Analysis of Linear Head Accelerations From Collegiate Football Impacts

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(ABSTRACT)

Sports related concussions result in 300,000 brain injuries in the United States each year. The purpose of this study was to utilize an in-helmet system that measures and records linear head accelerations to analyze head impacts from collegiate football. The Head Impact Telemetry (HIT) System is an in-helmet system with six spring mounted accelerometers and an antenna that transmits data via radio frequency to a sideline receiver and laptop computer system. The data reported by the HIT System includes the time of impact, location of impact, and linear acceleration resultant of the head center of gravity. The algorithm and in-helmet application for this system were validated in five series of tests. In particular, the validation emphasized that the HIT System measurements are of the player's head and not helmet accelerations as evidenced by the helmet acceleration being 16.6 (± 3.2) times greater than the peak head accelerations measured by the HIT System and dummy cg accelerometers. Using 130 head impacts in five different test configurations, the mean error in measuring peak linear acceleration was 0.01% (±18%). A total of 11,604 head impacts were recorded from the Virginia Tech football team throughout the 2003 and 2004 football seasons during 22 games and 62 practices from a total of 52 players. The acceleration data distribution was right skewed with a mean impact acceleration magnitude of 20.9 g and maximum value of 172.6 g. The HIC determined for each impact from

a 15 millisecond time period was similarly distributed with the mean equal to 17.9, and the maximum equal to 969.6. A total of three impacts with three different players resulted in concussions. These impacts had peak linear accelerations of 55.7 g, 136.7 g, and 117.6 g with HIC values of 120.6, 518.4, and 355.6 respectively. Each of these hits was among the highest recorded for each respective player. Although the incidence of injury data is limited, this study presents an extremely large data set from human head impacts that provides valuable insight into the lower bounds of mild traumatic brain injuries.

Table of Contents

Chapter 1

List of Figures

Chapter 1

List of Tables

CHAPTER 1

INTRODUCTION

Brain injuries occur in 1.5 million people in the United States each year, of which 75 percent are mild traumatic brain injuries (MTBIs) like concussions (Gerbeding, 2003). Sports related concussions constitute 300,000 of these injuries (McCrea, 2003). When Guskiewicz (2003) studied National Collegiate Athletic Association (NCAA) football players to examine the effects and recovery from a concussion, he found that 6% of the participants received a concussion in the years 1999 to 2001. The NCAA reported that in the 2002-2003 college football season 8% of all collegiate football injuries were brain injuries. This high speed and full contact sport has had an increasing rate of brain injury in the last seven years (McCrea, 2003).

Sports related concussion research has been advancing for years, leading to a greater insight into the cause and effect of a mild traumatic brain injury (ACRM 1993 and Gadd 1966). In the early 1970s, Moon and Reid (1971 & 1974) instrumented the headbands of suspension-style football helmets with a frequency modulation (FM) based accelerometer and electroencephalogram (EEG) system. Moon (1971) recorded several impacts up to 1000 g with no injury. Reid (1974) had one instrumented player during the 1970 Northwestern University football season receive 169 impacts in 418 plays with accelerations ranging from 40 g to 230 g. One concussion was recorded with a peak acceleration of 188 g and impact duration of 310 ms. Morrison (1983) utilized a similar system in the early 1980s at Penn State, although without the EEG capabilities. Impacts measured using his system yielded mean accelerations of 349 g for blocking and 280 g for tackling. While laying the ground for future research and providing a proof of concept, these studies were limited in their ability to measure head acceleration and to only a single player. In 2000, Naunheim (2000) instrumented hockey and football helmets with a padding-embedded tri-axial accelerometer. The average peak acceleration measured for the 158 impacts was 29.2 g, the subjects were high school level athletes, and no incidents of concussion or MTBI were recorded.

Recently, methods of head acceleration measurement have combined video analysis and dummy reenactments of impacts from game film. Newman (1999 & 2000) and Pellman (2003 & 2004) published a series of papers based on a National Football League (NFL) study of concussions in professional football. In this series they studied concussive impacts that were recorded on film from two or more different angles. They used this video data to reconstruct the angle of the impact, speed of the impact, and the resultant player kinematics. These data provided the necessary information to recreate the impact conditions with instrumented Hybrid-III dummies in the laboratory. There were 174 impacts analyzed for this data set. From the impacts in which a player received a brain injury the mean linear head acceleration was 98 g, while the cases without a brain injury had a mean linear head acceleration of 60 g.

King (2003) used the information from these NFL impacts to further quantify the brain kinematics using a computer model. From 53 NFL cases a logistic regression was computed with linear acceleration as the independent variable and the occurrence of an MTBI as the dependent variable. The outcome of this procedure predicted a 25%, 50%, and 75% probability of MTBI at a peak linear acceleration of 57.0 g, 79.3 g, and 98.4 g respectively. A regression analysis was also performed with the HIC determined for each impact from a 15 millisecond time period (HIC15) as the independent variable, estimating 25% probability of injury at 136, 50% at 235 and 75% at 333. These probabilities are predicted from 22 injury cases and 30 non injury cases.

The dependence on video reconstruction and dummy reenactments creates an indirect measure of head acceleration that is limited by dummy biofidelity assumptions and validation from 30 Hz video. In addition, previous research has not provided real-time information on the direction and magnitude of the impacts football players receive. Given the serious health concerns associated with concussions, there is a critical need for real-time measurement of head accelerations that can be readily applied to a large number of players. The purpose of this study is to utilize an in-helmet system that measures and records linear head accelerations and to record a large data set from human head impacts that provides valuable insight into the lower bounds of mild traumatic brain injuries.

METHODS

This study utilized the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). Tests at five facilities validated the algorithm and the accelerometer array in a standard football helmet. For two years, the 2003 and 2004 seasons, the system was used in conjunction with the Virginia Tech football team to monitor collegiate football head impacts during practices and games. The sports medicine team provided the clinical assessment necessary to determine brain injuries as a result of these impacts.

Instrumentation

The HIT System is a wireless system that provides real time data from impacts to a signal receiver and laptop computer system located on the sidelines. Each monitored player wears an in-helmet Player Unit (Figure 1) designed to fit a Riddell VSR-4 L or XL (Figure 2) football helmet. Spring mounted accelerometers keep constant contact with the head to ensure measurements are of head rather than helmet accelerations. The Player Unit consists of 6 linear accelerometers and 1 temperature sensor, a 903-927 MHz wireless transceiver, up to 120 impacts of on-board memory, and 8 bit data acquisition with 1000 Hz sampling. Data are collected for 40 ms when any single accelerometer detects an acceleration that exceeds a user-selected threshold set at 10 g. Since the trigger is from one accelerometer, it is possible to have the calculated linear or rotational components below the individual threshold. To ensure the entire waveform has been collected, 12 ms of data are stored pre-trigger with 28 ms post-trigger for a total of 40 ms. The acceleration data are time-stamped and wirelessly transmitted to a Sideline Controller interfaced to a laptop computer (Figure 3). Up to 64 players can be monitored simultaneously with a single Sideline Controller. If wireless communication is not present the individual helmet unit stores up to 120 impacts in non-volatile memory and is downloaded once communication has been re-established. The location of the impact and the magnitude of the linear acceleration of the head at any given time point are calculated from the individual accelerometer values and a previously described algorithm (Crisco, 2004).

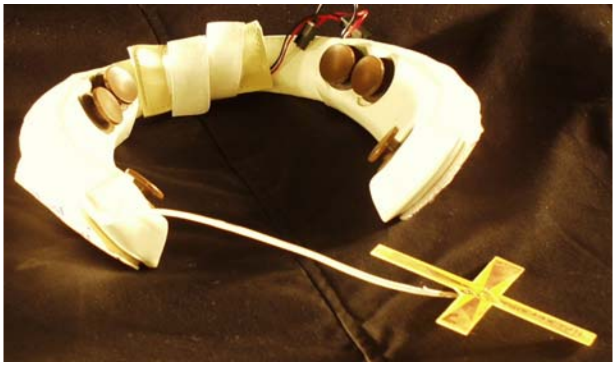


Figure 1: The Player Unit consisting of six accelerometers in spring-loaded holders, FM-antenna, and rechargeable battery pack.

Figure 2: Virginia Tech helmet with the Player Unit installed.

Figure 3: Sideline controller displays impact information upon downloading data.

There are different ways to quantify an impact using global measures. Most commonly, peak linear acceleration of the head center of gravity (cg) is measured. However the resultant linear acceleration of the cg over time has been used to calculate additional injury measures. The injury criteria widely used in motor vehicle research is Head Injury Criteria (HIC). An equation using the integral of acceleration with respect to a given amount of time, this study uses 15ms, is calculated over the entire duration of a pulse and the maximum value is the HIC for that impact (Hess 1981). This paper will report both measures to approximate the intensity of the impact. The coordinate system for the HIT System algorithm uses a positive x-axis out of the back of the head, positive y-axis out the right ear and positive z-axis out of the top of the head with (0,0,0) at the cg of the head. Azimuth is defined from -180 to 180 degrees as the angle of impact from the sagittal plane. Zero degrees is an impact directly in the sagittal plane from the rear of the head, negative values indicate any impact to the left side of the head and positive values correspond to impacts on the right side. Elevation is defined as the angle of impact from the Frankfurt plane. Positive angles represent impacts above this plane and negative angles represent impacts below this plane.

For a game environment, the field setup for this study had the receiver located at the 20 yard line of the visitor's sideline with approximately an 80 yard range. During practice the system was setup at a central location for all practice fields. The six single accelerometer traces and analysis are presented to the sideline unit controller on the laptop screen in four frames: the vector showing the current impact location, the event's impact location history, acceleration versus time graph of traces, and an acceleration magnitude history (Figure 4). Recorded data are time-stamped and then analyzed after a practice or game. The data time stamps are correlated with video footage taken during both practices and games in order to provide insight into the injury mechanisms for specific impacts. Full impact practices provide data that has a similar distribution to a game setting and therefore allowed for additional impact events to be collected.

