Using Laboratory Impact Devices to Quantify Football Helmet Performance

Tessa M. Reiber

Thesis submitted to the faculty of the Virginia Polytechnic Institute and State University in partial fulfillment of the requirements for the degree of

Master of Science
In
Biomedical Engineering

Steven Rowson, Chair
Stefan M. Duma
P. Gunnar Brolinson

April 30, 2019
Blacksburg, VA

Keywords: biomechanics, linear, rotational, acceleration, concussion, helmet design

Using Laboratory Impact Devices to Quantify Football Helmet Performance

Tessa M. Reiber

ACADEMIC ABSTRACT

When football originated in the 1800s, players wore no protective equipment. Between 1869 and 1905, there were 18 deaths and 159 serious injuries attributed to the sport. Following this, players began to wear protective equipment. The first use of a football helmet was in 1893, made of leather and designed to reduce the risk of skull fracture. Initially, football helmets were intended to protect a player against the most severe hits they would experience on the field. More recently, it has been shown that mild traumatic brain injuries, such as concussions, can induce long-term neurodegenerative processes. Since their introduction, helmets have transformed into plastic shells with padding designed to mitigate accelerations on the brain.

With the growing concern for player safety, regulating bodies, like the National Operating Committee on Standards for Athletic Equipment, have implemented standards for protective equipment, including football helmets. On top of these standards, there have been multiple methods developed to assess helmet performance with different testing apparatuses. Manufacturers are interested in how their helmet performs according to multiple testing methods. This could be costly if they do not have the proper testing equipment that a protocol utilizes. This thesis assesses the interchangeability of different test equipment to reproduce a testing protocol. The desire to perform well in testing standards has driven the improvement of helmet performance and continued design innovation. The second aim of this thesis is to evaluate helmet performance and its relationship with design changes in football helmets manufactured between 1980 and 2018.

Using Laboratory Impact Devices to Quantify Football Helmet Performance

Tessa M. Reiber

GENERAL AUDIENCE ABSTRACT

When football originated in the 1800s, players wore no protective equipment. Between 1869 and 1905, there were 18 deaths and 159 serious injuries attributed to the sport. Following this, players began to wear protective equipment. The first use of a football helmet was in 1893, made of leather and designed to reduce the risk of skull fracture. Initially, football helmets were intended to protect a player against the most severe hits they would experience on the field. More recently, it has been shown that mild traumatic brain injuries, such as concussions, can induce long-term neurodegenerative processes. Since their introduction, helmets have transformed into plastic shells with padding designed to mitigate accelerations on the brain.

With the growing concern for player safety, regulating bodies, like the National Operating Committee on Standards for Athletic Equipment, have implemented standards for protective equipment, including football helmets. On top of these standards, there have been multiple methods developed to assess helmet performance with different testing apparatuses. Manufacturers are interested in how their helmet performs according to multiple testing methods. This could be costly if they do not have the proper testing equipment that a protocol utilizes. This thesis assesses the interchangeability of different test equipment to reproduce a testing protocol. The desire to perform well in testing standards has driven the improvement of helmet performance and continued design innovation. Another aim of this thesis is to evaluate helmet performance and its relationship with design changes in football helmets manufactured between 1980 and 2018.

ACKNOWLEDGEMENTS

I would like to thank Dr. Steve Rowson, my advisor, for all of his mentorship and guidance throughout this process. I would also like to thank my committee members, Dr. Duma and Dr. Brolinson for their willingness to help.

Special thanks to my fellow graduate students, Emily Kieffer, Megan Bland, and Eamon Campolettano for their friendship and patience in answering all of my questions.

I am grateful for the help of all the undergraduate research assistants who have helped in testing and data collection.

I would like to extend my gratitude to Craig McNally for the equipment he developed for me to complete my work.

Finally, thanks to my family and friends for their endless encouragement and positivity.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

INTRODUCTION

Concussion research has become far more prevalent in recent years as the full long-term effects of a concussion have been realized. A concussion is classified as a mild traumatic brain injury (mTBI). Concussion are caused by rapid accelerations or decelerations experienced by the brain due to direct or indirect impact causing dysfunction [1]. It is speculated that these forces on the brain cause neuronal shearing. This shearing can cause ionic imbalances, neurotransmitter release, and changes in glucose metabolism [2]. Common symptoms of concussion include disorientation, headache, dizziness, and confusion [2]. In recent years, it has been shown that repeat concussions can have a lasting effect on the brain [3]. Also, a player that has suffered a concussion is more likely to experience another concussion than a player that has never had a concussion [4]. Concussions have become a higher public concern as research expands on the potential long-term consequences.

Between 2001 and 2009, 173,285 traumatic brain injuries due to sports or recreation activities were treated in patients that were 19 years old or younger [5]. In the same time period, the amount of emergency department visits for a TBI increased by 62% [5]. The sports that caused the highest number of patient visits were bicycling and football [5]. There are also 1.6 to 3.8 million concussions reported annually caused by a sport or recreational activity [6]. On top of this, many concussions go unreported. A study published in 2013 found an unreported or undiagnosed concussion rate of 30.5% [7]. This means that there is a higher prevalence of concussion in sports than what was previously reported in the literature.

It has been reported that football has the highest number of concussions out of 15 collegiate sports [8]. Since its origination in the 1800s, football has become one of the most popular sports in America. Initially, no protective equipment was worn by players. 18 deaths, and 159 serious injuries were attributed to the sport between 1869 and 1905 [9]. This sparked the desire to make football a safer game for players.

Due to the prevalence of injury, players began wearing protective equipment. The use of a helmet was first documented in an Army-Navy football game in 1893. The first football helmets were made of leather and designed to reduce the risk of skull fracture. A notable revision to the game was the use of helmets becoming mandatory by the National College Athletic Association (NCAA) in 1939, and by the National Football League (NFL) in 1940. Throughout the 1950s, facemasks started being used by players. Even with the added protective equipment, football deaths and injuries continued to increase. This resulted in the formation of the National Operating Committee on Standards for Athletic Equipment (NOCSAE) in 1969 to conduct head protection research. In 1973, NOCSAE published its first safety standards for football helmets [10]. These standards presented a pass/fail criteria intended to test the most severe impacts athletes would see on the field, limiting the occurrence of skull fracture.

In addition to the requirement of protective equipment, rule changes were implemented in the late 1900s. A new rule discouraged the use of the head in initial contact with another player. Following this and further rule changes, reports of head and neck injuries began to decrease in the 1970s [9]. Since, studies have shown that rule changes and training technique do, in fact, contribute to lower injury rates in youth football [11].

Modern helmet manufacturing companies began with the start of Riddell in 1929, which first specialized in football shoes. Riddell expanded into the helmet industry in 1939 with the first ever plastic football helmet on the market. After extensive helmet research, Riddell's Pac-3 became the first football helmet to pass the NOCSAE standards in 1973. Schutt joined the helmet industry in the mid-1980s. In 2002, Riddell released the Revolution football helmet, which was the first to address concussive risk reduction in its design.

Using data collected from instrumented helmets of football players, research has investigated the role of helmets in mitigating concussion risk. In 2011, the first Virginia Tech Varsity Football STAR ratings were released, which characterized helmet performance in a single metric. This methodology provided a ranking system based on concussion risk, expanding on the pass/fail criteria based on skull fracture limits. However, the STAR methodology does not replace the NOCSAE standards, as these standards ensure the most severe head impacts are assessed. The STAR rating system was able to provide consumers with a ranking system that addressed helmet performance in concussive impacts, and impacts regularly seen on the field, rather than only the most severe impacts.

Between 1988 and 2004, the number of reported concussions increased [8]. Though concussion rates have increased, it has been shown that helmet design can reduce the risk of concussion [12]. Although no helmet can completely protect a player from concussion, extensive helmet testing is conducted to ensure quality of the protective equipment. There are many different laboratory testing techniques and equipment used to evaluate helmet performance [10, 13]. Originally, testing protocols only evaluated linear accelerations, but it became evident that both linear and angular accelerations were involved in causing a concussion [14]. This led to the development of more advanced testing equipment. NOCSAE standards utilize a drop tower and a pneumatic ram. The NFL also implements their own methods of testing using a pneumatic ram. A pendulum is another common system used in helmet testing.

This thesis addresses two questions related to football helmets and helmet testing. First, the efficacy of reproducing protocol on different equipment was assessed. The second research objective was to evaluate the evolution of helmet performance and design. With this research, manufacturers and researchers will be able to reproduce helmet testing protocol on multiple testing systems. Also, the progression of helmet performance was analyzed to highlight that design modifications over time have led to improved helmet performance.

References

[1] S. P. Broglio *et al.*, "National Athletic Trainers' Association Position Statement: Management of Sport Concussion," *Journal of Athletic Training*, vol. 49, no. 2, pp. 245-265, 2014.

[2] C. C. Giza and D. A. Hovda, "The Neurometabolic Cascade of Concussion," *Journal of Athletic Training*, vol. 36, no. 3, pp. 228-235, 2001.

[3] B. E. Gavett, R. A. Stern, and A. C. McKee, "Chronic traumatic encephalopathy: a potential late effect of sport-related concussive and subconcussive head trauma," (in eng), *Clin Sports Med*, vol. 30, no. 1, pp. 179-88, xi, Jan 2011.

[4] K. M. Guskiewicz *et al.*, "Cumulative effects associated with recurrent concussion in collegiate football players: the NCAA Concussion Study," (in eng), *Jama*, vol. 290, no. 19, pp. 2549-55, Nov 19 2003.

[5] J. Gilchrist, K. Thomas, L. Xu, L. McGuire, and V. Coronado, "Nonfatal Traumatic Brain Injuries Related to Sports and Recreation Activities Among Persons Aged ≤ 19 Years -- Unites States, 2001-2009," *Center for Disease Control and Prevention Morbidity and Mortality Weekly Report*, vol. 60, no. 39, pp. 1337-1342, 2011.

[6] J. A. Langlois, W. Rutland-Brown, and M. M. Wald, "The epidemiology and impact of traumatic brain injury: a brief overview," (in eng), *J Head Trauma Rehabil*, vol. 21, no. 5, pp. 375-8, Sep-Oct 2006.

[7] W. P. M. III, R. C. Mannix, M. J. O'Brien, and M. W. Collins, "The Prevalence of Undiagnosed Concussions in Athletes," *Clin J Sport Med*, vol. 23, no. 5, pp. 339-342, 2013.
[8] J. M. Hootman, R. Dick, and J. Agel, "Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives," (in eng), *J Athl Train,* vol. 42, no. 2, pp. 311-9, Apr-Jun 2007.
[9] M. L. Levy, B. M. Ozgur, C. Berry, H. E. Aryan, and M. L. Apuzzo, "Birth and evolution of the football helmet," (in eng), *Neurosurgery,* vol. 55, no. 3, pp. 656-61; discussion 661-2, Sep 2004.
[10] N. O. C. o. S. f. A. Equipment, "Standard Performance Specification for Newly Manufactured Football Helmets," NOCSAE2018.
[11] B. R. Cobb *et al.*, "Head impact exposure in youth football: elementary school ages 9-12 years and the effect of practice structure," (in eng), *Ann Biomed Eng,* vol. 41, no. 12, pp. 2463-73, Dec 2013.
[12] S. Rowson *et al.*, "Can helmet design reduce the risk of concussion in football? Technical note," *Journal of Neurosurgery,* vol. 120, no. 4, pp. 919-922, 2014.
[13] A. Tyson and S. Rowson, "Adult Football STAR Methodology," Virginia Tech2018.
[14] T. A. Gennarelli, L. E. Thibault, and A. K. Ommaya, "Pathophysiologic Responses to Rotational and Translational Accelerations of the Head," *SAE Technical Paper Series,* vol. 720970, pp. 296-308, 1972.

CHAPTER 1

Assessing the Interchangeability of Different Impact Devices to Replicate a Football Helmet Testing Protocol

Abstract

There are different helmet testing protocols that use different types of impact equipment. The purpose of this study was to demonstrate the efficacy of replicating a single protocol across standardized equipment. A National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform was used with a Hybrid III neck on a linear sliding table for a series of impact tests on both a pendulum and a pneumatic ram. Impactor face and impactor mass were matched between systems. Three football helmets were evaluated using the Varsity Football STAR Methodology on both systems. The target velocities on the pneumatic ram were set to match the specified pendulum velocities of 3.1 m/s, 4.8 m/s, and 6.4 m/s. A total of 24 tests per helmet and system were conducted, consisting of four locations (front, front boss, side, back), three energy levels, and two trials at each test configuration. STAR values were computed to characterize overall helmet performance. The average percent difference in peak resultant linear and angular accelerations between the pendulum and the pneumatic ram were 1.6% ± 4.7% and 0.2% ± 5.9%, respectively. Average difference between STAR values from both systems was 0.143. These results are consistent with the variance observed within individual helmet models and demonstrate the efficacy of replicating different helmet impact protocols on a single impact system.

Introduction

Football was reported to have the greatest number of recorded concussions out of 15 collegiate sports, according to the National Collegiate Athletic Association's Injury Surveillance System [1]. Symptoms of a concussion are typically evident at the time of injury, and research suggests that repetitive concussions could have lasting effects on the brain, including chronic traumatic encephalopathy (CTE) [2]. The instrumentation of football players has provided good estimates of the linear and angular head accelerations associated with concussive impacts [3]. Helmet design is one strategy to reduce risk of concussion on the field, specifically by designing helmets to minimize the linear and angular accelerations resulting from impact [4, 5].

