2011

Head Impact Exposure in Collegiate Football Players

Joseph J. Crisco
Warren Alpert Medical School of Brown University and Rhode Island Hospital

Bethany J. Wilcox
The Warren Alpert Medical School of Brown University and Rhode Island Hospital

Jonathan G. Beckwith M.S.
Simbex

Jeffrey J. Chu M.S.
Simbex

Ann-Christine Duhaime M.D.
Dartmouth Hitchcock Medical Center

See next page for additional authors

Authors
Joseph J. Crisco, Bethany J. Wilcox, Jonathan G. Beckwith M.S., Jeffrey J. Chu M.S., Ann-Christine Duhaime M.D., Steve Rowson, Stefan M. Duma Ph.D., Arthur C. Maerlender, Thomas W. McAllister M.D., and Richard M. Greenwald Ph.D.

Published in final edited form as:
J Biomech. 2011 October 13; 44(15): 2673–2678. doi:10.1016/j.jbiomech.2011.08.003.

Head Impact Exposure in Collegiate Football Players

Joseph J. Crisco, Ph.D.[1], **Bethany J. Wilcox**[1], **Jonathan G. Beckwith, M.S.**[2], **Jeffrey J. Chu, M.S.**[2], **Ann-Christine Duhaime, M.D.**[3], **Steve Rowson**[4], **Stefan M. Duma, Ph.D.**[4], **Arthur C. Maerlender, PhD**[5], **Thomas W. McAllister, M.D.**[6], and **Richard M. Greenwald, Ph.D.**[2,7]

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

[2]Simbex, Lebanon, NH, USA

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

[4]Virginia Tech-Wake Forest, Center for Injury Biomechanics, Blacksburg VA, USA

[5]Pediatric Neuropsychological Services, Dartmouth Medical School, Lebanon, NH, USA

[6]Department of Psychiatry, Dartmouth Medical School, Lebanon, NH, USA

[7]Thayer School of Engineering, Dartmouth College, Hanover, NH, USA

Abstract

In American football, impacts to the helmet and the resulting head accelerations are the primary cause of concussion injury and potentially chronic brain injury. The purpose of this study was to quantify exposures to impacts to the head (frequency, location and magnitude) for individual collegiate football players and to investigate differences in head impact exposure by player position. A total of 314 players were enrolled at three institutions and 286,636 head impacts were recorded over three seasons. The 95th percentile peak linear and rotational acceleration and HITsp (a composite severity measure) were 62.7g, 4378 rad/s^2, and 32.6, respectively. These exposure measures as well as the frequency of impacts varied significantly by player position and by helmet impact location. Running backs (RB) and quarter backs (QB) received the greatest magnitude head impacts, while defensive line (DL), offensive line (OL) and line backers (LB) received the most frequent head impacts (more than twice as many than any other position). Impacts to the top of the helmet had the lowest peak rotational acceleration (2387 rad/s^2), but the greatest peak linear acceleration (72.4 g), and were the least frequent of all locations (13.7%) among all positions. OL and QB had the highest (49.2%) and the lowest (23%.7%) frequency, respectively, of front impacts. QB received the greatest magnitude (70.8g and 5428 rad/s^2) and the most frequent (44% and 38.9%) impacts to the back of the helmet. This study quantified head impact exposure in collegiate football, providing data that is critical to advancing the understanding of the biomechanics of concussive injuries and sub-concussive head impacts.

Correspondence Address: Joseph J. Crisco, Ph.D., Bioengineering Laboratory, Department of Orthopaedics, Alpert Medical School of Brown University and Rhode Island Hospital, CORO West, Suite 404, 1 Hoppin Street, Providence, Rhode Island, 02903, Phone: 401-444-4231, Fax: 401-444-4418, joseph_crisco@brown.edu.

Conflict of Interest Disclosure: Joseph J. Crisco, Richard M. Greenwald, and Simbex have a financial interest in the instruments (HIT System, Sideline Response System (Riddell, Inc)) that were used to collect the biomechanical data reported in this study.

