Evaluation of the Biomechanical Performance of Youth Football Helmets

David William Sproule

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Steven Rowson, Chair
Stefan M. Duma
Per Gunnar Brolinson

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ACADEMIC ABSTRACT

Youth and varsity football helmets are currently designed similarly and tested to the same impact standards from the National Operating Committee on Standards for Athletic Equipment (NOCSAE). Youth players have differences in anthropometry, physiology, impact exposure, and potentially injury tolerance that should be considered in future youth-specific helmets and standards. This thesis begins by investigating the current standards and relating them to on-field data. The standard drop tests represented the most severe on-field impacts, and the performance of the youth and varsity helmet did not differ. There likely is not a need for a youth-specific standard as the current standard has essentially eliminated the catastrophic head injuries it tests for. As more is known about concussion, standards specific to the youth population can be developed. The second portion of this thesis compares the impact performance between 8 matched youth and varsity helmet models, using linear acceleration, rotational acceleration, and concussion correlate. It was found that helmet performance did not differ between the youth and varsity helmets, likely attributed to testing to the same standard. The final portion of this feature is aimed at advancing STAR for youth and varsity football helmets by including linear and rotational head kinematics. For varsity helmets, an adult surrogate is used for impact tests which are weighted based on on-field data collected from collegiate football players. For youth helmets, a youth surrogate is used and tests are weighted based on data collected from youth players.

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GENERAL AUDIENCE ABSTRACT

The research presented in this thesis is intended to provide a reference point towards youth-specific football helmets and test standards. Currently, youth football helmets are designed similarly to varsity football helmets and are tested to the same standard. It is known that differences exist between youth and adult players in terms of the impacts they experience, the proportions of their body, and the maturity of the nervous system. However, it remains unknown as to how these differences should be expressed in youth-specific helmet design and impact standards. This thesis investigates the current test standard and relates it to population-specific on-field data. This analysis of both a youth and varsity helmet, suggest that there is no current benefit for a youth-specific standard until differences in concussion tolerance are better understood. This thesis goes on to compare the relative impact performance between multiple matched youth and varsity helmets, using a more realistic test setup than the current impact standards. Through this investigation it was found that there were no differences in performance between the youth and varsity helmets tested. This thesis then concludes by advancing STAR for youth and varsity football helmets, using a separate protocol specific to each population. Each test protocol is able to identify helmets that most effectively reduce the severity of head impacts in football.

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Chapter 1: Opening Remarks

Football is at the center of concussion research because of the popularity of the sport in the United States and the large number of concussive injuries that occur each year. The risk of head injuries in football can be mitigated through a combination of implementing and enforcing rule changes, teaching proper tackling technique, and improving helmet design. Each intervention takes a different approach towards reducing injury risk. Altering practice structure, eliminating high risk plays, and enforcing proper tackling technique are ways to reduce the number of head impacts that a player sustains. In contrast, improving helmet design is motivated by mitigating the energy transfer to the head. This strategy recognizes that head impacts cannot be entirely removed from the game of football, and seeks to reduce the severity of those that will still occur. The research in this thesis is motivated by improving helmets and helmet standards for the youth population.

Currently, players under the age of 14 wear youth football helmets and players over 14 wear varsity football helmets. The design of youth and varsity football helmets are similar, particularly in matched youth and varsity models. This can be attributed to youth and varsity football helmets being tested to the same pass/fail standard implemented by the National Operating Committee on Standards for Athletic Equipment (NOCSAE). Although it remains unknown how youth-specific helmets or helmet standards should differ some considerations should be made. For example, youth players have different anthropometry, impact exposure, and physiology due to developing nervous systems. These factors, along with others, play possible roles in differences in concussion tolerance in the youth population. As more research on this population is conducted these differences can be better understood and incorporated into youth-specific helmet design and helmet standards.

Research Objectives

The research in this thesis had several objectives. The first objective was to relate on-field football head impacts to the current standard used to evaluate youth and varsity football helmets. Through this we are able to provide perspective on proposed youth helmet standards as well as the current standards. The second objective was to investigate current differences in impact performance for youth and varsity football helmets. The final objective of this research was to advance the STAR methodology for youth and varsity football helmets. This methodology can be used to differentiate relative impact performance between helmets and identify helmets that most effectively reduce head acceleration.

Chapter 2: Football Helmet Impact Standards in Relation to On-Field Impacts

Abstract

Youth football helmets currently undergo the same impact testing and criteria as varsity helmets, although youth football players differ from their adult counterparts in anthropometry, physiology, and impact exposure. This study aimed to relate football helmet standards testing to on-field head impact magnitudes for youth and varsity football helmets. Head impact data, filtered to include only impacts to locations in the current National Operating Committee on Standards for Athletic Equipment (NOCSAE) standard, were collected for 48 collegiate players (ages 18 to 23) and 25 youth players (ages 9 to 11) using helmet-mounted accelerometer arrays. These on-field data were compared to a series of NOCSAE standard drop tests with a youth and varsity Riddell Speed helmet. In the on-field data, the adult players had a higher frequency of impact than the youth players, and a significant difference in head acceleration magnitude only existed at the top location ($p < 0.001$). In the laboratory drop tests, the only significant difference between the youth and varsity helmet was at the 3.46 m/s (61 cm) impact to the front location ($p = 0.0421$). Drop tests generated head accelerations within the top 10% of measured on-field impacts, at all locations and drop heights for both helmets, demonstrating that drop tests are representative of the most severe head impacts experienced by youth and adult football players on the field. Current standards have been very effective at eliminating skull fracture and severe brain injury in both populations. This analysis suggests that there is not currently a benefit for a youth-specific drop test standard for catastrophic brain injury. However, there may be such a need if helmet testing standards are updated to address concussion risk reduction, paired with a better understanding of differences in concussion tolerance between youth and adult populations.

Introduction

Of the estimated 5 million participants in organized football each year, approximately 70% are youth players between the ages of 6 and 13.[1] Currently, youth helmets, which are intended for players younger than 14 years old, undergo an identical impact testing protocol and are held to the same performance criteria as varsity helmets in the National Operating Committee on Standards for Athletic Equipment (NOCSAE) safety standard. Essentially all leagues require these performance criteria must be met for a helmet to be used. Although NOCSAE is in the process of developing a test standard specific to youth helmets,[2] there are many challenges in determining how a youth-specific helmet standard should differ from the current helmet standard.

Notable differences exist between adult and youth football players, where adult players are defined as those players recommended to wear a varsity helmet (age 14 and older) and youth players are those below age 14. Head mass and size are about 95% of adult size at 3.5 years old and stay relatively constant until around age 10, when the head size gradually increases to full adult size around age 16 to 17.[3] However, children have a smaller body resulting in a larger head-to-body size ratio, accompanied by reduced upper body and neck musculature and strength.[4] Concerns specific to brain injury in youth players include an immature central nervous system, thinner cranial bones, a larger subarachnoid space in which the brain can move, and differences in cerebral blood flow.[5] While concussions are a concern, the current NOCSAE standard only addresses catastrophic head injury, such as skull fracture and severe brain injury, and has been very effective in doing so.

The NOCSAE impact testing protocol involves a series of 27 drop tests per football helmet sample, and the helmet is tested without any type of facemask. Tests are done at seven locations: front, side, front boss, rear boss, rear, top, and a random location of the tester's choosing (Figure 1). Four impact velocities are evaluated: 3.46, 4.23, 4.88, and

5.46 m/s. Two additional tests are done to the side location at the highest test velocity, with the helmet temperature increased to 46^{0}C. A passing helmet requires no impact to exceed a Severity Index (SI) of 1200, and no 3.46-m/s velocity impacts to exceed 300 SI.[6] The SI is a function of weighted linear acceleration and duration, and can be described as a correlate of energy transfer to the head, where *a* is acceleration and *t* is time (Equation 1).

$$SI = \int a(t)^{2.5} dt \qquad (1)$$

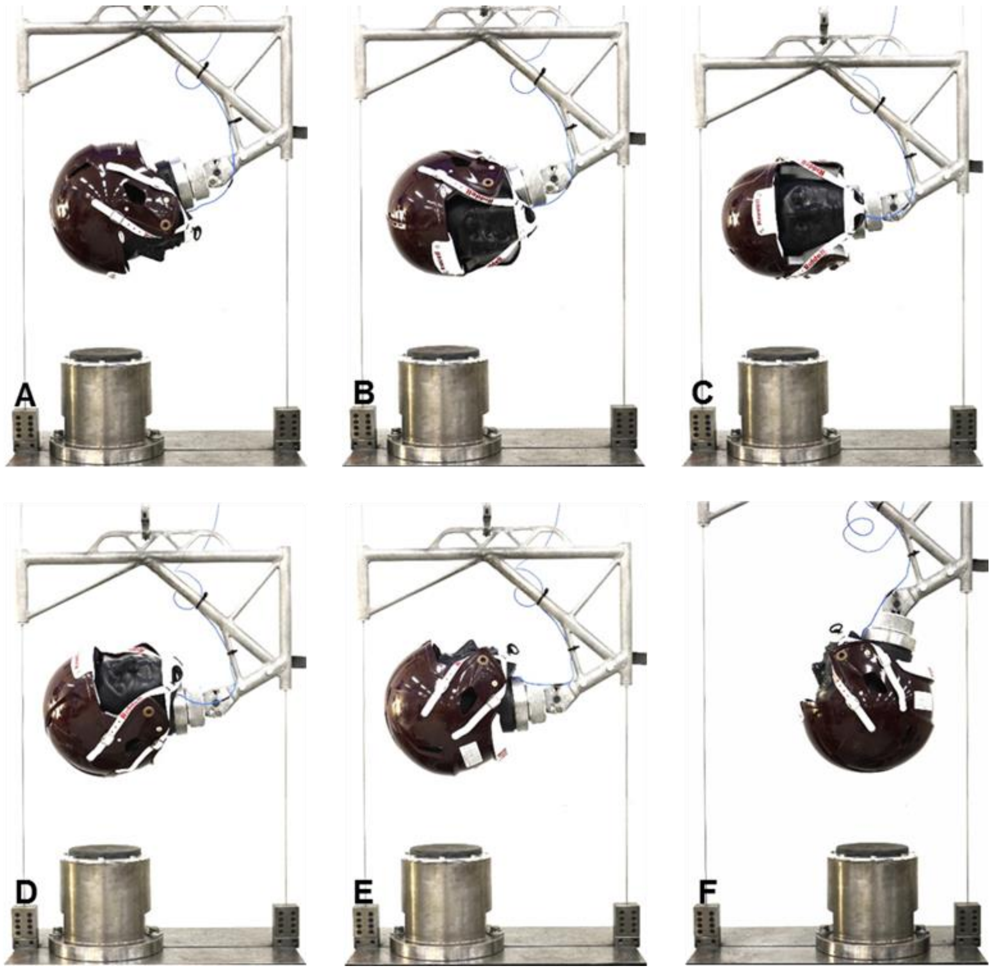

Figure 1: NOCSAE defined drop test impact locations (A) front (B) front boss (C) side (D) rear boss (E) rear (F) top. These locations are photographed on a medium sized NOCSAE headform with a head circumference of 57.6 cm.[6, 7]

Several studies investigating head impact exposure in youth football players of various age groups have demonstrated that the number of head impacts over the course of a single season and the severity of impacts increase with age (Table 1). Although high-severity impacts occur less frequently in younger players, impact magnitudes greater than 80 g are recorded in all age groups.[8] The amount of research investigating head impacts in youth football players available is limited in comparison to what is available for older populations.

Table 1: The number of head impacts sustained per season increases as players increase in age. Furthermore, the 95th percentile acceleration values increase with age. Values are reported as either average ± standard deviation or median [25th percentile – 75th percentile], depending on how they were reported in the literature. Variance was not reported for the collegiate athlete accelerations.

Age Group	# of Impacts	Linear Acceleration (g)		Rotational Acceleration (rad/s^2)	
		50th	95th	50th	95th
Youth 7-8[8]	161 ± 111	16 ± 2	38 ± 13	686 ± 169	2052 ± 664
Youth 9-12[9]	240 ± 147	18 ± 2	43 ± 7	856 ± 135	2034 ± 361
Youth 12-14[1]	275 ± 190	22 ± 2	54 ± 9	954 ± 122	2525 ± 450
High School[10]	340 [226-532]	22 ± 2	56 ± 11	953 ± 132	2519 ± 536
Collegiate[11]	420 [217-728]	21	63	1400	4378

Developing youth-specific helmets has proven challenging given the differences in head impact exposure, anthropometry, and possible differences in injury tolerance[1, 4, 5, 9, 11, 12] It is known that these differences exist, but it remains unknown how these differences should influence helmet design and standards testing. Furthermore, no data are available comparing the impact performance of youth and varsity helmets. This study aimed to relate NOCSAE standard drop tests to on-field head impacts for youth and adult football players. This relating was done through the collection of on-field head impact data for both adult and youth populations, and by investigating how these data compare to a set of drop tests using the helmets worn by these groups. The authors hypothesize that no differences exist between current youth and varsity helmets given that they are similar in design and that they are evaluated using the same NOCSAE testing standard. These data, in addition to providing context for the current standard, can provide insight towards the development of youth-specific helmet standards and improved youth-specific helmets, as this type of comparison has not been previously performed.

