432 rad/s^2 and rotational velocity of 36.5 rad/s. The average sub-concussive impact (n = 33) had a rotational acceleration of 4028 rad/s^2 and rotational velocity of 26.1 rad/s. Figure 19 compares the injury risk curve derived from the NFL data for rotational acceleration to the risk curve produced in this study. In comparison to the risk curve generated in this study, the NFL risk curve over-predicts injury risk at lower severities (risk < 50%) and produces similar values at higher severities (risk > 50%). The differences between the two risk curves can partially be attributed to the NFL data being biased towards concussive impacts. Furthermore, the NFL data were based on reconstructions from game film using Hybrid III ATDs. While the Hybrid III is often used to evaluate sports injury scenarios in the laboratory [45, 46], the neck of the HIII has limited biofidelity. The Hybrid III ATD reconstructions produced similar peak accelerations for concussive impacts, but generated higher rotational velocities. The temporal response of the Hybrid III neck to head impact is elongated due to its low stiffness [47]. Although the use of the Hybrid III has caveats, it remains a valuable tool when collecting data from humans is not feasible.

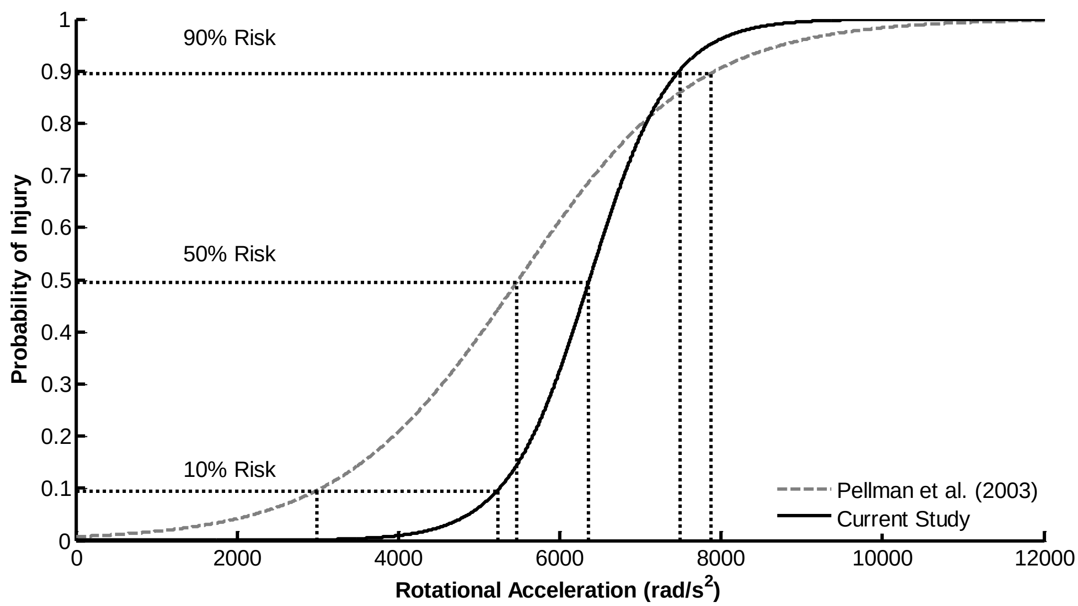

Figure 19: Comparison of the concussion risk curve generated in this study to that of Pellman et al. (2003). Nominal injury values of 10%, 50%, and 90% are emphasized to display differences between the two curves at varying severities.

Previous studies have generated rotational kinematic thresholds from scaled animal data for DAI. Although DAI is a more severe injury than the concussion injury analyzed in this study, there is value in comparing results. Ommaya [5] utilized a primate model and suggested an injury threshold of 4500 rad/s^2 when rotational velocity is less than 30 rad/s for sagittal plane rotation of the head. Additionally, Davidsson et al. [13] utilized a rat model and suggested a threshold of 10000 rad/s^2 with a rotational velocity of 19 rad/s for rearward sagittal plane rotation. For coronal plane rotation, Marguiles and Thibault [16] utilized a primate model and suggested a threshold of 16000 rad/s^2 with a rotational velocity of 46.5 rad/s. Figure 20 compares these published thresholds for DAI to the data collected from football players. The kinematics of these experiments had a negligible linear component, as they were designed to invoke pure rotation of the head. While theoretically possible, this phenomenon is likely rarely experienced in the real-world [12]. No head impact measured in football players was comprised of pure rotation. Moreover, these animal studies limited rotation to a single plane of the head, while the impacts measured from football players involved rotation in all three planes of the head simultaneously. With that said, the average concussive values of 5022 rad/s^2 and 22 rad/s generated in this study are most similar to that of Ommaya [5]. However, Ommaya's criterion was proposed to predict a life-threating brain injury. Ommaya's criterion was self-admittedly speculative for injury to humans due the scaling techniques used to transform the rhesus monkey data to human data [5, 48]. Similar caution should be exercised when drawing conclusions based on injury thresholds derived from Margulies and Thibault [16] and Davidsson et al. [13].

