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Head Impacts During High School Football: A Biomechanical Assessment

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Context: Little is known about the impact biomechanics sustained by players during interscholastic football.

Objective: To characterize the location and magnitude of impacts sustained by players during an interscholastic football season.

Design: Observational design.

Setting: On the field.

Patients or Other Participants: High school varsity football team (n = 35; age = 16.85 ± 0.75 years, height = 183.49 ± 5.31 cm, mass = 89.42 ± 12.88 kg).

Main Outcome Measure(s): Biomechanical variables (linear acceleration, rotational acceleration, jerk, force, impulse, and impact duration) related to head impacts were categorized by session type, player position, and helmet impact location.

Results: Differences in grouping variables were found for each impact descriptor. Impacts occurred more frequently and with greater intensity during games. Linear acceleration was greatest in defensive linemen and offensive skill players and when the impact occurred at the top of the helmet. The largest rotational acceleration occurred in defensive linemen and with impacts to the front of the helmet. Impacts with the highest-magnitude jerk, force, and impulse and shortest duration occurred in the offensive skill, defensive line, offensive line, and defensive skill players, respectively. Top-of-the-helmet impacts yielded the greatest magnitude for the same variables.

Conclusions: We are the first to provide a biomechanical characterization of head impacts in an interscholastic football team across a season of play. The intensity of game play manifested with more frequent and intense impacts. The highest-magnitude variables were distributed across all player groups, but impacts to the top of the helmet yielded the highest values. These high school football athletes appeared to sustain greater accelerations after impact than their older counterparts did. How this finding relates to concussion occurrence has yet to be elucidated.

Key Words: concussions, mild traumatic brain injuries, Head Impact Telemetry System, acceleration

Key Points

- The mean linear acceleration resulting from impacts recorded during both high school games and practices exceeded that reported at the collegiate level.
- Impacts to the top of the head yielded the greatest linear acceleration and impact force magnitude. Coaches must teach proper tackling techniques in order to reduce the risk of concussions and serious cervical spine injuries.
- High school offensive and defensive line players sustained the lowest-magnitude impacts but the highest number of impacts during games and practices.

Participation in sporting activities has been estimated to result in 1.6 million to 3.8 million brain injuries annually.[1] Injuries occurring during collegiate and professional football often receive the greatest attention, with an incidence rate of 4.8% to 6.3% of collegiate athletes[2,3] and 7.7% of National Football League (NFL) athletes.[4] On an annual basis, therefore, concussions occur to an estimated 3264 to 4284 of the 68 000 collegiate players and 130 of the 1700 NFL athletes. Interscholastic athletes (ie, high school), however, represent the single largest cohort of football players and account for the majority of sport-related concussions. In a given year, 3.6% to 5.6% of the 1.2 million interscholastic football athletes sustain concussions,[3,5] corresponding to an estimated 43 200 to 67 200 injuries. The true injury incidence is likely much higher, because 53% of concussed high school athletes are suspected of not reporting their injuries to medical personnel.[6] Despite the fact that the greatest number of concussions occur in high school football players,[7] our present understanding of the injury has focused on collegiate and professional athletes.

Concussion assessment has improved over the years, and evaluative measures of concussion-related symptoms, neurocognitive functioning, and postural control have been proposed as assessment devices.[8] The strength of these tests lies in providing objective information for withholding an injured athlete and in making return-to-play decisions. For example, the Standardized Assessment of Concussion[9] and the Balance Error Scoring System[10] have been used effectively as sideline assessment tools that are administered once an athlete with a suspected injury has been identified. Recognizing athletes with a suspected injury, however, remains problematic.[6] The Head Impact Telemetry System (HITS) (Simbex LLC, Lebanon, NH) is a novel wireless monitoring system with the potential to rapidly identify athletes who have sustained an impact to the helmet capable of producing an injury. The HITS provides real-time, postimpact data[11] to the clinician positioned on the sideline, but the novelty of the system and a dearth of information related to impact biomechanics presently limit utility of the HITS as a diagnostic tool.

Duma et al[12] were the first to report on the HITS when 38 athletes (up to 8 at a time) were equipped with the system for 10 games and 35 practices. The distribution of impact magnitudes was skewed positively, and the mean resultant linear acceleration was estimated at $32 \pm 25g$ ($1g = 9.8$ m/s^2). The investigators later expanded their sample and reported similar findings when up to 18 athletes (52 athletes total, according to the authors) were equipped with the HITS during the 2003 and 2004 football seasons.[13] During the 67 practices and 22 games, a total of 11 604 impacts were recorded. Across all players fitted with the system, an average of 135 impacts occurred during games and 129 impacts during practices, with a mean resultant linear acceleration of $20.9 \pm 18.7g$. With identical methods in a 2-year investigation at a different institution, the mean linear acceleration in collegiate athletes was $22.25 \pm 1.79g$ across all athletes and sessions. Further analyses showed different impact magnitudes for session type, player position, and helmet location.[14] Schnebel et al[15] were the first to evaluate impacts occurring during interscholastic football but failed to report the overall mean accelerations for the cohort.