Methodology

Head impact data were collected from the Virginia Tech football team, representing an adult population, along with two youth football teams between the ages of 9 and 11 using helmet-mounted accelerometer arrays (Head Impact Telemetry System, Simbex, Lebanon, NH, Sideline Response System, Riddell, Elyria, OH) in Riddell Speed helmets for the 2015 season.[1, 8, 9, 11, 13-18] A total of 25 total youth players (ages 9 to 11) and 48 adult players (ages 18 to 23) were instrumented. From the accelerometer arrays, linear acceleration, rotational acceleration (not used in this analysis), and location for each impact were recorded. Approval for this study was given by the Virginia Tech Institutional Review Board (IRB). Each player provided informed consent for participation, as well as parent/guardian consent for participants under the age of 18.

Data collected from the accelerometer arrays were mapped to the same impact locations used by NOCSAE in the standard.[6] NOCSAE defines these locations as points on the headform in relation to defined planes.[6, 7] These points were mapped to azimuth and elevation coordinates on the head, where azimuth is the horizontal angle and elevation is the vertical angle measured in degrees. The azimuth angles are defined by NOCSAE, and the elevation coordinates are defined by a distance above the basic and reference planes. The elevation angles were determined using these given distances along with the radius of the head, given a medium sized NOCSAE headform (head circumference of 57.6 cm).[7] To map these locations to the on-field data collected from the accelerometer arrays, each location was defined as a range of 15° in either direction for both the azimuth and elevation angles (Table 2). The on-field youth and adult data were then filtered to include only those impacts consistent with these NOCSAE standard location ranges.

Table 2: Impact locations that were mapped to on-field data. Each location was determined by azimuth and elevation coordinates defined by NOCSAE and given a range of 15^0 in either direction to produce an area of comparable impacts from on-field data, where 0^0 in either direction to produce an area of comparable impact from on-field data, where 0^0 is the front of the head at the center of gravity. [6]

Location	NOCSAE definition[7]	Azimuth	Elevation
Front	Located in the median plane approximately 1-in above the anterior intersection of the median and reference plane	0º	30.7º
Front Boss	A point approximately in the 45 degree plane from the median plane measured clockwise and located approximately above the reference plane.	45º	30.7º
Side	Located approximately at the intersection of the reference and Coronal planes on the right side of the headform	90º	17.8º
Rear Boss	A point approximately on the reference plane located approximately 135 degrees clockwise from the anterior intersection of the median and references planes	135º	17.8º
Rear	Approximately at the posterior intersection of the median and reference planes	180º	17.8º
Top	Located approximately at the intersection of the median and Coronal planes. The right hand carriage release ring should be used for this drop.	-	90º

In addition to the on-field data collection, laboratory-based NOCSAE standard drop tests were performed on the same youth and varsity large Riddell Speed helmet models as were used in the on-field data collection. Each helmet was tested at six impact locations (front, front boss, side, rear boss, rear, and top) (Figure 1) and four drop heights (61, 91, 122, and 152 cm) corresponding to the prescribed drop velocities of 3.46, 4.23, 4.88, and 5.46 m/s. Each impact configuration was repeated three times. The system was calibrated to NOCSAE specifications.[7] For each test, a tri-axial accelerometer at the center of gravity of a medium NOCSAE headform was sampled at 20,000 Hz for measuring linear acceleration in the x, y, and z directions, from which peak resultant linear acceleration and Severity Index were calculated. Data were filtered using a phaseless four-pole Butterworth low-pass filter with a cutoff frequency of 1650 Hz.

Differences in the on-field acceleration data were evaluated using a two-way ANOVA for helmet type and location. Differences in drop-test accelerations for the youth and varsity helmets were determined using a two-way ANOVA for helmet and location by drop height. A significance level of $p < 0.05$ was used to during statistical comparisons. In addition, a linear-regression model was fit to the drop test data to evaluate the youth and varsity helmet performance. To relate the standard drop-test conditions to the on-field head accelerations that are experienced by youth and adult football players, matched on-field head accelerations were found for each drop-test head acceleration and the representative on-field percentile impact was determined.

Results

A total of 3842 head impacts were collected for the youth players. For the average youth player, the median acceleration was 17.7 ± 2.2 g and the 95^{th} percentile acceleration was 48.8 ± 17.8 g. For the adult players, a total of 25,322 head impacts were collected over the course of the season, with the average player having a median acceleration of 20.1 ± 1.6 g and 95^{th} percentile acceleration of 60.1 ± 7.8 g. After the on-field data were filtered to include only locations which are evaluated in the NOCSAE standard--732 (19%) impacts for the youth data and 7043 (28%) impacts for the adult data were retained. In the filtered data, the on-field accelerations for the average youth player had a median of 19.3 ± 3.0 g and 95^{th} percentile of 44 ± 12.5 g, and the average adult player had a median impact of 19.9 ± 2.4 g and 95^{th} percentile of 57 ± 8.7 g. In comparing distributions of on-field acceleration data between youth and adult football players, a significant difference existed for the top location ($p < 0.001$). All other locations had no significant differences (Front, $p = 0.9997$, Front Boss, $p = 1$, Side, $p = 0.9945$, Rear Boss, $p = 1$, Rear, $p = 1$) (Figure 2).

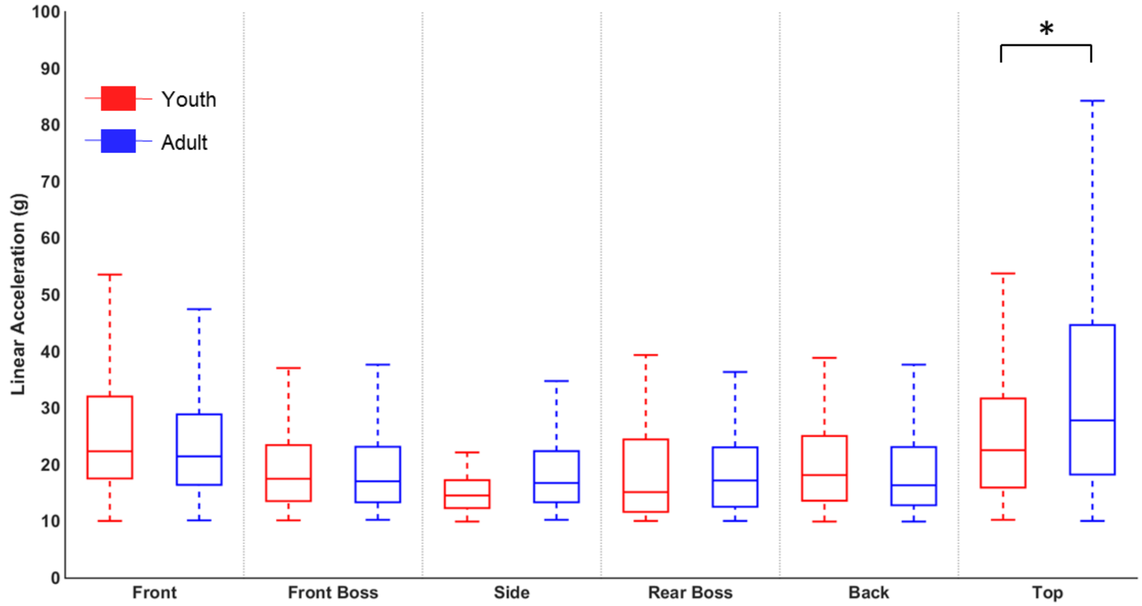

Figure 2: Distributions of on-field data in respect to NOCSAE locations for youth (black) and adult (grey) populations. The edges of the boxes are the 25th and 75th percentiles, with the middle representing the median. The whiskers represent 1.5 times the interquartile range. Significant differences were only found at the top location ($p < 0.0001$).

Drop tests yielded linear acceleration values between 44 and 125 g and SI values between 65 and 580, with the largest SI values occurring at the 5.46-m/s drop velocity in the top location. Acceleration and SI values generated by the youth helmet during drop testing were highly correlated to acceleration and SI values generated by the varsity helmet (Linear Acceleration, $y = 1.03x$, $R^2 = 0.95$) (SI, $y = 1.02x$, $R^2 = 0.97$). On average, the varsity helmet produced linear acceleration values 3% greater and SI values 2% greater than the youth helmet (Figure 3). Identical performance would result in a slope of 1 with a R^2 of 1. The only significant difference in acceleration between the youth and varsity helmets was at the 3.46-m/s impact to the front location ($p = 0.0421$) (Table 3). This same test condition also produced the only significant difference in SI ($p = 0.0256$).

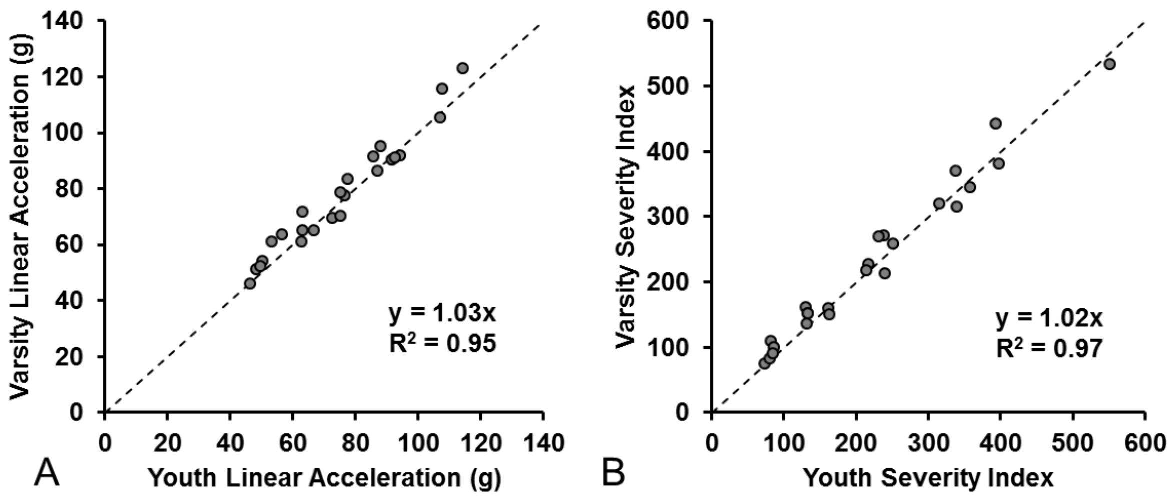

Figure 3: Comparison of (A) average linear acceleration and (B) average Severity Index for drop tests of a youth and varsity Riddell Speed helmet. Each point is a test condition with the dotted line representing identical performance of the youth and varsity helmet. On average, accelerations were 3% greater and Severity Index values were 2% greater in varsity helmets.

All accelerations measured in the drop tests were within the top 10% of impacts that were measured on-field for both youth and adult players (Table 3). Among all locations, the 3.46-m/s velocity ranged from the 93.1 percentile to the 98.9 percentile on-field acceleration for youth players and the 92.5 percentile to the 99.5 percentile on-field acceleration for adults. The 4.23-m/s velocity had accelerations in the top 3% of impacts for youth players and the top 4% for adult players across all locations. Among all locations for youth and adult players, the 4.88-m/s velocity resulted in accelerations that were within the top 2% of on-field head accelerations. The 5.46-m/s velocity produced accelerations that were within the top 0.5% of on-field head accelerations for all locations in youth and adult players, with one exception (adult top = 98.9 percentile).

Table 3: The equivalent on-field acceleration percentile was determined for each drop test acceleration for youth and varsity helmets. All the accelerations measured in the drop tests were within the top 10% of on-field accelerations in both youth and adult players. Significance was only found in the front location at the 3.46 m/s drop velocity (p = 0.0421).

	Drop Test		**On-Field**	
	Linear Acceleration (g)		**Percentile**	
	Youth	Varsity	Youth	Adult
Front				
3.46 m/s*	53 ± 2	61 ± 5	*93.1%*	*97.5%*
4.23 m/s	63 ± 4	72 ± 4	*97.5%*	*98.8%*
4.88 m/s	88 ± 4	95 ± 2	*99.3%*	*99.7%*
5.46 m/s	114 ± 2	123 ± 2	*99.9%*	*99.8%*
Side				
3.46 m/s	50 ± 2	52 ± 1	*98.4%*	*98.3%*
4.23 m/s	63 ± 1	65 ± 1	*99.4%*	*100.0%*
4.88 m/s	77 ± 1	78 ± 1	*100.0%*	*100.0%*
5.46 m/s	92 ± 5	91 ± 1	*100.0%*	*100.0%*
Rear				
3.46 m/s	48 ± 2	51 ± 3	*98.3%*	*96.0%*
4.23 m/s	67 ± 2	65 ± 3	*100.0%*	*98.4%*
4.88 m/s	77 ± 3	83 ± 7	*100.0%*	*99.0%*
5.46 m/s	94 ± 4	92 ± 1	*100.0%*	*99.5%*
Front Boss				
3.46 m/s	50 ± 1	54 ± 4	*98.9%*	*99.5%*
4.23 m/s	73 ± 11	70 ± 6	*100.0%*	*99.9%*
4.88 m/s	86 ± 5	92 ± 5	*100.0%*	*100.0%*
5.46 m/s	108 ± 5	116 ± 8	*100.0%*	*100.0%*
Rear Boss				
3.46 m/s	46 ± 3	46 ± 1	*94.5%*	*95.8%*
4.23 m/s	63 ± 9	61 ± 12	*97.5%*	*97.5%*
4.88 m/s	75 ± 7	70 ± 5	*98.6%*	*98.5%*
5.46 m/s	87 ± 7	86 ± 11	*99.6%*	*99.7%*
Top				
3.46 m/s	56 ± 3	64 ± 1	*94.9%*	*92.5%*
4.23 m/s	75 ± 4	79 ± 3	*99.0%*	*96.3%*
4.88 m/s	93 ± 3	91 ± 7	*99.9%*	*98.1%*
5.46 m/s	107 ± 1	105 ± 1	*100.0%*	*98.9%*

Discussion

This study related NOCSAE standard drop tests to on-field impacts measured by helmet-mounted accelerometer arrays for youth and adult football players. While some significant differences were identified, youth and varsity helmets did not differ greatly on the field or in the lab.

