ORIGINAL RESEARCH

Analysis of Real-time Head Accelerations in Collegiate Football Players

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Objective: To measure and analyze head accelerations during American collegiate football practices and games.

Methods: A newly developed in-helmet 6-accelerometer system that transmits data via radio frequency to a sideline receiver and laptop computer system was implemented. From the data transfer of these accelerometer traces, the sideline staff has real-time data including the head acceleration, the head injury criteria value, the severity index value, and the impact location. Data are presented for instrumented players for the entire 2003 football season, including practices and games.

Setting: American collegiate football.

Subjects: Thirty-eight players from Virginia Tech's varsity football team.

Main Outcome Measurements: Accelerations and pathomechanics of head impacts.

Results: A total of 3312 impacts were recorded over 35 practices and 10 games for 38 players. The average peak head acceleration, Gadd Severity Index, and Head Injury Criteria were 32 g ± 25 g, 36 g ± 91 g, and 26 g ± 64 g, respectively. One concussive event was observed with a peak acceleration of 81 g, a 267 Gadd Severity Index, and 200 Head Injury Criteria. Because the concussion was not reported until the day after of the event, a retrospective diagnosis based on his history and clinical evaluation suggested a mild concussion.

Conclusions: The primary finding of this study is that the helmet-mounted accelerometer system proved effective at collecting thousands of head impact events and providing contemporaneous head impact parameters that can be integrated with existing clinical evaluation techniques.

Key Words: acceleration, helmet, football, concussion, head

(*Clin J Sport Med* 2005;15:3–8)

Brain injuries occur in 1.5 million people in the United States each year, of which 75% are mild traumatic brain injuries (MTBIs) like concussions.[1] Sports-related concussions constitute 300,000 of these injuries.[2] In the 2002 to 2003 college football season, 8% of all injuries were concussions.[3] Football has the most total concussions of any sport and has had an increasing rate of injury in the last 7 years.[2]

Concussion research has been advancing for years, leading to a greater insight of the cause and effect of a MTBI.[4,5] In the early 1970s, Moon et al[6] and Reid et al[7,8] instrumented the headbands of suspension-style football helmets with a frequency modulation-based accelerometer and electroencephalogram system. Morrison[9] used a similar system in the early 1980s at Penn State, although without the electroencephalogram capabilities. 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. In 2000, Naunheim et al[10] instrumented hockey and football helmets with a padding-embedded triaxial accelerometer. The average peak acceleration measured for the 158 football helmets 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 used a combination of video analysis and dummy re-enactments of impacts from game film. Newman et al[11–13] and Pellman et al[14] 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 2 or more different angles. They used this video data to reconstruct the angle of the impact, the 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 (Denton ATD, Inc, Milan, OH) in the laboratory. While this study reported linear risk curves and rotational risk curves, these findings were limited to specific impacts with NFL players, and the methods are not widely applicable.

Received for publication May 2004; accepted September 2004.
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The Head Impact Telemetry System development was funded in part by NIH R43HD40743. Additional support for this project was provided by the Edward Via College of Osteopathic Medicine, the Virginia Tech College of Engineering and the Department of Sports Medicine, and Simbex.
Richard M. Greenwald, PhD, is president of Simbex LLC, and has a commercial interest in the technology used in this research study.
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The dependence on video reconstruction and dummy reenactments creates an indirect measure of head acceleration, and it prevents the data from being used clinically on the field. Despite the effort of previous investigators, this research has not given 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 article is to present an analysis of all head impact acceleration data collected from a full season of 1 American collegiate football team.

METHODS

This study used the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH), 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 (Fig. 1) designed to fit a Riddell VSR-4 L or XL (Fig. 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 sensors (6 linear accelerometers and 1 temperature), a wireless transceiver (903–927 MHz), on-board memory (up to 33 impacts), and data acquisition capabilities (8 bit; 1000 Hz/channel). Data are collected for 40 ms when any single accelerometer detects an acceleration that exceeds a user-selected threshold (10 g). Since the trigger is from 1 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 is stored pretrigger with 28 ms posttrigger for a total of 40 ms. The acceleration data are time-stamped (±5 ms resolution) and wirelessly transmitted to a sideline controller interfaced to a laptop. 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 33 impacts on nonvolatile memory and is downloaded once communication has been re-established.