The HITS technology has increased our biomechanical understanding of impacts sustained during collegiate-level football participation, but using both common and novel variables associated with blows to the head to characterize impacts sustained at the interscholastic level is a new approach. For instance, in addition to the often-reported linear and rotational accelerations that result from a head impact, other investigators of brain trauma biomechanics have evaluated variables such as force, velocity, impulse, and impact duration.[16] Many of these variables have yet to be applied to a sport context or examined in relation to more commonly reported values.

Although the incidence rate of concussion appears to be similar across levels of play, the absolute number of injuries occurring during high school football, where the disparity in medical coverage is the greatest, drives the need for a better understanding of impact biomechanics specific to the younger athlete. In particular, it is unclear whether the frequency and magnitude of impacts mimic those of the older and more mature players. Therefore, the purposes of our ongoing investigation were to characterize impacts to the head sustained by high school football athletes, using a variety of biomechanical variables, and to compare these outcomes across session type, playing position, and location of head impact.

METHODS

All varsity-level interscholastic football athletes (n = 35) from a single central Illinois high school (Class 3A) were recruited to participate in this study. Before data collection began, all athletes and their parents completed institutional review board–approved written informed assents and consents, respectively. The institutional review board also approved all study procedures. Each athlete then provided demographic information such as age, height, mass, year in school, primary position, and helmet age. The athletes were fitted with either a new or a 1-year-old Riddell Revolution helmet (Riddell/All American, Elyria, OH), and a HITS encoder was placed inside. Only 32 athletes wore a HITS-equipped helmet at any given time. Three athletes initially recruited to participate were lost to orthopaedic injuries but were replaced by junior varsity athletes. Data were collected across an entire season of football participation, including all preseason practices, all regular-season games and practices, and all postseason games and practices.

The HITS incorporates 2 components: an encoder unit located in the football helmet and a sideline computer. The encoder consists of 6 single-axis accelerometers, a telemetry unit, a data storage device, and an onboard battery pack, all encased in waterproof plastic and retrofitted within the padding of the football helmet (Figure 1). Helmets equipped with the HITS encoder look and function identically to other helmets and continue to meet National Operating Committee on Standards for Athletic Equipment (NOCSAE) standards for safety. Impact data are recorded at 1000 Hz and transmitted to the sideline computer for clinical use and data storage for later analysis. If the encoder-equipped helmet goes out of range (more than 137 m [150 yd]) of the computer, the onboard data storage unit will record up to 100 impacts. These impacts then download to the computer once the helmet is again within range. The sideline computer is connected to a radio receiver, which is housed within an environmentally resistant plastic case. Constant communication between the helmet and sideline receiver is achieved through a continually changing, Federal Communications Commis-

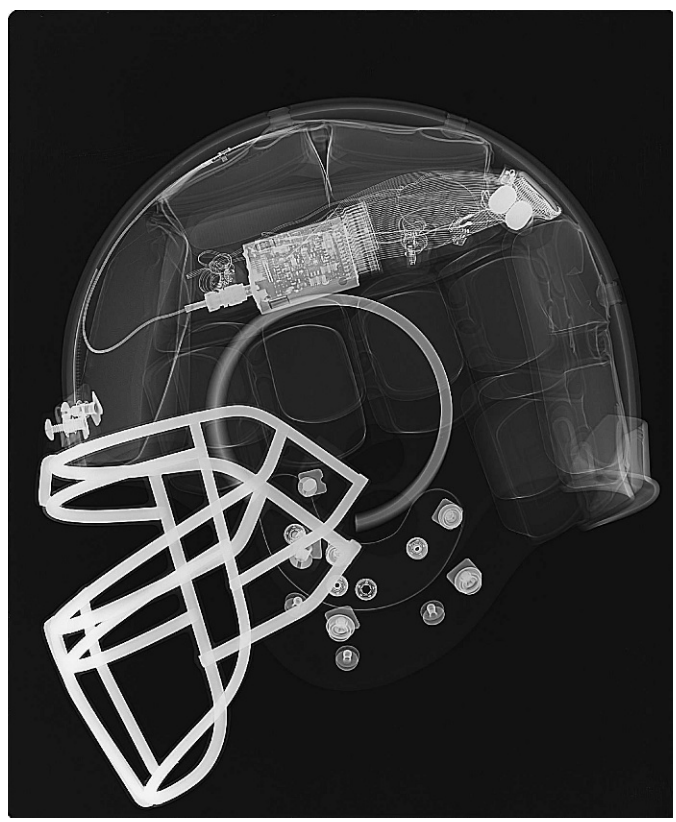

Figure 1. The Head Impact Telemetry System encoder placed within the padding of a football helmet around the crown of the head. (Photograph courtesy of Simbex LLC, Lebanon, NH.)

