Impact Biomechanics of the Head and Neck in Football

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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

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May 28, 2008
Blacksburg, VA

Keywords: concussion, stinger, helmet, linear, angular, acceleration

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Abstract

The research presented in the thesis explores the biomechanics of the head and neck during impacts in football. The research related to the head is geared towards advancing the current understanding of the mechanisms of mild traumatic brain injury, specifically by investigating head accelerations experienced by football players during impacts. To do this, a six degree of freedom sensor that could be integrated into existing football helmets and is capable of measuring linear and angular acceleration about each axis of the head was developed and validated. This sensor was then installed in the helmets of 10 Virginia Tech football players and data was recorded for every game and practice during the 2007 football season. A total 1712 impacts were recorded, creating a large and unbiased dataset. No instrumented player sustained a concussion during the 2007 season. From 2007 head acceleration dataset, 24 of the most severe impacts were modeled using a finite element head model, SIMon (Simulated Injury Monitor). Besides looking at head acceleration, the force transmitted to the mandible by chin straps in football helmets was investigated through impact testing. Little research has been conducted looking at the mandible-chin strap interface in the helmet, and this may be an area of helmet design that can be improved. The research presented in this thesis related to the neck is based on stingers. Football players wear neck collars to prevent stingers; however, their designs are largely based on empirical data, with little biomechanical testing. The load limiting capabilities of various neck collars were investigated through dynamic impact testing with anthropomorphic test devices. It was found that reductions in loads correlate with the degree to which each collar restricted motion of the head and neck. To investigate the differences in results that using different anthropomorphic test devices may present, the matched neck collar tests were performed with the Hybrid III and THOR-NT 50th percentile male dummies. The dummies exhibited the same trends, in that either a load was reduced or increased; however, each load was affected to a different degree.

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Chapter 1:
Head and Neck Injuries in Football

Opening Remarks

While catastrophic injuries in football are relatively rare, less severe injuries such as concussions and stingers occur frequently. These injuries are typically transient, in that their symptoms eventually resolve, but they are a major concern in competitive football due to their potential long term effects. Quantifying the biomechanics of these injuries presents a number of challenges, which are mainly a result of an inability to collect data from human volunteers at injurious levels. To study these injuries, unique methodologies that allow data to be collected in less than traditional ways must be utilized. The research presented in this thesis uses innovative methods that address the limitations of previous studies. All work is geared towards better understanding the injuries, in the hopes that the data will influence the design of protective equipment, resulting in a reduction of the occurrence of injuries in football.

Concussions

> "*Concussion is defined as a complex pathophysiological process affecting the brain, induced by traumatic biomechanical forces…*"

(Aubry et al., 2002)

A concussion can be caused by a direct blow to the head, neck, or elsewhere on the body that results in a force transmitted to the brain. Symptoms may include, but are not limited to, any of the following: headache, 'pressure in the head,' neck pain, dizziness, nausea or vomiting, vision and/or hearing problems, confusion, 'dazed' feeling, drowsiness, increased emotions, and amnesia. While symptoms typically resolve with 7-10 days, in some cases post-concussive symptoms may be prolonged or persistent (Aubry et al., 2002).

Each year in the United States, approximately 300,000 athletes sustain concussions while playing contact sports, with football having the largest occurrence (Thurman et al., 1998). This high incidence rate of concussions in football provides a unique opportunity to collect data that can characterize mild traumatic brain injury. The research presented in this thesis uses the football field as an experimental environment in which head acceleration data is collected by instrumenting football players' helmets with sensors.

Such data can have applications beyond the football field, as mild traumatic brain injuries are a major health concern in the United States. Quantifying human brain biomechanics as a result of impact will ultimately lead to a better understanding of human brain injury. This will influence all vehicle-related safety standards in terms of head injury. In addition, these data could serve as validation data for computation models, resulting in improved human tissue tolerance data.

Stingers

A stinger is most likely caused by injuring the upper trunk of the brachial plexus, which is made up of the C5 and C6 nerve roots (Robertson et al., 1979). This group of nerves runs from the cervical spine through the shoulder and into the upper arm, traveling directly under the clavicle. Stingers usually involve excessive hyperextension or lateral flexion of the head due to an impact, either with another player or with the ground. There are two main lateral flexion injury mechanisms: traction and compression. In a traction injury, the head is flexed laterally, and the brachial plexus ipsilateral to the impact is stretched. In a compression injury, lateral flexion combined with extension may lead to a pinching of the nerve roots when the foramina close on the contralateral side. This type of injury is usually very precise and local, while the stretching injury may occur anywhere along the plexus and is usually a more diffuse injury. Symptoms include numbness, pain, or a stinging or burning sensation in the shoulder and arm. Usually, these symptoms resolve within minutes (Clancy et al., 1977), but can escalate into long-term injuries (Hershman, 1990).

Stingers are a common injury in competitive football. Studies have shown lifetime injury incidences from 49% to 65% in college football (Clancy et al., 1977; Sallis et al., 1992). Many players will wear neck collars to prevent such injuries. These collar designs are based off empirical data, and few experiments have been conducted to quantify their effectiveness. The research presented in this thesis uses highly instrumented human surrogates to evaluate the load limiting capabilities of neck collars. By quantifying the effect of neck collars on the biomechanical response of the neck at injurious impact severities, insight to effective neck collar design can be acquired.

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Chapter 2:
A Six Degree of Freedom Head Acceleration Measurement Device for Use in Football

Abstract

Approximately 300,000 concussions are sustained by athletes playing contact sports in the US each year, with football having the largest occurrence. The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury. The goal of this study was to develop and validate a six degree of freedom (6DOF) measurement device that is designed to be integrated into existing football helmets. The new 6DOF sensor is capable of measuring linear and angular accelerations for each axis of the head. The 6DOF sensor consists of 12 accelerometers that integrate into football helmets. A novel algorithm processes the data from the accelerometers to determine linear and angular head accelerations. For validation, a football helmet equipped with the 6DOF sensor was fitted to a Hybrid III head instrumented with a 9 accelerometer array. The helmet was impacted using a pneumatic linear impactor in 5 locations at velocities ranging from 3.0 m/s to 9.0 m/s resulting in 114 tests. Hybrid III head accelerations were compared to that of the 6DOF sensor. Average errors for linear and angular head acceleration were 1% +/- 18% and 3% +/- 24%, respectively. The 6DOF HITS sensor can be used to quantify head accelerations experienced by football players considering a large dataset is collected.

Introduction

Each year, there are approximately 1.5 million traumatic brain injuries in the United States (Thurman et al., 1999); 75% of which are mild traumatic brain injuries (MTBI) (Sosin et al., 1996). About 300,000 of these concussions are sustained by athletes playing contact sports, with football having the largest occurrence (Thurman et al., 1998). The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data from humans to characterize MTBI. Competitive football has been used as an experimental environment for collecting human head acceleration data since the 1970's. Several studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games (Moon et al., 1971; Reid et al., 1971; Reid et al., 1974). Another study instrumented football helmets directly to measure helmet acceleration (Morrison, 1983). While laying the groundwork for future research and providing a proof of concept, these studies were limited in their ability to measure head acceleration and measured only a single player.

One study has quantified head accelerations experienced by football players by recreating concussive impacts. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on game video (Newman et al., 1999; Newman et al., 2000; Pellman et al., 2003a). This study was limited by a biased dataset, as only selected impacts could be reconstructed. More recently, a study has quantified head accelerations by instrumenting helmets worn by collegiate football players (Duma et al., 2005; Funk et al., 2007). In this study, a six accelerometer sensor was integrated into football helmets. These sensors recorded resultant linear head acceleration for every head impact a player experienced, producing a large and unbiased dataset. Over 27,000 head impacts were recorded over 4 seasons, 4 of which were concussive. The main limitation of this study is that angular acceleration was not directly measured by the 6 accelerometer sensor.

The 6 accelerometer sensor used by Duma et al. (2005) is part of the Head Impact Telemetry System (HITS), developed by Simbex, LLC (Lebanon, NH). The HITS sensor consists of 6 nonorthogonally mounted single-axis accelerometers which are positioned normally to the head. The packaging of the accelerometers includes an integrated radio board that communicates wirelessly with a computer on the sideline. All accelerometers are spring mounted so that they remain in contact with the head at all times and oriented normal to the head. The HITS sensor was shown to measure head acceleration by impacting a helmeted Hybrid III head and then comparing the accelerations of the helmet shell and Hybrid III head's center of gravity to the HITS sensor. The HITS sensor acceleration was in strong agreement with the Hybrid III head acceleration (Manoogian et al., 2006). Each time an impact occurs, data is transmitted from the sensor to the computer, which processes and displays data in real-time. Data are collected for 40 ms, of which 12 ms are pre-trigger and 28 ms are post-trigger data. HITS utilizes a novel algorithm for determining impact magnitude and direction (Crisco et al., 2004). This algorithm is capable of calculating resultant linear acceleration throughout time. HITS also estimates peak x and y axis rotational acceleration, but cannot completely model the head kinematics due to the unknown time histories of linear and angular acceleration for each axis.

In order to collect data from football players capable of accurately characterizing head kinematics, a sensor with the ability of measuring linear and angular acceleration about each axis of the head needed to be developed. The goal of this study was to develop and validate a six degree of freedom (6DOF) measurement device that is designed to be integrated into existing football helmets. Using a 6DOF sensor, human head acceleration data can be recorded that is capable of completely modeling the head kinematics resulting from impacts in football. Such data are ideal for the development of injury risk curves and the validation of computational models.

Methods

The new 6DOF HITS sensor is designed to be integrated into Riddell Revolution football helmets. The sensor is composed of two primary pieces: vinyl casing and fabric padding (Figure 1). Velcro is used to attach the vinyl casing of the sensor to the helmet between its padding. The vinyl casing serves as the housing for all the electronics, with exception to the accelerometers. The fabric pad contains the accelerometers inside. 12 accelerometers are enclosed in the fabric padding, positioned in orthogonally oriented pairs at 6 different locations. All accelerometers are orientated so that their sensing axes are tangential to the skull. The fabric pad also serves as a spring to keep the accelerometers in contact with the head. When the helmet is impacted, the padding inside the helmet compresses and the helmet shifts positions on the head. However, the fabric pad containing the accelerometers either compresses or expands to remain in contact with the player's head. This ensures that head acceleration, not helmet acceleration, is measured (Manoogian et al., 2006).

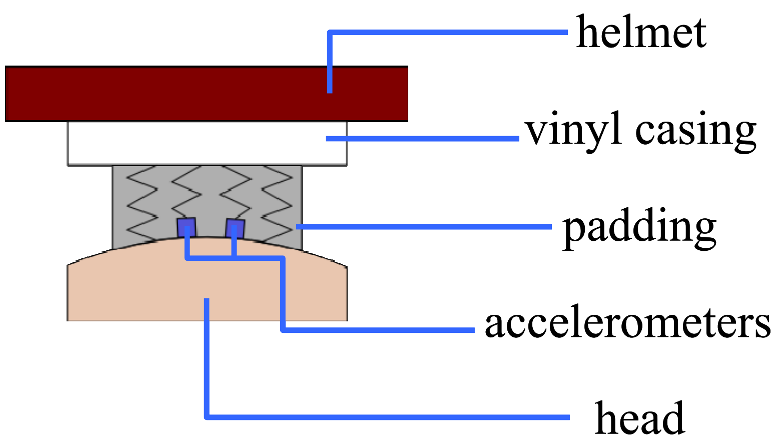

Figure 1: Schematic of 6DOF sensor.

[Image created by Steve Rowson]

The 6DOF HITS sensor utilizes 12 single-axis, high-g iMEMS accelerometers (ADXL193, Analog Devices, Norwood, MA). Data acquisition is triggered when any accelerometer exceeds 10 g. Data is collected for 40 ms at 1000 Hz, of which 8 ms are pre-trigger and 32 ms are post-trigger. After each impact is recorded, the data is sent to a computer via a 903-927 MHz wireless transceiver. If communication cannot be established with the computer, the sensor has enough memory to store up to 120 impacts. Stored impacts are transmitted to the computer once communication is reestablished. For each impact, linear and angular acceleration for each axis of the head are measured. In addition, impact location is recorded.

The algorithm that solves for linear and resultant acceleration of the head's center of gravity (CG) uses rigid body kinematics. Equation 1 sums the linear and rotational accelerations to calculate what each accelerometer should be reading. $\|a_i\|$ is the acceleration magnitude at each individual accelerometer, $\overrightarrow{r_{ai}}$ is the orientation of the sensing axis of each accelerometer, $\overrightarrow{H}$ is the head CG linear acceleration, $\vec{\alpha}$ is angular acceleration about the head CG, $\vec{r_i}$ is the accelerometer location relative to the head CG, and $\overrightarrow{\omega_i}$ is the angular velocity about the head CG.

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

Since the accelerometers are oriented tangentially to the skull, the centripetal acceleration term of Equation 1 is negligible, simplifying Equation 1 to Equation 2. Eliminating $\overrightarrow{\omega_i}$ simplifies the equation from a nonlinear differential equation to an equation that can be solved algebraically.

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

Since 12 accelerometers are in the sensor, the solution can be optimized. The algorithm uses an iterative optimization approach to solve for linear and angular acceleration (Chu et al., 2006).

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

A 6DOF HITS sensor was installed in a medium Riddell Revolution helmet, which was fitted on the Hybrid III head. The medium-sized jaw pads of the helmet were replaced with large jaw pads to better fit the narrow face of the Hybrid III head. A helmet positioning tool was used to ensure that the helmet's fit on the head was consistent. This tool used landmarks on the helmet and face to position the helmet on the head. The chin strap of the helmet was used to secure the helmet on the head. An air pump was used to inflate the padding of the helmet until the helmet could not change in position relative to the head.

The helmeted Hybrid III head was struck with the pneumatic linear impactor with several combinations of impact velocities and locations. Impact velocities were chosen so that they would simulate a range of impact severities typically experienced in tackling and blocking. The impact velocities ranged from 3.0 to 9.0 m/s and are based on the NFL reconstruction data (Pellman et al., 2003a). To account for the various ways a helmet can be struck, 5 impact locations were chosen based on NFL video analysis (Pellman et al., 2003b). Impact locations ranged from the front to the backside of the helmet (). Each configuration of impact velocity and location were tested in a minimum of 4 trials.

The Hybrid III is the industry standard of human surrogates for modeling human kinematics and dynamics in the automotive industry. For this reason, the Hybrid III's acceleration values are referenced in this study as the acceleration a human head would experience during such impacts. Therefore, the algorithm was optimized so that the 6DOF sensor produces acceleration values equivalent to that of the Hybrid III's.

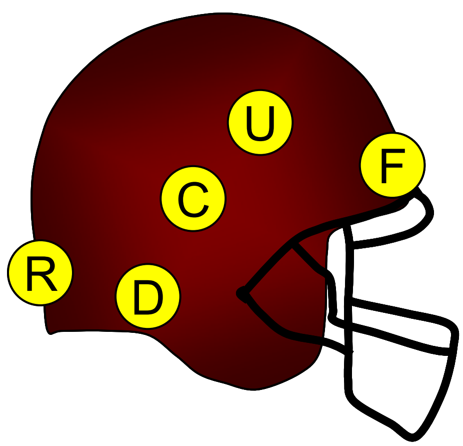

Figure 2: 5 locations on the helmet were impacted at a range of velocities.

[Image created by Steve Rowson]

The 3-2-2-2 accelerometer array in the Hybrid III head was used to calculate linear and angular accelerations about each axis. Figure 3 displays the configuration of the 3-2-2-2 array. Linear accelerations were defined by the acceleration along each axis at the center of gravity of the head (a_{x0}, a_{y0}, and a_{z0}). Equations 3, 4, and 5 were used to calculate angular accelerations about each axis (Padgaonkar et al., 1975). α_x is angular acceleration about the x-axis, α_y is angular acceleration about the y-axis, and α_z is angular acceleration about the z-axis.

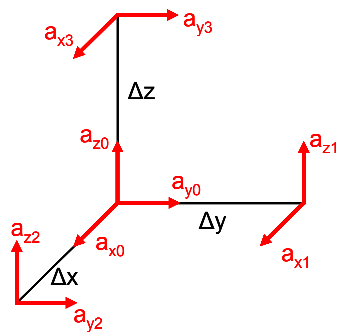

Figure 3: 3-2-2-2 accelerometer array configuration; where each red arrow represents an accelerometer.

[Image created by Steve Rowson]

$$\alpha_x = \frac{a_{z1} - a_{z0}}{2 \cdot \Delta y} - \frac{a_{y3} - a_{y0}}{2 \cdot \Delta z} \tag{3}$$

$$\alpha_y = \frac{a_{x3} - a_{x0}}{2 \cdot \Delta z} - \frac{a_{z2} - a_{z0}}{2 \cdot \Delta x} \tag{4}$$

$$\alpha_z = \frac{a_{y2} - a_{y0}}{2 \cdot \Delta x} - \frac{a_{x1} - a_{x0}}{2 \cdot \Delta y} \tag{5}$$

Results

The linear and angular accelerations computed from the 3-2-2-2 array in the Hybrid III head were compared to the accelerations produced by the 6DOF HITS sensor. Figure 4 compares the time series acceleration response for the Hybrid III and 6DOF HITS sensor for an impact to location D at 8.0 m/s. The 6DOF HITS measured acceleration closely matches that of the Hybrid III.

