HEAD IMPACT SEVERITY MEASURES FOR EVALUATING MILD TRAUMATIC BRAIN INJURY RISK EXPOSURE

Richard M. Greenwald, Ph.D.
Simbex,
Lebanon, New Hampshire

Joseph T. Gwin, B.S.
Simbex,
Lebanon, New Hampshire

Jeffrey J. Chu, M.S.
Simbex,
Lebanon, New Hampshire

Joseph J. Crisco, Ph.D
Department of Orthopaedics,
Brown Medical School,
Rhode Island Hospital,
Providence, Rhode Island

Reprint requests:
Richard M. Greenwald, Ph.D.,
Simbex,
10 Water Street,
Suite 410,
Lebanon, NH 03766.
Email: rgreenwald@simbex.com

Received, September 10, 2007.
Accepted, January 3, 2008.

OBJECTIVE: The aims of this study were to quantify the sensitivity of various biomechanical measures (linear acceleration, rotational acceleration, impact duration, and impact location) of head impact to the clinical diagnosis of concussion in United States football players and to develop a novel measure of head impact severity combining 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. Concussions were diagnosed by medical staff and later associated with impact data. Principal component analysis and a weighting coefficient based on impact location were used to transform correlated head impact measures into a new composite variable, weighted principal component score (wPCS). The predictive power of linear acceleration, rotational acceleration, head injury criterion, and wPCS was quantified using receiver operating characteristic curves. The null hypothesis, that a measure was no more predictive than guessing, was tested ($\alpha = 0.05$). In addition, receiver operating characteristic curves for wPCS and classical measures were directly compared to test the hypothesis that wPCS was more predictive of concussion than were classic measures ($\alpha = 0.05$).

RESULTS: When all of the 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 ($P < 0.03$); when compared with each other, wPCS was more predictive of concussion than were classical measures for the top 1 and 2% of all of the data ($P < 0.04$).

CONCLUSION: 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 the authors' knowledge to quantify improvements in the sensitivity of a biomechanical measure to incidence of concussion when impact location is considered.

KEY WORDS: Concussion, Football, Head impact biomechanics, Head impact tolerance, Mild traumatic brain injury, Sports injury prevention

Neurosurgery 62:789–798, 2008 DOI: 10.1227/01.NEU.0000311244.05104.96 www.neurosurgery-online.com

Head injuries in sports are occurring at an epidemic level. The Centers for Disease Control and Prevention estimates that as many as 300,000 sports-related mild traumatic brain injuries (MTBIs), also referred to as concussions, occur in the United States each year, with approximately one-third of these occurring during football games (4, 37). Among high school athletes, 62,800 MTBI incidents are reported annually (34). A large review of injuries in high school athletics demonstrated that 13.3% of reported football injuries were related to the head (33). In recent years, MTBI has become the major concern for clinicians dealing with sports-related brain injuries (1, 6, 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 after head impact has been linked to increased strain in brain tissue and to pressure waves and gradients within the cranium (2, 24, 29). Although these injury mecha-

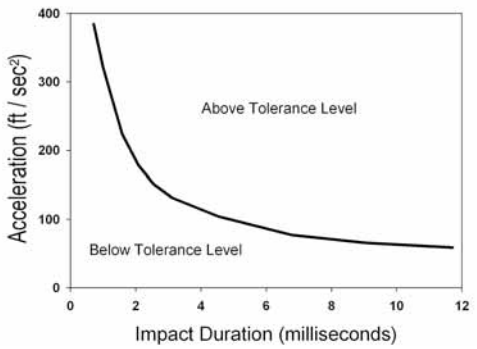


FIGURE 1. *The Wayne State Tolerance Curve (14) defines an injury threshold based on the onset of cranial fracture, which was thought to correlate with moderate to severe concussion.*

nisms are likely related to kinetic measures of impact severity (e.g., peak linear acceleration, peak rotational acceleration, and impact duration), it remains unclear whether 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, which was developed to better understand head injury acceleration tolerance in automotive crashes, defines an injury threshold using a linear acceleration versus impact duration curve (*Fig. 1*) (14). In this model, the onset of cranial 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) (13) and head injury criterion (HIC) (11, 39) were expansions of this work (*Appendix 1*). Each of these injury tolerance curves, developed from animal and cadaveric data, is specifically limited to severe brain injury (35). 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 (9, 15, 30, 31). Other data suggest that impact location is a key factor in evaluating susceptibility to MTBI. As early as 1983, Hodgson et al. (20) suggested that lateral impacts were the most likely to lead to concussion. Several studies using finite element analysis (23, 40, 41) have demonstrated lower head impact tolerance for lateral translational impacts than for anteroposterior or axial (top of the head) impacts in humans. More recently, Delaney et al. (8) prospectively analyzed 69 concussions in college-level football, hockey, and soccer players and reported that impacts to the side/temporal region were the most likely 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 (Denton ATD, Inc., Milan, OH). Newman et al. (27, 28) 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. (32) 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 nonconcussive 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. (40, 41) 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, which do not capture a representative sample of head impact biomechanics that occur in football.