This study involved the on-field measurement of head impacts for youth and adult football players over the course of one season, where the adult population consisted of college-aged players wearing varsity Riddell Speed helmets and the youth population consisted of 9-11 year olds wearing youth Riddell Speed helmets. The on-field data collected are similar to what have previously been reported in these populations.[9, 11, 15] Distributions of on-field acceleration data defined by the NOCSAE locations proved to differ only at the top location. While the central tendency of the on-field accelerations did not differ greatly, 95th percentile accelerations experienced by youth (44 ± 12.5 g) and adult players (60.1 ± 7.8 g) did. This difference between the 95th percentiles of the on-field accelerations points to adult players experiencing higher magnitude accelerations more frequently than youth players. This difference is masked in the analysis relating on-field accelerations to drop tests accelerations because the majority of the NOCSAE test conditions produce accelerations in the top 2% of on-field impacts, where the tails of the distributions converge.

Matched drop tests on a youth and varsity Riddell Speed football helmet were also performed using the current standard. Both helmets have a similar design and padding, with the only difference being the shell material, where the varsity helmet shell is made of polycarbonate and the youth helmet shell is made of acrylonitrile butadiene styrene (ABS).[19] Both materials have good impact resistance properties, but ABS is more compliant, has a lower tensile strength, is lighter, and is less expensive than polycarbonate**.** Considering this difference in helmet shell material, the youth and varsity

Riddell Speed helmets had no significant differences in 23 of the 24 matched drop test conditions, which is the result of being tested to the same standard and of the interior padding dominating performance. This padding compresses and attenuates the impact energy transfer to the head. The shell is meant to deflect and reduce focal loading, as well as prevent penetration. The interior padding dimensions were identical between the two helmets.

All the drop tests yielded acceleration values that were above the 90th percentile for both of the on-field data sets. The 5.46-m/s velocity produced head accelerations in the top 0.5% of the on-field head impacts, with one exception. This is appropriate because a helmet standard should test for the highest-severity impacts experienced in actual play. In NOCSAE's proposed draft of a youth helmet standard,[2] more tests are being recommended at each location, resulting in a possible 18 additional tests compared to the current standard. Additionally, a lower threshold of 600 SI, for 4.88-m/s drops is proposed, along with the current threshold of 1200 SI for all tests. A mass limit of 1.3 kg, including all accessories, is also proposed for this standard. The youth helmet tested in this study would meet the impact criteria in the proposed standard, but would not pass the proposed mass limit (m = 1.68 kg).

Given that there are no notable differences in the real-world equivalents of NOCSAE drop-test configurations between youth and varsity helmets, this analysis makes the case that there is not a need for a youth-specific drop-test standard. The NOCSAE football helmet standard was developed and implemented to remove catastrophic head injuries from the game of football and has been very effective in doing so. Given that there is no evidence of differences in catastrophic head-injury rates between youth and adult football players, little specific benefit would be provided to youth football players with the proposed youth drop-test standard.[20] However, if NOCSAE develops a standard specific to concussion, there would likely be a need for a youth-specific football helmet testing standard. This need will become more evident as more is learned about the differences in concussion tolerances between youth and adult populations. Importantly, the analysis presented in this paper could be used to inform such a standard by incorporating rotational acceleration in both on-field and laboratory data collections.

This study was limited in several ways. The standards that have been developed have been specific to skull fracture and other severe brain injuries. The implementation of these standards has been very effective in mitigating these types of injuries. However, rotational kinematics, which are known to contribute to concussion risk,[21] are not included in the current standards and were not tested here. This lack of consideration of the rotational kinematics is a limitation in the current standards, and NOCSAE is developing future standards to include rotational kinematics to reflect real-world head impacts and their relation to injury. Rotational acceleration was measured in the on-field data using the accelerometer mounted arrays, and descriptive values were equivalent to those that have been previously reported for collegiate and youth players in this age group.[9, 11] Second, only the Riddell Speed helmet was considered, so it is unknown if helmet performance would differ with other models. Third, only one sample of each helmet was used in testing. While there may be variation between samples of the same model, this variance is typically very small. Fourth, the helmet-mounted accelerometer arrays used, on average, overestimate linear acceleration by 1% and overestimate rotational acceleration by 6%. This error is higher in individual measurements, but is reduced in aggregate analyses as presented here.[9, 22, 23] Fifth, college football players were used to represent our adult population in this study, although varsity football helmets are recommended for players over the age of 14. These data should describe the bounds of differences between the ages of 12 to 17, where players transition from youth to varsity helmet models.

Conclusions

Current NOCSAE helmet standards are identical for both youth and varsity helmets. In the on-field data, the only difference in linear head accelerations that was found was in the top location with respect to NOCSAE defined locations. Additionally, only one drop-test configuration was significantly different in the performance of matched drop tests for youth and adult Riddell Speed football helmets. These data demonstrate that drop tests are representative of the most severe head impacts experienced by youth and adult football players on the field. Standards have been very effective at eliminating skull fracture and severe brain injury in both populations. This analysis suggests that there is not currently a need for a youth-specific drop-test standard. However, there may be such a need if helmet testing standards are updated to address concussion, paired with differences in concussion tolerance between youth and adult populations being better understood.

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Chapter 3: Comparison of Impact Performance between Youth and Varsity Football Helmets

Abstract

Current youth football helmets, intended for players under the age of 14 years old, are similar in design and are tested to same standard as varsity football helmets. This study evaluated the impact performance of matched youth and varsity football helmets. Eight helmet models were evaluated using an impact pendulum, with a modified National Operating Committee on Standards for Athletic Equipment (NOCSAE) medium sized headform mounted on a Hybrid III 50^{th} percentile neck. Four locations on the helmet shell at three impact velocities were tested for three trials, for a total of 576 impact tests. Linear acceleration, rotational acceleration, and a concussion correlate were recorded for each test and comparisons between the youth and varsity helmets were made. It was found that the age group the helmet is intended for did not have a significant effect on impact performance in linear acceleration, rotational acceleration, or concussion correlate. These results are likely due to the similarities in helmet design, resulting from being tested to the same standard. Although it is unknown how a youth helmet should differ from a varsity helmet, differences in impact exposure, anthropometry, physiology, and injury tolerance are factors to consider. These data serve as a reference point for future youth-specific helmet design and helmet standards.

Introduction

Football has a high incidence of concussion due to the physicality of the sport, and because of its popularity, accounts for a large proportion of sports-related concussions.[1] Ongoing research is looking at reducing concussion incidence in football through a variety of interventions. These interventions can be divided into one of three categories: adjusting rules of the game, enforcing proper tackling techniques, and improving the design of helmets.[2, 3] This study focuses on the aspect of helmet design by evaluating the relative performance of youth and varsity football helmets.

Youth football players, defined as under the age of 14, make up approximately 70% of the participants in football nationwide.[4] Youth players typically see fewer impacts per season than high school and collegiate players, mostly due to participation in fewer games and practices.[2-7] The number of impacts a player experiences per season increases as they get older. Youth players will sustain between medians of 148 and 213 head impacts per season, depending on their age group,[2, 4, 5] whereas collegiate players will experience a median of 420 head impacts per season.[6] All age groups sustain high severity impacts, but older players will experience high severity impacts more frequently.[2, 5, 6, 8] From a rules perspective, some youth leagues do not perform plays that pose a high risk of injury, such as kickoffs and punts.[9]

Youth helmets are intended for players under the age of 14, and varsity helmets are meant for players age 14 and older. Youth helmets are typically similar to their varsity counterparts in both design and liner materials. However, varsity helmet shells are typically composed of polycarbonate, whereas youth helmet shells are typically composed of acrylonitrile butadiene styrene (ABS). Both materials are used for their impact resistive properties, but ABS is cheaper, lighter, more compliant, and has a lower tensile strength. There are additional youth-specific helmets without varsity equivalents that are available for purchase which may present additional differences. Currently, all helmets, both youth and varsity, must pass the same set of impact performance criteria based on the standards from the National Operating Committee on Standards for Athletic Equipment (NOCSAE).[10] A set of youth-specific standards has been put forth by NOCSAE for youth football helmets with additional tests and performance criteria.[11] We have previously related the NOCSAE standard test conditions to on-field head impact measurements and found that drop tests represent similar impacts for both age levels. Given that catastrophic head injuries have been nearly eliminated from football, our data suggest different standards would have little effect. Once more is known about youth concussion, standards likely will need to differentiate impact performance criteria for youth helmets.[12]

The objective of this study was to investigate differences in impact performance between matched youth and varsity football helmets. There are no data currently available relating the performance of youth and varsity helmets, with respect to linear and rotational head kinematics, both of which contribute to concussion risk.[13] It is hypothesized that due to the similarity in the youth and varsity helmets that there will be no differences in their impact performance. These data have applications of improving helmet design and helmet standards, specifically in regards to the youth population.

Methodology

Helmet models that had a matched youth and varsity version at the time of the study were used. A total of 8 models fit this criterion: Riddell 360 (360), Schutt Air XP Pro (AXP), Schutt DNA Pro+ (DNA), Rawlings NRG Impulse (IMP), Riddell Speed (SPD), Riddell Speedflex (SPDF), Schutt Vengeance DCT (VEN), and Xenith X2E (X2E). One varsity and one youth helmet were purchased for each of these models.

An impact pendulum was used to evaluate the performance of each helmet, because of its reproducibility and repeatability.[14] The pendulum arm was 190.5 cm long with a 16.3 kg impacting mass fitted with a flat nylon impactor face. The impactor struck a helmeted, medium sized NOCSAE headform (57.6 cm circumference) which had been modified to couple with a Hybrid III 50th percentile neck. The NOCSAE headform was selected as the headform shape provides a more realistic helmet fit compared to a Hybrid III headform.[15] The modified NOCSAE head and neck assembly has been shown to produce a similar impact response to the Hybrid III.[16] The head and neck assembly was mounted to a sliding mass on a commonly used linear slide table (Biokinetics, Ottawa, Ontario, Canada), where the sliding mass is intended to simulate the mass of the torso.[17] The headform was instrumented with 3 linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and a triaxial angular rate sensor (ARS3 PRO-18K, DTS, Seal Beach, CA) at the center of gravity of the headform, which allowed linear and angular kinematic measurements with 6 degrees of freedom. All data were sampled at 20 kHz and filtered using a 4 pole phaseless Butterworth filter. Acceleration data were filtered using a cutoff frequency of 1650 Hz (CFC 1000) and angular rate data were filtered using a cutoff frequency of 255 Hz, which best matched rotation accelerations measured from a 9 accelerometer array.

Each helmet was tested at front, back, side, and top impact locations (Figure 1). These locations encompass a variety of shell impacts that could be experienced during play.[2, 6, 18] Each impact location is based on translation from the zero point, defined by where the tip of the nose of the headform contacts the center of the pendulum impactor face, and rotation about the y and z axes (Table 1). The coordinate system used is defined by SAE J211. The x-translation is variable, measured where the pendulum impactor face just touches the helmet when hanging. For each impact location, impact velocities of 3.0, 4.6, and 6.1 m/s were tested. The impact velocities used are inclusive of the broad range of head acceleration magnitudes experienced by football players, including sub-concussive and concussive impacts.[14] Furthermore, helmets were tested without a facemask. We have previously shown that the facemask does not make a significant difference in head acceleration for impacts to football helmet shells.[19] Every impact scenario was repeated for 3 trials to yield a total of 576 tests. For each test, peak resultant linear acceleration, peak resultant rotational acceleration, and a concussion correlate value were computed.

