

BIOMECHANICAL RISK ESTIMATES FOR MILD TRAUMATIC BRAIN INJURY

J. R. Funk
Biodynamic Research Corporation
San Antonio, Texas

S. M. Duma, S. J. Manoogian, S. Rowson
Impact Biomechanics Laboratory
Virginia Tech
Blacksburg, Virginia

ABSTRACT

The objective of this study was to characterize the risk of mild traumatic brain injury (MTBI) in living humans based on a large set of head impact data taken from American football players at the collegiate level. Real-time head accelerations were recorded from helmet-mounted accelerometers designed to stay in contact with the player's head. Over 27,000 head impacts were recorded, including four impacts resulting in MTBI. Parametric risk curves were developed by normalizing MTBI incidence data by head impact exposure data. An important finding of this research is that living humans, at least in the setting of collegiate football, sustain much more significant head impacts without apparent injury than previously thought. The following preliminary nominal injury assessment reference values associated with a 10% risk of MTBI are proposed: a peak linear head acceleration of 165 g, a HIC of 400, and a peak angular head acceleration of 9000 rad/s².

Biomechanical research on head injury has historically focused on characterizing serious injuries using cadaveric or animal models. These models are limited in their ability to predict MTBI, or concussion, in living humans. MTBI cannot be detected in cadavers, and injury thresholds derived from animal experiments cannot be directly applied to living humans. Although typically only an AIS 1 injury, MTBI occurs far more commonly than severe head injury and is a major public health problem (Kelly, 1999). Obtaining biomechanical data characterizing MTBI in living humans is difficult because it would obviously be unethical to intentionally inflict MTBI on volunteers in a laboratory setting. For that reason, studying athletes in contact sports is a promising avenue of biomechanical research on MTBI.

An important MTBI research goal is to establish biomechanical risk curves that can be used to predict the likelihood of injury based on head impact severity. MTBI risk curves would be useful in improving helmets, padding, and other countermeasures to reduce the incidence of concussion in sports, car crashes, and other settings. Although several studies have provided important information about the magnitude of injurious as well as non-injurious head impacts sustained by living humans, an accurate MTBI risk curve has yet to be established. Pellman et al. (2003) have provided the most extensive biomechanical incidence data concerning concussive head impacts in living humans to date. Based on video footage of National Football League (NFL) games, 31 head impacts were reconstructed using helmeted Hybrid III dummies. In 25 of these events, a player sustained a concussion. Pellman et al. (2003) presented risk curves for MTBI based on logistic regressions of the NFL data, and similar risk curves were published by King et al. (2003) and Zhang et al. (2004). We believe these risk curves are flawed because the statistical analysis requires unbiased exposure data (Altman, 1991; Evans, 2004), but the NFL data were intentionally biased towards injurious impacts. As a result, the NFL risk curves are far too conservative. The objective of the present study was to establish biomechanical risk curves for MTBI using a large set of unbiased head impact data collected from collegiate football players.

METHODS

Head impact data were obtained from instrumented helmets worn by Virginia Tech (VT) football players during the 2003 – 2006 seasons as described by Duma et al. (2005). Data were collected from 64 different players during games and full impact practices. All players gave written informed consent with Institutional Review Board approval from both Virginia Tech and the Edward Via Virginia College of Osteopathic Medicine. The helmets were instrumented with the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH), which consisted of six spring-mounted linear accelerometers designed to stay in contact with the player's head. Data from the accelerometers were sampled at 1000 Hz and transmitted wirelessly to a laptop computer on the sidelines in real time. Only impacts in which the signal from at least one of the accelerometers exceeded a user-defined threshold of 10 g were recorded. Algorithms were developed to remove extraneous head impact data recorded when the helmet was not on the player's head, and high severity impacts were confirmed based on team records and game video. The linear acceleration at the center of gravity of the head was estimated from the six accelerometer signals using an algorithm described by Crisco et al. (2004). The Head Injury Criterion (HIC) was calculated from the resultant linear acceleration at the head center of gravity.

Impacts were classified as injurious based on a diagnosis of concussion by the team physician.

The HIT system has been subjected to numerous validation tests in which an instrumented helmet was placed on a Hybrid III dummy head-neck complex and struck at various impact locations, directions, and severity levels. Over 100 validation tests were conducted with the 2006 f-HMAS MxEncoder, which is substantially similar to the HIT system instrumentation package used by Virginia Tech in this study. The validation tests showed that the HIT system accurately reproduces the shape and magnitude of the resultant linear acceleration pulse measured by the accelerometers mounted at the center of gravity of the Hybrid III head. Estimates of the measurement error of the HIT system were obtained by linear regression of the validation data. Compared to the dummy instrumentation, the HIT system overestimated the peak head acceleration by $8\% \pm 11\%$ and overestimated the HIC by $23\% \pm 28\%$. Although the HIT system was felt to have good accuracy overall, analyses were conducted using both the raw data and data that were corrected for measurement error in order to provide a more conservative estimate of injury risk. Individual measurements were corrected for measurement error simply by dividing the peak acceleration values by 1.08 and the HIC values by 1.23.