Recently developed technology (e.g., head impact telemetry [HIT] system; Simbex, Lebanon, NH, and sideline response system; Riddell, Chicago, IL) (7, 25) has enabled on-field measurement of head acceleration and impact location for all impacts in practices and games using helmets equipped 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 Polytechnic Institute and State University (Blacksburg, VA) during the 2003 and 2004 football seasons (3, 10). A total of 290 impacts with peak linear accelerations of greater than 75*g* were reported, but only three (1%) of these impacts were associated with clinical diagnosis of concussion. Note that 75*g* represents the 41% concussion tolerance level from the NFL data (32). An evaluation of 54,154 impacts from the University of Oklahoma, Oklahoma City, and 8326 impacts from Casady High School in Oklahoma City highlighted differences in impact profiles between the different levels of play (36). A total of 620 impacts with peak linear acceleration of greater than 98*g* were reported, but only six (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, in which 98*g* represents the 74% concussion tolerance level (32).

A data set consisting of 289,916 impacts collected from the National Collegiate Athletic Association instrumented helmets at 13 institutions (six Division I colleges and seven high schools) is available for analysis, providing 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 article is to quantify the sensitivity of various biomechanical measures of head impact (e.g., linear acceleration, rotational acceleration, impact duration, and 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) (21, 22), that account for these relationships among variables can be used. We hypothesize that computed indices that include multiple biomechanical measures, including impact location, will have a stronger correlation with the incidence of concussion compared with individual measures alone.

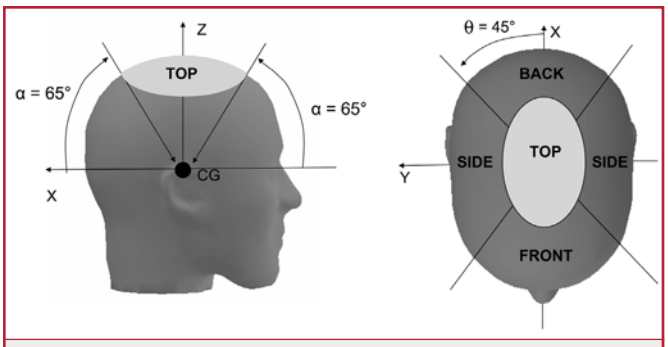


FIGURE 2. *Impacts were grouped into four bins based on impact location, which is defined by azimuth (θ) and elevation (α). Azimuth (θ) is defined from −180 degrees to 180 degrees with 0 degrees at the x axis and positive (θ) to the right side of the head. Elevation (α) is defined from 0 degrees (horizontal plane passing through the head center of gravity, CG) to 90 degrees (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 higher than 65 degrees were defined as top. The remaining impacts were grouped based on azimuth; −45 degrees to 45 degrees were back, ±45 degrees to ±135 degrees were side, and −135 degrees to 135 degrees were front.*

PATIENTS AND METHODS

On-field head impact data were collected during the 2004, 2005, and 2006 seasons (exclusive of 2006 post-season data) from 259 players at six NCAA Division I schools (n = 190,054) and 190 players at seven high schools (n = 99,862). All of the players wore Riddell football helmets instrumented with six linear accelerometers that recorded an acceleration time history of the head center of gravity (CG) for all of the impacts during practices and games (7, 25). Head linear acceleration, head rotational acceleration, impact location, impact duration, GSI, and HIC were computed for each of these impacts and stored for analysis (5, 7). 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 nonconcussive based on the evaluations completed by each participating institution. All of the concussions were graded according to the American Academy of Neurology guidelines.

For several classic biomechanical measures (linear acceleration, rotational acceleration, and HIC), receiver operating characteristic (ROC) (12, 38) 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 correctly identified by the measure (i.e., correct prediction level), and 1-specificity is the percentage of all noninjurious impacts 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 that was 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 versus 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 noninjurious impact. An area under the curve of 0.5 implies that a given biomechanical measure is as predictive of concussion as guessing, whereas an area under the curve of 1.0 implies that all of the 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. PCA (21, 22), a multivariate data analysis technique, was used for orthogonal transformation of correlated input biomechanical measures (linear acceleration, rotational acceleration, HIC, and GSI) into new, uncorrelated composite variables. To use 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 four bins based on impact location (top, side, front, or back) (*Fig. 2*). Elevation angle was defined as the angle between the projection of the impact direction vector through the estimated head CG and a horizontal plane through the CG (7, 10). Impacts occurring above 65 degrees 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 sides 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, and so forth.

For each impact, a weighted PCS (wPCS) was generated by multiplying PCS by a location coefficient based on the impact location bin. These location coefficients were determined by normalizing the 99th

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

	Linear acceleration, g	Rotational acceleration, rad/s^2	Head injury criterion
Top 1%[a]	103.4	6990.5	206.0
Top 2%	86.4	5805.	134.5
Top 5%	65.5	4368.3	67.9

[a] Top 1% = 99th percentile of all data for a given measure; top 2% = 98th percentile; top 5% = 95th percentile.

