

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

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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 article 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 four impact locations and six impact energy levels. Using 62,974 head acceleration data points collected from football players, the number of impacts experienced for one full season was translated to 24 drop test configurations. A new injury risk function was developed from 32 measured concussions and associated exposure data to assess risk of concussion for each impact. 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.

Keywords—Concussion, Mild traumatic brain injury, Acceleration, Risk, Exposure, HITS, Impact.

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.^{8,29} Studies showing the potential long-term effects of these injuries have put sports-related concussions under the national spotlight as a primary health concern.^{16,35,36} Furthermore, there is concern that repetitive sub-concussive head impacts in

sports may lead to neurocognitive deficits.^{18,19,25} 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.⁷ This article 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.²⁷ Head acceleration is thought to be indicative of the inertial response of the brain, and therefore is used to predict brain injury.²⁷ 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.^{23,28} 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. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) provides a set of voluntary standards that are designed to assess a helmet's ability to prevent skull fracture. 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 NOCSAE standards have done an excellent job of

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eliminating skull fractures in helmeted sports, but do not consider concussion or provide data to the public. 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.¹¹ While Duma *et al.*¹¹ were the first to instrument football players with the HIT System, other researchers have adopted this technology to investigate head impacts in football.^{4,21,40,43} The measurement and analysis of head acceleration data collected from these in-helmet accelerometer arrays have been well-validated and widely accepted.^{5,11,24,39} 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. Between all researchers utilizing the HIT System to collect head acceleration data from football players, 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 against concussions. This new evaluation system is designed to integrate the overall helmet impact performance into a singular metric that is derived from the head impact exposure and injury risk based on real-world impact measurements in football players. Head impact exposure is expressed as the number of impacts that are experienced by players in relation to specified drop test configurations. 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

The STAR Equation

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 to evaluate the relative performance of football helmets. The STAR equation was developed to relate the head impact exposure that a football

player experiences throughout one full season of participation to the injury risk associated with each head impact (Eq. 1). Fundamentally, this equation correlates every head impact that a football player experiences during one season of participation to 24 drop tests (four locations \times six drop heights) that can be performed using a NOCSAE-style drop tower and head form.³⁴ 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.

$$STAR = \sum_{L=1}^4 \sum_{H=1}^6 E(L, H) \cdot R(a), \quad (1)$$

where *STAR* is the Summation of Tests for the Analysis of Risk, *L* represents one of four impact locations (front, rear, side, or top), *H* represents one of six drop heights (60, 48, 36, 24, 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, and *a* represents peak resultant head acceleration resulting from each specific drop height and impact location.

The methods presented in this article 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.*⁶ quantified impact exposure in detail for three collegiate football teams over the duration of a single season; and reported that players can experience over 1400 impacts over the course of one season. 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.⁶ 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

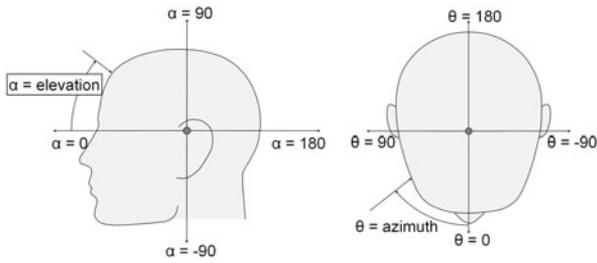


FIGURE 1. Definition of parameters used to group impact locations. Azimuth and elevation values for the impact vector were computed for each impact based on the head acceleration data measured.

impacts that were recorded between 2009 and 2010 with the HIT System at Virginia Tech.^{11,12} These impacts were analyzed to determine the head impact exposure on an impact location basis.⁵ Each recorded head impact was categorized into one of four general impact locations based on the computed azimuth and elevation of the impact vector (Fig. 1). Any impact with an elevation $>65^\circ$ was categorized as an impact to the top of the helmet. Impacts with elevations $<65^\circ$ and azimuths between 45° and -45° were categorized as impacts to the front of the helmet. Impacts with elevations $<65^\circ$ 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.

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, OH).^{11,12} Only one helmet type was used so that potential variations in helmet response between types would be removed for this part of the analysis. This analysis is geared toward quantifying impact exposure as impact energy input to the helmet, rather than resulting head acceleration from impact as a function of helmet response. These impacts were categorized into the aforementioned impact locations and analyzed to determine head impact severity distributions.

A Weibull distribution was fit to the peak linear head acceleration data for each impact location. The Weibull probability density function (pdf) takes the form of Eq. (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) (Eq. 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.

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

$$w_{cdf} = 1 - e^{-\left(\frac{x-\theta}{\beta}\right)^\alpha}, \quad (3)$$

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, and 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 three Large Riddell VSR4 helmets were tested on a medium NOCSAE head form using a NOCSAE-style drop tower. Drop heights ranged from 6 to 66 in. and were incremented by 6 in. For each of the three 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.¹⁵ A second 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 (Eq. 4).

