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Head Impact Severity Measures for Evaluating Mild Traumatic Brain Injury Risk Exposure

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Abstract

Objective—To quantify sensitivity of various biomechanical measures of head impact (linear acceleration, rotational acceleration, impact duration, impact location) to clinical diagnosis of concussion in American football players and to develop a novel measure of head impact severity which combines these measures into a single score that better predicts the incidence of concussion.

Methods—On-field head impact data were collected from 449 football players at 13 organizations (n = 289,916) using in-helmet systems of six single axis accelerometers.^{1,2,3,4,5} Concussions were diagnosed by medical staff and later associated with impact data. Principal Component Analysis^{6,7} and a weighting coefficient based on impact location were used to transform correlated head impact measures into a new composite variable (wPCS). The predictive power of linear acceleration, rotational acceleration, Head Injury Criteria, and wPCS was quantified using Receiver Operating Characteristic^{8,9,10} curves. The null hypothesis that a measure was no more predictive than guessing was tested ($\alpha=0.05$). Additionally, ROC curves for wPCS and classical measures were directly compared to test the hypothesis that wPCS was more predictive of concussion than classic measures ($\alpha=0.05$).

Results—When all impacts were considered, every biomechanical measure evaluated was statistically more predictive of concussion than guessing ($p < 0.005$). However, for the top 1% and 2% of impacts based on linear acceleration, a subset that consisted of 82% of all diagnosed concussions, only wPCS was significantly more predictive of concussion than guessing ($p < 0.03$), and, when compared to each other, wPCS was more predictive of concussion than classical measures for the top 1% and 2% of all data ($p < 0.04$).

Conclusions—A weighted combination of several biomechanical inputs, including impact location, is more predictive of concussion than a single biomechanical measure. This study is the first to quantify improvements in the sensitivity of a biomechanical measure to incidence of concussion when impact location is considered.

Keywords

Concussion; Football; Head impact tolerance; Head impact biomechanics; Mild traumatic brain injury; Sports injury prevention

Introduction

Head injuries in sports are occurring at an epidemic level. The Center for Disease Control and Prevention estimates that as many as 300,000 sports-related mild traumatic brain injuries (MTBI), also referred to as concussions, occur in the United States each year, with approximately 1/3 of these occurring in football.^{11,12} Among high school athletes 62,800 MTBI incidents are reported annually.¹³ A large review of injuries in high school athletics demonstrated that 13.3% of reported football injuries were head related.¹⁴ In recent years, MTBI has become the major concern for clinicians dealing with sport-related brain injuries.^{15,16,17} Correlating pathophysiology with clinical evaluation and treatment of injured athletes remains a challenge for practitioners and researchers alike. Early detection and accurate diagnosis are critical to the treatment of MTBI.

Brain injury following impact has been linked to increased strain in brain tissue as well as to pressure waves and gradients within the skull.^{18,19,20} While these injury mechanisms are likely related to kinetic measures of impact severity (e.g. peak linear acceleration, peak rotational acceleration, impact duration), it remains unclear if any single biomechanical measure is well correlated with the occurrence of MTBI.

Several injury tolerance criteria exist that associate linear head acceleration with severe brain injury. The Wayne State Tolerance Curve (WSTC), which was developed to better understand head injury acceleration tolerance in automotive crashes, defines an injury threshold using a linear acceleration versus impact time duration curve (Figure 1).²¹ In this model, the onset of skull fracture (rather than brain injury) is the injury criterion or curve boundary. Several impact severity measures based on weighted integrals of acceleration-time profiles, Gadd Severity Index (GSI)²² and Head Injury Criteria (HIC)^{23,24} were expansions of this work (Appendix A). Each of these injury tolerance curves, developed from animal and cadaveric data, is specifically limited to severe brain injury.²⁵ Impact measures specific to MTBI are lacking. Many researchers speculate that rotational acceleration of the head leads to strain on the brain tissue and therefore it may be the underlying mechanism for MTBI.^{26,27,28,29} Other data suggest that impact location is a key factor in evaluating susceptibility to MTBI. As early as 1983, Hodgson³⁰ suggested that lateral impacts were the most likely to lead to concussion. Several studies have demonstrated lower head impact tolerance for lateral translational impacts than anterior-posterior or axial (top of the head) impacts in humans, using finite element analysis.^{31,32,33} More recently, Delaney, et al.³⁴ prospectively analyzed 69 concussions to collegiate level football, hockey, and soccer players and reported that impacts to the side/temporal region were the most probable to result in concussion.