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 of the coefficients were, therefore, 1 or less. 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 was used to compare mean impact severities by impact location bin for each biomechanical measure using SPSS v13.0 software (SPSS, Chicago, IL) for all of the impacts and 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 of the data as well as the top 1, 2, and 5% of all of the impacts selected by linear acceleration magnitude (SPSS). 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 $\alpha = 0.05$. In addition, Hanley's method for direct comparison of ROC curves (18, 19) was used to test the hypothesis that wPCS was more predictive of concussion than were the classic biomechanical measures ($\alpha = 0.05$).

RESULTS

A total of 17 players (11 college and six 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, eight were to the front, three were to the top, five were to the side, and one was to the back of the head. Twelve of these concussions were diagnosed as Grade 1, three were Grade 2, and two were Grade 3, according to American Academy of Neurology guidelines. For the purposes of this article, concussion severity was not used as a covariate in the statistical analysis.

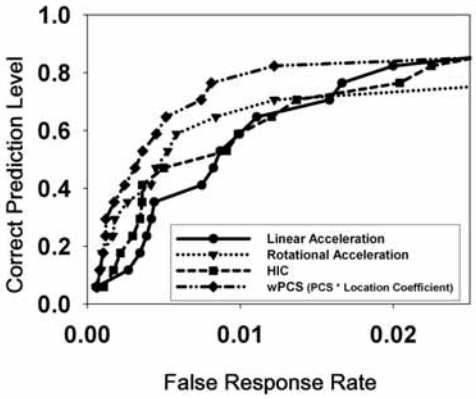


FIGURE 3. *Receiver operating characteristic (ROC) curves demonstrate the sensitivity of the independent measures of head impact severity (linear acceleration, rotational acceleration, head injury criterion (HIC), and weighted principal component score) to the incidence of concussion. Each point on a given curve represents a possible concussion threshold for which the correct prediction level and false response rate are calculated. The x axis is zoomed in to the clinically relevant portion of the ROC curves. Each curve begins at (0, 0) and ends at (1, 1).*

Linear acceleration measures exhibited the lowest false response rate (i.e., percentage of all of the noninjurious impacts that were incorrectly identified by the measure) among the classic biomechanical measures for correct prediction levels (i.e., percentage of concussions that were correctly predicted) above ~70% (*Fig. 3*). For example, at the 75% correct prediction level, linear acceleration measures resulted in a false response rate of 1.6% (n = 4755), whereas HIC and rotational acceleration measures resulted in false response rates of 1.9% (n = 5421) and 2.5% (n = 7161), respectively (*Table 2*). However, for correct prediction levels that were less

TABLE 2. Differences in false response rates for all measures at the 50, 75, and 90% correct prediction levels

	Classic biomechanical measures			Principal component-based measures	
	Linear acceleration, % (no.)[b]	Rotational acceleration, % (no.)	Head injury criterion, % (no.)	Principal component score, % (no.)	Weighted principal component score, % (no.)
50% correct prediction[a]	0.85 (2464)	0.48 (1392)	0.71 (2058)	0.58 (1687)	0.34 (977)
75% correct prediction	1.64 (4755)	2.47 (7161)	1.87 (5421)	1.51 (4389)	0.78 (258)
90% correct prediction	3.26 (9451)	16.40 (47,546)	3.15 (9132)	3.60 (10,437)	3.52 (10,205)

[a] Percentage of all of the injurious impacts that were identified correctly by the measure.
[b] Percentage of all of the noninjurious impacts that were identified incorrectly by the measure (false response rate) and number of false responses at a given prediction level.

than ~70%, rotational acceleration exhibited the lowest false response rate among the classic measures (*Fig. 3*). The threshold or impact severity associated with the 75% correct prediction level was 96*g* for linear acceleration, 7235 rad/second for rotational acceleration, and 160 for HIC (*Fig. 4*).

The PCA yielded the following equation (7) for PCS:

$$PCS = 10 \times ([0.4718 \times sGSI + 0.4742 \times sHIC + 0.4336 \times sLIN + 0.2164 \times sROT] + 2)$$

where sX = (X-mean[X])/(SD[X]), LIN = linear acceleration, ROT = rotational acceleration, HIC = head injury criterion, and GSI = Gadd severity index. The offset by 2 and scaling by 10 generated PCS greater than 0 and in the numerical range of the other classic measures studied. All four composite variables were used, capturing more than 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 demonstrated statistically significant differences in PCS by impact location for the top 1, 2, and 5% of all of the 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 a 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 of the the measures evaluated (*Fig. 3*). Specifically, wPCS resulted in a mean (SD) reduction in the false response rate of 58% (9%) across correct prediction levels of 20 to 80% when compared with linear acceleration (*Fig. 3*). There were statistically significant differences ($P < 0.05$) in the area under the ROC curve for all of the biomechanical measures compared with the area under the ROC = 0.5 (i.e., guessing) for all of the impacts and for the top 5% of all of the impacts, based on linear acceleration (*Table 4*). However, for the top 1 and 2% of all of the data there were statistically significant differences in the area under the ROC curve compared with the area under the ROC = 0.5 for wPCS ($P < 0.01$) only. For example, when the top 1% of all of the impacts were considered, the probability that a randomly selected MTBI case would have a higher linear acceleration than a randomly selected noninjurious impact was 0.56, which was not statistically different from a probability of 0.5, which would be achieved by guessing whether 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 were classic measures for the top 1 and 2% of all of the data ($P < 0.04$), but not for the top 5% of all of the data ($P = 0.31$). The top 1, 2, and 5% of all of the impacts contained 11, 15, and 16 of the 17 diagnosed concussions, respectively (*Table 4*). Similar results were found for all of the analyses when the location cutoff for the top impacts (elevation = 65 degrees) was increased or decreased by 5 degrees.

