

Published in final edited form as:

Ann Biomed Eng. 2012 January ; 40(1): 237–248. doi:10.1007/s10439-011-0422-2.

Measuring Head Kinematics in Football: Correlation Between the Head Impact Telemetry System and Hybrid III Headform

Jonathan G. Beckwith¹, Richard M. Greenwald^{1,2}, and Jeffrey J. Chu¹

¹Simbex, 10 Water Street, Suite 410, Lebanon, NH 03766, USA

²Thayer School of Engineering, Dartmouth College, Hanover, NH, USA

Abstract

Over the last decade, advances in technology have enabled researchers to evaluate concussion biomechanics through measurement of head impacts sustained during play using two primary methods: (1) laboratory reconstruction of open-field head contact, and (2) instrumented helmets. The purpose of this study was to correlate measures of head kinematics recorded by the Head Impact Telemetry (HIT) System (Simbex, NH) with those obtained from a Hybrid III (HIII) anthropometric headform under conditions that mimicked impacts occurring in the NFL. Linear regression analysis was performed to correlate peak linear acceleration, peak rotational acceleration, Gadd Severity Index (GSI), and Head Injury Criterion (HIC₁₅) obtained from the instrumented helmet and HIII. The average absolute location error between instrumented helmet impact location and the direction of HIII head linear acceleration were also calculated. The HIT System overestimated Hybrid III peak linear acceleration by 0.9% and underestimated peak rotational acceleration by 6.1% for impact sites and velocities previously identified by the NFL as occurring during play. Acceleration measures for all impacts were correlated; however, linear was higher ($r^2 = 0.903$) than rotational ($r^2 = 0.528$) primarily due to lower HIT System rotational acceleration estimates at the frontal facemask test site. Severity measures GSI and HIC were also found to be correlated, albeit less than peak linear acceleration, with the overall difference between the two systems being less than 6.1% for either measure. Mean absolute impact location difference between systems was $31.2 \pm 46.3^\circ$ (approximately 0.038 ± 0.050 m), which was less than the diameter of the impactor surface in the test. In instances of severe helmet deflection (2.54–7.62 cm off the head), the instrumented helmet accurately measured impact location but overpre-dicted all severity metrics recorded by the HIII. Results from this study indicate that measurements from the two methods of study are correlated and provide a link that can be used to better interpret findings from future study using either technology.

© 2011 Biomedical Engineering Society

Address correspondence to Jonathan G. Beckwith, Simbex, 10 Water Street, Suite 410, Lebanon, NH 03766, USA. jbeckwith@simbex.com.

CONFLICT OF INTEREST Jonathan Beckwith, Jeffrey Chu and Richard Greenwald, as employees or as beneficial owner of Simbex, have a financial interest in the HIT System technology used in this study.

Additional Contributions: David Viano, as co-chairman of the NFL's mTBI Committee at the time of this data collection, approved the initial protocol and funding to Biokinetics, but he did not participate in data collection, analysis, or interpretation of the data. At his request, following primary data collection and analysis by Simbex and Biokinetics, impact sites A' and A'' were added to the test protocol. These data were collected 7 months after the initial testing. The interpretation of the material presented within this communication is solely that of the authors and not necessarily reflective of the sponsor's views. The authors would like to thank Mr. Chris Withnall and Mr. Michael Wonnacott from Biokinetics and Associates, Ltd who provided technical expertise when needed and assisted with data collection. In addition, the authors would like to thank Joseph Crisco from Brown University for assisting with data interpretation.

Keywords

Head impact; Concussion; Biomechanics; Football

INTRODUCTION

The Centers for Disease Control and Prevention has concluded that mild traumatic brain injury (mTBI), also known as concussion, is at epidemic levels with an estimated 1.6–3.8 million cases occurring annually in the United States.⁷ Interest in understanding the pathophysiology of the injury, as well as potential short- and long-term effects, has never been greater. Recent studies suggest a link exists between concussion history and developing mild cognitive impairment, clinical depression, and early onset of Alzheimer's disease,^{21,22} creating a public and scientific debate about effective prevention strategies for concussions. A 2007 epidemiological study conducted by the National Collegiate Athletic Association (NCAA) on American Football revealed that concussion accounted for 6.8% of all game injuries and 5.5% of all practice injuries, making it the third most common game injury and fourth most common practice injury.¹⁴ Given the high number of injuries per athlete exposure and frequency of plays resulting in blows to the head, researchers have been using the football field as a living laboratory to study brain injury since the 1960s.^{28,40} To better understand the etiology of concussion and the relationship between head impacts and injury, however, a large number of measured impacts from multiple athletes, both concussed and non-concussed, are required.

In order to quantify concussion biomechanics in football, the National Football League (NFL) commissioned a series of studies to quantify the characteristics of impacts associated with concussion diagnosis.^{33,37–39} Using video reconstruction techniques, Pellman *et al.* were able to determine the impact velocity for 31 of 182 cases of head impact (both with and without diagnosis of concussion) captured on film. By matching the obtained impact velocity and observed direction, these 31 cases were re-enacted in a laboratory using Hybrid III (HIII) anthropomorphic test devices (ATDs). Head linear and rotational accelerations were measured from accelerometers in the ATD's head, permitting calculation of injury metrics such as Gadd Severity Index (GSI) and Head Injury Criterion (HIC).^{19,45} Due to practical limitations (e.g., complexity and cost of reconstruction), the dataset created by the NFL was biased toward injury events when compared with actual rates of injury per head impact sustained in play. Nonetheless, risk curves were generated from these data which have become commonly cited thresholds for concussion injury. In addition to the developed injury risk curves, several follow-up studies were also conducted to characterize the relationship between head impact measures and clinical outcome, including neuropsychological testing, symptomatology, and repeat injury.^{34–36}

To assess the accuracy of acceleration measures derived from the NFL studies, Newman *et al.* quantified the error associated with reconstruction variables, including effects of noise in the data acquisition system, processing inaccuracy, and determining the impact velocity from multiple camera angles. Of these variables, deriving input parameters from video was identified as the primary source of error, with the estimated difference between actual and reconstructed head acceleration potentially reaching levels of up to 17% for peak linear acceleration and 25% for peak rotational acceleration.³¹ In an additional study to quantify the effect of head, neck, and torso coupling, Beusenberg *et al.* found that if a player's neck coupling with the head (e.g., stiffness, strength, *etc.*) deviates from that of the HIII head and neck, peak linear acceleration can be altered by more than 15% and peak rotational acceleration will be drastically different.³ Due to the complexity of these reconstructions and the beneficial insight they provide, these errors were deemed to be within an acceptable

level, and output measures from the reconstructions were considered reasonable for use in estimating the actual head kinematics for this group of impacts.³¹

Subsequently, the NFL commissioned Biokinetics (Ottawa, Canada) to develop a linear impactor test device to replicate head impacts to a HIII dummy head and neck at the impact velocities derived from the laboratory reconstructions.³⁹ The linear impactor system has been proposed as a standardized helmet-testing device but has not yet been adopted by governing bodies or standards organizations.^{25,29} The benefit of a laboratory system for impacting helmets is to allow parametric evaluation of test parameters (e.g., helmet models, impact direction and energy, *etc.*) on biomechanical response (e.g., head acceleration, GSI, *etc.*) by pairing consistent test conditions with precision measurement devices and established mathematical techniques in a controlled environment. Similar to the full ATD reconstructions, though, the linear impactor test does not fully replicate on-field conditions as it standardizes the biomechanical response, which can affect biomechanical response to impact.²⁵ It has also proven to be challenging to achieve measurement repeatability across multiple test laboratories because of potential differences in impactor surfaces and HIII head and neck properties.