Several head acceleration tolerance criteria designed specifically for assessing concussion risk have been derived from indirect measures of head accelerations via laboratory simulations with Hybrid III anthropomorphic test dummies (ATD). Newman et al.^{35,36} proposed a new tolerance index, Head Impact Power, based on linear and rotational acceleration of the head during impact and on impact duration. Computation of this tolerance index requires inertial measurements (mass, density and geometry) of the head, which must be estimated and generalized for the target population. Head Impact Power, estimated through ATD reconstructions, was correlated with MTBI in a group of 24 professional football players representing 12 impacts (struck and striking players). Pellman et al.³⁷ performed a similar evaluation of impacts from National Football League (NFL) game videos, computing impact velocity and impact location from video and subsequently simulating the impacts in a laboratory using ATD. Linear and rotational head acceleration, GSI, and HIC were computed for striking and struck players from each simulated impact using a standard 3-2-2-2 accelerometer package located at the center of gravity of the ATD headform. Data from the laboratory impact reconstructions were categorized as concussive or non-concussive according

to on-field diagnosis at the time of impact. These data suggested that of the measured or computed variables, linear acceleration was the most highly correlated with the clinical diagnosis of MTBI. Zhang, et al.^{31,32} used biomechanical measures from these ATD simulations as inputs to a finite element analysis to predict brain response during impact, including intracranial pressure and brain shear stress, and concluded that the occurrence of concussion was best predicted by shear stress at the midbrain. These studies represented significant advances in the understanding of sports related MTBI. The ATD simulations in these studies were limited, however, to open field impacts because video from several angles was required for reconstruction analysis and to primarily injurious impacts that do not capture a representative sample of head impact biomechanics that occur in football.

Recently developed technology (Head Impact Telemetry (HIT) System, Simbex, Lebanon, NH; Sideline Response System, Riddell, Chicago, IL)^{1,2} has enabled on-field measurement of head acceleration and impact location for all impacts in practices and games using helmets instrumented with linear accelerometers (Analog Devices, Inc., Cambridge, MA). Results of *in vivo* head acceleration measurements have been reported for 11,604 impacts recorded at Virginia Tech (Blacksburg, VA) during the 2003 and 2004 football seasons.^{3,4} 290 impacts with peak linear accelerations > 75g were reported but only 3 (1%) of these impacts were associated with clinical diagnosis of concussion. Note that 75g represents the 41% concussion tolerance level from the NFL data³⁷. An evaluation of 54,154 impacts from the University of Oklahoma (Oklahoma City, OK) and 8,326 impacts from Casady High School (Oklahoma City, OK) highlighted differences in impact profiles between the different levels of play.⁵ 620 impacts with peak linear acceleration > 98g were reported but only 6 (1%) of those impacts were associated with clinical diagnosis of concussion. These results are not consistent with the logistic regression analysis from the NFL data where 98g represents the 74% concussion tolerance level.³⁷

A data set consisting of 289,916 impacts collected from instrumented helmets at 13 institutions (6 NCAA Division I and 7 high schools) are available for analysis and provide a unique opportunity to examine classic measures of impact severity (e.g. linear acceleration, rotational acceleration, HIC) and to evaluate how biomechanical measures of head impact correlate with concussion.

The purpose of this paper is to quantify the sensitivity of various biomechanical measures of head impact (e.g. linear acceleration, rotational acceleration, impact duration, impact location) to the clinical diagnosis of concussion in football players using the aforementioned database of head acceleration measures from impacts collected on the field. For a given impact, these biomechanical measures are interdependent and thus should not be treated as independent variables. Analytic methods, such as Principal Component Analysis (PCA),^{6,7} that account for these relationships among variables can be utilized. We hypothesize that computed indices which include multiple biomechanical measures, including impact location, will have a higher correlation with the incidence of concussion compared to individual measures alone.