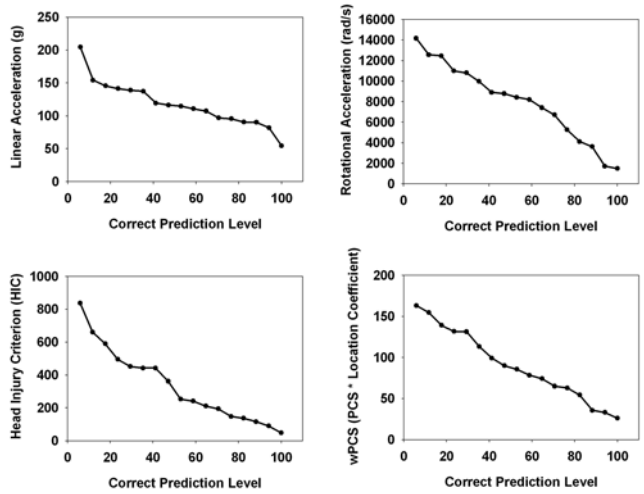


FIGURE 4. *Values of the biomechanical measures associated with each correct prediction level (i.e., percentage of concussions correctly predicted).*

TABLE 3. Differences in the distribution of principal component score as a function of impact location[a]

	Side		Front		Back		Top		*P* value
	No.	PCS cutoff (mean)	No.	PCS cutoff (mean)	No.	PCS cutoff (mean)	No.	PCS cutoff (mean)	
Top 1%	566	56.165 (90.7)	1248	59.318 (85.9)	708	90.326 (143.9)	377	116.36 (211.5)	<0.001
Top 2%	1131	44.354 (70.0)	2496	48.177 (69.5)	1417	67.63 (110.4)	754	86.8 (155.5)	<0.001
Top 5%	2828	32.561 (50.4)	6241	36.621 (52.6)	3542	45.181 (76.6)	1885	59.337 (104.3)	<0.001

[a] PCS, principal component score.

TABLE 4. Receiver operating characteristic curves of impact severity for all data based on linear acceleration[a]

	Linear acceleration		Rotational acceleration		HIC		wPCS		Concussions, no.
	Area under ROC curve[b]	*P* value[c]	Area under ROC curve[b]	*P* value[c]	Area under ROC curve[b]	*P* value[c]	Area under ROC curve[b]	*P* value[c]	
Top 1%	0.562	0.478	0.661	0.065	0.623	0.157	0.749	0.004[d]	11
Top 2%	0.637	0.066	0.689	0.074	0.651	0.078	0.764	<0.001[d]	15
Top 5%	0.834	<0.001[d]	0.768	<0.001[d]	0.816	<0.001[d]	0.854	<0.001[d]	16
All data	0.987	<0.001[d]	0.94	<0.001[d]	0.987	<0.001[d]	0.976	<0.001[d]	17

[a] ROC, receiver operating characteristic; HIC, head injury criterion; wPCS, weighted principal component score.
[b] The area under the ROC curve represents the probability that a randomly selected mild traumatic brain injury case is higher on the given severity scale than a randomly selected noninjurious impact.
[c] Null hypothesis: true area under the curve = 0.5; this would occur if an arbitrary measure such as guessing were used.
[d] Statistically significant.

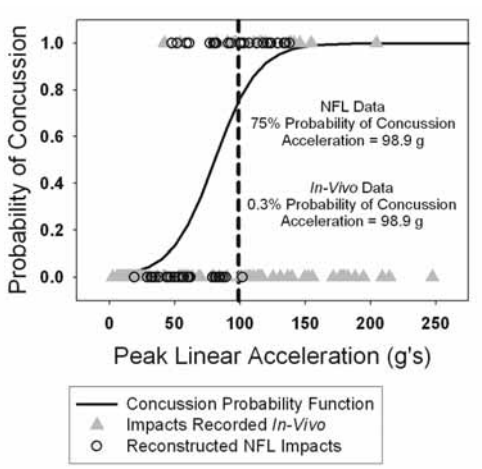


FIGURE 5. *Linear acceleration concussion probability function generated from National Football League impacts reconstructed in the laboratory, as in Pellman et al. (32), shown with a random sample of 50 noninjurious impacts and 13 concussive impacts recorded on the field using instrumented helmets.*

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 (8, 9, 23, 27, 28, 30, 31, 32, 39, 40, 41), and because of the on-field data collected with instrumented helmets (3, 10, 36) that exceeded the estimated 75% linear acceleration threshold for concussion defined by the recent NFL study (98.9*g*) (32), but did not result in a clinically diagnosed concussion (n = 3465), we believed that a novel measure of head impact severity would be both clinically and scientifically relevant. To optimize this new measure with respect to predicting concussion, we sought to understand the interrelations among measured and computed biomechanical variables and their correlation with concussion risk.