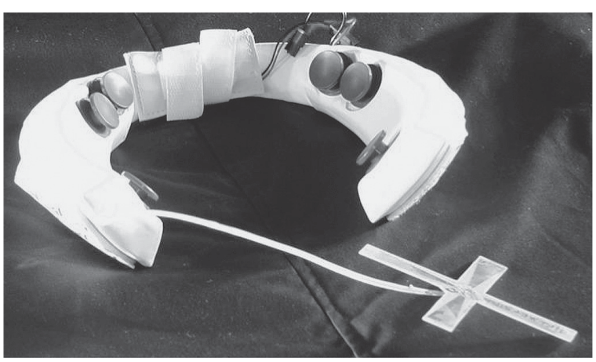

FIGURE 1. The player unit, consisting of 6 accelerometers in spring-loaded holders, frequency modulation antenna, and rechargeable battery pack.

FIGURE 2. Virginia Tech helmet with the player unit installed.

The impact data from the 6 accelerometers are processed using a novel algorithm that calculates head acceleration and impact location.[15] Using this algorithm, the peak linear head center of gravity time history, impact location, Gadd Severity Index (GSI), Head Injury Criteria (HIC), and sagittal and lateral peak rotational accelerations were calculated. GSI is the integral of acceleration with respect to time and therefore calculates a value for comparison of pulses with various durations.[5] An equation using the integral of acceleration with respect to a given amount of time, this study uses 36 ms and is calculated over the entire duration of a pulse, and the maximum value is the HIC for that impact.[16] 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 center of gravity of the head.

The HIT System was validated using a series of impact tests with a helmet-equipped Hybrid III dummy instrumented with a 3-2-2-2 head accelerometer array.[16a,17] The HIT System correlated well with the 3-2-2-2 data ($R^2 = 0.97$) and had a ±4% error for linear and rotational accelerations as well as HIC scores. In addition, as part of the validation process, linear drop tests were performed using a twin wire drop tower with a National Operating Committee on Standards for Athletic Equipment (NOCSAE) instrumented headform. An instrumented helmet was placed on the headform and impacted at locations ranging from −180° to 180° azimuth in 45° increments and at elevations ranging from 0° to 90° in 30° increments. Multiple drops were performed at each impact location. The estimated impact location and magnitude were repeatable at each impact location to within an average of 2.45° or approximately ±0.41 cm based on the average head radius of a NOCSAE headform. Accuracy was evaluated using marked impact locations and a 3-dimensional digitizer. The average impact location error for both azimuth and elevation was ±1.20 cm. Finally, the Riddell VSR4 helmets with the HIT System player unit installed were tested at a third-party laboratory and passed the NOCSAE football helmet performance standard required for all helmets worn at the collegiate level.

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. The 6 single accelerometer traces and analysis are presented to the sideline unit controller on the laptop screen in 4 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 (Fig. 3). 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 to provide insight into the injury mechanisms for specific impacts. Although the practices are not typically as intense as a game setting, the practices do allow for additional impact events to be collected.

During each practice and game, up to 8 players were instrumented and monitored simultaneously for the 2003 season. Players were selected each week by the medical staff to provide a representative cross-section of all football players' body types and positions. All players selected gave written informed consent with Institutional Review Board approval from both Virginia Tech and the Edward Via Virginia College of Osteopathic Medicine. Each player was instrumented for approximately a 2-week period that typically consisted of 6 practices and 2 games.

To analyze the mechanical data in conjunction with clinical aspects of concussion, the sports medicine staff maintains detailed player histories, including preseason neuropsychological records. HeadMinder (HeadMinder, Inc, New York, NY), a Web-based assessment program, is used preseason and postinjury to compare a player's baseline cognitive status with his postinjury evaluation. Following an injury, clinical indicators are closely observed and monitored by the medical staff. Clinical indicators include postural sway, confusion, retrograde amnesia, antegrade amnesia, and headache. When an injury initially appears more serious or symptoms remain persistent, more elaborate neuropsychological tests and neuroimaging procedures are performed as clinically indicated.

RESULTS

For the 2003 season, a total of 3312 valid head impacts were recorded, with 1198 occurring in 10 games and 2114 in 35 practices. The impacts were measured from 38 different players covering the majority of offensive and defensive positions.

For all recorded impacts, the average peak head acceleration was 32 g $\pm$ 25 g (range, 1–200 g). The majority, 89%, of the impacts were less than 60 g in a nonnormal distribution (Fig. 4). The average GSI was 36 $\pm$ 91 (range, 1–1599), and the average HIC was 26 $\pm$ 64 (range, 1–956; Table 1). Although more than 83% or 2749 of the impacts had a GSI and HIC value below 50, 9% or 298 impacts had a GSI value above 100, and 5% or 166 impacts had a HIC value over 100.