MTBI risk curves were calculated as a function of impact severity, which was characterized in terms of peak resultant linear head acceleration and HIC. Risk curves were developed by normalizing MTBI incidence data by the head impact exposure data. The HITS data were assumed to provide unbiased and mutually independent estimates of head impact exposure. This assumption was considered valid because of the prospective nature of the study and the fact that the instrumented players were chosen to encompass a wide range of body types and player positions. The distribution of head impact severities was expressed in terms of a probability density function (pdf) and cumulative distribution function (cdf) using a standard Weibull distribution:

$$pdf = f(x) = \frac{\alpha x^{\alpha-1}}{\beta^\alpha} e^{-\left(\frac{x}{\beta}\right)^\alpha}$$

(1a)

$$cdf = F(x) = 1 - e^{-\left(\frac{x}{\beta}\right)^\alpha}$$

(1b)

where α is the shape parameter, β is the scale parameter, and x is the impact severity in terms of either peak resultant head acceleration or HIC.

The HITS data were sorted and normalized to obtain a cdf of the head impact exposure data. The experimental cdf was mathematically

manipulated in two ways before curve fitting it to the Weibull distribution (equation 1b). First, in order to maximize the accuracy of the curve fit for the higher severity impacts that were of most interest, all impacts below 40 g were excluded when fitting the peak acceleration data, and all impacts with a HIC < 50 were excluded when fitting the HIC data. These relatively low severity impacts were deemed unlikely to cause MTBI, based on the NFL data (Pellman et al., 2003). The equation form for the Weibull distribution was adjusted to obtain the pdf and cdf over the severity region of interest $\theta < x < \infty$, where $\theta = 40$ g for the analysis of the peak acceleration data, and $\theta = 50$ for the analysis of the HIC data:

$$pdf_{\theta}(\theta < x < \infty) = f_{adj}(x) = \frac{f(x)}{1 - F(\theta)} = \frac{\alpha x^{\alpha-1}}{\beta^{\alpha}} e^{-\frac{\theta^{\alpha} - x^{\alpha}}{\beta^{\alpha}}}$$

(2a)

$$cdf_{\theta}(\theta < x < \infty) = F_{adj}(x) = \frac{F(x) - F(\theta)}{1 - F(\theta)} = 1 - e^{-\frac{\theta^{\alpha} - x^{\alpha}}{\beta^{\alpha}}}$$

(2b)

Second, rather than curve fitting equation (2b) directly to the experimental cdf of the head impact exposure (cdf_{exp}), the log transform of the complement of equation (2b) was fit to the log transform of the complement of the exposure cdf:

$$\frac{\theta^{\alpha} - x^{\alpha}}{\beta^{\alpha}} = \ln[1 - cdf_{exp}(\theta < x < \infty)]$$

(3)

This transformation was done in order to give greater weight to higher severity impacts when performing the least squares curve fit.

The above operations were applied to characterize the distribution of the raw HITS data. In addition to correcting individual HITS measurements for error due to bias, the distribution of impact severities was corrected for error due to scatter. The distribution of measurements reflects the sum of the true distribution of head impact severities and the distribution of the measurement error due to scatter. Because the probability distribution of the sum of two independent random variables is the convolution of each of their distributions, the true distribution of head impact severities can be obtained by deconvolving the measurement error from the measured distribution of head impacts:

$$f'(x) = \int_0^{\infty} f(z)g(x - z, z)dz$$

(4)

where f' is the measured distribution of head impacts (pdf_{exp}), f is the true distribution of head impacts, g is the distribution of measurement error

due to scatter, x is the measurement in terms of peak head acceleration or HIC, and z is a dummy variable. The equation form is not technically a true convolution, but it is still accurate. The measurement error due to scatter was characterized by an unbiased normal distribution having a standard deviation equal to a fixed ratio of the measured value (the coefficient of variation, c_v):

$$g(x - z, z) = \frac{1}{\sqrt{2\pi}c_v z} e^{-\frac{(x-z)^2}{2c_v^2 z^2}}$$

(5)

The coefficient of variation of the scatter error was 11% for the corrected peak acceleration data and 23% for the corrected HIC data. The cdf of $f^*(x)$ was obtained by numerically integrating $f^*(x)$, then adjusting $F^*(x)$ by normalizing it over the region $\theta < x < \infty$:

$$F'_{adj}(x) = \frac{\int_{\infty} f'(x) dx}{\int_{\theta} f'(x) dx}$$

(6)

Parameter estimates for α and β in the true distribution of impact severities were obtained by curve fitting the log transform of the complement of the adjusted cdf in equation (6) to the log transform of the exposure cdf:

$$\ln[1 - F'_{adj}(x)] = \ln[1 - cdf_{exp}(\theta < x < \infty)]$$

(7)

Once the overall distribution of head impacts was established for both the raw and corrected HITS data, these exposure data were normalized on a per player per play basis by combining the HITS data with player information obtained from the Virginia Tech football team. The probability that a player would sustain a head impact having a severity greater than θ was denoted by p_{θ} :

$$p_{\theta} = p(x_{hit} > \theta) = \frac{\# \text{ recorded impacts} > \theta}{\# \text{ instrumented player plays}}$$

(8)

where the number of recorded impacts greater than θ was obtained from the HITS data, and the number of player plays was the sum of the number of plays played by each instrumented player, which was obtained from team records.