Contrary to the recent NFL study (32), 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. In addition, the NFL study estimated that 75% of all impacts greater than 98.9*g* would result in concussion, whereas we collected on-field data for 3476 impacts of greater than 98.9*g*, only 11 of which (0.3%) were associated with clinical diagnosis of concussion (*Fig. 5*). A possible explanation for this disparity is that the NFL data were weighted heavily toward injurious impacts and were not representative of all impacts that occurred, including many that did not result in a concussion.

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 of the noninjurious impacts that were incorrectly identified by the measures). We used ROC curves to relate the false response rate to the correct prediction level (i.e., the percentage of all of the 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, that was derived through PCA. We found similar results for 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 to these industry-specific severity measures from both the automotive (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 HIC_{15} rather than HIC_{36} because

95% of the recorded head impacts had an impact duration of between 5.5 and 13.7 milliseconds; thus, HIC_{15} 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. HIC_{15} has been used in previous studies of helmet head impacts in football games (3, 10, 27, 28, 32, 40).

The ROC curves demonstrate that wPCS results in a lower false response rate across a wide range of correct prediction levels as compared with linear acceleration, rotational acceleration, and HIC. 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 were the 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 (40). However, regression analysis may not be appropriate because linear and rotational accelerations from head impacts in football are highly correlated (32), and multicollinearity of input variables is a known problem with regression (26). 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 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 percentage 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 data set 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 interrelations between inputs, are highlighted.

The increased sensitivity of biomechanical measures to the incidence of MTBI using wPCS is perhaps explained by the known relevance of impact duration (HIC and GSI) (13, 14, 39), rotational acceleration (9, 30, 31), and impact location (8, 20, 23, 40, 41) on susceptibility to concussion. Several studies have used finite element models (FEM) to model brain acceleration during impact (23, 40, 41). Kleiven (23) 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, whereas posteroanterior and lateral accelerations of the head CG with respect to the cranium resulted in the highest strains.

Other studies have used high-speed video and laboratory impact reconstructions with ATD to estimate the linear and rotational accelerations achieved during impact in American professional football. Previously published data have shown that the average linear acceleration for concussive impacts to the facemask ($78 \pm 18\ g$) was lower than the average linear accelerations for other locations (107–117g) (32). This may be a result of the stiffness of the facemask compared with 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 Kleiven (23), the average linear accelerations of impacts that resulted in concussions reported by Pellman et al. (32), and the location coefficients presented herein 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 research (32). 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. A 75% PPV or likelihood of concussion at a given threshold means that on average three of four impacts above this threshold will result in concussion. A 75% sensitivity or correct prediction level at a given threshold means that on average three of four concussions will result from impacts above the threshold. There are several variables that have not been accounted for in this analysis, including age, sex, 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, whereas others sustained a head injury after relatively few and relatively minor head impacts. Also noted was that for 76% of all of the concussions, wPCSs associated with subsequent medical diagnosis of concussion were among the highest wPCSs 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 of the noninjurious 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 before 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.

The present study has several limitations. Regarding the development of the new measure (wPCS), all 17 concussive impacts collected on the field were used as a training data set for creating the wPCS measure. The limited number of concussions prevented the authors from splitting the data into training and testing data sets. As concussion data become available for subsequent football seasons, wPCS and the location coefficients used will be further refined to improve their capability to assess concussion risk prospectively from on-field biomechanical measures of head impact. In addition, 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, the predictive capability of a formulation of wPCS with only linear and only rotational acceleration will be evaluated. 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) and lateral bending (rotation about the x axis) are estimated based on the geometrical properties of the head, but axial rotation (rotation about the z axis) is not (*Fig. 2*). However, these measurements are still representative of actual rotational acceleration (7). The algorithm used to compute head acceleration has recently been expanded to include all six degrees of freedom (unpublished data) and has been validated for field use in football. A second limitation of the data collection methods is that the instrumentation technology is available only 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 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 are required for both diagnosis and treatment (16). Nevertheless, the data described herein show strong correlations between biomechanical measures and concussion diagnosis. A widespread study that 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, and the occurrence of clinically diagnosed MTBI is recommended to refine this new impact tolerance measure (wPCS). This widespread study must specifically address MTBI and 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. To the authors' knowledge, this study is the first to quantify the sensitivity and specificity of an impact location–based head impact severity measure for predicting concussion. PCA and a weighting system based on impact location were used 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 that can then be combined into a single composite measure. Using the wPCS head impact severity measure, the rate of false responses was reduced at all of the correct prediction levels as compared to the classic biomechanical measures evaluated (linear acceleration, rotational acceleration, and HIC).

Disclosure

The authors have a financial interest in the HIT System technology used to collect data for this study.