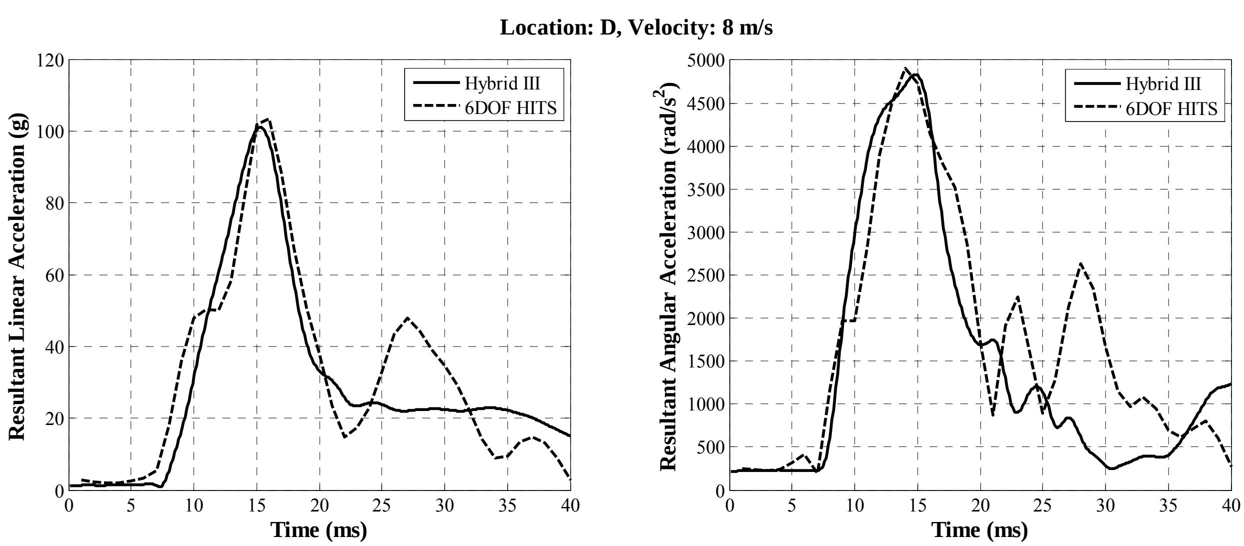

Figure 4: Resultant acceleration traces for a selected impact.

Figure 5 displays the linear relationship between the Hybrid III and 6DOF HITS peak resultant linear and angular accelerations. Peak resultant linear acceleration correlated strongly (R^2 = 0.88) between the Hybrid III and 6DOF HITS. The average error between the 6DOF HITS and

Hybrid III resultant linear acceleration is 1% +/- 18%. Peak resultant angular acceleration between the 6DOF HITS and Hybrid III also correlated strongly (R^2 = 0.85) with an average error of 3% +/- 24%.

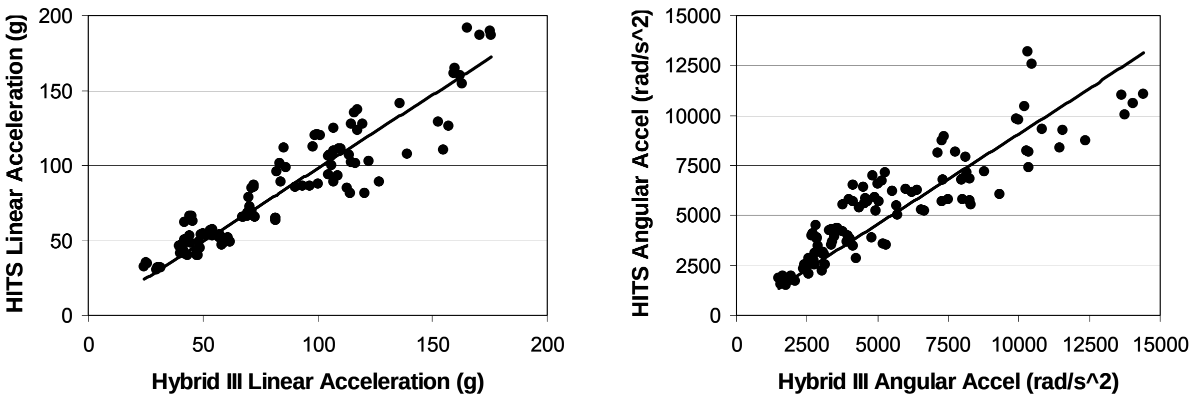

Figure 5: Linear relationship between 6DOF HITS and Hybrid III peak resultant accelerations. Linear (R^2 = 0.88 and angular (R^2 = 0.85) correlated strongly with Hybrid III head acceleration.

Discussion

Based on preliminary testing, facemask impacts were not included in this testing due to an unrealistic interaction between the impactor face and the facemask. When impacting the facemask, the facemask would bend, which is not typical of facemask impacts in football. Substantial stroking could also be seen on the chin strap in these impacts. This is where the chin strap slides through its grips, allowing the helmet's position relative to the head to change. This is also not typically seen in on-field football impacts. Upon inspection of high speed video of the preliminary tests, it could be seen that the interaction between the impactor face and facemask is not representative of a helmet-to-facemask impact. For these reasons, facemasks impacts were not included in testing.

Similar testing has been used to validate instrumented boxing headgear (IBH) for use during boxing competition (Beckwith et al., 2007). In this study, IBH acceleration was shown to correlate strongly with Hybrid III head acceleration for linear (R^2 = 0.91) and angular (R^2 = 0.91) acceleration. The 6DOF HITS sensor utilizes a similar algorithm (Chu et al., 2006) and technology as the IBH. Differences in accuracy between the two systems can be accounted to several factors. For one, the boxing head gear fits differently on the Hybrid III head than a football helmet. Boxing head gear has a tighter fit than a football helmet, which is one reason for the IBH's stronger correlation. In addition, the validation datasets for the IBH and 6DOF HITS contain different ranges of head acceleration. The differing acceleration ranges in the IBH and 6DOF HITS datasets contribute to the differences in correlation to Hybrid III head acceleration between the two systems, as the 6DOF HITS sensor was validated over a much larger range of head accelerations.

The maximum linear and angular accelerations in the IBH validation testing were 77.3 g and 6,433 rad/s^2, respectively. These maximum values are similar to the nominal injury values representing 50% risk of concussion published through analysis of the NFL data, which are 79 g and 5757 rad/s^2 (King et al., 2003). These values have been shown to be conservative estimates of concussion risk (Funk et al., 2007). Therefore, it was important to have a validation dataset for the 6DOF HITS sensor that encompasses a greater range of accelerations, as the ultimate goal of the sensor is to accurately measure head accelerations for all non-concussive and concussive head impacts an instrumented player may experience. The maximum linear and angular accelerations in the 6DOF HITS sensor's validation testing were 176 g and 14,431 rad/s^2, respectively. According to the NFL risk curves, these values would represent a 100% probability of concussion.

There is some inherent error within the measurement of head acceleration using the 6DOF sensor. Possible sources of error include the helmet changing in position relative to the head throughout an impact and non-ideal orientation of the accelerometers to the head. Error levels in the 6DOF acceleration measurements are within the range of error seen with other measurement devices and techniques. The NFL video analysis was reported to have error as high as 15% (Pellman et al., 2003b); while the original HITS technology has an error of 8% +/- 11% (Funk et al., 2007). In addition, chest bands, which are used to measure deflection of the chest, can have error as high as 10% (Rath et al., 2005). Considering the vast amounts of data that can be collected with the 6DOF HITS sensor on human volunteers and the error levels of other biomechanical experiments, the error inherent in the 6DOF sensor is acceptable.

While the 6DOF error is considered acceptable, additional methods can be applied to the dataset to minimize the statistical effects of data scatter. The standard deviations contribute to the majority of the error, as the average errors are minimal. Funk et al. (2007) presented a unique methodology for minimizing error due to scatter in a large dataset. Using this same technique, the effects scatter error in a 6DOF dataset can be incorporated into risk functions.

While the original HITS sensor is capable of measuring resultant linear acceleration and impact location, its applications are limited by its inabilities to measure angular acceleration and data traces for each individual axis. The 6DOF HITS sensor provides individual data traces for linear and angular head acceleration. This technology provides the opportunity to collect a large and unbiased dataset, since every head impact that an instrumented football player experiences would be recorded. Considering this hypothetical dataset contains non-injurious and concussive impacts, applications of the dataset would include the development of injury risk curves and the validation of computational head models. This could ultimately lead to a better understanding of the mechanisms of concussion.

Acknowledgements

Toyota Central Research and Development Labs and the National Highway Traffic and Safety Administration for funding the development of the 6DOF HITS sensor.

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Chapter 3:
Linear and Angular Head Acceleration Measurements in Collegiate Football

Abstract

Background: Each year, about 300,000 concussions are sustained by athletes playing contact sports, with football having the highest occurrence. The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data to characterize mild traumatic brain injury.

Hypothesis: Human head acceleration data for a range of impact severities can be collected by instrumenting the helmets of football players with accelerometers.

Study Design: Descriptive laboratory study.

Methods: The helmets of 10 Virginia Tech football players were instrumented with sensors for every game and practice for the 2007 football season. The sensors recorded linear and angular accelerations about each axis of the head. Data for each impact was downloaded wirelessly by a sideline computer shortly after each impact occurred.

Results: Data was collected for of 1712 impacts, creating a large and unbiased dataset. While the majority of the impacts were relatively low severity (< 30 g and < 2000 rad/s^2), 172 impacts were greater than 40 g and 143 impacts were greater than 3000 rad/s^2. No instrumented player sustained a concussion during the 2007 season.

Conclusions: A large and unbiased dataset was compiled by instrumenting the helmets of collegiate football players. Football provides a unique opportunity to collect head acceleration data of varying severity from human volunteers. Future collection of concussive data may advance the understanding of the mechanics of mild traumatic brain injury.

Clinical Relevance: With an increased understanding of the biomechanics of head impacts in football and human tolerance to head acceleration, better equipment can be designed to prevent head injuries.

Key Terms: concussion, brain injury, human tolerance.

Introduction

Each year, there are approximately 1.5 million traumatic brain injuries in the United States;[1] 75% of which are mild traumatic brain injuries (MTBI).[2] About 300,000 of these concussions are sustained by athletes playing contact sports, with football having the largest occurrence.[3] The high incidence rate of concussions in football provides a unique opportunity to collect biomechanical data to characterize MTBI. Several injury metrics are used to predict head injury; however, all the criteria use limited data from human volunteers. Head injury criterion (HIC), peak acceleration, and severity index (SI) are injury metrics derived from linear head acceleration and are primarily based on cadaver tests with skull fractures. Rotational acceleration injury thresholds are based mostly on primate tests with severe concussion, diffuse axonal injury (DAI), or intracranial bleed.

Competitive football has been used as an experimental environment for collecting human head acceleration data since the 1970's. Several studies have had football players wear headbands instrumented with accelerometers to measure head acceleration during football games.[4-6] Another study instrumented football helmets directly to measure helmet acceleration.[7] While laying the groundwork for future research and providing a proof of concept, these studies were limited in their ability to measure head acceleration and measured only a single player. More recently, hockey and football helmets were instrumented with accelerometers to measure linear head acceleration.[8] However, there were no incidents of mild traumatic brain injury in this study.

One study has quantified head accelerations experienced by football players by recreating concussive impacts. The National Football League (NFL) reconstructed injurious game impacts using Hybrid III dummies based on game video.[9-11] The authors recreated 31 impacts, 25 of which were concussive. From the data collected in the reconstructed impacts, injury risk curves were developed for MTBI. Nominal injury values determined in this study were a peak linear acceleration of 79 g, SI of 300, HIC of 250, and peak rotational acceleration of 5757 rad/s^2.[11, 12] The main limitation of this study is that the NFL data is biased towards injurious impacts.

Another study has quantified head accelerations by instrumenting helmets worn by collegiate football players.[13] In this study, a six accelerometer sensor was integrated into football helmets. These sensors recorded resultant linear head acceleration for every head impact a player experienced, producing an unbiased dataset. Over 27,000 head impacts were recorded over 4 seasons, 4 of which were concussive. Although there is a limited injury dataset, this study gives insight to the lower limits of human tolerance to head acceleration. Using a unique statistical analysis on this data, injury risk curves were developed.[14] The nominal injury values reported representing 10% risk of concussion were a peak linear acceleration of 165 g and HIC of 400. The main limitation of this study is that angular acceleration was not directly measured by the 6 accelerometer sensor.

The goal of this study was to utilize a newly developed six degree of freedom sensor to record 6 degree of freedom (6DOF) head accelerations for every head impact experienced by collegiate football players, producing a large and unbiased dataset. Data collected in this experiment has applications in validating computational models and creating risk curves based on finite element analysis of head impacts.

Methods

A new 6DOF sensor has been developed that is capable of measuring linear and angular accelerations for each axis of the head. The 6DOF sensor consists of 12 accelerometers and integrates into football helmets. A novel algorithm processes the data from the accelerometers to determine linear and angular head accelerations.[15] The sensor was validated through impact testing with a 50th percentile male Hybrid III anthropomorphic test dummy. The 6DOF sensor was shown to have and average error of 1% +/- 18% for linear acceleration and 3% +/- 24% for angular acceleration.

Using a similar methodology to Duma et al. (2005), 6DOF sensors were installed in the helmets of 10 Virginia Tech football players during the 2007 season. Each player that participated in the study gave written informed consent with Institutional Review Board approval from both Virginia Tech and the Edward Via College of Osteopathic Medicine. Head acceleration data were recorded for every practice and game each player participated in. During practices, the sideline computer was stationed next to the practice field and downloaded impacts as they happened throughout practice. During games, the sideline computer was set up at the 40 yard line and downloaded impacts as they happened throughout each game. X, y, and z linear and angular acceleration traces were recorded for every impact instrumented players experienced during games and practices throughout the 2007 Virginia Tech football season.

The coordinate system referenced in this paper is that of the SAE J211. The positive x-axis runs out of the face (perpendicular to the coronal plane), the positive y-axis runs out of the right ear (perpendicular to the sagittal plane), and the positive z-axis runs out of the bottom of the head (perpendicular to the transverse plane).

Results

A total of 1712 impacts were recorded during practices and games for the 10 instrumented players during the 2007 Virginia Tech football season. 570 of the recorded impacts occurred during games; while 1142 occurred during practices. Table 1 displays the total number of impacts each player experienced during games and practices. Not all instrumented players were starters on the football team, which explains the variation in the number of game impacts between players. No instrumented player sustained a concussion during the 2007 season.

Table 1: Total number of impacts for games and practices per player.

ID	Games	Practices	Total
55	14	129	143
58	303	315	618
66	48	103	151
71	9	31	40
72	0	122	122
74	0	35	35
79	0	99	99
82	66	163	229
91	130	122	252
93	0	23	23
Totals	570	1142	**1712**

Table 2 displays the frequency of impacts over specified resultant acceleration thresholds for linear and angular acceleration. For resultant linear acceleration, thresholds are in 20 g increments. The majority of the impacts were under 20 g in severity. 10% of the impacts were greater 40 g in severity. Of the 1712 impacts, 11 were greater than 79 g, which is the nominal injury value derived by the NFL study. For resultant angular acceleration, thresholds are in 1000 rad/s^2 increments. Roughly half of the impacts were less than 1000 rad/s^2 in severity. Only 143 of the impacts were greater than 3000 rad/s^2 in severity. Of the 1712 impacts, 14 were greater than 5757 rad/s^2, which is the nominal injury value derived in the NFL study.

Table 2: Frequency of impacts above specified resultant acceleration thresholds.

Linear Acceleration	Number of Impacts	Angular Acceleration	Number of Impacts
> 0 g	1712	> 0 rad/s^2	1712
> 20 g	684	> 1000 rad/s^2	875
> 40 g	172	> 2000 rad/s^2	339
> 60 g	52	> 3000 rad/s^2	143
> 80 g	11	> 4000 rad/s^2	57
>100 g	3	> 5000 rad/s^2	23
> 120 g	1	> 6000 rad/s^2	12
> 140 g	0	> 7000 rad/s^2	5
> 160 g	0	> 8000 rad/s^2	4
> 180 g	0	> 9000 rad/s^2	1

Figure 6 displays histograms of the distributions of resultant linear and angular acceleration. Linear accelerations ranged from 9 g to 135 g. The majority of the impacts were under 20 g in severity. At 20 g, the numbers of impacts begin to decrease quickly. By 70 g, the numbers of impacts decrease to values close to zero. Angular accelerations ranged from 107 rad/s^2 to 9922 rad/s^2. The majority of the impacts were under 2000 rad/s^2 in severity. At 1000 rad/s^2, the numbers of impacts begin to decrease quickly. By 5000 rad/s^2, the numbers of impacts decrease to values close to zero.