Football helmets are tested to evaluate their performance in a number of different ways. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) created a set of football helmet standards first published in 1973. These standards used a drop tower for helmet testing and an instrumented headform [6]. The initial testing protocol evaluated only linear accelerations, but has been very effective in mitigating catastrophic head injury in football players. It is suggested that both linear and angular accelerations are involved in causing a concussion [7]. This has led to the development of more advanced testing methods that use drop towers, pendulums, and pneumatic rams to evaluate linear and angular acceleration during helmet testing [6, 8]. In this study, we address the question of whether a single piece of equipment can be used to replicate different impact protocols.

The Varsity Football Summation of Tests for the Analysis of Risk (STAR) Methodology is a protocol that evaluates the ability of helmets to reduce linear and angular head acceleration over a range of impact conditions using an impact pendulum, and then summarizes test data into a single value characterizing overall helmet performance [8]. We aimed to reproduce data characterizing helmet performance from STAR testing using a pneumatic ram. Our hypothesis was that helmet performance would not differ between impact devices.

Methods

Three adult football helmets from different manufacturers were chosen for evaluation; Schutt F7VTD, Riddell Precision Fit, and Xenith X2E+. A medium NOCSAE headform was mated with a 50th percentile male Hybrid III neck and mounted on a 16 kg linear sliding table with 5 degrees of freedom (Biokinetics, Ottawa, Canada) for a series of impact tests on both a pendulum impactor and the pneumatic ram (Biokinetics). The headform was instrumented with a 6 degree of freedom sensor package, comprising of three linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and three angular rate sensors (ARS3 PRO-18K, DTS, Seal Beach, CA). In order to reduce variability introduced by replicate hardware, the same instrumentation and the same head and neck assembly were used throughout testing on both systems.

Originally, the pneumatic ram and pendulum impactor had differing masses. A 5.5 kg steel plate was added to the impacting end of the pneumatic ram to match the effective mass of the pendulum arm. A nylon impactor face (20.3 cm diameter, 12.7 cm radius of curvature) was adapted to the pneumatic ram to match the impactor face used for Varsity Football STAR testing on the pendulum (Figure 1.1) [8]. The effective impacting mass of both systems was 20 kg. By mass matching the systems, identical impact velocities could be used on both systems to generate similar accelerations.

Figure 1.1: The pendulum impactor face (left) with a 20.3 cm rounded nylon impactor face, with a 12.7 cm radius of curvature; and the modified pneumatic ram impactor face (right) with a 1.24 kg nylon face, 5.5 kg steel disk, and 13.2 kg rod, with the same geometries as the pendulum impactor face. The effective impacting mass of both systems was 20 kg.

All helmets were evaluated using the Varsity Football STAR methodology on both impact systems [8-10]. This consisted of running a total of 24 tests per helmet, with four locations, three energy levels, and two trials at each test configuration. Linear and rotational accelerations were measured for each impact at 3.09 m/s, 4.83 m/s, and 6.38 m/s on the front, side, front boss, and back locations (Table 1.1) [8]. A STAR value (Equation 1) [10] was used to summarize data from all tests into a single value characterizing helmet performance. STAR weights each impact condition by an exposure value (E) based on how frequently a player would experience an impact at that location (L) and energy level (V) on the field. Then, each value is multiplied by the risk (R) (Equation 2) associated with that impact location [9], which is computed from peak resultant linear (a) and angular accelerations (α). The product of risk and exposure is a predicted incidence value, and the addition of all incidence values for a helmet model represents the STAR value. Before testing, a velocity gate was used to verify matched velocities for the impact conditions on the pendulum impactor and pneumatic ram. The velocity gate was set up to take measurements right before impact with the helmet.

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

$$RISK = \frac{1}{1+e^{-(-10.2+0.0433*a+0.000873*\alpha-0.00000092*a*\alpha)}} \qquad \text{(Eq. 2)}$$

All helmets were tested with facemasks. After completing all tests, peak resultant linear and angular accelerations, and the resulting STAR values were computed in MATLAB and compared. Paired t-tests were performed on the data, comparing resulting accelerations on the pendulum and pneumatic ram at each test configuration. A significance threshold of $p < 0.05$ was used.

Table 1.1: NOCSAE headform linear slide table positions relative to the zero position for each impact location. The zero position was defined where the intersection of the midsagittal and transverse planes on the NOCSAE headform align with the center of the impactor face, when there is no rotation about the Y and Z axes. For each test, the x-axis position was set to where the helmet and the impactor face just touch, when the pendulum arm is hanging vertically. There was no rotation about the x-axis.

Location	Y (cm)	Z (cm)	Ry (deg)	Rz (deg)
Back	0	+4.8	0	-180
Front	0	+5.3	-20	0
Front Boss	0	+2.3	-25	+67.5
Side	-4	+5.8	-5	-100

Results

The paired differences of peak linear acceleration values between systems for each helmet model at each location and energy level can be seen in Figure 1.2. There were no significant differences between systems for any helmet ($p > 0.24$). The average difference for all helmets between the pendulum and pneumatic ram peak resultant linear acceleration values was -0.35 g ± 2.94 g. The corresponding percent difference, calculated with respect to pendulum values, was -1.55% ± 4.68%.

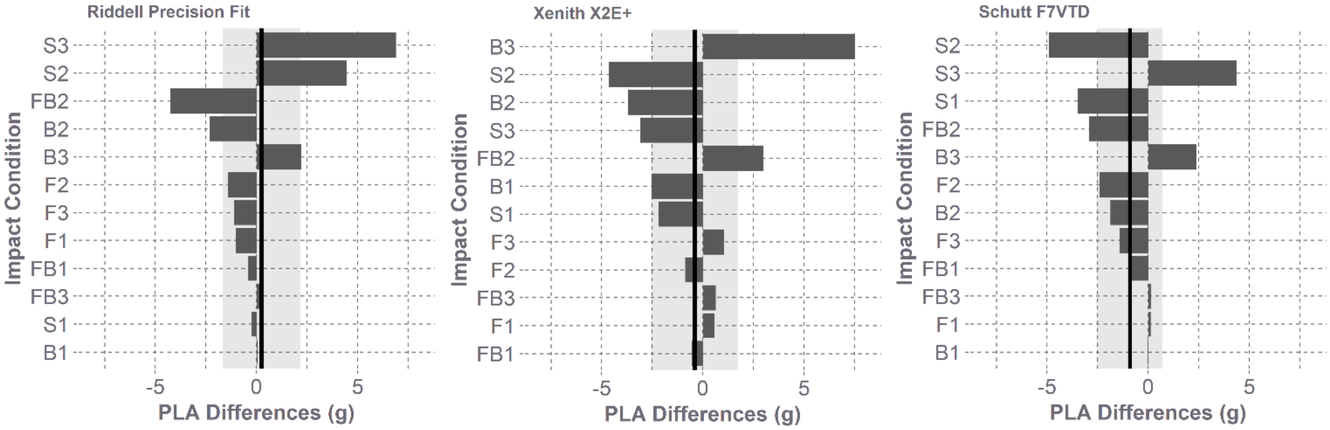

Figure 1.2: Paired differences of peak linear acceleration (PLA) values between pendulum and pneumatic ram values at each impact condition for each helmet model. The impact conditions are abbreviated as follows: front (F), front boss (FB), side (S), back (B), followed by energy level 1 (3.09 m/s), 2 (4.83 m/s), or 3 (6.38 m/s). The graphs show an approximately balanced distribution about zero, with an average difference value of 0.25, -0.4, -0.91 g for the Riddell Precision Fit, Xenith X2E+, and Schutt F7VTD, respectively.

Figure 1.3, below, shows the paired differences of peak rotational accelerations from the pendulum and pneumatic ram for each helmet. There were no significant differences between systems for any helmet ($p > 0.32$). For all helmets, the average difference between both systems was 16 rad/s^2 ± 206 rad/s^2. The corresponding percent difference was -0.21% ± 5.92%.

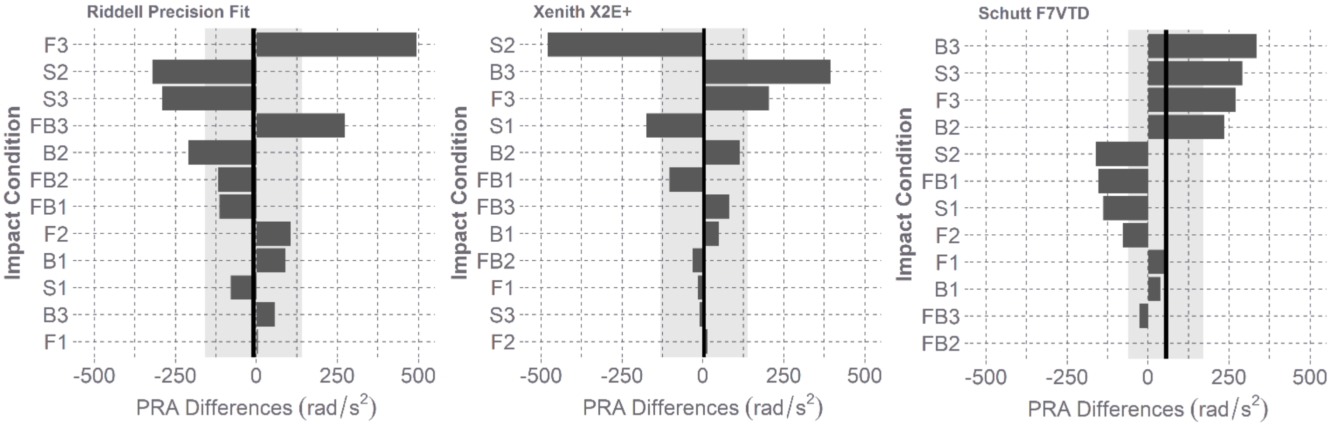

Figure 1.3: Paired differences of peak rotational acceleration (PRA) values between systems for each helmet model. The impact conditions are labeled using the same abbreviations as Figure 1.2. There is an approximately balanced distribution about zero, with an average difference value of -9, 3, and 55 rad/s^2 for the Riddell Precision Fit, Xenith X2E+, and Schutt F7VTD, respectively.

The average difference in incidence values for each test setup was 0.012 ± 0.058 (Figure 1.4). Similarly, there were no significant differences between systems for any helmet ($p > 0.35$).

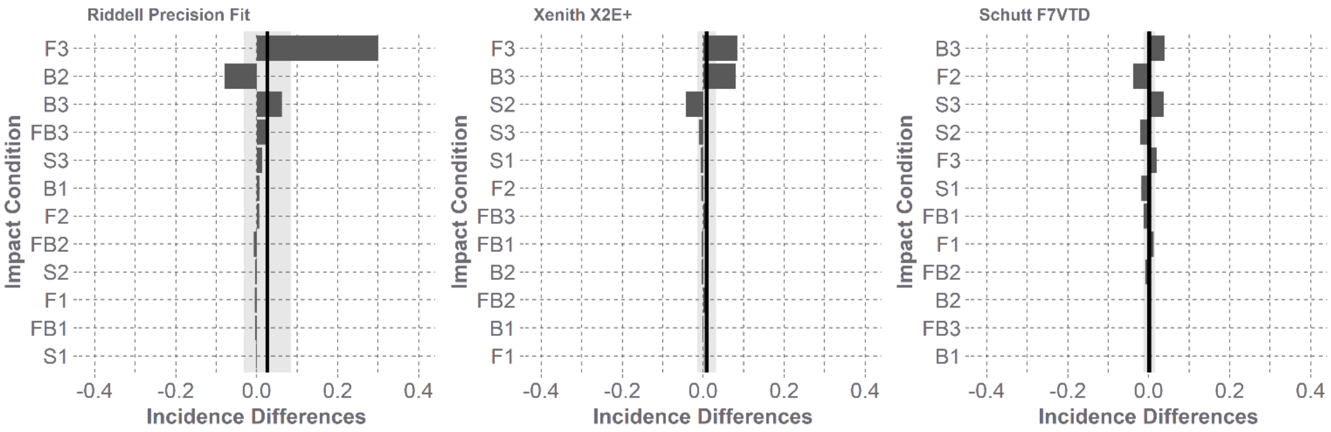

Figure 1.4: Paired differences of incidence values between systems for each helmet model and impact condition. Impact conditions are labeled using the same abbreviations as Figures 1.2 and 1.3. The average difference values were 0.03, 0.008, and 0.001, for the Riddell Precision Fit, Xenith X2E+, and the Schutt F7VTD, respectively.

Overall, the Varsity Football STAR values calculated for the three helmets were similar between the pneumatic ram and pendulum. The differences between STAR values from both systems were 0.01, 0.10, and 0.32 for the Schutt F7VTD, Xenith X2E+, and the Riddell Precision Fit, respectively (Table 1.2). The average difference in STAR values was 0.143 ± 0.159.

Table 1.2: STAR values calculated for both the pendulum and the pneumatic ram, and the difference between each. The average difference in STAR value was 0.143 ± 0.159.