Head Impact Telemetry (HIT) System technology (Simbex, Lebanon, NH; Sideline Response System, Riddell, Chicago, IL) records the frequency, location, and magnitude of impacts sustained by football players during live play,^{4,11,16,20,27,44} and was specifically developed to enable research to better understand the relationship between measured parameters of head kinematics and concussion. This technology incorporates an array of non-orthogonal accelerometers (Analog Devices, Inc., Cambridge, MA), data acquisition, and RF telemetry hardware into self-contained inserts placed in commercially available helmets. While physically connected to the helmet, the instrumented insert acts as an effective spring to maintain contact with the head during impact and to decouple head acceleration from helmet acceleration.²⁶ Because traditional mathematical approaches to calculate linear and rotational acceleration at the head center of gravity, such as those employed by the HIII, require precise mounting of accelerometers that is not practical for field implementation, the HIT System uses proprietary, simulated-annealing optimization algorithms to estimate linear and rotational acceleration.^{8,10} While this approach has been shown to be mathematically robust, on-field variables that could potentially affect the measured and computed output variables (e.g., linear and rotational acceleration, impact location, impact duration, *etc.*) include relative motion between the head and the sensors and separation of the sensors off the head during the impact. The first on-field use of the system occurred during the 2003 football season,¹⁷ and, since then, the HIT System has been validated in both laboratory and on-field settings.^{6,10,17,18,26} To date, usage has spread to all levels of play (youth, high school, collegiate, and professional) which has generated a head impact database that now includes more than 1.5 million recorded head impacts.⁴²

Given that data obtained from both the HIT System and the NFL studies are now commonly cited in the literature, it is important to understand the similarities and differences in output from these two approaches. To address this issue, the NFL's Mild Traumatic Brain Injury Committee requested a laboratory comparison of the output from the HIT System with the Hybrid III ATD under conditions that mimicked impacts occurring in the NFL that are associated with and without diagnosed concussion. The purpose of this study was to correlate measures of head impact biomechanics recorded by both systems when tested on the linear impactor with impact velocities and impact locations similar to those identified in the NFL laboratory reconstructions.

METHODS

Experimental Setup

The head of a 50th percentile HIII ATD (Humanetics, Plymouth MI) mounted to a standard HIII neck was used to replicate the response of a football player's head. Per manufacturer's specification, the cable in the HIII neck was tensioned to 1.1 Nm (10 in-lb.). Two nylon stockings were stretched over the HIII head before fitting the helmet to reduce friction and to provide a more realistic interface between the helmet and the headform skin. The HIII head-and-neck system was connected to a table fitted with linear bearings, permitting the head-and-neck assembly to translate in the direction of impact (Fig. 1). The sliding table allows for both more realistic kinematics and improved equipment durability while not affecting the head response, since the head acceleration impulse occurs before significant neck bending.³⁹

Impacts were delivered to the HIII using a pneumatically activated linear impactor (Biokinetics Ottawa, Canada; Fig. 1).^{29,39} The linear impactor uses a pressure storage vessel with attached high-speed solenoid valve to accelerate a ram by in-rush of high-pressure air into a pneumatic cylinder. The ram detaches from the cylinder to insure that the ram is not being driven at the moment of impact. An impacting face is attached to the ram via a metal plate. The impacting face is made of ultra-high molecular weight polyeth-ylene, and has the same curvature as that of a football helmet and a surface diameter of 127 mm (5 in). A 35-mm (1-3/8 in) layer of vinyl nitrile foam (VN600, DerTex, Saco, Maine), the same material used in many models of football helmets, was placed behind the impactor surface to mimic helmet-to-helmet contact. Impact velocity was measured by passing the impactor ram through a light gate just before contact with the helmet. The light gate also acted as a trigger for HIII data collection.

A new medium Revolution helmet (Riddell, Elyria, OH) with standard inflatable liner and 1.5-in. jaw pads was instrumented with a HIT System sensor insert. A G2EG-R style facemask was used for all impacts. The facemasks and attachment hardware were visually inspected between each impact and replaced with new equipment at first sign of deformation. The helmet was initially fit to the instrumented headform according to the manufacturer's fitting specifications and then re-adjusted between each trial to maintain consistent test conditions.

A high-speed camera was used to record the first impact trial at each reconstruction condition. Video was recorded at 1000 Hz and was triggered via the impactor light gate thereby synchronizing video with HIII data. High-speed video was used to confirm consistent helmet placement and to qualitatively review the quality of impact reconstruction.

Primary Impact Sites

Four primary impact sites, designated as A, B, C, and D (Fig. 2), were identified by Pellman *et al.*³⁷ as points of contact that most frequently result in concussion for NFL athletes. In that study, the average impact velocity for impacts associated with and without concussion was determined to be 9.3 ± 1.9 m/s and 7.0 ± 2.6 m/s, respectively. For this study, each site was impacted at four target speeds: 4.4, 7.4, 9.3, and 11.2 m/s. The highest three velocities correspond to the average speed, ± 1 standard deviation, of video-reconstructed impacts sustained before diagnosis of concussion in NFL players.³⁸ 4.4 m/s represents the average velocity, -1 standard deviation, of all sub-concussive impacts evaluated by the NFL. This lower impact velocity was selected to insure inclusion of conditions that may not result in injury, but occur most often in the field.^{20,24,30} Between three and five trials were conducted for each site and speed combination (Table 1).

Secondary Impact Sites

After initial testing and analyses were conducted, two additional impact sites, A' and A'' (Fig. 2), were added to the test protocol at the request of the then co-chairman of the NFL mTBI Committee to characterize HIT System measurement capability during conditions where extensive helmet movement was anticipated to occur. Sites A' and A'' were identified by Craig (2007) who, when reevaluating video from the initial NFL studies, suggested a sub-category of site A impacts existed that resulted in chinstrap loading.⁹ The two additional sites were variations of the original site A, located directly in line with the mid-sagittal plane of the headform, and tilted away from the impactor at -10° and -20° angles, respectively. Based on the recommendations made by Craig,⁹ and previous experience which predicted that facemask deformation and HIII neck damage were likely to occur, the 11.2 m/s speed was not used for A' and A'' tests. Three trials were conducted at each test condition except for A'' at 4.4 m/s which had four trials.

Laboratory Measurements

The HIII head was outfitted with nine accelerometers (Endevco Corp., San Juan Capistrano, CA) in a 3–2–2 configuration.³² This arrangement allows for the resolution of three-dimensional (3D) angular accelerations using linear accelerometers, and correction for centripetal accelerations.¹⁵ The accelerometer signals were filtered initially with a CFC1000 hardware filter in the signal-conditioning device. The data were collected using a custom Labview program, post-processed using an additional CFC180 filter,³¹ and used for calculation of linear and rotational head accelerations. Direction of the impact recorded by the HIII was determined by converting the 3D peak linear acceleration vector into a spherical coordinate space.

