Magnitude of Head Impact Exposures in Individual Collegiate Football Players

Joseph J. Crisco[1], **Bethany J. Wilcox**[1], **Jason T. Machan**[1], **Thomas W. McAllister**[2], **Ann-Christine Duhaime**[3], **Stefan M. Duma**[4], **Steven Rowson**[4], **Jonathan G. Beckwith**[5], **Jeffrey J. Chu**[5], and **Richard M. Greenwald**

[1]Bioengineering Laboratory, Department of Orthopaedics, the Warren Alpert Medical School of Brown University and Rhode Island Hospital, Providence, RI.

[2]Department of Psychiatry, Geisel School of Medicine at Dartmouth, Lebanon, NH.

[3]Pediatric Neurosurgery, Children's Hospital at Dartmouth, Dartmouth Hitchcock Medical Center, Hanover, NH.

[4]Center for Injury Biomechanics, Virginia Tech–Wake Forest, Blacksburg, VA.

[5]Simbex, Lebanon, NH.

[6]Simbex, Lebanon, NH, and with Thayer School of Engineering, Dartmouth College, Hanover, NH.

Abstract

The purpose of this study was to quantify the severity of head impacts sustained by individual collegiate football players and to investigate differences between impacts sustained during practice and game sessions, as well as by player position and impact location. Head impacts ($N = 184{,}358$) were analyzed for 254 collegiate players at three collegiate institutions. In practice, the 50th and 95th percentile values for individual players were 20.0 *g* and 49.5 *g* for peak linear acceleration, 1187 rad/s^2 and 3147 rad/s^2 for peak rotational acceleration, and 13.4 and 29.9 for HITsp, respectively. Only the 95th percentile HITsp increased significantly in games compared with practices (8.4%, *p* = .0002). Player position and impact location were the largest factors associated with differences in head impacts. Running backs consistently sustained the greatest impact magnitudes. Peak linear accelerations were greatest for impacts to the top of the helmet, whereas rotational accelerations were greatest for impacts to the front and back. The findings of this study provide essential data for future investigations that aim to establish the correlations between head impact exposure, acute brain injury, and long-term cognitive deficits.

Keywords

football; biomechanics; impacts; magnitude; acceleration

Concussion injuries are a growing and important health care problem in sports (Gerbeding, 2003), affecting approximately 5% of athletes of all ages and at all levels of participation (Collins et al., 1999; Gerbeding, 2003; Guskiewicz et al., 2000; Langlois et al., 2006;

Meehan et al., 2010; Powell & Barber-Foss, 1999a, 1999b; Shankar et al., 2007; Thurman et al., 1998). The current definition of concussion injury is a change in cognitive state preceded by an impact to the head (McCrory et al., 2009); however, to date, the specific association between the biomechanics of a head impact and a concussion injury remains unclear. Developing and documenting measures of head impact biomechanics is one critical step to understanding the cause of concussion injuries.

In sports, exposure to the risk of injury is often reported by the number of athlete-exposures (Dick et al., 2009), defined as one student athlete participating in one practice or competition in which he or she was exposed to the possibility of athletic injury. The athlete-exposure is valuable for assessing the risk of injury due to participation, but it does not account for specific injury mechanisms. With respect to concussion injuries, athlete-exposures cannot capture the magnitude or the frequency of head impacts. For example, two athletes who participate in the same number of games would have the same athlete-exposures, yet they would most likely be exposed to a different number and severity of head impacts.

In our approach to understanding the biomechanical basis of concussion injury we have defined *head impact exposure* as a multifactorial term that includes the frequency of head impacts (e.g., number of head impacts per practice), magnitude of the impacts (e.g., peak linear acceleration), the location (e.g., front of the helmet), and cumulative history of head impacts for an individual athlete. A multifactorial measure of exposure is critical at this juncture because a specific variable or combination of head impact variables that correlate with the risk of brain injury has not yet been determined.

There have been several efforts to measure head impacts in helmeted sports, dating back to the 1970s (Moon et al., 1971; Reid et al., 1971). These early efforts required football players to wear obtrusive data acquisition hardware that interfered with normal play. Consequently, these studies were limited by the number of athletes and the head impact data that were collected. More recently, an accelerometer-based system mounted inside football helmets, the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH, marketed commercially as Sideline Response System by Riddell, Elyria, OH) (Beckwith et al., 2007; Crisco et al., 2004; Manoogian et al., 2006), has been used to directly measure the magnitude of head acceleration and helmet impact location in football players (Broglio et al., 2009; Brolinson et al., 2006; Duma et al., 2005; Funk et al., 2007; Greenwald et al., 2008; Mihalik et al., 2007; Schnebel et al., 2007) during practices and games without interfering with normal play. These studies have provided new insights into the biomechanics of head impacts in football by examining the number of impacts and the magnitude of the resulting head accelerations aggregated within teams and player position.

Previously, we reported that the frequency of head impacts that individual collegiate football players were exposed to varied significantly with player position, team session (game vs. practice), and impact location (Crisco et al., 2010). The majority of players received between 200 and 400 head impacts per season, while some players were exposed to more than 1400. The average number of head impacts sustained in games was nearly three times greater than the number of impacts received in practices. While this study was the first to report the frequency of head impacts for individual collegiate players, the magnitude of those impacts was not reported.

The purpose of this study was to quantify the severity of head impacts to which individual collegiate football players were exposed during practices and games over two seasons. We tested the hypotheses that head impact magnitude differed by team, season, session, player position, and helmet impact location.