	Pendulum	Pneumatic Ram	Difference
F7VTD	1.48	1.47	0.01
X2EP	3.09	2.99	0.1
PFIT	4.33	4.01	0.32

Discussion

In this study, impact tests were completed on the pendulum impactor and pneumatic ram to determine reproducibility between systems. The average percent difference in peak resultant linear and angular accelerations between the pendulum and the pneumatic ram were -1.55% ± 4.68% and -0.21% ± 5.92%, respectively. Average difference between STAR values from both systems was 0.143 ± 0.159. These results demonstrate that the Varsity Football STAR protocol can be replicated on the pneumatic ram to produce similar kinematics and overall helmet performance scores. However, some variability was observed between test data collected by the impact devices. This can be partly explained by an individual helmet's inherent variance. It is possible that the padding in the helmet will interact differently in an impact, or that padding had shifted in a previous test. Additionally, slight differences in helmet positioning on the headform could alter impact location. In this way, the variation we see could be due to slight differences in positioning and padding interaction in an impact.

There was larger variability in risk and incidence values than linear and angular acceleration values. The risk curve assesses the risk of concussion due to combined linear and angular accelerations (Equation 2) [9]. When combining these values, it is important to note that the individual variances of linear and angular accelerations are also combined. In this way, the risk value inherently has greater variance than either linear or angular acceleration values. In addition, the curve's nonlinearity designates a range of combined accelerations where there is a steeper slope in the curve. At relatively low acceleration magnitudes, risk does not vary much with changes in magnitude. Where the slope becomes steeper, small changes in accelerations result in large differences in risk values. At these values, variability in linear and angular accelerations attribute to a magnified difference in risk and incidence values.

Additionally, the pendulum impactor and pneumatic ram have different approach paths. The pneumatic ram approaches the headform in a purely linear motion. In contrast, the pendulum swing is an arc, but this motion can be approximated as linear over short distances and at the bottom of the arc. Therefore, it can be approximated that the pendulum is moving linearly while it is in contact with the helmet during impact. The motion throughout the impact is linear, or can be approximated as such, for both systems, therefore this does not affect the resulting kinematics of the headform.

Overall, the study shows that the Varsity Football STAR testing protocol, originally performed on a pendulum impactor, can be completed on a pneumatic ram with an impactor face modification to achieve a good estimate of helmet performance defined by a STAR value, within 0.143 ± 0.159 on average. This study provides evidence that a single piece of equipment can potentially be used to test multiple football helmet testing protocols. This offers manufacturers the flexibility to conduct multiple testing protocols while avoiding cost of additional equipment or outsourcing.

There are some limitations in this study. Only three helmets were tested. Some helmets could produce differing variability between systems and tests. Also, only two different football helmet testing methods were evaluated. Future studies are needed to compare other testing protocols and equipment.

Conclusion

The goal of this study was to show that football helmet testing protocols can be reproduced on a pendulum impactor and a pneumatic ram. This was successful when replicating the Varsity Football STAR Methodology, in which a pendulum impactor is traditionally used, on a pneumatic ram. Similar linear and angular acceleration, risk, and overall helmet performance scores were produced on both systems using matched impact conditions with equivalent effective impacting masses. This provides flexibility to manufacturers who have equipment restraints.

References

[1] J. M. Hootman, R. Dick, and J. Agel, "Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives," (in eng), *J Athl Train,* vol. 42, no. 2, pp. 311-9, Apr-Jun 2007.

[2] B. E. Gavett, R. A. Stern, and A. C. McKee, "Chronic traumatic encephalopathy: a potential late effect of sport-related concussive and subconcussive head trauma," (in eng), *Clin Sports Med,* vol. 30, no. 1, pp. 179-88, xi, Jan 2011.

[3] S. Rowson *et al.*, "Rotational head kinematics in football impacts: an injury risk function for concussion," (in eng), *Ann Biomed Eng,* vol. 40, no. 1, pp. 1-13, Jan 2012.

[4] S. Rowson *et al.*, "Can helmet design reduce the risk of concussion in football? Technical note," *Journal of Neurosurgery,* vol. 120, no. 4, pp. 919-922, 2014.

[5] S. Rowson, R. W. Daniel, and S. M. Duma, "Biomechanical performance of leather and modern football helmets," (in eng), *J Neurosurg,* vol. 119, no. 3, pp. 805-9, Sep 2013.

[6] (2011). *NOCSAE DOC (ND)002-11m11a, Standard Performance Specification for Newly Manufactured Football Helmets*.

[7] T. A. Gennarelli, L. E. Thibault, and A. K. Ommaya, "Pathophysiologic Responses to Rotational and Translational Accelerations of the Head," *SAE Technical Paper Series,* vol. 720970, pp. 296-308, 1972.

[8] A. Tyson and S. Rowson, "Adult Football STAR Methodology," Virginia Tech2018.

[9] S. Rowson and S. M. Duma, "Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration," (in eng), *Ann Biomed Eng,* vol. 41, no. 5, pp. 873-82, May 2013.

[10] S. Rowson and S. M. Duma, "Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," (in eng), *Ann Biomed Eng,* vol. 39, no. 8, pp. 2130-40, Aug 2011.

CHAPTER 2

Relating Football Helmet Performance to Design Characteristics over Time

Abstract

Although helmets have been required in collegiate football since 1939, the first documented use of a football helmet during a game was in 1893. Since then, helmets have increased in size, weight, and pad thickness in efforts to reduce injury risk. With advances in helmet technology, interestingly, annual estimates of concussion incidence are near all-time highs, likely due to increased awareness. The purpose of this study was to assess overall helmet performance and its relationship with design changes in football helmets produced between 1980 and 2018. An impact pendulum was used for a series of tests to evaluate the performance of 15 helmet models. The Varsity Football STAR Methodology, a method to quantify helmet performance based on its ability to reduce concussion risk, was used to evaluate helmet performance. Helmets were tested on a modified National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform and impacted at four different locations (front, front boss, back, side), and three impact velocities (3.1 m/s, 4.8 m/s, and 6.4 m/s), with two trials at each impact configuration. Each helmet's weight, pad thickness, and helmet offset were measured. Resulting peak linear accelerations generally decreased in helmets manufactured after 2010 when compared to helmets manufactured before 1990, most notably at the highest impact velocity. Peak rotational accelerations remained relatively unchanged between eras, aside from the side location. Overall, concussion risk was observed to be lower in helmets introduced after 2010 compared to earlier helmets. Weight, pad thickness, and helmet offset all increased between 1980 and 2018; and were all associated with improved helmet performance. These results demonstrate how helmet performance has improved since the 1980s.

Introduction

Initially, football players wore no protective equipment. Between 1869 and 1905, 18 deaths and 159 serious injuries occurred in football [1], raising awareness for safety regulations in the sport. The first documented use of a helmet was in 1893 in the Army-Navy football game. Football helmets were originally made of leather and designed to protect against skull fracture. Riddell began manufacturing helmets in 1939, and released the first football helmet with a plastic shell. Plastic helmets were lighter and provided more room to incorporate padding than the original leather helmets. An innovation to plastic helmets came in the 1940s, with the addition of a suspension system in the helmet.

Since then, protective equipment, including helmets, has become a requirement of the game. Helmets were mandated for collegiate players in 1939, and National Football League (NFL) players in 1940. However, catastrophic head injuries continued to increase. In 1969, the National Operating Committee on Standards for Athletic Equipment (NOCSAE) was formed to conduct research and develop safety standards in the interest of improving head protection. By 1973, NOCSAE had published the first safety standards for football helmets [1, 2]. These standards were designed to test helmets based on skull fracture limits, and presented a pass/fail criteria. After the implementation of NOCSAE standards, head injuries in football decreased by 50% [3].

Schutt joined the helmet industry in the mid-1980s. In 2002, Riddell released the Revolution helmet, which was the first football helmet designed with the intent to reduce the risk of concussion [4]. Helmets continued to increase in size and weight, and added more energy-absorbing padding [5]. Riddell and Schutt have implemented different padding in their helmets;

Riddell has traditionally used vinyl nitrile foam, while Schutt has used thermoplastic urethane (TPU) padding. Along with innovation in padding material and design, helmet shell design has evolved. Helmet shells have become larger, and many now incorporate contours, ridges, and vent holes in the outer shell [5]. Panels have been placed in strategic locations on the shell to help mitigate forces and allow additional shell deformation in frequently impacted locations. In addition, more competitors joined the industry, including Rawlings, Xenith, SG, and VICIS. The increased competition has added pressure to design a better performing helmet, overall benefitting players' safety [1].

Even with the addition of safety regulations and helmet innovation, concussion rates have increased in football; reported concussions have shown an increase of 7% annually from 1988 to 2004 in collegiate level football [6]. This could be due to a heightened education and awareness of mild traumatic brain injuries (mTBIs) in recent years [7]. The definition of a concussion has also evolved over the years. Originally, a concussion was associated with severe and long-lasting symptoms. Generally, concussion symptoms included loss of consciousness, post-traumatic amnesia, and confusion [8-10]. Today, a concussion is defined by much less severe symptoms. Symptoms considered to be concussive include dizziness, nausea, sleep issues, and memory problems. Additionally, even brief occurrences of these symptoms are considered in concussion diagnosis [11]. The change in definition to include milder symptoms has also likely influenced the increase in reported concussions.

It has been shown that helmet design can decrease the risk of concussion [12]. Research comparing leather helmets and commonly used helmets in the 2000s demonstrated the superior performance of new helmets [13]. Previous research by Viano et al. found that helmets increased in size, weight, and performance between the 1970s and 2010s when tested at impact conditions relevant to the National Football League (NFL) [5]. The Virginia Tech Varsity Football STAR Methodology aggregates a range of real-world impact scenarios into a single metric of overall helmet performance in the context of concussion [14]. This methodology has motivated manufacturers to improve helmet design to mitigate accelerations over a range of impact severities experienced during play. The purpose of this study was to evaluate helmet performance of Riddell and Schutt helmets from the 1980s to the present, and to analyze design differences and their effect on performance when subjected to typical impact conditions seen in college and high school football.

Methods

In this study, a total of 15 varsity football helmets were analyzed. Helmets were chosen from two football helmet brands; Riddell and Schutt. The helmets were chosen based on the year they were manufactured, from 1980 to 2018 (Figure 2.1). These companies have developed numerous helmet models since 1980. The helmet models included serve as a representation of the evolution of football helmets over time. Helmets that are no longer manufactured were bought through Ebay. All helmets were visually in good condition, but the history of use was unknown for the helmets purchased from Ebay.

Riddell Pac-44 (1980)
Riddell M155 (1982)
Schutt Pro Air II (1987)
Schutt Air Advantage (1987)
Riddell VSR-4 (1992)
Riddell VSR-4 PL (1992)
Riddell Revolution (2002)
Schutt DNA (2004)
Riddell Speed (2008)
Schutt Air XP (2008)
Riddell 360 (2011)
Schutt Vengeance DCT (2014)
Riddell SpeedFlex (2014)
Riddell Precision-Fit (2017)
Schutt F7 LTD (2018)

Figure 2.1: Pictures of all helmets tested in order of manufacture year.

A pendulum impactor was used for testing (Figure 2.2). The pendulum had a total mass of 37 kg, consisting of a 15.5 kg anvil, and a 21.5 kg arm. The total rotational inertia of the pendulum was 72 kg-m2. The length of the pendulum arm was 190.5 cm. A 20.3 cm rounded nylon impactor face, with a 12.7 cm radius of curvature was used. This impactor face resembles the curve of a football helmet.

A modified medium NOCSAE headform with a 50% male Hybrid III neck was instrumented with a six degrees of freedom sensor package. A NOCSAE headform was used to optimize fit of the helmet [15]. The NOCSAE headform was modified by removing mass from the underside to maintain the position of the occipital condyle, and allow for proper mounting of the neck. An adaptor plate, of mass equal to the mass removed, was used to mate the headform and the neck while conserving the center of gravity and location of the occipital condyle on the headform [16]. This head and neck assembly has been shown to generate linear and rotational accelerations similar to the 50th percentile Hybrid III head and neck assembly [17]. The sensor package consisted of three linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and three angular rate sensors (ARS3 PRO-18K, DTS, Seal Beach, CA). Data was collected at 20,000 Hz. Angular velocity was differentiated to obtain angular acceleration data. Linear and angular accelerations were filtered at CFC 1000 Hz and CFC 155 Hz, respectively, and transformed to the CG of the headform. Additionally, the headform and neck were mounted on a 16 kg linear sliding table meant to represent the effective mass of the torso (Biokinetics, Ottawa, Canada).

Figure 2.2: The pendulum impactor used for testing. The impactor was a 20.3 cm rounded nylon impactor face, with a 12.7 cm radius of curvature, mimicking the curve of a football helmet.

All helmets were evaluated using the Varsity football Summation of Tests for the Analysis of Risk (STAR) protocol [18]. Each helmet was impacted a total of 24 times; two trials at three energy levels and four locations (Table 2.1). Front, front boss, back, and side locations were tested (Figure 2.3) [18]. All helmets were tested with facemasks. The impact velocities were 3.09 m/s, 4.83 m/s, and 6.38 m/s which are intended to represent the 80th, 95th, and 99th percentile impacts on-field.

Figure 2.3: The front, front boss, side, and back locations tested on the pendulum impactor.