Combining equations (2b) and (8), the probability that a player would sustain a head impact having a severity greater than x in a single play was calculated:

$$\begin{aligned}
 p(x_{hit} > x) &= p(x_{hit} > x | x_{hit} > \theta) p(x_{hit} > \theta) \\
 &= [1 - F_{adj}(x)] p_{\theta} = p_{\theta} e^{\frac{\theta^{\alpha} - x^{\alpha}}{\beta^{\alpha}}}
 \end{aligned}
 \tag{9}$$

It was assumed that each play could be modeled as an independent Bernoulli trial with regard to head impact severity, such that the probability of exactly k head impacts having a severity greater than x for a single player exposed to n plays could be calculated from the binomial distribution:

$$p_n(x_{hit} > x, k) = \binom{n}{k} [p(x_{hit} > x)]^k [1 - p(x_{hit} > x)]^{n-k}
 \tag{10}$$

The probability that the most severe head impact for a single player over n plays was greater than x was obtained from the complement of the probability that no head impacts would be greater than x (binomial distribution with $k = 0$):

$$p_n(x_{max} > x) = 1 - [1 - p(x_{hit} > x)]^n = 1 - \left(1 - p_{\theta} e^{\frac{\theta^{\alpha} - x^{\alpha}}{\beta^{\alpha}}} \right)^n \tag{11}$$

There are two equivalent approaches that can be taken to determine the expected number of concussions per player per play (p_{inj}). Both involve calculating the probability of a player sustaining a concussion at a particular impact severity level x in a given play and then integrating over all x to obtain the overall risk of concussion per play for that player:

$$p_{inj} = \int_0^{\infty} p_{inc}(x) dx
 \tag{12}$$

In the first approach, the probability of a concussion incidence for a player having an injury tolerance $x_{inj} = x$ is given by:

$$p_{inc}(x) = p(x_{inj} = x) p(x_{hit} > x)
 \tag{13a}$$

In the second approach, the probability of a concussion incidence for an impact of severity $x_{hit} = x$ is given by:

$$p_{inc}(x) = p(x_{hit} = x) p(x_{inj} < x)
 \tag{13b}$$

The incidence probability functions $p_{inc}(x)$ calculated by these two approaches are not equivalent. In the first approach (equation 13a), the probability of a concussion incidence is given in terms of the injury tolerance of the player ($x = x_{inj}$). In the second approach (equation 13b), the probability of a concussion incidence is given in terms of the severity of the injurious impact experienced by the player ($x = x_{hit}$). When a player is injured, it is because the severity of the head impact exceeded the injury tolerance of the individual by some amount ($x_{hit} > x_{inj}$). Therefore, the distribution of incidence probabilities calculated in terms of injury tolerances (equation 13a) will be shifted to the left somewhat (lower values of x) compared to the incidence probability calculated in terms of injurious impact severities (equation 13b). This reflects the fact that biomechanical data from injured players are left-censored. In spite of this difference, it can be shown using integration by parts that the area under each curve is the same:

$$\int_0^{\infty} p(x_{inj} = x)p(x_{hit} > x)dx = \int_0^{\infty} p(x_{hit} = x)p(x_{inj} < x)dx$$

(14)

Therefore, the expected number of concussions per player play can be calculated using either approach. The expected number of concussions after n player plays is given by:

$$E(n) = np_{inj}$$

(15)

It was desired to fit the injury risk curve $p(x_{inj} < x)$ to a cumulative Weibull distribution with parameters α_{inj} and β_{inj} (equation 1b). The injury risk curve was calculated by normalizing the injury incidence data by the head impact exposure data. The injury incidence equation (13b) was rewritten by substituting the derivative of the complement of equation (9) for the head impact exposure function and equation (1b) for the injury risk function:

$$p_{inc}(x) = p_{\theta} \frac{\alpha x^{\alpha-1}}{\beta^{\alpha}} e^{-\frac{x^{\alpha}}{\beta^{\alpha}}} \left[1 - e^{-\left(\frac{x}{\beta_{inj}}\right)^{\alpha_{inj}}} \right]$$

(16)

Peak head acceleration and HIC values associated with injurious impacts were sorted and normalized to obtain a cdf of the injury incidence data (cdf_{inc}). Parameter estimates for α_{inj} and β_{inj} were obtained by performing a least squares fit of the normalized integral of equation (16) to the incidence cdf:

$$\frac{\int_0^x p_{inc}(x)dx}{\int_0^\infty p_{inc}(x)dx} = \frac{1}{p_{inj}} \int_0^x p_\theta \frac{\alpha x^{\alpha-1}}{\beta^\alpha} e^{-\frac{\theta^\alpha - x^\alpha}{\beta^\alpha}} \left[1 - e^{-\left(\frac{x}{\beta_{inj}}\right)^{\alpha_{inj}}} \right] dx = cdf_{inc}(x)$$

(17)

The integral in equation (17) has no closed-form solution, so it was solved numerically subject to the constraint given in equation (12).