Methods

Players from three National Collegiate Athletic Association (NCAA) football programs (Brown University, Dartmouth College, and Virginia Tech) were provided the opportunity to participate in this IRB-approved observational study after informed consent was obtained. During the 2007 and 2008 fall football seasons, a total of 254 male players from the three teams, denoted arbitrarily as Team A ($n = 85$ players), Team B ($n = 83$ players), and Team C ($n = 86$ players), participated in this study. Of these players, 116 were monitored in both seasons. This participant turnover was expected, and due primarily to typical roster fluctuations on a collegiate team (e.g., graduation, incoming freshman, and injuries). Each player was assigned a unique identification number and categorized in one of eight position units defined by the team staff as the player's primary position: defensive line (DL, $n = 39$), linebacker (LB, $n = 38$), defensive back (DB, $n = 46$), offensive line (OL, $n = 60$), offensive running back (RB, $n = 29$), wide receiver (WR, $n = 26$), quarterback (QB, $n = 10$), and Special Teams (ST, $n = 6$).

All players wore Riddell (Riddell, Chicago, IL) football helmets instrumented with the HIT System (Figure 1A), a device capable of recording the acceleration–time history of an impact from six linear acceler-ometers at 1000 Hz. Impact data from all participating institutions were uploaded to a secure central server with a consolidated database, and subsequently exported for statistical analysis. Data were reduced in postprocessing to exclude any impact event with a peak resultant linear head acceleration less than 10 g (Mihalik et al., 2007) to eliminate events that had been determined during initial system development to be inconsequential, nonimpact events (e.g., running and jumping). Any impact event in which the acceleration–time history pattern of the six linear accelerometers did not match the theoretical pattern for rigid body head acceleration (Crisco et al., 2004), such as a spike in a single accelerometer signal that can occur when a player removes his helmet and throws or kicks it, was also excluded. These data reduction methods have been previously verified (Brolinson et al., 2006; Duma et al., 2005; Funk et al., 2007; Manoogian et al., 2006), as was the accuracy of the HIT algorithm (Crisco et al., 2004). Laboratory tests have determined that the linear and rotational accelerations measured by the HIT system were within ± 4% of a helmet-equipped Hybrid III dummy (Duma et al., 2005).

A team session (session) was defined as either a formal team practice (players wore protective equipment with the potential of head contact) or a game (competitions and scrimmages). An individual player was defined to have participated in a session when at least one head impact was recorded for that given player. Impacts that were recorded outside the time of the team session, as defined by the team staff, were excluded from the analysis.

Head impact magnitude was quantified by peak linear acceleration (g) and peak rotational acceleration (rad/s^2). Each recorded impact event was processed using a simulated annealing optimization algorithm to solve for the linear acceleration magnitude at the head center of gravity (CG) (Crisco et al., 2004). Peak rotational acceleration was calculated as the vector product of peak linear acceleration and a point of rotation estimated to be 10 cm inferior to the CG of the head. Laboratory testing has confirmed that this location is consistent with the impact response of the Hybrid III dummy (Duma et al., 2005). Helmet impact location for each impact was computed as azimuth and elevation angles in an anatomical coordinate system relative to the CG of the head (Crisco et al., 2004) and then categorized into one of five helmet impact locations: front (F), left (L), right (R), back (B), and top (T) (Figure 1B). Four equally spaced regions centered on the anatomical midsagittal and coronal planes defined front, left, right, and back impact locations. All impacts occurring above an elevation angle of 65°, where 0° elevation was defined as a horizontal plane through the center of gravity of the head, were defined as impacts to the top of the helmet. In addition, a

nondimensional measure of head impact severity, HITsp (Greenwald et al., 2008), was computed. HITsp transforms the computed head impact measures of peak linear and peak angular acceleration into a single latent variable using principal component analysis, and applies a weighting factor based on impact location (Greenwald et al., 2008).

Statistical Analysis

The 50th and 95th percentile values of the peak linear and peak rotational acceleration were first calculated across the entire study, independent of player. For analysis, individual players' 50th and 95th percentiles were calculated for each impact location (front, left, right, top, and back) within all their practices and within all their games. For HITsp, the 50th and 95th percentiles were calculated for practices and for games without consideration of location. HITsp was not analyzed among impact locations because impact location is a factor in computing HITsp values. The 50th and 95th percentile values were positively skewed (normality test failed, $p < .05$), making general linear models that assume normally distributed variances inappropriate. Therefore, generalized estimating equations for log-normally distributed data were used to model the 50th and 95th percentiles, with repeated measures within players treated as having correlated error with a heterogeneous compound symmetrical variance–covariance matrix for session type × location, block diagonal by season. For peak linear and rotational accelerations, the predictive factors were team, season, season × team, session type, impact location, player position, and the two- and three-way interactions among session type, impact location, and player position. The interactions between season and team were also included to allow for differences in the changes across seasons between institutions. For HITsp, the factors were team, season, team × season, session, player position, and session × player position.

Statistical significance was set at $\alpha = .05$. Given the large number of hypotheses, to minimize type II error, α was only adjusted for multiple comparisons within families of effects (e.g., differences by player position were adjusted without consideration of differences by helmet impact location). These post hoc tests used the Holm-simulated adjustment procedure. All statistical analyses were performed using SAS version 9.2 (SAS Institute, Cary, NC).

Results

Impacts Across Study

A total of 184,358 head impacts, recorded during 412 sessions (330 practices and 82 games), were included in the analysis. The distributions of each measure were heavily skewed ($p < .001$) toward lower values (Figure 2). Across the study, independent of player, the 50th percentile values for peak linear acceleration, peak rotational acceleration, and HITsp were 20.3 g, 1392 rad/ s^2, and 13.7, respectively. The 95th percentile values for peak linear acceleration, peak rotational acceleration, and HITsp were 62.2 g, 4289 rad/s^2, and 32.1, respectively.

Impacts Among Players

A player's position and helmet impact location were the largest factors associated with the differences in the magnitudes of 50th and 95th percentile peak linear acceleration, peak rotational acceleration, and HITsp. Season (2007 vs. 2008) and team (A vs. B vs. C) also differed significantly, but the differences were smaller (Table 1).