Table 2.1: Locations of the linear sliding table relative to the zero position. The zero position is defined at the intersection of the midsagittal and transverse planes with the impactor face when the NOCSAE headform has no rotation in the X and Y axes. Before every test, the distance in the X direction was determined by moving the headform to be in contact with the impactor face when the pendulum arm is hanging vertically.

Location	Y (cm)	Z (cm)	Ry (deg)	Rz (deg)
Back	0	+4.8	0	-180
Front	0	+5.3	-20	0
Front Boss	0	+2.3	-25	+67.5
Side	-4	+5.8	-5	-100

For each test, linear and rotational accelerations were measured and resultant peak values calculated. A STAR value (Equation 1) was calculated to evaluate helmets based on their performance [19]. Each impact condition is weighted by an exposure value (E), which is based on the frequency a player would experience an impact at that location (L) and velocity (V) on the field. The exposure value is then multiplied by a risk value (R) (Equation 2), which is the risk of concussion at a particular peak linear (a) and angular acceleration (α) at each respective location [20].

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

$$RISK = \frac{1}{1+e^{-(-10.2+0.0433*a+0.000873*\alpha-0.00000092*a*\alpha)}} \qquad \text{(Eq. 2)}$$

Multiple factors of helmet design were also evaluated. Pad thicknesses, helmet weight, and helmet offset were measured. Pad thickness at each impact location tested was measured using calipers for each helmet. Helmet weights were measured with the facemask they were tested with. To measure helmet offset, photographs of each helmet mounted on the NOCSAE headform and Hybrid III neck were taken at zero and 90 degrees; with zero degrees oriented facing the camera, and 90 degrees being a side profile of the right side of the helmet. Photographs of the bare NOCSAE headform were taken at the same angles. Using MATLAB, the bare headform picture was subtracted from the helmeted headform picture at their respective angles (Figure 2.4). Offset was calculated by determining the distance between points on the outside of the helmet shell and points on the outside of the headform. In MATLAB, radial lines were drawn every ten degrees from the center of the headform, to cover the entire area that the helmet covered the headform.

Coordinates where the lines intersected the edge of the helmet and headform were identified. This was done with both the zero and 90 degree subtracted pictures. The distances between the edges of the headform and helmet were averaged at each orientation. These values were then averaged to quantify the offset value for each helmet.

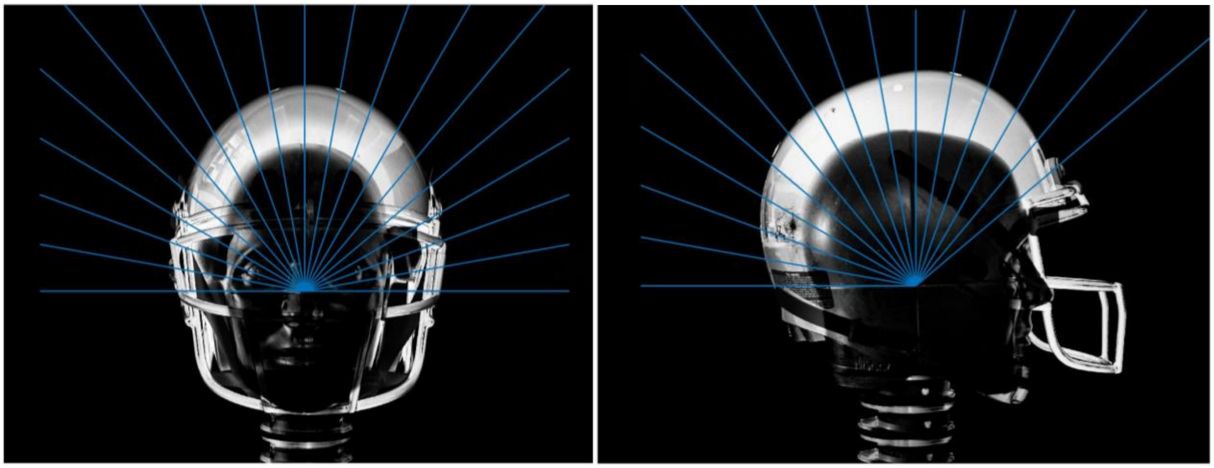

Figure 2.4: Example of the final product of subtraction of a bare NOCSAE headform picture from a helmeted headform picture. This technique was used to find the average offset of each helmet.

To compare results over time, two eras were analyzed. The helmets manufactured before 1990, and the helmets manufactured after 2010 were compared. An ANOVA was used to assess the differences between eras for peak linear and peak rotational accelerations. Spearman's correlation was used to assess the correlation of design features with helmet performance variables over time. A significance threshold of $p < 0.05$ was used for all statistical analysis.

Results

Overall, peak linear acceleration decreased for all impact conditions over time (Figure 2.5). Peak linear acceleration of the headform in matched impact conditions decreased for helmets manufactured after 2010 compared to helmets manufactured before 1990 (Figure 2.6). Overall, the average decrease in PLA over all energy levels and locations was 30.04%. The largest decrease between eras was 49.9% at the 6.4 m/s impact velocity at the side location (Table 2.2).

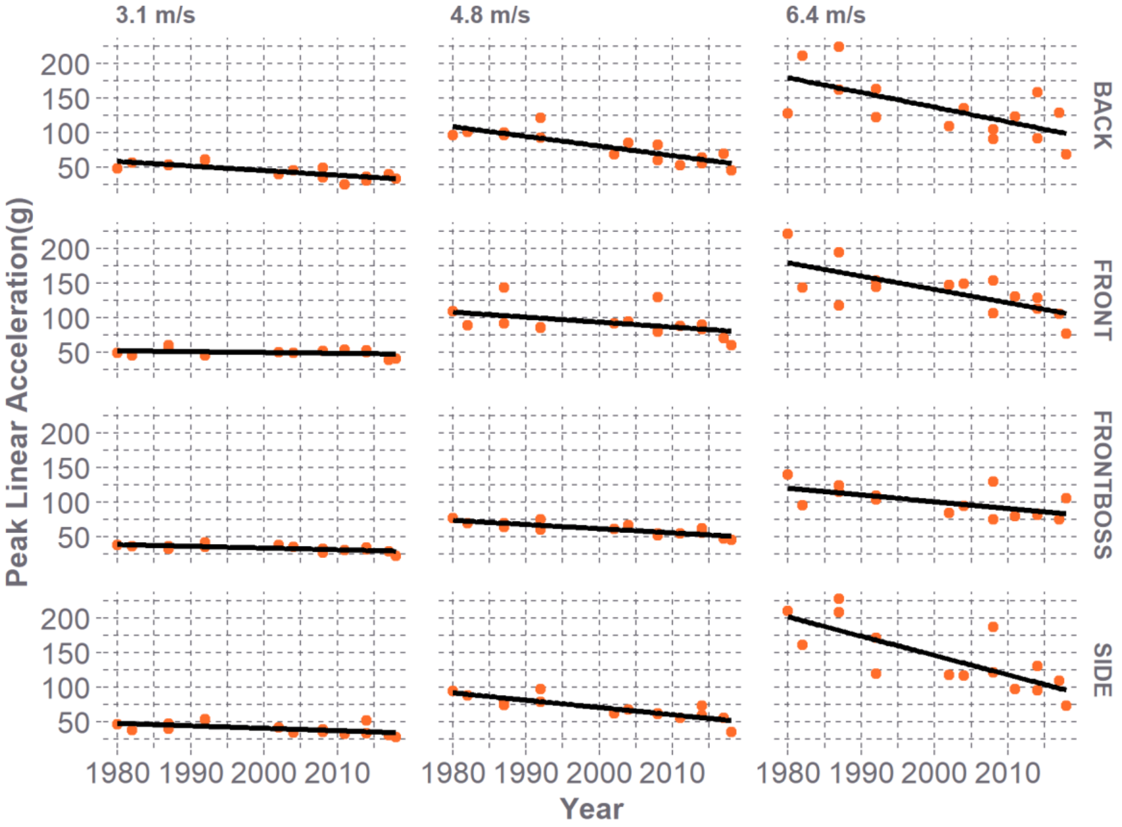

Figure 2.5: Peak linear accelerations of helmets by year for each testing configuration. While all show a decrease over the years, the most variation is seen at the 6.4 m/s impact velocity. All correlations were significant ($\rho > -0.5735$, $p < 0.0254$), except for the 3.1 m/s impact velocity at the front and front boss locations.

An ANOVA was conducted to compare the data between eras (before 1990 and after 2010); the results showed that location, energy level, and era and their interactions are significant ($p < 0.0049$). A post-hoc Tukey HSD test showed that there were significant differences in peak linear accelerations between eras at each location and the 6.4 m/s impact velocity ($p < 0.027$), with an average effect size of 64.9 g. The back location was also significant at the 4.8 m/s impact velocity ($p = 0.00082$). The average effect sizes at the 3.1 m/s and 4.8 m/s impact velocities were 9.8 g and 29.8 g, respectively.

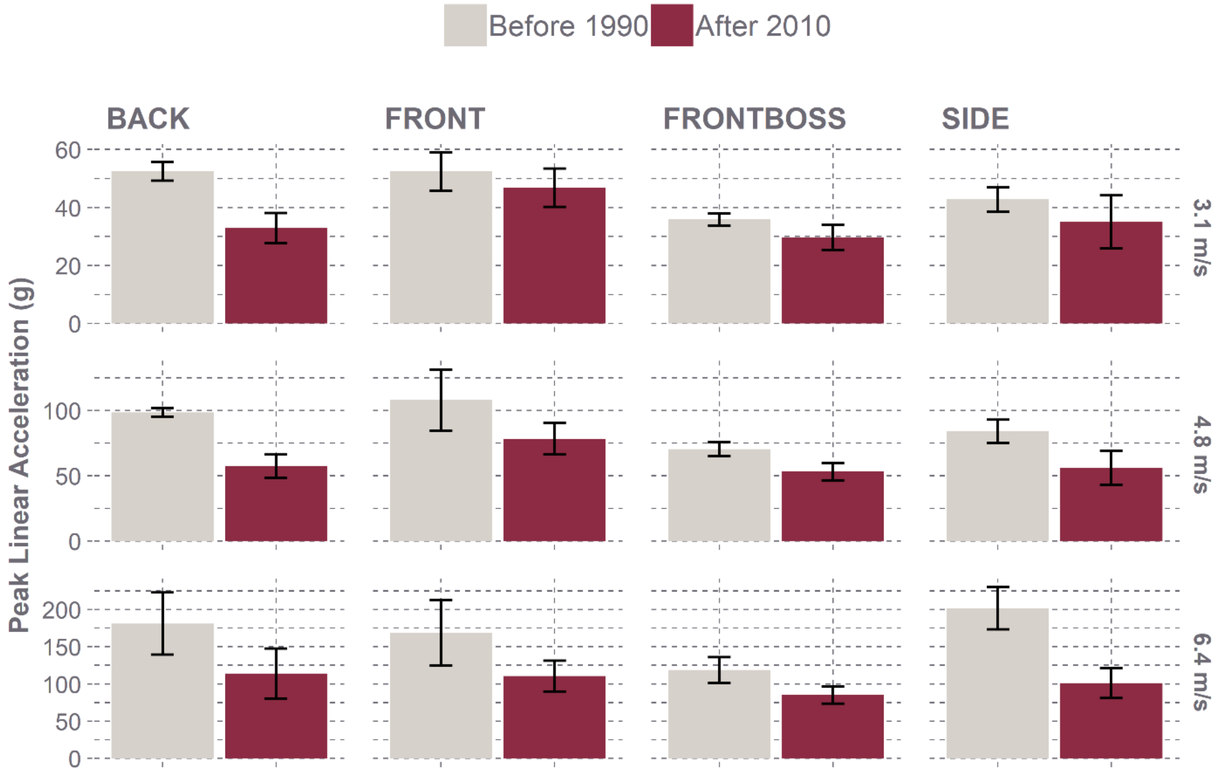

Figure 2.6: The average peak linear acceleration of all helmets manufactured before 1990 and all helmets manufactured after 2010.

Table 2.2: The percent decreases in peak linear acceleration of all helmets manufactured before 1990 compared to helmets manufactured after 2010 for all locations and impact velocities. In all cases, peak linear acceleration decreased.

Energy Level	Back	Front	Front boss	Side
1	37.3%	10.7%	17.1%	18.2%
2	41.8%	27.6%	24.5%	33.7%
3	37.1%	34.5%	28.1%	49.9%

Peak rotational accelerations decreased over time for all impact velocities at the front boss, side, and back location (Figure 2.7). The average percent decrease of peak rotational acceleration over all locations and impact velocities between eras was 14.72% (Table 2.3). There are noticeable differences in peak rotational acceleration values at the back and side locations (Figure 2.8). The largest decrease was at the side location for energy level three, with a 6745 rad/s^2 decrease between eras (Figure 2.8).

An ANOVA on the data between eras showed significant differences for all locations, energy levels, eras, and their interactions ($p < 0.001$). A post-hoc Tukey HSD test showed significant differences between eras for the side location at energy levels two and three ($p < 0.001$). The average effect sizes at the 3.1 m/s, 4.8 m/s, and 6.4 m/s were 257 rad/s^2, 886 rad/s^2, and 2566 rad/s^2, respectively.