Keywords

biomechanics; frequency; severity; acceleration; helmet; concussion

INTRODUCTION

Impacts to the head are commonly identified as the cause of concussion injury during athletic play (CDC, 1997; McCrory, et al., 2009; Thurman, et al., 1998) while repetitive head impacts, even those with no acute symptoms or signs, often referred to as sub-concussive impacts, have been suggested as a possible cause of chronic brain injury (Janda, et al., 2002). At present, the relationships between head impacts and these brain injuries are not well understood. For example, studies utilizing surrogate reconstructions of documented concussive hits in the National Football League have proposed that the risk of concussion injury is associated with the peak linear acceleration of the head (Pellman, et al., 2003b). Others have postulated that the threshold for concussive injury may be difficult to establish because of the varying magnitudes and locations of impacts resulting in concussion, as well as other factors such as the frequency of sub-concussive impacts and the number of prior concussions (Guskiewicz and Mihalik, 2011). This lack of consensus may be due in part to the challenges of measuring and analyzing head impacts. It also has been suggested that the location of the impact and the direction of the resulting head motion is a factor in the mechanism of concussion injury (Pellman, et al., 2003a). Greenwald et al. (Greenwald, et al., 2008) determined that a weighted measure, HITsp that incorporates linear acceleration, rotational acceleration, impact duration and impact location, was more predictive of concussion diagnosis than any single biomechanical measure. Accordingly, head impact exposure is a risk factor for concussion injury that needs to be quantified, with implications for pathophysiology and for prevention. In our approach to understanding the biomechanical basis of concussion (Crisco, et al., 2010; Crisco, et al., 2011) we have defined “head impact exposure” as a multi-factorial term that includes the frequency of head impacts (e.g. number of head impacts per season), magnitude of the impacts (e.g. peak linear acceleration), impact location (e.g. front of the helmet), and cumulative history of head impacts for an individual athlete. A multi-factorial measure of exposure is critical at this time because the mechanism of acute and chronic brain injury is still not completely understood. Thus, this study is motivated by the need to fully understand and to rigorously quantify measures of head impact exposure.

There have been several efforts to measure head impacts in helmeted sports dating back to the 1970’s (Moon, et al., 1971; Reid, et al., 1974). These early efforts were limited by the available technology, requiring football players to wear obtrusive data acquisition hardware that allowed data collection on only a few athletes in a few sessions. More recently, an accelerometer-based system mounted inside of 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; Duma and Rowson, 2011; 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 positions, and at different levels of play. Previously we analyzed the frequency (Crisco, et al., 2010) and magnitude (Crisco, et al., 2011) of head impacts for individual collegiate football players, but did not examine the relationships between these measures of exposure.

Building upon these previous studies, and expanding the data collection to three seasons, the purpose of the current study was to examine head impact exposure by quantifying the frequency of the impacts, the location of the impacts on the helmet, and the magnitude of impacts to individual collegiate football players among various player positions. Specifically, we tested the hypotheses that head impact frequency, location and magnitude would not differ by player position.

METHODS

During the 2007, 2008 and 2009 fall football seasons, a total of 314 players from three National Collegiate Athletic Association (NCAA) football programs (Brown University, Dartmouth College, and Virginia Tech) participated in this observational study after informed consent was obtained with institutional review board approval. Of these players, 146, 106 and 62 were monitored during one, two and three seasons, respectively. This participant turnover was expected, and due primarily to typical roster fluctuations on a collegiate team. Each player was assigned a unique identification number and categorized into one of eight position units defined by the team staff as the player's primary position: defensive line (DL, n = 49), linebacker (LB, n = 47), defensive back (DB, n = 55), offensive line including tight ends (OL, n = 75), offensive running back (RB, n = 37), wide receiver (WR, n = 30), quarterback (QB, n = 14), and Special Teams (ST, n = 7), which were not included in this analysis because of the relatively low number of players.

All players wore Riddell VSR-4, Revolution, or Speed (Riddell, Chicago IL) football helmet models that were instrumented with the HIT System. The HIT System is an accelerometer-based device that computes linear and rotational acceleration of the center of gravity (CG) of the head, as well as impact location on the helmet (Beckwith, et al., 2007; Crisco, et al., 2004; Manoogian, et al., 2006). The HIT System is specifically designed to measure head accelerations by elastically coupling the accelerometers to the head, isolating them from the helmet shell. Data were reduced in post-processing to exclude any acceleration event with peak resultant linear head acceleration less than 10g in order to eliminate head accelerations from non-impact events (e.g. running, jumping, etc.) (Ng, et al., 2006). Data reduction methods are described in detail elsewhere (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 impact tests of a Hybrid III dummy fitted with all helmets instrumented with the HIT System determined that the linear and rotational accelerations measured by the HIT System were within ± 4% of those measured concurrently by the internally instrumented Hybrid III headform (Duma, et al., 2005).

Head impact exposure was defined for each individual player using measures of impact frequency, location and magnitude. 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 within the specified time of the team session. Five measures of impact frequency were computed: practice impacts was the total number of head impacts for a player during all practices; game impacts was the total number of head impacts for a player during all games; impacts per season was the total number of head impacts for a player during all team sessions in a single season; impacts per practice was the average number of head impacts for a player during practices; and impacts per game was the average of the number of head impacts for a player during games.

Impact locations to the helmet and facemask were computed as azimuth and elevation angles in an anatomical coordinate system relative to the center of gravity of the head (Crisco, et al., 2004) and then categorized as front, side (left and right), back, and top. Front, left, right and back impact locations were four equally spaced regions centered on the mid-sagittal plane All impacts above an elevation angle of 65° from a horizontal plane through the CG of the head were defined as impacts to the top of the helmet (Greenwald, et al., 2008).