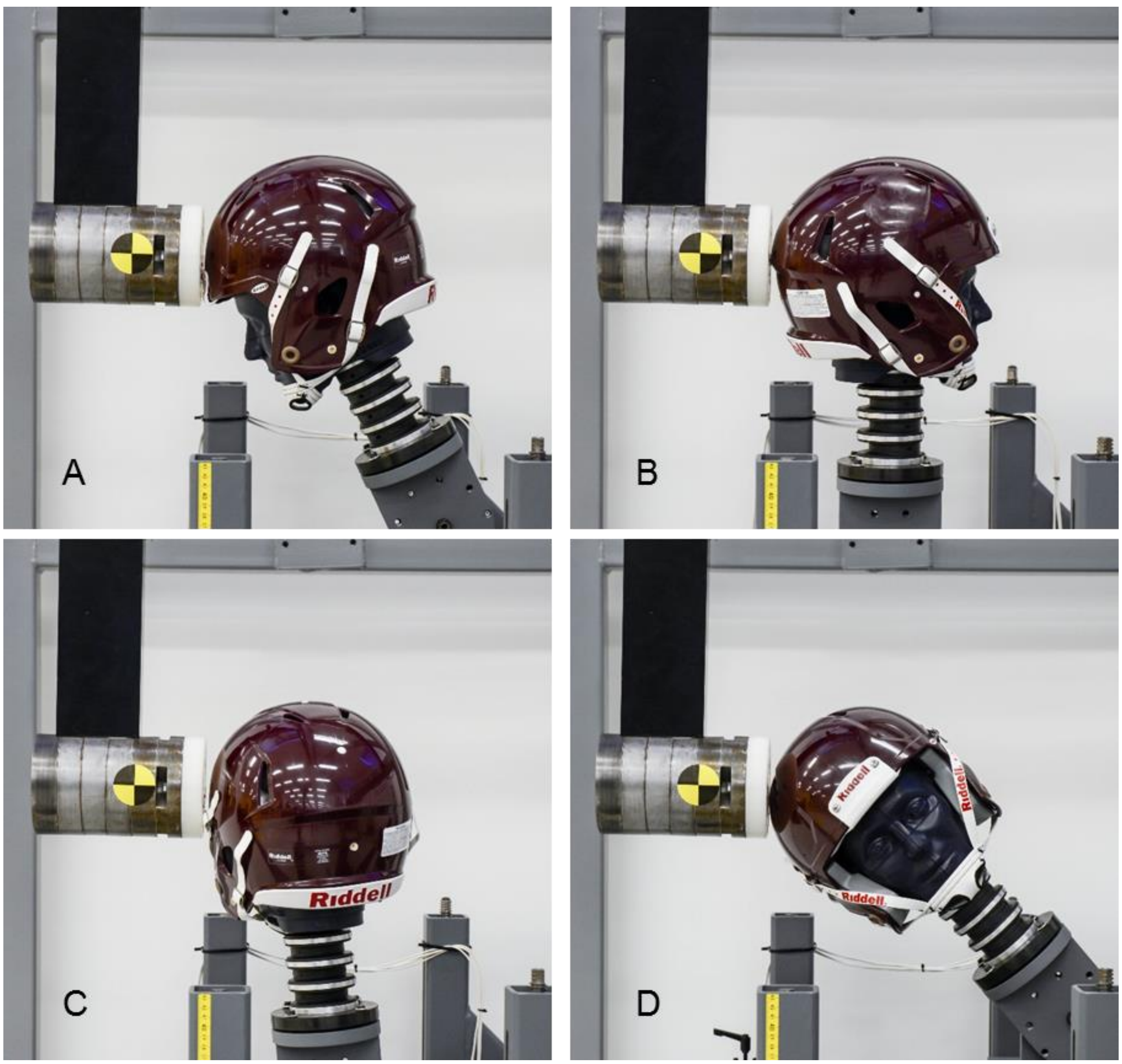

Figure 1: (A) Front, (B) back, (C) side, and (D) top locations were tested at 3.0, 4.6, and 6.1 m/s impact velocities. Matched youth and varsity helmets were tested for three trails of each impact scenario without a facemask.

Table 1: Measurements for locations based on the zero point defined by the tip of the nose contacting the center of the pendulum. Directions are based off the SAE coordinate system.

	Y (cm)	Z (cm)	Ry (^{0})	Rz (^{0})
Front	0	3.6	25	0
Back	0	-1.0	0	180
Top (right)	2.4	6.5	40	-90
Side (left)	-3.4	9.0	5	80

Concussion correlate was used to describe overall impact severity and is calculated using both linear acceleration (*a*) and rotational acceleration (α) (Equation 1). This metric can give both positive and negative numbers, with increasing values being indicative of increased impact severity, and has been shown to be a good predictor for AIS 2+ brain injury compared to other metrics.[20] The concussion correlate is based on an analysis of on-field head acceleration data collected from football players consisting of injurious and non-injurious head impacts.[13] Concussion correlate is used in place of the risk function here due to potential differences in youth and adult injury tolerance.

$$CC = -10.2 + 0.0433 \cdot a + 0.000873 \cdot \alpha - 0.00000092 \cdot a \cdot \alpha \qquad (1)$$

A three factor ANOVA with repeated measures was used to describe significant effects. Random effects were assumed for helmet model in the analysis. Differences were assessed using factors of age group, location, energy level, and all interaction terms. When necessary, Tukey HSD tests were used for post-hoc analysis. Statistical significance was set at $p < 0.05$ for all comparisons. Additionally, a linear regression model was fit to linear acceleration, rotational acceleration, and concussion correlate to characterize overall differences between youth and varsity helmets.

Results

The 3.0 m/s impacts generated an average linear acceleration of 37.9 ± 10.7 g, average rotational acceleration of 2036 ± 448 rad/s^2, and average concussion correlate of -6.85 ± 0.64. The 4.6 m/s impacts generated an average linear acceleration of 73.5 ± 15.0 g, average rotational acceleration of 3949 ± 924 rad/s^2, and average concussion correlate of -3.84 ± 1.10. For the 6.1 m/s impacts, averages were 114.3 ± 22.5 g for linear acceleration, 6117 ± 1684 rad/s^2 for rotational acceleration, and -0.57 ± 1.83 for concussion correlate. Across all energy levels and locations, the youth helmets produced an average linear acceleration of 74.7 ± 35.3 g, average rotational acceleration of 3996 ± 2017 rad/s^2, and average concussion correlate of -3.81 ± 2.88. These averages are comparable to the varsity helmets, where averages observed were 75.7 ± 35.7 g for linear acceleration, 4071 ± 2024 for rotational acceleration, and -3.71 ± 2.87 for concussion correlate.

The factor of helmet age group (youth vs. varsity) did not have a significant effect on linear acceleration ($p = 0.4768$), rotational acceleration ($p = 0.4714$), or concussion correlate ($p = 0.4351$). The differences between the youth and varsity helmets was insignificant across helmet models for linear acceleration, rotational acceleration, and concussion correlate (Figure 2). Matched differences were calculated by subtracting the average values for each impact configuration for the youth helmet from the matched varsity helmet values. Location had a significant effect on linear acceleration ($p = 0.0010$), rotational acceleration ($p = 0.0028$), and concussion correlate ($p = 0.0346$). The factor of energy level had a significant effect on all measurements ($p < 0.0001$). Additionally, the interaction of location and energy level had a significant effect on linear acceleration ($p = 0.0015$), rotational acceleration ($p < 0.0001$), and concussion correlate ($p < 0.0001$). The interaction between factors of age group and location had no significant effects on linear acceleration ($p = 0.0910$) or concussion correlate ($p = 0.2848$). A significant effect was found for the age group and location interaction for rotational acceleration ($p = 0.0382$), however post hoc tests found no significant effects when matching locations between age groups. For the age group and energy level interaction no significant effect was found on linear acceleration ($p = 0.1709$), rotational acceleration ($p = 0.6516$), or concussion correlate ($p = 0.3011$).

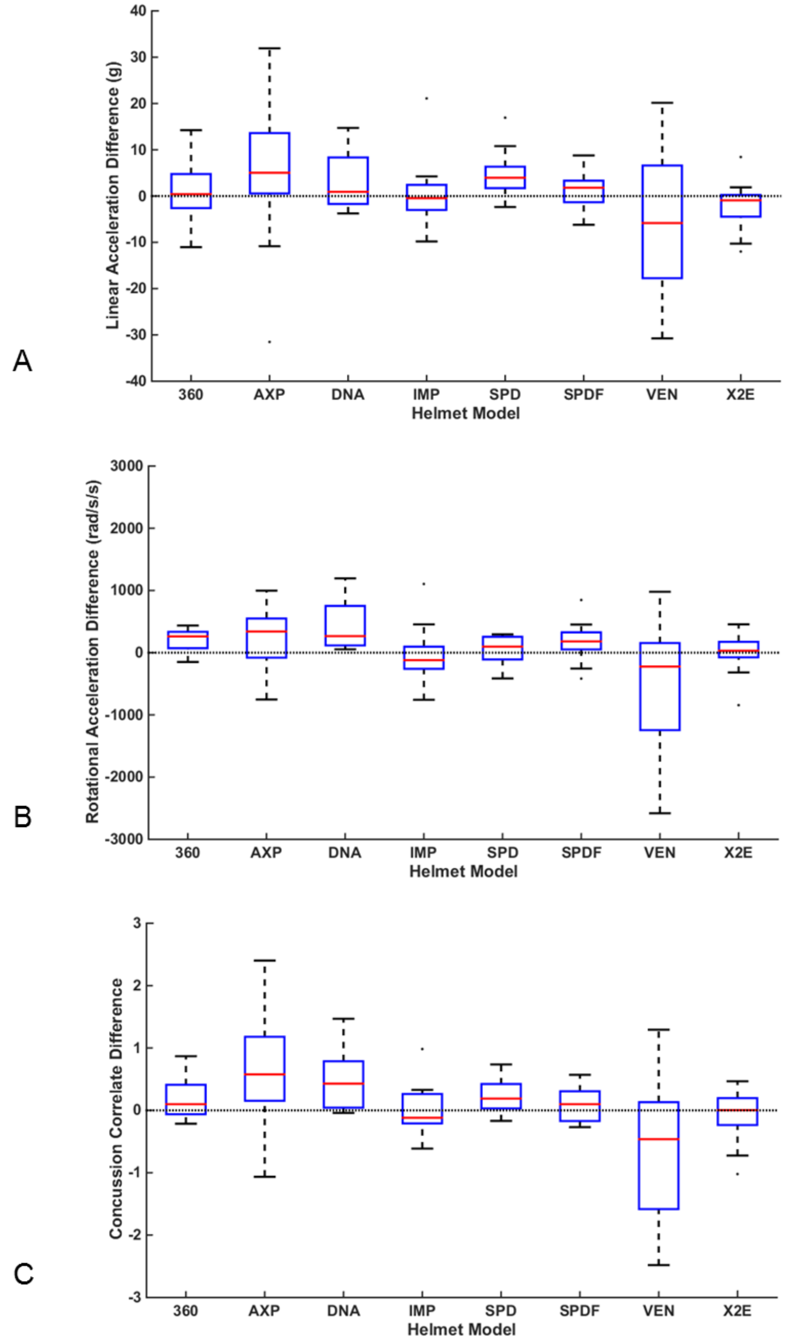

Figure 2: Distribution of matched differences for (A) linear acceleration (B) rotational acceleration and (C) concussion correlate, where differences were calculated for matched trials by subtracting the youth measurement from the varsity measurement. The middle of the box represents the median, with the edges giving the 25th and 75th percentiles. Whiskers represent 1.5 times the interquartile range.

The impact response for the youth and varsity helmets were highly correlated in linear acceleration, rotational acceleration, and concussion correlate. On average, linear acceleration for youth helmets were 98% that of varsity helmets ($R^2 = 0.9223$, $p < 0.0001$). Rotational acceleration, on average, for youth helmets was also 98% that of varsity helmets ($R^2 = 0.9235$, $p < 0.0001$). On average, concussion correlate for youth helmets was 101% of varsity helmets ($R^2 = 0.8961$, $p < 0.0001$) (Figure 3). For reference, a perfectly symmetrical data set would have a slope of 1, with an R^2 value of 1.

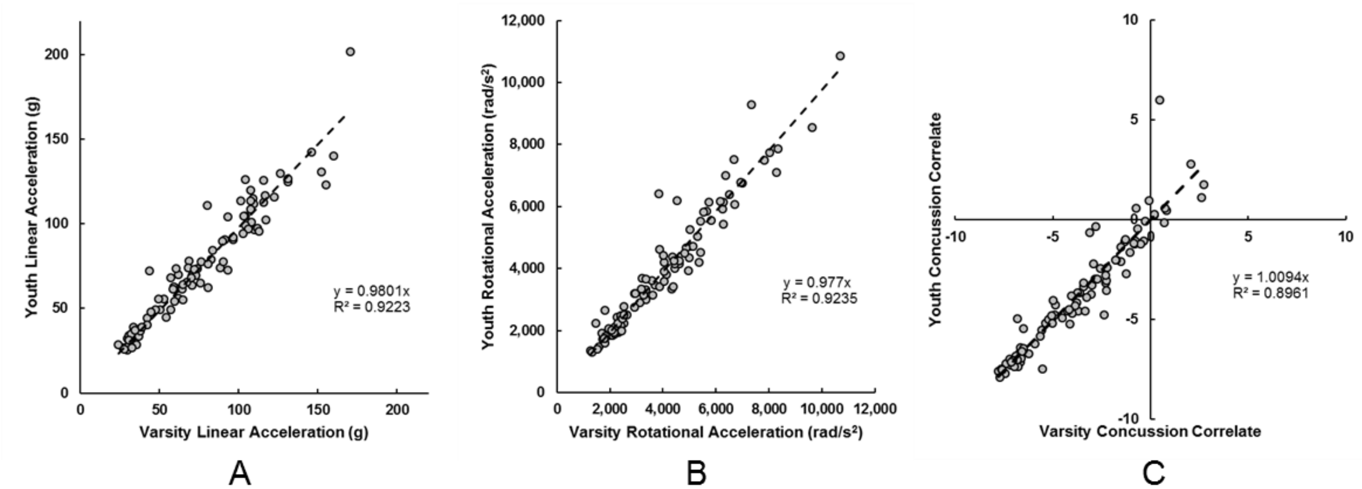

Figure 3: Comparison for matched youth and varsity helmets for (A) linear acceleration, (B) rotational acceleration, and (C) concussion correlate. This indicates that performance was highly correlated for matched helmets in that all lines of best fit have slopes near 1 and have high R^2 values.