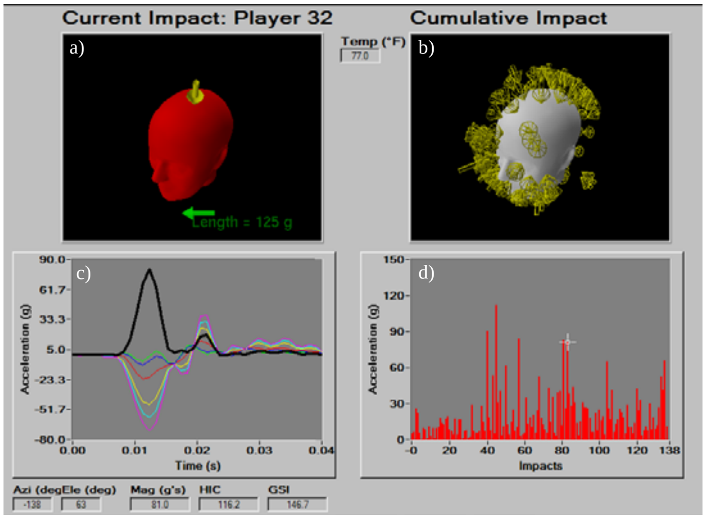


Figure 4: Sideline controller laptop screen. a) A directional vector indicates the most recent hit location. b) Directional vectors for one game or practice. c) Six accelerometer traces and the computed linear acceleration (black) are plotted versus time for 40 ms. d) Peak linear acceleration magnitude history is presented in bar chart format.

Sensor Validation

Five series of tests including 130 impacts were conducted to validate the algorithm and helmet application of the HIT System used at Virginia Tech. These test methods and results validate the head acceleration measurements presented in this paper (Appendix A). First, the general response of the algorithm in an ideal configuration was tested through computational and experimental methods (Crisco, 2004). Since these tests showed the six accelerometer configuration to reasonably predict the cg accelerations, the system was fitted to a standard Riddell VSR-4 L helmet. Second, testing was completed using a pneumatic linear impactor to ensure the spring mounted accelerometers produced desired results on the Hybrid III headform (Figure 5). Third, a pendulum impactor was used to simulate helmet to helmet low speed impacts and measure helmet accelerations. Fourth, an ASTM impact system was used to collect impacts at different locations on a standard test rig designed to fit a large football helmet. Lastly, high velocity tests were performed with a pendulum to test several impact locations to further validate the acceleration and impact location measured by the HIT System. Additional accelerometers were placed on the inside of the helmet shell at the point of contact to measure helmet accelerations during the helmet to helmet pendulum impacts. Overall the peak acceleration for the helmet at the point of contact was 16.6 (± 3.2) times greater than the head cg peak linear acceleration for that impact. The peak magnitude, time to peak, and waveform shape measured by the HIT System matched the head cg acceleration values and not the helmet shell acceleration values (Figure 6). Supplementary tests were completed to comply with helmet standards. To demonstrate that the HIT System did not influence the safety performance of the helmet, Riddell VSR4 helmets with the HIT System Player Unit installed were tested at a third-party laboratory and passed the National Operating Committee on Standards for Athletic Equipment (NOCSAE) football helmet performance standard required for all helmets worn at the collegiate level.

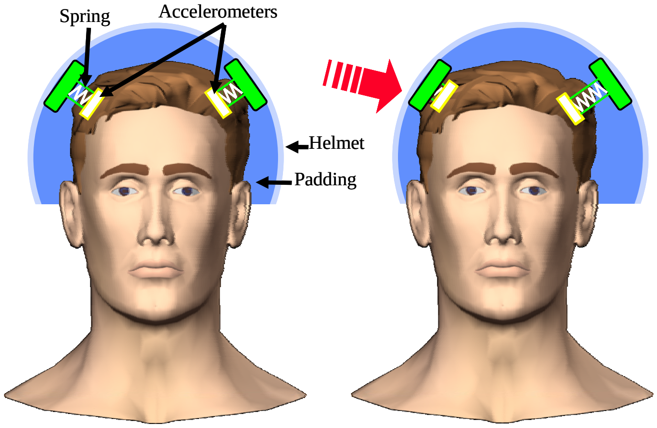


Figure 5: Spring mounted accelerometers guarantee measurements are of head and not helmet accelerations.

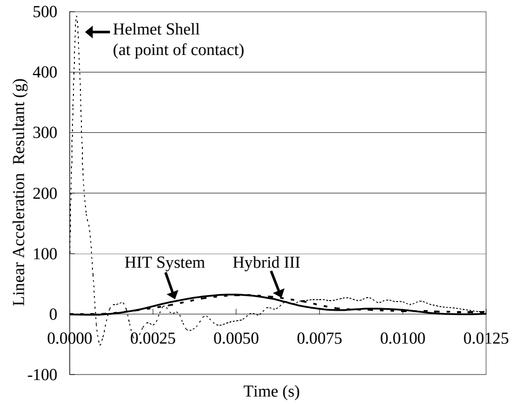
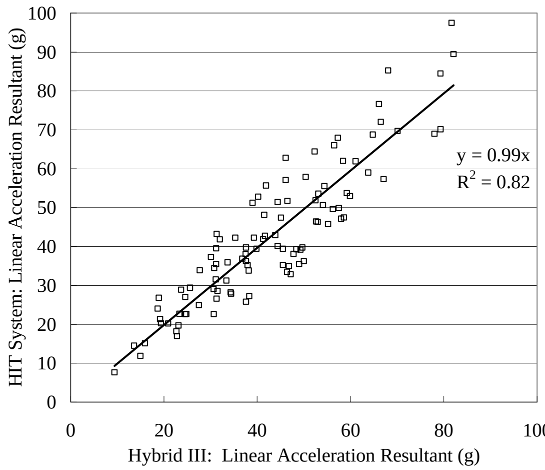


Figure 6: The peak helmet acceleration is an average of 16.6 (±3.2) times greater than the head cg acceleration.

All validation tests were plotted together to establish an overall range of accuracy for the HIT system measurements of head cg peak acceleration, head injury criteria, and impact location. The mean error in measuring peak head cg linear acceleration is 0.01% ±18% (Figure 7). The mean error in measuring HIC values is 30% ±43% (Figure 8). For measuring the location, the azimuth error was 10° ± 9°and the elevation error was 21° ±15°. While these values are not 100% accurate, I suggest that the error is acceptable given standard experimental error and the restraints imposed by using a system on a large population. However, the 30% error suggests HIC values should be more carefully considered.



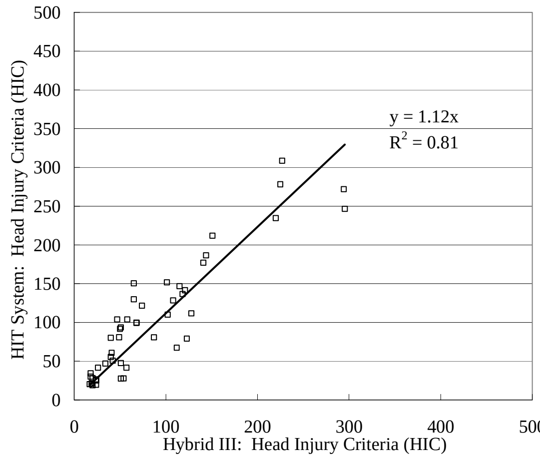


Figure 7: Overall the peak acceleration of the HIT System correlated to the peak acceleration at the Hybrid III cg with a linear slope of 0.99 (R^2=0.82).

Figure 8: Overall the calculated HIC from the HIT System correlated to the calculated HIC from the Hybrid III cg with a linear slope of 1.12 (R^2=0.81).

<u>Field Data Collection</u>

During each practice and game up to eighteen players were instrumented and monitored simultaneously for the 2003 and 2004 seasons. Over the two year period a total of 52 players were selected by the medical staff to provide a representative sample of all football players' body types and positions. Players were selected from seven position categories: defensive backs, defensive linemen, offensive linemen, running backs, wide receivers, line backers, and quarterbacks. In 2003 each player was instrumented for approximately a two week period that typically consisted of six practices and two games. However for the 2004 season each player selected was instrumented for the entire season. Data collection days included 67 practices and 22 games for both seasons. All players selected gave written informed consent with Institutional

Review Board approval from both the Virginia Tech and the Edward Via Virginia College of Osteopathic Medicine.

Clinical Injury Assessment

To analyze the mechanical data in conjunction with clinical aspects of concussion, the sports medicine staff maintains detailed player histories, including pre-season neuropsychological records. HeadMinder™, a web based neuropsychological assessment program, is used pre-season and post injury to compare a player's baseline cognitive status to his post injury evaluation (Erlanger, 2003). Following an injury, clinical indicators are closely observed and monitored by the medical staff. Clinical indicators include postural sway, confusion, retrograde amnesia, anterograde amnesia, and headache. A diagnosis is made based on the level of brain injury on a scale that ranges from Grade 1 to Grade 3 where Grade 3 is the most severe level (Kelly, 1997). When an injury initially appears more serious or symptoms persist, more elaborate neuropsychological tests and neuroimaging procedures are performed as clinically indicated.

RESULTS

Overall Head Impacts

For the 2003 and 2004 seasons a total of 11,604 head impacts were recorded with 2,970 occurring in 22 games and 8,634 in 67 practices. The impacts were measured from 52 different players covering the majority of offensive and defensive positions. The acceleration data distribution was right skewed with a median impact acceleration magnitude of 15.3, a mean value of 20.9 g (±18.7), and a maximum value of 172.6 g (Figure 9). The HIC15 was similarly distributed with the median equal to 3.1, mean equal to 17.9 (±50.4), and the maximum equal to 969.6 (Figure 10). Although the majority of impacts have low peak acceleration values, 290 of the impacts have a peak acceleration of 75 g or greater. Similarly, 487 of the impacts have a HIC of 100 or greater.