Methods

On-field head impact data were collected during the 2004, 2005, and 2006 seasons[†] from 259 players at 6 NCAA Division I schools (n=190,054) and 190 players at 7 high schools (n=99,862). All players wore Riddell football helmets (Riddell, Chicago IL) instrumented with six linear accelerometers that recorded an acceleration time history of the head center of gravity (CG) for all impacts during practices and games.^{1,2} Head linear acceleration, head rotational

[†]2004 and 2005 data contains pre-season, regular season, and post season. At the time of this report, data from the 2006 collegiate post-season were not available for analysis.

acceleration, impact location, impact duration, GSI, and HIC were computed for each of these impacts and stored for analysis.^{1, 38} Concussions were diagnosed by team medical staff and were later associated with impact data by cross referencing observed impact time and observed impact location (from video when available or from sideline observations) with recorded impact times and recorded impact locations. For the purposes of this study, each impact was defined as concussive or non-concussive based on the evaluations completed by each participating institution. All concussions were graded according to the American Academy Of Neurology (AAN) guidelines.

For several classic biomechanical measures (linear acceleration, rotational acceleration, HIC), Receiver Operating Characteristic (ROC)^{8,9} curves were developed from the data set. This type of curve defines the relationship between sensitivity and 1-specificity for each biomechanical measure. Sensitivity is the percentage of all concussions that were correctly identified by the measure (i.e. “correct prediction level”) and 1-specificity is the percentage of all non-injurious impacts that were incorrectly identified as concussions by the measure (i.e. “false response rate”). Varying the value of the biomechanical measure that defined the tolerance level to concussion injury alters the relationship between the correct prediction level and the false response rate.

The biomechanical measure most sensitive to the prediction of concussion injury was defined as the measure that minimized the false response rate across a wide range of correct prediction levels (i.e. percentage of concussions correctly identified). The area under each specific ROC curve (correct prediction level vs. false response rate) is a measure of the predictive value of that specific biomechanical measure. This area represents the probability that a randomly selected concussive impact will have a higher severity score, than a randomly selected non-injurious impact. An area under the curve of 0.5 implies that a given biomechanical measure is as predictive of concussion as guessing, while area under the curve of 1.0 implies that all concussions would be predicted by that measure with a zero false response rate.

A new biomechanical measure of impact severity was developed and subjected to the same analysis as the individual biomechanical measures. Principal Component Analysis (PCA),^{6,7} a multivariate data analysis technique, was used to orthogonally transform correlated input biomechanical measures (linear acceleration, rotational acceleration, HIC and GSI) into new uncorrelated composite variables. In order to utilize PCA, the input biomechanical measures were first mean-centered and scaled by the variance of each measure. For each impact a Principal Component Score (PCS) was calculated as the sum of each composite variable weighted according to the variance explained (i.e. the eigenvalues of the covariance matrix of the mean centered and normalized data). The resultant PCS is effectively a weighted sum of linear acceleration, rotational acceleration, HIC and GSI, with objectively defined weights.

The impact data were sorted into one of four bins (top, side, front, or back) based on impact location (Figure 2). Elevation angle was defined as the angle between the projection of the impact direction vector through the estimated head center of gravity (CG) and a horizontal plane through the CG.^{1,3} Impacts occurring above 65° elevation were defined as top. The remaining impacts were divided into four equally spaced bins centered on the midsagittal and coronal planes. Impacts to the right and left side of the head were grouped together as side impacts. The distribution of recorded impact severities within each impact location bin was computed for each biomechanical measure. The top 1% of all impacts was defined as all impacts greater than or equal to the 99th percentile for a given measure, the top 2% was all impacts greater than or equal to the 98th percentile, etc.

For each impact, a weighted PCS (wPCS) was generated by multiplying PCS by a location coefficient that was based on impact location bin. These location coefficients were determined

by normalizing 99th percentile of PCS for each impact location (defined as x_n , where n=front, side, top, back). The impact location bin with the lowest PCS at the 99th percentile (defined as x_o) was assigned a coefficient of 1 and the other coefficients were determined by x_o/x_n . All coefficients were therefore ≤ 1 . To test the significance of changes in impact severity as a function of impact location, and thus to justify the use of location coefficients, a univariate analysis of variance statistic was used to compare mean impact severities by impact location bin for each biomechanical measure using SPSS (Chicago, IL) for all impacts as well as for the top 1%, 2%, and 5% of all impacts (Table 1). The significance level was set at $\alpha = 0.05$.

ROC curves were generated for each biomechanical measure (linear acceleration, rotational acceleration, HIC, wPCS) using all data as well as the top 1%, 2%, and 5% of all impacts selected by linear acceleration magnitude (SPSS, Chicago, IL). For each ROC curve the null hypothesis, that the true area under the curve = 0.5, was tested and an asymptotic significance value (p-value) was reported. Significance was set at $\alpha=0.05$. Additionally, Hanley's method for direct comparison of ROC curves³⁹ was used to test the hypothesis that wPCS was more predictive of concussion than the classic biomechanical measures ($\alpha=0.05$).