Season and Team—Head impact magnitudes decreased from the 2007 to the 2008 season. Significant decreases were in 50th percentile peak linear acceleration (1.8 g, 8.5%), rotational acceleration (99.2 rad/s^2, 8.0%) and HITsp (0.93, 6.7%) (Table 1). Among teams,

there were no differences in the 50th peak linear accelerations, 50th and 95th peak rotational accelerations, or 95th HITsp (Table 1). Differences between teams were marginally significant but small (< 3 g) for 95th percentile peak linear accelerations, with individual team confidence intervals overlapping: 50.8 g (95%CI: 48.6–53.0 g), 47.7 g (95%CI: 45.8–9.8 g), and 50.2 g (95%CI: 48.1–52.4 g) for Teams A, B, and C, respectively. Similarly, the differences in 50th percentile HITsp with Team A (12.9, 95%CI: 12.6–13.1) were significantly less than Teams B (13.6, 95%CI: 13.4–13.8) and C (13.6, 95%CI: 13.4– 13.9), who did not differ. The interactions between season and team were not statistically significant (Table 1).

Practice vs. Games—There were no increases (< 1%) from practices to games in the head impact magnitude, except in the 95th percentile HITsp, which increased significantly in games (Table 1). In practice, the 50th and 95th percentile values were 20.0 g (95%CI: 19.7–20.3 g) and 49.5g (95%CI: 48.0–51.1 g) for peak linear acceleration, 1187 rad/s^2 (95%CI: 1166–1210 rad/s^2) and 3147 rad/s^2 (95%CI: 3043–3253 rad/s^2) for peak rotational acceleration, and 13.4 (95%CI: 13.3–13.6) and 29.9 (95%CI: 29.1–30.8) for HITsp, respectively. In games, the 50th and 95th percentile values were 20.2 g (95%CI: 19.8–20.5 g) and 49.6 g (95%CI: 47.7–51.5 g) for peak linear acceleration, 1197 rad/s^2 (95%CI: 1170–1225 rad/ s^2) and 3145 rad/s^2 (95%CI: 3011–3285 rad/s^2) for peak rotational acceleration, and 13.3 (95%CI: 13.1–13.5) and 32.4 (95%CI: 30.9–33.9) for HITsp, respectively. The 95th percentile HITsp during practices increased significantly, by 8.4%, when compared with games (Table 1).

Positions—There were statistically significant differences in head impact magnitude among player position (Table 1; Figure 3). There were no statistically significant differences in the 50th peak linear acceleration among running backs (RB), linebackers (LB), or quarterbacks (QB), whereas each had significantly greater 50th percentile peak linear acceleration than offensive linemen (OL) and wide receivers (WR) (Figure 3A). The 50th percentile peak linear acceleration was also significantly greater for RB than for defensive backs (DB). These significant differences ranged between 5% (LB over OL) to 16% (QB and RB over ST). Running back had the greatest 95th peak linear acceleration, followed by LB (3.5% less than RB), and DB (8.8% less than RB). The 95th peak linear acceleration for RB, LB, and DB were significantly greater than for OL, WR, and ST. These differences ranged from 13% (DB over OL) to 42% (RB over ST). The DL were 11% less than RB, and DL were significantly greater than WR by 20%. The pattern of differences between player positions in peak rotational acceleration was essentially the same as the pattern of differences between player positions in peak linear acceleration (Figure 3B).

Defensive back, LB, OL, and RB each had significantly greater 50th percentile HITsp than ST and WR (Figure 3C). There were no statistically significant differences in 50th HITsp among DB, LB, OL, and RB. Running back also had significantly greater 95th HITsp than DL, OL, and WR. These differences ranged from 18.5% (RB over DL) to 25% (RB over WR). The 95th percentile HITsp was 19% greater for RB than for QB, but this difference was not statistically significant.

Impact Locations—The magnitude of the head impacts differed significantly among impact locations (Table 1). Peak linear accelerations were greatest for impacts to the top of the helmet, followed by the front, back, and sides (Figure 4A). The 50th percentile peak linear acceleration differed significantly among all helmet impact locations, except between the left and right sides. For the 95th percentile data, the statistical relationships were similar to the 50th percentile except that impacts to the front were not significantly different from those to the back. The 95th percentile peak linear acceleration for impacts to the top were

9.6% (95%CI: 5.9–14%) and 11.5% (95%CI: 6.6–16.1%) greater than the front and the back, respectively, and approximately 30% (95%CI: 26–35%) greater than the sides.

Impacts to the front and to the back (which did not differ from each other) had the greatest peak rotational acceleration (Figure 4B). The 95th percentile peak rotational acceleration for the front and back were both significantly greater (approximately 33%) than for the left and right sides (which did not differ from each other) and for the top (approximately 109% greater).

In games, impacts to the front of the helmet tended to have greater 95th percentile peak linear acceleration than in practice. The 95th percentile peak linear acceleration for impacts to the front of the helmet increased the most among all helmet impact locations in games compared with practice for most player positions (17% for DB, 11.5% for DL, 12.5% for LB, 8.6% for OL). For WR and ST, the increases varied by location, and there was no clear pattern. There were no notable increases for the 95th percentile peak rotational acceleration in games compared with practice for the various impact locations.

Impact Location by Position—There were statistically significant differences in the magnitude of impacts among helmet locations by player position (Table 1). Linebacker and RB, which did not differ significantly from each other, had the greatest 95th percentile peak linear acceleration for front impacts (Table 2). Running back had the greatest 95th percentile peak linear acceleration for the side impact location among all player positions, and this value was significantly greater than OL by approximately 29% and WR by 57%. Quarterback had the greatest 95th percentile peak linear acceleration for impacts to the back of the head among all player positions, but the differences were not statistically significant, and ranged from approximately 34% for ST to 7% for RB.