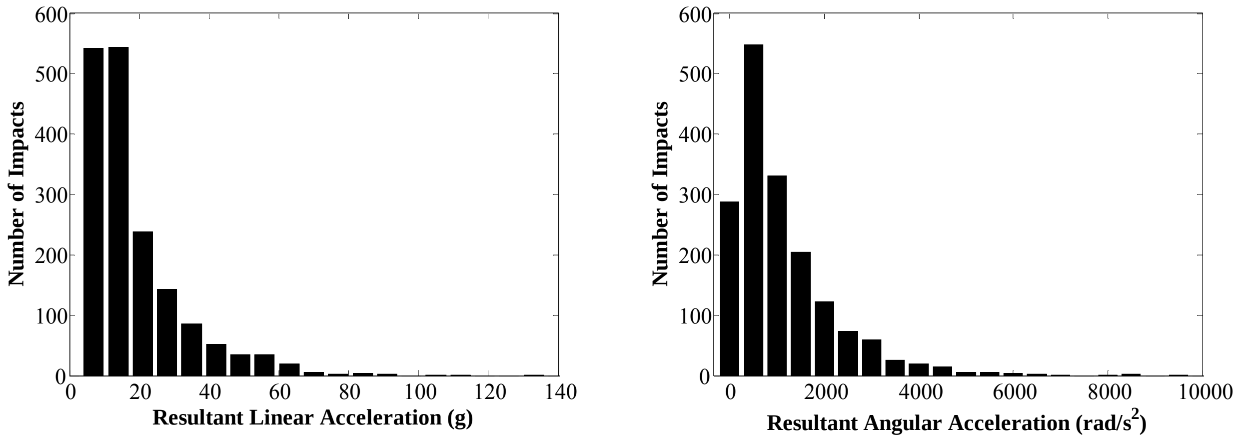

Figure 6: Distributions of linear and angular accelerations.

Figure 7 compares the distributions of each axis's peak resultant linear acceleration for every recorded impact. The distributions of peak resultant linear acceleration are similar for each axis; in that they all peak near 10 g, and then the frequencies begin to decrease quickly. However, some differences can be observed through examining the plots. Linear acceleration along the y axis peaked under 10 g more frequently than x and z axis acceleration. In addition, y axis acceleration peaked greater than 20 g less frequently than x and z axis acceleration. Linear acceleration along the z axis peaked greater than 20 g more often than x and y axis acceleration. X axis acceleration frequencies were often between that of the y and z axes' accelerations. Large peak linear accelerations (>60 g) were most common along the z axis; while low peak linear accelerations (<10 g) were most common along the y axis. X axis peak acceleration frequencies fell between that of the y and z axes' accelerations.

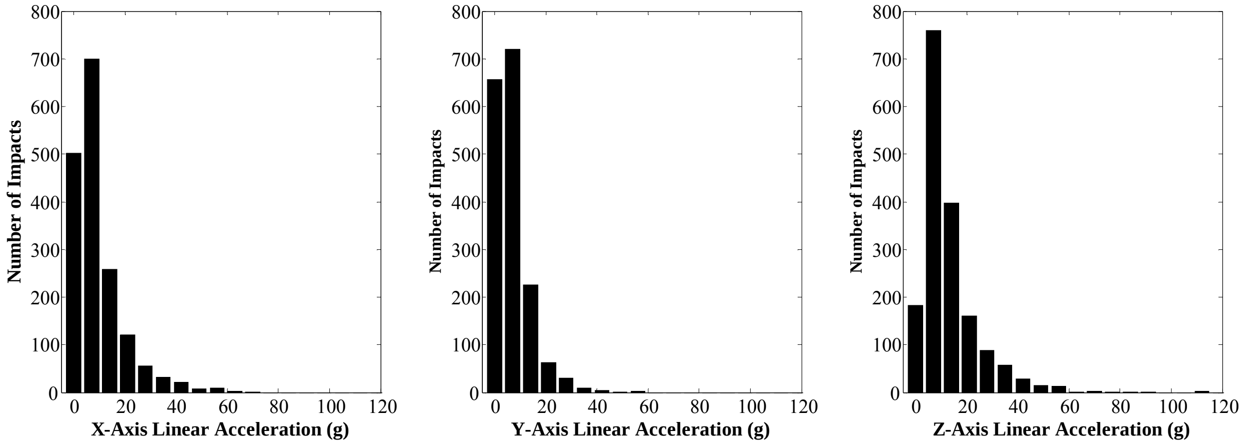

Figure 7: Distributions of linear accelerations about each axis of the head.

Figure 8 compares the distributions of each axis's peak resultant angular acceleration for every recorded impact. The distributions of peak resultant angular acceleration are similar for the x and y axes; in that they both peak at less than 500 rad/s^2, and then the frequencies begin to decrease quickly. The distribution of angular acceleration about the z axis is different than that of the x and y axes. Angular acceleration about the z axis peaked greater than 1000 rad/s^2 more often than x and y axis angular acceleration. Large peak angular accelerations (>4000 rad/s^2) were most common about the z axis; while low peak low angular accelerations (<1000 rad/s^2) were more common about the x and y axes.

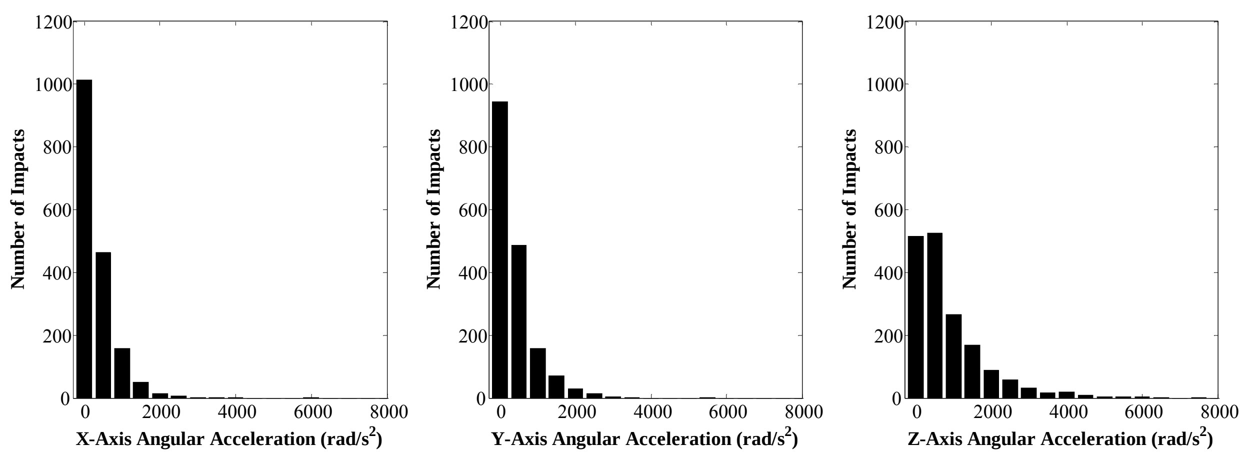

Figure 8: Distributions of angular acceleration about each axis of the head.

For each impact, azimuth and elevation are recorded to identify where the helmet was impacted. Azimuth (θ) is defined as the angle between the impact location and negative x axis in the x-y plane. Elevation (α) is defined as the angle between the impact location and the x-y plane. The head was divided into sections to generalize each impact location. Figure 9 displays the groups impacts based on their impact location. Impacts to the front of the helmet were most common with 704 impacts. A total of 573 impacts were to the sides of the helmet. Back and top impacts were the least common with 220 and 215 impacts, respectively.

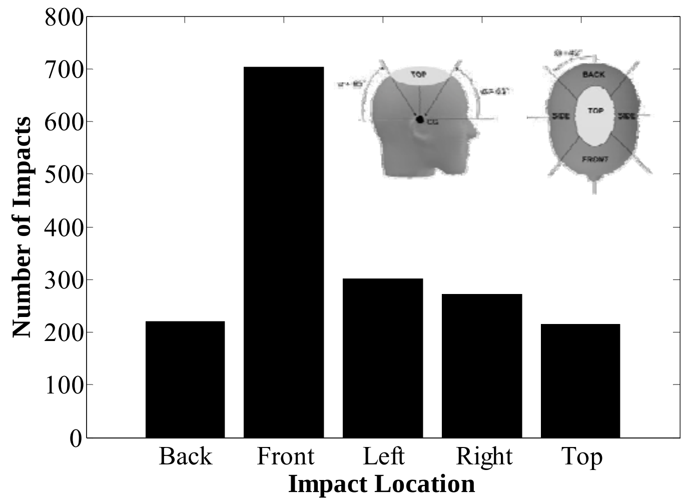

Figure 9: Distribution of impact locations broken into back, front, left, right, and top bins. Bins are defined in the top right corner of the histogram.

The average duration of the 1712 impacts was 14 ms (Figure 10). Impact duration and acceleration magnitude are ultimately responsible for the change in velocity of the head. Change in linear head velocity and angular velocity are displayed in Figure 11 as a function of their respective peak accelerations. Delta V ranged from 0.3 m/s to 6.1 m/s and did not correlate well with peak linear acceleration (R^2 = 0.49). Angular velocity ranged from 0.5 rad/s to 42.5 rad/s and did not correlate strongly with peak angular acceleration (R^2 = 0.68).

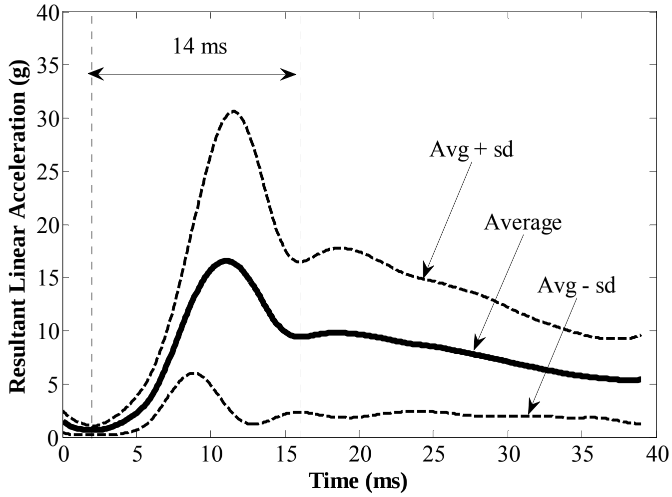

Figure 10: Average linear acceleration response for the 1712 impacts. The average impact duration was 14 ms.

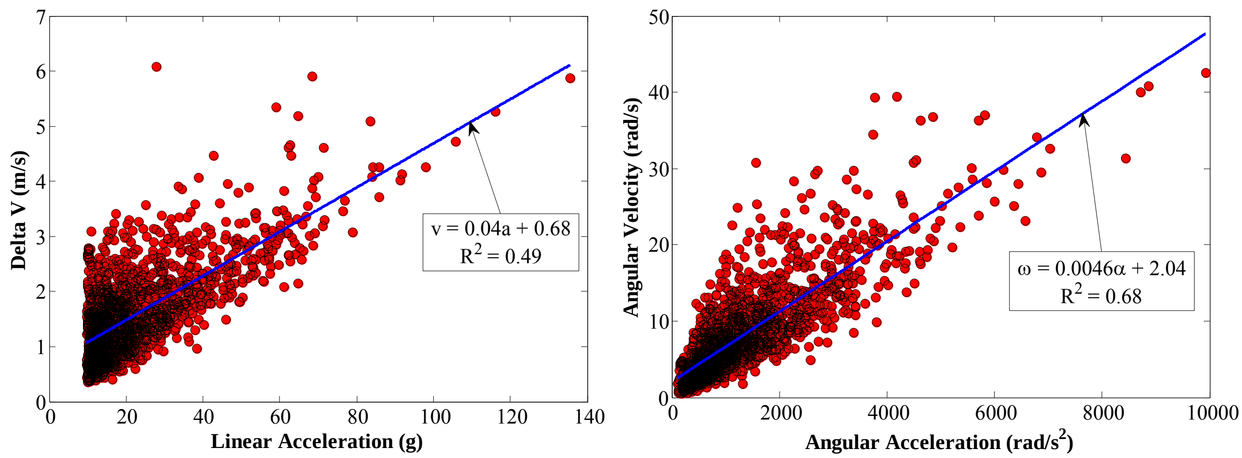

Figure 11: Relationship between change in velocity (Delta V) of the head versus peak linear acceleration (left). Relationship between angular velocity and angular acceleration (right).

Discussion

While the original HITS sensor is capable of measuring resultant linear acceleration and impact location, its applications are limited by its inabilities to measure angular acceleration and data traces for each individual axis. The 6DOF sensor provides individual data traces for linear and angular acceleration. Using this sensor, a large and unbiased dataset on human head acceleration was compiled. The biomechanical response of the human head to impact can be investigated at the organ level by examining the acceleration data. Computational models can be used to determine the tissue level response of the head to impact by examining the resulting stresses and strains of each impact.

The 6DOF dataset shows that football players routinely experience head accelerations up to 40 g and 3000 rad/s^2. While no concussions were measured, 11 impacts had peak linear accelerations greater than 79 g and 14 impacts had peak angular accelerations greater than 5757 rad/s^2, which are the nominal injury values for concussions based on the NFL data. This is consistent with data collected by Virginia Tech using the original HITS sensor throughout the 2003-2007 seasons. Out of 504 impacts with peak linear accelerations greater than 80 g, only 4 resulted in concussions. The injury values reported by the NFL are biased towards injurious impacts, which results in conservative injury risk values. Had the NFL reconstructed more impacts that did not result in concussion, perhaps their values for 50% probability of concussion would be higher. Due to the fact that no concussions were sustained by instrumented players in this study, risk curves cannot be developed. By collecting data over the next several seasons using the 6DOF sensor, it is expected that several concussions will be recorded due to an increased number of instrumented players. The resulting dataset can then be used to create risk curves that more accurately predict risk of concussion in football.

Over 27,000 impacts were recorded using the original HITS sensor throughout the 2003-2006 Virginia Tech football seasons.[13, 14] Figure 12 compares the resultant linear accelerations collected throughout the 2003-2006 seasons using the original HITS sensor and 2007 season using the 6DOF sensor. Since there were a different number of impacts in each dataset, frequencies were normalized to be a percentage. The distributions are very similar to one another. The main difference is that the 2003-2006 dataset collected a higher percentage impacts greater than 20g.

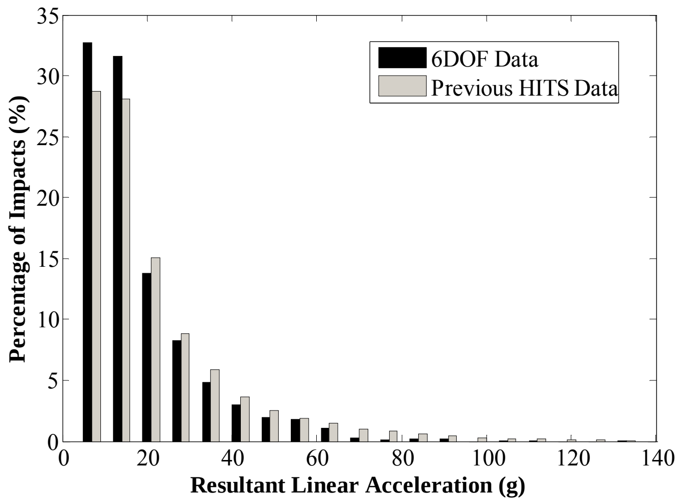

Figure 12: Comparison of linear accelerations collected with the 6DOF sensor during 2007 Virginia Tech football season and the original HITS sensor during the 2003-2006 seasons.

When comparing the accelerations about each axis, large angular accelerations about the z axis were most common (Figure 8). A hypothesis for this is that the high z angular accelerations are due to the large moment arm resulting from the facemask's distance away from the center of gravity of the head. The majority of impacts were to the front of the helmet.

Figure 13 plots peak resultant angular acceleration against peak resultant linear acceleration for every recorded impact. This plot suggests that there is no strong correlation between peak linear and angular acceleration. Figure 13 has data points with low linear and low angular acceleration, data points with low linear and high angular acceleration, data points with high linear and low angular acceleration, and data points with high linear and high angular acceleration. There is too much scatter for any correlation to exist.

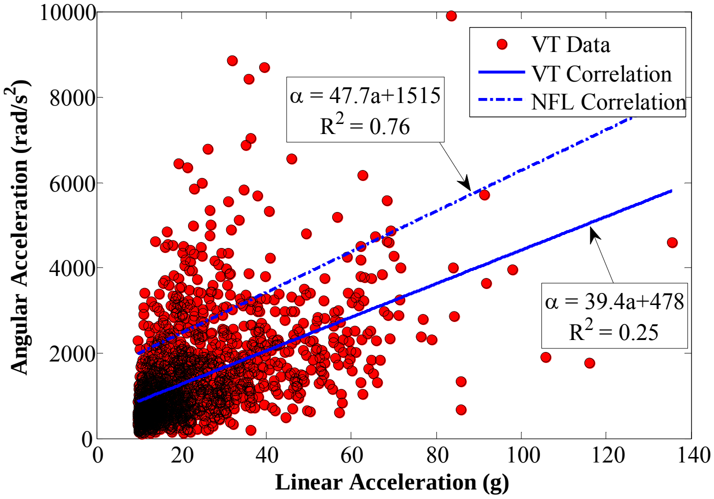

Figure 13: Peak angular acceleration as a function of peak linear acceleration. The VT data suggests that no correlation exists (R^2 = 0.25). The dashed line overlays the correlation reported by the NFL study.

The 6DOF data produced in this study suggests that there is no strong correlation (R^2 = 0.25) between linear and angular acceleration. This is inconsistent with relationships reported in the literature. Pellman et al. (2003a) reported a linear relationship between linear and angular acceleration (R^2 = 0.76). That correlation should only be applied to the NFL data, as only selected impacts in the concussive severity range were included in that study. It is possible for no correlation to exist between linear and angular acceleration when looking at a complete range of impacts. Varying impact location and principle direction of force for the same linear acceleration input can result in different angular accelerations. Theoretically, it is possible for impacts to exist with pure linear acceleration (no angular acceleration) and with pure angular acceleration (no linear acceleration). Pellman et al. (2003a) reported a correlation for specific impacts. When looking at a full range of impacts with varying impact location, principle direction of force, and impact severity; the 6DOF data suggests that no correlation exists.