Discussion

This study is the first to biomechanically compare the relative performance of matched youth and varsity football helmets. It was found that the youth and varsity helmets did not differ in impact performance. It should be noted however that youth and adult football players differ in impact exposure, anthropometry, and brain physiology. These differences are likely associated with differences in concussion tolerance.[21]

Differences between youth and varsity helmets were likely not observed due to matched helmets being similar in design and youth and varsity helmets being tested to the same standard. Age group was found not to have any effect on the impact performance in any of the helmet models tested (Figure 2). Some helmets had more variance in their matched differences. This can likely be attributed to the different energy mitigation strategies that different helmets employ, some of which are susceptible to more variance than others.

The design characteristics of a youth helmet are similar to the helmets used in the adult game, where the difference between matched models is the material used for the helmet shell. For each of the helmets used here, the varsity version used polycarbonate for the shell material and the youth version utilized ABS, with the exception of the Rawlings NRG Impulse which used polycarbonate for both models. The use of ABS in place of polycarbonate produces a youth helmet that is about 5% lighter than its varsity counterpart for these helmets. In the Rawlings and Schutt models the varsity helmets used a 7/8" jaw pad where the youth helmets use a 1 1/8" jaw pad. The jaw padding, however, is not impacted in any of our impact locations and these pads are interchangeable with different size pads for comfort. Otherwise, the liner dimensions were identical between matched helmets. The similar validation requirements for youth and adult football helmets have led to matched models that are very similar in design and performance.

The linear and rotational values recorded in this study are indicative of the range of on-field acceleration values for both youth and adult football players previously reported, where both see a similar range of acceleration values.[2, 4-6, 22] Additionally, these data include the range of accelerations in which concussions have been found to occur. For adult football players, numerous concussions have been recorded using helmet mounted accelerometer arrays, with the average injury occurring with average linear and rotational accelerations of 100 g and 5000 rad/s^2.[13, 23, 24]. Fewer concussions have been recorded in youth football, with injury measurements of 26 g and 1152 rad/s^2, 64 g and 2830 rad/s^2, 58 g and 4548 rad/s^2, and 95 g and 3148 rad/s^2.[2, 8] Further work needs to be done to gain a better understanding of the biomechanics of concussion in youth football, which will further inform youth-specific helmet design.

Although it is unknown how a youth helmet should differ from a varsity helmet, some considerations offer insight to the challenges of designing youth-specific helmets. Head mass and size only differ slightly between youth and adult players, as the head is already about 95% fully grown around three and a half years old. The head then fully matures to full adult size between the ages of 10 and 17.[25] A child's smaller body, however, means the head-to-body size ratio is much greater compared to a fully grown adult. Additionally, a child will have reduced strength and musculature in their neck and upper body.[26] There are concerns that youth players are more susceptible to concussion than adult players. It is still relatively unknown how concussion differs physiologically in the youth population, but some concerns include a developing nervous system, thinner cranial bones, differences in blood flow to the brain, and a larger subarachnoid space.[26, 27] Biomechanically, youth players may also have a lower concussion tolerance.[21] However, due to the limited sample of concussions in youth players, the differences in injury tolerance are not fully understood. Although older players experience more impacts per season, it is important to note that younger players do still experience high magnitude head impacts, just at a lower frequency compared to older players.[2, 3, 5]

This study was limited in several ways. First, performance differences of only matched youth and varsity helmets were investigated. Although no differences were found, it is unknown how the performance of helmets without youth or varsity counterparts may differ. Second, the test setup used here was specific to an average adult male and may not best represent the impact response of a youth player. These differences were not considered in the current study in order to make an effective comparison in helmet performance. Third, concussion correlate was calculated for both the youth and varsity helmets although this injury metric was developed using data from collegiate football players.[13] This measurement is still useful as it provides a severity summary value that considers both linear and rotational acceleration. It is unknown how the injury tolerance of youth players differ from adult players, and no youth specific injury metrics have been developed.

Conclusions

To evaluate the relative biomechanical performance of youth and varsity football helmets, eight helmet models with matched youth and varsity versions were evaluated through a series of impact tests using a pendulum impactor. No differences were found between the youth and varsity helmets in these impact tests, likely due to the similarity in design between helmets and both being tested to the same standard. It is currently unknown how youth and varsity helmets should differ. These data serve as a reference point for future youth-specific football helmet design and standards.

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Chapter 4: STAR for Youth and Varsity Football Helmets: Characterizing Helmet Performance Using Linear and Rotational Acceleration

Abstract

The STAR evaluation system was developed to differentiate performance between football helmets in efforts to inform consumers on relative impact performance. This system is intended to identify helmets that best reduce head acceleration during impact. STAR was developed through relating on-field head impact exposure to a series of laboratory tests, each weighted based on how frequently they occur on the field. The objective of this study was to advance STAR for varsity football helmets and to develop a new methodology for evaluating youth football helmets. A youth test surrogate was developed through scaling an existing adult surrogate using anthropometric data for 10-12 year old boys. A comparison between the adult and youth surrogates was conducted under the STAR test conditions, consisting of impacts to front, front boss, side, and back locations at 3.0, 4.6, and 6.1 m/s impact velocities. Differences were found in impact response between the youth and adult surrogates, although these differences were small in magnitude. Bivariate cumulative distributions of linear and rotational head acceleration collected from youth and varsity football players were used to relate STAR impact conditions to head impact exposure for each population. The STAR impact protocol was performed on two matched youth and varsity football helmets to provide a comparison of population-specific methodologies. Due to the impact exposure of the youth players, the STAR values for youth football helmets had an inherently smaller STAR value, and the lower severity impacts had a higher relative weighting compared to the varsity helmets. This comparison demonstrated that youth-specific helmets do differ in performance from one another, and the STAR evaluation can identify those differences. The methodologies

have limitations, however, by giving objective measures based on linear and rotational head kinematics to differentiate helmet performance, valuable information can be provided to direct consumers to safer helmets.

Introduction

Currently, football helmets are produced as either varsity or youth, in which varsity helmets are recommended for players over the age of 14 and youth helmets for those players below. Youth and football helmets are designed similarly and matched helmets do not differ in impact performance (Chapter 3). Both varsity and youth helmets must meet the same set of performance criteria put forth by the National Operating Committee on Standards for Athletic Equipment (NOCSAE).[1] This standard implements a pass/fail criteria to prevent catastrophic injuries in football, and has essentially eliminated these types of injuries from the game.[2] Youth specific standards have been proposed by NOCSAE,[3] but are likely unnecessary until the standards are adapted to be more specific to concussion.[4] While all helmets that meet NOCSAE's performance criteria are given a seal of approval, some helmets attenuate impacts more effectively than others.[5]

It was this observation that led to the development of the STAR evaluation system to compare relative football helmet performance. The STAR system related on-field impacts to a series of laboratory impact tests and characterized performance using two fundamental principles: 1) helmets that lower acceleration will reduce concussion risk and 2) each laboratory impact is weighted based on how frequently a player would experience that impact on the field. The STAR system was derived from on-field impact data collected from collegiate football players instrumented with helmet-mounted accelerometer arrays. The laboratory tests were based on the NOCSAE standard drop tests.[6] The impact performance of all available varsity football helmets has been characterized and disseminated to the public using a five star scale to rate each helmet. This was the first effort to provide consumers with knowledge of the relative performance of different football helmets, and directly affected the market. A study comparing on-field concussion rate between two helmets supported the findings of STAR, showing a better rated helmet was associated with a lower concussion rate than a lower rated helmet.[7] While STAR advanced helmet design, it did not consider rotational acceleration, which is an important factor for concussion.[8, 9]

Building on the football methodology, STAR was later adapted to hockey helmets.[10] STAR for hockey evaluated linear and rotational head acceleration using a headform mounted to a biofidelic neck. Using data collected directly from hockey players[11-13], as well as dummy impact tests performed in a hockey rink,[10] an impact pendulum was developed to replicate impacts experienced by hockey players. Both linear and rotational acceleration resulting from impacts were used to characterize concussion risk.[9] This evaluation system is more representative of real world head impacts and demonstrated a wide range of performance in existing hockey helmets.

The purpose of this investigation was to advance the STAR methodology for football helmets by incorporating rotational acceleration to evaluate varsity and youth helmets. This methodology evaluates linear and rotational head acceleration resulting from a series of pendulum impacts to an adult or youth surrogate. The youth and adult methodologies are based on head impact data collected from each population.

Methodology

Laboratory Impact System

The adult surrogate (Figure 1) previously developed for evaluating hockey helmets[10] was used to advance the impact methods of STAR for varsity football helmets. This includes a medium sized NOCSAE headform adapted to fit a Hybrid III 50th percentile neck.[14] The NOCSAE headform was selected as it provides a more realistic helmet fit than the Hybrid III headform and provides a similar impact response.[14, 15] The head and neck assembly is mounted on a sliding mass which simulates the effective mass of a torso.[16] This sliding mass is constructed from steel and has a mass of 16 kg. The headform is instrumented with 3 linear accelerometers and a triaxial angular rate sensor at its center of gravity to measure linear acceleration and rotational velocity.

The adult surrogate was scaled to develop a youth-specific surrogate (Figure 1). The relative masses for each surrogate are given in Table 1. The red NOCSAE headform was chosen to emulate a youth player because majority of youth players wear size medium football helmets. The human head is about 95% full size by three and a half years old, and gradually grows to full adult size between the ages of 10 and 16.[17, 18] This 5% difference is equivalent to the difference between a large and medium sized helmet, where a large helmet is intended for head circumferences between 55.9 and 59.7 cm, and a medium helmet is intended for circumferences between 51.8 and 55.9 cm. A 50th percentile 10 year old has a head circumference of 53.3 cm and a 50th percentile 12 year old has a head circumference of slightly above this at 53.9 cm, compared to a full grown 50th percentile male head circumference of 55.9 cm.[19] The red NOCSAE headform, meant for medium sized helmets, has a circumference of 53.4 cm whereas the blue NOCSAE headform, intended for large sized helmets, has a circumference of 57.6 cm (Table 1).[20]

Youth athletes have weaker necks compared to their adult counterparts.[21-26] The Hybrid III 50th percentile neck was used in the adult surrogate,[10] and is considered biofidelic.[27] Using scaling techniques, necks have similarly been developed for a 10 year old dummy and 5th percentile female dummy, where the 5th percentile female has a similar stature as a 12 year old adolescent.[26] The 10 year old neck is molded butyl rubber with a center cable (mass = 0.80 kg), whereas the 5th percentile female neck is constructed of butyl rubber, segmented by aluminum discs with a center cable to limit strain on the rubber (mass = 0.91 kg). For the surrogate, the 5th percentile female neck was selected as it more closely resembles the 50th percentile neck (mass = 1.54 kg) (Table 1) in its design (Figure 1).[26, 28]

The sliding mass that the head and neck assembly is mounted to is intended to simulate the mass of the upper torso.[16] The sliding torso mass of the adult surrogate is 16 kg and constructed of steel. Youth players are between ages 6 and 14, representing a wide range of body masses. Using CDC growth charts, a 50th percentile 6 year old boy has a total body mass of 21.0 kg and a 50th percentile 14 year old boy has a total body mass of 51.0 kg. If you average these two extremes you get a mass of 36.0 kg. Applying the ratio of a 50th percentile male to the adult surrogate torso, a 8 kg sliding mass was selected, representative of a 38.1 kg body mass (Table 1). The 8 kg sliding mass corresponds to a 38.1 kg simulated total body mass, which is between a 50th percentile 11 year old and 50th percentile 12 year old (Table 1). To decrease the mass, the sliding mass was built out of aluminum instead of steel. (Figure 1)

Figure 1: The adult surrogate (left) features a blue NOCSAE headform, 50^{th} percentile Hybrid III neck, and 16 kg steel sliding mass. The youth surrogate (right) was scaled to include a red NOCSAE headform, 5^{th} percentile Hybrid III neck, and 8 kg aluminum sliding mass.

Table 1: Representative masses of the adult and youth surrogates. The adult body mass is based off the 50th percentile Hybrid III dummy and the youth body mass is the equivalent of an 11 or 12 year old male. Each body mass is scaled to give a sliding mass estimate of the effective upper torso mass. The head circumference, head mass, and neck size are based off the specifications for the corresponding components of each surrogate. [20, 28, 29]

	Adult Surrogate	Youth Surrogate
Simulated Body Mass (kg)	77.7	38.1
Head Circumference (cm)	57.6	53.4
Head Mass (kg)	4.9	4.12
Neck Mass (kg)	1.54	0.91
Sliding Mass (kg)	16	8

An impact pendulum was used to simulate head impacts to evaluate youth and varsity football helmets. The pendulum arm, constructed of aluminum, is 190.5 cm, with a 16.3 kg impacting mass connected to its end. The pendulum arm has a total mass of 36.3 kg and moment of inertia of 72 kg m^2. For football testing, a hemispherical nylon impactor face 12.7 cm in diameter was used to represent the shell geometry, simulating helmet-to-helmet impacts. This impacting face is rigid, as the use of a compliant impactor face potentially masks differences in comparisons of helmet performance.[30] A pendulum impactor is used for the STAR testing because it has excellent repeatability and reproducibly, which is important when comparing helmet performance.[16]

Comparison of Impact Response

The impact response of the youth and adult surrogates were compared. The pendulum was used for impacts to the front, front boss, side, and back locations with impact velocities of 3.0, 4.6, and 6.1 m/s. These locations and velocities were selected as they encompass a wide range of impacts that players might experience on the field.[31-38] Locations on each headform were matched between surrogates to impact similar points on each headform (Figure 2, Table 2). Both surrogates were evaluated through four trials for each impact configuration without a helmet. In place of the rigid hemispherical impactor face, a 40 mm vinyl nitrile pad was used as the impactor face. This vinyl nitrile pad acts to simulate a helmet that provides identical characteristics for each surrogate. Testing with helmets could introduce variability, as different helmet sizes would be required for each surrogate.