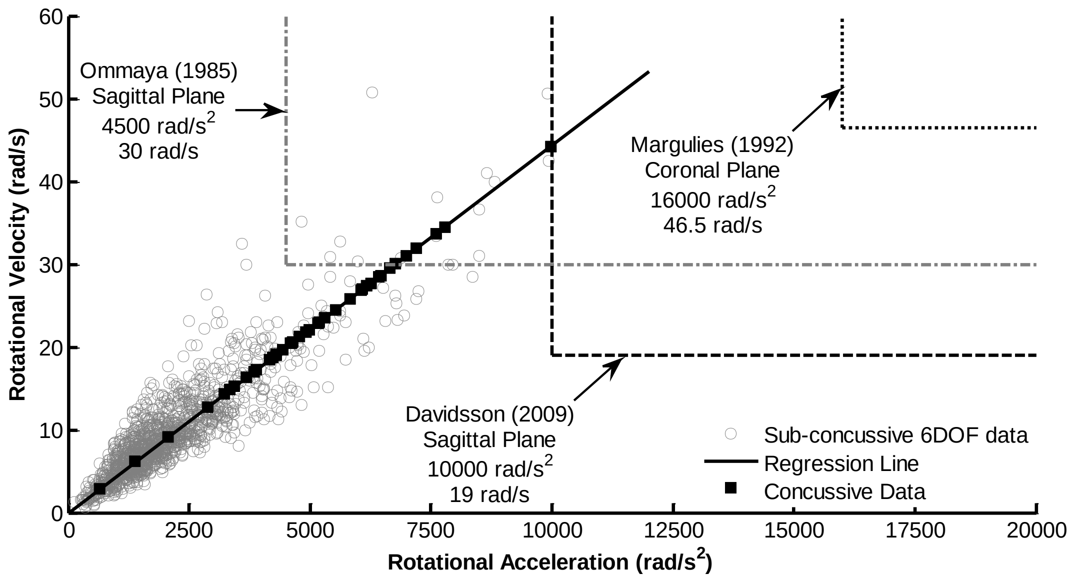

Figure 20: Comparison of sub-concussive and concussive data collected from football players to DAI thresholds derived from animal data that were scaled to reflect human data.

While the 6DOF measurement device was used to measure rotational acceleration, Equation 1 was used to calculate rotational acceleration for impacts recorded with the HIT System. Equation 1 calculates rotational acceleration from the resultant linear acceleration along the anterior-posterior and medial-lateral axes of the head, the inertial properties of the head, and a calculated average direction of force. Since rotational acceleration for the HIT System is determined from the acceleration vector of the head CG in the transverse plane, this analysis is insensitive to transverse rotation and only considers sagittal and coronal plane rotation. Of the impacts recorded, 67.5% were to the front or back of the helmet; indicating that the majority of impacts were dominated by sagittal plane rotation. These data are consistent with those previously reported [28, 33]. For this reason, the moment of inertia used in Equation 1 is for the sagittal plane that was established from cadaver data [37, 49]. Notably, linear acceleration along the inferior-superior axis of the head is not considered in Equation 1. Impacts that had the largest accelerations along this axis likely had little rotation due to the impact force being transmitted through (or near) the head CG and neck. For this reason, impacts to the top of the helmet were separated from the main analysis, as this study focuses on the rotational kinematics. Figure 21 compares the linear and rotational accelerations associated with concussion for impacts that were generalized into three groups: sagittal rotation, coronal rotation, and impacts to the top of the helmet. Table 19 compares the average linear acceleration, rotational acceleration, and rotational velocity for each of the three groups. Although the linear accelerations for each impact mode were very similar, rotational kinematics for impacts to the top of the helmet were substantially lower than impacts to the front, back, or sides of the helmet. This supports the notion that both linear and rotational components of acceleration contribute to concussion [18].