Rotational acceleration about the x-axis and y-axis displayed a similarly left-skewed distribution as the linear acceleration data (Fig. 5). Overall, there were a greater number of larger y-axis rotational accelerations versus the x-axis rotational accelerations. The mean rotations were 905 rad/s^2 $\pm$ 1075 rad/s^2 (range, 1–11,348 rad/s^2) and 2020 rad/s^2 $\pm$ 2042 rad/s^2 (range, 1–18,477 rad/s^2) about the x-axis and y-axis, respectively.

Impact locations varied but were also clustered into areas with high density of impacts that appeared to depend upon the separate field positions as well as individual player technique (Fig. 6). A single game's data display these observations. The receiver exhibited a line of impacts in the sagittal

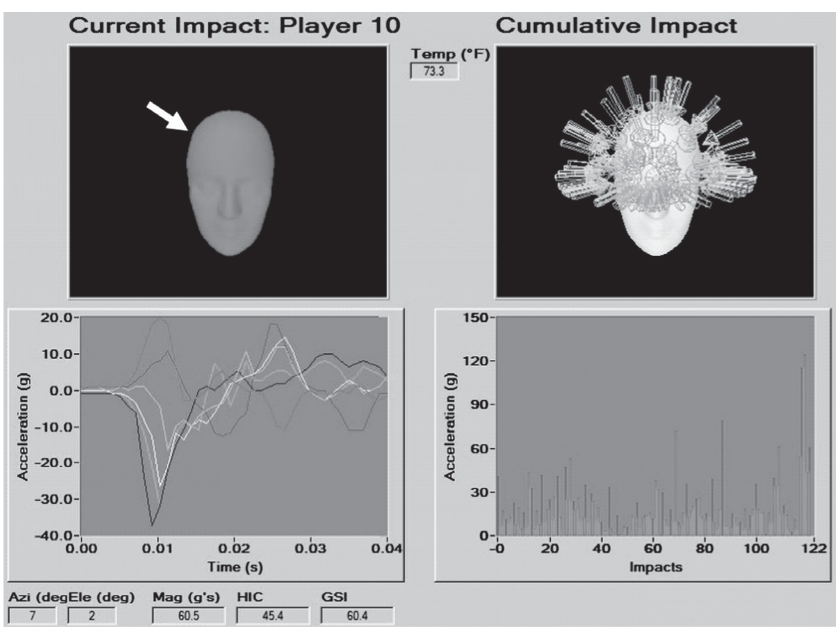

FIGURE 3. Sideline controller laptop screen. A, A directional vector indicates the most recent hit location. B, Directional vectors for 1 athlete exposure (ie, 1 game or practice). C, Six accelerometer traces are plotted versus time for 40 ms. D, Acceleration magnitude history is presented in bar chart format.

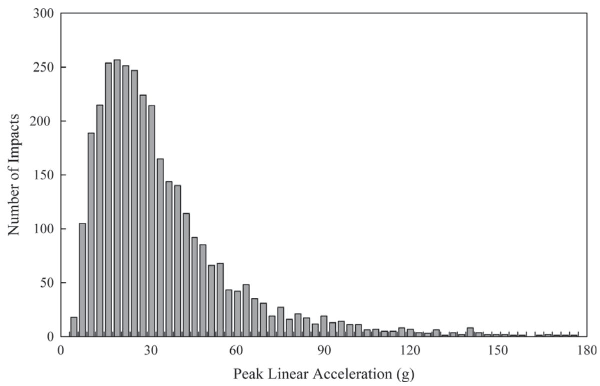

FIGURE 4. Distribution of peak linear accelerations of the head center of gravity for all instrumented players from the 2003 season (1 hit not shown at 200 g).

plane with very few lateral impacts. Conversely, the fullback exhibited a lateral band of impacts circling the equator of the head, with a second concentration of impacts at the forehead. While displaying impact distributions similar to a fullback, the linebacker and defensive lineman had more impacts to the rear and upper halves of the helmet.