Results

A total of 17 players (11 collegiate and 6 high school), of the 449 in this study, sustained concussions that were diagnosed by medical staff. For each of these diagnosed concussions, a single significant impact was identified as the concussive event based on the collected head impact data and the onset of clinical symptoms. Of these 17 impacts, 8 were to the front, 3 to the top, 5 to the side, and 1 to the back of the head. 12 of these concussions were diagnosed as Grade 1, 3 were Grade 2, and 2 were Grade 3, according to the American Academy Of Neurology (AAN) guidelines. For the purposes of this paper concussion severity was not used as a covariate in the statistical analysis.

Linear acceleration measures exhibited the lowest false response rate (i.e. percentage of all non-injurious impacts that were incorrectly identified by the measure) among the classic biomechanical measures for correct prediction levels (i.e. percent of concussions correctly predicted) above ~70% (Figure 3). For example, at the 75% correct prediction level, linear acceleration measures resulted in a false response rate of 1.6% (n=4,755), while HIC and rotational acceleration measures resulted in false response rates of 1.9% (n=5,421) and 2.5% (n=7,161), respectively (Table 2). However, for correct prediction levels less than ~70% rotational acceleration exhibited the lowest false response rate among the classic measures (Figure 3). The threshold or impact severity associated with the 75% correct prediction level was 96g for linear acceleration, 7,235 rad/sec for rotational acceleration, and 160 for HIC (Figure 4).

Principal Component Analysis yielded the following equation for Principal Component Score (PCS):

$$PCS = 10 \cdot ((0.4718 \cdot sGSI + 0.4742 \cdot sHIC + 0.4336 \cdot sLIN + 0.2164 \cdot sROT) + 2) \quad (1)$$

where: $sX = (X - \text{mean}(X)) / (SD(X))$, LIN = linear acceleration, ROT = rotational acceleration, HIC = Head Impact Criteria, GSI = Gadd Severity Index. The offset by 2 and scaling by 10 generated PCS greater than zero and in the numerical range of the other classic measures studied. All four composite variables were utilized, capturing >99.9% of the variability in the data set. At the 50%, 75%, and 90% correct prediction levels PCS resulted in false response rates of 0.58%, 1.15%, and 3.60% respectively (Table 2).

Of the 289,916 impacts recorded, 124,817 (43.1%) were to the front of the head, 70,835 (24.4%) were to the back of the head, 56,566 (19.5%) were to the sides of the head, and 37,698 (13.0%) were to the top of the head. A univariate analysis of variance statistic demonstrated statistically significant differences in PCS by impact location for the top 1%, 2% and 5% of all impacts. Specifically, the top 1%, 2%, and 5% of all impacts to the top of the head had a higher PCS than impacts to the back of the head, which had higher PCS than impacts to the side and front. The top 1%, 2%, and 5% of all impacts to the side and front were statistically indistinguishable (Table 3).

Impact location weighting coefficients were derived based on the 99th percentile PCS for each location bin. The coefficients were 1.00, 0.95, 0.62, and 0.48 for side, front, back, and top impacts, respectively. Similar results were found when the location coefficient was based on the 95th or 98th percentiles. When PCS was weighted based on impact location, the new measure (wPCS) had the lowest false response rate of all the measures evaluated (Figure 3). Specifically, wPCS resulted in a mean (s.d.) reduction in the false response rate of 58% (9%) across correct prediction levels of 20% to 80% when compared to linear acceleration (Figure 3). There were statistically significant differences ($p < 0.05$) in the area under the ROC curve for all biomechanical measures compared to area under the ROC = 0.5 (e.g. guessing) for all impacts and for the top 5% of all impacts (based on linear acceleration) (Table 4). However, for the top 1% and 2% of all data there were statistically significant differences in the area under the ROC curve compared to area under the ROC = 0.5 for wPCS ($p < 0.01$) only. For example, when the top 1% of all impacts were considered, the probability that a randomly selected MTBI case would have a higher linear acceleration than a randomly selected non-injurious impact was 0.56, which was not statistically different than a probability of 0.5 which would be achieved by guessing if each impact was injurious or not. The same probability for wPCS was 0.75, which was statistically significant. Direct comparison of ROC curves demonstrated that wPCS was more predictive of concussion than classic measures for the top 1% and 2% of all data ($p < 0.04$), but not for the top 5% of all data ($p = 0.31$). The top 1%, 2%, and 5% of all impacts contained 11, 15, and 16 of the 17 diagnosed concussions, respectively (Table 4). Similar results were found for all analyses when the location cutoff for top impacts (elevation = 65°) was increased or decreased by 5°.