TABLE 1. Integral bounds used to determine the number of impacts associated with each drop height for each impact location.

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

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.

$$H = p_1 a^2 + p_2 a + p_3, \quad (4)$$

where p_1 , p_2 , and p_3 are regression coefficients, H is drop height, and 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, 24, 36, 48, 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 1 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 with head accelerations <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.^{12,21} 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.

Injury Risk

The sub-concussive head acceleration data set 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 three separate studies using identical data collection protocols with the HIT System to create a data set of 32 clinically diagnosed concussions.^{3,12,21} These data sets 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 Eq. (2). Concussive impacts were found to be normally distributed, and therefore were fit to a normal probability density function (Eq. 5).

$$n_{pdf} = \frac{1}{\sigma\sqrt{2\pi}} e^{-\frac{(x-\mu)^2}{2\sigma^2}}, \quad (5)$$

where n_{pdf} is the normal probability density function, μ is the mean, and 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.¹ For athletes in the National Football League (NFL), an injury incidence rate of 0.41 concussions per game was considered for professional athletes.³⁷ 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.⁶ 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.³⁰ 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 (Eq. 6). The regression coefficients were determined using a generalized linear model technique.

$$R(a) = \frac{1}{1 + e^{-(\alpha + \beta x)}}, \quad (6)$$

where $R(a)$ 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, 24, 36, 48, 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 (Eq. 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 (Fig. 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.

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

A total of 132 drop tests were performed investigating average head acceleration as a function of drop

height and impact location for three Riddell VSR4 helmets. Table 3 displays the regression coefficients and R^2 values for the relationship between head acceleration and drop height for these tests (Eq. 4). Figure 4 displays how the number of impacts for each drop height was determined from the Weibull distribution for the front impact location, while Table 4 displays the number of impacts for each drop height for all impact locations.

Injury Risk

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

STAR Value Assessment

Table 6 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.

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 dominating 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

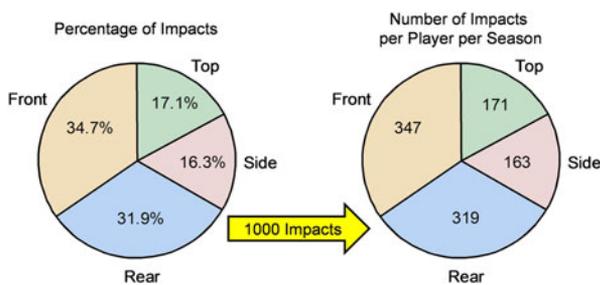


FIGURE 2. Percentage of impacts associated with each impact location (left). Assuming 1000 impacts per player per season, the number of impacts to each location was determined (right).

TABLE 2. 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

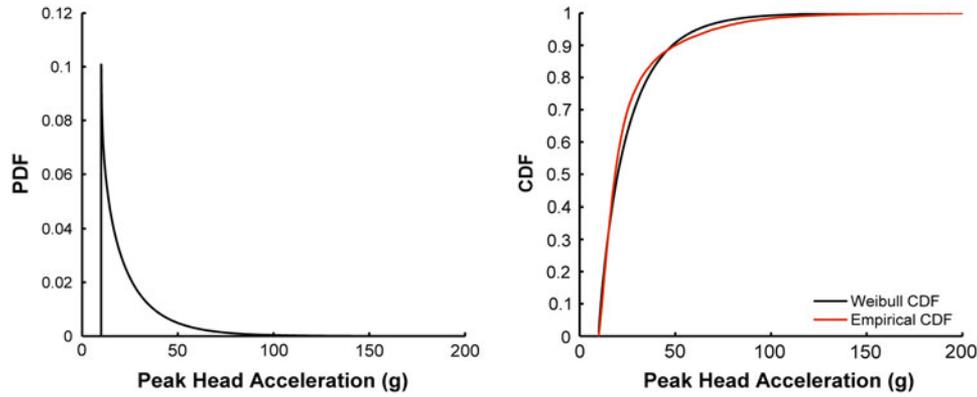


FIGURE 3. Weibull probability density function fitted to head accelerations resulting from impacts to the front of the helmet (left). Quality of fit can be investigated through comparison of the Weibull cumulative density function to an empirical cumulative density function of the same data (right).

TABLE 3. The second order polynomial regression coefficients and R^2 values for drop height and head acceleration relationships (Eq. 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

TABLE 4. 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 90th percentile player.

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

Note: impacts < 19 g are not considered in the STAR testing protocol.