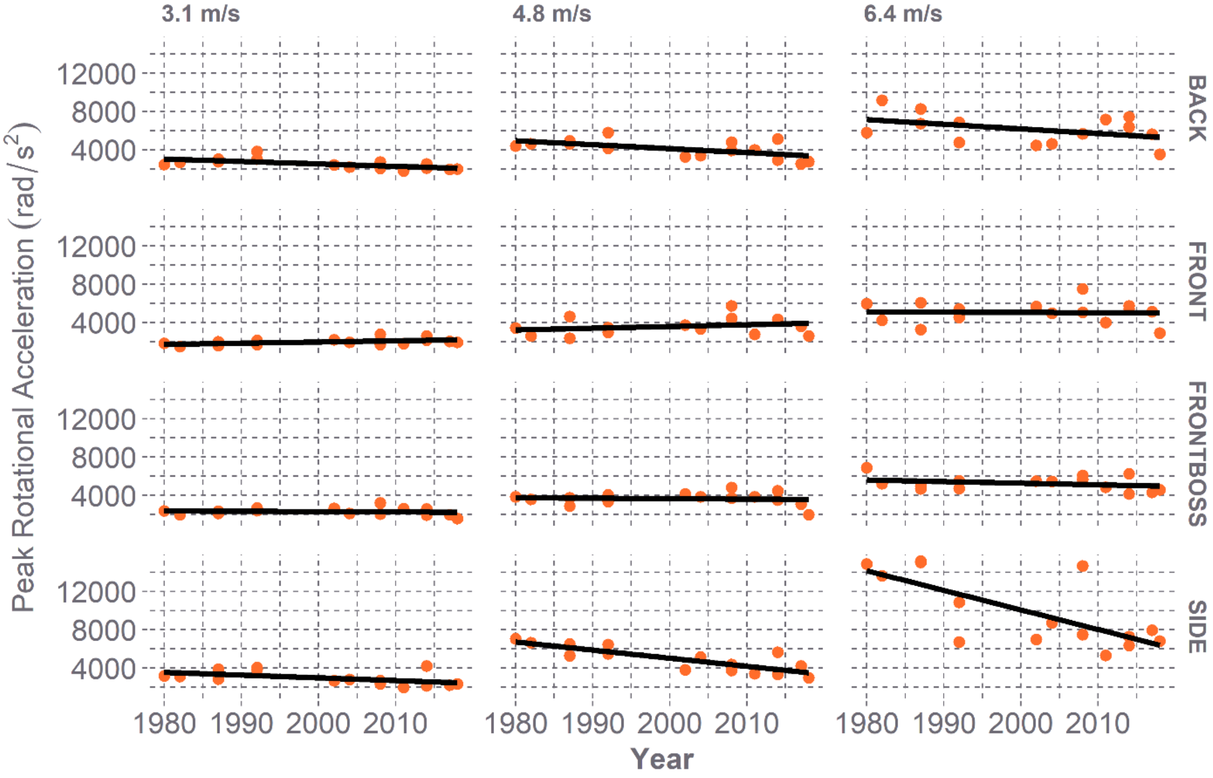

Figure 2.7: Peak rotational acceleration of helmets by year for each testing configuration. Correlations at all impact velocities of the side location, and the 4.8 m/s and 6.4 m/s impact velocities of the back location were significant ($\rho > -0.5358$, $p < 0.0395$).

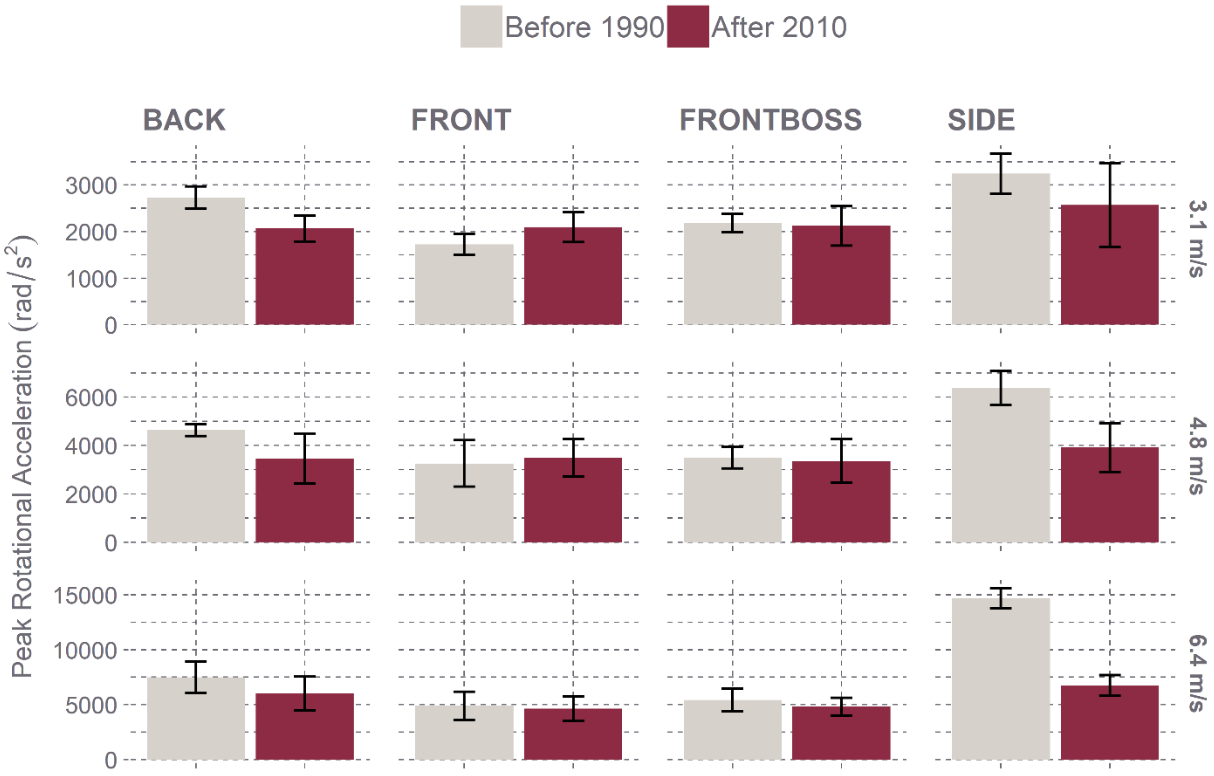

Figure 2.8: The average peak rotational acceleration of all helmets manufactured before 1990 compared to helmets manufactured after 2010 for all locations and energy levels.

Table 2.3: The percent decreases in peak rotational acceleration of all helmets manufactured before 1990 compared to helmets manufactured after 2010 for all locations and energy levels. Most test configurations showed a decrease in peak rotational acceleration between eras, except for the first and second energy level at the front location.

Energy Level	Back	Front	Front boss	Side
1	24.27%	-21.33%	2.13%	20.82%
2	25.37%	-7.39%	3.99%	38.67%
3	19.70%	5.29%	11.06%	54.03%

STAR values also decreased over time (Figure 2.9). The difference between average STAR values between eras was significant (p = 0.001). The average STAR value of helmets manufactured before 1990 was 17.45. The average STAR value of helmets manufactured after 2010 was 5.60. Average STAR values dropped 68% between 1990 and 2010.

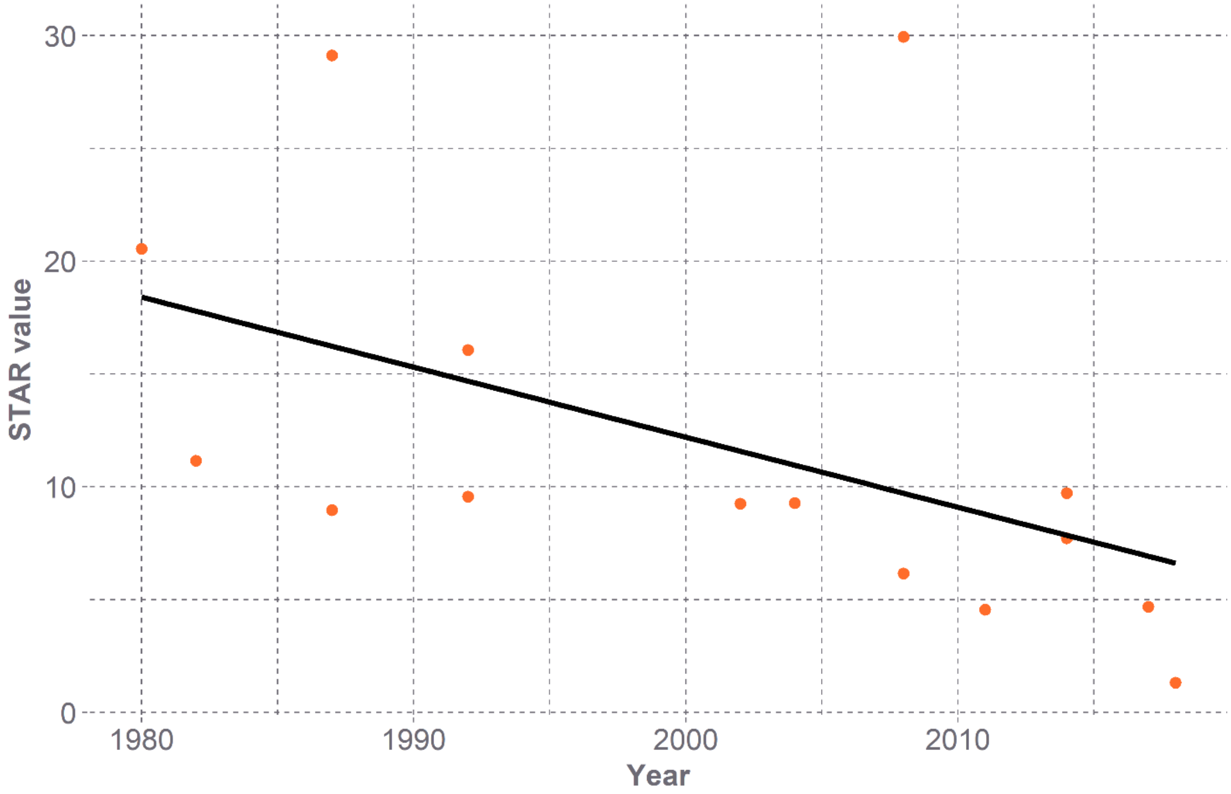

Figure 2.9: Individual helmet incidence values by year. STAR values have decreased over time (ρ = -0.64, p = 0.01).

Average helmet offset increased over time (Figure 2.10). The average offset of helmets manufactured before 1990 was 1.68 inches. The average offset of helmets manufactured after 2010 was 2.14 inches. This shows a 0.46 inch increase of helmet offset between time periods (27.3%).

The Schutt F7 LTD, manufactured in 2018, showed a 0.9 inch increase in average offset compared to the Riddell Pac-44, manufactured in 1980 (Figure 2.11).

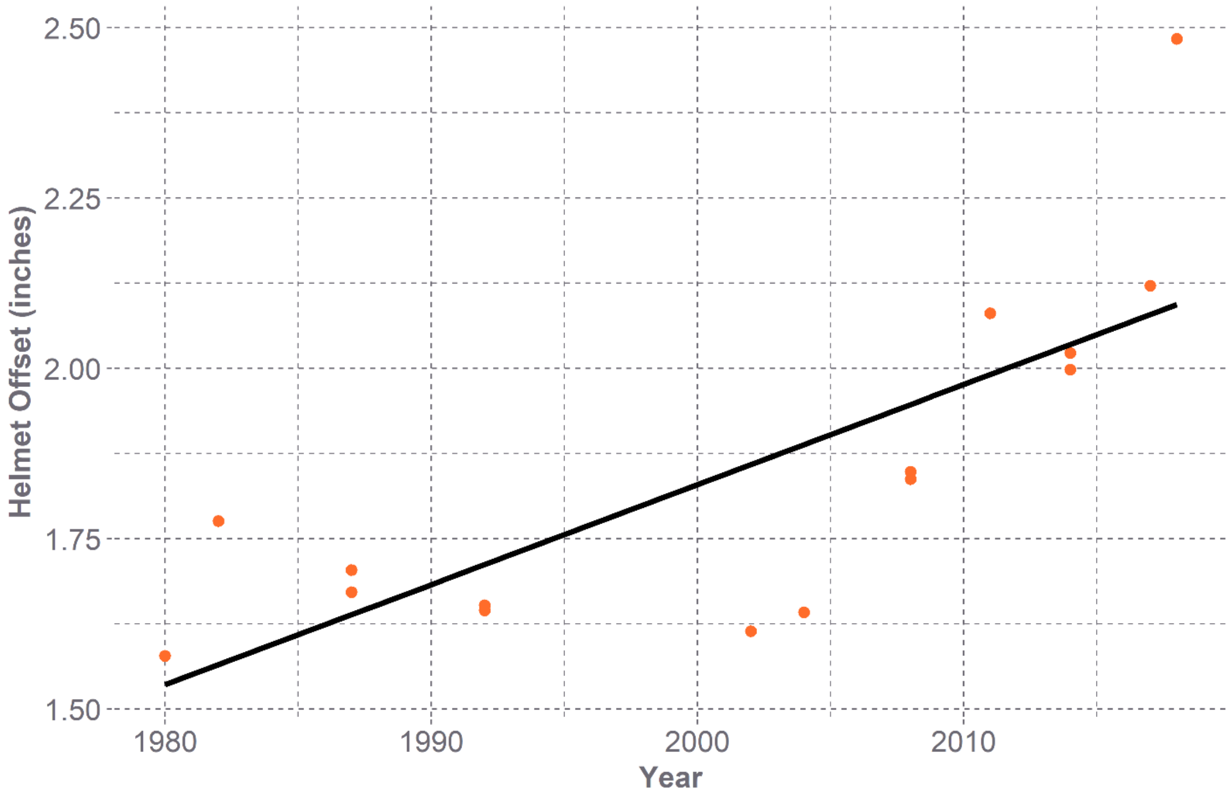

Figure 2.10: Average helmet offset of each helmet by year. Average helmet offset increased from 1980 to 2018 (ρ = 0.79, p = 0.0004).