Discussion

We sought to quantify the performance of various biomechanical measures for predicting the incidence of MTBI. In light of recent research indicating that several biomechanical factors (linear acceleration, rotational acceleration, impact duration, and impact location) are linked to mechanisms of concussion^{18,26,27,28,31,32,33,34,35,36,37} and because of the on-field data collected with instrumented helmets^{3,4,5} that exceeded the estimated 75% linear acceleration threshold for concussion defined by the recent NFL study (98.9g)³⁷ but that did not result in a clinically diagnosed concussion ($n = 3,465$), we believed that a novel measure of head impact severity would be both clinically and scientifically relevant. In order to optimize this new measure with respect to predicting concussion, we sought to understand the inter-relationships among measured and computed biomechanical variables and their correlation with concussion risk.

Contrary to the recent NFL study³⁷ which concluded that linear acceleration was the best classic measure, we found that for correct prediction levels above ~70%, linear acceleration measures exhibited the lowest false response, but for correct prediction levels less than ~70% rotational acceleration exhibited the lowest false response rate among the classic measures. Additionally, the NFL study estimated that 75% of all impacts that were greater than 98.9g would result in concussion, while we collected on-field data for 3,476 impacts > 98.9g, only 11 of which (0.3%) were associated with clinical diagnosis of concussion (Figure 5). A possible explanation for

this disparity is that the NFL data were heavily weighted toward injurious impacts and were not representative of all impacts that occurred, including many impacts that did not result in a concussion.

In order to evaluate the clinical relevance of using various biomechanical measures of head impact to predict the clinical diagnosis of concussion, it is important to consider the rate of false responses (the percentage of all non-injurious impacts that were incorrectly identified by the measures). We utilized ROC curves to relate the false response rate to the correct prediction level (i.e. the percentage of all concussions that were correctly identified by the measures) as a function of various independent biomechanical measures of head impact. The measures evaluated were peak linear acceleration, peak rotational acceleration, HIC, GSI and a new measure (wPCS) which was derived through principal component analysis. We found similar results for both GSI and HIC, and report only the HIC risk curve based on a strong linear relationship between these two measures ($r^2 = 0.93$). Despite this linear relationship, it is important to include both HIC and GSI as inputs to PCA so that the new measure (wPCS) can be related back these industry specific severity measures from both the automotive industry (HIC) and sports industries (GSI). Any correlation between HIC and GSI is captured by PCA and is not reflected in the resultant head impact severity measure (wPCS). We elected to use HIC₁₅, rather than HIC₃₆, because 95% of the recorded head impacts had an impact duration between 5.5 ms and 13.7 ms, thus HIC₁₅ was sufficient to capture the characteristics of the impact while minimizing error associated with noise in the accelerometer signals outside the impact window which would tend to increase the value of HIC. HIC₁₅ has been used in previously studies of helmet head impacts in football.^{3,4,31,35,36,37}

The ROC curves demonstrate that wPCS results in a lower false response rate across a wide range of correct prediction levels compared to linear acceleration, rotational acceleration, or HIC. This new measure, wPCS, is based on PCA of linear acceleration, rotational acceleration, HIC, and GSI, weighted according to impact location. The location coefficients were derived from the 99th percentile PCS for impacts in each location bin. In other words, the data were normalized so that the 99th percentile wPCS was the same for each impact location. We elected to normalize the data based on the higher end of the impact magnitude scale because these impacts are most likely to be associated with concussion. Similar location coefficients were derived from the 98th and 95th percentiles, with no significant change in the overall analysis. Multiplying by these coefficients normalizes the data to account for variance in impact severity as a function of impact location. This is relevant because the impact locations with the highest impact magnitudes (top and back) did not have the most concussions.