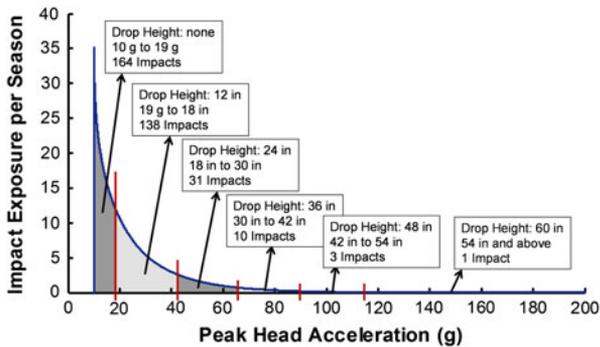


FIGURE 4. Number of impacts associated with each drop height for the front impact location, with integral bounds displayed for each drop height.

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 data set.¹⁴

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.

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.*²¹ reported that the average collegiate football player experiences 950 impacts per season. In addition, Schnebel *et al.*⁴³ found the average collegiate player to experience 1353

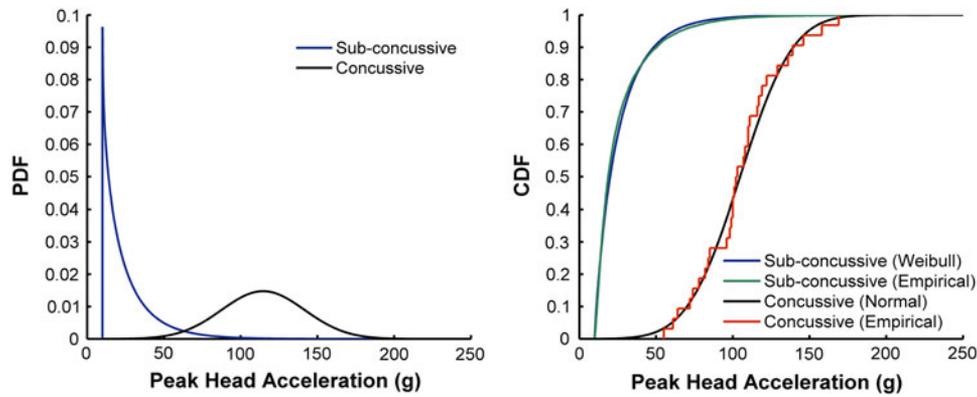


FIGURE 5. Weibull probability density function for all sub-concussive impacts and normal probability density function for all concussive impacts (left). Comparison of distribution fits for sub-concussive and concussive data to empirical data using cumulative density functions (right).

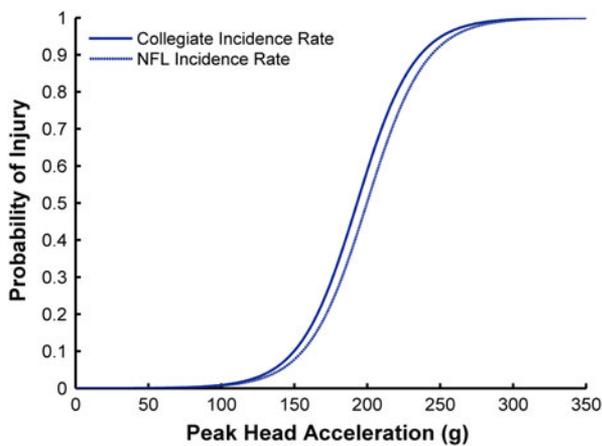


FIGURE 6. Injury risk curves based on the collegiate and NFL injury rates using resultant peak linear head acceleration.

TABLE 5. Logistic regression parameters for the college and NFL injury incidence rate-based risk curves (Eq. 6).

	α	β
Collegiate	-9.805	0.051
Professional	-9.928	0.497

impacts per season. However, it should be noted that these numbers may not be directly applicable to high school athletes. Broglio *et al.*⁴ measured an average of 565 impacts per player in high school athletes. 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 7).

Furthermore, the head impact exposure per player per impact location is nearly identical to other published studies. Table 8 compares the impact location

TABLE 6. 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

weightings used in the STAR evaluation system to that of another study that utilized the same data collection methodologies.³¹ Moreover, these results are consistent with a study of three collegiate teams.⁶ It is important to note that exposure on a player position basis was not considered although it has been shown the impact frequency and magnitude vary by position.^{6,31} Since close to every player on the field is instrumented, the exposure values reported in this

