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Characterization of Concussive Events in Professional American Football Using Videogrammetry

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Abstract—Sports concussions offer a unique opportunity to study head kinematics associated with mild traumatic brain injury. In this study, a model-based image matching (MBIM) approach was employed to analyze video footage of 57 concussions which occurred in National Football League (NFL) games. By utilizing at least two camera views, higher frame rate footage (> 60 images s^{-1}), and laser scans of the field and helmets involved in each case, it was possible to calculate the change in velocity of the helmet during impact in six degrees of freedom. The average impact velocity for these concussive events was 8.9 ± 2.0 m s^{-1}. The average changes in translational and rotational velocity for the concussed players' helmets were 6.6 ± 2.1 m s^{-1} and 29 ± 13 rad s^{-1}, respectively. The average change in translational velocity was higher for helmet-to-ground ($n = 16$) impacts compared to helmet-to-helmet ($n = 30$) or helmet-to-shoulder ($n = 11$) events ($p < 0.001$), while helmet-to-shoulder impacts had a smaller change in rotational velocity compared to the other impact sources ($p < 0.001$). By quantifying the impact velocities and locations associated with concussive impacts in professional American football, this study provides information that may be used to improve upon current helmet testing methodologies.

Keywords—Concussion, Helmet, Kinematics, Biomechanics.

INTRODUCTION

Concussive events in professional American football offer a unique opportunity to study the head kinematics associated with mild traumatic brain injury. While incidence of concussion has been associated with contact sports at all levels of play,[10,23,37,46] concussions sustained by professional football players in NFL games are generally well-documented with multiple views of high-quality video footage.[26] The precise biomechanical mechanisms causing these concussions are still unclear, as are the risk levels associated with varying measures of head impact severity. Therefore, additional investigation into the head kinematics associated with concussive impacts is warranted.

Researchers have previously utilized two methods to measure head motion in American football impacts: wearable sensors and videogrammetry. A primary advantage of wearable sensors is that they theoretically record a census of all head impacts. The measurement of injury and non-injury impacts without selection bias is necessary when formulating injury risk curves.[13] In addition, wearable sensor data can be collected at a higher sampling rate than permitted by video, making it possible to accurately measure accelerations. Current wearable sensor systems include, but are not limited to, those mounted in headbands,[2,34] helmets,[40] mouthguards/retainers,[5,7,28,39,48] and some have been skin-, tooth-, or ear-mounted.[8,33,42] While newer sensor systems show substantial improvements in accuracy,[39] many of the existing on-field systems have documented issues with accuracy of measured magnitudes and impact count.[44] In one study, a mouthguard sensor recorded an impact count nine times higher than the number of head impacts observed on video.[25] Poor accuracy in certain impact conditions has also been noted for the Head Impact Telemetry System (HITS), which is a commonly used helmet-mounted sensor system available only in Riddell helmets.[21,42]

Videogrammetry, the process of obtaining kinematic measurements from video images, is an alter-

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native method for studying the biomechanics of concussion. The sample rate of information gained from videogrammetry is not as fast as wearable sensor data, but it allows for post hoc analysis of events of interest. The sampling rate of video (tens to hundreds of images per second) is generally far lower than wearable sensors (hundreds to thousands of samples per second). While high-quality video can yield accurate measurements of helmet velocity,[31,35] the sampling rate is generally too low to accurately calculate accelerations. In addition, videogrammetry (especially model-based image matching) is a labor-intensive process.[4,29] The time-intensive nature of videogrammetry results in a focus on severe impacts and/or concussion cases.[35] This selection bias renders the data set useless for calculating injury risk curves, where an unbiased sampling of both injurious and non-injurious head impacts is required. However, videogrammetry can be performed retrospectively and allows for detailed analysis of events of interest (e.g., concussion-causing impacts).