Previous research has shown improved correlation between biomechanical measures of head impact severity and concussion when linear and rotational accelerations, instead of linear acceleration alone, were used as inputs to a multivariate regression.³¹ However, regression analysis may not be appropriate because linear and rotational accelerations from head impacts in football are highly correlated³⁷ and multicollinearity of input variables is a known problem with regression.⁴⁰ It is important to note that PCA is not a predictive approach. Rather, PCA is a statistical tool that is used to transform correlated input variables into new uncorrelated composite variables. These new uncorrelated composite variables can then be combined into a single score (wPCS) or evaluated independently using predictive techniques such as multivariate logistic regression. We combined the outputs of PCA into a single score (wPCS) based on the percent of the variance explained by each output. This is a method of noise reduction. The variables that capture the largest portions of the variance in the dataset are weighted highest. PCA is also unique in that it allows for visualization of impacts in a unique solution space in which trends in the data, which may have otherwise been masked by noise or inter-relationships between inputs, are highlighted.

The increased sensitivity of biomechanical measures to incidence of MTBI using wPCS is perhaps explained by the known relevance of impact duration (HIC and GSI),^{21,22,23} rotational acceleration^{26,27,28} and impact location^{30,31,32,33,34,37} on susceptibility to concussion. Several studies have utilized finite element models (FEM) to model brain acceleration during impact.^{31,32,33} Kleiven³³ reported that strain in the bridging veins, which could be correlated with occurrence and severity of subdural hematoma, varied with impact direction for pure translational and pure rotational impulses simulated with an experimentally validated FEM. Specifically, for pure translational impulses, axial (top) impacts resulted in the lowest strain levels while posterior-anterior and lateral accelerations of the head CG with respect to the skull resulted in the highest strains.

Other studies have utilized high speed video and laboratory impact reconstructions using ATD to estimate the linear and rotational accelerations achieved during impact in professional American football. Previously published data has shown that the average linear acceleration for concussive impacts to the facemask ($78 \pm 18g$) was lower than the average linear accelerations for other locations (107 to 117g).³⁷ This may be a result of the stiffness of the facemask compared to the helmet shell, interaction between the facemask and the shell, and/or physical properties of the head and neck.

The coefficients used to scale translational acceleration values, as reported by Kleiven³³, the average linear accelerations of impacts that resulted in concussions reported by Pellman³⁷, and the location coefficients presented herein all describe an effect of impact location on the likelihood of sustaining a concussion for a given impact. The effect of impact location was derived through different analytic means, but similar results were found for each study. Specifically, each study demonstrated that impacts to the top of the head were less likely to cause concussion than lateral or frontal impacts of the same impact accelerations. Top impacts may be less likely to cause concussion because the associated rotational acceleration is decreased due to the reduced lever arm of the impact vector with respect to the centers of rotation of the cervical spine or because the cervical spine absorbs impact energy thus reducing strain in the bridging veins and brain tissue. The aim of this study was not to evaluate the effect of impact location on the mechanisms of brain injury, but rather to quantify improvements in the sensitivity of biomechanical measures to the incidence of concussion when impact location is taken into account.

It is important to note that no current biomechanical measure of the severity of a single impact can predict concussions in football with a positive predictive value (PPV) (e.g. likelihood of concussion) as high as 75%, as suggested by previous research³⁷. The highest PPV for linear acceleration was 0.3% at the 50% correct prediction level. At this correct prediction level PPV for wPCS was 0.9%. It is important to note the difference between PPV or likelihood of concussion, as reported in previous research, and correct prediction level or sensitivity, as reported here. 75% PPV or likelihood of concussion at a given threshold means that on average 3 of 4 impacts above this threshold will result in concussion. 75% sensitivity or correct prediction level at a given threshold means that on average 3 of 4 concussions will result from impacts above the threshold. There are several variables that have not been accounted for in this analysis, including age, gender, and both impact and concussion histories for each individual player. For example, we noted a large variation in false response rates by player. In other words, certain players sustained high magnitude head impacts frequently and did not sustain a concussion while others sustained a head injury after relatively few and relatively minor head impacts. We also noted that for 76% of all concussions, wPCS that were associated with subsequent medical diagnosis of concussion were among the highest wPCS sustained by the injured player throughout the season (above the 99th percentile wPCS for that player). This suggests that each player may have a unique head injury tolerance that is reflected in the distribution of impact severity measures for all non-injurious impacts for that player. Future

work will include quantifying the effects of a particular player's impact history as well as the effect of severe impacts several days prior to injury. An analysis of correlation between concussion grade (severity) and biomechanical measures of head impact was beyond the scope of this study but will be addressed in future manuscripts.