Impact magnitude was quantified by peak linear acceleration (g) and peak rotational acceleration (rad/s^2) (Crisco, et al., 2004). Peak rotational acceleration was calculated as the vector product of peak linear acceleration and a point of rotation 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). Additionally, a non-dimensional 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). It thus serves as a measure of impact severity, with weight given to factors shown in previous head injury research (linear and rotational acceleration, impact duration and location (Gadd, 1966; Gennarelli, et al., 1972; Hodgson, 1970; Pellman, et al., 2003a)) to predict increased likelihood of clinical or structural injury. Impacts were further reduced for analysis by computing the 95th percentile value of all seasonal impacts for each individual player.

Statistical Analysis

Results were expressed as median values and [25–75% interquartile range], because each study variable was not normally distributed (Shapiro-Wilk test; $P < 0.001$). The significance of the differences among player positions in impact frequency (impacts per season) and in severity measures (95th percentile peak linear acceleration, 95th percentile rotational acceleration, and 95th HITsp) were examined separately using a Kruskal-Wallis one-way ANOVA on ranks with a Dunn's post-hoc test for all pairwise comparisons. Statistical significance was set at $\alpha = 0.05$ and the reported P values are those for the post hoc test. An identical approach was used to examine the significance of the differences among player positions in frequency and the 95th percentile peak linear and rotational acceleration at each location. Statistical comparison among impact location were performed with a Friedman repeated measures ANOVA on ranks. All statistical analyses were performed using SigmaPlot (Systat Software, Chicago, IL).

RESULTS

A total of 286,636 head impacts were analyzed in this study. These data were collected during a median of 50 [28–76.5] practices and 12 [6–20] games (including scrimmages) for all players. Impact magnitudes across the study were heavily skewed to lower values ($P < 0.001$) with a 50th and 95th percentile peak linear acceleration of 20.5g and 62.7g, respectively, 50th and 95th percentile peak rotational acceleration of 1400 rad/s^2 and 4378 rad/s^2, respectively, and 50th and 95th percentile HITsp of 13.8 and 32.6 (Figure 1). The total number of impacts received by an individual player during a single season was a median of 420 [217–728], with a maximum of 2492. The total number of impacts that players received in a single season during practices was 250 [131–453], with a maximum of 1807, and during games was 128 [47–259], with a maximum of 1683. The frequency of impacts players received were further analyzed by normalizing the number of impacts by the number of sessions for each individual player because of differences in team schedules and player attendance.

After grouping players by their primary position and analyzing impacts over all locations, the number of impacts per season ranged from a median of 149 [96–341] for QB to 718 [468– 1012] for DL (Figure 2). Across all player positions, there was a linear increase in median impacts per practice with median impacts per season (slope = 0.02, R^2 = 0.934) and in median impacts per game with median impacts per season (slope = 0.04, R^2 = 0.929). DL, LB and OL received the highest number of impacts per season, and QB and WR the lowest. The median 718 impacts per season received by DL were significantly ($P < 0.05$) more than QB, WR (157 [114–245]), RB (326 [256–457]), and DB (306 [204–419]), but not different than LB (592 [364–815]) or OL (543 [264–948]). LB received significantly ($P < 0.05$) more impacts per season than QB, WR, DB and RB. OL received significantly ($P < 0.05$) more impacts per season than QB, WR and DB.

RB received the impacts with greatest magnitude accelerations and highest HITsp values. The 95th percentile peak linear and rotational acceleration for RB were significantly ($P < 0.05$) greater than OL, DL and DB (Figure 3 and Table 1). The 95th percentile HITsp for RB and LB were also significantly ($P < 0.05$) greater than OL and DL (Table 1). Although OL and DL received the most frequent impacts per season, the magnitudes of the impacts were the least of all player positions. The median 95th percentile peak linear and rotational accelerations values were greatest for QB, but these were not significantly different from the other player positions (Figure 3).

The magnitudes of impacts to the front of the helmet were significantly ($P < 0.05$) greater for RB than for OL, DL, WR and DB (Figure 4). LB, which were not different from RB, received significantly ($P < 0.05$) greater magnitude front impacts than OL and DL. Although the magnitudes were the lowest, OL received significantly ($P < 0.05$) more front impacts than QB, WR, DL, and LB. QB received significantly ($P < 0.05$) fewer front impacts than all player positions except WR. The magnitude of impacts to the side of the helmet was the greatest for RB, which were significantly greater than OL, DL and WR. Side impacts were significantly ($P < 0.05$) less frequent and less severe for OL than all player positions. The median 95th percentile peak linear and rotational accelerations values associated with impacts to the top of the helmet were greatest for QB, but these were not significantly different among player positions, while RB received impacts that had significantly ($P < 0.005$) greater peak linear accelerations and significantly ($P < 0.02$) greater peak rotational accelerations than OL. For all player positions, top impacts were the least frequent ($P<0.05$) impact location (approximately 13% of all head impacts), but were associated with the greatest ($P<0.05$) peak linear acceleration magnitudes of all impact locations. In contrast, peak rotational accelerations associated with top impacts had significantly ($P<0.05$) lower magnitudes than all locations for all player positions (Table 2). Impacts to the back of the helmet tended to be the highest magnitude for the QB and WR, and were significantly ($P < 0.05$) more frequent for QB and WR than for all other positions.