The head acceleration data produced in this study was collected by measuring helmeted head impacts on human volunteers. While having applications in real world scenarios with padded impacts, this data may not be able to accurately model head impacts resulting in skull fractures. The resulting average duration of these helmeted head impacts was 14 ms. This is in agreement with the 15 ms impact duration experienced by the Hybrid III dummies in the NFL study.[11] This 14-15 ms range of impact duration is unique to helmeted head impacts. When comparing this to real world durations of impact in motor vehicle crashes, it is between that of head impacts to vehicle structures (> 6 ms) and an airbag with seatbelt restraints (< 40 ms).

In this study, angular velocities ranged from 0.5 rad/s to 42.5 rad/s. The 6DOF data is in agreement with the NFL data because all impacts fell within or below the range of the angular velocities of the non-injured players (9.8 rad/s to 55.8 rad/s).[11] Concussive impacts in the NFL data ranged from 12.8 rad/s to 80.9 rad/s. While concussive impacts can occur within the 6DOF dataset's angular velocity range, the fact that so many impacts were experienced within this range with no reported injuries suggests that human tolerance of angular velocity may be greater than previously thought.

There is some inherent error within the measurement of head acceleration using the 6DOF sensor. The 6DOF sensor has an average error of 1% +/- 18% and 3% +/- 24% for linear and angular acceleration, respectively. However, this error is within the accepted range of other measurement devices. The video analysis conducted for the NFL reconstructions reported error as high as 15%,[16] the original HITS sensor has an error of 8% +/- 11%, and chest bands used to measure chest deflection can have error as high as 10%.[17] Considering the vast amounts of data that can be collected with the 6DOF HITS sensor on human volunteers and the error levels of other biomechanical experiments, the error inherent in the 6DOF sensor is acceptable. While the 6DOF error is considered acceptable, additional methods can be used be applied to the dataset to minimize the statistical effects of data scatter. The standard deviations contribute to the majority of the error, as the average errors are minimal. Funk et al. (2007) presented a unique methodology for adjusting a large dataset to minimize the effect of data scatter. Using this same technique, the effects of data scatter in a 6DOF dataset can be incorporated into risk functions. This is primarily possibly given the thousands of data in the sample.

Conclusion

Football presents a unique opportunity to quantify the biomechanical response of the head to impact with human volunteers. The helmets of 10 Virginia Tech football players were instrumented with 6DOF sensors throughout the 2007 season, resulting in head acceleration data for 1712 impacts. The dataset is large and unbiased, as head impacts were recorded for every game and practice during the 2007 season. Future collection of concussive data will result in a better understanding of mild traumatic brain injury. With an increased understanding of the biomechanics of head impacts in football and human tolerance to head acceleration, better equipment can be designed to prevent head injuries.

Acknowledgements

The authors would like to thank Toyota Central Research and Development Labs, Inc and the National Highway Safety and Traffic Administration for funding this research.

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Chapter 4:
Computational Modeling of Football Head Acceleration Data Using SIMon

Abstract

Contact sports account for approximately 300,000 concussions each year in the United States, with football having the highest occurrence of any sport. Using a football field as an experimental environment provides a unique opportunity to collect biomechanical data to characterize concussions. In this study, linear and angular head acceleration data was collected from 10 collegiate football players. A total of 1712 impacts were collected during games and practices throughout the 2007 season. No instrumented player sustained a concussion. 24 of the most severe impacts were modeled using the Simulated Injury Monitor (SIMon) finite element head model. Probability of concussion was calculated based on previously published cumulative strain damage measure risk curves. Probabilities of concussions ranged from 3% to 29%. The angular kinematics had a greater effect of concussion risk than the translational kinematics. The injury risks in this study were compared to injury risks of other studies. The low risks of concussions estimated through SIMon simulations support the fact that no instrumented player sustained a concussion during data collection, while other risk curves estimated high probabilities of concussion for these impacts.

Introduction

Traumatic brain injuries are the leading cause of death and disability in the United States, with approximately 1.5 million cases each year (Thurman et al., 1999). Up to 75% of these injuries are mild traumatic brain injuries (MTBI), which is a less severe form of traumatic brain injury (Sosin et al., 1996). While MTBI is typically not life threatening like traumatic brain injury, it is still a major health concern, partly due to its potential long term effects. In the past, MTBI has been a difficult subject to study due to the difficulty of collecting brain injury data from human volunteers. Previous research has primarily used indirect methods for collecting data to characterize MTBI. Such methods have included utilizing animals, cadavers, and anthropomorphic test devices. Several injury metrics are used to predict head injury based on these methods. Head injury criterion (HIC), peak acceleration, and severity index (SI) are injury metrics derived from linear head acceleration and are primarily based on cadaver tests with skull fractures. Rotational acceleration injury thresholds are based mostly on primate tests with severe concussion, diffuse axonal injury (DAI), or intracranial bleed.

Contact sports account for approximately 300,000 concussions each year in the United States, with football having the highest occurrence of any sport (Thurman et al., 1998). Several researchers have identified competitive football as a unique experimental environment to collect head acceleration data. The NFL reconstructed a number of concussive impacts using Hybrid III dummies based on game video (Pellman et al., 2003a). From the data collected in the reconstructed impacts, injury risk curves were developed for MTBI. Nominal injury values determined in this study were a peak linear acceleration of 79 g, SI of 300, HIC of 250, and peak rotational acceleration of 5757 rad/s^2 (King et al., 2003; Pellman et al., 2003a). These risk curves are ultimately flawed because the NFL data is biased towards injurious impacts. Another study instrumented the helmets of collegiate football players to measure every head impact instrumented players experienced during games and practices (Duma et al., 2005). Data was collected over 4 years and a large and unbiased dataset was produced, including 4 concussive impacts. Using this data, MTBI risk curves for linear acceleration were produced, where a 165 g impact represented a 10% probability of concussion (Funk et al., 2007). The main limitation of this study was that only resultant linear head acceleration was recorded, resulting in data that could not completely model the kinematics of the head for impacts. This is an issue because angular acceleration has been thought to be the primary cause of brain injury by many.

Several researchers have simulated the NFL data with various computational models (King et al., 2003; Kleiven, 2007; Zhang et al., 2004). These studies were interested in the tissue level response of the brain, rather than the global inputs to the head. From the modeled impacts, the authors sought to identify the best predictor of brain injury. The present study choose to use the publicly available Simulated Injury Monitor (SIMon) finite element head model (FEHM) to model head impacts that human volunteers (football players) experienced.

SIMon was developed as a tool to assess the potential of traumatic brain injury in car crashes. Data collected from anthropomorphic test devices during crash tests serves as input to the model.

SIMon uses three injury metrics to assess whether a brain injury would or would not occur for each crash test: cumulative strain damage measure (CSDM) - a correlate to concussion/diffuse axonal injury, dilation damage measure – a correlate for contusions, and relative damage motion measure – a correlate for acute subdural hematoma. Data from animal experiments were used to determine injury thresholds for each injury metric computer by SIMon (Takhounts et al., 2003). SIMon is not a very complex model, as it only models the rigid skull, the dura-CSF layer, the brain, the falx cerebri, and the bridging veins. SIMon does not contain any specialized structures of the brain, which allows it to be scaled independently in three dimensions. This allows the SIMon to be generalized to other cases, and is why SIMon's injury metric could be validated using animal data.

The goal of this study was to model head acceleration data of severe head impacts from human volunteers with SIMon. The resulting data could be used to estimate injury risk for each specific impact and compare that to previous studies that define risk.

Methods

Data Collection

The helmets of 10 Virginia Tech football players were instrumented with a newly developed 6 degree of freedom (6DOF) head acceleration measurement device for the 2007 football season. Each player that participated in the study gave written informed consent with Institutional Review Board approval from both Virginia Tech and the Edward Via College of Osteopathic Medicine. The 6DOF sensors are capable of measuring linear and angular acceleration about each axis of the head for every head impact an instrumented player may experience during games and practices. The 6DOF sensor consists of 12 single-axis, high-g accelerometers (ADXL193, Analog Devices, Norwood, MA) that are enclosed in padding and integrated into existing Riddell Revolution football helmets (Elyria, Oh). The sensor is designed so that the accelerometers remain in contact with the head at all times. This ensures that head acceleration, not helmet acceleration, is measured (Manoogian et al., 2006).

In addition to the 12 accelerometers, the 6DOF sensor is equipped with on-board data acquisition and a wireless transceiver. Data acquisition is triggered anytime an accelerometer exceeds 10 g.

Data is collected for 40 ms at 1000 Hz, of which 8 ms are pre-trigger and 32 ms are post-trigger. Each recorded impact is downloaded wirelessly by a sideline computer via the 6DOF sensor's transceiver. Linear and angular accelerations are computed through post-processing using a novel algorithm (Chu et al., 2006). 10 Virginia Tech football players were instrumented for every game and practice during the 2007 football season. A total of 1712 impacts were recorded. All data was up-sampled to 10 kHz by linear interpolation and then filtered to SAE J211 specification. For each impact, the time series responses of linear and angular acceleration were recorded for each axis of the head.

Impact Selection

No instrumented player sustained a concussion during the 2007 Virginia Tech football season. Of the collected impacts, peak resultant linear head accelerations ranged from 10 g to 135 g and peak resultant angular accelerations ranged from 107 rad/s^2 to 9922 rad/s^2. A total of 24 of the most severe impacts were selected to model with the SIMon FEHM. Selection criteria were based on impacts that exceeded the nominal injury values based on the NFL data (King et al., 2003; Pellman et al., 2003a). 11 impacts had peak resultant linear accelerations greater than 79 g. 14 impacts had peak resultant angular accelerations greater than 5757 rad/s^2. Only 1 impact exceeded both these values. The 24 selected impacts represent 6 different players.

SIMon Modeling

SIMon version 3.0 was used in this study. The SIMon FEHM accepts three dimensional head kinematic data from a Hybrid III anthropomorphic test device (ATD) as input. The data may be in the form of either a nine accelerometer array package (NAP) or three linear head accelerations and three angular head velocities. Since the 6DOF data is in the form of three linear head accelerations and three angular head accelerations, the data was converted to the output of an instrumented Hybrid III ATD in the form of a NAP. The form of NAP data was chosen over angular velocities to eliminate any potential error numerical integration of the angular acceleration data may have introduced. Once the 24 impacts were converted to the nine accelerations a NAP would measure, all impacts were modeled in SIMon.

SIMon computes several metrics that correlate to real world injuries. This study focuses on cumulative strain damage measure (CSDM). CSDM represents the fraction of the brain that experiences strains greater than a specified threshold. Takhounts et al. (2003) developed injury risk curves for CSDM based on data from animal experiments. From SIMon simulations of these data, Takhounts et al. (2003) reported that a strain threshold of 0.15 was shown to be the best predictor of injury for CSDM. In this study, the probability of concussion based on SIMon simulations was determined for each of the 24 modeled impacts using a CSDM strain threshold of 0.15.

Results

Table 3 displays the resultant head kinematics for all 24 modeled impacts with their respective risk of concussion based on CSDM (strain = 0.15). Linear accelerations ranged from 19 g to 135 g. Change in translational velocity, or delta v, ranged from 1.0 m/s to 5.9 m/s. Angular accelerations ranged from 668 rad/s^2 to 9919 rad/s^2. Angular velocities ranged from 4 rad/s to 43 rad/s. 7 of these impacts had probabilities of concussion greater than or equal to 10% based on SIMon simulations.

Table 3: Kinematics of the 24 modeled impacts and their respective probability of concussion based on CSDM (strain = 0.15).

Impact	Linear Acceleration (g)	Delta V (m/s)	Angular Acceleration (rad/s^2)	Angular Velocity (rad/s)	CSDM	Probability of Concussion
114	86	4.3	668	4	0.0%	**3%**
13	91	4.0	5707	23	0.0%	**3%**
1306	86	3.7	1346	7	0.0%	**4%**
265	116	2.3	1758	6	0.0%	**4%**
357	106	4.7	1916	10	0.0%	**4%**
1118	84	4.3	2865	14	0.0%	**4%**
550	91	4.1	3645	19	0.1%	**4%**
480	98	4.3	3950	15	0.0%	**4%**
1067	84	4.1	4014	21	0.1%	**4%**
400	135	5.9	4602	18	0.0%	**4%**
1099	63	4.7	6165	30	4.2%	**5%**
245	46	3.3	6565	23	5.2%	**6%**
456	23	1.3	5846	28	8.4%	**7%**
251	25	1.3	5998	26	7.5%	**7%**
962	21	1.0	6352	25	7.9%	**7%**
286	19	1.5	6446	28	8.5%	**7%**
1538	35	1.8	6870	30	12.4%	**8%**
116	36	2.9	8427	31	16.4%	**10%**
241	36	2.0	7034	33	17.4%	**11%**
1537	26	1.3	6783	34	19.6%	**12%**
472	84	5.1	9919	43	23.4%	**14%**
136	39	1.9	8709	40	35.3%	**25%**
135	32	2.1	8863	41	35.0%	**25%**
724	35	2.9	5824	37	39.1%	**29%**

Figure 14 displays the probability of concussion based on CSDM (strain = 0.15) as a function of the kinematic data. Linear acceleration and delta v had no correlation with probability of concussion. Angular acceleration and angular velocity follow similar trends. As angular acceleration and angular velocity increase, the probability of concussion calculated from CSDM increases. Angular velocity has a stronger relationship with probability of concussion than angular acceleration.

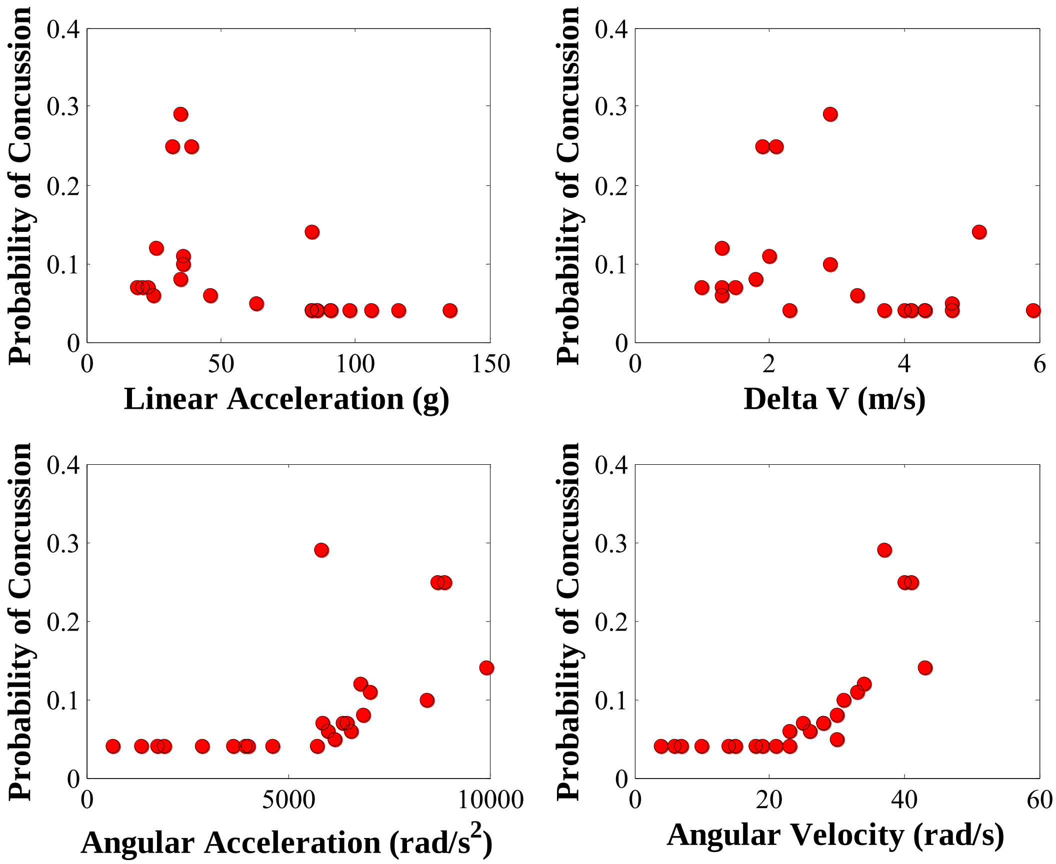

Figure 14: Probability of concussion based on CSDM (strain = 0.15) as a function of the kinematic factors: linear acceleration, delta v, angular acceleration, and angular velocity.

Discussion

Any impact that represented greater than a 50% risk of concussion based on the NFL data for linear and/or angular acceleration was modeled using SIMon. While the NFL risk curves predict a great risk of concussion for many of these impacts, probabilities of concussion based on SIMon simulations ranged from 3% to 29%, with 17 of these impacts having less than 10% risk. The fact that no instrumented player sustained a concussion supports why SIMon is predicting such low probabilities of concussion for these relatively severe impacts. SIMon quantifies the tissue level response of the head to impact because CSDM values are a product of the complete kinematics of the head throughout impact. Injury risk curves based on single kinematic variables only look at part of the organ level response to impact. Although using a computational model takes into account the complete kinematics of the head, single kinematic variables may still be useful as predictors of head injury.