Two different data sources for the MTBI incidence cdf were utilized to calculate the injury risk curves: the Virginia Tech HITS data (n = 4) and the NFL data (Pellman et al., 2003) (n = 25). In all analyses, the head impact exposure was estimated from Virginia Tech HITS data and the expected number of concussions per player play (p_{inj}) was estimated from the epidemiological data of Pellman et al. (2004). Pellman et al. (2004) reported a mean concussion rate of 0.2 per team per game in the NFL, which equates to approximately one MTBI per 8000 player plays ($p_{inj} = 0.000125$) assuming a nominal value of 1600 player plays per game.

The HIT system was not capable of accurately measuring resultant angular head acceleration. However, it was desired to estimate MTBI risk in terms of peak angular head acceleration. Pellman et al. (2003) reported that peak linear and peak angular accelerations were well correlated ($R^2 = 0.76$), and provided the following equation relating these two parameters:

$$\alpha_{peak} \left(rad / s^2 \right) = 47.7 \cdot a_{peak} \left(g \right) + 1515$$

(18)

MTBI injury risk was therefore mapped from peak linear head acceleration to peak angular head acceleration using equation (18).

RESULTS

Data from 27,319 head impacts in games and practices were collected. One third to one half of all player plays during games were captured by the HIT system each season. The measured impact exposures followed a roughly exponential distribution in terms of both peak head acceleration and HIC (Figures 1 and 2). The vast majority of the impacts were of low severity. Nonetheless, there were 3,956 impacts in which the raw peak head acceleration was greater than 40 g and 1,662 impacts in which the raw HIC was greater than 50. There were four impacts in which an instrumented player sustained a concussion. In these impacts, the raw peak head accelerations were 81 g, 139 g, 172 g, and 200 g (mean 148 ± 51 g), and the raw HIC values were 200, 589, 661, and 859 (mean 557 ± 331). In three out of the four concussed players,

the injurious impact was the most severe impact experienced by that particular player in terms of both peak acceleration and HIC. In the fourth concussed player, who was injured at the lowest impact severity, the injurious impact was the third most severe for that player in terms of HIC, and the eleventh most severe impact in terms of peak head acceleration.

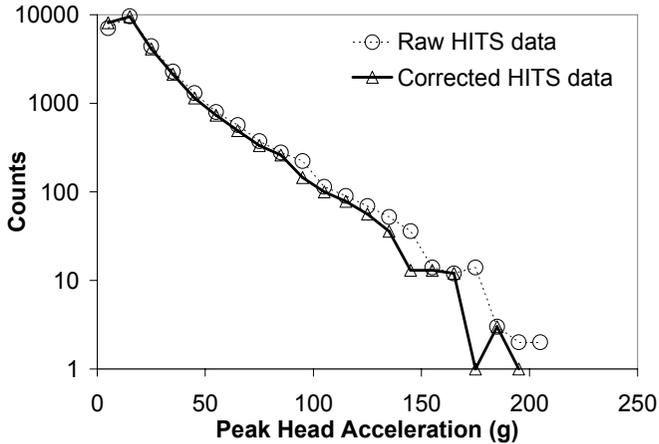


Figure 1 – Number of head impacts at various severity levels grouped in increments of 10 g.

The injurious head impacts in the present study were associated with higher impact severities than the NFL study. Even after correcting for measurement error, the mean peak head acceleration associated with MTBI was higher in the present study than in the NFL study (137 ± 47 g vs. 98 ± 28 g, $p = 0.20$). The mean corrected HIC associated with MTBI was also higher in the present study than in the NFL study (470 ± 225 vs. 381 ± 197 , $p = 0.50$). However, possibly due to the small number of injury data points in the present study, neither of these differences was statistically significant using two-tailed t-tests.

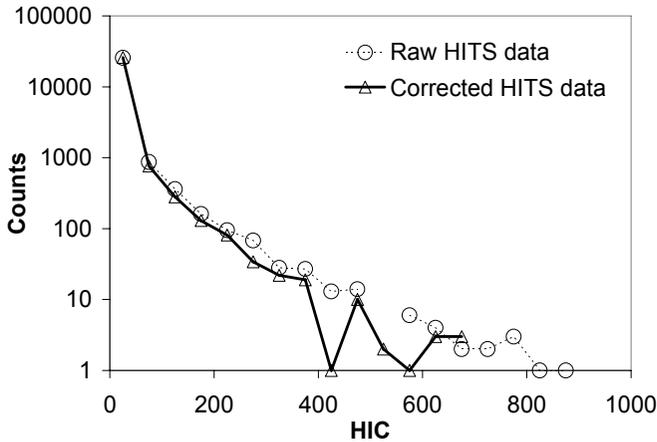


Figure 2 – Number of head impacts at various severity levels grouped in HIC increments of 50.