DISCUSSION

The purpose of this study was to quantify head impact exposure in individual collegiate football players and then examine the relationships between head impact frequency, location and magnitude as a function of player position. Quantifying head impact exposures is a critical step in achieving our long-term goals of understanding the biomechanical basis for mild traumatic brain injuries (concussion injuries), correlating head impact exposure with the clinical variables associated with these injuries, and understanding the acute and long-term effect of repeated sub-concussive impacts.

Player position had the most significant effect on head impact exposure in this study of collegiate football players. These differences across player position were considerably greater than the differences we previously reported among the teams and season in both frequency (Crisco, et al., 2010) and severity (Crisco, et al., 2011) of head impacts. The difference among teams in the median head impact frequency was approximately 100 impacts per season. This difference may simply be due to random effects or possibly to the structure of the practice plan and the philosophies of the coaching staff. The increase in impacts per game over impacts per practice was approximately a factor of two for each player position (Figure 2), which is less than, but still consistent with, our previous study of a smaller cohort (Crisco, et al., 2010). Our values for the number of head impacts per practice and impacts per game bracket those values reported by others when we consider the range of our values among our player positions (Brolinson, et al., 2006; Duma, et al., 2005; Mihalik, et al., 2007; Schnebel, et al., 2007). Recently, Guskiewicz et al. (Guskiewicz and Mihalik, 2011) reported that the average collegiate football player experiences 950 impacts per season. Schnebel et al. (Schnebel, et al., 2007) reported an average of 1353 impacts per season per player in collegiate football players, while in high school players Broglio et al. (Broglio, et al., 2009) reported 565 impacts per season per player and Schnebel et al. (Schnebel, et al., 2007) reported 520 impacts per season per player. The discrepancy with the median of 420 impacts per season that we report may be associated with differences in player participation, but may also be due in part to their computations used to determine the average exposure per player, which was simply dividing the total number of impact recorded in their study by the number of players in their study. Our values for frequency are computed for each individual player.

While DL, LB, and OL were found to have the lowest head impact magnitudes of all player positions, they had the greatest number of head impacts. This is in agreement with previous reports that offensive and defensive linemen sustain the most frequent head impacts (Broglio, et al., 2009; Mihalik, et al., 2007; Schnebel, et al., 2007), with relatively low severity (Schnebel, et al., 2007). QB received severe impacts (Figure 3) but these were not statistically greater than the other positions, most likely because of the low sample size and the large distributions of values. Mihalik et al. reported that OL received higher linear accelerations than DL and DB (Mihalik, et al., 2007); however, it should be noted that these comparisons were based upon the mean values, where as our comparison were based upon those of individual players and their median 95th percentile value because of the non-normality of the positively skewed dataset (Figure 1). In agreement with anecdotal observations of football games, OL received the largest percentage of impacts to the front of the helmet and the smallest to the back, while QB received the largest percentage of impacts to the back of the helmet, with magnitudes comparable to the greatest front and side impacts of all positions. RB had the greatest magnitude impacts to the front, side and top of the helmet relative to other position, and this is consistent with previous reports (Mihalik, et al., 2007; Schnebel, et al., 2007). As previously reported (Broglio, et al., 2009; Crisco, et al., 2011; Mihalik, et al., 2007; Mihalik, et al., 2008), impacts to the top of the helmet generated the highest peak linear accelerations, while being associated with the least rotational acceleration among all player positions. The lower rotational accelerations associated with impacts to the top of the helmet is most likely due to the reduced length of the lever arm for neck sagittal and coronal rotations with impact vectors aligned with the central axis of the cervical spine.

This study has several limitations. Our study only captured the primary position as reported by the team. The number of players who play both offense and defense is low at the collegiate level, and the error due to this is considered to be small compared to the total number of impacts recorded. The differences among teams and season were small compared with the differences among player positions that we reported in this study. Post-processed linear head CG acceleration less than 10g were excluded from our analysis. This threshold reduces the number of impacts recorded but eliminates acceleration levels typically experienced by athletes during normal, non-injurious activities such as quickly standing up, running, and jumping (Ng, et al., 2006). The HIT System provides the location of the impact in spherical coordinates, but we reduced these coordinate to four locations (front, side, top

NIH-PA Author Manuscript

NIH-PA Author Manuscript

and back). Impacts to the front of the helmet can be analyzed further by those located on the helmet and on the facemask. We found the median 95th percentile peak linear acceleration and rotational acceleration values were within 1 g and 100 rad/s^2, respectively, among these impact locations, so all front impacts were grouped. Head impact exposure metrics developed from this data set, including a highly ranked Division I team and two Ivy League teams, may be representative of different levels of collegiate football, but these data may not be readily extrapolated to other levels of play, such as high school or youth football. Finally, this study does not examine the relationships between head impact exposure and either diagnosis of concussion injury or return to play following brain injury diagnosis. We do note that injury rates have been reported higher in games than in practices (Shankar, et al., 2007). While we have reported a two-fold increase in the number of head impacts per game over practice and an increase in 95th percentile HITsp in games over practices, the correlations between these factors and concussion rates remains to be demonstrated.