The 95th peak rotational acceleration from impacts to the front of the helmet was greatest for RB and LB, which did not differ (Table 3). Running back and LB had significantly greater peak rotational accelerations at the front location than OL by approximately 15%. Running back and LB had significantly greater peak rotational acceleration at the side impact location than OL by approximately 36% and WR by approximately 60%.

Discussion

In this study, we examined the magnitude (peak linear acceleration and peak rotational acceleration) and HITsp severity of the head impacts received players from three NCAA collegiate football teams. In our previous study, we found that the median number of head impacts for individual players per practice and per game ranged from 4.8 to 7.5 and 12.1 to 16.3, respectively (Crisco et al., 2010). In addition, we found that offensive linemen had a higher percentage of impacts to the front than to the back of the helmet, whereas quarterbacks had a higher percentage to the back than to the front of the helmet. The present study focused on impact magnitude and the relationships among magnitude, session, player position, and impact location.

The question remains as to what magnitudes and/or quantities of impact accelerations are important clinically, both acutely and cumulatively. The magnitude of the majority of the impacts received by college football players in this study were less than 20 g and 1389 rad/ s^2. The 95th percentile values for peak linear acceleration and peak rotational acceleration were 61 g and 4245 rad/s^2, respectively. These values are below brain injury tolerance levels commonly cited in the literature. Pellman et al. (2003) suggested that impacts greater than 98 g had an 80% probability of resulting in a concussion in NFL players based on 25 laboratory reconstructions of NFL impacts that resulted in a diagnosed concussion. Zhang et

al. (2004) predicted that linear accelerations of 66, 82, and 106 g, and rotational accelerations of 4600, 5900, and 7900 rad/s^2 were associated with 25%, 50%, and 80% probability of clinical diagnosis of concussion based on computer modeling of the NFL data using the Wayne State brain injury model. We propose that impact magnitude, frequency, and location are all critical measures of head impact exposure, and theorize they can be used to more accurately quantify the risk of sustaining concussion or other potentially clinically consequential brain injuries in helmeted athletes than measures based upon participation levels. The current study is one important step in quantifying this risk.

Head impact magnitudes were generally not different for games compared with practices. Broglio et al. (2009) reported that head impacts during games resulted in greater mean linear and rotational accelerations than in practice for high school athletes; however, these differences were small (approximately 1.5 g and 200 rad/ s^2). Their mean values for practices (23.3 ± 14.5 g and 1469 ± 1055 rad/s^2) were similar to our 50th percentile values for individual players. We are cautious about using these values for comparison because of the skewness in the data. We did not detect differences in the magnitude of impacts between games and practices for individual players, except for the 95th percentile HITsp values, although it has been reported that injury rates are greater in games than in practices (Shankar et al., 2007).

A strong association was found between head impact magnitude and player position. This is not unexpected given the different skills and strategies of play required for each position. Running backs (RB) had significantly greater 50th percentile peak linear and peak rotational acceleration than offensive lineman (OL), wide receivers (WR), special teams (ST), and defensive backs (DB), as well as the greatest 95th percentile peak linear acceleration and HITsp. These findings are consistent with those of Mihalik et al. (2007), who reported that offensive backs (OB) were more likely to sustain impacts greater than 80 g than defensive lineman (DL), defensive backs (DB), offensive lineman (OL), linebackers (LB), and wide receivers (WR). Similarly, Schnebel et al. (2007) reported that skilled players (QB, RB, WR, LB, DB) were more likely to sustain impacts with greater magnitude than lineman (OL, DL). Broglio et al. (2009) reported that high school linemen had greater peak rotational acceleration when compared with offensive and defense skill positions. They also reported that defensive linemen and offensive skill players sustained similar-magnitude linear accelerations, but only the defensive line players had greater linear accelerations than the defensive skill and offensive line players. Whether these differences with our findings are due to differences between collegiate and high school players or to our approach in analyzing specific positions require further studies in both populations.

In addition to player position, head impact magnitudes differed by location. Consistent with previous findings (Mihalik et al., 2007, 2008) and (Broglio et al., 2009), peak linear accelerations were greatest to the top of the helmet. Broglio et al. reported that front impacts resulted in greater rotational accelerations than any other impact location (Broglio, et al., 2009). We found no difference in peak rotational accelerations between impacts to the back and to the front of the helmet, but we did find that these impact locations resulted in greater peak rotational accelerations than any other impact location.

The interaction of player position and helmet impact location had a significant effect on head impact magnitude. Linebackers and running backs (LB, RB) had the greatest 95th percentile peak linear and peak rotational accelerations for impacts to the front of the helmet. Running backs (RB) also had the greatest 95th percentile peak linear acceleration for side impacts. For all positions (except for ST), impacts to the top of the helmet had the greatest peak linear acceleration. These data may prove useful in developing a rationale for position-specific helmet designs or protective strategies in the future.

In the current study, we focused our analysis on head impact magnitude for individual players. Our previous study (Crisco et al., 2010) focused specifically on head impact frequency and helmet impact location. Further analysis is needed to determine the relationship between impact frequency and magnitude. If such a relationship is found, it may be possible to reduce head impact exposures, and hence the risk of concussion injury, without significantly altering the sport. In addition, the risk of sustaining a concussion injury may depend not only on a threshold to a single impact but also on an individual's impact history. We are examining cumulative measures of repetitive head impacts as another measure of head impact exposure. Signs and symptom reports from medically diagnosed concussion injuries and other measures of cognitive deficits were not reported here, as a suitably sized cohort of players with diagnosed injuries is still being accumulated and analyzed.