sion–approved radio frequency range (903 to 927 MHz). Data received by the sideline computer are processed through a novel algorithm to determine the location and magnitude of each impact to the head.[11] Accuracy of the HITS was established by comparing impact outcomes from HITS-equipped football helmets with similarly protected Hybrid III crash dummies equipped with a 3-2-2-2 accelerometer array. The correlation between the HITS and Hybrid III dummies was high (r = 0.98), with a 4% error rate when estimating both linear and angular accelerations.[12] Similarly, the location of impact was nearly identical under the NOCSAE testing protocol, where the HITS accurately identified the location of impact within ±0.41 cm.[12]

Data related to a head impact were recorded when a single accelerometer exceeded the preset 15*g* threshold. Following this event, the 8 milliseconds before impact and the 32 milliseconds after impact were transmitted and stored. Because the data storage is triggered by a single accelerometer, the resultant linear acceleration might fall below 15*g*. To reduce the inclusion of errant impacts in the data set, an investigator reviewed all impacts on a daily basis. Athletes with high-magnitude impacts were questioned about the nature of the impact. For example, if the helmet was dropped, thrown, or swung against another helmet, the impact was marked and later removed from the data set and all analyses. The data available from the computer contained all pertinent impact data, including peak linear acceleration, rotational acceleration (derived from the x-axis and y-axis angular accelerations), impact location, and date and time stamp.

Data Analysis

Data were grouped based on the session type (practice and game), impact location, and athlete position. Impacts were categorized into front, back, side, and top. Those impacts occurring above 60° of elevation over the head's center of mass were categorized as *top*. Those below 60° of elevation were divided into quadrants based on the azimuth location of impact (front, back, or side; Figure 2). Player positions were grouped into offensive line (center, guard, or offensive tackle; n = 8), offensive skill (quarterback, receiver, tight end, running back, or full back; n = 15), defensive line (defensive end, nose tackle, or defensive tackle; n = 7), and defensive skill (linebacker, corner, or safety; n = 4). As is common to many high school teams, some athletes played both offensive and defensive positions (n = 8, 23%). For the purposes of this investigation, players self-identified their primary position, which was used for analyses.

Automated features within the HITS software provide the resultant linear and angular acceleration and location of each impact. Additionally, a custom MATLAB (version 6.5.1; The MathWorks, Inc, Natick, MA) program was written to evaluate each impact and calculate the following variables: body mass index (BMI), estimated head mass,[17] number of impacts per session per athlete, estimated peak head jerk (m/s^3), estimated impact force (N), impact impulse (N/s), and impact duration (ms). A representative impact curve is shown in Figure 3. Impulse is the amount of force applied over a period of time, with greater impulse resulting in a greater change in momentum of the struck object. Larger changes in momentum are speculated to present a greater injury risk.[18] Impulse was calculated by summing the area under the acceleration curve using the trapezoidal estimation method and multiplying that value by impact duration. Impact duration was determined as the difference in time from point A to point C. Jerk is a measure of velocity, defined as the change in acceleration divided by the change in time: m·s^{-2}·s^{-1}. Not all of the impact measurements could be processed with our custom program; fewer than 3% (n = 580) were excluded from the analyses.

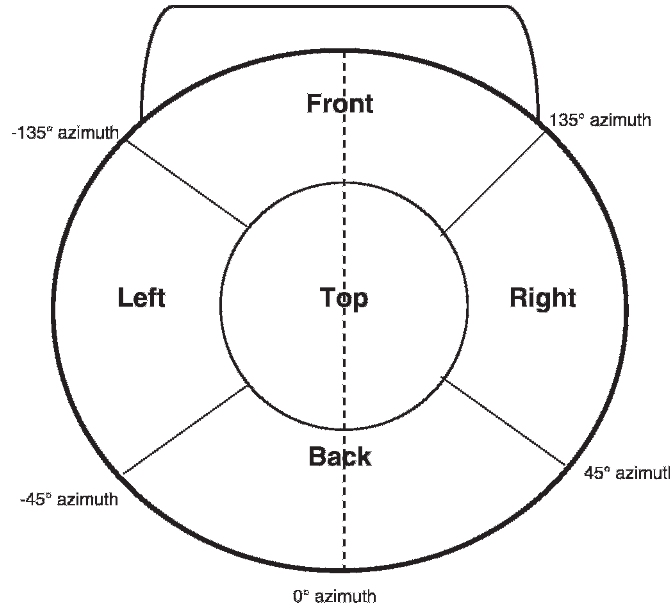

Figure 2. Distribution of impact locations across the helmet. Impacts to the right and left sides were grouped together.

Before analysis, we visually inspected the dependent variables (linear acceleration, rotational acceleration, jerk, force, impulse, and duration), revealing a positively skewed distribution. To control for the violation of normality in the subsequent analyses, these data were converted with a natural log function, and the statistical analyses were conducted on the converted data. All data presented below are in the original metric. Further inspection of the data indicated a violation to the homogeneity of variance for all dependent variables when grouped by player position and helmet impact location ($P < .01$). No homogeneity violations were noted when the data were grouped by session type ($P > .01$).