APPENDIX 1. Equations for Head Injury Criterion and Gadd Severity Index[a]

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

where a(t) = linear acceleration of the head center of gravity
$t_2 - t_1 \leq 15$ milliseconds

[a] Head injury criterion can be calculated over any duration. Throughout this article, head injury criterion is calculated over a 15-ms window.

REFERENCES

1. Aubry M, Cantu R, Dvorak J, Graf-Baumann T, Johnston K, Kelly J, Lovell M, McCrory P, Meeuwisse W, Schamasch P; Concussion in Sport Group: Summary and agreement statement of the first International Conference on Concussion in Sport, Vienna 2001. Recommendations for the improvement of safety and health of athletes who may suffer concussive injuries. **Br J Sports Med** 36:6–10, 2002.
2. Bandak FA, Eppinger RH: A three dimensional finite element analysis of the human brain under combined rotational and translational accelerations, in *Proceedings of the 38th Stapp Car Crash Conference*. Warrendale, PA, Society of Automotive Engineers, 1994, vol. 279, pp 145–163.
3. Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald R, Duma S: Analysis of linear head accelerations from collegiate football impacts. **Curr Sports Med Rep** 5:23–28, 2006.
4. Centers for Disease Control and Prevention (CDC): Sports-related recurrent brain injuries: United States. **Morb Mortal Wkly Rep** 46:224–227, 1997.
5. Chu JJ, Beckwith JG, Crisco JJ, Greenwald RM: A novel algorithm to measure linear and rotational head acceleration using single-axis accelerometers. Presented at the 5th World Congress of Biomechanics, Munich, Germany, July 29–August 4, 2006.
6. Collins MW, Lovell MR, Mckeag DB: Current issues in managing sport-related concussion. **JAMA** 282:2283–2285, 1999.
7. Crisco JJ, Chu JJ, Greenwald RM: An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single-axis accelerometers. **J Biomech Eng** 126:849–854, 2004.
8. Delaney SJ, Puni V, Rouah F: Mechanisms of injury for concussions in university football, ice hockey, and soccer: A pilot study. **Clin J Sport Med** 16:162–165, 2006.
9. DiMasi F, Marcus JH, Eppinger RH: 3-D anatomic brain model of relating cortical strains to automotive crash loadings. Presented at the 13th International Technical Conference on Experimental Safety Vehicles, Paris, France, November 4–7, 1993.
10. 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** 15:3–8, 2005.
11. Eppinger RH: A Prospective View of Head Injury Research, in Bandak FA, Eppinger RH, Ommaya AK (eds): *Traumatic Brain Injury: Bioscience and Mechanics*. Mary Ann Liebert Inc, New York, 1996, pp 19–38.
12. Faraggi D, Reiser B: Estimation of the area under the ROC curve. **Stat Med** 21:3093–3106, 2002.
13. Gadd CW: Use of a weighted-impulse criterion for estimating injury hazard, in *10th Stapp Car Crash Confrence.* New York, Society of Automotive Engineers, 1966, pp 164–174.
14. Gurdjian ES, Roberts VL, Thomas, LM: Tolerance curves of acceleration and intracranial pressure protective index in experimental head injury. **J Trauma** 6:600–604, 1966.
15. Gurdjian ES, Webster JE, Lissner HR: Observations on the mechanism of brain concussion, contusion, and laceration. **Surg Gynecol Obstet** 101:688–690, 1955.
16. Guskiewicz KM, Bruce SL, Cantu RC, Ferrara MS, Kelly JP, McCrea M, Putukian M, Valovich McLeod TC: National Athletic Trainers' Association Position Statement: Management of sport-related concussion. **J Athl Train** 39:280–297, 2004.
17. Guskiewicz KM, Weaver NL, Padua DA, Garrett WE Jr: Epidemiology of concussion in collegiate and high school football players. **Am J Sports Med** 28:643–650, 2000.
18. Hanley JA, McNeil BJ: A method of comparing areas under receiver operating characteristic curves derived from the same cases. **Radiology** 148: 839–843, 1983.
19. Hanley JA, McNeil BJ: The meaning and use of the area under receiver operator characteristic (ROC) curve. **Radiology** 143:29–36, 1982.
20. Hodgson VR, Thomas LM, Khalil TB: The role of impact location in reversible cerebral concussion. Presented at the 27th Stapp Car Crash Conference, SAE 831618, Warrendale, Society of Automotive Engineers, 1983.
21. Jackson JE: *A User's Guide to Principal Components*. New York, John Wiley & Sons, 1991, pp 189–232.
22. Jollife IT: *Principal Component Analysis*. New York, Springer-Verlag, 1986, pp 167–195.
23. Kleiven S: Influence of impact direction on the human head in prediction of subdural hematoma. **J Neurotrauma** 20:365–379, 2003.
24. Margulies SS, Thibault LE: A proposed tolerance criterion for diffuse axonal injury in man. **J Biomech** 25:917–923, 1992.
25. 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** 42:383–388, 2006.
26. Neter J, Kutner MH, Nachtsheim CJ, Wasserman W: *Applied Linear Statistical Models*. Boston, McGraw-Hill, 1996, ed 4, pp 289–295.
27. Newman JA, Barr C, Beusenberg M, Fournier E, Shewchenko N, Welbourne E, Withnall C: A new biomechanical assessment of mild traumatic brain injury, part 2: Results and conclusions. *Proceedings of International Research Conference on the Biomechanics of Impacts (*IRCOBI*).* Montpellier, 2000, pp 223–230.
28. Newman JA, Shewchenko N, Welbourne E: A Proposed New Biomechanical Head Injury Assessment Function-The Maximum Power Index. Presented at the 44th Stapp Car Crash Conference, New York, New York, November 2000.
29. Nusholtz G, Lux P, Kaiker P, Janicki M: Head impact response-skull deformation and angular accelerations, in Backaitis SH (ed): *Biomechanics of Impact Injury and Injury Tolerance of the Head-Neck Complex*. Warrendale, Society of Automotive Engineers, 1993, pp 159–192.
30. Ommaya AK, Gennarelli TA: Cerebral concussion and traumatic unconciousness. Correlation of experimental and clinical observations of blunt head injuries. **Brain** 97:633–654, 1974.
31. Ommaya AK, Hirsch AE, Martinez JL: The role of "whiplash" in cerebral concussion. *Proceedings of the 10th Stapp Car Crash Conference*, Holloman Air Force Base, NM. Warrendale, Society of Automotive Engineers, 1966, pp 197–203.
32. Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF: Concussion in professional football: Reconstruction of game impacts and injuries. **Neurosurgery** 53:799–814, 2003.
33. Powell JW, Barber-Foss KD: Injury patterns in selected high school sports: A review of the 1995–1997 seasons. **J Athl Train** 34:277–284, 1999.
34. Powell JW, Barber-Foss KD: Traumatic brain injury in high school athletics. **JAMA** 282:958–963, 1999.
35. Prasad P, Mertz HJ: The Position of the United States Delegation to the ISO Working Group 6 on the Use of HIC in the Automotive Environment, SAE 851246. Warrendale, Society of Automotive Engineers, 1985.
36. 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** 60:490–495, 2006.
37. Thurman DJ, Branche CM, Sniezek JE: The epidemiology of sports-related traumatic brain injuries in the United States: Recent developments. **J Head Trauma Rehabil** 13:1–8, 1998.
38. Tosteson TD, Buonaccorsi JP, Demidenko E, Wells WA: Measurement error and confidence intervals for ROC curves. **Biom J** 47:409–416, 2005.
39. Versace J: A review of the severity index, SAE 710881, in *Proceedings of the 15th Stapp Car Crash Conference*. Warrendale, Society of Automotive Engineers, 1971, pp 771–796.
40. Zhang L, Yang KH, King AI: A proposed injury threshold for mild traumatic brain injury. **J Biomech Eng** 126:226–236, 2004.
41. Zhang L, Yang KH, King AI: Comparison of brain responses between frontal and lateral impacts by finite element modeling. **J Neurotrauma** 18:21–30, 2001.