Instrumented Helmet Measurements

Accelerometer data were collected by the instrumented helmet's on-board microprocessor at 1000 Hz over a 40-ms window for each event. Data collection occurred when any of the six accelerometers exceeded a pre-filtered threshold of 10 g (8-ms pre-trigger and 32-ms post-trigger). An on-board 400 Hz low-pass filter was applied to the data after a 0.5-Hz AC hardware filter removed any DC offset. Data were then transmitted wirelessly to a laptop computer where each event was processed using a simulated-annealing optimization algorithm that solved for linear acceleration magnitude at the head center of gravity.¹⁰ Impact location was estimated by taking the reflection of the 3D direction vector of peak linear acceleration which is optimized from the peak output of each of the six accelerometers. The directional vector is then converted to a spherical head coordinate system (azimuth and elevation) that has been previously reported by Greenwald *et al.*²⁰ Rotational acceleration at the time of peak linear acceleration was calculated as described by Rowson *et al.* using the modeled equations of motion for a force acting on the head, the anterior–posterior and medial–lateral components of the peak linear acceleration vector, and the directly measured relationship between linear and rotational acceleration or on-field head impacts.⁴³

Data Analysis

Linear acceleration time series data from both the HIT System and HIII were processed in real time to obtain the resultant linear acceleration for each impact. Because a simultaneous trigger was not feasible, the linear acceleration resultants were synchronized post-processing by minimizing the RMS error between resultants.^{1,41} HIII data were then truncated to a 40-ms time window to allow direct time series comparison with the instrumented helmet. Resultant linear acceleration data from both systems were utilized to calculate two impact

metrics used for relating time-weighted head acceleration to risk of injury, GSI and HIC₁₅. GSI was calculated for each impact event¹⁹:

$$\text{GSI} = \int^T a(t)^{2.5} dt \quad (1)$$

where $a(t)$ is the linear resultant acceleration of the head CG, and T is the impact duration. HIC₁₅ was evaluated over an incremental time window^{38,45} of maximum duration $t_2 - t_1 = 15$ ms:

$$\text{HIC}_{15} = \left\{ (t_2 - t_1) \left[\int_{t_1}^{t_2} a(t) dt / (t_2 - t_1) \right]^{2.5} \right\}_{\max} \quad (2)$$

where t_1 and t_2 are the initial and final points of the time window and $a(t)$ is the linear resultant acceleration of the CG.

To establish a correlation between severity measures from the HIT System and HIII, linear regression analysis (Microsoft Excel 2010) was performed on all acceleration and severity measures—peak linear acceleration, peak rotational acceleration, GSI, and HIC₁₅.¹ All regressions were performed both on the entire dataset and by impact site using Eq. (3):

$$y = mx + y_0 \quad (3)$$

where x is the HIII measure, y is the HIT System measure, and the linear slope, m , is the relationship between the measurements. For all conditions, y_0 was constrained to be zero since both systems have a baseline output of zero when not impacted. The coefficient of determination (r^2) was also calculated for each regression as a measure of goodness of fit. Finally, the average absolute location difference between the estimated instrumented helmet location and the direction vector (in spherical coordinates) of the peak HIII linear acceleration was calculated for each impact. The overall average difference between the two measures along with the difference by impact site was calculated.

RESULTS

Primary Impact Sites

Fifty-four impacts were measured by the instrumented helmet and HIII over four target velocities and impact sites previously defined by the NFL as potentially injurious (Locations A–D). A high correlation for peak linear acceleration was found between the two systems, $r^2 = 0.903$, with a slope of 1.009 for the linear fit trendline (Fig. 3) indicating a 0.9% overprediction of the HIT System compared to the HIII for all locations and velocities. Analysis by impact site again showed high correlation between the two systems ($r^2 = 0.822$ – 0.987) with no linear slope deviating more than 8.4% of nominal for any impact location (Table 2).

Peak rotational acceleration was correlated for B, C, and D sites ($r^2 = 0.710$ – 0.981) but not for A location ($r^2 = 0.415$). For all impact sites, the slope of the fit trendline was 0.939 ($r^2 = 0.528$); however, the coefficient of correlation ($r^2 = 0.800$) improved when only considering sites B, C, and D (Fig. 3).

The coefficient of correlation between the two measurement systems for GSI and HIC₁₅ were $r^2 = 0.846$ and 0.787 , respectively with no individual impact site correlation lower than $r^2 = 0.609$. Slopes of the fit trendlines for all impact events were 0.948 and 1.061 for GSI

and HIC_{15} , respectively (Table 2). Impact location recorded by the HIT System had an average absolute location difference of $31.2 \pm 46.3^\circ$ for all impacts when compared to the acceleration direction of the HIII (Table 2). Most of this error can be attributed to six impacts to the A site, which were symmetrically opposite of the expected. When considering location measurement without the A impact site, the average difference for remaining sites was only $13.2 \pm 6.3^\circ$, well within the overall 0.127-m diameter of the impactor surface (10° absolute error equates approximately to 0.0127 m). The average impact location as estimated by both systems can be seen in Fig. 4 for all impact locations.

Secondary Impact Sites

For the 19 additional frontal impacts at sites A' and A'', the HIT System had a moderate overall peak linear acceleration correlation with the HIII ($r^2 = 0.556$) and greatly overpredicted the HIII as indicated by a factor of 2.673. While the HIT System overpredicted the HIII at both sites, linear acceleration was highly correlated for site A' while no correlation was identified for site A'' (Table 2). Rotational acceleration, GSI, and HIC_{15} all followed similar trends of overall moderate correlation ($r^2 = 0.438, 0.609,$ and 0.649), high correlation for site A', no correlation for site A'', and the HIT System overpredicting the HIII for each measure ($m = 1.627, 5.346,$ and 3.835). Impact location recorded by the HIT System for A' and A'' locations had an average absolute location difference of $42.1 \pm 13.4^\circ$, which was consistent with the results at the A, B, C, and D locations (Table 3). The source of this discrepancy can be attributed to location estimates from the HIII that did not correspond to the point of impact for either impact site, but rather went directly through the nose.

DISCUSSION

Researchers have long sought to determine the relationship between head kinematics following impact and the pathophysiology of traumatic brain injury. One approach to understanding this relationship for mTBI has been to treat the football playing field, where athletes may sustain more than 2000 impacts to the head during a season,¹¹⁻¹³ as a living laboratory. Specifically, researchers have monitored the impacts sustained by football players and have attempted to correlate their impact exposure with signs and symptoms of injury. Over the last decade, two primary approaches have been implemented for quantifying head kinematics following impacts experienced during play: laboratory reconstruction of impacts recorded on video and on-field measurement using instrumented helmets. The aim of this study was to correlate HIT System output with Hybrid III ATD output in a laboratory test using a linear impactor system under realistic input conditions.