Five concussions were sustained on the team during the 2003 season, with 1 of these in an instrumented player. A more specific analysis of the data for the 1 instrumented player who sustained a concussion revealed that he sustained a mild concussion as well as a neck injury from 2 of the 33 impacts in 1 game (Fig. 7). The concussion occurred during his second impact after kickoff, and it had a value of 81 g, 267 GSI, 200 HIC, 5600 rad/s^2 x-axis rotation, and 5590 rad/s^2 y-axis rotation. The location of the impact was on the right side, 164° from the middle of the back of the head, in the facemask region and in plane with the center of gravity of the head (Fig. 8A). This impact occurred as the opposing team's player struck this player straight on, causing him to fall backward and hit the ground. Although clinical symptoms were later discovered, he did not experience loss of consciousness, and he stayed in the game. When reviewing the game film with his coach, it was revealed that immediately following this impact, the player could not remember the plays or where he should set up on the field. 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.[18] Later in the game, he received a neck injury as a result of the 16th impact of the game: 86 g, 165 HIC, 200 GSI, 4132 rad/s^2 x-axis rotation, and 6638 rad/s^2 y-axis rotation. This hit location was 31° to the left of the middle of the back of the head and 11° above the center of gravity plane (Fig. 8C). Because of the physical discomfort, he was immediately aware of this injury and sought attention from the medical staff.

DISCUSSION

The purpose of this study was to measure and analyze head accelerations during American collegiate football practices and games. Through the 2003 season, the HIT System proved effective at collecting thousands of head impact events and providing real-time on-field data analysis. A total of 3312 impacts were collected from 38 players on 35 practice and 10 game days. An immediate benefit was that the impact severity and location were available on the field. As the season progressed, acceleration distributions were available for all impacts monitored during practices and games.

In this study, the average linear acceleration was 32 g ± 25 g of the 3311 impacts that did not result in a concussion. This is considerably lower than the average by Pellman et al[14] for noninjured 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 with the selection by Pellman et al[14] of more severe open-field impacts. The average data from the current study are very similar to the 29.2 g average head impact acceleration measured by Naunheim et al[10] for high school football players, which included all impacts and not just the most severe. The study by Naunheim et al[10] instrumented 2 football players for a total of 132 impacts.

For concussive impacts, the peak linear head acceleration average Pellman et al[14] published was 98 g ± 28 g. These were the average data obtained from Hybrid III dummy reconstructions of concussive impacts from the NFL. While there were 583 impacts in the present study with a linear acceleration peak above 70 g, the 1 concussion observed falls in the range of Pellman et al[14] at 81 g. Moreover, Pellman et al[19]

TABLE 1. Frequency of Peak Linear Acceleration, GSI, and HIC Values for All Instrumented Players from the 2003 Season

Range		0–30	30–60	60–90	90–120	120–150	150–180	180–210	>210
Linear acceleration (g)	Number of impacts	1974	959	245	87	36	10	1	0
	Percent of total impacts	59.6%	29.0%	7.4%	2.6%	1.1%	0.3%	0.0%	0.0%
GSI	Number of impacts	2504	357	144	70	50	41	26	120
	Percent of total impacts	75.6%	10.8%	4.3%	2.1%	1.5%	1.2%	0.8%	3.6%
HIC	Number of impacts	2710	278	105	61	41	28	16	73
	Percent of total impacts	81.8%	8.4%	3.2%	1.8%	1.2%	0.8%	0.5%	2.2%

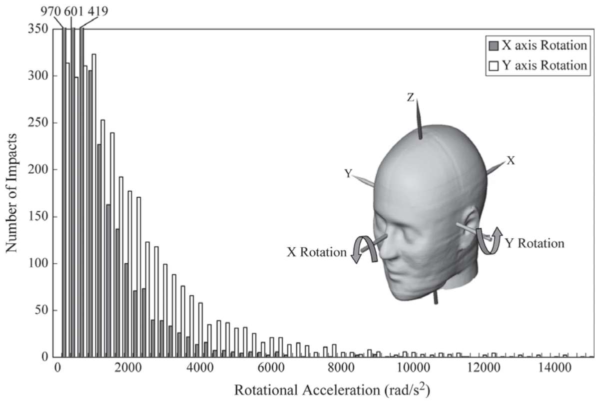

FIGURE 5. The x-axis and y-axis peak rotational acceleration values for all instrumented players from the 2003 season (1 y-axis rotation value of 18,477 rad/s^2 not shown).

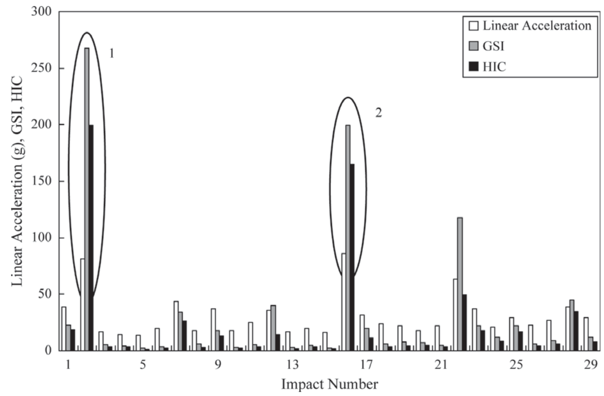

FIGURE 7. One player's head impacts for 1 game in which he sustained a concussion (1) and stayed in the game, and a neck stinger (2) for which he came out briefly before playing the remainder of the game.