We first performed a χ^2 goodness-of-fit assessment for impact frequency based on session type, player position, and helmet location. This technique adjusts the number of expected events by accounting for the numbers of game and practice sessions and athletes per position group. A second set of analyses included assessments of the dependent variables by session type, player position, and helmet location. Analyses of variance were used to evaluate differences in the demographic variables and dependent variables based on session type. When main effects were found, post hoc comparisons were corrected using the Bonferroni method. To control for the homogeneity of variance violation when the dependent variables were grouped by player position and helmet location, we implemented the Brown-Forsythe technique and the

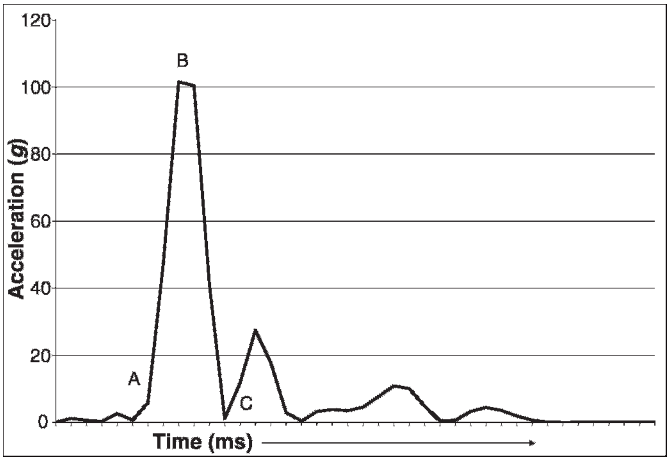

Figure 3. Representative impact curve. A number of variables were calculated from the impact data and participant demographics. Jerk = (B_a−A_a)/Δ_{B-A} time. Force = B_a * head mass. Impulse = [Area under curve from A_a to C_a] * time. Impact duration = time from A_t to C_t. *a* indicates acceleration values; *t*, time values.

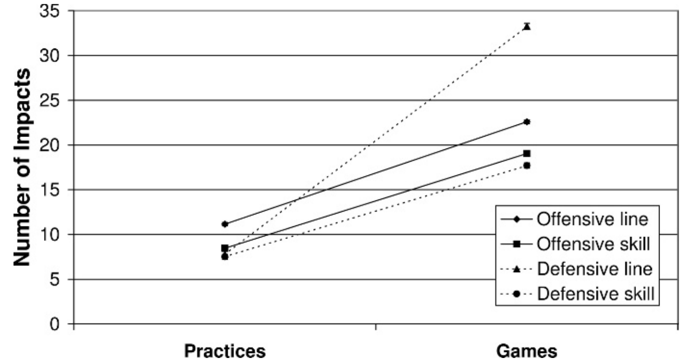

Figure 4. Mean number of impacts sustained during practices and games by player positions.

Games-Howell post hoc analysis. Finally, Spearman correlations were calculated to better understand the relationship between the commonly used variables of linear and rotational acceleration and other measures of the impacts. All analyses were completed with SPSS (version 15.0; SPSS Inc, Chicago, IL) with statistical significance set at $P < .01$.

RESULTS

During the 2007 football season, we collected data across 68 sessions, which included 55 practice days and 13 games. A total of 19 224 impacts were included in the analyses. Some impacts (n = 82) were excluded from the data set for events not directly related to sport participation, such as dropping or throwing a helmet. No differences were noted among the players' ages, heights, and helmet ages ($P > .01$; Table 1). However, differences were seen for player mass ($F_{3,30}$ = 19.757, $P < .01$), BMI ($F_{3,30}$ = 13.20, $P < .01$), and head mass ($F_{3,30}$ = 16.44, $P < .01$). For these variables, the offensive line players were larger than the offensive skill ($P < .01$) and defensive skill ($P < .01$) players, and the offensive skill players were smaller than the defensive linemen ($P < .01$).

The mean number of impacts incurred by all players during all sessions was 15.87 (SD = 17.87). The frequencies of impacts by player position for both session types are presented in Figure 4. More impacts occurred during games (mean = 24.54 ± 22.41) than during practice sessions (mean = 9.16 ± 8.64) for all athletes (χ^2_1 = 7452.88, $P < .01$). Further analyses indicated differences (χ^2_3 = 1405.46, $P < .01$) among the number of impacts by player position for all sessions. The defensive line players sustained the greatest number of impacts per session, with impacts occurring 1.04, 1.69, and 1.98 times more frequently than those experienced by the offensive linemen, offensive skill players, and defensive skill players, respectively. Finally, differences in impact frequency by helmet location were present (χ^2_3 = 4815.02, $P < .01$), with impacts to the front of the helmet 1.79, 2.88, and 3.32 times more frequent than those to the back, side, or top of the helmet, respectively.