Acknowledgments

We appreciate and acknowledge the researchers and institutions from which the data were collected, Lindley Brainard and Wendy Chamberlain, Simbex, Mike Goforth, ATC, Virginia Tech Sports Medicine, Steve Rowson, MS, Virginia Tech, Dave Dieter, Edward Via Virginia College of Osteopathic Medicine, Jeff Frechette ATC, Scott Roy ATC, and Michael Derosier, ATC, Dartmouth College Sports Medicine, Mary Hynes, R.N., MPH, and Nadee Siriwardana, Dartmouth Medical School, Russell Fiore, MEd, ATC and David J. Murray, ATC, Brown University.

Funding: NIH R01HD048638, RO1NS055020, R25GM083270 and R25GM083270-S1, and the National Operating Committee on Standards for Athletic Equipment (NOCSAE 04-07).

REFERENCES

Beckwith JG, Chu JJ, Greenwald RM. Validation of a noninvasive system for measuring head acceleration for use during boxing competition. J Appl Biomech. 2007; 23(3):238–244. [PubMed: 18089922]

Broglio SP, Sosnoff JJ, Shin S, He X, Alcaraz C, Zimmerman J. Head impacts during high school football: a biomechanical assessment. J Athl Train. 2009; 44(4):342–349. [PubMed: 19593415]

Brolinson PG, Manoogian S, McNeely D, Goforth M, Greenwald RM, Duma SM. Analysis of linear head accelerations from collegiate football impacts. Curr Sports Med Rep. 2006; 5(1):23–28. [PubMed: 16483513]

CDC. Sports-related recurrent brain injuries--United States. MMWR Morb Mortal Wkly Rep. 1997; 46:224–227. [PubMed: 9082176]

Crisco JJ, Chu JJ, Greenwald RM. An algorithm for estimating acceleration magnitude and impact location using multiple nonorthogonal single-axis accelerometers. J Biomech Eng. 2004; 126(6): 849–854. [PubMed: 15796345]

Crisco JJ, Fiore R, Beckwith JG, Chu JJ, Brolinson PG, Duma S, McAllister TW, Duhaime AC, Greenwald RM. Frequency and location of head impact exposures in individual collegiate football players. J Athl Train. 2010; 45(6):549–559. [PubMed: 21062178]

Crisco JJ, Wilcox BJ, Machan JT, McAllister TW, Duhaime AC, Duma S, Rowson S, Beckwith JG, Chu JJ, Greenwald RM. Head Impact Severity Measures in Collegiate Football Players. J Applied Biomechanics. 2011 (In Press).

Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, Greenwald RM, Chu JJ, Crisco JJ. Analysis of real-time head accelerations in collegiate football players. Clin J Sport Med. 2005; 15(1):3–8. [PubMed: 15654184]

Duma SM, Rowson S. Past, present, and future of head injury research. Exerc Sport Sci Rev. 2011; 39(1):2–3. [PubMed: 21183827]

Funk JR, Duma SM, Manoogian SJ, Rowson S. Biomechanical risk estimates for mild traumatic brain injury. Annu Proc Assoc Adv Automot Med. 2007; 51:343–361. [PubMed: 18184501]

Gadd, CW. Use of a weighted-impulse criterion for estimating injury hazard; Proc. 10th Stapp Car Crash Conf; New York: Society of Automotive Engineers; 1966. p. 164-174.

Gennarelli, TA.; Thibault, LE.; Ommaya, AK. Pathophysiologic responses of rotational and translational accelerations of the head; Proceedings of the 16th STAPP Car Crash Conference; New York: Society of Automotive Engineers; 1972. p. 296-308.

Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 2008; 62(4):789–798. discussion 798. [PubMed: 18496184]

Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. Exerc Sport Sci Rev. 2011; 39(1):4–11. [PubMed: 21088602]

Hodgson VR. Testing the Validity and Limitations of the Severity Index. Stapp Car Crash J. 1970

Janda DH, Bir CA, Cheney AL. An evaluation of the cumulative concussive effect of soccer heading in the youth population. Inj Control Saf Promot. 2002; 9(1):25–31. [PubMed: 12462161]

Manoogian S, McNeely D, Duma S, Brolinson G, Greenwald R. Head acceleration is less than 10 percent of helmet acceleration in football impacts. Biomed Sci Instrum. 2006; 42:383–388. [PubMed: 16817638]

McCrory P, Meeuwisse W, Johnston K, Dvorak J, Aubry M, Molloy M, Cantu R. Consensus statement on concussion in sport: the 3rd International Conference on Concussion in Sport held in Zurich, November 2008. J Athl Train. 2009; 44(4):434–448. [PubMed: 19593427]

Mihalik JP, Bell DR, Marshall SW, Guskiewicz KM. Measurement of head impacts in collegiate football players: an investigation of positional and event-type differences. Neurosurgery. 2007; 61(6):1229–1235. [PubMed: 18162902]

Mihalik JP, Guskiewicz KM, Jeffries JA, Greenwald RM, Marshall SW. Characteristics of head impacts sustained by youth ice hockey players. IMechE. 2008; 222:45–52.