Acknowledgments

This work was supported in part by an award from the National Center for Medical Rehabilitation Research at the National Institute of Child Health and Human Development at the National Institutes of Health (Grant R44HD40473) and research and development support from Riddell, Inc. (Chicago, IL). Accelerometers were provided by or purchased from Analog Devices (Cambridge, MA). We appreciate and acknowledge the researchers and institutions from whom the data were collected, including Stefan Duma, Ph.D., Virginia Tech–Wake Forest Center for Injury Biomechanics; P. Gunnar Brolinson, D.O., Virginia Polytechnic Institute and State University and Edward Via Virginia College of Osteopathic Medicine; Mike Goforth, A.T.C., Virginia Polytechnic Institute and State University; Brock Schnebel, M.D., McBride Clinic, Inc., Oklahoma City, OK, and University of Oklahoma; Scott Anderson, A.T.C., University of Oklahoma; Ron Gatlin, A.T.C., Casady High School, Oklahoma City, OK; Tom McAllister, M.D., Dartmouth Hitchcock Medical Center, Hanover,

NH; Jeff Frechette, A.T.C., and Scott Roy, A.T.C., Dartmouth College; Steven Broglio, Ph.D., A.T.C., Department of Kinesiology and Community Health, University of Illinois, Chicago; Brian Lund, A.T.C., and Dean Kleinschmidt, A.T.C., University of Indiana; Steven Erickson, M.D., and Bryce Nalepa, A.T.C., Arizona State University; Mickey Collins, Ph.D., Center for Sports Medicine, University of Pittsburgh Medical Center, Pittsburgh, PA; Jesse Townsend, A.T.C., Greensburg Salam High School, Greensburg, PA; Jeff Cienick, A.T.C., Blackhawk High School, Beaver Falls, PA; John Burnett, A.T.C., Karns City High School, Karns City, PA; John Panos, A.T.C., Fox Chapel High School, Pittsburgh, PA; Chris D'Antionio, A.T.C., Mt. Lebanon High School, Mt. Lebanon, PA; and Don Short, A.T.C., Hopewell High School, Aliquippa, PA.