In summary, we found that an individual collegiate football player receives head impacts of varying magnitudes during play, and that the magnitudes of these impacts are heavily skewed toward lower values. Interestingly, the magnitude of impacts during games was not significantly greater than the magnitude of the impacts during practices except for the 95th percentile HITsp. We also found that there were significant differences in the magnitude of impacts among different player positions. Impact location was found to be a factor strongly associated with the differences in magnitude of peak linear acceleration and peak rotational acceleration. The interaction of player position and impact location also yielded statistically significant differences in impact magnitude. This study has provided a detailed description of the magnitude of head impact exposures for collegiate football players that will be critical for establishing the relationship between head impact biomechanics and the risk of concussion injury and for developing appropriate concussion injury prevention strategies.

Acknowledgments

References

Beckwith JG, Chu JJ, Greenwald RM. Validation of a noninvasive system for measuring head acceleration for use during boxing competition. Journal of Applied Biomechanics. 2007; 23:238–244. PubMed. [PubMed: 18089922]

Broglio SP, Sosnoff JJ, Shin S, He X, Alcaraz C, Zimmerman J. Head impacts during high school football: a biomechanical assessment. Journal of Athletic Training. 2009; 44:342–349. PubMed. [PubMed: 19593415]

Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald RM, Duma SM. Analysis of linear head accelerations from collegiate football impacts. Current Sports Medicine Reports. 2006; 5:23–28. PubMed. [PubMed: 16483513]

Collins MW, Lovell MR, McKeag DB. Current issues in managing sports-related concussion. Journal of the American Medical Association. 1999; 282:2283–2285. PubMed. [PubMed: 10612307]

Crisco JJ, Chu JJ, Greenwald RM. An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single-axis accelerometers. Journal of Biomechanical Engineering. 2004; 126:849–854. PubMed. [PubMed: 15796345]

Crisco JJ, Fiore R, Beckwith JG, Chu JJ, Brolinson PG, Duma SM, et al. Frequency and location of head impact exposures in individual collegiate football players. Journal of Athletic Training. 2010; 45:549–559. PubMed. [PubMed: 21062178]

Dick R, Agel J, Marshall SW. National Collegiate Athletic Association Injury Surveillance System commentaries: Introduction and methods. Journal of Athletic Training. 2009; 44:173–182. PubMed.

Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, et al. Analysis of real-time head accelerations in collegiate football players. Clinical Journal of Sport Medicine. 2005; 15:3–8. PubMed. [PubMed: 15654184]

Funk JR, Duma SM, Manoogian SJ, &Rowson S. Biomechanical risk estimates for mild traumatic brain injury. Annual Proceedings— Association for the Advancement of Automotive Medicine. 2007; 51:343–361.

Gerbeding, JL. Report to congress on mild traumatic brain injury in the United States: Steps to preventing a serious public health problem. Atlanta, Georgia: Center for Disease Control and Prevention; 2003. p. 1-5.

Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Journal of Neurosurgery. 2008; 62:789–798. PubMed discussion 798.

Guskiewicz KM, Weaver NL, Padua DA, Garrett WE Jr. Epidemiology of concussion in collegiate and high school football players. American Journal of Sports Medicine. 2000; 28:643–650. PubMed. [PubMed: 11032218]

Langlois JA, Rutland-Brown W, &Wald MM. The epidemiology and impact of traumatic brain injury: a brief overview. The Journal of Head Trauma Rehabilitation. 2006; 21:375–378. PubMed. [PubMed: 16983222]

Manoogian S, McNeely D, Duma SM, Brolinson G, Greenwald RM. Head acceleration is less than 10 percent of helmet acceleration in football impacts. Biomedical Sciences Instrumentation Journal. 2006; 42:383–388. PubMed.

McCrory P, Meeuwisse W, Johnston K, Dvorak J, Aubry M, Molloy M, et al. Consensus statement on concussion in sport: the 3rd International Conference on Concussion in Sport held in Zurich, November. Journal of Athletic Training. 2009; 44:434–448. PubMed. [PubMed: 19593427]

Meehan WP, d'Hemecourt P 3rd, Comstock RD. High school concussions in the 2008–2009 academic year: Mechanism, symptoms, and management. The American Journal of Sports Medicine. 2010

Mihalik JP, Bell DR, Marshall SW, Guskiewicz KM. Measurement of head impacts in collegiate football players: An investigation of positional and event-type differences. Journal of Neurosurgery. 2007; 61:1229–1235. PubMed discussion 1235.

Mihalik JP, Guskiewicz KM, Jeffries JA, Greenwald RM, Marshall SW. Characteristics of head impacts sustained by youth ice hockey players. Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology. 2008; 222:45–52.

Moon DW, Beedle CW, Kovacic CR. Peak head acceleration of athletes during competition-football. Scandinavian Journal of Medicine & Science in Sports. 1971; 3:44–50.

Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF. Concussion in professional football: Reconstruction of game impacts and injuries. Journal of Neurosurgery. 2003; 53:799–812. discussion 812–794.

Powell JW, Barber-Foss KD. Injury patterns in selected high school sports: A review of the 1995–1997 seasons. Journal of Athletic Training. 1999a; 34:277–284. PubMed. [PubMed: 16558577]

Powell JW, Barber-Foss KD. Traumatic brain injury in high school athletes. Journal of the American Medical Association. 1999b; 282:958–963. PubMed. [PubMed: 10485681]

Reid SE, Tarkington JA, Epstein HM, O'Dea TJ. Brain tolerance to impact in football. Surgery, Gynecology & Obstetrics-. Journal of the American College of Surgeons. 1971; 133:929–936.

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. Journal of Neurosurgery. 2007; 60:490–495. discussion 495–496.PubMed.