Measures of peak linear acceleration, GSI, and HIC_{15} obtained from the instrumented helmet and HIII were found to be correlated ($r^2 = 0.903, 0.846,$ and 0.787), and the overall system-to-system relationship was less than 6.1% for each measure (0.9%, 5.2%, and 6.1%, respectively). Higher deviations for both GSI and HIC_{15} were not unexpected given the reliance of these measures on the entire 40 ms linear acceleration resultant which leads to more potential points of deviation than maximum value measures. While little difference in correlation was found between impact sites for peak linear acceleration, site A resulted in the lowest correlation for both GSI and HIC_{15} . A review of high-speed video data for these impacts provided insight to the discrepancy between the A site and other impact sites. For the A site, the impactor ram drove the helmet downward and counter clockwise toward the HIII before head acceleration began (approximately 5 ms of helmet contact before head acceleration). Once the impactor surface settled into the opening of the facemask, the helmet and headform moved simultaneously. This helmet movement prior to head acceleration caused a discrepancy in the linear acceleration resultant time series between the two systems from which GSI and HIC_{15} are calculated.

In these tests, the instrumented helmet underestimated peak rotational acceleration compared to the HIII by 6.1%. The overall correlation between the two systems, however, was only moderate and lower than the other impact measures. This can be attributed primarily to measurements at the A site which had low correlation with the HIII ($r^2 = 0.415$) when compared with the other primary sites B, C, and D, ($r^2 = 0.800$). The processing algorithm used in conjunction with the instrumented helmet in this study calculates rotational acceleration at the time of peak linear acceleration from the equations of motion for force applied on the head.⁴³ The unknown parameters of head mass, head moment of inertia, and the point of rotation about the neck are determined by the relationship of peak linear and rotational acceleration of on-field impacts obtained from a similar device paired with a full six degree of freedom processing algorithm.^{8,41} Compared to the full six degree of freedom helmet, this approach reduces cost of the instrumented helmet and the computation time required for processing head acceleration measures, thus allowing researchers to conduct studies on a larger sample of players. This approach, however, also limits rotational acceleration calculations to two degrees of freedom since acceleration about the axis extending through the neck cannot be calculated. Similar to findings from previous studies evaluating the accuracy of rotational acceleration measures from the HIII head and neck,^{3,31} results from these tests indicate that the HIT System estimation of rotational acceleration is an accurate approach when analyzing and reporting results based on impact distributions; however, care should be taken when reporting rotational acceleration from either system for single impact events.

The average difference between the direction of head acceleration estimated by the instrumented helmet and HIII was 31.2° for all primary test impacts. Based on HIII dimensions, a 10° offset correlates to approximately 0.0127 m of head circumference, and, given the impactor size (diameter = 0.127 m), these results indicate the instrumented helmet measures are well within practical reason for on-field studies. The instrumented helmet identified the direct opposite acceleration for six of the twelve site A impacts causing a higher mean difference at this site compared to all others. It is important to note that the HIT System algorithm used in this study optimizes for impact location based on the temporal relationship of peak acceleration values obtained from each of the six embedded accelerometers. While the resultant head acceleration corresponds with the head CG, the estimation of location best approximates impact location on the helmet. In most cases the impact location and direction of acceleration is the same, but, there are occasions where it is not. A good example of this can be seen in the average estimated impact locations of both the HIII and HIT System at sites A' and A'' (Fig. 5). While the HIT System correctly identifies the impact location as being low on the facemask, the direction at peak acceleration measured by the HIII is directly through the nose. Incorrect identification of acceleration direction for the six site A impacts is related to the facemask and impactor surface interaction previously described. Initial contact caused the helmet to move downward before head acceleration-mimicking motion typically resulting from an impact to the back of the head. It is unclear if this phenomenon is limited to the conditions present in these specific laboratory tests or if there is the potential for misidentifying impact location in the field; however, on-field performance of the HIT System has previously been verified through video review without description of inaccurate location measurements and, anecdotally, have not been recorded under similar test conditions at other independent test laboratories.

Nineteen additional frontal impacts were measured at two sites (A' and A'') with increased neck angle relative to the impactor, causing primary loading of the chinstrap and large deflection of the helmet off the head. In these tests, the aim was to challenge the HIT System's ability to retain contact with the head during the impact. For these 19 additional impacts, the instrumented helmet had high correlation with the HIII for all A' measures (r^2

≥ 0.883) and low correlation for all A'' measures ($r^2 \leq 0.308$). All four severity metrics were greatly overpredicted by the instrumented helmet. As with the A location, high-speed video illustrated excessive helmet departure from the head 6 ms before initial head acceleration (Fig. 6). Maximum lift of the helmet off the head was estimated to be in the range of 1–3 in. depending on impact velocity. For the highest impact speeds, the entire gap between the face and facemask was closed as the bottom facemask bar was driven into the HIII chin. From these observations, it can be assumed that the instrumented helmet was potentially no longer in contact with the head at the time of initial acceleration and most likely not in contact at the time of peak acceleration. The large overestimation of linear acceleration would therefore be attributed to a high component of helmet acceleration artifact measured by the instrumented helmet.²⁶ To account for helmet movement that will inevitably occur during play, the instrumented helmet is designed to isolate head acceleration from helmet acceleration via the use of spring-like actuators.²⁶ Accelerometers are embedded in the actuators so that head contact is maintained even during helmet movement. These actuators can account for the majority of helmet movement that occurs during a typical impact, however, in the event that the helmet is dislodged from the head before the acceleration pulse, as was the case in these laboratory tests, the system most likely will overestimate head acceleration. This represents a practical limitation of the system that must be considered by adopters of the technology.

The results from this study predict that impacts to the A' and A'' tests measured by the instrumented helmet will result in much higher accelerations (e.g., 2–5 times the actual value) for a given impact velocity when compared to other impact locations. It is important to note, however, that, independent of the large differences in acceleration magnitudes recorded between the systems at A' and A'', the impact location was closely identified. While impacts will inevitably occur on the field at these sites, we do not believe that the A' and A'' sites and, in some instances, the A site, as currently tested using the linear impactor represent realistic cases of helmet impact and helmet motion that occurs on the football field. Using data collected with the HIT System, Mihalik *et al.* have reported statistically lower average peak linear acceleration for impacts to the frontal region of the head than to the top.²⁷ Similarly, Greenwald *et al.* reported frontal impacts are statistically indistinguishable in terms of impact severity from side of the head impacts and are much lower than either impacts to the top or back of the head following analysis of on-field head impact data from 13 high school and collegiate football teams.²⁰ In the most extensive data collection reported to date, Crisco *et al.* recorded head impacts with the HIT System from three collegiate football teams over three seasons, resulting in 286,636 recorded impacts.¹² Front of the head impacts (i.e., front 90° quadrant of the head and below an elevation level of 65° that generally relates to mid-forehead)²⁰ accounted for 41.7% of these events. To compare with the laboratory tests reported here, data from the cited study were separated into three general levels of elevation to estimate forehead, mid facemask (similar to A and B sites), and low facemask (similar to A' and A'' sites) impacts (Table 4). Impacts to the lower portion of the facemask do occur on the field (13.3% of all Front impacts, 5.5% of all impacts); however, a comparison of median and top 95th percentile levels of linear and rotational acceleration by impact elevation does not support the A' and A'' laboratory test results which suggest that the HIT System will consistently measure low facemask impacts by a factor of 2–3 times higher than those of mid facemask and helmet shell impacts. While the A' and A'' impact sites do provide insight into potential limitations of using instrumented helmets, the discrepancy in on-field results with those predicted by linear impactor tests also demonstrates the difficulty of creating a biofidelic reconstruction of live events in a laboratory setting.