observed that the lowest average linear acceleration to cause a concussion was a 78.5-g impact to the facemask. Although the 1 concussive event measured in this study agrees with the data by Pellman et al,[19] the large number of noninjury impacts with high acceleration values do not. According to Pellman et al,[14] the nominal tolerance of a concussion is GSI = 300 and HIC = 250. In our study, 71 impacts had a GSI value above this tolerance, and 55 impacts had HIC values above 250, but we observed no reported concussions for these specific impacts.

The MTBI tolerance estimates defined by King et al[20] suggest a 75% chance of MTBI for HIC of 333, linear acceleration of 98 g, and angular acceleration resultant of 7130 rad/s^2. These values were developed using logistic regression analyses on the data by Pellman et al.[14] A total of 25 impacts from all of the data collected for the 2003 season meet all 3 of these criteria; however, none of these impacts resulted in an observed concussion. The 1 concussion observed in the 2003 season and recorded by the HIT system had a HIC of 200, linear acceleration of 81 g, and 2-dimensional angular acceleration resultant of 7912 rad/s^2.

One explanation for this disagreement may be the underreporting of concussions by players. A total of 5 concussions were reported by the Virginia Tech football players throughout the 2003 season. A previous study by McCrea et al[21] 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

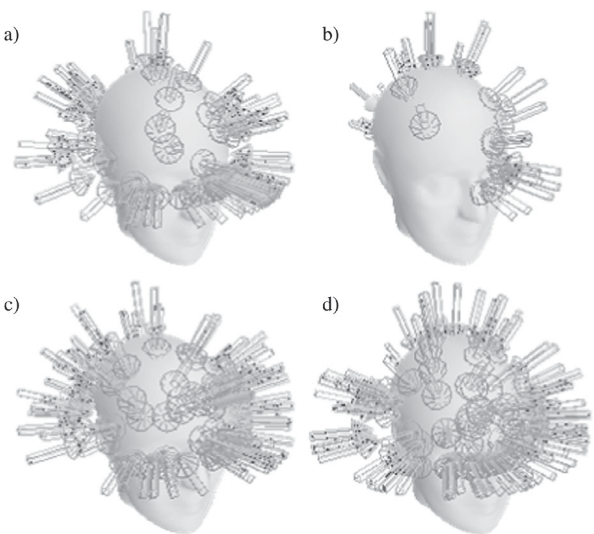

FIGURE 6. One game's cumulative impact vectors for 4 positions. A, Fullback. B, Receiver. C, Defensive lineman. D, Linebacker.

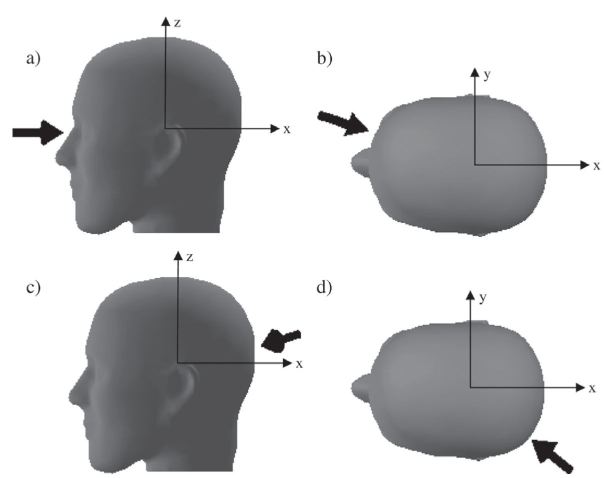

FIGURE 8. The impact vectors associated with both imacts highlighted in Figure 7. A, Concussion, xz-plane. B, Concussion, xy-plane. C, Neck stinger, xz-plane. D, Neck stinger, xy-plane.

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.[22] As a result, these players would not seek medical attention and continue to play. This was precisely the type of behavior we observed in the 1 concussive event we did record. It is therefore likely that some of the players in our current study may be represented in this unreported category.

While the acceleration data from this study can be correlated to brain injury, the data themselves are not descriptive of a tissue-level cause for brain injury such as tissue strain. However, it is anticipated that through expansion of the number of players instrumented and the number of seasons data are collected, more injury events will be observed and head acceleration data recorded. The increased number of injury events will allow for more thorough conclusions to be made on the accuracy of previously published injury thresholds. 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.

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