Shankar PR, Fields SK, Collins CL, Dick RW, Comstock RD. Epidemiology of high school and collegiate football injuries in the United States, 2005–2006. American Journal of Sports Medicine. 2007; 35:1295–1303. PubMed. [PubMed: 17369559]

Thurman DJ, Branche CM, Sniezek JE. The epidemiology of sports-related traumatic brain injuries in the United States: recent developments. The Journal of Head Trauma Rehabilitation. 1998; 13:1–8. PubMed. [PubMed: 9575252]

Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. Journal of Biomechanical Engineering. 2004; 126:226–236. PubMed. [PubMed: 15179853]

Figure 1.
Football players wore helmets instrumented with the HIT system that was specifically designed to record head accelerations as a result of an impact to the helmet without interfering with play. The HIT System (**A**) comprises an in-helmet unit containing six accelerometers (a), battery (b), transmitting and logging electronics (e), a sideline receiver, and a laptop computer. An illustration (**B**) of the regions that defined the front (F), right side (R), back (B) and top (T) helmet impact locations.
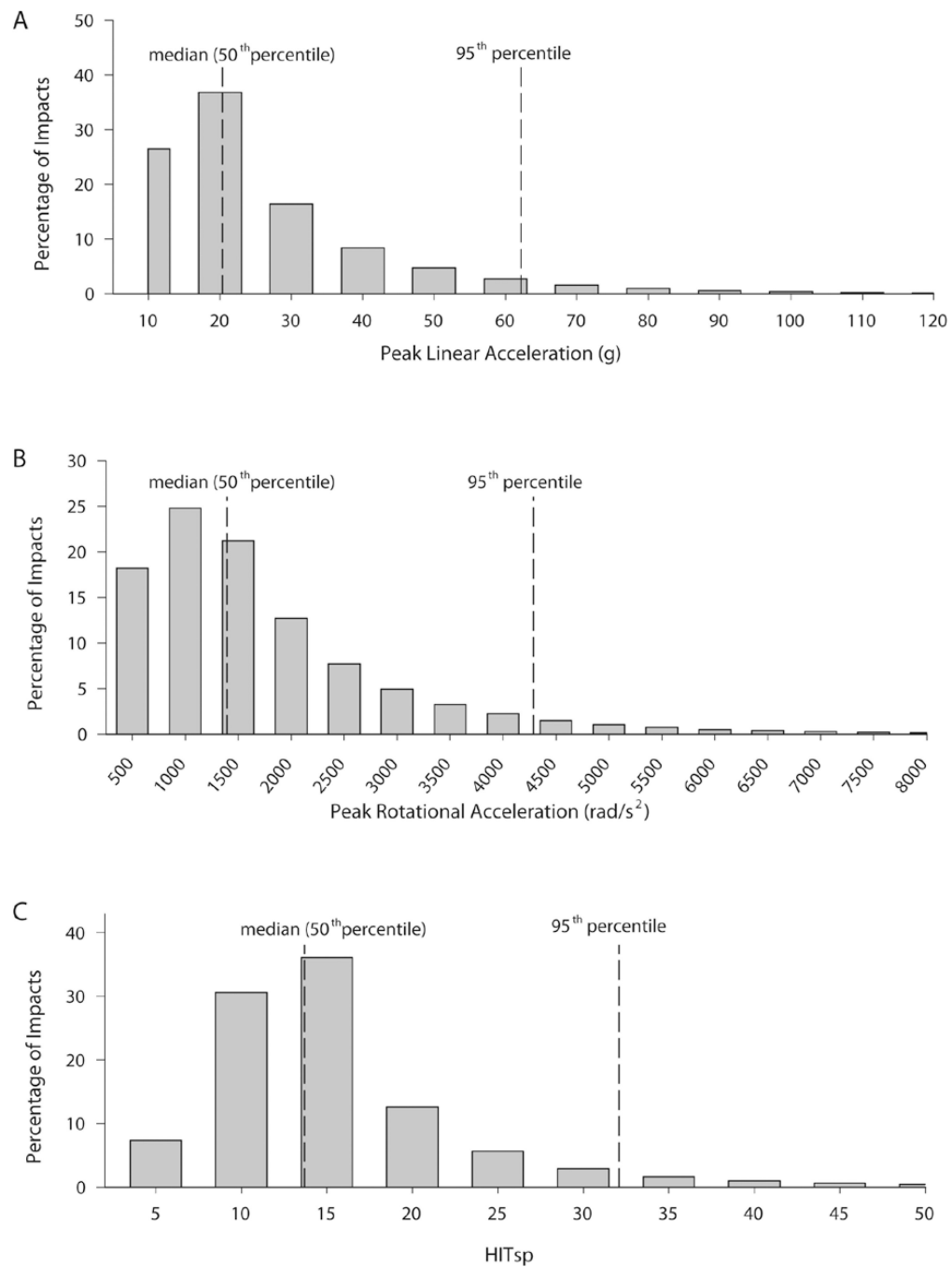
Figure 2.