We chose to use regression analysis as the primary means for correlating output between the HIT System and HIII. One of the primary benefits of the HIT System is its ability to record

large amounts of *in vivo* impact data that are inclusive of all on-field scenarios that can be experienced by different playing positions and skill levels.^{5,11,12,27,42,44} It can reasonably be assumed that each of the 286,636 impacts reported in the previously described study by Crisco *et al.* represents a specific combination of input conditions that could influence the player's head kinematics. Because of this variability in conditions, researchers have primarily chosen to present data from the HIT System in the form of impact distributions inclusive of a variety of impact conditions (e.g., impact location, impact severity, playing position, *etc.*).

By utilizing regression analysis, an expected relationship between the overall distributions of output from the two systems can be obtained as well as an indicator for the strength of that relationship. This relationship, though, should only be assumed for conditions included in the test distribution. For example, considering the A, B, C, and D impact sites and associated test velocities were previously found to be representative of concussion in the NFL, it would be expected from these test results that HIII measures obtained by Pellman *et al.* would closely match those recorded by the HIT System for concussive injuries when compared as a group. This close relationship has been confirmed in several studies,^{4,23,42} including one analysis that found no statistical difference in peak linear acceleration, peak rotational acceleration, HIC₁₅, or GSI between 55 on-field impacts associated with concussion and those obtained from the 25 reconstructions of concussive impacts conducted by the NFL.² The relationships provided by regression analysis do not, however, provide an estimate of how the two systems may differ for individual impacts. One approach to obtaining this estimate would be to calculate the percent difference for each impact. For all trials conducted at the primary impact sites, this method would result in linear acceleration differences between 0.1–38.9% for individual impacts simulated in this study. While there is benefit to this approach, it has only limited utility considering that the association can only be applied for the exact input conditions, which are rarely known for on-field impacts. Because the exact differences for single impact measures between the HIT System and HIII remains unclear because of the impracticality of evaluating all possible on-field scenarios in the laboratory, researchers should use caution when analyzing results from isolated events. In general, though, the relationship found through regression suggests distributions of measurements obtained on-the field with the HIT System will be nearly identical to those obtained in the laboratory as long as the laboratory tests reflect field conditions. This relationship appears to be strong even for distributions of relatively few data points, as shown by the 54 impacts evaluated in this study.

Athletic environments, American football in particular, have long been viewed as a potentially rich testbed for understanding concussions. Over the last decade, two methods have been employed to obtain head acceleration measures occurring from actual impact events. Laboratory reconstructions utilize precision measurement equipment to recreate on-field impacts but are limited by cost, technical expertise, and the necessary assumptions of the human surrogate models. The HIT System is an on-field measurement system that allows for large-scale data collection by actively measuring head acceleration of athletes during play. Given that the current knowledge base has been developed using both techniques, the current study was required to establish the relationship between these two measurement methodologies. From these laboratory tests, a relationship has been established between the two systems, particularly when considering peak metrics such as linear and rotational acceleration for distributions of impacts; however, it is important for users of HIT System technology to understand the practical limitation of ensuring proper fit of a player's helmet. This limitation could result in error for single impacts that is similar to those previously reported for laboratory reconstructions. Results from this study indicate that measurements from the two methods of study are correlated and provide a link that can be used to better interpret findings from future study using either technology.

Acknowledgments

The authors gratefully acknowledge the funding for this study provided by the National Institutes of Health (R01HD048638). HIT System technology described in this study was developed with support from the National Institutes of Health (R44HD04743 and R01HD048638) and from Riddell, Inc. (Chicago, IL). Helmets and associated hardware were also provided by Riddell, Inc. The study protocol was developed in conjunction with the National Football League's Committee on Mild Traumatic Brain Injury, which provided funding to Biokinetics to operate the linear impactor. Following data collection at the said facility, data from both systems were mutually exchanged.

REFERENCES

1. Beckwith JG, Chu JJ, Greenwald RM. Validation of a noninvasive system for measuring head acceleration for use during boxing competition. *J. Appl. Biomech.* 2007; 23(3):238–244. [PubMed: 18089922]
2. Beckwith, JG.; Chu, JJ.; Crisco, JJ.; McAllister, TW.; Duma, SM.; Brolinson, PG.; Greenwald, RM. Severity of head impacts resulting in mild traumatic brain injury. American Society of Biomechanics. Penn State University; State College, PA: 2009. p. 1144
3. Beusenberg, M.; Shewchenko, N.; Newman, JA.; de Lange, R.; Cappon, H. Head, neck, and body coupling in reconstructions of helmeted head impacts. International Research Council on the Biomechanics of Injury. The Isle of Man; United Kingdom: 2001.
4. Broglio SP, Schnebel B, Sosnoff JJ, Shin S, Fend X, He X, Zimmerman J. Biomechanical properties of concussions in high school football. *Med. Sci. Sports Exerc.* 2010; 42(11):2064–2071. [PubMed: 20351593]
5. Broglio SP, Sosnoff JJ, Shin S, He X, Alcaraz C, Zimmerman J. Head impacts during high school football: a biomechanical assessment. *J. Athl. Train.* 2009; 44(4):342–349. [PubMed: 19593415]
6. Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald RM, Duma SM. Analysis of linear head accelerations from collegiate football impacts. *Curr. Sports Med. Rep.* 2006; 5(1):23–28. [PubMed: 16483513]
7. Centers for Disease Control and Prevention (CDC). Report to congress on mild traumatic brain injury in the United States: steps to prevent a serious public health problem. National Center for injury Prevention and Control; Atlanta, GA: 2003. p. 1–45.
8. Chu J, Beckwith JG, Crisco JJ, Greenwald RM. A novel algorithm to measure linear and rotational acceleration using single-axis accelerometers. 5th World Congress of Biomechanics. Munich, Germany. *J. Biomech.* 2006:S534.
9. Craig, MJ. Dissertation. Wayne State University; Detroit, MI: 2007. Biomechanics of jaw loading in football helmet impacts.
10. Crisco JJ, Chu JJ, Greenwald RM. An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single-axis accelerometers. *J. Biomech. Eng.* 2004; 126(6): 849–854. [PubMed: 15796345]
11. Crisco JJ, Fiore R, Beckwith JG, Chu JJ, Brolinson PG, Duma S, McAllister TW, Duhaime AC, Greenwald RM. Frequency and location of head impact exposures in individual collegiate football players. *J. Athl. Train.* 2010; 45(6):549–559. [PubMed: 21062178]
12. Crisco JJ, Wilcox BJ, Beckwith JG, Chu JJ, Duhaime AC, Rowson S, Duma SM, Maerlender AC, McAllister TW, Greenwald RM. Head impact exposure in collegiate football players. *J. Biomech.* 2011 Epub ahead of print.
13. Crisco JJ, Wilcox BJ, Machan JT, McAllister TW, Duhaime AC, Duma SM, Rowson S, Beckwith JG, Chu JJ, Greenwald RM. Head impact severity measures in collegiate football players. *J. Appl. Biomech.* 2011 (accepted).
14. Dick R, Ferrara MS, Agel J, Courson R, Marshall SW, Hanley MJ, Reifsteck F. Descriptive epidemiology of collegiate men's football injuries: national collegiate athletic association injury surveillance system, 1988–1989 through 2003–2004. *J. Athl. Train.* 2007; 42(2):221–233. [PubMed: 17710170]