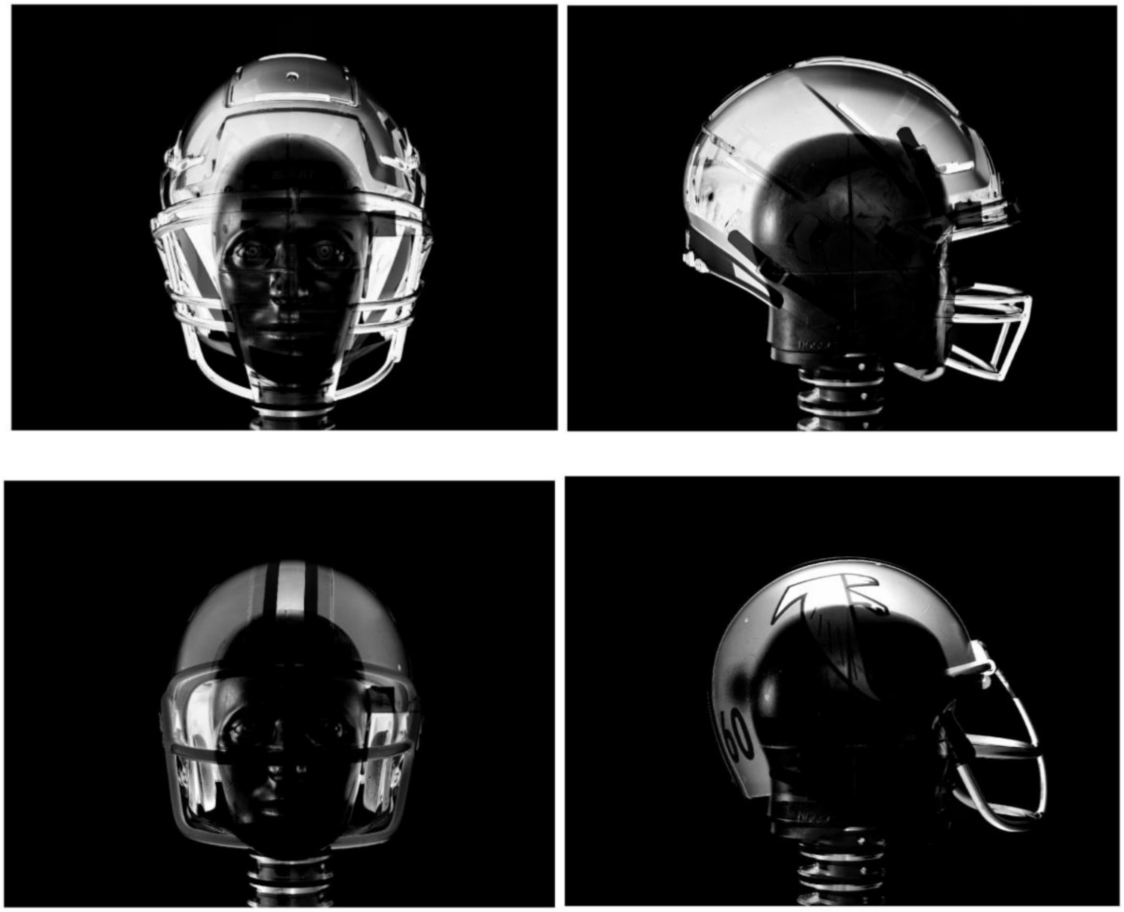

Figure 2.11: Offset of the Schutt F7 LTD (top) compared to the Riddell Pac-44 (bottom). There is a noticeable difference in offset between helmets at both orientations, corresponding to a 0.9 inch difference in average helmet offset.

Average helmet offset was negatively correlated with STAR value (Figure 2.12). As helmet offset increased, the corresponding STAR value decreased.

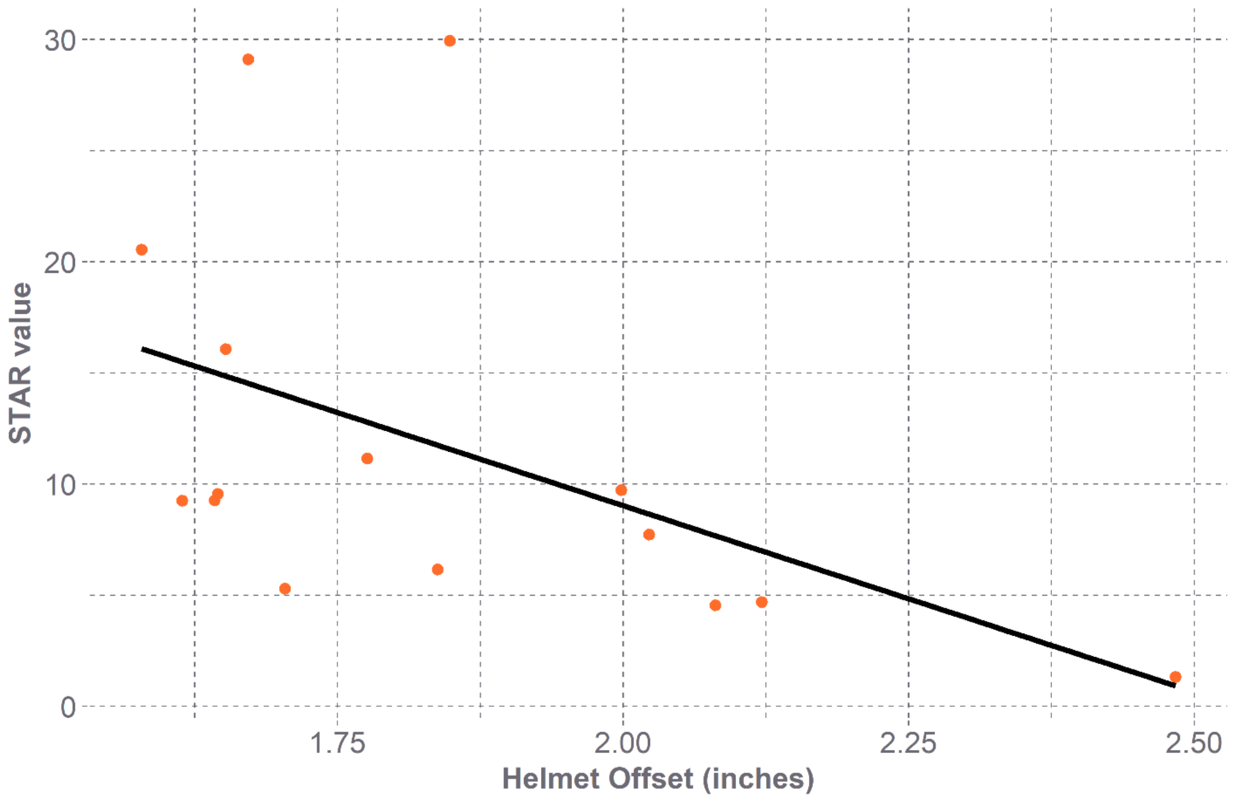

Figure 2.12: Helmet offset versus STAR values. With greater helmet offset, STAR values decreased (ρ = -0.55, p = 0.037).

Helmet weight increased over time (Figure 2.13). The average weight of helmets manufactured before 1990 was 3.74 pounds. The average weight of helmets manufactured after 2010 was 4.41 pounds. Helmet weight increased by 0.6725 pounds (17.98%) between the time periods. STAR value is negatively correlated with weight (Figure 2.14). As weight increased, STAR value decreased.

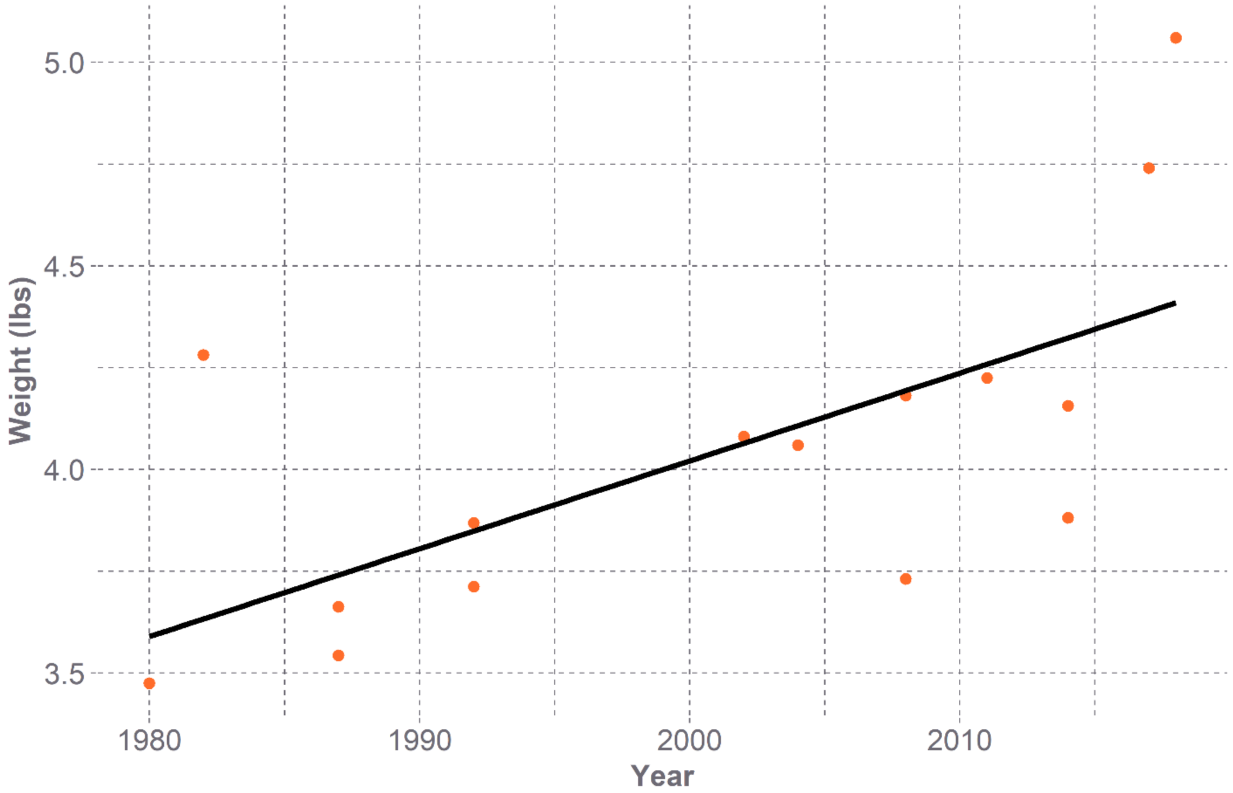

Figure 2.13: Helmet weight with facemask over manufacture year. Helmet weight increased over time (ρ = 0.66, p = 0.0075).