Moon DW, Beedle CW, Kovacic CR. Peak head acceleration of athletes during competition--football. Med Sci Sports. 1971; 3(1):44–50. [PubMed: 5095335]

Ng TP, Bussone WR, Duma SM. The effect of gender and body size on linear accelerations of the head observed during daily activities. Biomed Sci Instrum. 2006; 42:25–30. [PubMed: 16817580]

Pellman EJ, Viano DC, Tucker AM, Casson IR. Concussion in professional football: location and direction of helmet impacts-Part 2. Neurosurgery. 2003a; 53(6):1328–1340. discussion 1340-1321. [PubMed: 14633299]

Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF. Concussion in professional football: reconstruction of game impacts and injuries. Neurosurgery. 2003b; 53(4):799–812. discussion 812-794. [PubMed: 14519212]

Reid SE, Epstein HM, O'Dea TJ, Louis MW, Reid SE Jr. Head protection in football. J Sports Med. 1974; 2(2):86–92. [PubMed: 1186180]

Schnebel B, Gwin JT, Anderson S, Gatlin R. In vivo study of head impacts in football: a comparison of National Collegiate Athletic Association Division I versus high school impacts. Neurosurgery. 2007; 60(3):490–495. [PubMed: 17327793]

Shankar PR, Fields SK, Collins CL, Dick RW, Comstock RD. Epidemiology of high school and collegiate football injuries in the United States, 2005–2006. Am J Sports Med. 2007; 35(8):1295–1303. [PubMed: 17369559]

Thurman DJ, Branche CM, Sniezek JE. The epidemiology of sports-related traumatic brain injuries in the United States: recent developments. J Head Trauma Rehabil. 1998; 13(2):1–8. [PubMed: 9575252]

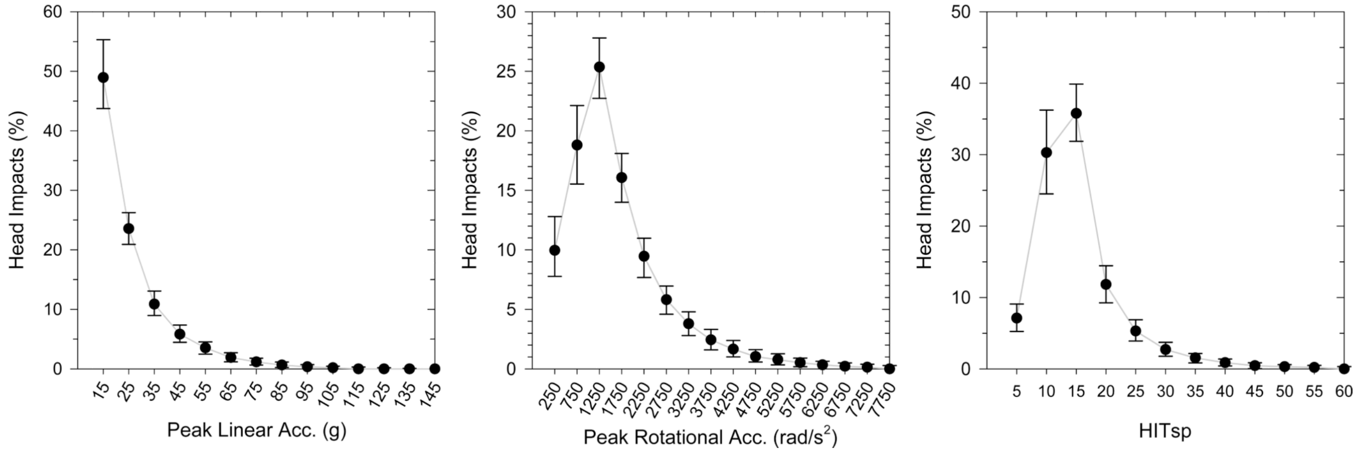

Figure 1.
Study wide peak linear acceleration (g), peak rotational acceleration (rad/s^2) and HITsp distributions of head impacts. Data are a percentage of all impacts for individual players with median [25–75%] values plotted at each bin in the distribution.