Linear head accelerations after impact by player position and helmet location are presented in Figure 5. Differences by session type were noted ($F_{1,19\,222}$ = 50.23, $P < .01$), with game impacts (mean = 24.76 ± 15.72 *g*) resulting in higher linear accelerations than practice impacts (mean = 23.26 ± 14.48*g*). Main effects for resultant linear accelerations were seen for player position ($F_{3,10\,149.78}$ = 9.53, $P < .01$) and impact location ($F_{3,12\,039.58}$ = 243.17, $P < .01$). Post hoc analyses revealed that the 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 ($P < .01$). For impact location, hits to the top of the helmet produced linear accelerations that were greater than those at all other locations ($P < .01$). In descending order, the magnitude of resultant linear accelerations was top, front, back, and side.

Rotational head accelerations after impact by player position and helmet location are presented in Figure 6. Analyses of the resultant rotational acceleration also showed a difference by session type ($F_{1,19\,220}$ = 89.39, $P < .01$), with game impacts (mean = 1669.79 ± 1249.41 rad/s^2) generating more rotational acceleration than practice impacts (mean = 1468.58 ± 1055.00 rad/s^2). Additional analyses showed main effects for player position ($F_{3,10\,159.69}$ = 23.00, $P < .01$) and impact location ($F_{3,11\,893.57}$ = 421.43, $P < .01$). Follow-up testing revealed that the offensive and defensive line players sustained similar magnitudes of rotational accelerations,

Table 1. Demographic Information for the High School Football Athletes (Mean ± SD)

Players	Age, y	Height, cm	Mass, kg	Body Mass Index	Head Mass, kg[a]
Offensive line (n = 8)	16.97 ± 0.54	184.85 ± 4.74	100.12 ± 10.14	29.37 ± 3.40	5.65 ± 0.18
Offensive skill (n = 15)	16.96 ± 0.95	181.17 ± 5.52	76.81 ± 6.26	23.44 ± 2.14	5.20 ± 0.15
Defensive line (n = 7)	16.78 ± 0.48	185.16 ± 3.77	95.91 ± 7.17	27.97 ± 1.90	5.58 ± 0.15
Defensive skill (n = 4)	16.09 ± 0.80	182.78 ± 6.90	81.64 ± 4.50	24.47 ± 1.34	5.31 ± 0.16
All athletes (n = 34)	16.85 ± 0.75	183.49 ± 5.31	89.42 ± 12.88	26.54 ± 3.59	5.45 ± 0.26

[a] Head mass was estimated from the participants' heights and weights, based on recommendations by Zatsiorsky.[17]

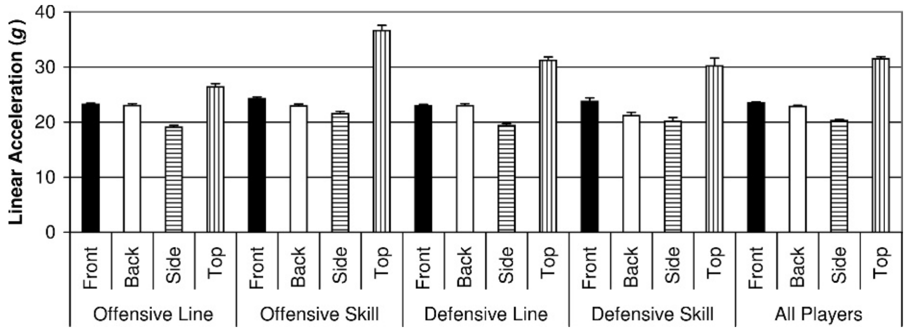

Figure 5. Resultant linear acceleration by player position and impact location.

which were greater than those of the offensive and defensive skill players ($P < .01$). The post hoc impact location analyses demonstrated that impacts to the front of the helmet were greater than to all other helmet areas ($P < .01$). In descending order, the greatest rotational acceleration resulted from impacts to the front, back, side, and top.

The final analyses included evaluations of the maximum head jerk, impact force, impact impulse, and duration of impact variables (Table 2). Compared with practices, larger magnitudes of all variables ($P < .01$) resulted during game play. We found a main effect for all variables by player position ($P < .01$), with follow-up analysis indicating that the offensive line players had less head jerk than the offensive skill and defensive line players ($P < .01$) but did not differ from the defensive skill players ($P > .01$). Forces of impact did not differ between the offensive and defensive skill players ($P > .01$), but they were both less than those of the offensive and defensive linemen ($P < .01$). The evaluation of impact impulse revealed equivalent values for the offensive and defensive lines, which were both greater than those for the offensive and defensive skill players ($P < .01$). The duration of impact was longer in offensive line players than in all other athletes, followed by the defensive line, who had longer-duration impacts than the offensive and defensive skill players ($P < .01$). The offensive and defensive skill players did not differ for duration of impact ($P > .01$).