Data were collected from 3 linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and a triaxial angular rate sensor (ARS3 PRO-18K, DTS, Seal Beach, CA) at 20 kHz and filtered using a 4 pole phaseless Butterworth filter, with a cutoff frequencies of 1650 Hz (CFC 1000) for linear data and 255 Hz for angular rate data. Additionally, the pendulum was instrumented with a linear accelerometer to measure the pendulum's change in velocity during impact. Impact response was characterized using peak resultant linear acceleration, change in linear velocity, peak resultant rotational acceleration, change in rotational velocity, change in pendulum velocity, and impact duration. At each energy level, two factor ANOVAs were conducted for each response variable investigating the effects of surrogate type and location. Post-hoc Tukey HSD tests were conducted to compare the youth and adult surrogates in each impact scenario. A significance level of $p < 0.05$ was used for all comparisons.

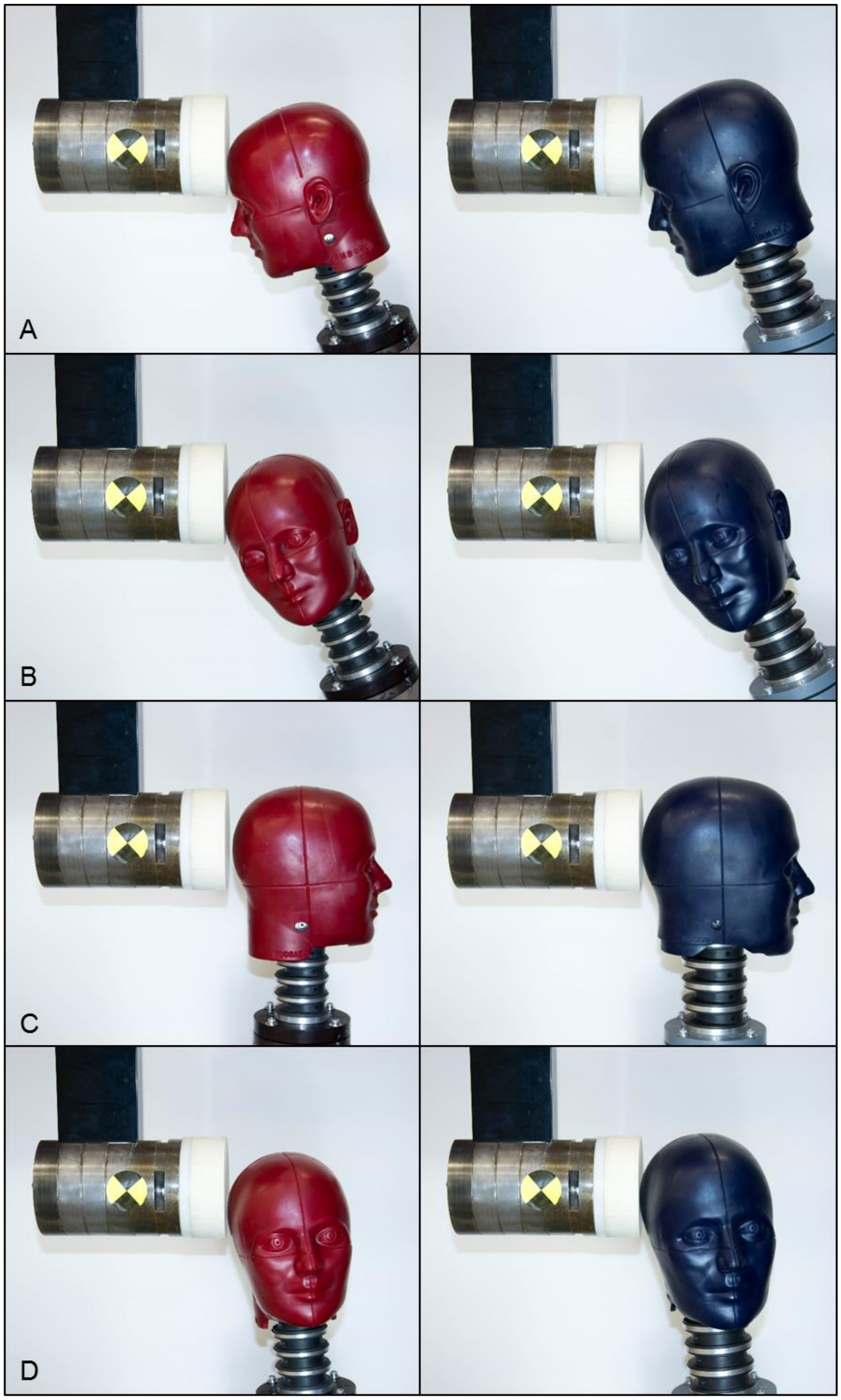

Figure 2: To compare the impact response between the youth (left) and adult (right) surrogates, impacts were conducted with a 40 mm vinyl nitrile pad to a bare headform with impact velocities of 3.0, 4.6, and 6.1 m/s at (A) front, (B) front boss, (C) back, and (D) side locations.

Table 2: Locations used in the STAR evaluation systems for youth and varsity football helmets. Measurement directions are based on the SAE J211 coordinate system. The reference point for each surrogate is where the tip of nose is in contact with the center of the curved pendulum impactor face.

	Adult			
	Y (cm)	Z (cm)	Ry (^{0})	Rz (^{0})
Front	0	6.5	20	0
Front Boss	0	4	25	67.5
Back	0	6.5	0	180
Side	-4	7.5	5	100
	Youth			
	Y (cm)	Z (cm)	Ry (^{0})	Rz (^{0})
Front	0	5.5	20	0
Front Boss	0	4.5	25	67.5
Back	0	6	0	180
Side	-4	8	5	100

Mapping of On-Field Exposure

Bivariate empirical cumulative distribution functions (CDF) of linear and rotational acceleration were determined from on-field data collected from football players wearing varsity and youth helmets. For all games and practices in one full season, linear and rotational acceleration were collected using helmet mounted accelerometer arrays. Any impact resulting in an acceleration over a 10 g threshold was included in the dataset. Data were collected from 25 youth players between the ages of 9 and 11 wearing Riddell Speed youth helmets and 48 collegiate players between the ages of 18 and 22 wearing Riddell Speed varsity helmets.[4] Bivariate CDFs were developed for youth and varsity players. STAR impact conditions were used to produce linear and rotational head accelerations for the adult surrogate equipped with a Riddell Speed varsity helmet and the youth surrogate equipped with a Riddell Speed youth helmet. Each impact condition was tested 4 times and then averaged. The bivariate CDFs were used to relate measured laboratory accelerations for each impact condition to on-field acceleration percentiles. Exposure weightings for each impact energy were determined by averaging the accelerations between locations and multiplying the percentage of impacts in the CDFs at or below that acceleration by the number of impacts a youth or collegiate player sustains per season. Weightings were rounded to the nearest non-zero integer.

Summarizing Helmet Performance

Both the STAR for youth football helmets and STAR for varsity football use the same equation used in STAR for hockey helmets (Equation 1). In this equation, E represents the exposure as a function of location (L) and impact velocity (V), and R represents the risk of concussion, as a function of linear (a) and rotational (α) acceleration. Locations impacted in STAR for youth and varsity football helmets are front, front boss, side, and back. Exposure as a function of location and energy is sport and population-specific.

$$STAR = \sum_{L=1}^{4} \sum_{\theta=1}^{3} E(L,V) * R(a,\alpha) \qquad (1)$$

The risk function for concussion (Equation 2) considers both linear (a) and rotational acceleration (α) and is derived from a multivariate logistic regression analysis of head acceleration data from collegiate football players.[9] While risk has not been characterized for youth players, it is still a valuable injury metric that summarizes linear and rotational acceleration into a single value that can be used to identify helmets that best reduce head acceleration. As research progresses, a youth specific risk function can be developed and implemented.

$$R(a,\alpha) = \frac{1}{1+e^{-(-10.2+0.0433 \cdot a+0.000873 \cdot \alpha-0.00000092 \cdot a\alpha)}} \qquad (2)$$

The STAR evaluation system protocol calls for each helmet to be tested twice at each impact velocity (3.0, 4.6, and 6.1 m/s) and location (front, front boss, back, and side). The acceleration values for each test are to be averaged to for each impact scenario. Two helmets of each model are to be evaluated and their STAR values averaged to produce a final STAR value.

Demonstrative STAR tests

STAR tests were performed on two matched youth and varsity helmet models. Both youth helmets were size medium, and both varsity helmets were size large. The youth helmets were tested using the youth-specific surrogate and weightings, and the varsity helmets were tested using the adult surrogate and weightings. Matched front, front boss, back, and side locations were tested at 3.0, 4.6, and 6.1 m/s impact velocities. Two trials were performed at each impact location and averaged. A STAR value was then determined for each helmet.

Results

Comparison of Impact Response

Surrogate type had a significant effect on peak linear acceleration (all impact velocities, $p < 0.0001$) (Figure 3). The youth surrogate had significantly higher linear accelerations at the front ($p = 0.0303$) and back ($p = 0.0199$) for the 3.0 m/s impact, front ($p < 0.0001$) and back ($p < 0.0001$) for the 4.6 m/s impact, and back ($p < 0.0001$) for the 6.1 m/s impact. These differences ranged from 4.7 to 7.0 g with an average of 5.9 g between the youth and adult surrogate for the locations that were significant. For each energy level, matched differences were calculated by subtracting the averages across locations for the adult surrogate from the averages across locations for the youth surrogate. All locations at all energy levels were found to differ between the youth and adult surrogate for linear velocity ($p < 0.0001$), where on average, the youth surrogate produced 0.24 m/s higher linear velocities for matched locations, with a range of 0.12 to 0.32 m/s (Figure 3).

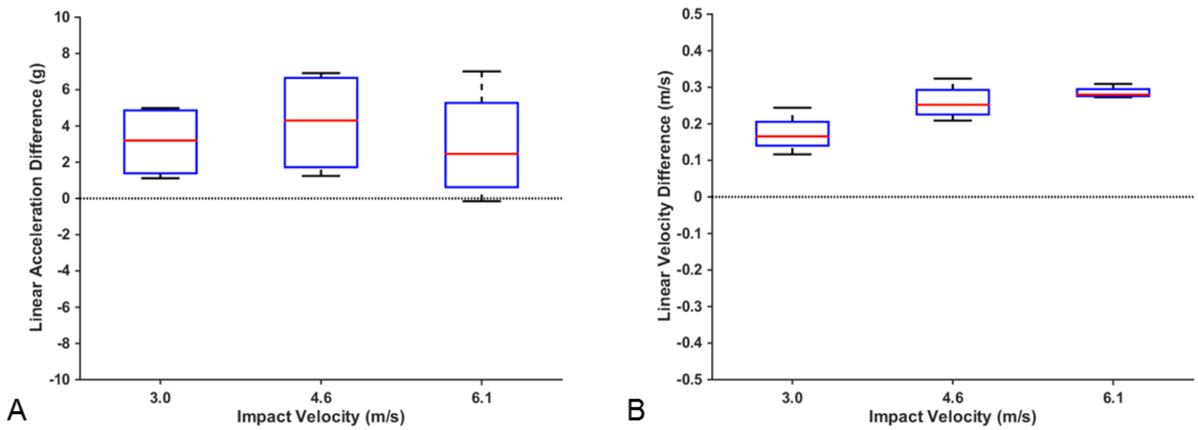

Figure 3: Matched differences between youth and adult surrogates for (A) linear acceleration and (B) linear velocity. For both linear acceleration ($p < 0.0001$) and linear velocity ($p < 0.0001$), the surrogate had a significant effect for all energy levels.

Surrogate type was found to have a significant effect on rotational acceleration ($p < 0.0001$) (Figure 4). The youth surrogate had significantly higher values for the back ($p = 0.0274$) in the 3.0 m/s impacts, the front ($p = 0.0011$), side ($p < 0.0001$), and back ($p < 0.0001$) in the 4.6 m/s impacts, and the front ($p = 0.0011$), side ($p < 0.0001$), and back ($p < 0.0001$) in the 6.1 m/s impacts. For the impact scenarios in which significant differences were found, there was an average difference of 496 rad/s^2, and ranged from 335 to 676 rad/s^2. Rotational velocity was higher for the youth surrogate for all impact conditions ($p < 0.0001$). These location-specific differences ranged from 1.9 to 5.1 rad/s, with an average difference of 3.3 rad/s (Figure 4).