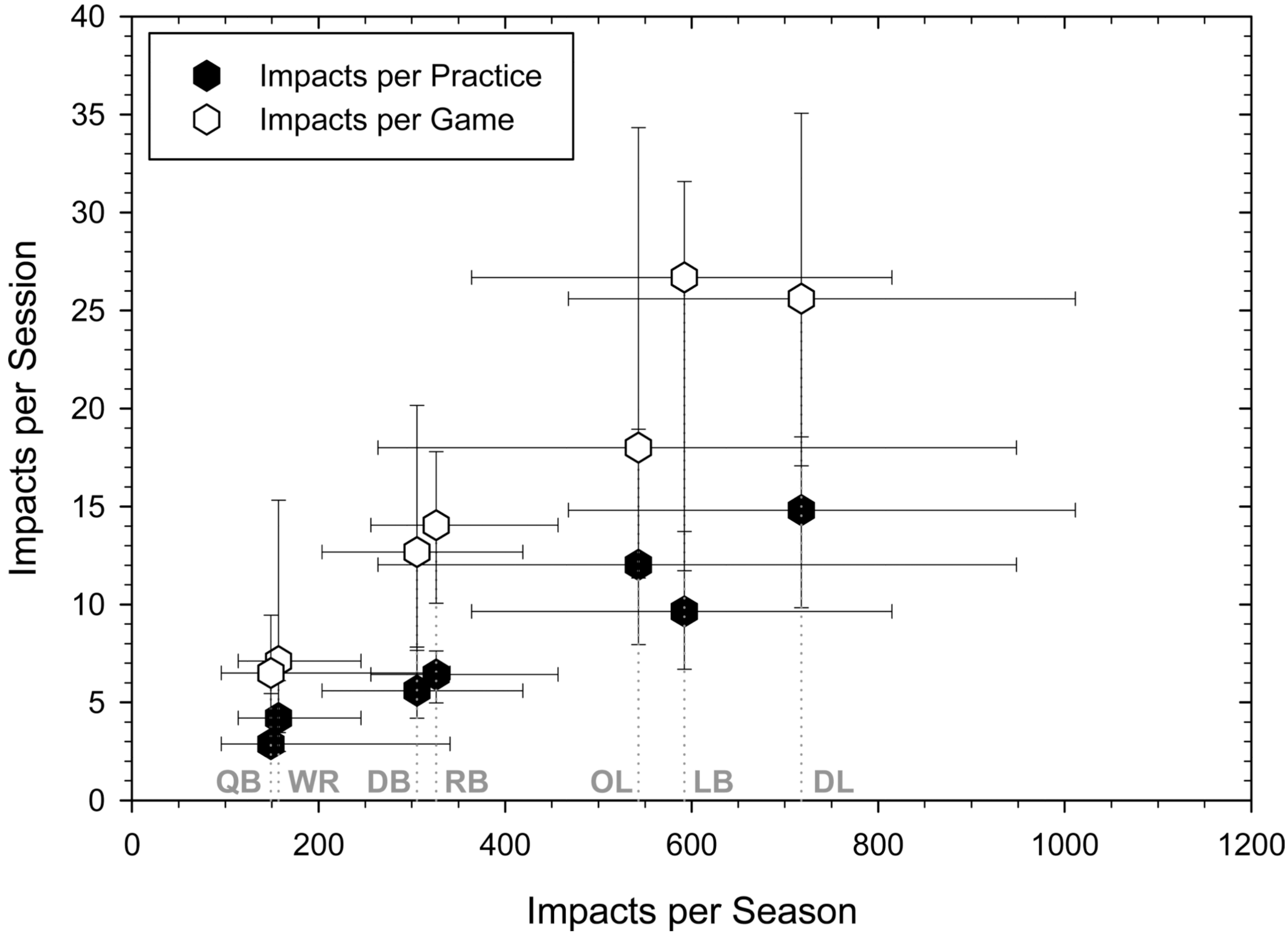

Figure 2.
After categorizing by player position, the median [25% – 75%] frequency of head impacts per practice and head impacts per game were linearly correlated with the median [25% – 75%] frequency of head impacts per season (slope = 0.02, R^2 = 0.934 and slope = 0.04, R^2 = 0.929, respectively). The vertical dotted lines identify the positions associate with each median value.