On evaluating differences between the calculated variables and impact location, we noted main effects for all variables ($P < .01$). The post hoc assessment for maximum head jerk indicated that impacts to the top of the helmet generated more jerk than all other impact locations ($P > .01$). Impacts to the side of the helmet produced less jerk than all other locations ($P < .01$). No difference was seen between impacts to the front or back of the helmet ($P > .01$). Impact force was greater at the top of the helmet than at all other locations ($P < .01$), with differences occurring at all locations. In descending order, the greatest force was generated when blows occurred to the top, front, back, and side. Impact impulse was also greatest at the top of the helmet ($P < .01$), with differences occurring at all helmet locations. The greatest impulse values occurred in the same order: top, front, back, and side. Compared with all other locations, the duration was shortest ($P < .01$) for impacts to the top of the helmet and longest ($P < .01$) for impacts to the side of the helmet. Impacts to the front and back of the helmet did not differ in duration ($P > .01$).

Last, the correlation analyses indicated strong relationships between the resultant linear acceleration and the maximum jerk ($r_{s(18\,649)} = 0.90$, $P < .01$), maximum force ($r_{s(18\,649)} = 0.98$, $P < .01$), and impulse ($r_{s(18\,649)} = 0.85$, P $< .01$). Significant but weaker correlations were noted between rotational acceleration and jerk ($r_{s(18\,649)} = 0.55$, $P < .01$), force ($r_{s(18\,649)} = 0.68$, $P < .01$), and impulse ($r_{s(18\,649)} = 0.67$, $P < .01$). Significant weak relationships were seen between impact duration and linear ($r_{s(18\,649)} = -0.12$, $P < .01$) and rotational acceleration ($r_{s(18\,649)} = 0.10$, $P < .01$).

DISCUSSION

Our goal was to describe head impacts in interscholastic football athletes across a season of play. In addition to the previously reported assessments of resultant linear and

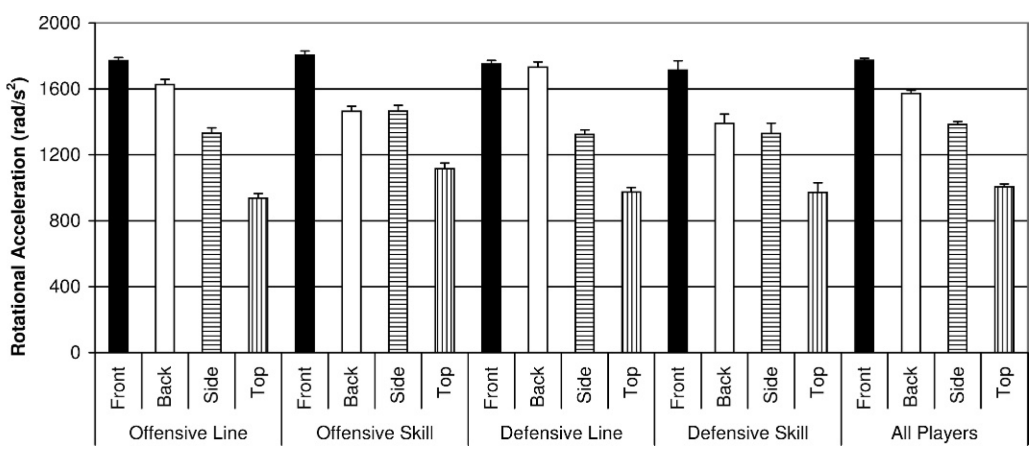

Figure 6. Resultant rotational acceleration by player position and impact location.

Table 2. Calculated Variables by Session Type, Player Position, and Impact Location (Mean ± SD)

Variable		Maximum Head Jerk, m/s³	Impact Force, N	Impulse, N·s	Impact Duration, ms
Session type[a]					
	Practices	10.04 ± 9.87	1281.67 ± 966.74	4.73 ± 3.28	9.06 ± 3.02
	Games	10.81 ± 9.97	1357.72 ± 1090.63	4.91 ± 3.90	8.90 ± 3.00
Player position					
	Offensive line	9.73 ± 10.40	1314.14 ± 931.68	4.94 ± 3.11	9.19 ± 2.90
	Offensive skill	11.01 ± 11.52	1318.22 ± 1257.57	4.76 ± 4.43	8.86 ± 3.16
	Defensive line	10.31 ± 7.24	1332.08 ± 783.49	4.81 ± 2.71	8.98 ± 2.88
	Defensive skill	10.25 ± 8.29	1242.98 ± 939.21	4.50 ± 3.56	8.83 ± 3.17
Helmet impact location					
	Front	9.85 ± 8.16	1285.87 ± 932.53	4.79 ± 3.43	9.04 ± 2.77
	Back	10.01 ± 11.38	1262.57 ± 100.82	4.64 ± 3.25	9.01 ± 3.03
	Side	8.14 ± 6.74	1093.96 ± 756.57	4.42 ± 3.15	9.84 ± 3.52
	Top	15.42 ± 13.26	1767.35 ± 1418.18	5.65 ± 4.74	7.78 ± 2.71
Overall					
	All sessions, players, and locations	10.38 ± 9.92	1314.88 ± 1023.36	4.81 ± 3.57	8.99 ± 3.01