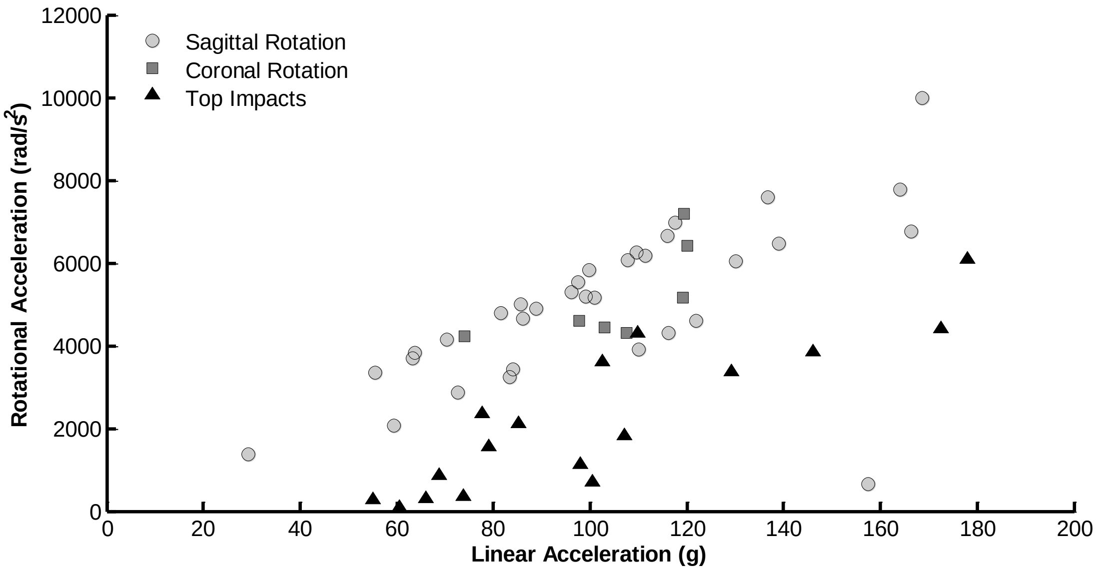

Figure 21: Linear and rotational accelerations for concussive impacts grouped by impact mode. Impacts to the top of the helmet had similar peak linear accelerations and lower rotational accelerations than other impact locations.

Table 19: Average concussive linear acceleration and rotational kinematics for impacts that were either primarily sagittal plane rotation (front and rear impact locations), primarily coronal plane rotation (side impact locations), or to the top of the helmet.

	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Rotational Velocity (rad/s)
Sagittal plane rotation	102.7 ± 33.6	4986 ± 1909	22.1 ± 8.5
Coronal plane rotation	105.8 ± 16.6	5192 ± 1166	23.0 ± 5.2
Impacts to helmet top	100.6 ± 37.1	2192 ± 1790	9.7 ± 7.9

Linear and rotational acceleration have traditionally been examined independently of one another, even though both can contribute to brain injury [12, 18]. This is largely due to linear acceleration and rotational acceleration being correlated to different injury mechanisms. Studies have suggested that linear acceleration is correlated to the intracranial pressure response [7, 41, 50, 51], and that rotational kinematics are correlated to the strain response of the brain [7, 42, 52]. Brain injuries due to linear acceleration are typically focal in nature, while brain injuries due to rotational acceleration are typically diffuse in nature, but can also produce focal damage [5]. Ommaya [5] suggested the use of dual criteria when predicting brain injury due to head kinematics: his own rotational kinematic criterion and the maximum strain criteria [53, 54], which is based on linear acceleration. In this same light, several researchers have suggested that combined linear and rotational kinematic parameters are likely to have the greatest predictive capabilities of concussion [55, 56]. With the increased understanding of injury risk related to single biomechanical parameters, more work should be conducted investigating the combined role of linear and rotational kinematics in producing injury.

The kinematics associated with concussion appear to be clearly defined as a non-zero normal distribution, which indicates there is a correlation between mechanical input and clinical outcome. However, there were many impacts with accelerations at concussive levels that did not result in injury. This suggests that there are great variations in genetic traits and individual tolerances between humans, and that there may also be better predictors of brain injury than head kinematics. Ultimately when evaluating the safety of consumer products, it is desirable to perform a test in a laboratory with a human surrogate and predict brain injury from head kinematics, more advanced methods may offer improved predictive capabilities. Using the head kinematics of the human surrogate as input to finite element models of the head, the tissue level response can be quantified, and the strain or pressure response (or any other parameter of interest) of the brain can be used to assess injury [42, 51, 52]. However, before this is possible, the best injury predictors must be determined and validated using injury data, such as the data presented in this study. Unfortunately, these predictors are likely to be model-specific, as each model may find a different parameter that best predicts injury.