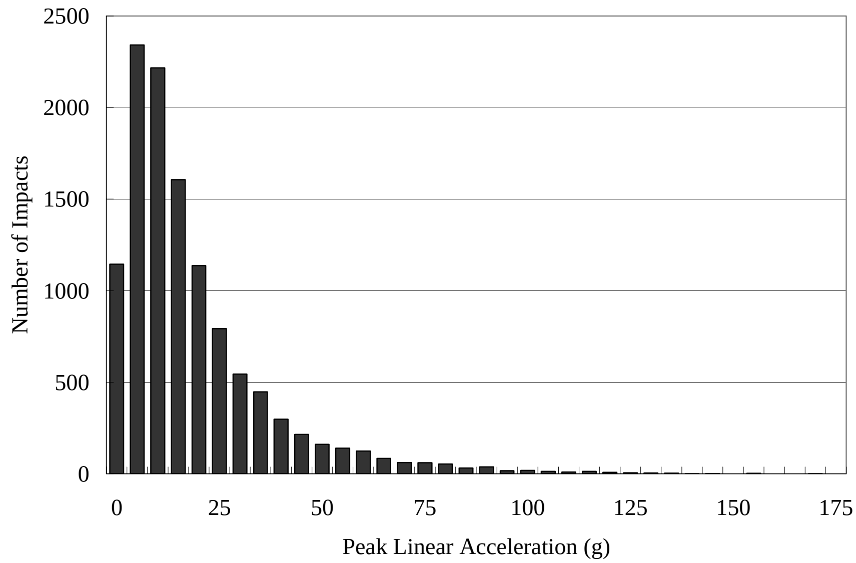


Figure 9: The distribution of peak linear accelerations is right skewed with 15.3 g as the median impact and a maximum impact magnitude of 172.6 g.

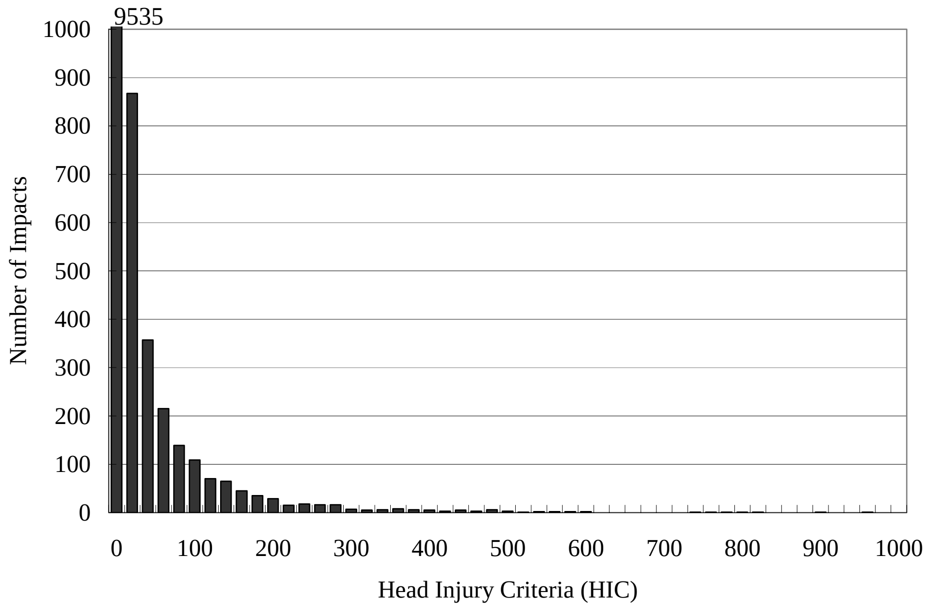


Figure 10: The distribution of HIC is right skewed with 3.1 as the median impact and a maximum HIC of 969.9.

<u>By Impact Location</u>

Four regions have been defined to classify the general location of an impact. Any impact with an elevation greater than 45 degrees above the Frankfurt plane is considered to impact the top of the head. The rest of the impacts are either to the front, side, or back of the head. Using the azimuth angle, the back is any impact from -45 to 45 degrees, the side is 45 to 135 or -135 to 45 degrees and the front is -180 to -135 or 135 to 180 degrees. The front region has the most impacts, 3,946, out of the four locations. The impact location with the highest median linear acceleration peak and HIC values is the back of the head. The top location has the fewest number of impacts, 1,581, and the lowest median linear acceleration peak and HIC values (Table 1).

Table 1: The distribution of impact frequency and magnitude to each of the four helmet regions indicated the most impacts were to the front of the helmet and the most severe impacts were to the back of the helmet.

	Front	Back	Side	Top
Number of Impacts	3946	3667	2410	1581
Median Linear Acceleration Peak (g)	14.2	17.9	14.7	13.0
Maximum Linear Acceleration Peak (g)	136.7	172.6	114.7	93.6
Median Head Injury Criteria (HIC)	2.5	4.4	2.8	2.7
Maximum Head Injury Criteria (HIC)	576.9	969.6	442.9	413.8

By Player Position

Football players were also grouped by position. The position indicates the player's body type as well as their form for hitting and receiving hits and frequency of hits. When the data are considered in seven position groups, there are natural trends in impact location, frequency and magnitude. Impact locations varied but were also clustered into areas with a high density of impacts that appeared to depend upon the player position as well as individual player technique (Figure 11). A single game's data display these observations. The defensive back exhibited a line of impacts in the sagittal plane with very few lateral impacts. Similarly, the running back exhibited the same band of impacts with a heavy concentration at the facemask. While displaying similar impact distributions to a running back, the defensive lineman had more variation in impact location including side impacts. The linebacker received mostly front and side impacts with a small concentration on the forehead. The overall data confirms these trends. The specific distribution of each position's impacts among the four locations indicates the defensive linemen and linebackers have a more even distribution with lateral impacts than the other six positions (Figure 12).

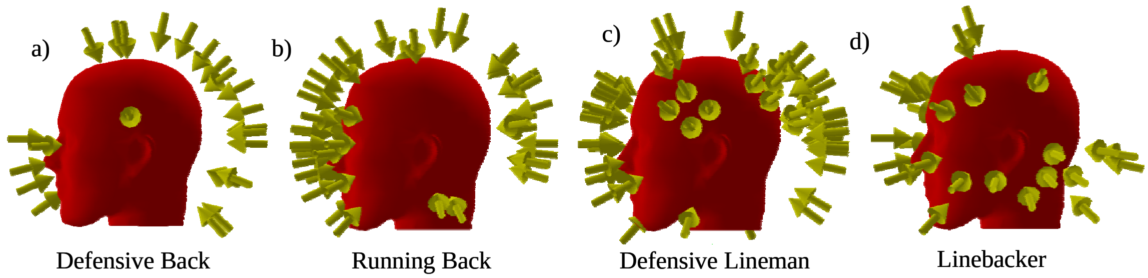


Figure 11: The defensive back and running back had mostly sagittal plane impacts while the defensive lineman and linebacker had a more even distribution with lateral impacts.

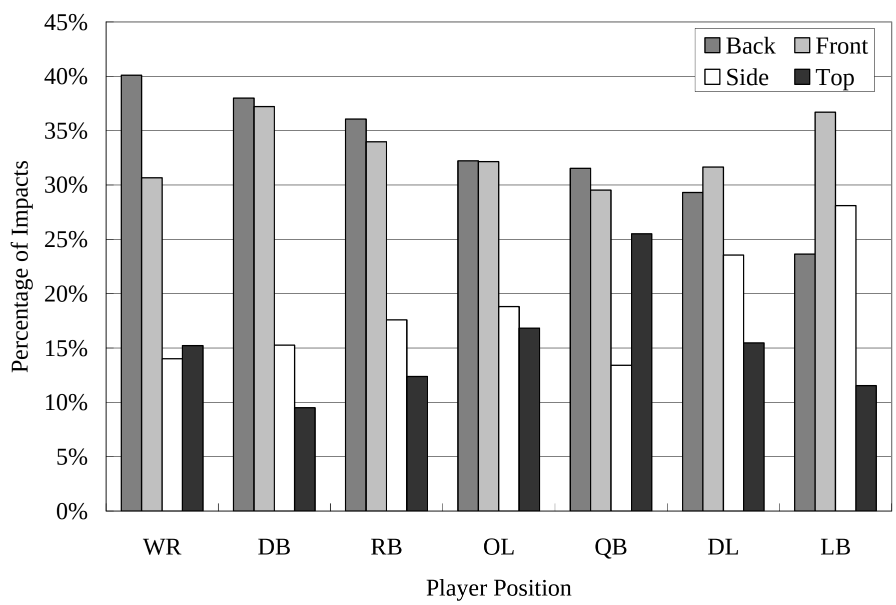


Figure 12: The highest percentage of impacts were to the back of the head for the wide receiver while the linebacker positions received the most impacts to the front of the head.

Although the distribution of high acceleration impacts varies by position, each position is exposed to high acceleration impacts. There are noticeable differences in impact frequency and magnitude. The frequency of impacts varies based on the role of the position. Some positions are more likely to receive one or more impacts every play while others are not. Therefore, the defensive linemen and linebackers have the highest average number (29 ±14) of impacts per game, and the wide receivers and offensive linemen have a low average of impacts per game (14 ±9). The variation in the magnitude of impacts distributed to these positions can be quantified by comparing the median linear acceleration peak and HIC values. Offensive lineman and linebackers have high median values for both acceleration and HIC while the defensive backs and quarterbacks have low median values (Table 2). All positions had maximum impacts of 115.1 g or higher.

Table 2: The distribution of players, impacts, and impact magnitude for each position is shown.

	DL	LB	WR	RB	OL	DB	QB
Number of Players	6	9	10	7	12	5	3
Number of Impacts	2610	2555	1801	1769	1409	1271	149
Median Linear Acceleration Peak (g)	13.9	19.0	14.5	13.4	17.1	13.0	12.6
Maximum Linear Acceleration Peak (g)	157.8	172.6	146.2	155.1	159.5	124.7	115.1
Median Head Injury Criteria (HIC)	2.5	5.4	2.8	2.4	4.0	2.1	1.8
Maximum Head Injury Criteria (HIC)	907.8	969.6	760.8	787.4	828.2	418.3	360.9

Brain Injuries

A more specific analysis of the data was performed for the three instrumented players that sustained a concussion during play. Video and information from the medical team was used to confirm the time and circumstances of the impact. The equivalent impact from the HIT System recorded the linear acceleration, direction, and HIC for each injury. This information is reported along with the clinical symptoms and diagnosis.