[a] For all variables, magnitudes during games were greater than during practices ($P < .01$).

rotational acceleration, we also sought to calculate and provide a statistical analysis of variables thought to better describe these impacts. The most notable finding from this investigation is that the mean linear acceleration resulting from impacts recorded during games (24.76*g*) and practices (23.26*g*) exceeded that reported at the collegiate level. Using identical methods, Mihalik et al[14] reported that the average linear acceleration of a collegiate football team was 22.25*g* across all session types, and Brolinson et al[13] found a mean linear acceleration of 20.9*g* in a similar collegiate sample. Furthermore, Mihalik et al[14] removed impacts of less than 10*g* from the data set, deeming them inconsequential. Employing the same technique here would further increase the mean linear acceleration to 24.98*g* across all sessions. Whether the small differences in linear acceleration are clinically meaningful is not entirely clear. That is, we are unable to discern if these differences may influence the risk for concussive injury, but we speculate that the increased magnitude would likely escalate the potential for injury in the high school athlete. This potential is of particular importance considering that medical coverage is available at fewer than 50% of high schools,[19] where the risk for catastrophic injury appears to be the highest.[20]

The distribution and magnitude of impacts across helmet locations also differed by the level of play. In collegiate football athletes,[14] blows to the front of the helmet were 10% less frequent and resulted in 3*g* less force than in our interscholastic players. Conversely, impacts to the top and back of the helmet in the collegiate athletes occurred more frequently but resulted in slightly less linear acceleration (1 to 2*g*). Most concerning in the high school athletes were the impacts to the top of the head, which yielded the greatest linear acceleration (Figure 5) and impact force magnitude (Table 2). The increased impact intensity to this area of the helmet likely elevates the risk of concussion but also increases the propensity for more severe cervical injury.[21] This finding highlights the need for coaching proper tackling techniques, such that the athlete keeps his head up and avoids contact with the top of the helmet. An explanation of the differences between collegiate and interscholastic football athletes is beyond the scope of this investigation. However, we hypothesize that physical maturation and the associated neck strength and endurance discrepancy between the levels of play may be a factor. On average, collegiate athletes are reported to weigh 15 kg more than our athletes but stand only 3 cm taller.[22] Thus, collegiate athletes may have a more developed musculature system that is better able to control head motion after impact.

Using a variety of methods, prior investigators have attempted to identify the resultant linear acceleration necessary to cause a concussion. Initial attempts to quantify head impacts were completed by placing a single triaxial accelerometer within the padding of football helmets. Reid et al[23] reported impacts ranged from 40 to 230*g* in a collegiate football player, with a single concussion resulting from a 188*g* linear acceleration. Naunheim et al[24] reported the mean peak linear acceleration of interscholastic football impacts as 29.2*g* but no concussions occurred. These studies provided initial assessments of football impacts but were limited by their small sample sizes; also, the reported linear accelerations likely represent motion of the helmet and not the athlete's head. More recently, in a sample of concussive impacts in 13 collegiate football athletes, the HITS quantified mean linear acceleration at 102.77*g*.[25] In perhaps the most extensive attempt to establish a threshold for concussion, researchers[26] acquired video footage of 31 concussive blows that occurred during professional football games. After the impacts were reconstructed in a laboratory setting with Hybrid III models, the mean linear acceleration was noted to be 98*g*. The authors proposed that a linear acceleration of 70 to 75*g* was necessary to induce concussion.[26] Using these criteria, our data included 271 impacts (1.4%) that exceeded the 70*g* level and 78 impacts (0.4%) that exceeded the 98*g* level, slightly fewer impacts than previously were reported at the high school level.[15] However, only 5 concussive injuries were diagnosed (although not included in our analyses), suggesting that the magnitude of impact reported by the NFL Mild Traumatic Brain Injury Committee may not apply to the high school athlete or that other variables need to be considered when establishing a threshold for concussion (or both).[27]

Despite the differences in the mean linear accelerations occurring at different levels of the sport, the differences in impact magnitude among the high school athletes may be linked to the probability of concussive injury. The incidence of concussion is not equal across all player positions: The highest injury rates in high school football occur in quarterbacks (1.3 per 100 team-game positions) and running backs (0.74 per 100 team-game positions),[5] and wide receivers sustain more severe injuries.[3] Our classification method would designate these athletes as offensive skill players, who, we found, sustained the greatest magnitude of linear acceleration after an impact (Figure 5). The higher magnitudes of impacts may result from the full-speed, open-field impacts that commonly take place at these player positions. The highest injury rates at the collegiate level occurred to the offensive line (0.95/1000 athlete-exposures) and defensive skill players (0.93/1000 athlete-exposures), but these findings do not correspond with our data.[2] Our data suggest that high school offensive and defensive line athletes in these positions sustained the lowest-magnitude impacts but also experienced the greatest number of impacts during games and practices (Figure 4). Impact frequencies of our data by position group show that the 15 offensive and defensive line players (44% of all players) sustained 11 035 impacts (57% of all hits), whereas the 19 skill players (56% of all players) sustained 8189 impacts (43% of all hits). The higher impact frequency endured by the linemen likely results from their involvement in every play, whereas the magnitude of these impacts is likely lowest due to the short distance between the offensive and defensive line players and the subsequent low impact speed. Furthermore, given the nature of high school football, in which the number of players is limited, many athletes in our sample played both offensive and defensive positions, possibly increasing impact exposure and injury risk.