Simulations of impacts with high linear accelerations and low angular accelerations produced some of the lowest probabilities of concussion, as seen in Table 3. In contrast, impacts with low linear accelerations and high angular acceleration resulted in some of the higher probabilities of concussion. This is illustrated in Figure 14, which displays the probability of concussion based on the SIMon simulations as a function of linear acceleration, delta v, angular acceleration, and angular velocity. The lack of a correlation between the linear kinematics of the head and probability of injury is a result of the material properties of the SIMon brain and the validation used to correlate injury risk to CSDM. SIMon models the brain as being incompressible (Poisson's ratio $v = 0.49999$); which means the brain will not experience any strain when only linear acceleration is present, and thus having no effect on CSDM. In addition, the animal experiments used to determine injury risk induced concussions by subjecting the animals to angular acceleration and correlated highly with angular velocity. This can be seen by the strong correlations for the angular kinematics in Figure 14, particularly angular velocity.

Several studies have developed injury risk curves based on football-related data. The NFL reconstructed concussive impacts with Hybrid III dummies (Pellman et al., 2003a). From the data collected in this study, injury risk curves for linear and angular acceleration were created (King et al., 2003). Another study used head acceleration data from Virginia Tech football players to create concussion risk curves for linear acceleration (Funk et al., 2007). Table 4 compares the associated risk of concussion of the selected 24 impacts for the NFL, VT, and SIMon risk curves.

Table 4: Comparison of concussion risk for the modeled impacts based on risk curves published for the NFL data, Virginia Tech data, and SIMon.

Impact	Linear Acceleration	Angular Acceleration	NFL Linear	NFL Angular	VT Linear	SIMon CSDM
114	86	668	60%	< 2%	1%	**3%**
13	91	5707	66%	49%	1%	**3%**
1306	86	1346	60%	3%	1%	**4%**
265	116	1758	90%	4%	2%	**4%**
357	106	1916	83%	4%	1%	**4%**
1118	84	2865	57%	9%	1%	**4%**
550	91	3645	67%	16%	1%	**4%**
480	98	3950	75%	19%	1%	**4%**
1067	84	4014	57%	20%	1%	**4%**
400	135	4602	96%	28%	3%	**4%**
1099	63	6165	28%	58%	0%	**5%**
245	46	6565	12%	66%	0%	**6%**
456	23	5846	4%	52%	0%	**7%**
251	25	5998	4%	55%	0%	**7%**
962	21	6352	3%	62%	0%	**7%**
286	19	6446	3%	63%	0%	**7%**
1538	35	6870	7%	71%	0%	**8%**
116	36	8427	7%	89%	0%	**10%**
241	36	7034	7%	73%	0%	**11%**
1537	26	6783	4%	69%	0%	**12%**
472	84	9919	57%	> 96%	1%	**14%**
136	39	8709	8%	91%	0%	**25%**
135	32	8863	6%	92%	0%	**25%**
724	35	5824	7%	51%	0%	**29%**

When comparing NFL linear acceleration risk values to that of SIMon for the selected impacts, there is no agreement between their concussion risk assessments. This may be a result of linear acceleration having a minimal effect on the SIMon simulations' risk assessment. However, it is possible that linear acceleration may be a good predictor of concussion. The NFL data are biased towards injury, thus resulting in risk curves that overestimate risk. Funk et al. (2007) presented linear acceleration risk curves for MTBI based on a large and unbiased dataset. This resulted in a less conservative risk assessment, in which a 165 g represented a 10% risk of MTBI.

Interestingly, the probability of concussion based on angular acceleration using the NFL risk curve compares more favorably with that produced by the SIMon simulations. While the probabilities don't match, similar trends can be seen, in that probability of concussion generally increases with increased angular acceleration with both methods. SIMon injury risk does not begin to increases until the NFL angular acceleration based risk approaches 50%. While the NFL risk curves predicted probabilities of concussions over 70% for a number of impacts, the maximum probability of MTBI based on the SIMon simulations was 29%. The lower injury probabilities estimated by SIMon support the fact that no instrumented player sustained a concussion during data collection.

Various methods of assessing concussion risk were compared to SIMon in this study. It should not be assumed that the SIMon simulations accurately represent risk; even though CSDM with a strain threshold of 0.15 has been shown to correlate strongly with probability of concussion based on animal experiments (Takhounts et al., 2003). SIMon is inherently flawed by modeling the brain as incompressible and validation data based on angular acceleration induced concussions. With saying that, risk values based on the complete kinematics of the head are more likely to produce superior risk estimates than that of single kinematic variables. Further research should be conducted using an expanded dataset that includes concussive and non-concussive impacts with a greater range of linear and angular accelerations. Such data can be used to assess the accuracy of the injury probabilities based on SIMon simulations, as well as other existing injury risk curves.

Conclusion

This study modeled head acceleration data from human volunteers during relatively severe head impacts using the publicly available finite element head model, SIMon. The influence of the linear and angular head kinematics on injury risk estimated from SIMon simulations was investigated. 24 impacts that represented greater than a 50% risk of concussion based on the NFL risk curves were modeled using SIMon. Probabilities of concussions based on the SIMon simulations ranged from 3% to 29%. The low risks of concussions estimated through SIMon simulations support the fact that no instrumented player sustained a concussion during data collection. While it is apparent that the NFL risk curves overestimate injury risk, it is unclear how accurate the predictive capabilities of SIMon and the VT linear risk curve are. An expanded dataset that includes concussive impacts from human volunteers can be used to help validate computational models, such as SIMon, and be responsible for a better understanding of the mechanisms of mild traumatic brain injury.

Acknowledgements

The authors would like to thank the National Highway Safety and Traffic Administration for funding this research.

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Chapter 5:
Force Transmission to the Mandible by Chin Straps during Head Impacts in Football

Abstract

The objective of this study was to determine the force transmitted to the mandible from the chin strap in football helmets for head impacts. A total of 32 tests were performed comparing front and side impact locations. Each location was tested at two impact velocities (6.5 m/s and 9.0 m/s). Different combinations of neck collars and shoulder pads were tested at each speed and location to account for potential equipment variability between football players. A 50th percentile male Hybrid III dummy was equipped with a helmet, shoulder pads, and various neck collars. Tension load cells were installed on the left and right sides of the chin straps. From the tension values in the chin strap, the force transmitted to the mandible was calculated. With the front impact location, the average peak mandible load was 568 +/- 80 N at 6.5 m/s and 806 +/- 64 N at 9.0 m/s. With the side impact location, the average peak mandible load was 142 +/- 80 N at 6.5 m/s and 275 +/- 84 N at 9.0 m/s. Although there are some assumptions, these values represent a good estimation of the forces acting on the mandible for head impacts in football.

Keywords: chin, strap, jaw, Hybrid III, football, impact, concussion, mandible

Introduction

The facemask of football helmets has been shown to be a common impact location to cause concussions. In fact, it has been shown that concussive impacts to the facemask occur with a lower peak head acceleration than impacts to the helmet shell [1]. The facemask of a helmet protrudes further from the head than the helmet shell, producing a larger moment arm when impacted. This results in higher rotations; which may cause higher forces affecting the midbrain, increasing the risk of concussion [2]. Also, this type of impact directly loads the mandible through the chin strap. Loads to the mandible may increase the risk of concussion because the mandible protrudes forward of the head center of gravity, resulting in higher rotations relative to the other anatomical structures of the head [2]. Many football players will wear mouth guards to reduce the risk of concussions. It is thought that these mouth guards dampen the response of the mandible impacting the head. Several studies have attempted to quantify the effectiveness of these mouth pieces, and have not been able to identify a reduction in concussions [3, 4]. Studies have also been conducted to assess the effectiveness of a football helmet's ability to reduce head acceleration [5]. However, these studies do not take into account the direct loading of the mandible through the chin strap, which is considered a mechanism of concussion.

Although studies have quantified the effectiveness of helmets and mouth guards, little research has been conducted to look at how the football helmet interacts with the mandible via the chin strap. The purpose of the chin strap is to keep the helmet on a player's head. Most chin straps have little-to-no padding and directly transfer impact energy to the mandible. The goal of this study was to determine the force transmitted to the mandible from the chin strap for head impacts in football.

Methods

An instrumented 50th percentile male Hybrid III test dummy was used to determine the force transmitted to the mandible by the chin strap of a football helmet during impacts to the helmet. The dummy was equipped with a set of Douglas CP25 shoulder pads and a large Riddell VSR4 helmet for all tests. The front and side of the helmet were impacted using a linear pneumatic impactor, and the tension in the chin strap was recorded.

In order to obtain a range of data that accounted for potential equipment variability between football players, several variables were included in the dummy configuration. Many football players wear protective neck collars that limit motion of the neck in an attempt to reduce the probability of experiencing a stinger [6]. To account for the possibility of players wearing protective neck collars, 4 neck collar configurations were used in this set of testing. The dummy was either equipped with McDavid's Cowboy Collar, the Bullock Collar, the Kerr Collar, or no collar. These neck collars are discussed in detail by Rowson [7]. In an effort to simulate actual football impacts and to account for the various ways a player may get hit, different shoulder pad positions were also tested. This involved testing the shoulder pads in a normal and raised position. In order to raise the shoulder pads, shoulder implants were made for the dummy using expanding polyurethane foam. These implants were secured on the shoulders of the dummy for the raised shoulder pad tests.

Impact velocities were chosen so that they would simulate moderate to high severity impacts that are typically seen in football games. The impact velocities for this study were 6.5 m/s and 9.0 m/s and are based on NFL reconstructions performed by Pellman [8]. The helmet was impacted in two locations: the front and side of the helmet (Figure 15). These impact locations were chosen with the intentions of the front impacts resulting in large mandible loads and the side impacts resulting in relatively small mandible loads. The different combinations of the configuration variables [2 locations x 2 speeds x 4 collars x 2 shoulder pad positions] resulted in 32 tests, or 16 tests for each impact location. For each impact, the position of the dummy relative to the impactor was precisely controlled using alignment targets. In addition, a helmet positioning tool was used to ensure that the position of the helmet on the dummy was consistent.

The impacts were performed using the same pneumatic linear impactor as described by Rowson [7]. The impactor was instrumented with a load cell (Denton 1968 LC-91, Rochester, MI) on the impactor arm. A light gate (Omron E3S-AT11, Schaumburg, IL) was used to measure the velocity of the impactor arm as it contacted the dummy. All instrumentation was sampled at 10,000 Hz and processed in accordance with SAE J211. A digital high speed color camera (Phantom V4, Wayne, NJ) recorded each test at 1000 frames per second.

Load cells were attached to the left (Denton 6370 LC-78, Rochester, MI) and right (Denton 6370 LC-79, Rochester, MI) sides of the chin strap. These load cells are designed similarly to seat belt load cells, in that they measure tension. Figure 16 is a diagram depicting how the tension is measured in this particular load cell. The chin strap is intertwined between three bars that are spaced closely together, making an S-shape when there is no tension in the strap. When the chin strap is put under tension, it pushes the three bars away from the strap. The force the chin strap exerts on the bars is measured and recorded as the tension in the chin strap.

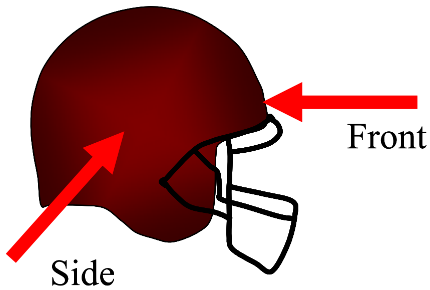

Figure 15: Front and side impact locations.

[Image created by Steve Rowson]

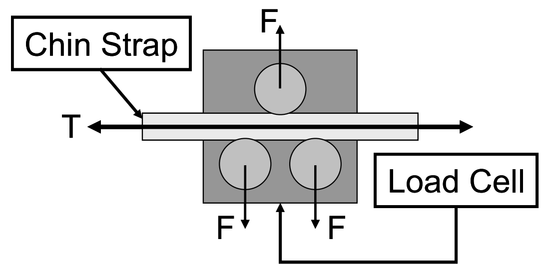

Figure 16: Tension in the chin strap results in a measurable force on the bars of the load cell.

[Image created by Steve Rowson]

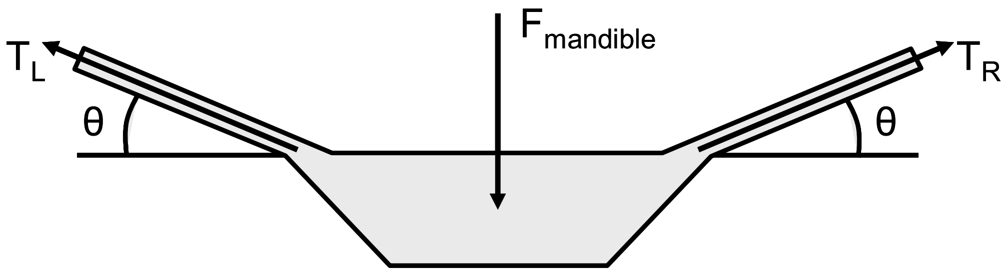

Figure 17: Free body diagram of the forces acting on the chin strap.

[Image created by Steve Rowson]

$$F_{mandible} = T_L \sin\theta + T_R \sin\theta \qquad (1)$$

Knowing the magnitude and direction of the tension in the chin strap, it is possible to estimate the force being transmitted to the mandible from the chin strap. Figure 17 is a free body diagram of the forces acting on the chin strap. Equation 1 is the solution for the force being transmitted to the mandible, where T_L is the tension in the left strap, T_R is the tension in the right strap, $F_{mandible}$ is the force transmitted to the mandible, and θ is the angle of the chin strap with respect to the plane perpendicular to the mandible of the dummy. θ was measured and found to be 21 degrees. This angle is assumed to remain constant throughout each test. This assumption may result in a low estimate of the force transmitted to the mandible, as the angle will increase as the helmet translates away from the impactor.

Results

Table 5 displays the left and right chin strap tensions, the calculated mandible load, and impactor force for each testing configuration for the front impacts. Table 13 displays these same data for the side impacts. For tests side12 and side16, the data could not be analyzed due to bad signal quality.

Figure 18 and Figure 19 compare the time series loading curves of the mandible for the front and side impact locations. Figure 4 is a plot of the average mandible load +/- one standard deviation for all the front impacts at 6.5 m/s. Figure 5 shows the same for the side impacts. There are distinct differences between the loading paths of the front and side impact locations. Impacts to the front of the helmet result in higher mandible loads than impacts to the side of the helmet. Also, the front impacts generated a relatively clean and consistent signal, while the side impacts had much more variability between signals, as seen with the larger standard deviations.

Table 5: Peak values for all front impact configurations.

Test ID	Location	Collar	Shoulder Pad Position	Impact Velocity (m/s)	Left Strap Tension (N)	Right Strap Tension (N)	Mandible Load (N)	Impactor Force (N)
front1	Front	None	Normal	6.5	854	711	560	4410
front2	Front	None	Normal	9.0	1314	1189	896	9674
front3	Front	None	Raised	6.5	955	834	641	4578
front4	Front	None	Raised	9.0	1244	1047	813	10621
front5	Front	Cowboy	Normal	6.5	808	706	534	4529
front6	Front	Cowboy	Normal	9.0	1316	1038	844	10689
front7	Front	Cowboy	Raised	6.5	720	614	477	4416
front8	Front	Cowboy	Raised	9.0	1150	939	748	10928
front9	Front	Kerr	Normal	6.5	1064	768	653	4579
front10	Front	Kerr	Normal	9.0	1282	1105	800	10962
front11	Front	Kerr	Raised	6.5	606	659	446	4647
front12	Front	Kerr	Raised	9.0	1060	868	690	10534
front13	Front	Bullock	Normal	6.5	958	875	656	4539
front14	Front	Bullock	Normal	9.0	1247	1140	855	10969
front15	Front	Bullock	Raised	6.5	863	738	573	4351
front16	Front	Bullock	Raised	9.0	1171	1081	807	10921

Table 6: Peak values for all side impact configurations.

Test ID	Location	Collar	Shoulder Pad Position	Impact Velocity (m/s)	Left Strap Tension (N)	Right Strap Tension (N)	Mandible Load (N)	Impactor Force (N)
side1	Side	None	Normal	6.5	90	117	25	3482
side2	Side	None	Normal	9.0	603	285	317	6256
side3	Side	None	Raised	6.5	456	208	187	3406
side4	Side	None	Raised	9.0	691	206	305	6393
side5	Side	Cowboy	Normal	6.5	378	200	175	3455
side6	Side	Cowboy	Normal	9.0	798	332	387	6241
side7	Side	Cowboy	Raised	6.5	287	104	107	3379
side8	Side	Cowboy	Raised	9.0	543	167	156	6403
side9	Side	Kerr	Normal	6.5	295	254	121	3430
side10	Side	Kerr	Normal	9.0	505	268	198	6606
side11	Side	Kerr	Raised	6.5	200	115	107	3432
side12	Side	Kerr	Raised	9.0	-----	-----	-----	-----
side13	Side	Bullock	Normal	6.5	296	251	123	3358
side14	Side	Bullock	Normal	9.0	698	238	291	6359
side15	Side	Bullock	Raised	6.5	568	274	299	3421
side16	Side	Bullock	Raised	9.0	-----	-----	-----	-----

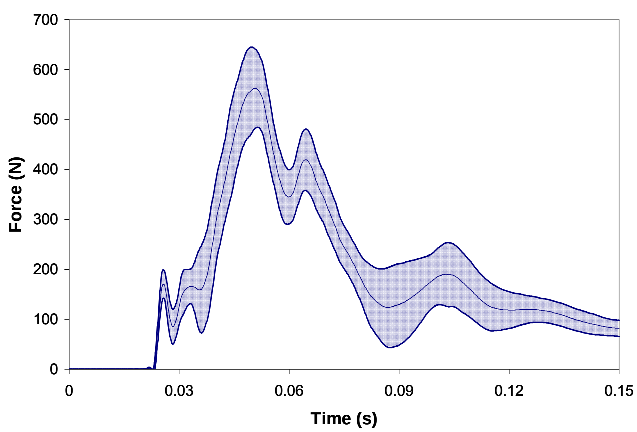

Figure 18: Average force vs time plot for front impacts at 6.5 m/s.