<u>Case 1</u>

In 2003 during a game, the second impact after the first half kickoff one player sustained a mild concussion as well as a neck injury from two of the 29 recorded impacts (Figure 13). This linebacker was blocking and received an impact that had a peak linear acceleration value of 55.7 g and HIC equal to 120.1. The location of the impact was on the right side, 164 degrees from the middle of the back of the head, in the facemask region and in-plane with the center of gravity of the head. This impact occurred as the opposing team's player struck this player straight-on, causing him to fall backwards and hit the ground. He did not experience loss of consciousness nor any other motor impairment and therefore stayed in the game. Although clinical symptoms were later discovered, he did not initially recognize his injury as a concussion. When reviewing the game film the following day with his position coach, he noted that immediately following this impact he could not remember the plays or where he should set up on the field. He was then referred to the medical staff for evaluation. Because the concussion was not reported until after the day of the event, a retrospective diagnosis was made suggesting the concussion was a Grade 1 or possibly a Grade 2 based on his history and clinical evaluation. Later in the game he received a neck injury as a result of the 16^{th} impact of the game: 74.0 g and 167.0 HIC. This hit location was 31 degrees to the left of the middle of the back of the head and 11 degrees above the center of gravity plane. Because of the immediate physical discomfort, he was aware of this injury and sought attention from the medical staff. This player had no prior history of concussions but did have a history of neck injury. This player had 125 recorded impacts during

2003 and 2004 but only 10 of these impacts had a peak acceleration or HIC value greater than the concussive impact. None of these ten were to the front of the head.

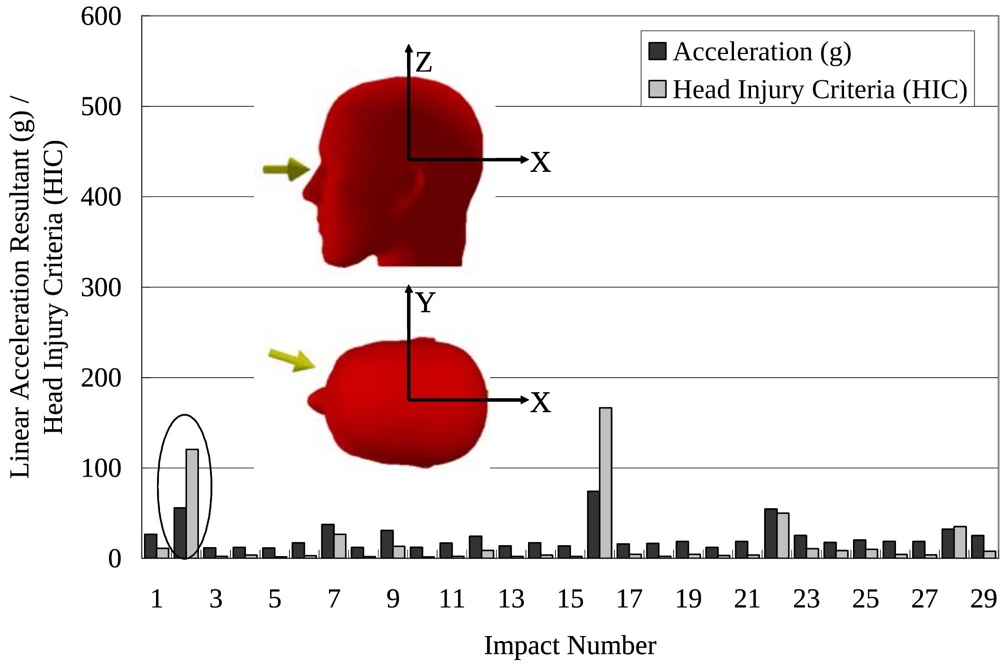


Figure 13: The second impact of the game (circled) resulted in a concussion when the player experienced a 55.7 g impact with a HIC of 120.1.

<u>Case 2</u>

During a fall practice in 2004, an offensive lineman was concussed by an impact from another player. This impact was a head to head impact during a full contact drill. The resultant linear acceleration peak from this impact was 117.6 g at 174 degrees from the back of the head and 9 degrees above the transverse cg plane (Figure 14). The corresponding HIC value was 355.6. Although he knew it was a hard hit, he did not recognize it as a concussion until after practice when the symptoms progressed. He suffered a post-traumatic headache and slight instability on the Rhomberg test. The physician diagnosed it as a Grade 1 concussion. According to the

HeadMinder™ evaluation post injury his cognitive functioning was not affected by the concussion compared to his baseline study. Prior to this incident the player had a history of four concussions. From 441 total impacts recorded for this player, only four had a peak acceleration or HIC value greater than this concussive impact. One of these four was to the front of the head. His symptoms resolved quickly and he had no further clinical symptoms for the remainder of the competitive season.

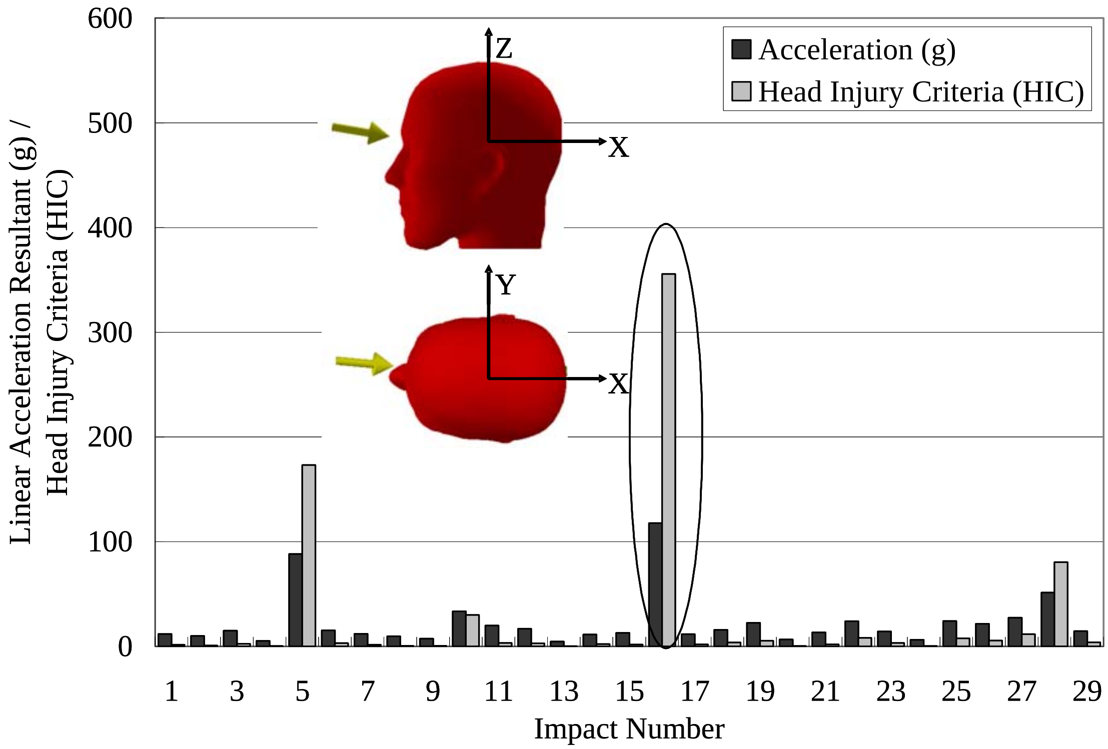


Figure 14: An impact during a full speed practice caused an injury for this player when an impact from another player resulted in a 117.6 g peak head acceleration.

Case 3

In this case, the impact occurred as the wide receiver was running with the ball and was tackled at high speed early in the first quarter. The striking player landed on the wide receiver's head as they reached the ground together. Consequently, the direction of the impact was in line with the sagittal plane on the front of the head 23 degrees above the transverse cg plane. Linear acceleration for this impact peaked at 136.7 g with a HIC value of 518.4 (Figure 15). The struck player had immediate knowledge of his concussion and came out of the game and was evaluated on the sideline by the medical staff per their usual concussion protocol. His symptoms cleared rapidly, within 15 minutes, and he had no recurrent symptoms with provocative testing. He ultimately returned to the game in the third quarter. This is consistent with return to play protocol for a Grade 1 concussion, which was the classification for this injury. Later in the game he sustained two additional high energy impacts, impact numbers 25 and 33. In the immediate post game evaluation he reported no further symptoms and had a normal neurological examination. At the time of this evaluation the medical staff was not informed of the data on these subsequent impacts. The following day, his symptoms returned and included headache, dizziness, mild nausea and confusion. A repeat clinical evaluation at that time resulted in a diagnosis of post-concussion syndrome. The contribution of these two additional impacts to his post-concussive syndrome is unknown but likely of clinical significance. The HeadMinder™ report indicated a severe decrease in the post injury cognitive status from the baseline test taken preseason. He was held out of competition for two weeks during which time his clinical symptoms resolved and his Headminder™ score returned to baseline. Prior to this injury he had a history of one concussion. For this player, only two impacts out of a total of 556 had a peak acceleration or HIC value greater than the reported concussive impact. From these two impacts only one was also to the front region of the head. He had no further clinical symptoms for the remainder of the competitive season.

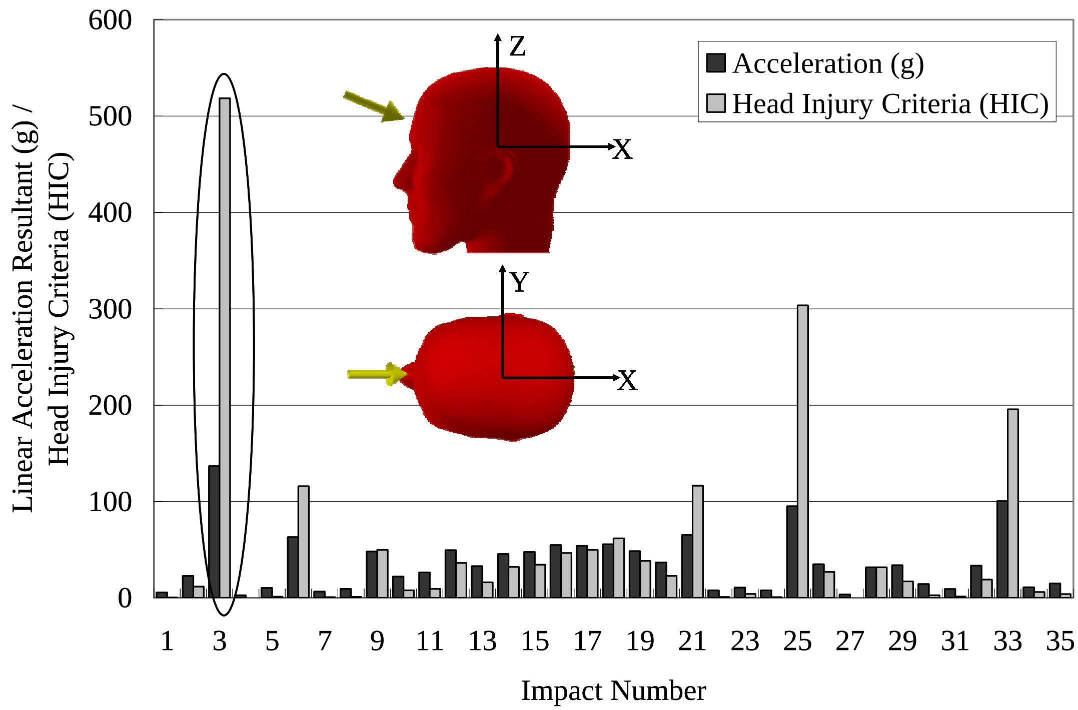


Figure 15: This player recognized having received a significant head impact and sought medical attention after receiving an impact that caused a peak linear acceleration of 136.7 g.