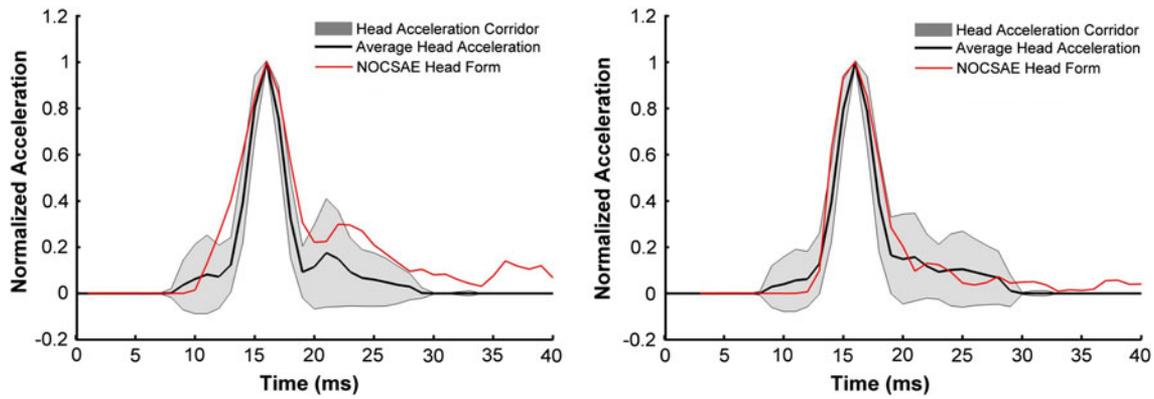


FIGURE 7. Comparison of typical normalized NOCSAE head form impact response to football head impact acceleration corridors for the front (left) and side (right) impact locations. All acceleration curves were normalized to their peak values.

TABLE 7. 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

study represent an average exposure for all player positions. Moreover, most players play multiple positions and no position-specific helmets currently exist.

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

TABLE 8. 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 <i>et al.</i> ³¹
Front	34.7	35.9
Rear	31.9	30.9
Side	16.3	14.4
Top	17.1	18.8

TABLE 9. Published injury incidence rates expressed as concussions per 1000 athletic exposures in collegiate football.^{1,9,20}

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

participating in at least one play of one game or practice. Table 9 is a summary of published injury incidences rates for concussion in football.^{1,9,20} Booher *et al.*¹ reported higher injury incidence rates because that study included injuries that did not result in loss of playing time, where Guskiewicz *et al.*²⁰ and Dick *et al.*⁹ only considered injuries associated with loss of playing time. In addition, Pellman *et al.*³⁷ reported an injury incidence rate of 0.41 concussions per game in the NFL. 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.^{2,13,40,46}

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,¹⁰ such approaches are only representative of the data that are experimentally recorded.²⁶ Moreover, Kent and Funk²⁶ have shown that parametric injury risk functions typically fall within the 95% confidence intervals of non-parametric injury risk functions for any given data set. 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.*³⁸ created injury risk curves based on reconstructed concussive impacts from NFL football games. 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.*³⁸ was 98 ± 27 g, which is nearly identical to the average concussive linear acceleration collected from human volunteers in this study (105 ± 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.²² Funk *et al.*¹⁴ quantified risk using unique statistical analysis of data collected from human volunteers. However, that risk curve was based on only four concussive data points, and was thought to underestimate risk. For head accelerations that are associated with <10% risk, the

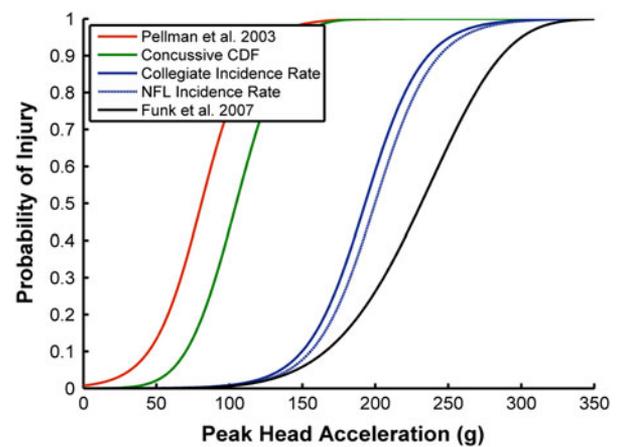


FIGURE 8. Comparison of injury risk curves generated in this study to that of previously published research. The cumulative density function of the normally distributed concussive data set (concussive cdf) is displayed to illustrate the head accelerations associated with only concussive impacts.

risk curve from Funk *et al.*¹⁴ 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.

The injury risk curves utilized in the STAR equation currently only incorporate linear acceleration. It should be noted that these linear accelerations are associated with impact durations of 8–12 ms. Although rotational acceleration is not considered in this analysis, work is underway investigating human tolerance to rotational kinematics.⁴⁰ 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.¹⁷ 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.^{32,33} Finite element models will likely be an important factor in understanding this relationship.⁴⁵ 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.^{41,42,44} 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 four impact locations and six 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. The STAR system is conceptual and its ability to differentiate between helmet designs will be addressed in subsequent work. 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.

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