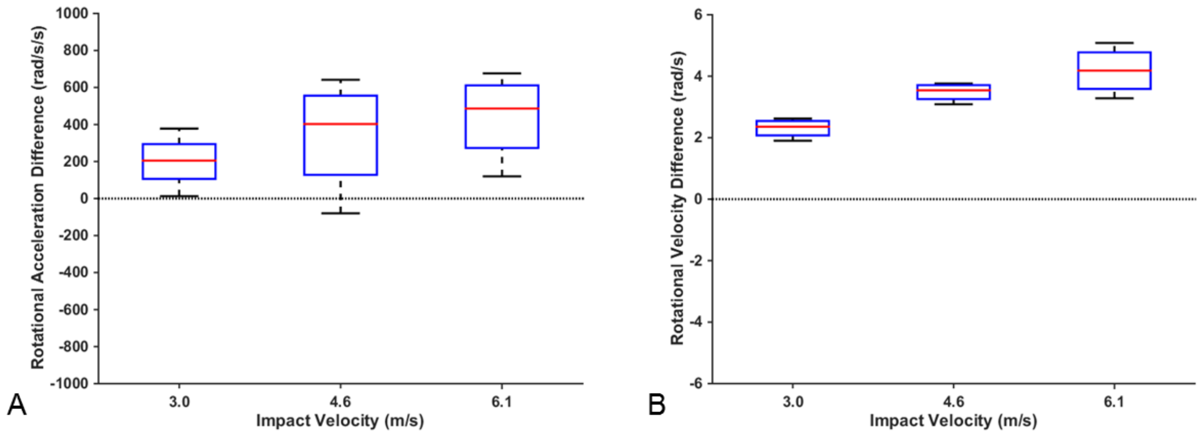

Figure 4: Matched differences between youth and adult surrogates for (A) rotational acceleration and (B) rotational velocity. For both rotational acceleration (3.0 m/s, $p = 0.0009$, 4.6 m/s, $p < 0.0001$, 6.1 m/s, $p < 0.0001$) and rotational velocity ($p < 0.0001$), the surrogate type had a significant effect for all energy levels.

A significant effect on impact duration was found for the 4.6 m/s ($p = 0.0169$) and 6.1 m/s ($p < 0.0001$) (Figure 5). Significant differences were found in the front boss location for the medium ($p < 0.0001$) and high ($p < 0.0001$) energy levels with the youth surrogate having an average of 1.4 ms longer duration, with a range from 1.3 to 1.5 ms. Pendulum change in velocity was significantly higher for the adult surrogate for all tests ($p < 0.0001$), with the differences ranging from 0.08 to 0.32 m/s with an average of 0.18 m/s.

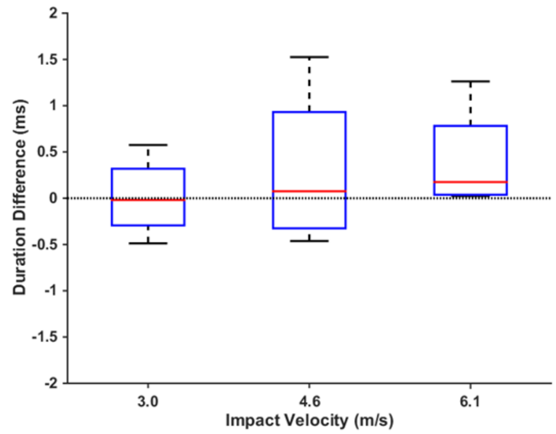

Figure 5: Matched differences between youth and adult surrogates for impact duration. No significant effects for surrogate type were found at 3.0 m/s impact velocity ($p = 0.9489$), but effects were found at 4.6 m/s ($p = 0.0169$) and 6.1 m/s ($p < 0.0001$) impact velocities. The magnitude of these differences however, were relatively small.

The magnitude of most of these differences were relatively small. Overall, the youth and adult surrogates produced time series acceleration responses that were similar between the two. Additionally, the tests were repeatable through trials, as there is little variation in the response corridors (Figure 6). The response corridors were developed by determining the maximum and minimum value at each time point across each of the trials.

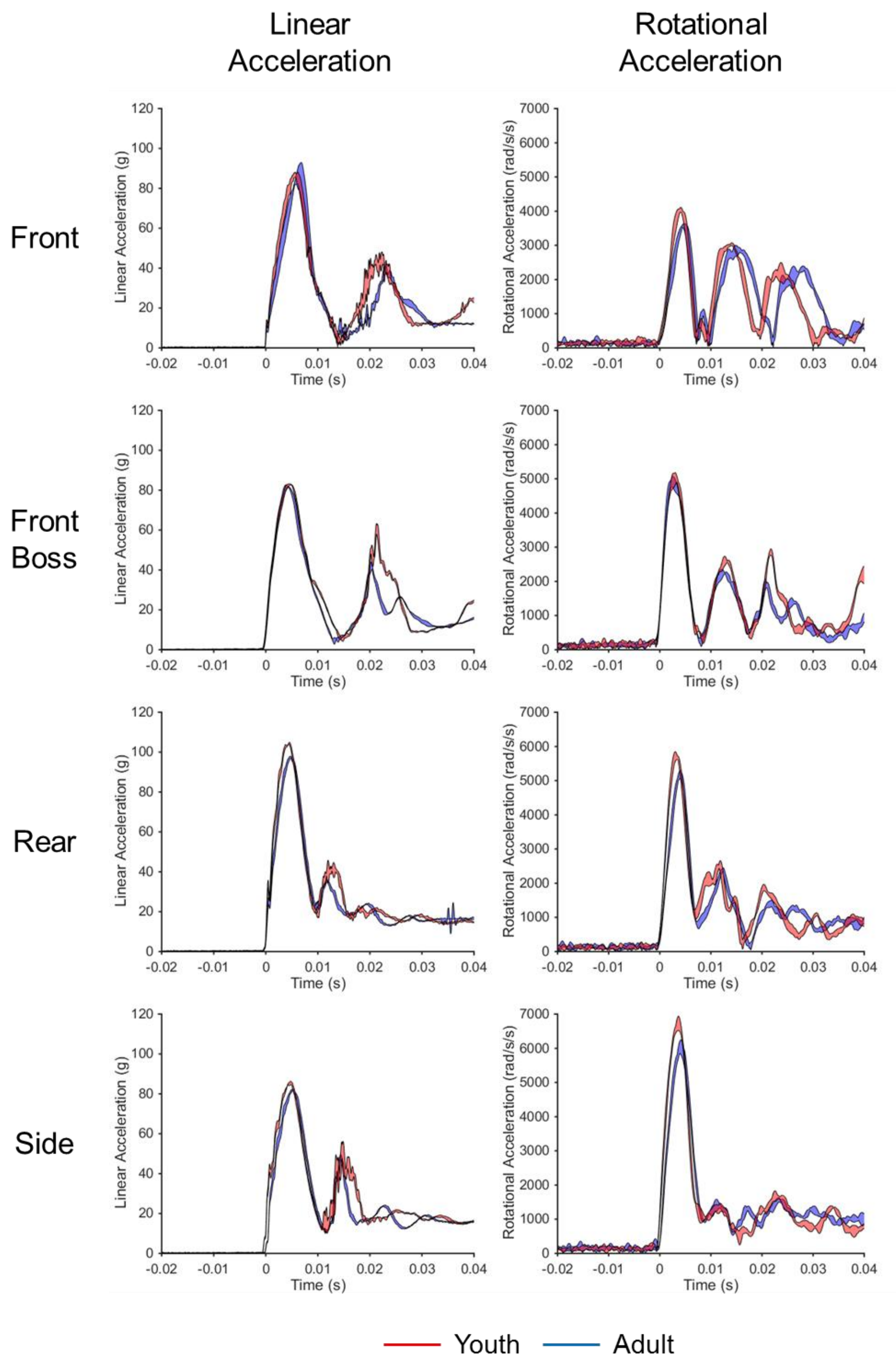

Figure 6: Response corridors for each location at the 6.1 m/s impact velocity. Corridors were developed taking the maximum and minimum value from each time point using all four trials. The youth surrogate is in red and the adult surrogate is show in blue, with similar responses evident between the two.

Mapping of On-Field Exposure

A total of 3842 head impacts were included in the dataset from the youth players. The average player in this youth dataset had median accelerations of 17.7 ± 2.2 g and 867 ± 116 rad/s^2, and 95th percentile accelerations of 48.8 ± 17.8 g and 2563 ± 1058 rad/s^2. In the collegiate dataset, 25,322 impacts were included. The average player had median accelerations of 20.1 ± 4.5 g and 911 ± 147 rad/s^2, and 95th percentile accelerations of 59.3 ± 10.9 g and 2487 ± 570 rad/s^2. The bivariate CDF for both datasets were produced and are shown with 50th, 75th, 90th, 95th, and 99th percentile contour lines (Figure 7). Additionally, acceleration values from the tests with the youth and varsity Riddell Speed helmet are overlaid onto the corresponding CDFs to illustrate the on-field acceleration percentile equivalents of laboratory impacts.

For the laboratory tests of the youth Riddell Speed helmet, averages between locations for the 3.0 m/s impacts were 43.5 g and 2803 rad/s^2, corresponding to a 92.4 percentile impact. For the 4.6 m/s impact, the averages of 75.2 g and 4353 rad/s^2 were equivalent to a 98.9 percentile impact, and the 6.1 m/s impact had averages of 131.5 g and 7667 rad/s^2 corresponding to a 99.9 percentile impact. Comparatively, the varsity Riddell Speed helmet averaged 35.7 g and 2147 rad/s^2 (79.3 percentile) for the 3.0 m/s impact, 65.5 g and 4015 rad/s^2 (96.4 percentile) for the 4.6 m/s impact, and 108.7 g and 6273.4 rad/s^2 (99.6 percentile) for the 6.1 m/s impact. These percentiles were multiplied by the average number of impacts a player in each age group would experience in one season (youth = 240, collegiate = 420), [33, 34] and divided by four to determine a weighting for individual test conditions. For youth, exposure mappings to impact velocities for each location were 55 impacts for 3.0 m/s impacts, 4 for 4.6 m/s impacts, and 1 for 6.1 m/s impacts. For varsity, exposure mappings to impact velocities for each location were 83 for 3.0 m/s impacts, 18 for 4.6 m/s impacts, and 4 for 6.1 m/s impacts.

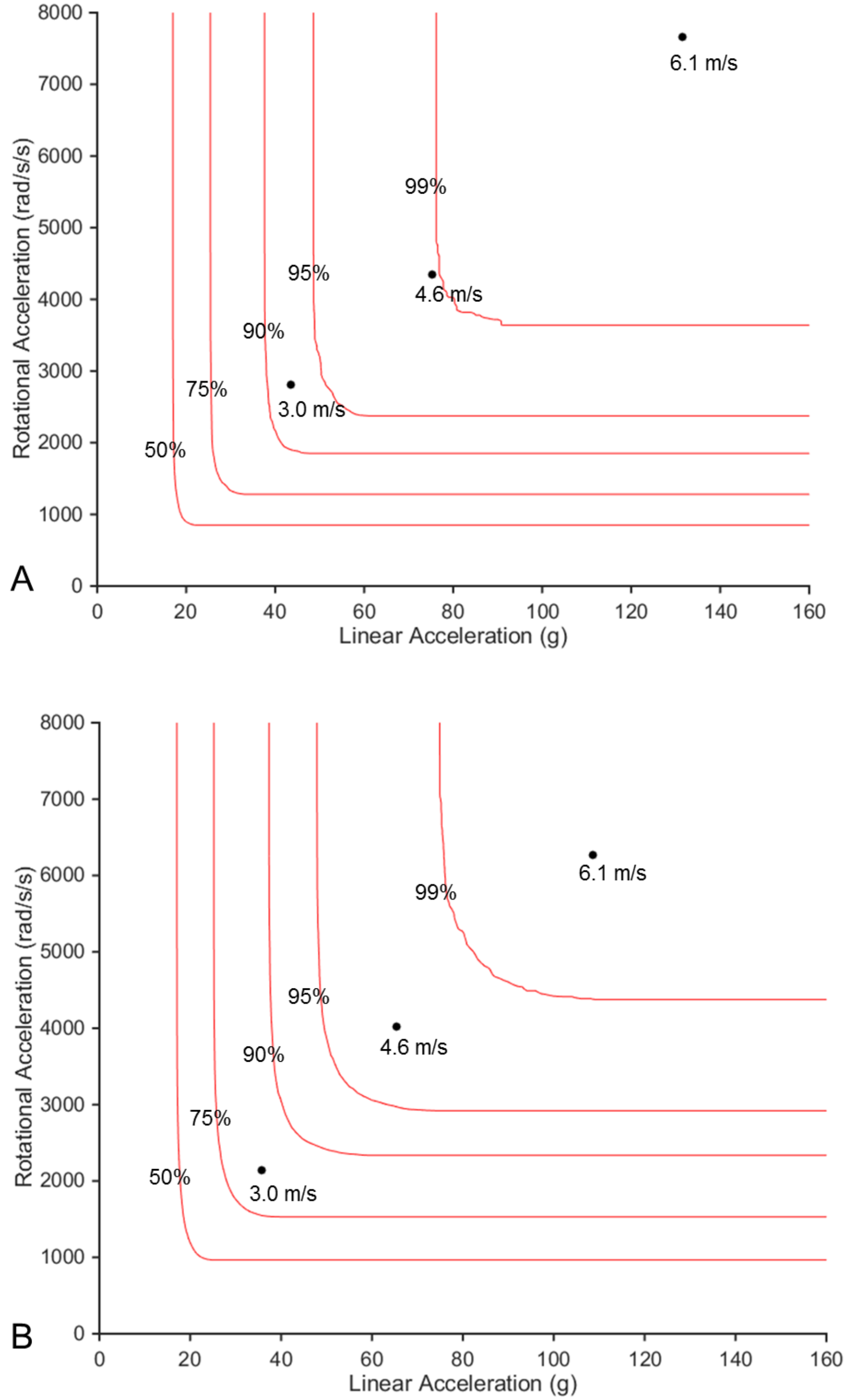

Figure 7: Bivariate CDFs for (A) youth football and (B) collegiate football. 50th, 75th, 90th, 95th, and 99th percentile contour lines are shown for on-field impact for players wearing a youth or varsity Riddell Speed helmet. The average of the laboratory impacts to the same helmet models are overlaid for each impact velocity.