15. DiMasi, FP. Transformation of nine-accelerometer-package (nap) data for replicating headpart kinematics and dynamic loading: Technical report dot hs-808 282. National Highway Traffic Safety Administration; [Internet][cited. Available]
16. Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, Greenwald RM, Chu JJ, Crisco JJ. Analysis of real-time head accelerations in collegiate football players. *Clin. J. Sport. Med.* 2005; 15(1):3–8. [PubMed: 15654184]
17. Duma, S.; Manoogian, S.; Brolinson, G.; Goforth, M.; Donnenwerth, J.; Greenwald, R.; Chu, J.; Bussone, B.; Stitzel, J.; Crisco, J. Measuring real time head accelerations in collegiate football players. American Society of Biomechanics; 2004.
18. Funk JR, Duma SM, Manoogian SJ, Rowson S. Biomechanical risk estimates for mild traumatic brain injury. *Annu Proc Assoc Adv Automot Med.* 2007; 51:343–361. [PubMed: 18184501]
19. Gadd, CW. Use of a weighted-impulse criterion for estimating injury hazard. 10th Stapp Car Crash Conference; New York. Society of Automotive Engineers; 1966. p. 164-174.
20. Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery.* 2008; 62(4):789–798. [PubMed: 18496184]
21. Guskiewicz KM, Marshall SW, Bailes J, McCrea M, Cantu RC, Randolph C, Jordan BD. Association between recurrent concussion and late-life cognitive impairment in retired professional football players. *Neurosurgery.* 2005; 57(4):719–726. [PubMed: 16239884]
22. Guskiewicz KM, Marshall SW, Bailes J, McCrea M, Harding HP Jr, Matthews A, Mihalik JR, Cantu RC. Recurrent concussion and risk of depression in retired professional football players. *Med. Sci. Sports Exerc.* 2007; 39(6):903–909. [PubMed: 17545878]
23. Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. *Exerc. Sport. Sci. Rev.* 2011; 39(1):4–11. [PubMed: 21088602]
24. Guskiewicz KM, Mihalik JP, Shankar V, Marshall SW, Crowell DH, Oliaro SM, Ciocca MF, Hooker DN. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. *Neurosurgery.* 2007; 61(6):1244–1252. [PubMed: 18162904]
25. Gwin JT, Chu JJ, Diamond SG, Halstead PD, Crisco JJ, Greenwald RM. An investigation of the noxae linear impactor test method based on in vivo measures of head impact acceleration in American football. *J. Biomech. Eng.* 2010; 132(1):011006. [PubMed: 20524744]
26. Manoogian S, McNeely D, Duma S, Brolinson G, Greenwald R. Head acceleration is less than 10 percent of helmet acceleration in football impacts. *Biomed. Sci. Instrum.* 2006; 42:383–388. [PubMed: 16817638]
27. Mihalik JP, Bell DRM, Marshall SW, Guskiewicz KM. Measurement of head impacts in collegiate football players: an investigation of positional and event-type differences. *Neurosurgery.* 2007; 61(6):1229–1235. [PubMed: 18162902]
28. Moon DW, Beedle CW, Kovacic CR. Peak head acceleration of athletes during competition—football. *Med. Sci. Sports.* 1971; 3(1):44–50. [PubMed: 5095335]
29. National Committee on Standards for Athletic Equipment. Nocsae standard linear impactor test method and equipment used in evaluating the performance characteristics of protective headgear and faceguards. 2006. p. 1-6.Document #081-04m04 (Proposed Status)
30. Naunheim RS, Standeven J, Richter C, Lewis LM. Comparison of impact data in hockey, football, and soccer. *J. Trauma.* 2000; 48(5):938–941. [PubMed: 10823540]
31. Newman JA, Beusenberg MC, Shewchenko N, Withnall C, Fournier E. Verification of biomechanical methods employed in a comprehensive study of mild traumatic brain injury and the effectiveness of American football helmets. *J. Biomech.* 2005; 38(7):1469–1481. [PubMed: 15922758]
32. Padgaonkar AJ, Krieger KW, King AI. Measurement of angular accelerations of a rigid body using linear accelerometers. *J. Appl. Mech.* 1975; 42(3):552–556.
33. Pellman EJ. Background on the national football league’s research on concussion in professional football. *Neurosurgery.* 2003; 53(4):797–798. [PubMed: 14560733]
34. Pellman EJ, Powell JW, Viano DC, Casson IR, Tucker AM, Feuer H, Lovell M, Waeckerle JF, Robertson DW. Concussion in professional football: Epidemiological features of game injuries and review of the literature—part 3. *Neurosurgery.* 2004; 54(1):81–96. [PubMed: 14683544]

35. Pellman EJ, Viano DC, Casson IR, Arfken C, Powell J. Concussion in professional football: Injuries involving 7 or more days out—part 5. *Neurosurgery*. 2004; 55(5):1100–1119. [PubMed: 15509317]
36. Pellman EJ, Viano DC, Casson IR, Tucker AM, Waeckerle JF, Powell JW, Feuer H. Concussion in professional football: Repeat injuries—part 4. *Neurosurgery*. 2004; 55(4):860–876. [PubMed: 15458594]
37. Pellman EJ, Viano DC, Tucker AM, Casson IR. Concussion in professional football: Location and direction of helmet impacts—part 2. *Neurosurgery*. 2003; 53(6):1328–1341. [PubMed: 14633299]
38. Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF. Concussion in professional football: Reconstruction of game impacts and injuries. *Neurosurgery*. 2003; 53(4):799–814. [PubMed: 14519212]
39. Pellman EJ, Viano DC, Withnall C, Shewchenko N, Bir CA, Halstead D. Concussion in professional football: helmet testing to assess impact performance—part 11. *Neurosurgery*. 2006; 58(1):78–96. [PubMed: 16385332]
40. Reid S, Tarkington JA, Epstein H, O’Dea T. Brain tolerance to impact in football. *Surg. Gynecol. Obstet.* 1971; 133(6):929–936. [PubMed: 5117389]
41. Rowson S, Beckwith JG, Chu JJ, Leonard DS, Greenwald RM, Duma SM. A six degree of freedom head acceleration measurement device for use in football. *J. Appl. Biomech.* 2011; 27(1): 8–14. [PubMed: 21451177]
42. Rowson S, Duma SM. Development of the star evaluation system for football helmets: integrating player head impact exposure and risk of concussion. *Ann. Biomed. Eng.* 2011; 39(8):2130–2140. [PubMed: 21553135]
43. Rowson S, Duma SM, Beckwith JG, Chu JJ, Greenwald RM, Crisco JJ, Brolinson PG, Duhaime AC, McAllister TW, Maerlender AC. Rotational head kinematics in football impacts: an injury risk function for concussion. *Ann. Biomed. Eng.* 2011 (in review).
44. Schnebel B, Gwin JT, Anderson S, Gatlin R. In vivo study of head impacts in football: a comparison of national collegiate athletic association division I versus high school impacts. *Neurosurgery*. 2007; 60(3):490–496. [PubMed: 17327793]
45. Versace, J. A review of the severity index. 15th Stapp Car Crash Conference; Warrendale. Society of Automotive Engineers; 1971. p. 771-796.