DISCUSSION

From these real-time head acceleration measurements a new data set has been established for an extensive number of impact situations that resulted in no brain injury. This research can lead to defining more specific brain injury thresholds for individuals as well as target high magnitude impact regions for protective equipment design. There has also been a clinical advantage of the HIT System because of the on field application with immediate notification of impact severity.

Previous Research

In this study the average linear acceleration was 20.1 g ±18.7 g for the 11601 impacts that did not result in a concussion. This is considerably lower than Pellman's (2003) average for non-injured players of 60 g ±24 g. However, this is expected given the current study's inclusion of all head impacts during the practices and games compared to Pellman's selection of more severe open-field impacts. The average data from the current study are more similar to the 29.2 g average head impact acceleration measured by Naunheim (2000) for high school football players which included all impacts and not just the most severe. Naunheim's (2000) study instrumented two football players for a total of 132 impacts.

Pellman (2003a) published 98 g ±28 g as the average peak linear head acceleration for concussive impacts.. This was data obtained from Hybrid III dummy reconstructions of concussive impacts from the NFL. While there were 348 impacts in the present study with a linear acceleration peak above 70 g, only two of these were concussions. Moreover, Pellman (2003) observed that the lowest average linear acceleration to cause a concussion was a 78.5 g impact to the facemask. Although the concussive events measured in this study generally agree with Pellman's data, the large number of non-injury impacts with high acceleration values does not. From the NFL data, King (2003) compiled an injury risk curve for linear acceleration as a predictor of mild traumatic brain injury. This curve predicts a much higher percentage of injury risk than was recorded at Virginia Tech. Using the NFL's level of a 75% risk of injury or greater, 85 impacts were recorded with two resulting in brain injuries (Figure 16).

One explanation for this disagreement may be the underreporting of concussions by players. Eight concussions were reported by the Virginia Tech football players throughout 2003 and 2004. A previous study by McCrea (2004) found that only 47.3% of high school football players reported their injury after sustaining a minor concussion, presumably Grade 1 or Grade 2. Moreover, other investigators have indicated that athletes often do not realize they have suffered a concussion. One recent study of collegiate football players utilizing self-reported symptoms of concussion indicated that only 23.4% of the players realized that their symptoms represented a concussion (Delaney 2002). As a result, these players would not seek medical attention and continued to play. This was precisely the type of behavior observed in two of the concussive events we recorded. It is therefore likely that some of the players in our current study may be represented in this unreported category.

According to Pellman (2003) the nominal tolerance of a concussion is HIC of 250. In our study, 123 impacts had HIC values above 250, and two of these impacts were clinically diagnosed concussions. The MTBI tolerance estimates defined by King (2003) suggest a 75% chance of MTBI for HIC of 333. These values were developed using logistic regression analyses on the Pellman (2003) data. The three concussions observed in the 2003 and 2004 seasons and recorded by the HIT system have a HIC of 120, 355, and 518 (Figure 17).

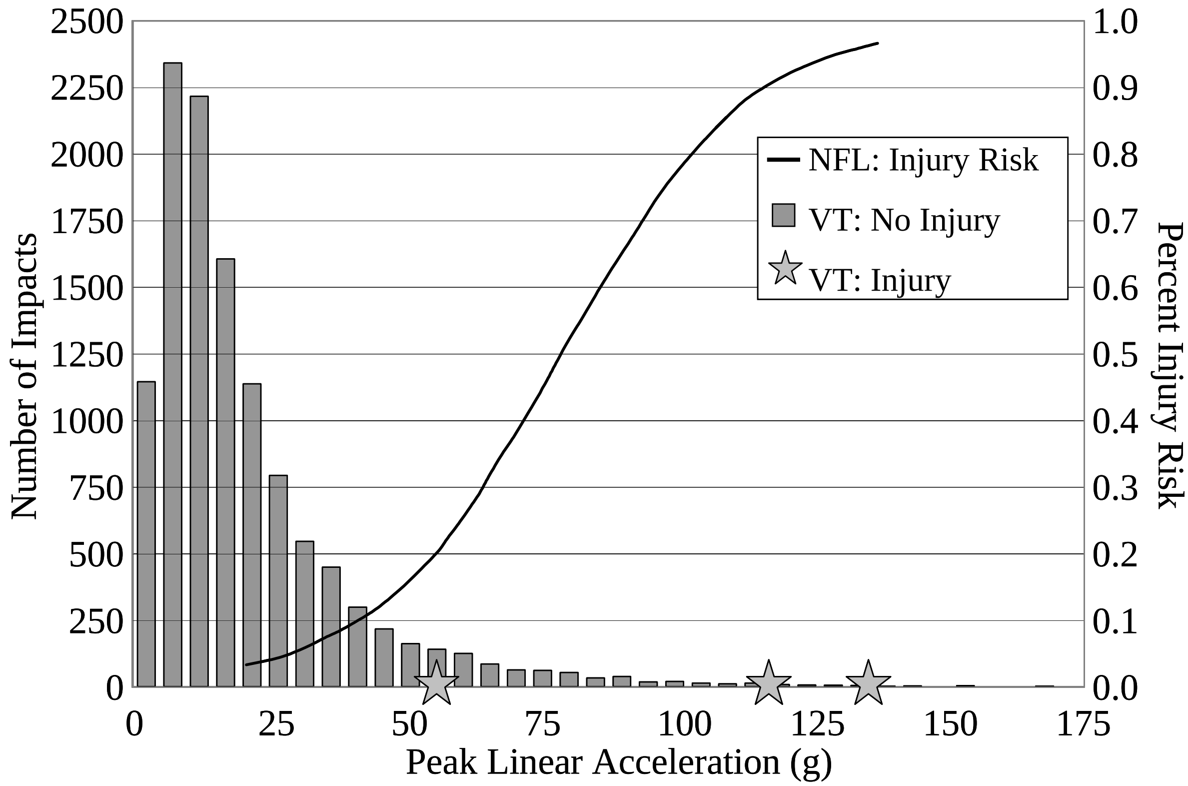


Figure 16: From the NFL data, King (2003) predicts a much higher risk of injury than was recorded at Virginia Tech.

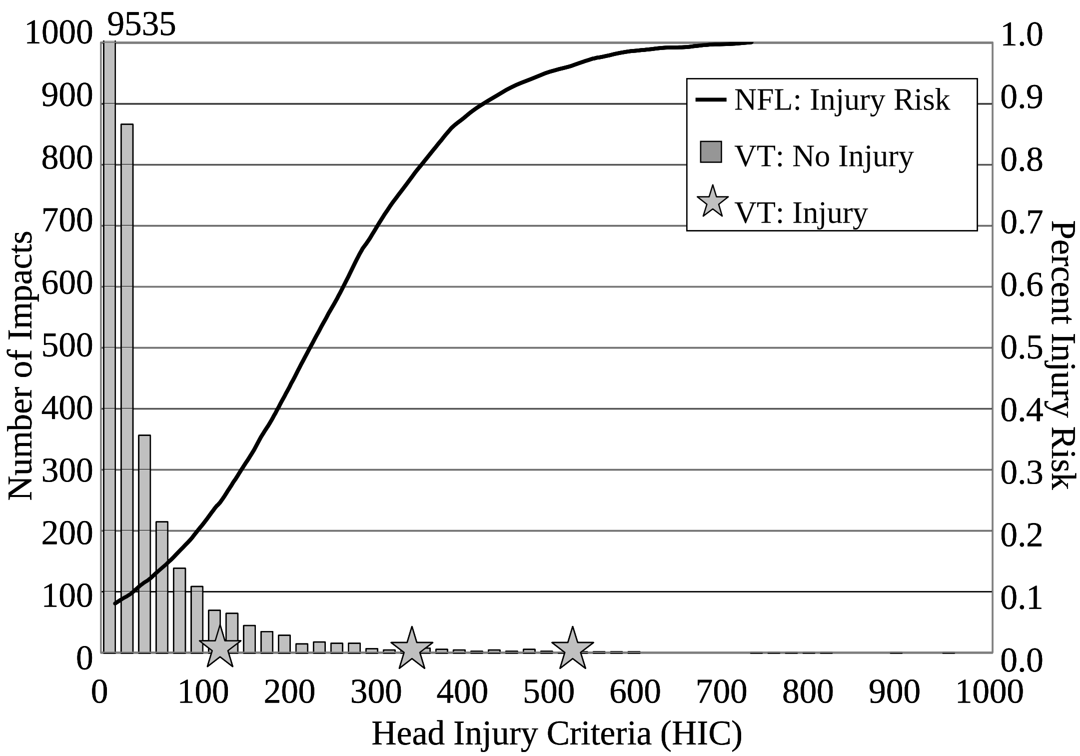


Figure 17: The MTBI tolerance estimates defined by King (2003) suggest a 75% chance of MTBI for a HIC of 333.

Limitations

Major challenges in measuring head acceleration during play include the need to place the sensors in contact with the head throughout the impact, to isolate the sensors from measuring the helmet deformation that occurs during impact, while keeping the system fully contained within the helmet for ease of use and to not affect the game in any way. The spring-loaded head mounted accelerometer system (HMAS) developed applies less than 1 lbf of pressure against the surface of the head, and remains in contact with the head throughout the 5-20 ms duration of a helmet impact. The data demonstrate that the system is recording head acceleration and not helmet acceleration, overcoming limitations of the earliest attempts in this area (Mecham, 1999).