COMMENTS

The head impact telemetry system jointly developed by Simbex Corporation (Lebanon, NH) and helmet manufacturer Riddell (Chicago, IL) provides an interesting and potentially valuable system for obtaining biomechanical data with the use of implanted accelerometers in helmets. The goal of this technology is to determine how biomechanical measures of head impact correlate with concussion. Of 449 high school and collegiate athletes monitored, 17 were diagnosed by the medical staff as experiencing a concussive blow. Of the 17 concussive blows, 8 were to the front of the head, 3 were to the top of the head, 5 were to the side of the head, and 1 was to the back of the head.

Linear acceleration, rotational acceleration, impact duration, and impact location are all biomechanical factors that have been linked to concussion. In the National Football League study reported by Pellman et al. (1), 98.9 *g* of linear acceleration was thought to be the concussive threshold. In this study, the authors did not duplicate that. They collected 3476 impacts with linear acceleration greater than 98.9 *g* and found that only 11 (0.3 versus 75% in Pellman et al.'s series) were associated with a clinical diagnosis of concussion.

The possibility of false positives, the small number of concussive impacts, and the methodology used are all important variables that could skew the authors' results. However, their conclusion that a composite variable of the various forces and accelerations is more sensitive an indicator that linear acceleration alone seems reasonable. This report adds additional biomechanical observations on concussion to the growing literature in this field.

Joseph C. Maroon
Pittsburgh, Pennsylvania

1. Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF: Concussion in professional football: Reconstruction of game impacts and injuries. **Neurosurgery** 53:799–814, 2003.

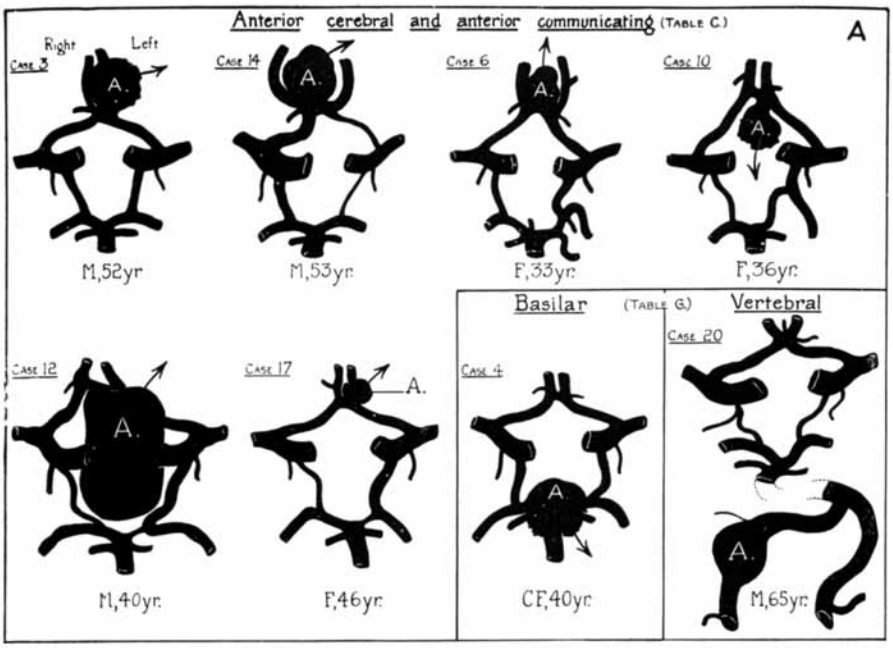
The heart of this study lies in interpreting how biomechanical data can best be used to predict concussion in the context of a dataset in which 17 concussions were identified in a sample of more than a quarter million instrumented head impacts during games of American football. Ideally, a well-designed, exquisitely sensitive, incontestably valid, and highly reliable predictive algorithm can be found that signifies the exact moment each of those 17 events transpired. The authors have located a decently performing statistical tool for this purpose in the form of a weighted composite score derived from principal component analysis, a well-known method of handling complex data. More work on the properties and performance of this statistical approach and others is recommended, because the present study's sample of known concussions is small.

David L. McArthur
Epidemiologist
Los Angeles, California

Daniel F. Kelly
Santa Monica, California

This small, preliminary study attempts to use in-helmet accelerometers (six per helmet) to collect and interpret data about concussions in American football players. Readers must remember that the authors have a financial interest in the company that makes this product; however, the data must speak for themselves. The greatest difficulty with these types of studies is summarizing the large amount of highly complex information that is collected and then simplifying it into an easily usable scheme. The authors' creation of a single composite variable (weighted Principal Component Score) is a first step in this direction. This composite measure incorporates impact location, linear acceleration, rotational acceleration, and impact duration. Future studies involving more athletes in different sports and at different skill levels (e.g., high school, college, professional) will undoubtedly test the usefulness of this variable and generate refinements and modifications.

Alex B. Valadka
Houston, Texas



Aneurysms of the anterior cerebral, anterior communicating, basilar, and vertebral arteries. From Dandy WE: *The Brain*. New York, Harper & Row, Publishers, 1969. Reprinted from *Lewis' Practice of Surgery*, Prior, 1933.