Demonstrative STAR tests

The STAR value for the varsity version of Helmet A was 5.864 and the STAR value for the youth version was 2.268. The varsity version of Helmet B produced a STAR value of 3.035 and the youth version produced a STAR value of 1.314 (Table 3). The values for the youth helmets are lower due to the lower impact exposure of youth players, expressed by the weighting values. The 3.0 m/s impact velocity contributed 8% of the total STAR value for the varsity helmets, while the 4.6 m/s impact velocity contributed 18%, and the 6.1 m/s impact velocity contributed 74%. For the youth helmet STAR values, the 3.0 m/s impact velocity contributed 17%, the 4.6 m/s impact velocity contributed 21%, and the 6.1 m/s impact velocity contributed 62% (Figure 8).

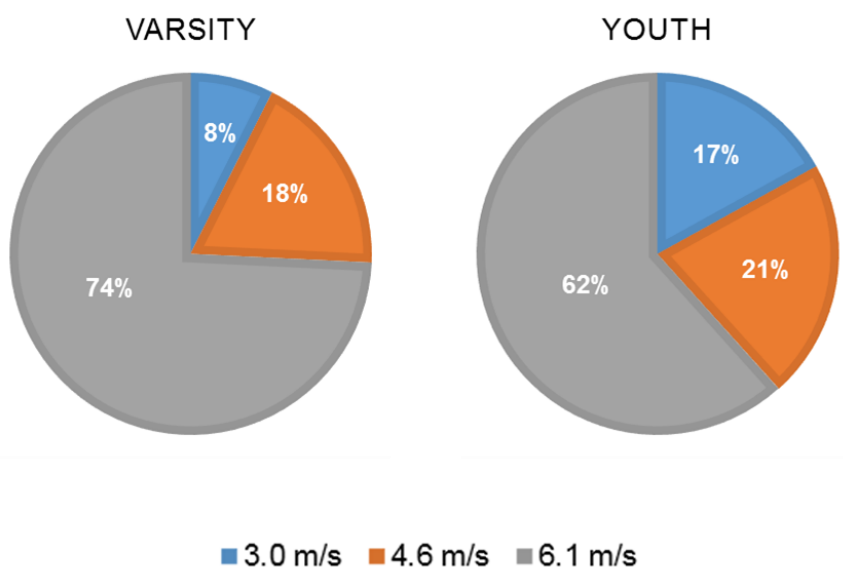

Figure 8: Pie charts illustrating the relative contributions of each impact velocity to the overall STAR value for the varsity football helmets (left) and the youth football helmets (right).

Table 3: STAR calculations for two pairs of matched youth and varsity helmets using their respective methodologies. Varsity helmet A had a STAR value of 5.864, and youth helmet A had a STAR value of 2.268. Varsity helmet B had a STAR value of 3.035, and youth helmet B had a STAR value of 1.314. It should be noted that the smaller values for the youth helmets are not necessarily due to better helmets, but are because of the decreased impact exposure in youth football.

VARSITY HELMET A

Impact Location	Velocity (m/s)	Peak a (g)	Peak α (rad/s/s)	Risk of Injury (%)	Exposure per season	STAR
Front	3.0	41	2335	0.16	83	0.129
Front	4.6	67	3555	1.20	18	0.216
Front	6.1	94	4760	8.51	4	0.341
Front Boss	3.0	30	2099	0.08	83	0.068
Front Boss	4.6	59	3752	1.00	18	0.181
Front Boss	6.1	81	5026	6.51	4	0.260
Back	3.0	41	1933	0.11	83	0.090
Back	4.6	70	4559	3.03	18	0.545
Back	6.1	100	7428	47.98	4	1.919
Side	3.0	39	2896	0.22	83	0.185
Side	4.6	58	4395	1.68	18	0.302
Side	6.1	104	6822	40.70	4	1.628
					STAR	***5.864***

VARSITY HELMET B

Impact Location	Velocity (m/s)	Peak a (g)	Peak α (rad/s/s)	Risk of Injury (%)	Exposure per season	STAR
Front	3.0	39	1341	0.06	83	0.051
Front	4.6	69	2300	0.47	18	0.084
Front	6.1	116	3875	9.73	4	0.389
Front Boss	3.0	27	1551	0.05	83	0.037
Front Boss	4.6	48	2384	0.21	18	0.038
Front Boss	6.1	72	3182	1.07	4	0.043
Back	3.0	33	1653	0.06	83	0.053
Back	4.6	60	3152	0.66	18	0.120
Back	6.1	100	4330	7.56	4	0.303
Side	3.0	32	1978	0.08	83	0.066
Side	4.6	67	3638	1.26	18	0.226
Side	6.1	109	6587	40.60	4	1.624
					STAR	***3.035***

YOUTH HELMET A

Impact Location	Velocity (m/s)	Peak a (g)	Peak α (rad/s/s)	Risk of Injury (%)	Exposure per season	STAR
Front	3.0	47	2297	0.19	55	0.107
Front	4.6	77	3990	2.54	4	0.102
Front	6.1	110	5138	18.54	1	0.185
Front Boss	3.0	34	2291	0.11	55	0.062
Front Boss	4.6	50	3381	0.53	4	0.021
Front Boss	6.1	94	4954	9.51	1	0.095
Back	3.0	36	2614	0.16	55	0.086
Back	4.6	70	6259	10.88	4	0.435
Back	6.1	105	8667	74.43	1	0.744
Side	3.0	42	2881	0.25	55	0.138
Side	4.6	63	4256	1.76	4	0.071
Side	6.1	97	6083	22.28	1	0.223
					STAR	***2.268***

YOUTH HELMET B

Impact Location	Velocity (m/s)	Peak a (g)	Peak α (rad/s/s)	Risk of Injury (%)	Exposure per season	STAR
Front	3.0	38	1730	0.08	55	0.046
Front	4.6	75	3288	1.34	4	0.053
Front	6.1	110	4991	16.77	1	0.168
Front Boss	3.0	29	1772	0.06	55	0.032
Front Boss	4.6	54	3108	0.50	4	0.020
Front Boss	6.1	74	3897	2.09	1	0.021
Back	3.0	38	1965	0.10	55	0.054
Back	4.6	64	3594	1.09	4	0.044
Back	6.1	98	5202	13.18	1	0.132
Side	3.0	40	2432	0.16	55	0.087
Side	4.6	65	4306	2.02	4	0.081
Side	6.1	113	7286	57.71	1	0.577
					STAR	***1.314***

Discussion

The purpose of this paper is to further advance STAR by creating protocols specific to youth and varsity football helmets. The inclusion of both linear and rotational acceleration results in a more robust approach to evaluating helmets. A helmet that more effectively reduces the acceleration of the head will reduce concussion risk on average.[8, 9, 39-41] The current NOCSAE standard was developed to reduce the incidence of catastrophic head injuries in football and has led to the elimination of these type of injuries from the game.[2] STAR is intended to only provide additional information and does not intend replace the current standard. Only those football helmets that have the NOCSAE seal will ever be tested using STAR.

All STAR methodologies are based on two overarching principles. The first is that helmets that reduce linear and rotational acceleration will also decrease the risk of concussion. Although all current head injury standards are based solely on linear acceleration, all head impacts have a rotational component, and both linear and rotational kinematics should be considered in determining injury risk.[8, 9] The second principle is that the laboratory tests are weighted relative to the frequency of impact on the field. In STAR for youth and varsity football, the weightings are derived from a full season of population specific on-field data collection.

Previously, STAR has weighted each impact configuration separately. The updated methods weight on impact velocity alone, with all locations weighted equally. This improvement was made as it is possible to reduce protection in a portion of a helmet that had a small weighting and still score well in the evaluation. By evenly weighting by location, it ensures that a helmet has good protection throughout. STAR for hockey helmets was recently updated to reflect this change.

STAR values are not comparable between youth and varsity helmets because of the differences in on-field exposure. The STAR values for youth helmets are inherently less than STAR values for varsity helmets due to the decreased impact exposure of youth players, and not due to youth football helmets performing better than varsity football helmets. The relative proportions that contribute to STAR values for youth and varsity helmets are unique between methodologies in that lower severity impacts are weighted greater in STAR for youth helmets (Figure 8). This is supported by the on-field data showing the frequency of high magnitude impact is less in youth players.[33-35, 38]

The concussion risk function used to calculate incidence is based on head impact data collected from collegiate football players.[9] This risk calculation likely does not align with the injury tolerance of youth athletes, and should not be directly related to injury outcomes. However, it is used in the youth methodology as it provides a metric that summarizes both linear and rotational acceleration into a single severity measure. As more is understood about concussion in youth players, the methodology can be updated to include a youth-specific risk function. STAR is designed to identify helmets that best reduce head acceleration, and less acceleration is assumed to be associated with a lower risk of injury.[2, 9, 40, 42]

For a constant linear acceleration, rotational acceleration may vary between helmets, and the updated methods can account for this. Through the addition of rotational acceleration, real-world head impacts are more accurately replicated. It is anticipated that varsity helmets that performed well using the original STAR methodology will continue to perform well in the updated methodology. However, helmets that reduce rotational acceleration for constant linear accelerations can be identified.

The underlying data used to make the weightings for the laboratory tests are not perfect. The helmet mounted accelerometer arrays are prone to some error in measurement for both linear and rotational acceleration.[43, 44] However, this method remains the best way to get on-field head impact measurements. The error is reduced as multiple measurements are used and translates through the analyses here giving a good indication of relative impact severities.

The impact response appears to be biofidelic for both surrogates. The impact durations range from 10 to 15 ms for tests conducted in these analyses. These value are in similar to what has been reported in standard drop tests (8 -12 ms),[6] laboratory recreations (15 ms),[40] and on-field measurements (8-14 ms).[32, 33, 45, 46] Furthermore, all the peak linear and rotational acceleration values of the impacts performed lie within the range of acceleration values observed through laboratory recreations of real world impacts[39] as well as data collected on the field.[32-35, 47]

Previously, STAR evaluations have tested a top location. The top location impact has been replaced in STAR for youth and varsity football helmets with a front boss impact that is toward the crown of the helmet. The reasoning behind this was to limit axial loading of the neck. The Hybrid III neck was not developed to be biofidelic in pure compression.[27] The top location is tested in the standard, and therefore still has minimum performance requirements.

Although differences were observed in head kinematics, these differences were relatively small, and the impact responses were nearly identical for linear and rotational acceleration (Figure 6). The pendulum change in velocity was significantly higher for the adult surrogate, meaning that the energy delivered to the surrogates is lower for the youth surrogate, due to the lower inertia. This indicates that the impacts are not identical between surrogates.

The methodologies presented have several limitations. By weighting each location equally, the STAR values are not necessarily directly representative of the actual impact exposures of an average player. Additionally, there is some error associated with the measurements taken for the on-field data. The risk function used is a good predictor of concussion for adults because it was initially developed from collegiate football players.[9] However, this does not accurately describe injury tolerance in the youth population, as data are lacking. We do know enough currently to compare helmet performance for youth players, due to the translational principle that a helmet that reduces head acceleration will also reduce concussion risk. As more is known about concussion in youth players, the methodology can be adjusted to more accurately represent risk.

Conclusions

This paper presents STAR methodology including linear and rotational head acceleration for youth and varsity football helmets. STAR for youth football helmets differs in that a youth specific surrogate is used, with a smaller head, weaker neck, and scaled sliding mass. Values specific to each population, developed from on-field data, are used to weight each STAR impact test. Weightings are specific to impact velocity, but are the same across locations. STAR for youth and varsity football helmets was demonstrated by performing each protocol on two matched youth and varsity helmets. This information will help to inform consumers in purchasing football helmets by directing them towards helmets that more effectively mitigate head accelerations.

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Chapter 5: Closing Remarks

Research Summary

The work presented in this thesis is intended to inform future youth football helmet design, as well as evaluate current helmets. The current impact standard was analyzed for youth and varsity helmets by relating impact tests to on-field data collected for a youth and collegiate population. Data suggest that a youth-specific standard is unnecessary until standards develop to include concussion and more is understood about concussion tolerances in youth players. Youth helmets were also compared to respective varsity helmet counterparts in an effort to observe differences in impact performance. No differences were found between the youth and varsity helmets in linear head acceleration, rotational head acceleration, or concussion correlate.

STAR methodology was presented for youth and varsity football helmets to both include linear and rotational head acceleration. A youth-specific surrogate was developed through scaling of the previously used adult surrogate. Exemplar tests were performed for matched youth and varsity helmets. Each impact test was weighted from on-field data collected for youth and adult players. These tests demonstrated the differences between the youth and adult methodologies, and validated the ability of the STAR evaluation to identify differences in helmet performance.

Publication Outline

Table 1: Summary of progress for publication in regards to each chapter in this thesis.

Chapter	Title	Journal	Status
2	Football Helmet Impact Standards in Relation to On-Field Impacts	Journal of Sports Engineering and Technology	Accepted
3	Comparison of Impact Performance between Youth and Varsity Football Helmets	Journal of Sports Engineering and Technology	In Preparation
4	STAR for Youth and Varsity Football Helmets: Characterizing Helmet Performance Using Linear and Rotation Head Acceleration	TBD	In Preparation