A limitation of the current system is that it requires a properly fitted helmet to prevent decoupling between the head and the HMAS units. An early concern was that hair type and style might interfere with the coupling between head and HMAS. This has not been supported by the laboratory or field data, which demonstrate no significant effect on head acceleration measures as a function of hair type or style. The key element appears to be a well fitted helmet that ensures the accelerometers are in contact with the head during the impact. Although the spring mounted sensors stay in contact with the head in a correctly worn helmet, an improperly worn helmet could allow decoupling of the head and the accelerometers. This would result in spring or even helmet acceleration contributing to the measured head acceleration.

The error in the HIT System HIC measurement is attributed to higher accelerations recorded following the peak acceleration in the 10-15 ms period of the impact. It may be that the use of a 15 ms window is not appropriate for these data. More research on the use of HIC15 and its relevance to MTBI in football is being pursued. HIC values may be considered an approximation of impact severity and not a direct measurement of injury. Nonetheless, they are still being presented because of their uniqueness in the biomechanics field as real-time indicators of head impact severity.

Although there are several testing standards and headforms available, the validation tests were chosen to best represent on-field impacts that allow for cg measurement. Both the ASTM headform and the Hybrid III 50th percentile male headform were fitted with a helmet and tested. A pneumatic piston, drop system, and pendulum were used to create impact velocities. The impacting surfaces varied including a helmeted headform, a custom designed face, and a standard drop plate. Further discussion of these validations is presented in the Appendix.

Another limitation of this study is the participants studied. Collegiate football players are commonly exposed to intense play conditions with severe impacts. Current conditioning programs are designed to improve a player's ability to give and receive impacts without sustaining injury. Significant emphasis is placed on neck and shoulder strengthening for these individuals. If an individual could not tolerate this environment they would be unable to participate. Therefore the data presented in this study are collected from individuals who have chosen to expose themselves to a high impact environment and can endure it on a weekly basis. Their tolerance for injury may be higher than the general population as result of this sport's specific training process.

Relative Error

It is important to consider the relative error of the HIT System with other measurements used in biomechanical testing. Although the HIT system has a mean 0.01% error and it tracks linearly with validation data, the standard deviation of 18% presents more error (Figure 7). However, this error is on the same order of magnitude as other safety testing instruments such as the chestband and the Simulation Model Nonlinear (SIMON). In a dynamic evaluation of a chestband's ability to measure maximum intrusion the error reported was 10% below the actual value for dynamic tests (Rath, 2005). In a detailed analysis of the SIMON model, the path error range was 6.2% to 14.8% and a delta velocity error range was -1.6% to 7.5% (Day, 2004). A better comparison is the NFL impact reenactments which had 15% error. Moreover, their data were being validated to 30 Hz video while the HIT System data are being validated to 1000 Hz head cg accelerometers (Pellman, 2003a). According to the 11,604 impacts which recorded 40 ms at 1000 Hz, the event is much shorter than 33 ms. In summary, relative to other systems, and given the stringent design limitations of an in-helmet system that can be widely implemented, the HIT system measures linear acceleration reasonably accurately.

Rotational Acceleration

A predictor of brain injury, in addition to peak linear acceleration and HIC, is the peak rotational acceleration of the head. However, the placement of accelerometers is limited within the helmet to areas that do not compromise the protective nature of the helmet. As a result, only two of the three rotational accelerations can be reasonably approximated and these have not been thoroughly validated. This measurement in addition to the peak linear acceleration would provide more information in predicting brain injury. However, peak linear acceleration alone has been proven to be an indicator of injury. King (2003) measured linear and angular accelerations in his model for the condition of the headform wearing a helmet and not wearing a helmet. The linear acceleration was an average of 29% lower for the headform wearing a commercially available football helmet. On the contrary, rotational acceleration was not significantly different with the helmet compared to without the helmet. With the logistic regression chi-squared values he determined that HIC and linear acceleration are better predictors of head injury than angular acceleration. Future versions of the HIT System may include additional accelerometers in an effort to quantify the three rotational accelerations.

CONCLUSION

During two years at Virginia Tech the HIT System was used to collect 11,604 real-time head accelerations in collegiate football players. The results from these impacts are consistent with those previously measured from high school and NFL football impacts. However, the current study contributes a substantially larger number of game and practice impacts with hundreds of impacts for each of the 52 participants. Additionally, this study includes clinical evaluations as well as video recordings to supplement these measurements for a complete injury analysis. Validation of this system indicates the error in the peak linear accelerations reported is within reasonable experimental error. The three brain injuries reported had a peak linear head acceleration of 55.7 g, 136.7 g, and 117.6 g with HIC values of 120.6, 518.4, and 355.6 respectively. It is anticipated that through expansion of the number of players instrumented, and the number of seasons data is collected, more injury events will be observed and head acceleration data recorded. In addition, through continued use and expansion of this system, it is anticipated that on-field impact measurements will help improve head injury criteria and clinical evaluation techniques as well as enhance return to play decision making.

ACKNOWLEDGMENTS

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APPENDIX A: VALIDATION

Five series of tests including 130 impacts were completed to validate the algorithm and helmet application of the HIT System used at Virginia Tech. First, the general response of the algorithm in an ideal configuration was tested through computational and experimental methods. Since these tests showed the six accelerometer configuration to reasonably predict the cg accelerations, the system was fitted to a standard Riddell VSR-4 L helmet. Second, testing was completed using a pneumatic linear impactor to ensure the spring mounted accelerometers produced desired results on the Hybrid III headform (Figure 5). Third, a pendulum impactor was used to simulate helmet to helmet low speed impacts and measure helmet accelerations. Fourth, an ASTM impact system was used to collect impacts at different locations on a standard test rig designed to fit a large football helmet. Lastly, high velocity tests were done with a pendulum to test several impact locations to further validate the acceleration and impact location measured by the HIT System. These test methods and results are presented to validate the data presented in this paper.

Validation 1: Hemisphere Impacts and Computations

The novel algorithm used in the HIT System was developed specifically to calculate linear acceleration and impact location using multiple singe-axis accelerometers (Crisco, 2004). Unlike previous linear acceleration calculations these accelerometers are not constrained to be orthogonal to each other but rather simply normal to the object's surface. Experimental and computational tests are presented for the 11 HMAS configuration noting that I indicate the 6 HMAS configuration, used in the HIT System, has similar values. For the experimental tests, a hemisphere with embedded accelerometers was used on an ASTM drop fixture to test the algorithm (Figure A. 1). The test matrix included 45 degree increments in azimuth and 20 degree increments in elevation to test 33 possible impact locations. The error of the embedded accelerometers resolved using the algorithm compared to the triaxial accelerometer cube at the origin was less than ±10% for peak acceleration at each test location. HIC is solely dependent on acceleration and time. Therefore, because this algorithm can precisely determine the cg acceleration over 40 ms, we can assume the HIC measured by the embedded accelerometers on the surface is the same as the HIC determined from the tri-axial cube at the cg (Figure A. 2). The error in measured impact location was less than ±10 degrees for azimuth and less than ±15 degrees for elevation. In addition to experimental tests, computational analysis of the algorithm was also performed. Resultant head cg acceleration and impact location were estimated for a test matrix of specific impact locations and impact magnitudes equal to 100 g. The impact magnitude error had a mean equal to zero with standard deviation less than 10%. The impact location error had a mean equal to zero with standard deviation less than 10 degrees for the azimuth and less than 6 degrees for the elevation. Thus, the experimental error and computational error from the algorithm agree in predicting the impact magnitude and location azimuth within 10% and 10 degrees respectively. Experimentally the algorithm was within 15 degrees of predicting the location elevation. A limitation of this procedure is that the object tested is a hemisphere and the human head is not. To apply the algorithm to a non-spherical shape, an additional term in the algorithm was mathematically determined to adjust for the sensing axis not being the surface normal.

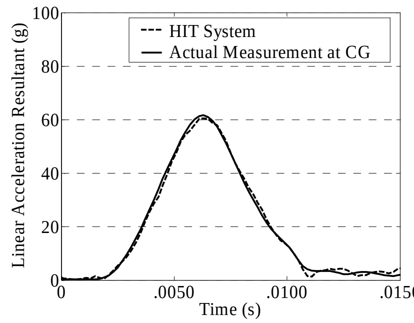


Figure A. 1: A solid hemisphere with embedded accelerometers was impacted to verify the algorithm.

Figure A. 2: The six accelerometers calculated the same acceleration as the actual cg for the duration of the impact.

Validation 2: Pneumatic Impactor Tests

Validation testing was performed at Wayne State University to test the HIT System's ability to measure headform cg acceleration when the 6 HMAS were installed in a Riddell VSR-4 Large football helmet. A standard triaxial accelerometer cube mounted in a HIII headform measured the cg acceleration while a pneumatically driven linear ram impactor made impacts against the helmeted HIII at predetermined locations (Figure A. 3). The impacting face was a rigid metal with a 4 mm polycarbonate outer surface, spherical in shape with a radius of 127 mm, covered with foam padding and a thin polycarbonate shell designed to simulate a striking player's helmet. The impactor mass was 11.3 kg, representing the effective mass of a striking player as determined from actual video recordings of impacts (Pellman, 2003). Impactor velocity was 6 m/s ±0.1 m/s. The HIII upper body was mounted to a hydraulic lift table that not only permitted vertical positioning but was also part of a trolley system to allow for linear motion after impact. The three tested impact locations included rear oblique, lateral, and lateral above the cg plane. The maximum linear acceleration ranged from 60 g to 80 g in these 5 tests. The ratio of the measured HIT System peak acceleration to the actual HIII peak acceleration was 0.89 (R^2=0.98)

(Figure A. 4 and Figure A. **5**). The head injury criteria measured by the HIT System versus the HIC calculated from the HIII cg had a linear slope of 0.95 (R^2=0.66) (Figure A. 6). Limitation was that only 5 impacts were performed and only a one impact velocity.