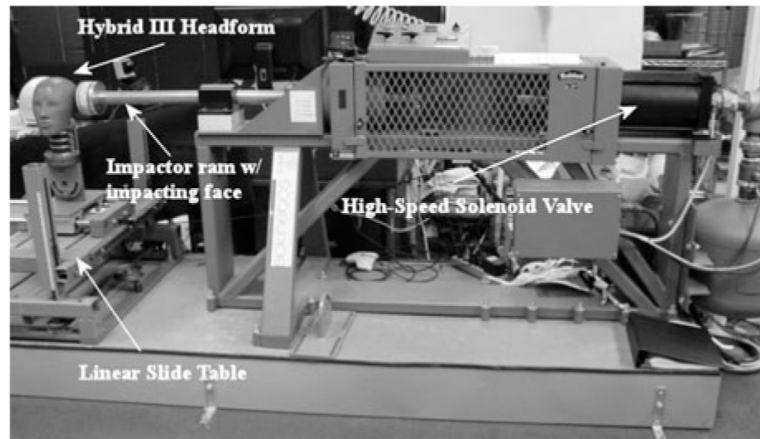


FIGURE 1.

A pneumatic linear impactor (Biokinetics, Inc) was employed to replicate on-field reconstruction of head impacts. The ram mass, impactor surface, and target velocities were selected to best simulate on-field head impacts occurring in the National Football League.

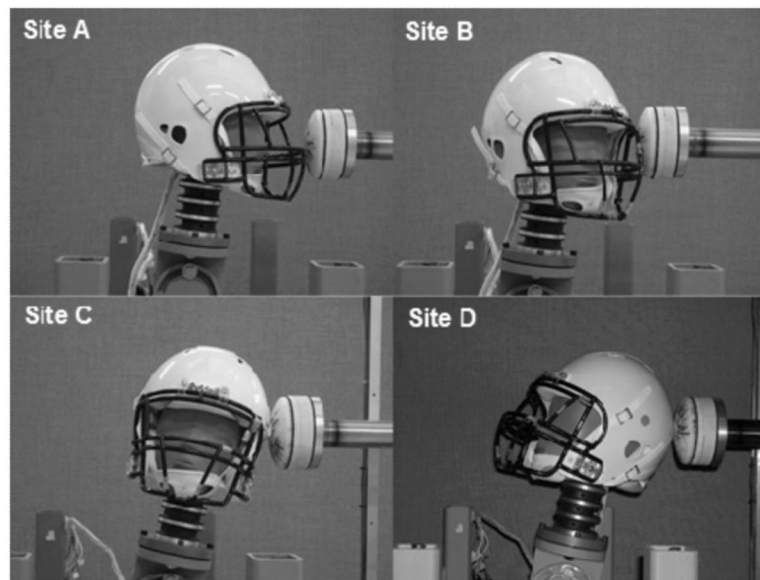


FIGURE 2.

Four primary impact sites (A, B, C, and D) were identified as points of contact that most frequently result in mTBI for NFL athletes. Each site was impacted at four target speeds: 4.4, 7.4, 9.3, and 11.2 m/s.

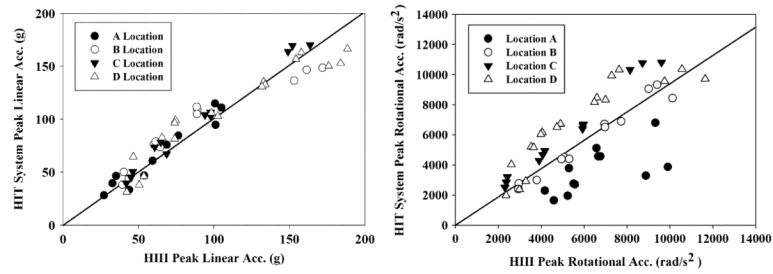


FIGURE 3.

The HIT System overestimated Hybrid III peak linear acceleration by 0.9% and underestimated peak rotational acceleration by 6.1% for all primary impact sites. Acceleration measures for all impacts were correlated; however, linear acceleration was higher ($r^2 = 0.903$) than rotational ($r^2 = 0.528$) primarily due to lower HIT System rotational acceleration estimates at impact site A.

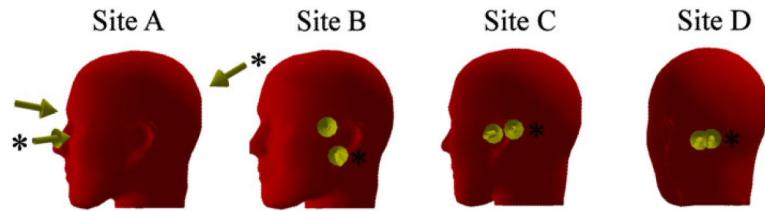


FIGURE 4.

The average absolute difference between impact location estimated by the instrumented helmet and HIII for all impacts was $31.2 \pm 46.3^\circ$ for all primary impact sites (A, B, C, and D). Arrows shown on the headforms above, identify the average impact location by both systems for each impact sites. A * is placed to the left of the arrow indicating impact location estimated by the instrumented helmet. Six of the twelve impacts to site A were identified by the instrumented helmet as being symmetrically opposite of the actual location, and so two arrows are provided at that location for the instrumented helmet to differentiate those events.

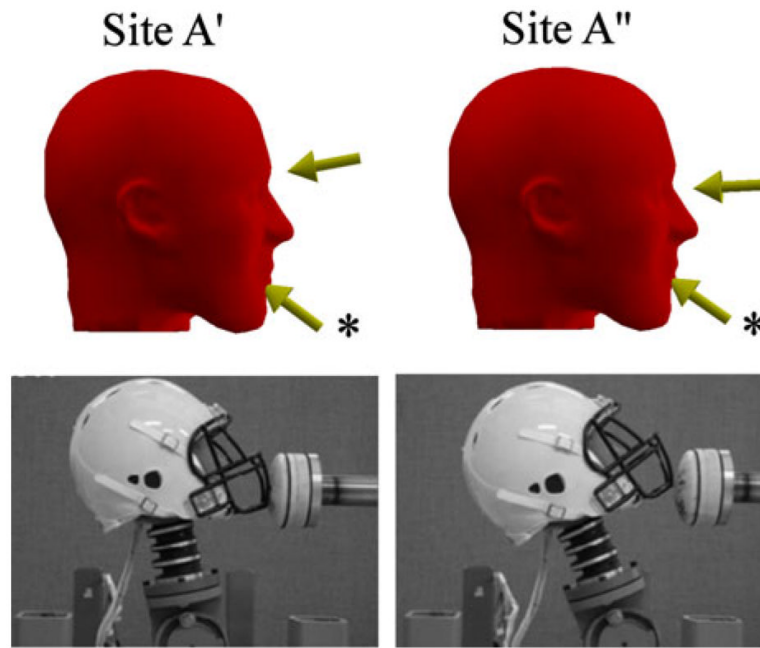


FIGURE 5.