The distribution of the head impact measures of peak linear acceleration (**A**), peak rotational accelerations (**B**), and HITsp (**C**) were heavily skewed toward lower magnitudes. These distributions, and their associated 50th and 95th percentile values, were computed by aggregating all impacts (*N*= 184,358) recorded in the study. We note the bin size for peak linear acceleration is 10 *g*, except for the first, which binned values of 10 *g* to 15 *g*.
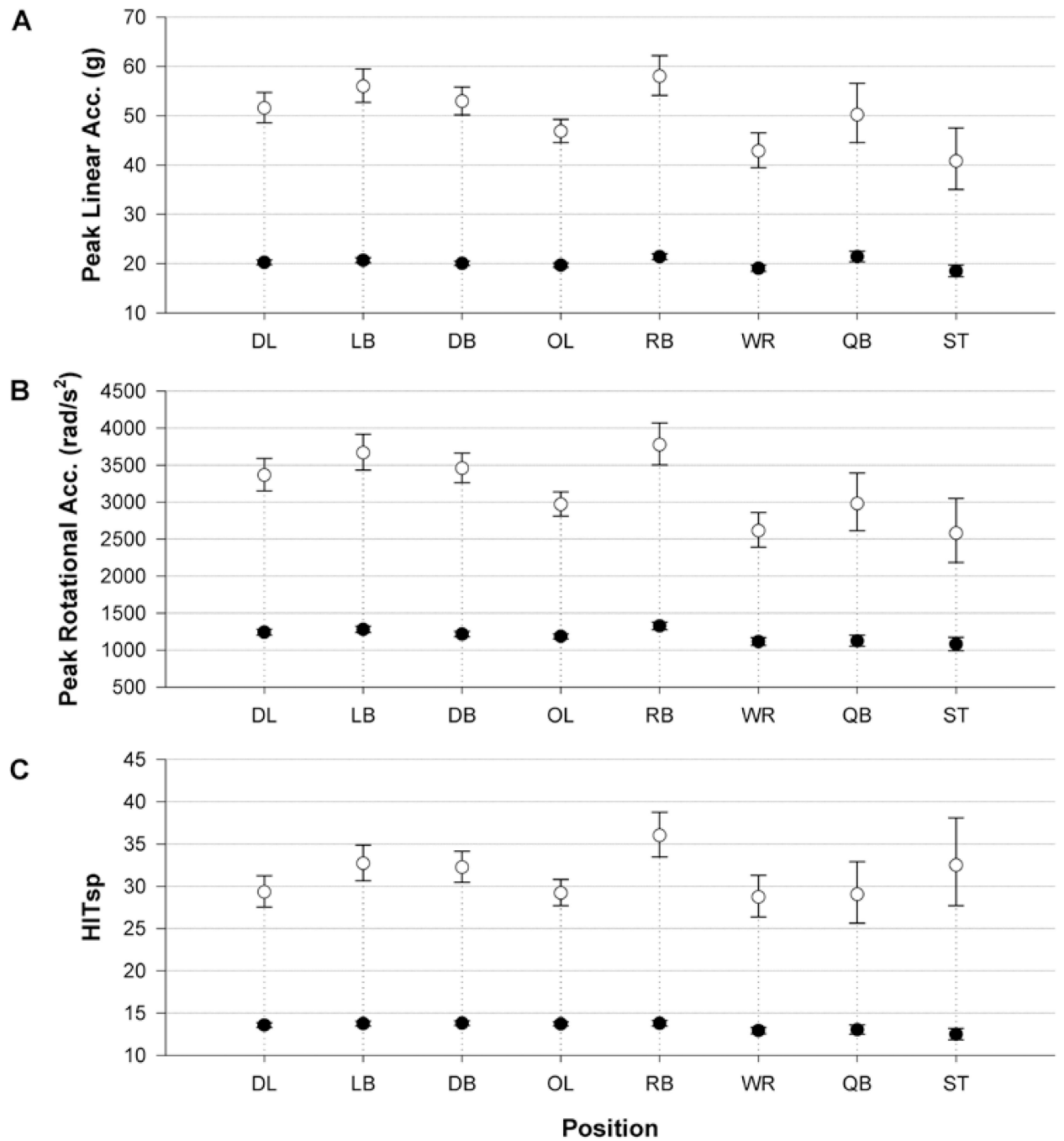
Figure 3.
The 50th percentile (filled circles) and the 95th percentile (open circles) of peak linear acceleration (**A**), peak rotational acceleration (**B**) and HITsp (**C**) differed significantly ($p < .0001$) among player positions. The error bars represent the lower and upper 95% confidence intervals for individual players of that position.
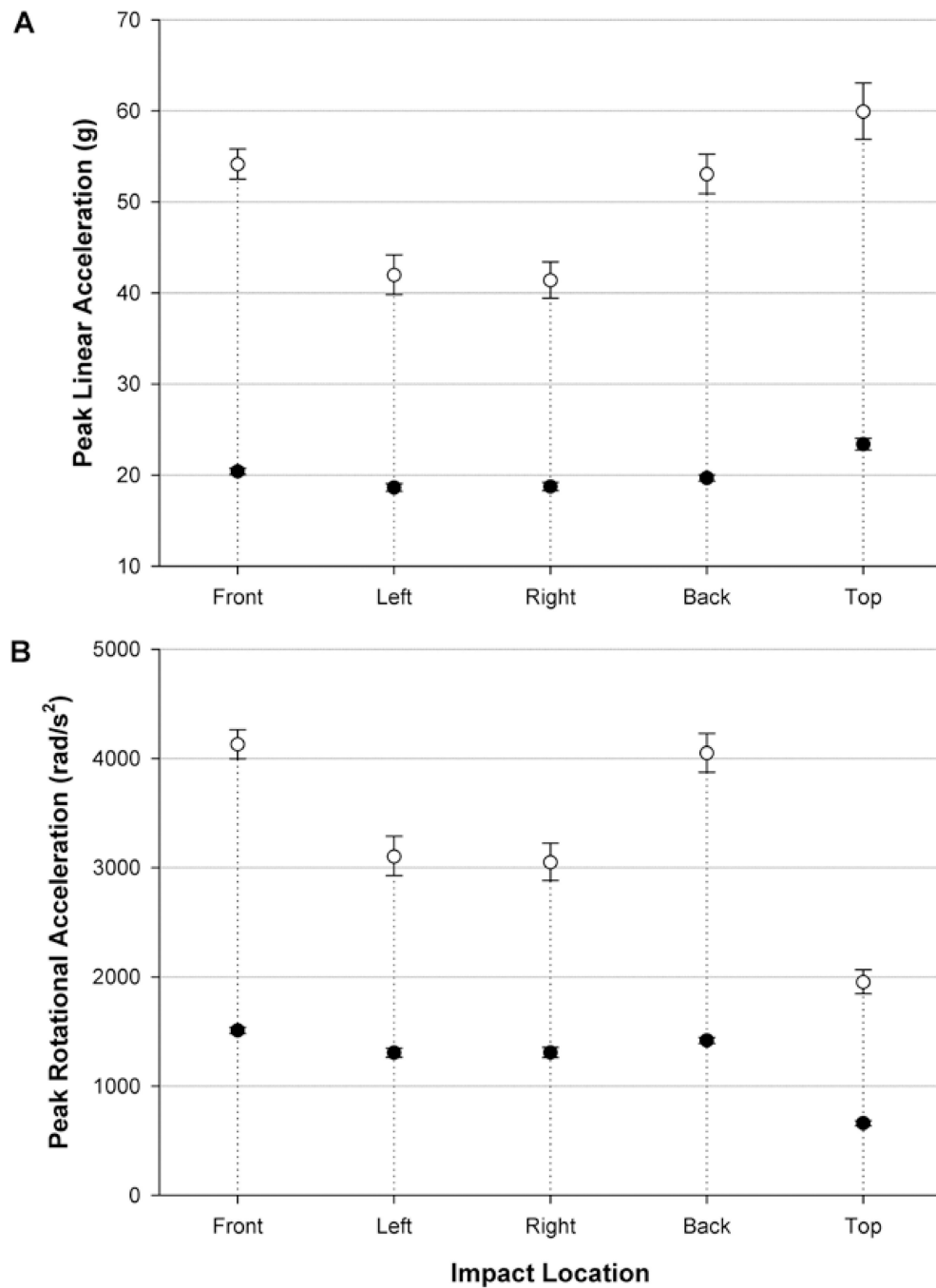
Figure 4.
The 50th percentile (filled circles) and the 95th percentile (open circles) of peak linear acceleration (**A**) and peak rotational acceleration (**B**) differed significantly ($p < .0001$) among impact location. The error bars represent the lower and upper 95% confidence intervals for individual players of that impact location.