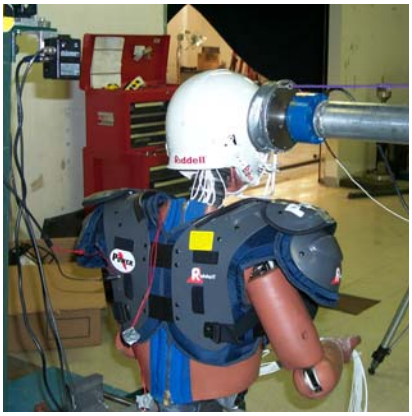


Figure A. 3: A pneumatically driven linear ram impactor made impacts against the helmeted HIII.

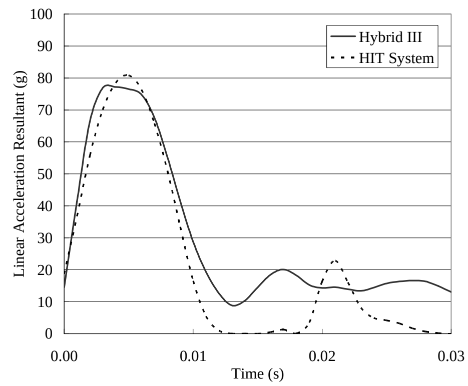


Figure A. 4: The HIT System in the helmet calculated the same acceleration as the actual cg for the duration of the impact.

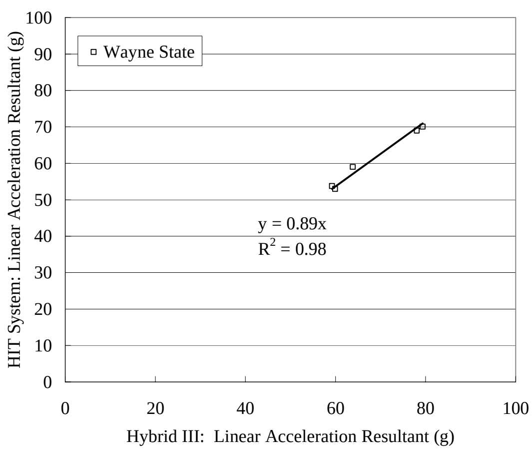


Figure A. 5: The peak acceleration of the HIT System correlated to the peak acceleration at the Hybrid III cg with a linear slope of 0.89 (R^2=0.98).

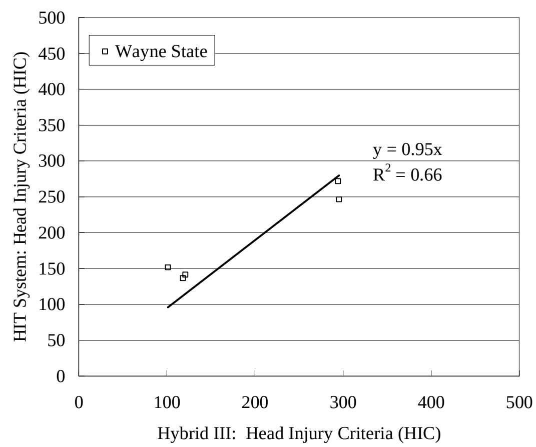


Figure A. 6: The calculated HIC from the HIT System correlated to the calculated HIC from the Hybrid III cg with a linear slope of 0.95 (R^2=0.66).

Validation 3: Helmet to Helmet Pendulum Impacts

More validation tests were done at the Virginia Tech Center for Injury Biomechanics laboratory to test helmet to helmet impacts as well as evaluate helmet versus head acceleration. A custom designed pendulum was set up with a HII head and neck attached at the end so the impact to the striking helmet occurred in the sagittal plane on the front of the helmet above the facemask. A full HIII was placed in a wheel chair and instrumented with the HIT System in a Riddell VSR-4 L helmet and a triaxial accelerometer cube at the head cg (Figure A. 7). The test matrix included three impact velocities of 2.0 m/s, 3.5 m/s and 5.0 m/s at four different locations. The impact locations were on the side, back, top and just above the facemask on the front. For these 32 tests, the ratio of the HIT System peak acceleration measured value to the actual HIII peak cg acceleration was 1.07 (R^2=0.81) (Figure A. 8).

Figure A. 7: The HIT System in the helmet calculated the same acceleration as the actual cg for the duration of the impact.

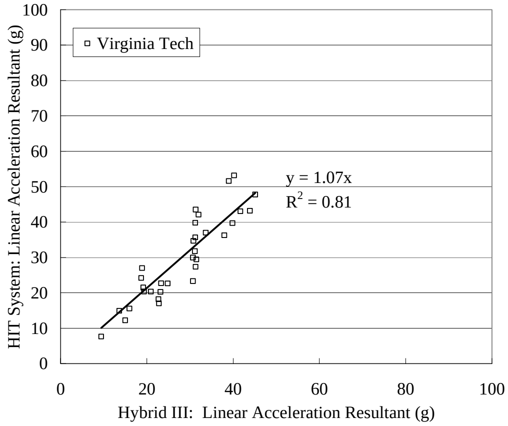


Figure A. 8: The peak acceleration of the HIT System correlated to the peak acceleration at the Hybrid III cg with a linear slope of 1.07 (R^2=0.81).

Additional accelerometers were placed on the inside of the helmet shell at the point of contact to measure helmet accelerations. Overall the peak acceleration for the helmet at the point of contact was 16.6 (±3.2) times greater than the head cg peak linear acceleration for that impact. Additionally, the time from contact to peak head acceleration, 5.82ms (±0.78ms), is longer than time to peak helmet acceleration, 0.42ms (±0.17 ms). The peak magnitude, time to peak, and waveform shape of the HIT System measure match the head cg acceleration values and not the helmet shell acceleration values (Figure A. 9).

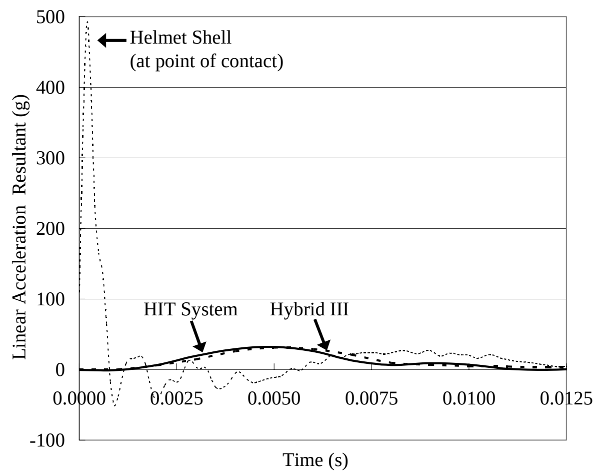


Figure A. 9: The HIT System measures peak magnitude, time to peak, and waveform shape which match the head cg acceleration values and not the helmet shell acceleration values.

Validation 4: Helmet Drop Impacts

Tests were completed using the standard ASTM headform and drop system. The ASTM procedure has been a leader in the development of helmet standards. This headform is designed to fit and test a helmet. The Player Unit was installed in a Riddell VSR-4 L helmet and placed on a size O headform which was instrumented with a triaxial accelerometer cube to measure resultant cg acceleration. The helmet and headform were impacted by using the ASTM twin wire drop test system. Various impact locations were all tested at a target cg acceleration of 50

g. From these 16 impacts the ratio of measured cg peak acceleration to actual cg peak acceleration was 0.8 (R^2=0.78) (Figure A. 11).

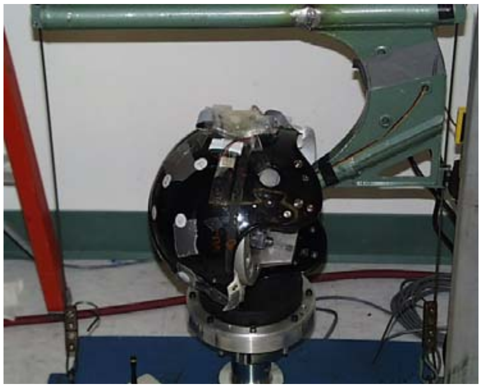


Figure A. 10: The ASTM drop system was used to impact the helmeted ASTM headform at several locations.

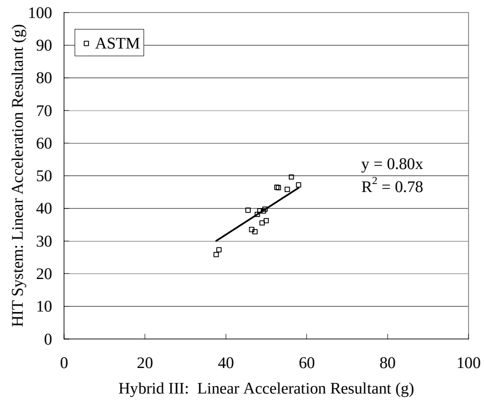


Figure A. 11: The peak acceleration of the HIT System correlated to the peak acceleration at the ASTM cg with a linear slope of 0.80 (R^2=0.78).

<u>Validation 5: Impactor to Helmet Pendulum Impacts</u>

Validation tests were performed using a Helmet Impact Pendulum (Biokinetics) to compare the HIT System calculated max linear acceleration, head injury criteria, and location of impact for several impacts at different speeds and locations. The Helmet Impact Pendulum system uses a Hybrid III 50th percentile neck and head mounted to a frame which is adjustable for an impact at any location on the helmet (Figure A. 12). The impactor face used for these tests was constructed of a UHWPE shell with VN600 Dertex foam, VSR4 liner foam, behind it. This impactor face was tested at Biokinetics to best represent helmet to helmet impacts. Because a Riddell VSR-4 L helmet is not designed to fit a HIII headform, 1” jaw pads were used in the helmet to have a secure fit. Before each impact the helmet was aligned on the headform and the chinstraps were tightened to represent a correctly worn helmet. The maximum pendulum height correlated to a speed of 7.0 m/s. Two additional heights were used for 3.0 m/s and 5.0 m/s. Six impact locations were designated as high frequency impact regions by Biokinetics based on Pellman's previous research. Impact location A (-168,-13) is the facemask region just below the cg, B (-132,1) is the region of the facemask close to where it meets the helmet shell and almost at the level of the cg, C (-57,-4) is just posterior of the ear and slightly below the cg, D (-25,26) is above is the cg and off center from the sagittal plane on the back of the head, E (1,36) is in the sagittal plane on the back of the head above the cg, F (-179,42) is above the facemask in the sagittal plane on the front of the head (Figure A. 13). The test matrix, a total of 44 tests, did not include high impact velocities for the facemask region because those impacts would damage the facemask and resources and time were limited.