It was found that over the course of a season, most Virginia Tech football players were exposed to significant head impacts. For each play, it was found that the average player had an approximately 4.8% chance of experiencing a head impact having a peak acceleration greater than 40 g, and approximately a 2.0% chance of sustaining a HIC greater than 50 (p_0 in Table 1). Equation (11) predicts that after 1000 plays, a player is almost certain to have experienced at least one head impact with a peak acceleration greater than 100 g or a HIC greater than 200 (Figures 3 and 4). The mean number of game plays per player was approximately 250 per season, but some players played in as many as 800 game plays in a season.

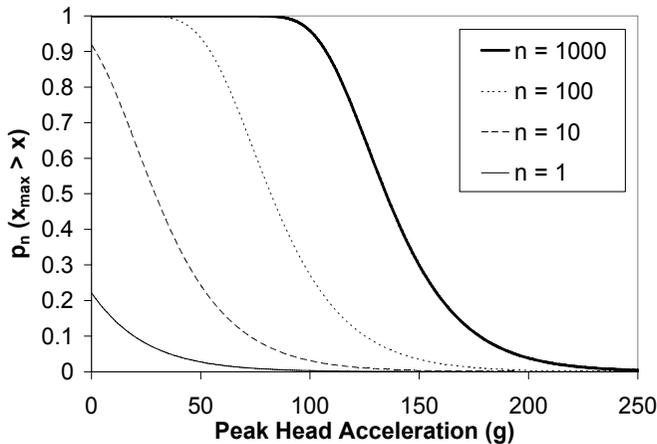


Figure 3 –Probability distribution of the most severe head impact as a function of the number of plays n (from corrected acceleration data).

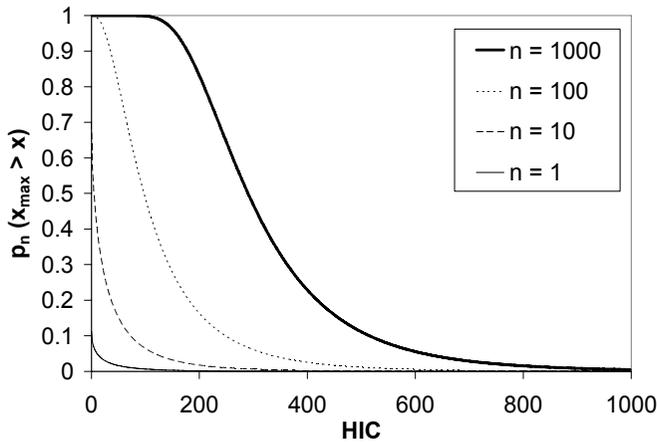


Figure 4 –Probability distribution of the most severe head impact as a function of the number of plays n (from corrected HIC data).

Nominal injury assessment reference values (IARVs) representing a 10% risk of MTBI were 185 g, a HIC of 525, and 10,000 rad/s^2 based on the raw HITS data, and 165 g, a HIC of 400, and 9,000 rad/s^2 based on the corrected HITS data. Injury risk curves fit to the NFL incidence data combined with the VT exposure data predicted 10% risk values of approximately 200 g, a HIC of 420, and 11,000 rad/s^2 (Table 2). The MTBI risk curves calculated in this study tended to be similar at low levels of risk and then diverge at higher risk levels (Figures 5 and 6). Correcting the peak head acceleration and HIC values for measurement error in the HIT system made the risk curves more conservative. The accuracy of the injury risk curves was assessed by examining how well the predicted incidence of MTBI matched the actual data. As described previously, all injury risk curves were constrained to predict an incidence of 0.2 concussions per team per game as reported in the NFL study (Pellman et al., 2004). The predicted cumulative distribution of injury-producing impact severities generally matched the actual incidence data quite well ($R^2 > 0.9$) (Figures 7 and 8).

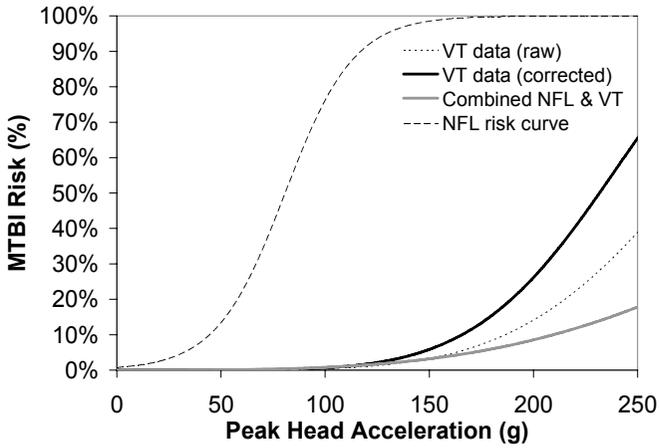


Figure 5 – Injury risk curves in terms of peak head acceleration.