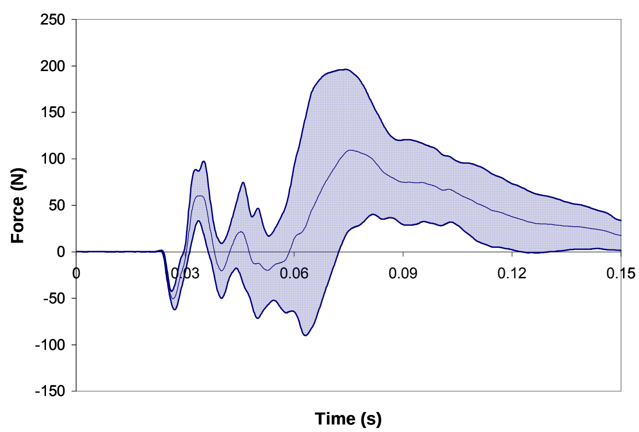

Figure 19: Average force vs time plot for side impacts at 6.5 m/s.

Figure 20 displays the average peak mandible load for each impact velocity at each location. With the front impact location, the average peak mandible load was 568 +/- 80 N at 6.5 m/s and 806 +/- 64 N at 9.0 m/s. With the side impact location, the average peak mandible load was 142 +/- 80 N at 6.5 m/s and 275 +/- 84 N at 9.0 m/s.

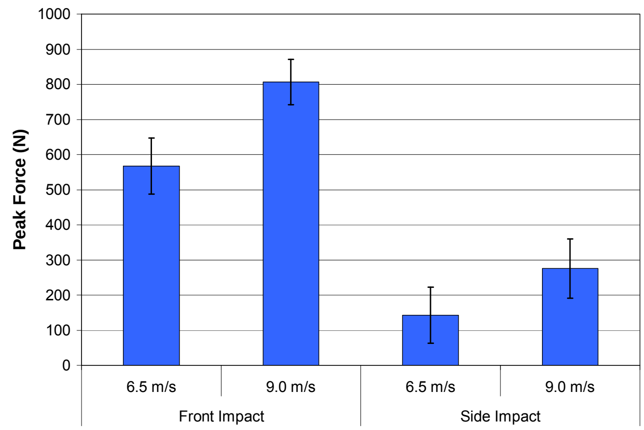

Figure 20: Average peak mandible load for each location at each impact velocity.

Discussion

Impacts to the front of the helmet generated higher mandible loads than any impact to the side of the helmet. The chin strap is not padded and remains bound to the mandible at all times. In contrast, the helmet is designed with padding; and when that padding is compressed, the helmet will translate in the direction of the impact as much the padding is compressed. In addition, if the helmet does not perfectly fit on the head, there will be additional room for the helmet to translate. In a front impact, this directly increases tension in the chin strap. In a side impact, the helmet translates away from the opposite side of the head that is being impacted. This reduces tension in chin strap on the side of the helmet being impacted and increases tension in the chin strap on the opposite side. This can be seen in Table 2, in which the left chin strap tension is greater than the right chin strap tension. Figure 5 also displays this type of loading. There is pretension in the chin strap when the helmet is at rest on the head, which is considered to be zero-tension in this study. When the side of the helmet is impacted, the tension is reduced in the chin strap on the impacted side, resulting in the sensor generating a negative tension. In the first few milliseconds of a side impact, the reduction in tension can be seen in Figure 5 by the negative force. However, once the helmet translates away from the opposite side of the head, the tension in the chin strap on the opposite side dominates the mandible force, as seen by the positive peak.

The Hybrid III head and neck have limited biofidelity and cannot perfectly emulate the head and neck response of a human. However, it is the best available surrogate for a human. The automotive industry considers the Hybrid III the gold standard when testing to predict injury in crash tests. In the past, the Hybrid III has also been used for applications beyond the automotive industry, such as football testing. Pellman (2003) reconstructed concussive football impacts with Hybrid III dummies using multiple angles of game video [8]. The Hybrid III dummy may not be perfectly biofidelic, but it is commonly accepted as a human surrogate and these types of tests have been performed in the past [9].

Conclusion

A series of 32 tests were performed to determine the force that is transmitted to the mandible from the chin strap of football helmets for head impacts in football. Each impact location was tested at two impact velocities with various combinations of neck collars and shoulder pad positions. The force transmitted to the mandible could be calculated based on the tension in the chin strap. Impacts to the front of the helmet generated higher mandible forces than impacts to the side of the helmet. With the front impact location, the average peak mandible load was 568 +/- 80 N at 6.5 m/s and 806 +/- 64 N at 9.0 m/s. With the side impact location, the average peak mandible load was 87 +/- 36 N at 6.5 m/s and 170 +/- 80 N at 9.0 m/s. Although there are some assumptions, these values represent a good estimation of the forces acting on the mandible for head impacts in football.

Acknowledgements

The authors would like to thank Robert A. Denton Inc. for providing the load cells used in this study.

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Chapter 6:
Biomechanical Analysis of Football Neck Collars

Abstract

Objective: To determine the load limiting capabilities of protective neck collars used in football through dynamic impact testing.

Design: A 50th percentile male Hybrid III dummy was utilized in 48 dynamic impact tests comparing the Cowboy Collar, Bullock Collar, and Kerr Collar. A control and each collar was tested at two velocities (5 m/s and 7 m/s), three impact locations (front, top, and side of the helmet), and two shoulder pad positions (normal and raised).

Setting: Research laboratory.

Patients: None.

Interventions: None. Independent variables were the neck collars, impact velocity, and shoulder pad position.

Main Outcome Measurements: In addition to range of motion, upper and lower neck forces and moments were measured.

Results: With the top impact location, it was found that the Kerr Collar and Bullock Collar reduced head accelerations and force transmission through the neck. With the front impact location, all the collars reduced lower neck moment. The Kerr Collar was also capable of reducing the lower neck force and upper neck moment. With the side impact location, the Kerr Collar substantially reduced lower neck moment.

Conclusions: These reductions in loads correlate with the degree to which each collar restricted the motion of the head and neck. By restricting the range of motion of the neck and redistributing load to the shoulders, neck loads can be effectively lowered.

Key Words: stinger, neck collar, football, brachial plexus

Introduction

Neck injuries in football can vary from the rare catastrophic event, to the much more frequent but less severe neck stinger. Stingers are a common injury in competitive football. Studies have shown lifetime injury incidences from 49% to 65% in college football.[1, 2] Many players will wear neck collars to prevent such injuries. These collar designs are based off empirical data, and few experiments have been conducted to quantify their effectiveness.

A stinger is most likely caused by injuring the upper trunk of the brachial plexus, which is made up of the C5 and C6 nerve roots.[3] This group of nerves runs from the cervical spine through the shoulder and into the upper arm, traveling directly under the clavicle. Stingers usually involve excessive hyperextension or lateral flexion of the head due to an impact, either with another player or with the ground. Symptoms include numbness, pain, or a stinging or burning sensation in the shoulder and arm. Usually, these symptoms resolve within minutes.[1] However, this simple neurapraxia can escalate into an axonotmesis (damage to the axon or myelin sheath) that lasts for days or months, or a neurotmesis (complete disruption of the nerve) that is permanent.[4]

There are two main lateral flexion injury mechanisms: traction and compression. In a traction injury, the head is flexed laterally, and the brachial plexus ipsilateral to the impact is stretched. In a compression injury, lateral flexion combined with extension may lead to a pinching of the nerve roots when the foramina close on the contralateral side.[2] This type of injury is usually very precise and local, while the stretching injury may occur anywhere along the plexus and is usually a more diffuse injury.

The neck collars that are worn by football players to prevent injuries were most often designed and put into use without biomechanical testing. Two researchers have attempted to quantify the effectiveness of these collars in reducing range of motion in the lateral flexion and extension planes: Hovis in 1994 and Gorden in 2003. Hovis and his collaborators outfitted a subject with a helmet and various shoulder pad/collar combinations. A pulley system was used to apply a quasi-static load to the subject's head to produce either hyperextension or lateral flexion of the neck. The collars provided reductions of 33% to 48%, in hyperextension of the neck. The study found no difference in reduction of motion for lateral flexion of the neck.[5] Gorden took a similar approach in analyzing football neck collars, but opted to apply a force with a hand-held pressure transducer. The test subjects were fitted with a helmet, shoulder pads, and a variety of neck collars. In the front-loading position, the researchers found that all collars permitted significantly less hyperextension than the shoulder pads alone. In the lateral loading tests, it was found that the collars did not significantly affect the active motion of the head.[6] The objective of this study was to perform a biomechanical analysis of neck collars through dynamic testing.

Methods

Three different neck collars were evaluated in this study: the Cowboy Collar (manufactured by McDavid), the Bullock Collar (designed by Virginia Tech head team physician, Richard Bullock), and the Kerr Collar (prototype designed by Patrick Kerr). The Cowboy Collar consists of a molded polyurethane foam collar that gets laced into the shoulder pads. The Cowboy Collar is designed to limit extension of the neck much more so than lateral flexion. The Bullock Collar consists of a high-density foam collar with a rigid plastic insert that is strapped to the shoulder pads. The Bullock Collar is designed to prevent hyperextension of the neck, with some restriction to lateral flexion. The Kerr Collar consists of a rigid synthetic mold that rests on the shoulders that is laced into the shoulder pads. The Kerr Collar is designed so that the base of the helmet contacts the collar, thus restricting motion in multiple planes.

An instrumented 50th percentile male Hybrid III test dummy was used to assess the effectiveness of these neck collars. The dummy was suited with a set of Douglas CP25 shoulder pads and a medium Riddell VSR4 helmet for all tests (Figure 21). A pneumatic linear impactor was used to strike the helmet. A total of 48 tests were performed where neck collar, impact velocity, impact location, and shoulder pad position were varied. The impacting velocities of stingers have not been studied or determined; therefore impact velocities were chosen so that they would encompass the impact velocities typical of tackling and blocking. The impacting speeds used were 5 m/s, and 7 m/s.[7] The locations impacted were the side, front, and top of the helmet (Figure 22). The shoulder pads were tested in a normal and raised position. The raised shoulder pad position was meant to simulate a player assuming a tackling posture, in which the shoulders are naturally raised in anticipation of an impact. In order to raise the shoulder pads, shoulder implants made of expanding polyurethane foam were secured to the shoulders of the dummy.

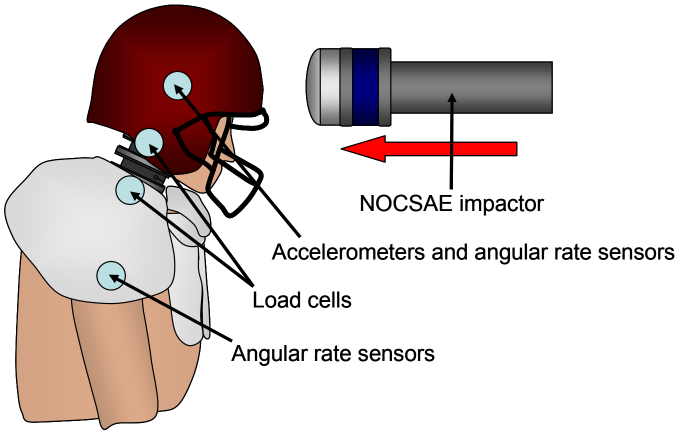

Figure 21: Dummy equipped with shoulder pads, helmet, and instrumentation.

[Image created by Steve Rowson]

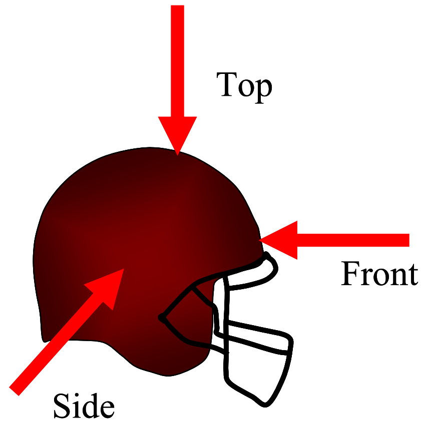

Figure 22: Top, front, and side impact locations.

[Image created by Steve Rowson]

The dummy was fitted with three single-axis orthogonally mounted accelerometers (Endevco B40351 B40234 B40740, 2000 G, San Juan Capistrano, CA) and a tri-axial angular rate sensor (IES 3103, 4800 deg/s, Braunschweig, Germany) in the center of gravity of the head. The chest of the dummy was also fitted with an angular rate sensor (ATA Sensors ARS-06S, 600 rad/s, Albuquerque, NM). The dummy was instrumented with angular rate sensors during the 5 m/s tests. The neck was instrumented with upper and lower neck load cells (Denton 1716A LC-592 and 1794A LC-242, Rochester, MI). The impactor arm was instrumented with a load cell (Denton 1968 LC-91, Rochester, MI) and an accelerometer (Endevco B40592, 2000 G, San Juan Capistrano, CA). A light gate (Omron E3S-AT11, Schaumburg, IL) was used to measure the velocity of the impactor arm as it contacted the dummy. All instrumentation was sampled at 10,000 Hz and processed in accordance with SAE J211. In addition, a high-speed video camera (Phantom V4, Wayne, NJ) recorded each test at 1000 frames per second. All impacts were performed with a pneumatic linear impactor.[8] The impacting surface was designed to replicate the impacting characteristics of a typical football helmet, and is identical to the impacting surface used in the new proposed NOCSAE standard for football helmet testing.[7]

The head and neck's range of motion was calculated from the angular rate data using a technique for the upper extremities described by Hall.[9] While described for the upper extremity, this method is utilized more widely to quantify the kinematics of other anatomical structures. In this study, the method was adapted for the cervical spine and validated using high speed video.

Angular rate data was used to determine range of motion due to its high sampling resolution and exact alignment of axes.

The coordinate system referenced in this paper is that of the Hybrid III. The positive x-axis runs out of the face (perpendicular to the coronal plane), the positive y-axis runs out of the right ear (perpendicular to the sagittal plane), and the positive z-axis runs out of the bottom of the head (perpendicular to the transverse plane). The metrics used to evaluate neck response in the results are force, moment, and range of motion. Force can be described as the influence on a body which causes it to accelerate. Moment can be described as the measure of a force's tendency to produce rotation about an axis. The range of motion reported in the results is the maximum change in angle the z-axis of the head makes with the transverse plane.

Results

An impact to the top of the helmet promotes axial compression of the neck. The most relevant data in a top impact test are the resultant head acceleration, force transmission reduction in the lower neck, and force transmission reduction in the upper neck. Force transmission reduction in the upper and lower neck was calculated as a percentage. This percentage represents the fraction of the impactor load not experienced in the upper and lower neck. Upper and lower neck moments were not reported because an impact to the top of the helmet does not provoke rotation of the head and neck. Table 7 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 8 presents the results of the top impact, raised shoulder pads configuration.

A front impact promotes extension of the neck. The most relevant data in a front impact test are the resultant head acceleration, upper and lower neck forces along the x-axis, and upper and lower neck moment about the y-axis. Table 9 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 10 presents the results of the side impact, raised shoulder pads configuration. In some cases, it can be observed that peak force decreases while peak moment increases relative to the control tests. Although this is uncommon, it is physically possible. This is due to the fact that the peak values for each occur at different times throughout the response to impact, with peak force occurring before peak moment. These values are related to the translation and bending phases of the response to impact. The addition of neck collar can affect one or both of these phases, resulting in changes in peak force and/or peak moment.

A side impact promotes lateral bending of the neck. The most relevant data in a side impact test are the resultant head acceleration, upper and lower neck forces along the y-axis, and upper and lower neck moment about the x-axis. Table 11 displays the peak values for each collar in the normal shoulder pad position when tested at each velocity. Table 12 presents the results of the side impact, raised shoulder pads configuration.

Table 7: Peak Values for Top Impact, Normal Shoulder Pad Configuration.

Top Impact Location					
	Normal Shoulder Pad Position	**Neck Collars**			
		Control	**Cowboy**	**Bullock**	**Kerr**
5 m/s	Test ID	top1	top2	top3	top4
	Actual Velocity [m/s]	4.90	4.97	4.93	4.95
	Resultant Head Acceleration [G]	31	31	28	24
	Impactor Force [N]	3533	3535	3630	3751
	Lower Neck Force (Fz) [N]	3550	3553	3181	3197
	Lower Neck % Force Reduction [%]	0	0	10	15
	Upper Neck Force (Fz) [N]	4210	4211	3751	3651
	Upper Neck % Force Reduction [%]	0	0	0	3
7 m/s	Test ID	top5	top6	top7	top8
	Actual Velocity [m/s]	6.44	6.41	6.49	6.25
	Resultant Head Acceleration [G]	42	41	34	34
	Impactor Force [N]	5203	5378	5260	6821
	Lower Neck Force (Fz) [N]	4496	4410	4039	4677
	Lower Neck % Force Reduction [%]	14	18	23	31
	Upper Neck Force (Fz) [N]	5334	5240	4718	5609
	Upper Neck % Force Reduction [%]	0	3	10	18

Table 8: Peak Values for Top Impact, Raised Shoulder Pad Configuration.