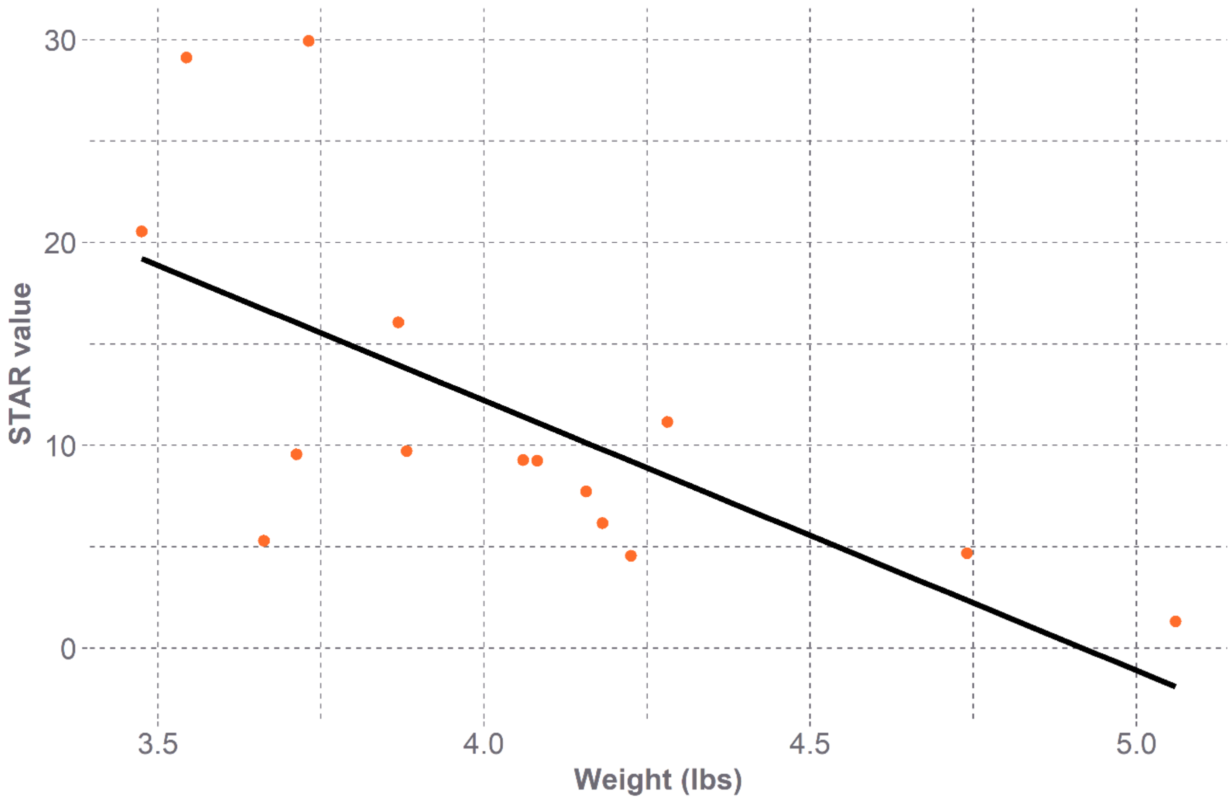

Figure 2.14: Helmet weight versus STAR values. As helmet weight increased, STAR values decreased (ρ = -0.6714, p = 0.0077).

Pad thickness also increased over time at all locations (Figure 2.15). The average pad thickness of helmets manufactured before 1990 was 0.944 inches. The average pad thickness of helmets manufactured after 2010 was 1.53. The average pad thickness increased by 0.59 inches between eras (62%). Pad thickness increased over all locations between eras. There was an average increase of 46.7%, 77.7%, 62.9%, and 51.7%, at the front, back, side, and front boss locations, respectively.

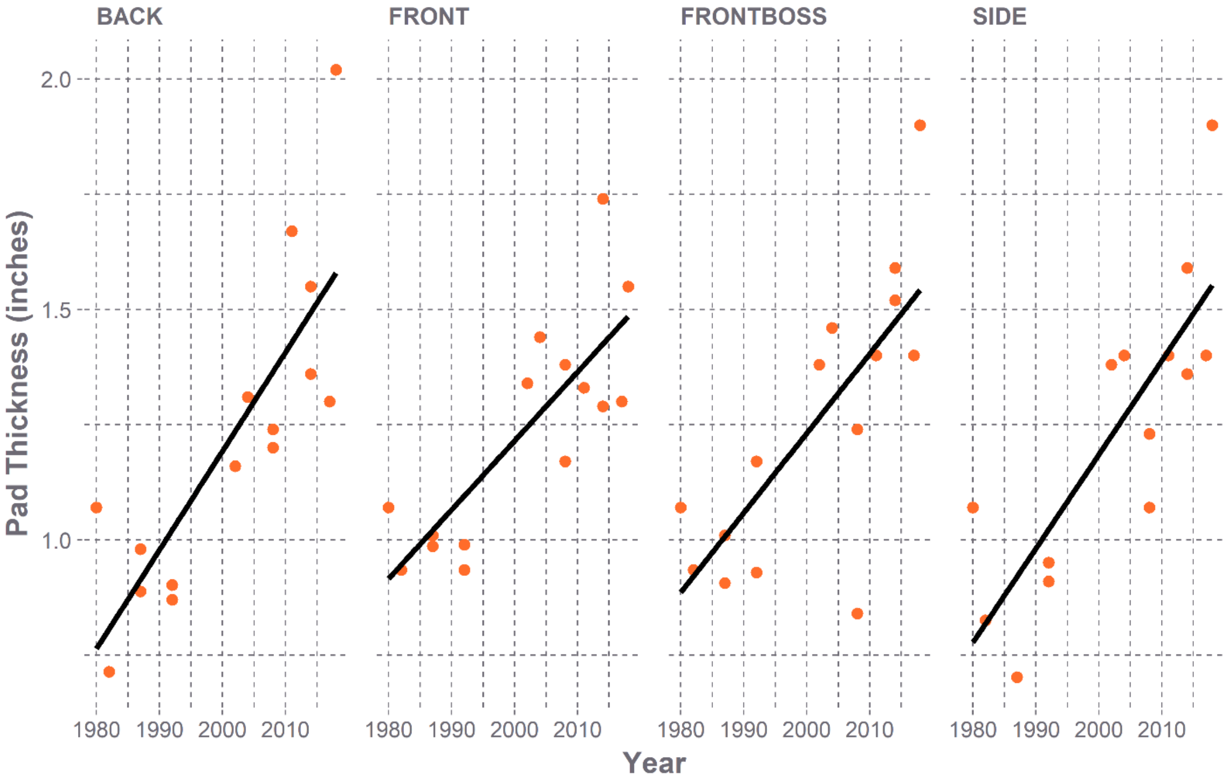

Figure 2.15: Helmet pad thickness at each location by year. Overall, pad thickness has increased at all locations between 1980 and 2018 (ρ > 0.73, p < 1.808E-8).

At the back location, all impact velocities have a significant correlation between a decrease in peak linear acceleration with an increase in pad thickness (ρ < -0.59, p < 0.022). All impact velocities at the side location also show a significant correlation between peak linear acceleration and pad thickness (ρ < -0.70, p < 0.0028). The 6.4 m/s impact velocity level at the front boss location also shows a significant correlation (ρ = -0.61, p = 0.015). Although the other locations and impact velocities do not show significance, there is an association between increased pad thickness and decreased peak linear acceleration (Figure 2.16).

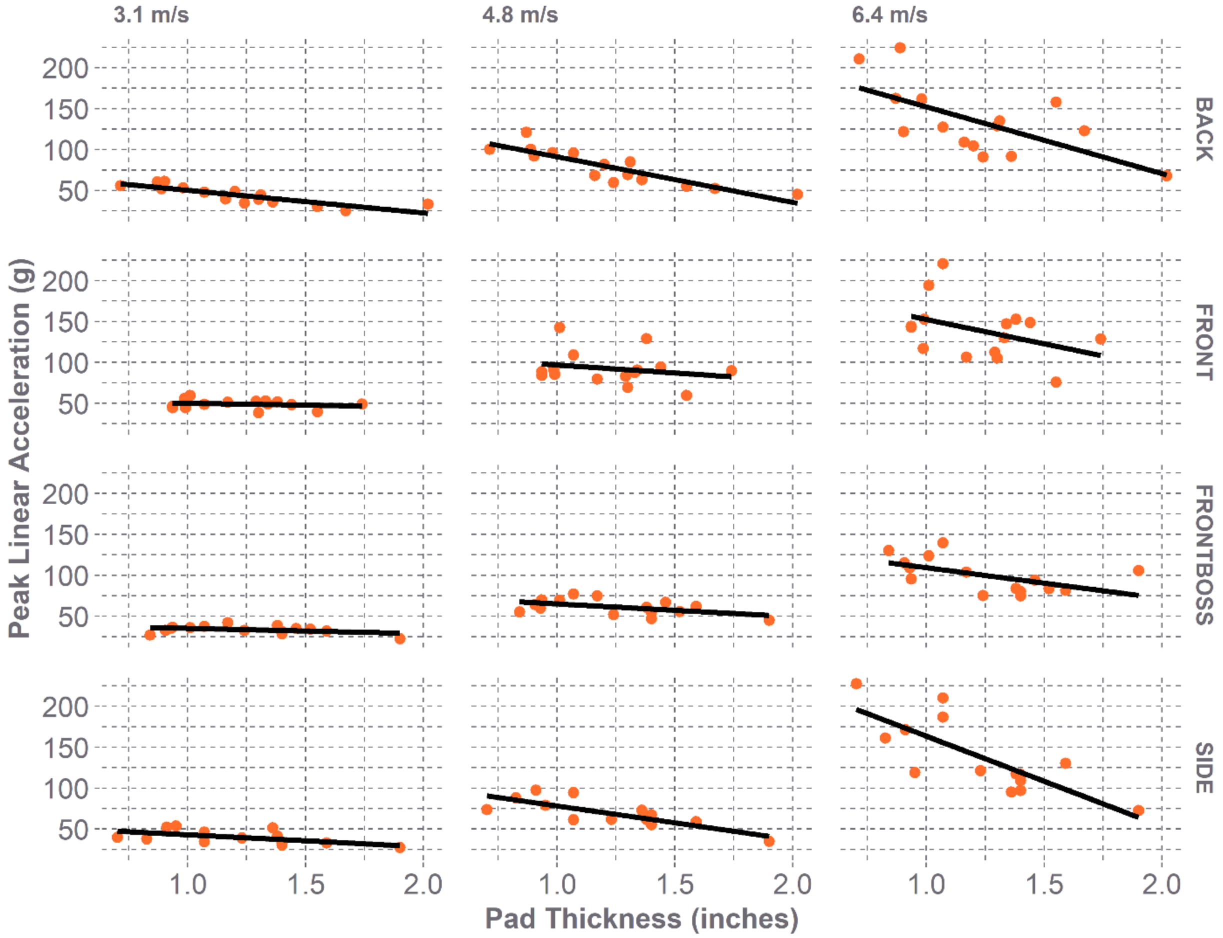

Figure 2.16: Pad thickness at each location versus peak linear acceleration at each location and energy level. Overall, there was a decrease in peak linear acceleration associated with an increase in pad thickness.

All impact velocities at the side location have a significant negative correlation between pad thickness and peak rotational acceleration ($\rho < -0.57$, $p < 0.03$). At the back location, energy levels one and two have significant correlation ($\rho < -0.59$, $p < 0.02$). While most of the cases show a negative correlation, there are some locations and energy levels which appear to show no correlation, or a slight positive correlation between pad thickness and peak rotational acceleration

(Figure 2.17). STAR is negatively correlated with average pad thickness (Figure 2.18). As pad thickness increased, STAR value decreased.

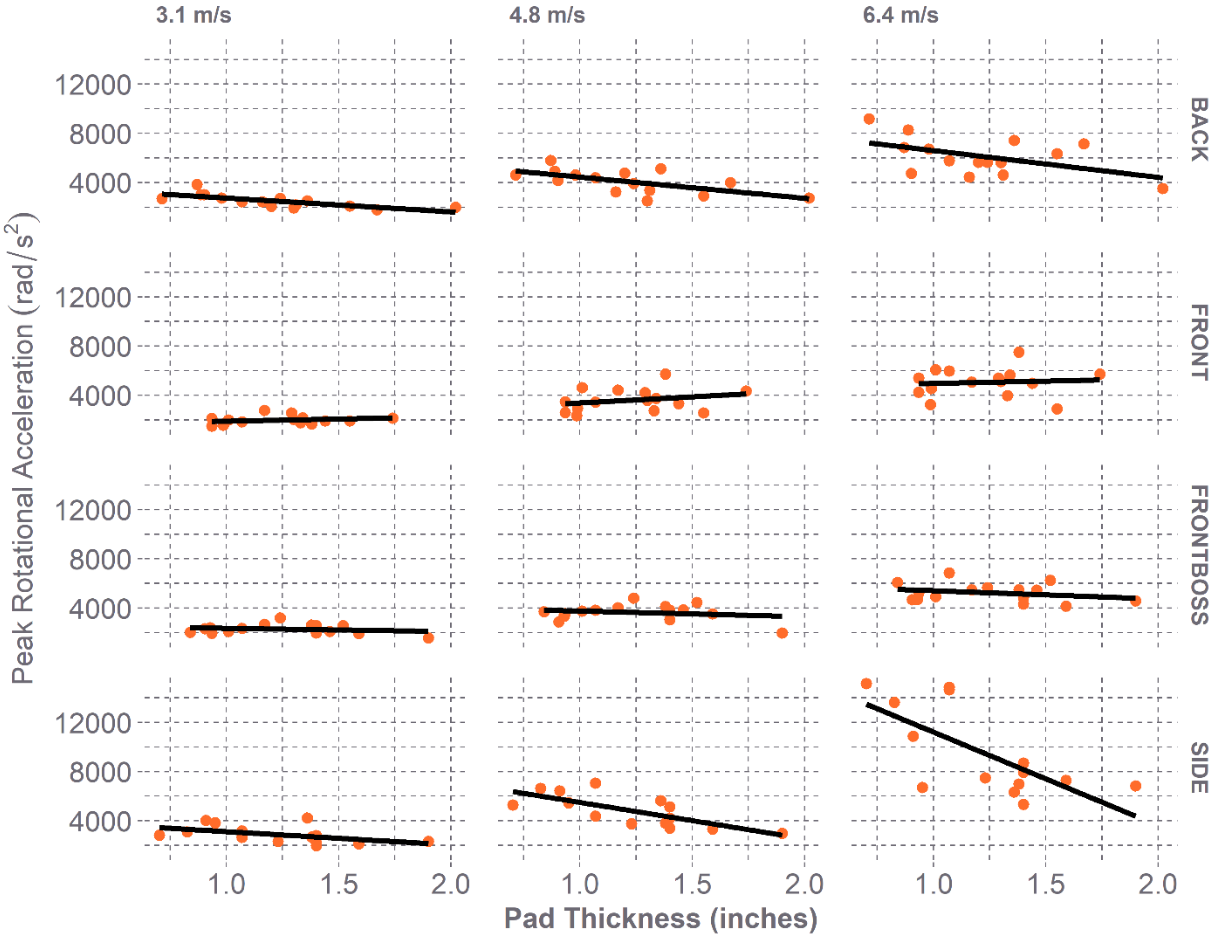

Figure 2.17: Pad thickness at each location versus peak rotational acceleration at each location and energy level. Generally, there is a negative correlation observed in the data.