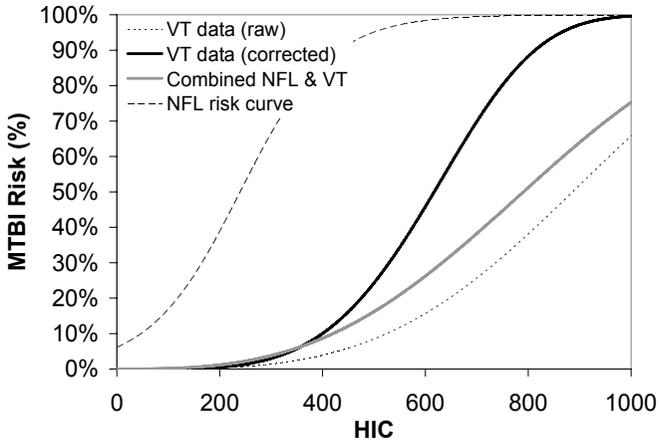


Figure 6 – Injury risk curves in terms of HIC.

Table 1 – Parameters characterizing the distributions of head impact exposure.

	Peak G		HIC	
	Raw	Corrected	Raw	Corrected
α	0.99	1.03	0.51	0.55
β	25.23	24.41	14.34	14.88
θ	40	36.92	50	40.69
p_0	0.048	0.048	0.020	0.020

Table 2 – Parameters characterizing the MTBI injury risk curves.

Exposure data	VT raw	VT corrected	VT corrected
Incidence data	VT raw	VT corrected	NFL study
Peak translational head acceleration (g)			
α_{inj}	5.27	5.63	3.53
β_{inj}	286	247	397
1% risk	119	109	107
5% risk	162	145	171
10% risk	186	165	209
Head Injury Criterion (HIC)			
α_{inj}	3.60	4.34	2.98
β_{inj}	981	671	894
1% risk	273	232	191
5% risk	430	338	330
10% risk	525	400	420
Peak angular head acceleration (rad/s ²) (using eq. 18)			
1% risk	7191	6714	6619
5% risk	9242	8432	9672
10% risk	10,387	9386	11,484

In all cases, the injury risk curves calculated using the Virginia Tech exposure data were substantially less conservative than the NFL risk curves. The nominal values of 85 g, a HIC of 250, and 6000 rad/s² that are associated with an approximately 50% risk of injury according to the NFL risk curves (Zhang et al., 2004) were associated with approximately a 1% or lower risk of MTBI based on the risk curves developed in the present study (Table 2). The NFL risk curves, when multiplied by the corrected Virginia Tech exposure data, predicted 26 concussions per game based on peak head acceleration data and 94 concussions per game based on HIC data, which far exceeds the actual concussion incidence of 0.2 concussions per game reported in the same NFL study (Pellman et al., 2004).

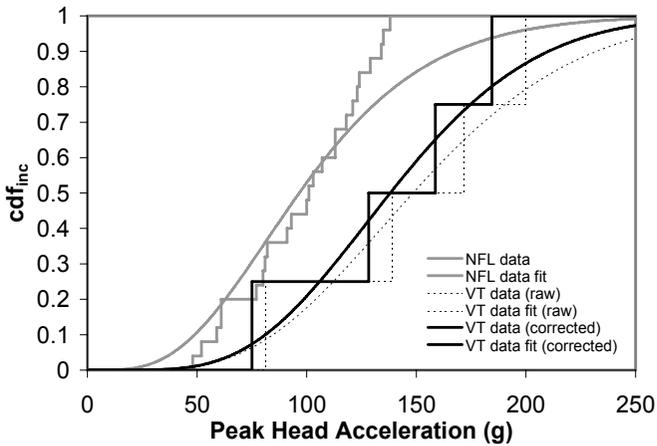


Figure 7 – Cumulative injury incidence in terms of peak head acceleration.

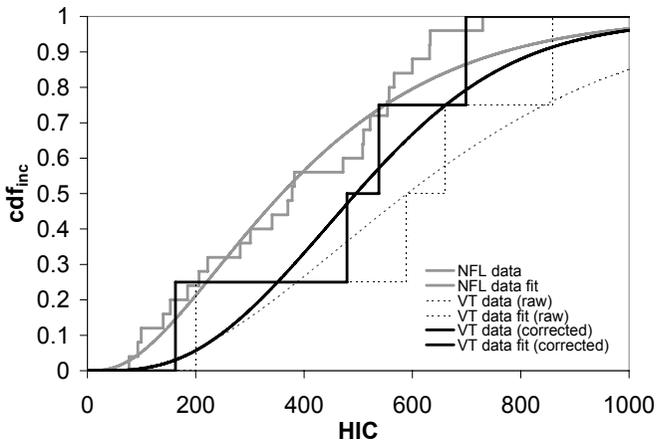


Figure 8 – Cumulative injury incidence in terms of HIC.

DISCUSSION

MTBI risk curves were developed using biomechanical head impact data collected from collegiate football players. The injury risk curves developed in the present study are drastically different from the injury risk curves developed from the NFL data and reported by Pellman et al. (2003), King et al. (2003), and Zhang et al. (2004). MTBI risk curves from the present study are more accurate because they utilized a very large data set that provided an unbiased estimate of head impact exposure. Exposure data were unbiased because the present study was designed as a prospective cohort study (Altman, 1991). The experimental design of the NFL study was a case control study. As such,

the NFL data set was intentionally biased towards injurious impacts. The fact that the MTBI injury rate (p_{inj}) predicted by multiplying the NFL risk curve by the VT exposure data is over 100 times the actual injury rate observed in the same NFL study (Pellman et al., 2004) demonstrates that the NFL risk curve is not properly normalized for head impact exposure. Other studies have also documented that athletes regularly sustain significant head impacts without apparent injury at severity levels that would be expected to cause concussion based on the NFL risk curves (Pincemaille et al., 1989; Naunheim et al., 2000).