Figure A. 12: The Helmet Impact Pendulum was used to impact the helmeted Hybrid III head and neck at several locations and velocities.

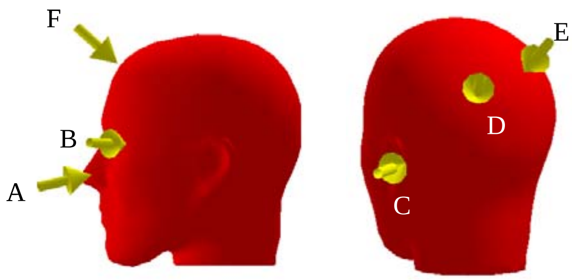


Figure A. 13: Six locations were tested with the pendulum to helmet test series.

The mounting table was set for the correct location of the impact and the point where the pendulum hit was marked and used as the target location for that impact. When this location was compared to the calculated HIT System azimuth and elevation, the average azimuth error was 10°±9°and the elevation error was 21°±15°. The peak linear resultant acceleration and the head injury criteria calculated from the 40ms time trace were evaluated by comparing the HIII accelerometer array to the HIT System values (Figure A. 14and Figure A. 15). For the linear acceleration, a linear fit indicates the HIT System measured the peak as 6% greater than the actual HIII peak (R^2=0.83) (Figure A. 16). The head injury criteria calculated from the HIT System was 22% greater than the value calculated from the HIII data (R^2=0.81) (Figure A. 17).

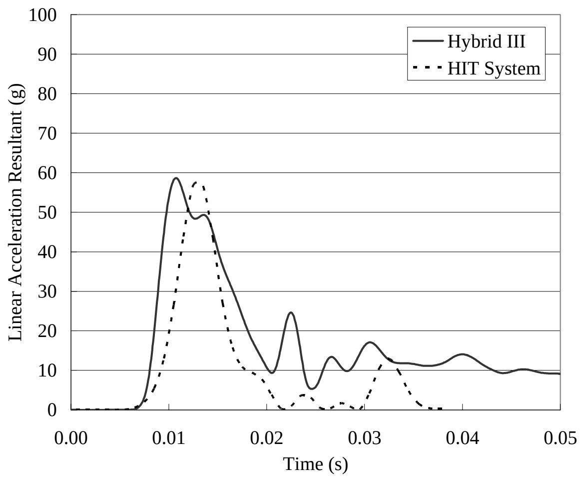


Figure A. 14: The HIT System's measured acceleration followed closely to the actual cg for the duration of the impact.

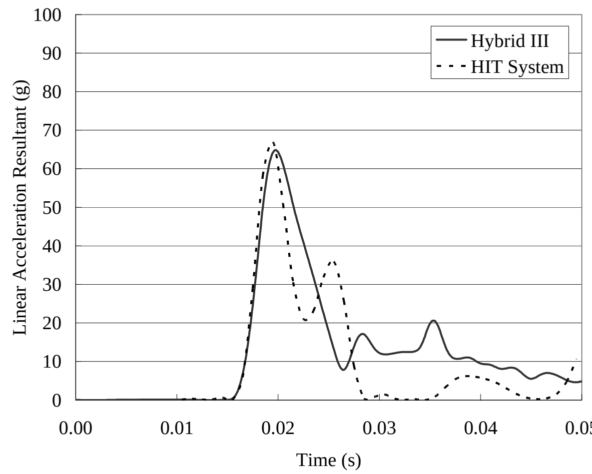


Figure A. 15: The HIT System's measured acceleration followed closely to the actual cg for the duration of the impact.

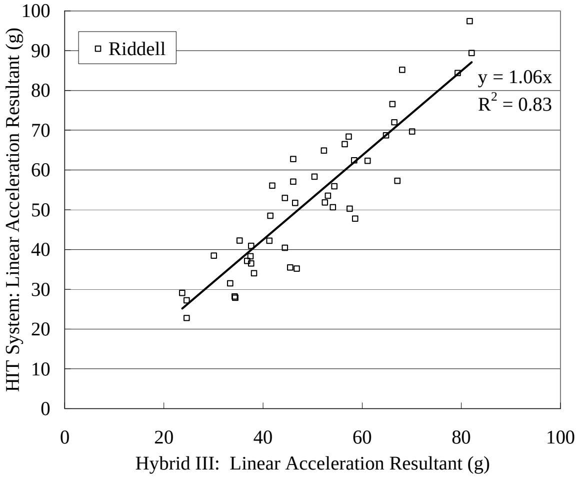


Figure A. 16: The peak acceleration of the HIT System correlated to the peak acceleration at the Hybrid III cg with a linear slope of 1.06 (R^2=0.83).

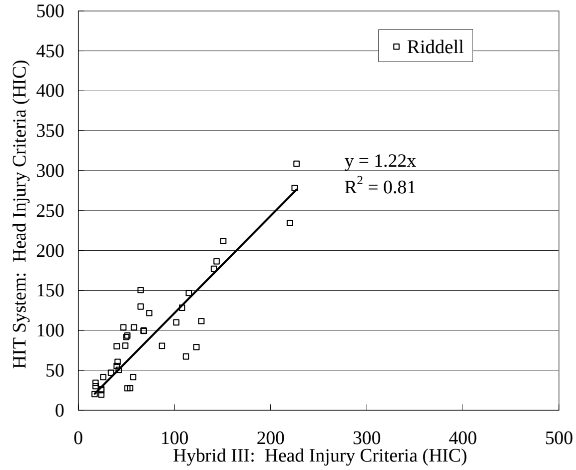


Figure A. 17: The calculated HIC from the HIT System correlated to the calculated HIC from the Hybrid III cg with a linear slope of 1.22 (R^2=0.81).

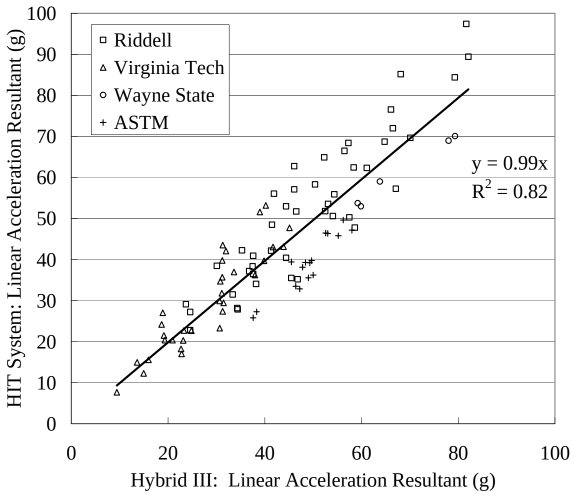
Summary

In conclusion, a variety of experimental test setups were used to validate the HIT System method of measuring head cg acceleration. The simplest case of a solid object with embedded accelerometers experimentally showed the precision of the algorithm predicting acceleration magnitude and impact location. The system was then tested in a football helmet application. The results are evidence that with either a pendulum or linear impactor striking the helmet the HIT System would closely estimate the peak acceleration and the 40 ms acceleration trace of the head cg. Furthermore, the HIT System and the accelerometers in the HIII head measure a similar head acceleration response which varies greatly in magnitude and shape from the helmet response.

Although there are several testing standards and headforms available, these tests best represent on-field impacts that allow for cg measurement. The headform used in certifying helmets is the NOCSAE headform. The anthropometric accuracy of this headform makes it a better fit for a helmet than a HIII headform. However, the neck is a rigid attachment so it is not designed to be attached to a surrogate body or model a human neck response. Additionally, each NOCSAE headform must be calibrated since the nature of the design reduces its repeatability. In combination with the NOCSAE headform and testing procedure, the ASTM procedure has led the field for helmet standards development. The ASTM testing procedure was used for one test series to validate the system's capabilities in measuring cg peak acceleration. Providing a more thorough evaluation, the Hybrid III 50th percentile male headform was chosen for other three tests presented. This surrogate can model humanlike head and neck response. Using the Hybrid III also allowed for full upper body and full body testing of impact response. The headform has a designated cg where a triaxial accelerometer can easily be mounted. Because this headform is not designed for a helmet, the tests used larger jaw pads to fill the space around the jaw and ensure a secure helmet fit. All of these configurations were accurately setup with previously tested impactor faces, a secure helmet fit, and realistic impactor velocities.

All 130 validation tests have been plotted together to establish an overall range of accuracy for the HIT system in measuring head cg peak acceleration, head injury criteria, and impact location. The mean error in measuring peak head cg linear acceleration is 0.01% ± 18% (Figure A. 18). The mean error in measuring HIC is 30% ±43% (Figure A. 19). For measuring the location, the azimuth error was 10°±9°and the elevation error was 21°±15°. While these values are not 100%

accurate, I submit that an 18% error is acceptable given standard experimental error. However, the 30% error suggests HIC values should be more carefully considered.



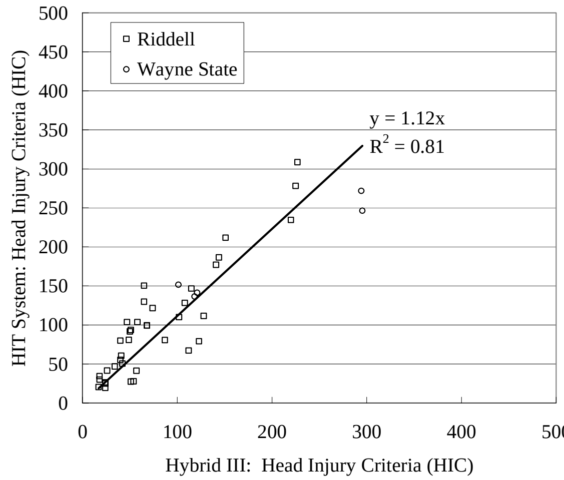


Figure A. 18: Overall the peak acceleration of the HIT System correlated to the peak acceleration at the Hybrid III cg with a linear slope of 0.99 (R^2=0.82).

Figure A. 19: Overall the calculated HIC from the HIT System correlated to the calculated HIC from the Hybrid III cg with a linear slope of 1.12 (R^2=0.81).