The HIT System's processing algorithm to estimate impact location best approximates the impact site on the helmet. The average location difference between the HIT System and HIII for sites A' and A'' was $42.1 \pm 13.4^\circ$; however, the HIT System correctly identified the impact location as being low on the facemask, while the estimate of impact location from HIII was directly through the nose.

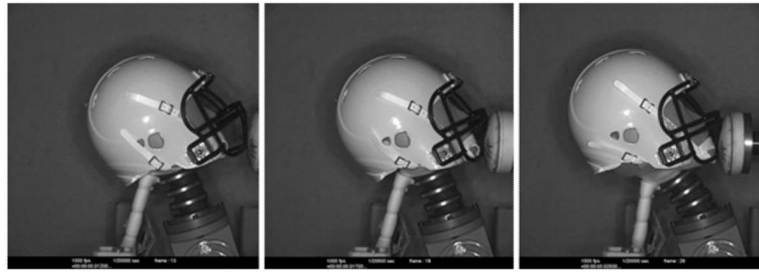


FIGURE 6.

High speed video of a 9 m/s impact at the A'' site. Top Row: (1) Initial contact occurs, (2) 1 ms after contact—facemask compresses, (3) 6 ms after contact—facemask is bent, helmet has lifted off the head, and the head begins to accelerate. Bottom Row: (4) 10 ms after contact—impactor surface pushes the facemask into the chin, (5) 16 ms after contact—the helmet experiences maximum deflection off the head.

TABLE 1

Kinematic parameters were recorded from instrumented helmets and a Hybrid III ATD following 54 linear impactor tests to correlate output measures from on-field and laboratory measurement systems.

| Target velocity (m/s) | Number of trials per primary impact site | | | |
|-----------------------|--|----|----|----|
| | A | B | C | D |
| 4.4 | 3 | 3 | 3 | 4 |
| 7.4 | 3 | 3 | 3 | 4 |
| 9.3 | 3 | 3 | 3 | 5 |
| 11.2 | 3 | 3 | 3 | 5 |
| Total trials | 12 | 12 | 12 | 18 |

Test conditions are representative of sites and velocities previously identified as being most associated with concussion in the NFL. The number of trials conducted by impact site and target impactor velocity is provided.

TABLE 2

Results from linear regression analysis between HIT System and HIII impact measures.

| | A | | B | | C | | D | | Overall A-B-C-D | |
|-------------------|------------------------|----------------|-----------|----------------|-----------|----------------|-----------|----------------|------------------------|----------------|
| | m | r ² | m | r ² | m | r ² | m | r ² | m | r ² |
| Linear Acc. | 1.055 | 0.930 | 0.995 | 0.822 | 1.084 | 0.987 | 0.969 | 0.875 | 1.009 | 0.903 |
| Rotational Acc. | 0.549 | 0.415 | 0.917 | 0.961 | 1.170 | 0.981 | 1.125 | 0.710 | 0.939 | 0.528 |
| GSI | 0.457 | 0.701 | 0.781 | 0.996 | 1.232 | 0.947 | 1.004 | 0.880 | 0.948 | 0.846 |
| HIC ₁₅ | 0.409 | 0.609 | 0.782 | 0.987 | 1.505 | 0.932 | 1.109 | 0.840 | 1.061 | 0.787 |
| | Ave (deg) ^a | Std (deg) | Ave (deg) | Std (deg) | Ave (deg) | Std (deg) | Ave (deg) | Std (deg) | Ave (deg) ^b | Std (deg) |
| Acc. direction | 94.5 | 67.6 | 18.3 | 9.1 | 13.2 | 1.9 | 9.7 | 3.1 | 31.2 | 46.3 |

The relationship between measures (m) and coefficient of determination (r²) are shown for each measure at all impact sites that most frequently relate to mTBI. The average angle difference (degrees) between the direction of head acceleration calculated from each system is also shown by impact site and for all events.

Bold values highlight the overall relationship between the two systems whereas the standard font is for within-impact-site comparisons.

^aHIT System measure was directionally opposite for 6 of 12 impacts to site A.

^bOverall error between directions of acceleration for sites B-D was 13.2 (6.3).

TABLE 3

Results from linear regression analysis between HIT system and HIII impact measures.

| | A' | | A'' | | Overall A'-A'' | |
|-------------------|-----------|----------------|-----------|------------------|----------------|----------------|
| | m | r ² | m | r ² | m | r ² |
| Linear Acc. | 2.918 | 0.928 | 2.419 | N/A ^a | 2.673 | 0.556 |
| Rotational Acc. | 2.234 | 0.883 | 1.671 | 0.308 | 1.927 | 0.438 |
| GSI | 5.400 | 0.918 | 5.268 | 0.270 | 5.341 | 0.609 |
| HIC ₁₅ | 4.360 | 0.974 | 3.147 | N/A ^a | 3.839 | 0.649 |
| | Ave (deg) | SID (deg) | Ave (deg) | SID (deg) | Ave (deg) | Std (deg) |
| Acc. Direction | 47.5 | 6.9 | 38.2 | 16.2 | 42.6 | 13.2 |

The relationship between measures (m) and coefficient of determination (r²) are shown for two sites (A' and A'') evaluated to determine HIT System performance during extreme helmet deflection.

Bold values highlight the overall relationship between the two systems whereas the standard font is for within-impact-site comparisons.

^a Coefficients of determination were indeterminate due to poor correlation.

TABLE 4

On-field data support that impacts to the lower portion of the facemask do occur; however, a comparison of median and top 95th percentile levels of linear and rotational accelerations by elevation does not support the A' and A'' laboratory test results, which suggests that the HIT System will consistently measure low facemask impacts by a factor of 2–3 times higher than those of mid facemask and helmet shell impacts.

| Category of front impacts | Elevation of impacts (deg) | % of all impacts (<i>n</i> = 286,636) | Peak linear Acc. (g) | | Peak rotational Acc. (rad/s ²) | |
|---------------------------|----------------------------|---|----------------------|----------|--|----------|
| | | | Median | Top 95th | Median | Top 95th |
| Forehead | 65 to 40 | 9.6 | 21.1 | 64.7 | 1,437 | 4,428 |
| Mid-facemask | 40 to -20 | 18.7 | 20.7 | 63.7 | 1,413 | 4,395 |
| Low-facemask | <-20 | 13.3 | 20.8 | 64.4 | 1,428 | 4,440 |