In light of the more complete exposure data obtained in the present study, it was concluded that the risk of injury at the severity levels where most injuries occur (peak head acceleration ~ 100 g, HIC ~ 400) is actually low, not high. This same phenomenon occurs and is widely recognized in the field of traffic safety. For example, although roughly half of all fatal crashes occur at velocity changes of 30 mph or less, the risk of fatality in a 30 mph delta-V crash is on the order of 10% (Evans, 2004). Because there are so many low severity car crashes and low severity head impacts in the field, the incidence of injury at these levels is amplified, which belies the fact that they are actually relatively low risk events.

The biomechanical risk estimates for MTBI presented here are in agreement with previous head injury research. In terms of peak translational head acceleration and HIC, the risk curves associated with MTBI, which is typically an AIS 1 injury, are more conservative than published risk curves associated with more serious head injuries such as skull fracture or serious brain injury (Prasad and Mertz, 1985; Mertz et al., 1996). Because angular head acceleration was not directly measured in the present study, those MTBI risk estimates should be viewed more cautiously. The NFL data (Pellman et al., 2003) showed a good correlation between peak linear and peak angular head acceleration ($R^2 = 0.76$), and equation (18) provides an unbiased mapping between those two parameters. Although this mapping may not be accurate for each individual impact, it should be accurate on average, which is why we believe it is an appropriate methodology in this case. Furthermore, the levels of peak angular head acceleration associated with a low risk of MTBI (Table 2) are in agreement with the findings of Thibault et al. (1989). In their animal experiments, no concussions were reported below a scaled value of 4500 rad/s² and approximately 10% of their concussions occurred below a scaled value of 9000 rad/s². The present study provides further evidence that the scaled value of 1800 rad/s² proposed as a 50% risk of concussion by Ommaya and Hirsch (1971) is erroneous.

The fact that many players sustained higher peak head accelerations and HICs without injury than some of the concussed players does not imply that peak head acceleration and HIC were poor predictors of MTBI. On the contrary, peak head acceleration and HIC were both found to be good predictors of MTBI in this study. Both

parameters identified the injurious impact as being the most severe impact or among the most severe impacts sustained by the injured player. The variation in injury tolerance between individuals explains why high severity impacts cause MTBI in some players but not others.

The key to obtaining unbiased head impact exposure data in the present study was the HIT system, which records all head impacts experienced by instrumented players. The measurement error in the peak head acceleration calculated by the HIT system was found to be $8\% \pm 11\%$ in validation testing. This level of error is comparable to many other indirect biomechanical measurement devices. Small errors in the magnitude and duration of the acceleration pulse measured by the HIT system were magnified in the HIC calculation, which raises the acceleration to the 2.5 power and is sensitive to changes in the pulse shape and possibly sampling rate. Measurement error in the HITS data was accounted for by correcting individual data points for bias error and by deconvolving the measured distributions to remove the effect of scatter error. Having said that, deconvolution of the scatter error resulted in only a very minor adjustment to the measured distributions in this study.

In addition to measurement error in the HIT system, the MTBI risk curves calculated in this study are also subject to sampling error. Sampling errors would include extraneous head impact data recorded when the helmet was not on the player's head, as well as errors due to a small sample size for injurious or high severity impacts. The head impact exposure function (equation 2a) is derived from a curve fit of several thousand head impacts, and is therefore insensitive to the presence of a small number of extraneous data points. Likewise, the MTBI risk curves were relatively insensitive to changes in the incidence cdf and the expected injury rate (p_{inj}). Another factor that can affect the injury sample is the definition of concussion, which is somewhat ambiguous. Although concussion severity occurs along a continuum, the diagnosis of concussion is a yes or no decision that is made by the treating physician. In the present study, all concussions were minor and none of the injured players lost consciousness. Other researchers might define concussion more liberally or conservatively, and their injury sample and risk estimates would vary accordingly. In addition, some concussions may go unrecognized or unreported. Of course, it is not possible to quantify the number of unreported concussions. We assumed that because the players are closely monitored, only very minor concussions would be missed.

Even after accounting for measurement error in the HIT system, a difference was observed between the mean peak head acceleration and HIC values among concussed players in the present study and the NFL study. Though not statistically significant, the difference is probably greater than what would be expected between two presumably similar populations of football players. The HIT system has the advantage of

measuring injurious impacts directly on the injured player, rather than retrospectively in a dummy reconstruction. It is possible that a lack of dummy biofidelity with respect to head mass, body mass, and neck stiffness introduced error into the NFL biomechanical data. In spite of the differences in the VT and NFL injury incidence data, the injury risk curves calculated from each data set were relatively similar at low risk levels (Figures 5 and 6) due to the high exposure at those levels of impact severity. Therefore, the NFL data corroborates the risk curves developed from the VT data and suggests that the analysis is robust at low levels of risk in spite of the relative paucity of injury data in the VT data set.