In assessing the calculated variables, the offensive and defensive line players were estimated to have the largest head masses. The role that head mass has in the concussion injury mechanism is not thoroughly understood, but finite element analysis models have been used to provide indirect evidence supporting the relationship between increased head mass and greater intracranial pressure after impact.[28] Increased pressure is thought to be related to the degree of structural damage in the cerebral tissue after impact.[29] Conversely, smaller head mass is associated with higher linear and rotational acceleration values,[30] which may make the athlete more susceptible to injury. Notably, the offensive skill players enrolled in this investigation had the smallest head masses and the largest linear accelerations and head jerks after impact. The change in head velocity of these athletes was 11.01 m/s, exceeding the 7.2 m/s in concussed professional athletes reported by the NFL Mild Traumatic Brain Injury Committee. Furthermore, the duration of impact experienced by the concussed NFL athletes was 15 milliseconds,[26] almost twice as long as the average we recorded. Finally, the use of impulse as a variable to quantify impacts has been reported previously in 100 concussed rugby athletes.[31] Using 2-dimensional analyses of the rugby impacts, the mean impulse of the impacts was estimated at 29 N·s. A direct comparison with our data cannot be made, because we did not include any concussive impacts, but our nonconcussive impacts fell well below this value. A disparity between the NFL and rugby impact reconstructions and the data collected here is evident. The reasons for these differences, however, are not completely apparent. Estimating head mass from only anthropometric variables may not necessarily represent the player's effective mass at the time of impact. The effective mass at impact is equivalent to the combined mass of the head and body linked through the tensed neck musculature and acting as a single unit. The additional mass results in a lower resultant acceleration, although any neck motion at impact would lessen the influence of the body mass. Additional differences in findings include our method, which permitted the tracking of all impacts (concussive and nonconcussive) during all sessions (practices and games). The method employed by the NFL[26] and rugby[31] investigations was limited by including only concussive impacts captured on video that were later reconstructed or evaluated.

Further investigations of head impacts resulting in concussion are needed to establish the relevance of the calculated variables to injury biomechanics. The strong relationships between linear acceleration and jerk, force, and impulse indicate the potential for redundancy when examining impact biomechanics. The high correlations likely result from the calculated variables being founded in the recorded linear acceleration. The relationship between these variables and rotational acceleration is not as strong, indicating that rotational acceleration provides novel information when describing and quantifying an impact. Similarly, the weak relationships between impact duration and linear and rotational acceleration also suggest the contribution of unique information.

Despite our unique findings, certain limitations are present. Although we collected a myriad of impacts, the large number of data points also may have influenced some of our significant findings. Further, when comparing these results with those at the collegiate level, one should consider the different distributions of player positions and the time frame at which those data were collected. Finally, future investigators should attempt to quantify the playing time and skill level of the athletes. These variables will likely play a role in the number of impacts and, potentially, the severity of impacts incurred.

This project is the first report in an ongoing analysis of head impacts and concussions incurred by interscholastic football players. Our intent is to disseminate descriptive biomechanical information related to impacts sustained during a season of interscholastic football. The findings from this investigation demonstrate clear differences between impacts occurring on the interscholastic football field and those occurring at the collegiate and professional levels. Through ongoing data collection and analyses, we believe that a better understanding of football-related impacts can be provided for the benefit of sports medicine professionals and so that equipment manufacturers can take the appropriate steps to improve safety equipment. Evaluating impacts to the head in sports other than football will prove more challenging, but the HITS has been adapted for soccer, boxing, and ice hockey athletes. Our understanding of impact forces and biomechanics in sports in which headgear is not commonly worn during practices and competitions (eg, soccer, basketball) will, unfortunately, lag behind.

The ultimate goal of this research is to develop criteria that will allow the HITS or a similar system to be used as a sideline diagnostic tool for concussive injuries. However, until more investigations can be conducted to better understand all the biomechanical components associated with sport concussion, the HITS remains an instrument of research. In a limited capacity, the HITS may be used as ''an extra set of eyes'' to identify those athletes who have sustained significant blows to the head and warrant follow-up evaluation by sports medicine professionals. The clinical examination, coupled with tests of concussion-related symptoms, neurocognitive function, and postural control, remains the most sensitive assessment of this complex injury.[32]

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