Table 1

Post hoc analysis summary of statistical differences (*considered significant) by season, team, session, position, and their interactions

Factors	Peak Linear Acceleration (g)		Peak Rotational Acceleration (rad/s²)		HITsp	
	50th	95th	50th	95th	50th	95th
Season	< 0.0001*	0.1134	< 0.0001*	0.0643	< 0.0001*	0.0641
Team	0.2306	0.0468*	0.0875	0.3134	< 0.0001*	0.1862
Season × Team	0.0762	0.6151	0.1941	0.4386	0.7386	0.1999
Session	0.3972	0.9797	0.4915	0.9832	0.3507	0.0002*
Position	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*	0.0001*	< 0.0001*
Session × Position	0.1335	0.0980	0.0869	0.1579	0.0045*	0.2317
Location	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*		
Session × Location	0.0282*	0.2276	0.0005*	0.4000		
Position × Location	< 0.0001*	< 0.0001*	< 0.0001*	< 0.0001*		
Session × Position × Location	0.0514	0.0147*	0.0276*	0.4481		

Table 2

The 95th percentile (95%CI) peak linear acceleration (g) for individual players of the various positions and helmet impact locations

Impact Location	Player Position							
	DL	LB	DB	OL	RB	WR	QB	ST
Front	53.6 (3.2)	60.7 (3.7)	55.9 (3)	52.6 (2.7)	60.1 (4.2)	46.3 (3.9)	52 (6.7)	53.2(8.7)
Left	45.3 (4.7)	48.7 (5.1)	44.8(4.1)	36.4 (3.2)	54.1 (6.7)	33.8(5.1)	41.2(9)	35.5 (10.4)
Right	40.9 (3.9)	49.1 (4.8)	46.3 (4)	38.5(3.1)	52.1 (5.9)	33.2 (4.7)	39.3 (8.3)	35.3(9.1)
Back	55.4 (4.7)	54.9 (4.7)	55.3 (4.1)	50.7 (3.6)	56.1 (5.5)	49.1 (5.7)	59.8(10.3)	44.5 (9.7)
Top	66.0 (7)	68.9 (7.3)	64.4(6.1)	60.5 (5.4)	68.9 (8.6)	56.4 (8.6)	63.3 (13.9)	38 (10.7)

Note. DL = defensive line; LB = linebacker; DB = defensive back; OL = offensive line; RB = offensive running back; WR = wide receiver; QB = quarterback; ST = special teams.

Table 3

The 95th percentile (95%CI) peak rotational acceleration (rad/s^2) for individual players of the various positions and helmet impact locations

Impact Location	Player Position							
	DL	LB	DB	OL	RB	WR	QB	ST
Front	4070 (256)	4615 (294)	4359 (245)	3992(211)	4727 (348)	3376 (293)	3746 (504)	4316(739)
Left	3532(411)	3817 (454)	3367(348)	2769 (272)	4098 (567)	2434 (416)	2845 (703)	2396 (818)
Right	3125 (356)	3875 (450)	3615 (368)	2732 (262)	3995 (538)	2346 (391)	2681 (681)	2474 (758)
Back	4236 (384)	4197 (384)	4145 (332)	3916 (296)	4307 (456)	3750 (467)	4346 (805)	3555 (838)
Top	2265 (264)	2314(271)	2245 (232)	1953(191)	2298(315)	1682 (284)	1890 (459)	1257 (386)

Note. DL = defensive line; LB = linebacker; DB = defensive back; OL = offensive line; RB = offensive running back; WR = wide receiver; QB = quarterback; ST = special teams.