It should be emphasized that the injury risk curves presented here are only a first attempt at calculating injury assessment risk values from limited injury data. Parametrically fitting cumulative incidence and exposure data allowed smooth injury risk curves to be calculated from sparse data. The injury risk curves (Figures 5 and 6) are most accurate in the low risk region where the exposure data are densest, and cannot be used to estimate impact severities associated with higher levels of risk. For example, 50% risk levels for concussion cannot be reliably calculated because the exposure data are extremely sparse in that region. It is anticipated that with the addition of more injury and exposure data in the future, the MTBI risk curves will be refined and confidence in the injury risk curves will be extended to higher levels of risk. Future work will also examine the effect of impact direction, player position, and other factors thought to affect impact exposure and injury risk.

The MTBI risk curves derived from the raw HITS data can be put to direct clinical use by physicians and other sports medicine personnel who are using the HIT system with an instrumentation package comparable to the one in this study. We believe that the MTBI risk curves based on the corrected HITS data have applicability for predicting injury in the general population for the case of padded head impacts. There is some concern that football players may have a higher concussion tolerance than other people because they are young, male, large, athletic, and generally impact-tolerant. However, these attributes are not likely to have a large effect on concussion tolerance. Although age and gender have been shown to affect injury tolerances in other parts of the body, no significant age or gender effect has been demonstrated with regard to brain injury tolerance. The fact that football players are large and athletic has no bearing on their brain injury tolerance, because the injury tolerance is expressed in terms of head acceleration, which effectively normalizes for head mass and muscle strength. It is possible that there is some selection for MTBI tolerance among football players who make it to the collegiate level. However, concussions in football occur comparatively rarely and generally have a better outcome than orthopaedic injuries, so it is doubtful that a low average concussion tolerance becomes the limiting factor in the progression of an athlete's career very often. In spite of these concerns and limitations, we believe

the present study provides good preliminary estimates of the impact severities associated with a low risk of MTBI that are based on some of the best available biomechanical and epidemiological data for living humans.

ACKNOWLEDGMENTS

The Head Impact Telemetry System development was funded in part by NIH R43HD40743. Additional support for this project was provided by the Edward Via College of Osteopathic Medicine, the Virginia Tech College of Engineering and the Department of Sports Medicine, and Simbex.

REFERENCES

- Altman DG. Practical Statistics for Medical Research. Chapman & Hall/CRC: London, 1991.
- 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.
- Duma SM, Manoogian SJ, Bussone WR, et al., “Analysis of Real-time Head Accelerations in Collegiate Football Players,” *Clin J Sports Med*, 15(1): 3 – 8, 2005.
- Evans L. Traffic Safety. Science Serving Society: Bloomfield Hills, MI, 2004.
- Kelly JP, “Traumatic Brain Injury and Concussion in Sports,” *JAMA*, 282(10): 989 – 991, 1999.
- King AI, Yang KH, Zhang L, Hardy W, Viano DC, “Is Head Injury Caused by Linear or Angular Acceleration?” *Proceedings of the International Research Conference on the Biomechanics of Impact (IRCOBI)*, 2003.
- Mertz HJ, Prasad P, Nusholtz G, “Head Injury Risk Assessment for Forehead Impacts,” Paper 960099, *Society of Automotive Engineers*, 21 pp., 1996.
- Naunheim RS, Standeven J, Richter C, et al., “Comparison of Impact Data in Hockey, Football, and Soccer,” *J Trauma*, 48(5): 938 – 941, 2000.

- Ommaya AK, Hirsch AE, "Tolerances for Cerebral Concussion From Head Impact and Whiplash in Primates," *J Biomech*, 4: 13 – 21, 1971.
- Pellman EJ, Viano DC, Tucker AM, et al., "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," *Neurosurgery*, 53(4): 799 – 814, 2003.
- Pellman EJ, Powell JW, Viano DC, et al., "Concussion in Professional Football: Epidemiological Features of Game Injuries and Review of the Literature – Part 3," *Neurosurgery*, 54(1): 81 – 96, 2004.
- Pincemaille Y, Trosseille X, Mack P, et al., "Some New Data Related to Human Tolerance Obtained from Volunteer Boxers," *Proc 33rd Stapp Car Crash Conf*, Paper 892435, pp. 177 – 190, 1989.
- 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," Paper 851246, *Society of Automotive Engineers*, 11 pp., 1985.
- Thibault L, Gennerelli T, Margulies S, "Animal, Physical and Analytical Models for Use in the Development of Improved Head Injury Criteria," Final Report No. DOT HS 807 481 to NHTSA, 283 pp., 1989.
- Zhang L, Yang KH, King AI, "A Proposed Injury Threshold for Mild Traumatic Brain Injury," *J Biomech Eng*, 126: 226-236, 2004.