There are several limitations of the current study. Regarding the development of the new measure (wPCS), all 17 concussive impacts collected on the field were used as a training dataset for creating the wPCS measure. The limited number of concussions prevented us from splitting the data into training and testing datasets. As concussion data become available for subsequent football seasons, wPCS and the location coefficients used will be further refined to improve their capability to prospectively assess concussion risk from on-field biomechanical measures of head impact. Additionally, statistical evaluation of the contribution of each independent variable to wPCS will be evaluated in an attempt to minimize the number of input variables required. For example, we will evaluate the predictive capability of a formulation of wPCS with only linear and only rotational acceleration. Regarding the data collection methods, *in vivo* measures of rotational acceleration using the instrumented helmets may be inexact because rotation was assumed to happen about a fixed point in the neck, limiting measurements to two dimensions. Specifically, flexion/extension (rotation about the y-axis, Figure 2) and lateral bending (rotation about the x-axis, Figure 2) are estimated based the geometrical properties of the head while axial rotation (rotation about the z-axis, Figure 2) is not. However, these measurements are still representative of actual rotational acceleration.¹ The algorithm used to compute head acceleration has recently been expanded to include all six degrees of freedom³⁸ and has been validated for field use in football. A second limitation of the data collection methods is that the instrumentation technology is only currently available for Riddell helmets, limiting any evaluation of the effect of helmet type on the measures of impact severity. Third, this study was limited to male athletes. Future studies will be expanded to include both male and female athletes in soccer, hockey and boxing. Fourth, as with any in-vivo study of head injury, it is possible that players sustained minor head injuries that were never reported or diagnosed. Finally, grouping data into distinct bins based on impact region may not be optimal. Future work will involve the development of a continuous variable for impact location risk that will be used in place of the location coefficients, which are currently based on location bins.

We do not advocate the use of measures of impact severity and/or impact history as a diagnostic tool for concussion. Concussions are difficult to diagnose and visual symptoms checks and direct medical attention by qualified personnel is required for both diagnosis and treatment.⁴¹ Nevertheless, the data described herein show strong correlations between biomechanical measures and concussion diagnosis. A widespread study which correlates individual impact history with changes in neurocognitive test results, neurological changes as seen via imaging methods including, but not limited to, functional magnetic resonance imaging (fMRI), and the occurrence of clinically diagnosed MTBI is recommended to refine this new impact tolerance measure (wPCS), which specifically addresses MTBI, as well as to quantify the effects of cumulative impacts and individual impact histories on susceptibility to concussion.

Conclusion

This work provides evidence that a single biomechanical measure, such as linear acceleration, is not the most sensitive biomechanical measure for determining concussion risk. A composite variable that contains aspects of linear acceleration, rotational acceleration, impact duration, and impact location is more sensitive to incidence of MTBI. This study is the first to quantify the sensitivity and specificity of an impact location based head impact severity measure for predicting concussion. We use PCA and a weighting system based on impact location to define a new measure (wPCS). This method accounts for the multicollinearity of the input variables

(linear acceleration, rotational acceleration, HIC, and GSI) by transforming the data into new uncorrelated composite variables which can then be combined into a single composite measure. Using this new head impact severity measure (wPCS), the rate of false responses was reduced at all correct prediction levels as compared to the classic biomechanical measures evaluated (linear acceleration, rotational acceleration, and HIC).

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Appendix A: Equations for Head Injury Criterion (HIC) and Gadd Severity Index (GSI)

$$HIC^* = (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right]^{5/2} \quad GSI = \int_0^T a(t)^{2.5} dt$$

where $a(t)$ = linear acceleration of the head center of gravity

$t_2 - t_1 \leq 15$ ms

* HIC can be calculated over any duration of time. Throughout this paper HIC is calculated over a 15 ms window.

Figure 1.

The Wayne State Tolerance Curve²¹ defines an injury threshold based on the onset of skull fracture which was thought to correlate with moderate to severe concussion.

Figure 2.

Impacts were grouped into 5 bins based on impact location which is defined by azimuth (θ) and elevation (α). Azimuth (θ) is defined from -180° to 180° with 0° at the X axis and positive (θ) to the right side of the head. Elevation (α) is defined from 0° (horizontal plane passing through the head center of gravity, CG) to 90° (crown of the head at the Z axis). The XZ plane represents the midsagittal plane with the positive X corresponding to the caudal direction. The XY plane represents the coronal plane with positive Y to the right side of the head. Impacts with elevation > 65 were defined as top. The remaining impacts were grouped based on azimuth; -45° to 45° were back, $\pm 45^\circ$ to $\pm 135^\circ$ were side, and -135° to 135° were front.