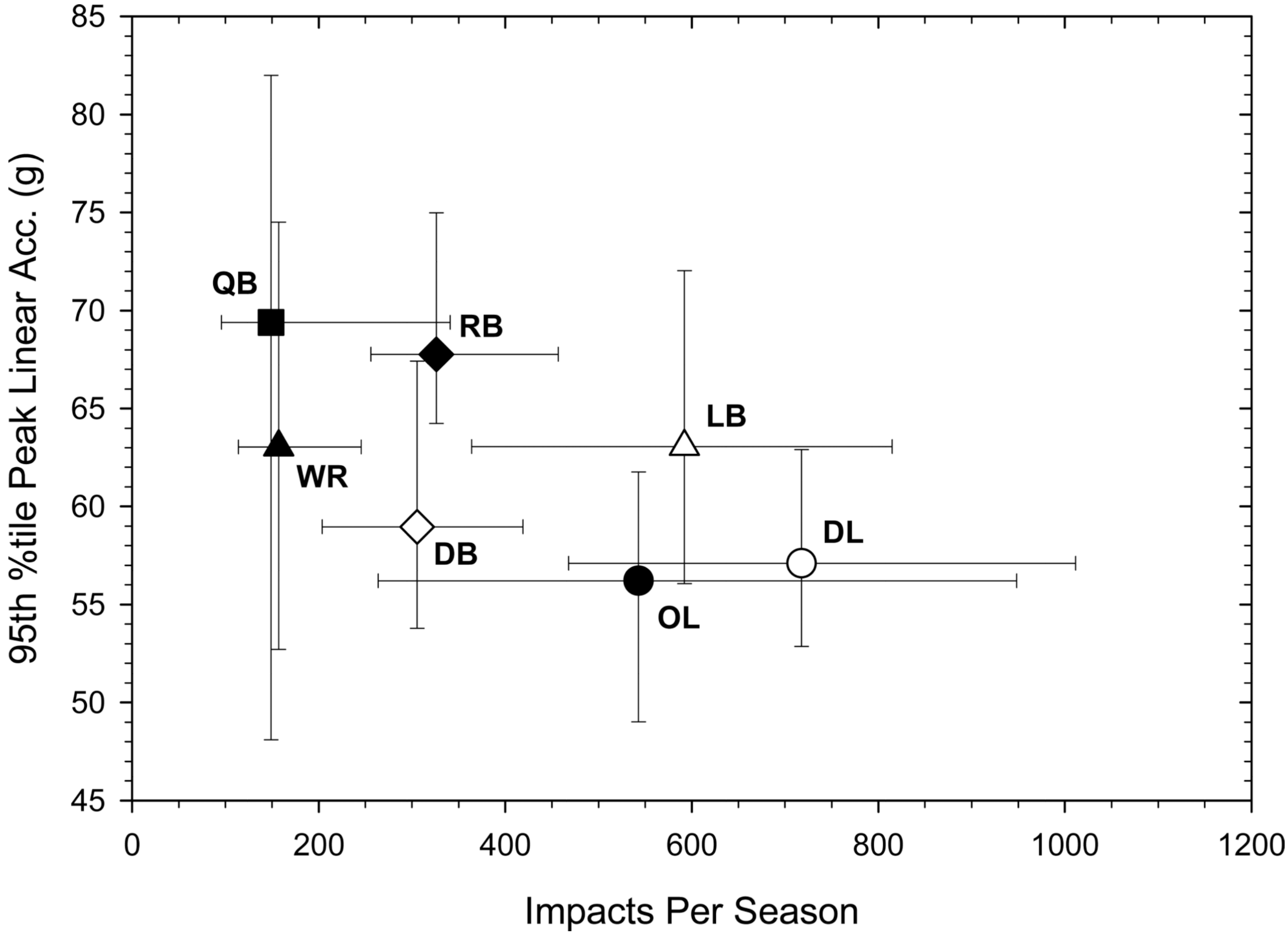

Figure 3.
The median [25%– 75%] of the 95th percentile of peak linear acceleration (g) as a function of the median [25%– 75%] number of head impacts per season and categorized by player position. Analogous values for peak rotational acceleration (rad/s^2) and HITsp are provided in Table 1.

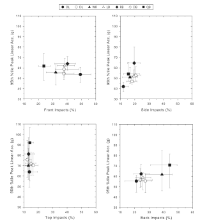

Figure 4.
The median [25%– 75%] of the 95th percentile of peak linear acceleration (g) as a function of the median [25%– 75%] frequency of impacts at each helmet location and categorized by player position. Analogous values for peak rotational acceleration (rad/s^2) are provided in Table 2.

Table 1

The median [25–75%] of the 95th percentile peak rotational acceleration (rad/s^2) and HITsp among the various player positions. The values for 95th peak linear acceleration and the number of impacts per season are plotted in Figure 3.

	Player Position						
	QB	**WR**	**DB**	**RB**	**OL**	**LB**	**DL**
Peak Rotational Acc. (rad/s^2)	4956 [3728-5451]	4297 [3478-5122]	4269 [3968-4665]	4869 [4477-5343]	3799 [3401-4380]	4359 [3823-4837]	3891 [3493-4388]
HITsp	34.5 [25.6-39.3]	31 [24.5-35.8]	31.9 [29.4-34.6]	36.1 [32.6-40.2]	29 [25.3-32.1]	32.6 [30.3-36.6]	28.9 [27-32.5]

Table 2

The median [25–75%] of the 95th percentile peak rotational acceleration (rad/s^2) for each player position and impact location. Values for 95th peak linear acceleration and the percentage of impacts at each location are plotted in Figure 4.

	Player Position						
Impact Location	**QB**	**WR**	**DB**	**RB**	**OL**	**LB**	**DL**
Front	4548 [3399-5500]	3917 [2785-4914]	4573 [3946-4920]	4953 [4441-5786]	4033 [3487-4606]	4810 [4359-5376]	4129 [3534-4582]
Side	3956 [3236-4928]	3739 [3166-4469]	4086 [3636-4729]	5089 [4140-5815]	3129 [2638-3504]	4166 [3672-4682]	3637 [3378-3783]
Top	2802 [1536-3314]	2219 [1599-2921]	2695 [1931-3276]	2688 [2243-3270]	2125 [1717-2626]	2339 [1997-2812]	2487 [2093-2944]
Back	5428 [3985-6156]	4930 [3595-6365]	4184 [3487-5003]	4801 [3845-5492]	4145 [3456-4784]	4242 [3561-5047]	4074 [3766-5116]