This study has several limitations. First, it should be noted that linear acceleration was measured using the HIT System and rotational acceleration was calculated from a linear acceleration vector, the inertial properties of the head, and an average direction of force. Although rotational acceleration was not directly measured, the calculation provides a good estimate. Second, there is measurement error associated with both the HIT System and 6DOF measurement device. However, the average errors of these devices are on the order of 1-4%. While there may be greater errors associated with individual data points, these errors are of little consequence when working with the overall data distributions. Third, many concussions sustained while participating in football are unreported or undiagnosed. This study makes an attempt to account for unreported concussions in our injury incidence calculation, but the under-reporting of concussions may bias our data. Finally, although every impact was composed of linear and rotational kinematics, this study investigates rotational kinematics independent of linear acceleration. More work is needed investigating the combined contribution of linear and rotational kinematics to brain injury.

The significance of this study lies within methods that collect biomechanical head impact data from humans at potentially injurious severities and pair these data with clinical outcome. Large sub-concussive and concussive datasets were analyzed and characterized. This study addresses the limitations of earlier experiments, in that data were measured directly from humans and a large and unbiased dataset was produced. Valuable insight to the rotational kinematics associated with concussion in humans has been presented. With an increased understanding of the kinematics associated with injury, engineering analyses can be used to evaluate and influence product design to reduce injury incidence [45, 57].

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Chapter 5:

Research Summary and Resulting Publications

Research Summary

The research in this dissertation aimed to biomechanically characterize sports-related concussion with respect to linear and rotational acceleration. The output of this research has provided valuable insight to the head kinematics associated with concussion. Aspects of this research have direct applications towards product safety evaluation methods.

The research presented in this dissertation has yielded the following:

1. A risk function for concussion based on linear head acceleration.
2. An evaluation tool for football helmets that consumers can utilize to make educated decisions when considering impact performance when purchasing a helmet.
3. A custom in-helmet accelerometer array was developed that can be used to measure linear and rotational head acceleration for each axis of the head.
4. A risk function for concussion based on rotational acceleration.

Expected Publications

The research presented in this dissertation is intended to be published in several journals. Table 20 displays the publications for each chapter in this presentation. All chapters are currently in their publication forms.

Table 20: Expected publications from this dissertation.

Chapter	Title	Journal
2	Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion	Annals of Biomedical Engineering
3	A Six Degree of Freedom Head Acceleration Measurement Device for Use in Football	Journal of Applied Biomechanics
4	Rotational Kinematics Associated with Concussions in Humans	Journal of Biomechanical Engineering

Current Publication Status

Journal Publications

1. Rowson, S., Beckwith, J.G., Chu, J.J., Leonard, D.S., Greenwald, R.M., and Duma, S.M., A Six Degree of Freedom Head Acceleration Measurement Device for Use in Football. **J Appl Biomech**, 27(1): p. 8-14, 2011.
2. Rowson, S., McNally, C., and Duma, S.M., Can footwear affect achilles tendon loading? **Clin J Sport Med**, 20(5): p. 344-9, 2010.
3. Rowson, S., Brolinson, G., Goforth, M., Dietter, D., and Duma, S.M., Linear and angular head acceleration measurements in collegiate football. **J Biomech Eng**, 131(6): p. 061016. 2009.
4. Rowson, S., McNeely, D.E., Brolinson, P.G., and Duma, S.M., Biomechanical analysis of football neck collars. **Clin J Sport Med**, 18(4): p. 316-21, 2008.

In-Preparation Journal Publications

1. Rowson S and Duma S.M., Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion. **Annals of Biomedical Engineering**, 2011 (submitted).
2. Rowson S, Duma S.M., et al., Rotational Kinematics Associated with Concussion in Humans. **Journal of Biomechanical Engineering**, 2011.
3. Rowson S, Duma S.M., et al. Head Impact Severity Likelihood during Competitive Football Play. **Medicine and Science in Sports and Exercise,** 2011.
4. Rowson S, Beeman S, Hulse M, Strom B, Field R, and Duma S.M., Head Impact Response from Child Toy Swords. **Pediatrics**, 2011.

Refereed Conference Publications

1. Rowson S, McNally C, and Duma SM, In situ measurement of Achilles tendon tension during dorsiflexion, **Biomedical Sciences Instrumentation**, Vol. 45, pp. 18-23, 2009.
2. Rowson S, Goforth MW, Dietter D, Brolinson PG, and Duma SM, Correlating cumulative sub-concussive impacts in football with player performance, **Biomedical Sciences Instrumentation**, Vol. 45, pp. 113-118, 2009.
3. Rowson S, McNeely DE, and Duma SM, Force transmission to the mandible by chin straps during head impacts in football, **Biomedical Sciences Instrumentation**, Vol. 44, pp. 195-200, 2008.
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5. Rowson S, McNeely DE, and Duma SM, Lateral bending biomechanical analysis of neck protection devices used in football, **Biomedical Sciences Instrumentation**, Vol. 43, pp. 200-205, 2007.