Figure 3.

Receiver Operator Characteristic (ROC) curves demonstrate the sensitivity of the independent measures (e.g. linear acceleration, rotational acceleration, HIC, weighted PCS) to the incidence of concussion. Each point on a curve represents a possible concussion threshold (for a given head impact severity measure) for which the correct prediction level and false response rate are calculated. The x-axis in the figure shown is zoomed in to the clinically relevant portion of the ROC curves. Each curve begins at (0,0) and ends at (1,1).

Figure 4.

Values of the biomechanical measures that are associated with each correct prediction level (i.e. percentage of concussions correctly predicted).

Figure 5.

Linear acceleration concussion probability function generated from NFL impacts reconstructed in the laboratory, as in Pellman³⁷, shown with 17 concussive impacts recorded *in-vivo* and a random sample of 100 controls (non-injurious impacts). In total there were 289,899 controls of which 3,476 were > 98.9g (the 75% concussion probability level based on NFL data).

Table 1

Cutoff values for the top 1%, 2%, and 5% for each classic biomechanical measure

	Linear Acceleration	Rotational Acceleration	HIC
Top 1%*	103.4 g	6990.5 rad/s ²	206.0
Top 2%	86.4 g	5805.1 rad/s ²	134.5
Top 5%	65.5 g	4368.3 rad/s ²	67.9

* Cutoff values for the top X% (i.e. Top 1% = 99th percentile of all data for a given measure)

Table 2

Differences in the false response rate for the classic biomechanical measures at the 50%, 75% and 90% correct prediction levels.

Correct Prediction Level	Classic Biomechanical Measures				Principal Component Based Measures			
	Linear Acceleration		Rotational Acceleration		HIC		PCS	
	False Response Rate	N	False Response Rate	N	False Response Rate	N	False Response Rate	N
50%	0.85%	2464	0.48%	1392	0.71%	2058	0.58%	1687
75%	1.64%	4755	2.47%	7161	1.87%	5421	1.51%	4389
90%	3.26%	9451	6.40%	47546	3.15%	9132	3.60%	10437

* percentage of all injurious impacts that were correctly identified by the measure

**

percentage of all non-injurious impacts that were incorrectly identified by the measure at a given correct prediction level

number of false responses at a given correct prediction level for a given measure

Table 3

Differences in the distribution of PCS as a function of impact location.

	Side			Front			Back			Top			Significance
	n	cutoff	mean PCS	n	cutoff	mean PCS	n	cutoff	mean PCS	n	cutoff	mean PCS	
Top 1%	566	56.165	90.7	1248	59.318	85.9	708	90.326	143.9	377	116.36	211.5	P<0.001
Top 2%	1131	44.354	70	2496	48.177	69.5	1417	67.63	110.4	754	86.8	155.5	P<0.001
Top 5%	2828	32.561	50.4	6241	36.621	52.6	3542	45.181	76.6	1885	59.337	104.3	P<0.001

Table 4

ROC curves were generated for all data as well as for the top 1%, 2%, and 5% of impact severity for all data based on linear acceleration.

	Linear Acceleration			Rotational Acceleration			HIC			wPCS			Number of Concussions
	Area under ROC curve ^a	p-value ^b	Area under ROC curve ^a	p-value ^b	Area under ROC curve ^a	p-value ^b	Area under ROC curve ^a	p-value ^b	Area under ROC curve ^a	p-value ^b	Area under ROC curve ^a	p-value ^b	
Top 1%	0.562	0.478	0.661	0.065	0.623	0.157	0.749	0.004*	0.749	0.004*	0.749	0.004*	11
Top 2%	0.637	0.066	0.689	0.074	0.651	0.078	0.764	<0.001*	0.764	<0.001*	0.764	<0.001*	15
Top 5%	0.834	<0.001*	0.768	<0.001*	0.816	<0.001*	0.854	<0.001*	0.854	<0.001*	0.854	<0.001*	16
All Data	0.987	<0.001*	0.94	<0.001*	0.987	<0.001*	0.976	<0.001*	0.976	<0.001*	0.976	<0.001*	17

^aThe area under the ROC curve represents the probability that a randomly MTBI case is higher on the given severity scale than a randomly selected non-injurious impact

^bNull hypothesis: true area under the ROC curve = 0.5 (this would occur if an arbitrary measure such as guessing was used)