Top Impact Location					
	Raised Shoulder Pad Position	**Neck Collar**			
		Control	**Cowboy**	**Bullock**	**Kerr**
5 m/s	Test ID	top9	top10	top11	top12
	Actual Velocity [m/s]	4.92	4.93	4.92	4.87
	Resultant Head Acceleration [G]	29	30	28	22
	Impactor Force [N]	3574	3554	3451	3546
	Lower Neck Force (Fz) [N]	3205	3301	3387	2331
	Lower Neck % Force Reduction [%]	10	7	2	34
	Upper Neck Force (Fz) [N]	3781	3875	4005	2761
	Upper Neck % Force Reduction [%]	0	0	0	22
7 m/s	Test ID	top13	top14	top15	top16
	Actual Velocity [m/s]	6.58	6.41	6.36	6.25
	Resultant Head Acceleration [G]	41	39	32	32
	Impactor Force [N]	5006	4823	5196	6226
	Lower Neck Force (Fz) [N]	4262	4315	3729	3525
	Lower Neck % Force Reduction [%]	15	11	28	43
	Upper Neck Force (Fz) [N]	5034	5079	4380	4322
	Upper Neck % Force Reduction [%]	0	0	16	31

Table 9: Peak Values for Front Impact, Normal Shoulder Pad Configuration.

	Front Impact Location				
	Normal Shoulder Pad Position	Neck Collar			
		Control	Cowboy	Bullock	Kerr
5 m/s	Test ID	front1	front2	front3	front4
	Actual Velocity [m/s]	4.90	4.97	4.92	4.92
	Impactor Force [N]	3081	3043	2772	2894
	Resultant Head Acceleration [G]	49	48	47	43
	Lower Neck Force (Fx) [N]	611	621	567	387
	Lower Neck Moment (My) [N*m]	86	82	77	89
	Upper Neck Force (Fx) [N]	613	594	559	558
	Upper Neck Moment (My) [N*m]	51	49	54	35
	Range of Motion [deg]	25	23	19	15
7 m/s	Test ID	front5	front6	front7	front8
	Actual Velocity [m/s]	6.52	6.52	6.55	6.25
	Impactor Force [N]	4410	4530	4579	4870
	Resultant Head Acceleration [G]	60	62	65	70
	Lower Neck Force (Fx) [N]	676	650	707	533
	Lower Neck Moment (My) [N*m]	133	127	117	139
	Upper Neck Force (Fx) [N]	704	710	741	773
	Upper Neck Moment (My) [N*m]	64	67	73	41

Table 10: Peak Values for Front Impact, Raised Shoulder Pad Configuration.

	Front Impact Location				
	Raised Shoulder Pad Position	Neck Collar			
		Control	Cowboy	Bullock	Kerr
5 m/s	Test ID	front9	front10	front11	front12
	Actual Velocity [m/s]	4.89	4.98	4.85	4.90
	Impactor Force [N]	2828	2911	2886	2811
	Resultant Head Acceleration [G]	49	46	49	42
	Lower Neck Force (Fx) [N]	602	623	560	415
	Lower Neck Moment (My) [N*m]	78	72	87	54
	Upper Neck Force (Fx) [N]	566	577	583	505
	Upper Neck Moment (My) [N*m]	49	49	50	25
	Range of Motion [deg]	19	16	17	12
7 m/s	Test ID	front13	front14	front15	front16
	Actual Velocity [m/s]	6.61	6.58	6.52	6.25
	Impactor Force [N]	4577	4416	4351	4580
	Resultant Head Acceleration [G]	66	67	59	66
	Lower Neck Force (Fx) [N]	742	673	589	625
	Lower Neck Moment (My) [N*m]	125	105	123	90
	Upper Neck Force (Fx) [N]	747	661	674	726
	Upper Neck Moment (My) [N*m]	73	60	56	32

Table 11: Peak Values for Side Impact, Normal Shoulder Pad Configuration.

Side Impact Location					
Normal Shoulder Pad Position		Neck Collar			
		Control	Cowboy	Bullock	Kerr
5 m/s	Test ID	side1	side2	side3	side4
	Actual Velocity [m/s]	4.90	4.93	4.92	4.95
	Impactor Force [N]	2781	2893	2843	3040
	Resultant Head Acceleration [G]	54	64	61	61
	Lower Neck Force (Fy) [N]	452	442	430	398
	Lower Neck Moment (Mx) [N*m]	127	117	112	110
	Upper Neck Force (Fy) [N]	530	526	501	554
	Upper Neck Moment (Mx) [N*m]	35	34	30	35
	Range of Motion [deg]	39	37	37	25
7 m/s	Test ID	side5	side6	side7	side8
	Actual Velocity [m/s]	6.85	6.82	6.84	6.85
	Impactor Force [N]	3651	3968	3732	3646
	Resultant Head Acceleration [G]	75	79	78	75
	Lower Neck Force (Fy) [N]	607	626	537	557
	Lower Neck Moment (Mx) [N*m]	154	153	157	136
	Upper Neck Force (Fy) [N]	621	676	579	619
	Upper Neck Moment (Mx) [N*m]	57	61	59	53

Table 12: Peak Values for Side Impact, Raised Shoulder Pad Configuration.

Side Impact Location					
Raised Shoulder Pad Position		Neck Collar			
		Control	Cowboy	Bullock	Kerr
5 m/s	Test ID	side9	side10	side11	side12
	Actual Velocity [m/s]	4.89	4.98	4.85	4.92
	Impactor Force [N]	2611	2720	2494	2797
	Resultant Head Acceleration [G]	54	54	51	57
	Lower Neck Force (Fy) [N]	419	421	427	398
	Lower Neck Moment (Mx) [N*m]	112	108	112	91
	Upper Neck Force (Fy) [N]	449	430	471	475
	Upper Neck Moment (Mx) [N*m]	31	35	30	36
	Range of Motion [deg]	33	34	33	17
7 m/s	Test ID	side13	side14	side15	side16
	Actual Velocity [m/s]	6.81	6.82	6.85	6.88
	Impactor Force [N]	3830	3796	3860	3549
	Resultant Head Acceleration [G]	73	74	80	69
	Lower Neck Force (Fy) [N]	584	517	631	530
	Lower Neck Moment (Mx) [N*m]	149	132	146	118
	Upper Neck Force (Fy) [N]	572	491	650	406
	Upper Neck Moment (Mx) [N*m]	55	47	54	58

Discussion

In a top impact, the Kerr Collar provided the most protection, as measured by the load cells and accelerometers. The Bullock Collar provided some protection, while the Cowboy Collar did not protect the dummy from experiencing high neck loads. The Kerr Collar reduced the head acceleration and force transmission due to its unique design. The Kerr Collar is designed to contact the base of the helmet during an impact. In a top impact, this data suggests that the Kerr Collar redirects some of the load to the shoulders, on which the collar rests. The stiffness of the collar prevents the neck from further compression. Interestingly, impactor force actually increased when impacting the dummy equipped with the Kerr Collar. This is due to the base of the helmet contacting the collar, creating a more rigid system. Neither the Bullock Collar nor Cowboy Collar prevents the neck from compressing. However, the Bullock Collar was capable of reducing a small portion of the load in some configurations. This is most likely due to the back of the helmet contacting the collar.

The Kerr Collar also provided the most protection during an impact to the front of the helmet. It reduced upper neck moment and lower neck force in all configurations. The Kerr Collar also reduced the lower neck moment, but only in the raised configuration. Upon inspection of the high speed video, the collar restricts the range of motion of the head and neck by contacting the base of the helmet during the impact. This contact between the helmet and collar is responsible for the lower loads.

The Kerr Collar typically performed better in the raised position because it contacts the collar sooner and restricts more motion. This is true for any of the collars in the raised position. The Cowboy Collar and Bullock Collar also provided protection for the dummy throughout the front impacts. The reductions of loads were not as large and consistent as the Kerr Collar, but they were capable of reducing loads in some configurations.

In a side impact, none of the collars substantially reduced loads in multiple configurations. Only the Kerr Collar reduced the lower neck moment. Again, this is due to the base of the helmet contacting the collar, restricting the range of motion. This movement restriction is most noticeable in the high speed video. The Cowboy Collar and Bullock Collar provided no side impact protection.

The Kerr Collar performs differently than the other collars tested because it contacts the base of the helmet, which restricts motion of the head and neck. The Cowboy Collar and Bullock Collar are designed to prevent hyperextension of the neck. Therefore, the Cowboy Collar and Bullock Collar only reduce loads in front impacts. Restriction of motion correlates with load reductions for each of the collars. In the future, manufacturers should consider restricting the motion of the head and neck in more orientations than just hyperextension when designing collars. This restriction of motion should lead toward distributing loads to the shoulders, rather than the head and neck. It was also evident that the collars generally performed better when the shoulder pads were in the raised shoulder pad positions. This is mainly due to earlier contact with the neck collars.

Limitations

The Hybrid III neck has limited biofidelity. The complex anatomy and musculature of the human cervical spine distribute and reduce loads experienced in the neck. A mechanical model, such as the Hybrid III, cannot account for this; as it models the cervical spine as butyl rubber segmented by aluminum discs. While the Hybrid III neck responds similarly to a human's in a frontal crash test, it cannot perfectly replicate the complicated kinematics that occur during such impacts, and thus is an imperfect model. Even though there are biofidelity issues with the Hybrid III test dummy, it is the best available surrogate for a human. It is the standard for the automotive industry when testing to predict injuries in crash tests. The Hybrid III has also been used for various other applications, including football testing. Pellman reconstructed concussive football impacts with Hybrid III dummies using multiple angles of game video.[10] The Hybrid III dummy may not be perfectly biofidelic, but it is commonly accepted as a human surrogate and these types of tests have been performed in the past.[11] Moreover, the effect of the limited biofidelity is reduced by examining relative trends in the data, which is the goal of this paper.

Since only one test was done for each collar/shoulder pad/speed configuration, repeatability tests were conducted in order to determine which differences in performance could be considered significant. These tests proved there was some variability inherent in the test setup. However, the variance was an acceptable amount and some data channels proved more sensitive than others. Five consecutive tests in the control configuration were performed to assess the repeatability of this experiment. Coefficients of variation were determined by dividing the standard deviation by the average peak value for each data channel. Head acceleration was the most sensitive data channel with a coefficient of variation of 6.7%. Upper and lower neck moments had 1.6% and 2.9% coefficients of variation between the tests, respectively. Upper and lower neck forces had coefficients of variation of 4.5% and 3.9%, respectively. The neck collars' performance differences were greater than the coefficients of variation; and therefore could be used to determine differences between collars. In addition, these curves followed the same data-traces through time, which allowed for comparison between characteristics of the data throughout time.

Conclusion

A series of 48 tests were performed to assess the dynamic biomechanical effects of neck collars used in competitive football. Each neck collar was tested at two different impact speeds, at three different impact locations, and two different shoulder pad positions. With the top impact location, it was found that the Kerr Collar and Bullock Collar reduced head accelerations and force transmission through the neck. However, the Kerr Collar produced greater reductions in force transmission. The Cowboy Collar produced no reductions in a top impact. With the front impact location, all the collars reduced lower neck moment, while the Kerr Collar was also capable of reducing the lower neck force and upper neck moment. With the side impact location, the Kerr Collar produced the greatest lower neck moment reductions. These reductions in loads correlate with how much each collar restricted the motion of the head and neck. Overall, the collars performed better when the shoulder pads were in the raised configuration.

Acknowledgements

Thanks to Mike Goforth from the Virginia Tech Sports Medicine Department for providing the Center with equipment for this study.

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Chapter 7: Differences in Hybrid III and THOR-NT Neck Response in Extension Using Matched Tests with Football Neck Collars

Abstract

Anthropometric test devices have been used in sports injury biomechanics research. This study addresses the differences in the head and neck response of the Hybrid III and THOR-NT 50th percentile male crash test dummies when used to evaluate the load limiting capabilities of football neck collars. 24 matched tests were performed with the Hybrid III and THOR-NT; in which they were equipped with shoulder pads, a helmet, and various neck collars. The dummies were then impacted on the front of the helmet using a pneumatic linear impactor to promote extension of the neck. Results from these tests indicate that the Hybrid III generates greater loads than the THOR-NT due to its stiffer neck. The Hybrid III was also more sensitive to impact velocity. The neck collars had different effects on each dummy, typically affecting the Hybrid III's response more. Even though this study looks at a specific application, it highlights differences in neck response between the Hybrid III and THOR-NT.

Keywords: Hybrid III, THOR-NT, football, neck, collar, spine, injury

Introduction

Anthropometric test devices, such as the Hybrid III and THOR-NT, have a wide spread of applications beyond the automotive industry. These dummies are used when human volunteers or cadavers are not a viable option. A common alternate application that test dummies are used for is sports injury biomechanics research. The Hybrid III dummy has been used to evaluate sports injury prevention devices, as well as to determine the loads experienced by players during sporting events. Using video of various football games, Pellman (2003) reconstructed helmet-to-helmet concussive impacts with two Hybrid III dummies [1]. Hybrid III head and neck assemblies have also been used to evaluate football helmets. Manoogian (2006) compared the helmet acceleration relative to the head acceleration by impacting a helmeted Hybrid III head [2]. The Hybrid III has also been used to evaluate the effectiveness of neck protection devices used in football [3].

In the past, the Hybrid III has been the gold standard for the automotive industry when used to predict injury. Therefore, the majority of sports injury biomechanics research has utilized the Hybrid III dummy as a human surrogate. However, the recent introduction of an advanced dummy, the THOR-NT, has provided another dummy that may be used for such testing.

When examining the head and neck response of the Hybrid III and THOR-NT, it is important to understand the differences in design between the two dummies. The Hybrid III neck models the neck as butyl rubber segmented by aluminum discs. A steel cable runs through the center of the neck, which serves to control extension and flexion. The THOR-NT neck assembly consists of a series of elliptically shaped rubber pucks that are segmented by aluminum discs. The THOR-NT neck also contains a center cable. The major difference in design between the two necks is that the THOR-NT is designed to account for the muscle effects of the neck. Compression springs located on the front and rear of the THOR-NT neck are used to model the muscle effects of the neck. Research has shown that the Hybrid III has a stiffer neck response than the THOR-NT [4]; while the THOR-NT produces a more biofidelic response than the Hybrid III [5].

This study focuses on comparing the response of the Hybrid III and THOR-NT during a sports injury biomechanics application. Specifically, it uses a methodology used to assess the load limiting capabilities of neck collars used in football. Neck collars are worn by football players to prevent neck injuries, such as stingers [6, 7]. Due to the severity of injurious impacts in football, human volunteers cannot be tested using dynamic impact testing. Therefore, anthropometric dummies must be used to determine the load limiting capabilities of these collars in a dynamic impact environment. However, with various test dummies available, it is important to identify differences in neck response between the dummies. This study addresses the differences in neck response between the Hybrid III and THOR-NT while being used to evaluate the effectiveness of such collars.

Methods

A series of 24 matched tests were performed on the 50th percentile male Hybrid III and the 50th percentile male THOR-NT anthropometric test devices. Each dummy was suited with a set of

Douglas CP25 shoulder pads and a large Riddell VSR4 helmet for all tests. A pneumatic linear impactor was used to strike the front of the helmet at two impact velocities. Tests were first performed with neither dummy wearing a neck collar. Following these control tests, the same tests were performed with the dummies wearing various neck collars.

Two different neck collars were used in this comparison: the Cowboy Collar and the Bullock Collar. These neck collars are discussed in detail by Rowson [3]. The impact location was just above where the facemask meets the helmet shell. Impacts at this location forced the neck of each dummy into extension. In an effort to simulate actual football impacts and to account for the various ways a player may get hit, different shoulder pad positions were also tested. This involved testing the shoulder pads in a normal and raised position. In order to raise the shoulder pads, shoulder implants were made for each dummy using expanding polyurethane foam. These implants were secured on the shoulders of the dummy for the raised shoulder pad tests

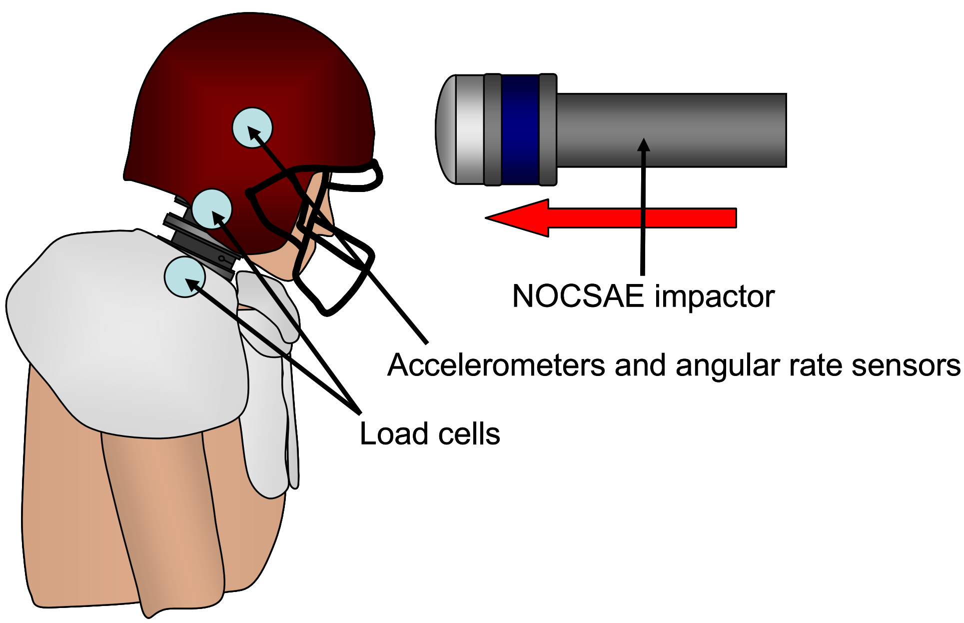

Figure 23: Testing setup.