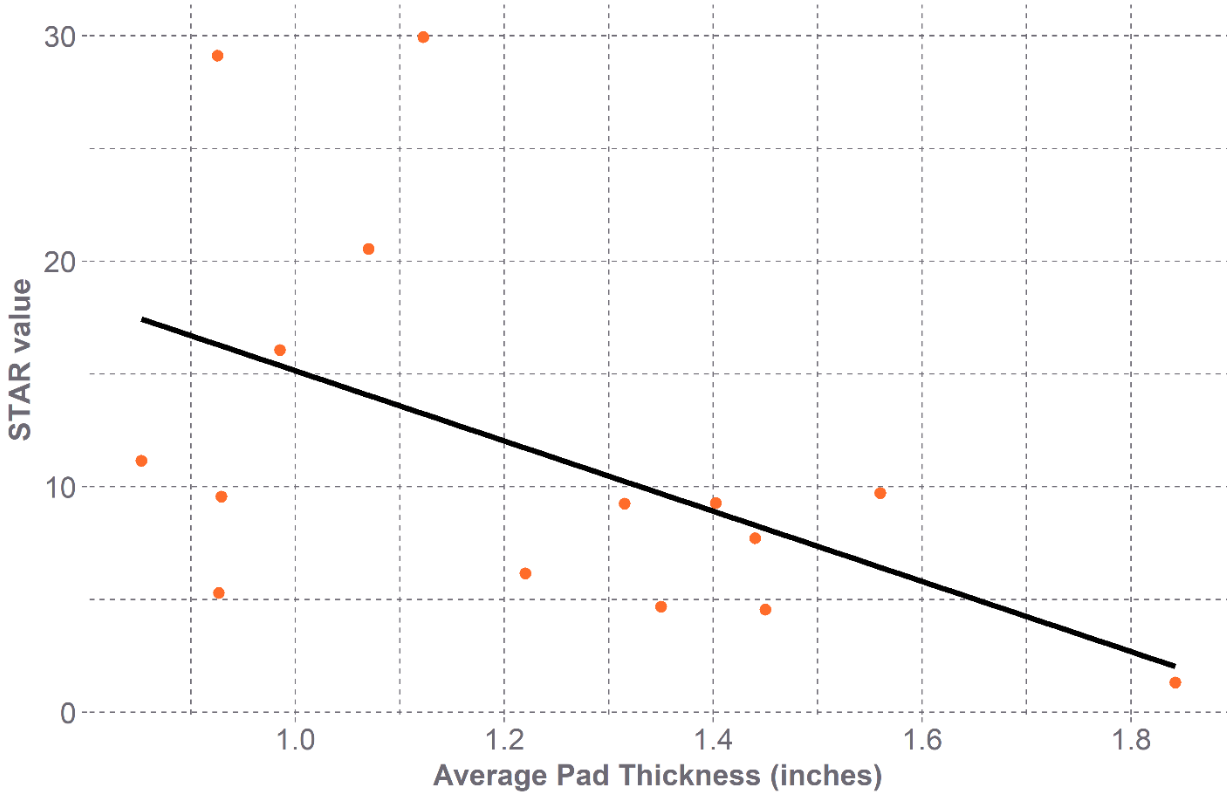

Figure 2.18: Average pad thickness versus STAR values for all helmets. There is a negative correlation (ρ = -0.56, p = 0.03)

Discussion

In this study, helmet performance and design were evaluated over time. Average peak linear acceleration decreased 30.04% between helmets manufactured before 1990 compared to helmets manufactured after 2010. Average peak rotational acceleration decreased 14.72% between eras. Between eras, average STAR values of helmets decreased 68%. Helmet average pad thickness, weight, and offset increased by 62%, 27.3%, and 17.98%, respectively, between time periods. All three helmet design features evaluated were positively correlated with helmet performance.

The difference in average peak linear acceleration of helmets manufactured before 1990 compared to helmets manufactured after 2010 was significant at all locations at the 6.4 m/s test condition. The largest changes were seen in energy level three because at higher energies, the same percentage change amounts to a greater magnitude change. Also, helmets have traditionally been designed and tested to mitigate accelerations at higher severities, so larger differences would be expected at higher impact energies. Although the other energy levels at each location do not show a significant change between time periods, the peak linear acceleration values still decreased between 10.72% and 41.80%.

The average peak rotational acceleration from before 1990 to after 2010 was significant at the side location at energy levels two and three. Significant differences were not observed at other locations, however, helmets between eras reduced peak rotational acceleration values at all locations, except for the front location at the 3.1 m/s and 4.8 m/s impact velocities. This lack of significance can be partially attributed to the inherently more variable measurement of rotational acceleration. Further, we evaluated a small sample of fifteen helmets, and only included two brands of helmets. The lack of significant decrease in peak rotational accelerations for all locations could be due to the design attributes of Schutt and Riddell helmets. Other helmet technologies not included in the study could have a greater effect on resulting rotational accelerations of the headform. Elements that enable shearing or provide non-normal compression would mitigate rotational accelerations.

There was a large difference in average STAR values between eras (68% decrease). Even with a small difference in an acceleration value at a specific test configuration, cumulative differences add up to large differences considering the hundreds of impacts that a player might experience during a season of play. There are two helmets with exceptionally high incidence values (Figure 2.9). These points correspond to the Schutt Air Advantage and the Schutt Air XP. Both of the incidence values are skewed by the helmet performance at the front location. The Schutt Air Advantage is in the lower 13%, 13%, 40%, for helmet weight, average pad thickness, and average offset out of the helmets tested, respectively. The Schutt Air XP is in the lower 33%, 47%, 66%, for helmet weight, average pad thickness, and average offset, respectively. With this in mind, there are likely other factors, like pad stiffness, which also play a role in why these helmets are separated from the group in incidence value.

Pad stiffness was not analyzed in this study, though it has been shown to have an effect on helmet performance. A softer pad is more effective at attenuating impact energy, given that it does not bottom out. At higher impact velocities, an adequately stiff pad is needed to prevent maximum compression [21]. Helmet design is an optimization problem of balancing pad stiffness with pad thickness. However, not all thick pads are as compliant as they could be to minimize accelerations. Contact area with the head also influences pad stiffness. If the pad has less contact area with the head, the force will be concentrated over the smaller area, influencing the performance of the pad. Pad stiffness could account for some of the unexplained variability in the helmet design characteristic analysis. When comparing helmets that have less difference in pad thickness, pad stiffness will arguably have a greater influence on helmet performance.

Overall, it can be concluded that helmet performance has continued to improve since the first NOCSAE certified football helmets release in the mid-1970s. Based on the helmets evaluated in this study, STAR values decreased 68% between NOCSAE certified helmets manufactured before 1990, and helmets manufactured after 2010. These results agree with on field data that compares the Riddell VSR4, first manufactured in 1992, to the Riddell Revolution, first manufactured in 2002. Collegiate players wearing the Riddell VSR4 were more likely to sustain a concussion than those wearing a Riddell Revolution when controlling for exposure [12].

There are several limitations in this study. All helmets tested that were manufactured before 1992 were acquired from independent sellers. In this way, our sample size was limited by which helmets could be found in good condition. Also, because of this, we do not know the amount of playing time each helmet bought from independent sellers has seen. However, football helmets are made to endure multiple impacts. Aside from usage, it is uncertain how the padding could have degraded since it was manufactured. All pads in helmets tested appeared to be in good condition, but it is possible the energy-absorbing characteristics had changed since being manufactured.

Conclusion

The purpose of this study was to evaluate helmet performance and design components over time. Testing was done on a pendulum impactor, and the Varsity Football STAR Methodology was used to analyze helmet performance. Overall helmet performance, quantified by a STAR value, improved over time. Peak linear acceleration significantly decreased between eras at the 6.4 m/s velocity, and peak rotational acceleration significantly decreased at the side location at 4.8 and 6.4 m/s. Helmet offset, pad thickness, and weight increased between eras. All of these components were also found to be negatively correlated with STAR values. In conclusion, helmet performance has improved over time and has been affected by changing helmet pad thickness, weight, and offset.

References

[1] M. L. Levy, B. M. Ozgur, C. Berry, H. E. Aryan, and M. L. Apuzzo, "Birth and evolution of the football helmet," (in eng), *Neurosurgery,* vol. 55, no. 3, pp. 656-61; discussion 661-2, Sep 2004.

[2] (2011). *NOCSAE DOC (ND)002-11m11a, Standard Performance Specification for Newly Manufactured Football Helmets*.

[3] "SAE J885 - human tolerance."

[4] M. Collins, M. R. Lovell, G. L. Iverson, T. Ide, and J. Maroon, "Examining concussion rates and return to play in high school football players wearing newer helmet technology: a three-year prospective cohort study," (in eng), *Neurosurgery,* vol. 58, no. 2, pp. 275-86; discussion 275-86, Feb 2006.

[5] D. C. Viano and D. Halstead, "Change in size and impact performance of football helmets from the 1970s to 2010," (in eng), *Ann Biomed Eng,* vol. 40, no. 1, pp. 175-84, Jan 2012.

[6] J. M. Hootman, R. Dick, and J. Agel, "Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives," (in eng), *J Athl Train,* vol. 42, no. 2, pp. 311-9, Apr-Jun 2007.

[7] K. Sarmiento, J. Mitchko, C. Klein, and S. Wong, "Evaluation of the Centers for Disease COntrol and Prevention's concussion initiative for high school coaches: "Heads Up:Concussion in High School Sports"," *J Sch Health,* vol. 80, no. 3, pp. 112-118, 2010.

[8] Congress of Neurological Surgeons, "Proceedings of the Congress of Neurological Surgeons in 1964: Report of the Ad Hoc Committee to Study Head Injury Nomeclature," *Clin Neurosurg,* vol. 12, pp. 386-394, 1966.

[9] R. C. Cantu, "Guidelines for return to contact sports after a cerebral concussion," *Physician Sportsmed,* vol. 14, no. 10, pp. 75-83, 1986.

[10] J. Kelly, J. Nichols, C. Filley, K. Lillehei, D. Rubinstein, and B. Kleinschmidt-DeMasters, "Concussion in Sports: Guidelines for the Prevention of Catastophic Outsomes," *JAMA,* vol. 266, pp. 2867-2869, 1991.

[11] P. McCrory *et al.*, "Consensus statement on Concussion in Sport 3rd International Conference on Concussion in Sport held in Zurich, November 2008," (in eng), *Clin J Sport Med,* vol. 19, no. 3, pp. 185-200, May 2009.

[12] S. Rowson *et al.*, "Can helmet design reduce the risk of concussion in football? Technical note," *Journal of Neurosurgery,* vol. 120, no. 4, pp. 919-922, 2014.

[13] S. Rowson, R. W. Daniel, and S. M. Duma, "Biomechanical performance of leather and modern football helmets," (in eng), *J Neurosurg,* vol. 119, no. 3, pp. 805-9, Sep 2013.

[14] S. Rowson and S. M. Duma, "Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," *Annals of Biomedical Engineering,* journal article vol. 39, no. 8, pp. 2130-2140, August 01 2011.

[15] B. R. Cobb, A. MacAlister, T. J. Young, A. R. Kemper, S. Rowson, and S. M. Duma, "Quantitative comparison of Hybrid III and National Operating Committee on Standards for Athletic Equipment headform shape characteristics and implications on football helmet fit," *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, vol. 229, no. 1, pp. 39-46, 2015.

[16] B. Rowson, S. Rowson, and S. M. Duma, "Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets," *Annals of Biomedical Engineering,* vol. 43, no. 10, pp. 2429-2443, 2015.

[17] B. R. Cobb, A. M. Zadnik, and S. Rowson, "Comparative analysis of helmeted impact response of Hybrid III and National Operating Committee on Standards for Athletic Equipment headforms," *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology,* vol. 230, no. 1, pp. 50-60, 2015.

[18] A. Tyson and S. Rowson, "Adult Football STAR Methodology," Virginia Tech2018.

[19] S. Rowson and S. M. Duma, "Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion," (in eng), *Ann Biomed Eng,* vol. 39, no. 8, pp. 2130-40, Aug 2011.

[20] S. Rowson and S. M. Duma, "Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration," (in eng), *Ann Biomed Eng,* vol. 41, no. 5, pp. 873-82, May 2013.

[21] C. Bradfield *et al.*, "Combat Helmet Suspension System Stiffness Influences Linear Head Acceleration and White Matter Tissue Strains: Implication for Future Helmet Design," *Military Medicine,* vol. 183, pp. 276-286, 2018.