[Image created by Steve Rowson]

The helmet was positioned on each dummy using a custom helmet positioning tool. This positioning tool used facial features on the Hybrid III's face to consistently fit the helmet on the head. Since the THOR-NT's head does not contain any facial features, targets were place on the face of the THOR-NT, allowing use of the positioning tool for a consistent fit. Each dummy was placed in a sitting position in front of the impactor on a custom milling table. The milling table allowed each dummy's location relative to the impactor to be precisely controlled. A target was place on the desired impact location, and the dummies were moved so that the target lined up with the impactor face. This method of controlling the orientations of the dummies resulted in consistent and matching impact locations for each test.

The impact velocities were 5 m/s and 7.5 m/s. These were chosen to represent a range of moderately severe impacts, and fall within the lower end of the spectrum of impact velocities described by Pellman [1]. The impacts were performed with the pneumatic linear impactor described by Rowson [3]. The impactor arm was instrumented with a load cell (Denton 1968 LC-91, Rochester, MI) and an accelerometer (Endevco B40592, 2000 G, San Juan Capistrano, CA). A light gate (Omron E3S-AT11, Schaumburg, IL) was used to measure the velocity of the impactor arm as it contacted the dummy. A high-speed video camera (Phantom V4, Wayne, NJ) recorded each test at 1000 frames per second. All instrumentation was sampled at 10,000 Hz and processed in accordance with SAE J211.

The Hybrid III was fitted with three single-axis orthogonally mounted accelerometers (Endevco B40351 B40234 B40740, 2000 G, San Juan Capistrano, CA) at the center of gravity of the head. Its neck was instrumented with upper and lower neck load cells (Denton 1716A LC-592 and 1794A LC-242, Rochester, MI) which provided forces and moments for each axis. The THOR-NT was also instrumented with three single-axis accelerometers (Endevco B40351 B16828 B17366, 2000G, San Juan Capistrano, CA) in the center of gravity of the head. Its neck was instrumented with upper and lower neck load cells (Denton LC-77 and LC-79 Rochester, MI) as well as with front and rear load cells attached to the compression springs (Denton LC-76 and LC-78, Rochester, MI).

In order to compare the loads of the neck upper neck of the Hybrid III and THOR-NT, the loads for each dummy was summarized about the occipital condyle pin. To do this, the forces and moments in the neck were transformed to the coordinate system of the head. Then, the forces and moments were summed about the occipital condyle pin.

Results

The impact locations and velocities of each test were precisely controlled, with impact velocities averaging 5.32 ± 0.05 m/s and 7.66 ± 0.06 m/s. At 5.3 m/s, the Hybrid III had an average head acceleration of 40.9 ± 2.6 g and that of the THOR-NT was 43.3 ± 1.6 g. At 7.7 m/s, the Hybrid III averaged 113.9 ± 4.9 g and the THOR-NT averaged 77.2 ± 2.9 g.

Figure 24 displays the upper neck force experienced in each dummy as a function of impact velocity. At 5.3 m/s, the Hybrid III had an average upper neck force of 541.2 ± 20.2 N and that of the THOR-NT was 272.1 ± 29.6 N. At 7.7 m/s, the Hybrid III averaged 881.3 ± 54.8 N and the THOR-NT averaged 382.7 ± 63.1 N. Figure 25 displays the upper neck moment of each dummy as a function of impact velocity. At 5.3 m/s, the Hybrid III had an average upper neck moment of 32.0 ± 5.0 N*m and the THOR-NT averaged 17.2 ± 3.5 N*m. At 7.7 m/s, the Hybrid III averaged 55.6 ± 9.7 N*m and the THOR-NT averaged 31.3 ± 6.4 N*m.

Figure 26 displays the lower neck force experienced in each dummy as a function of impact velocity. At 5.3 m/s, the Hybrid III had an average lower neck force of 314.1 ± 38.4 N and that of the THOR-NT was 553.1 ± 62.8 N. At 7.7 m/s, the Hybrid III averaged 632.2 ± 73.0 N and the THOR-NT averaged 652.0 ± 89.8 N. Figure 27 displays the lower neck moment of each dummy as a function of impact velocity. At 5.3 m/s, the Hybrid III had an average lower neck moment of 128.9 ± 16.6 N*m and the THOR-NT averaged 43.8 ± 2.8 N*m. At 7.7 m/s, the Hybrid III averaged 244.7 ± 23.2 N*m and the THOR-NT averaged 66.9 ± 1.9 N*m.

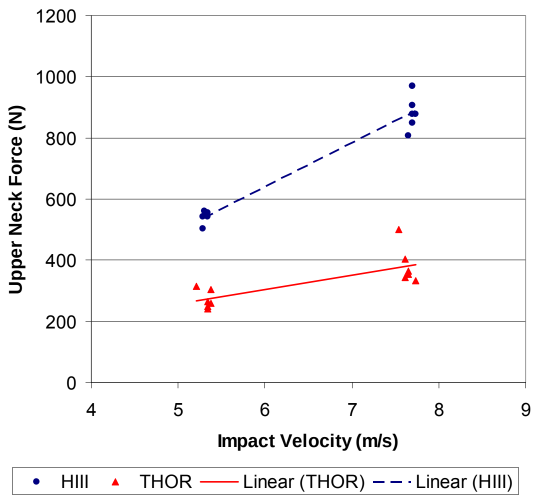

Figure 24: Upper neck force as a function of impact velocity.

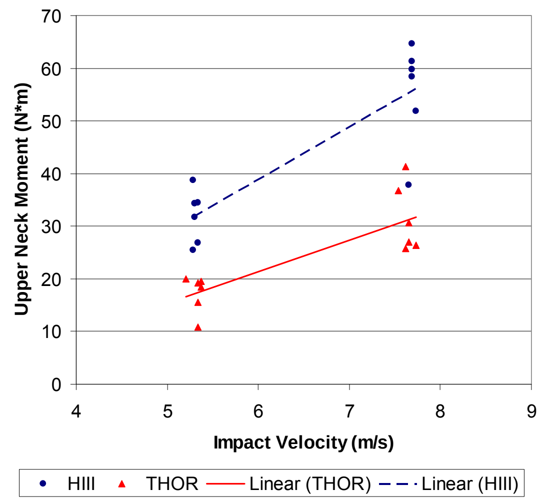

Figure 25: Upper neck moment as a function of impact velocity.

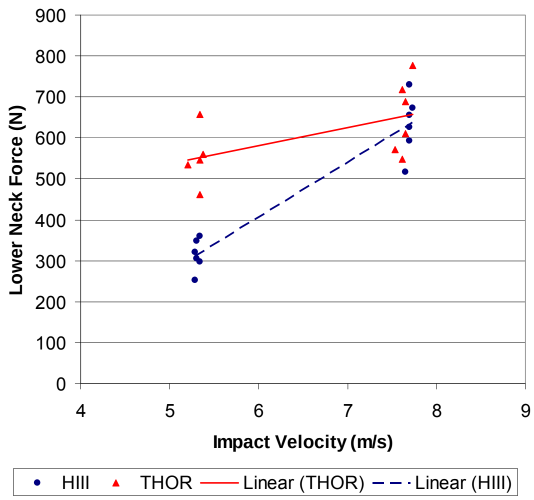

Figure 26: Lower neck force as a function of impact velocity.

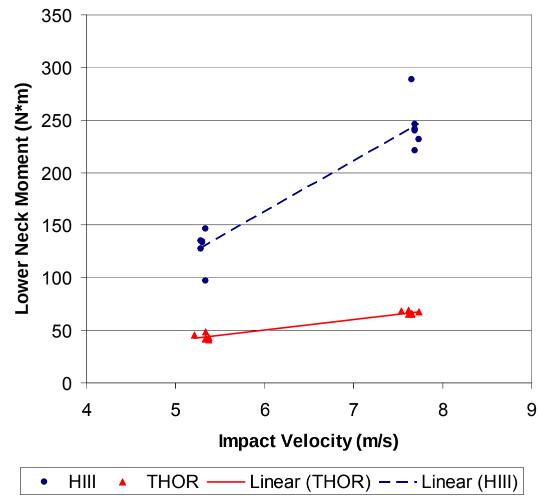

Figure 27: Lower neck moment as a function of impact velocity.

Figure 28 and Figure 29 describe the overall effect of adding the neck collars to the shoulder pads for both the Hybrid III and THOR-NT. In these figures, Accel is resultant head acceleration, Ufx is upper neck force, Umy is upper neck moment, Lfx is lower neck force, and Lmy is lower neck moment. These values were calculated by normalizing the neck collar value to the impactor force, and then dividing that value by the control value normalized to the control impactor force. This resulted in a percent of the control for each neck collar for peak resultant head acceleration, upper neck force, upper neck moment, lower neck force, and lower neck moment. These percent reductions were then averaged for both impact velocities and shoulder pad positions to look at the overall effect of adding neck collars to the shoulder pads. The effect of adding the neck collars generally reduced all loads except lower neck force, which slightly increased with the addition of the neck collars. However, the amount each neck collar reduced these loads varied between the Hybrid III and THOR-NT.

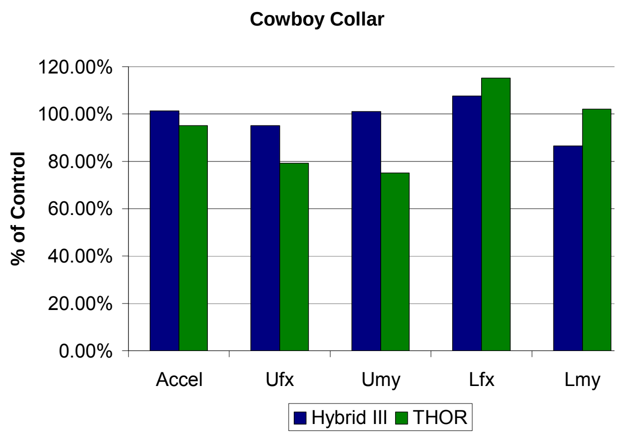

Figure 28: Effect of adding the Cowboy Collar to the shoulder pads for both the Hybrid III and THOR-NT.

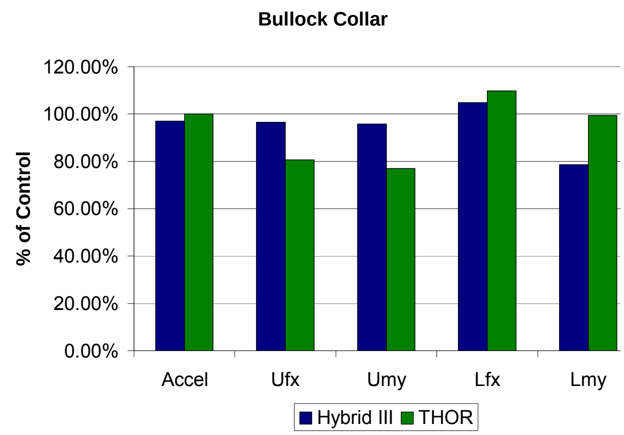

Figure 29: Effect of adding the Bullock Collar to the shoulder pads for both the Hybrid III and THOR-NT.

Discussion

With the 5.3 m/s impact velocity, the Hybrid III and THOR-NT had no difference in head acceleration. However, the Hybrid III produces higher head accelerations than the THOR-NT at 7.7 m/s. This implies that the head acceleration of the Hybrid III is more sensitive to impact velocity than the THOR-NT is. The small standard deviations show that neck collar and shoulder position do not greatly affect head acceleration for either dummy.

In general, the Hybrid III produced greater loads in the upper neck than the THOR-NT. The Hybrid III produced higher forces in the upper neck than the THOR-NT in these tests. The trend lines in Figure 24 show that upper neck force is more sensitive to impact velocity for the Hybrid III than it is for the THOR-NT. There is some scatter for the upper neck force of each dummy. This suggests that the neck collars had an effect on upper neck force. The Hybrid III also produced higher upper neck moments than the THOR-NT. The upper neck moment trend lines are similar in slope, as seen in Figure 25; this suggests that the sensitivities of upper neck moment to impact velocity for both dummies are similar. The neck collars and shoulder pad position had a large effect on the upper neck moment for the Hybrid III, which is evident by the large amount of scatter. The THOR-NT had much less scatter than that of the Hybrid III, therefore the neck collars affected the upper neck moment of the THOR-NT less than that of the Hybrid III.

The lower neck forces experienced by the Hybrid III and THOR-NT are similar to one another. At 5.3 m/s, the THOR-NT averaged a lower neck force of lesser magnitude than the Hybrid III. At 7.7 m/s, the Hybrid III and THOR-NT had no difference in lower neck force. The scatter in Figure 26 suggests that neck collar and shoulder pad position had an effect on lower neck force for both dummies. The Hybrid III experienced greater lower neck moments than the THOR-NT at both impact velocities. The lower neck moment of the Hybrid III is also more sensitive to the impact velocity than the THOR-NT. In addition, the Hybrid III also produced greater lower neck moment scatter than the THOR-NT. This implies that the neck collars and should pad position had a greater effect on the lower neck moment of the Hybrid III than the THOR-NT.

Figure 28 and Figure 29 demonstrate the differences the addition of neck collars resulted in for each dummy. Although the each dummy exhibited the same trends, in that either a load was reduced or increased, each load was reduced or increased to a different degree. This is due to the neck collars interacting differently with each dummy's head, neck, and shoulder assemblies. The THOR-NT has been shown to have a more human-like neck response in extension than the Hybrid III [5]. Therefore, the authors would suggest using the THOR-NT when looking at neck response in extension. The Hybrid III consistently generated greater loads than the THOR-NT in this set of testing due to its stiffer neck.

Conclusion

For this application, the dummies exhibited differences in upper and lower neck forces and moments. The Hybrid III was typically more sensitive to impact velocity, as its trend line slopes were most often greater than that of the THOR-NT's. In addition, the neck collars and shoulder pad position affected both dummies, but typically had a greater effect on the Hybrid III. Upper neck moment and lower neck force were most affected by the presence of the neck collars and position of the shoulder pads. Due to the stiffer neck design of the Hybrid III, the Hybrid III generally experienced greater neck loads when compared to the THOR-NT. Even though this testing looks at the specific application of using dummies to evaluate neck collars used in football, it highlights that the Hybrid III and THOR-NT's necks perform differently during dynamic impact testing.

Acknowledgements

The authors would like to thank Peter Martin of NHTSA for providing the THOR-NT used in this study.

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Chapter 8:
Closing Remarks

Research Summary

The research presented in this thesis investigates the biomechanics of the head and neck during impacts in football. Using innovative methodologies to study various types of loading, insightful data was collected. Each study addressed the methodological limitations of previous research. By instrumenting football helmets with sensors for every game and practice in a football season, a large and unbiased dataset of human head acceleration was compiled. By measuring chin strap tension during an impact to a football helmet, a model could be used to estimate the force transmitted to the mandible. By using highly instrumented human surrogates, the load limiting capabilities of neck collars could be quantified at potentially injurious impact severities. While a novel start, the methodologies of each study can be utilized in the future to create expanded datasets. Such data should have applications in future football protective equipment design.

While the neck collars and chin strap studies may be limited to football, the head acceleration study has applications beyond the football field, particularly with expanding the head acceleration dataset to include concussive impacts. Quantifying human brain biomechanics as a result of impact will ultimately lead to a better understanding of human brain injury. This will influence all vehicle-related safety standards in terms of head injury. In addition, these data could serve as validation data for computation models, resulting in improved human tissue tolerance data. Eventual conclusions drawn from such data may influence vehicle safety design, as well as sport and automotive helmet design.

Publication Outline

The research presented in this thesis is intended to be published in several journals and presented at various conferences. Table 13 displays the publication destinations for each research topic. All chapters are in their publication forms, with the exception of Chapter 4, which will be part of a larger study.

Table 13: Publication plan for research presented in this thesis.

Chapter	Title	Journal / (*Conference)*
2	Validation of a Six Degree of Freedom Head Acceleration Device for Use in Football	Journal of Applied Biomechanics (*NHTSA Human Subjects Workshop 2007*)*
3	Linear and Angular Head Acceleration Measurements in Collegiate Football	American Journal of Sports Medicine
4	Computational Modeling of Football Head Acceleration Data Using SIMon	Stapp Car Crash Journal
5	Force Transmission to the Mandible by Chin Straps during Head Impacts in Football	(*Biomedical Sciences Instrumentation*)*
6	Biomechanical Analysis of Football Neck Collars	Clinical Journal of Sports Medicine* (*Biomedical Sciences Instrumentation*)*
7	Differences in Hybrid III and THOR-NT Neck Response in Extension Using Matched Tests with Football Neck Collars	(*Biomedical Sciences Instrumentation*)*

* Accepted