Past video analysis efforts in other sports have focused primarily on cataloging the circumstances of the concussions.[1,3,12,16,20,24] Videogrammetry has been used to determine the closing speed between the helmet and a collision partner in concussion-causing impacts in Australian Rules football and rugby,[27] hockey,[32,50] rugby,[47] skiing,[49] and professional football.[36] The closing speed and impact location were used as inputs to an analytical, computational, or physical reconstruction that then determined the head kinematics.[15,18,27,32,36,50] In the case of American football, video analysis must be supplemented by some form of reconstruction in order to obtain head kinematics because the players are helmeted, which prevents the head from being directly visible on video. The accuracy of these reconstructions could be improved by matching not only the initial impact speed and location, but also the translational and rotational motion of the helmet during and after impact.

None of the studies cited above used videogrammetry to measure head or helmet motion during or after impact, nor did they measure the rotational motion of the helmet. In this study, high-quality video of 57 concussive events that occurred in NFL games was obtained after a detailed video review process during which the concussed players and collision partners were identified.[26] A model-based image matching (MBIM) methodology was applied to determine not only closing speed and impact location, but also the change in translational and rotational velocity of the helmet in six degrees of freedom. In addition, the torso alignment of each player was measured as an indicator of effective mass and the differences in concussive head kinematics based on the impact source (helmet, shoulder, or ground) were investigated. The authors hypothesize that initial conditions for helmet impacts and the resulting helmet kinematics may vary by impact source, with ground impacts being the most unique of the three impact sources studied.

MATERIALS AND METHODS

Game footage from professional football games was acquired for 367 concussion cases diagnosed by the NFL Game Day Concussion Diagnosis and Management Protocol during the 2015–2017 NFL seasons,[11] and recorded in the NFL Electronic Medical Record (EMR). The NFL EMR was collectively bargained between the NFL and the NFL Players Association to serve, in part, as a workplace injury recording system. NFL players sign authorization forms for the data that are provided to this system to be used in furtherance of certain research that is approved by the League and the Union. Injury plays for the concussions resulting from helmet, shoulder, and ground impacts were identified through a peer-reviewed video analysis process.[26] For cases in which multiple impacts were identified during the injury play, 'primary' and 'secondary' exposure designations were given to each impact, with the 'primary' exposure defined as the most severe impact.[26]

The game footage was surveyed to acquire frame rate and resolution details for each available camera view. Typical camera locations for NFL games were described in a previous study, though the number of cameras and quality of footage varied from game to game.[4] Frame rate was defined by the number of images captured per second (ips) and resolution was assessed in terms of pixels per helmet. From the 367 concussive events, 57 cases were selected for use in a videogrammetry study based on quality of the camera views with good visibility of the players' helmets throughout the impact. For use in the study, at least two camera views of the impact were required, and cases were selected by giving preference to impacts with at least one camera view with a frame rate exceeding 180 ips. An effort was also made to select cases with impact locations that were roughly representative of the overall distribution of impacts characterized by a video review study of concussive impacts in the NFL.[26] For the cases selected, the resolution of the images ranged from 64 to 126,700 pixels per helmet, while frame rates ranged from 48 to 628 ips. In all 57 cases chosen for analysis, the concussed player sustained a direct impact to the helmet. The collision partner was another helmet in 30 (53%) cases, the ground in 16 (28%) cases, and the shoulder in 11 (19%) cases. For helmet-to-helmet impacts, both the concussed player and the uninjured collision partners' helmet kinematics were quantified.

Videogrammetry was performed on the selected cases using the techniques described in detail by previous studies.[4,29,30] To summarize those techniques, game-style footage was first stabilized to remove the effects of camera motion. Next, the position and orientation of each camera was calculated using a camera-matching procedure.[4] Three-dimensional scans of the stadiums where the concussive events occurred were used to calibrate the dimensions in the virtual views generated by camera-matching (Figs. 1a and 1b). Next, three-dimensional scans of each of the helmets involved in the impacts were used as part of a MBIM technique to estimate the position and orientation of the helmets in six degrees of freedom (Figs. 1c and 1d). Position and orientation of the players' torsos were characterized by overlaying a three-dimensional, size-scaled ellipsoid shape on the players' torsos in each camera view for the frame corresponding to the time of impact (Fig. 2). The length of the ellipsoids were scaled using the players' waist-back length estimated to be 26.5% of the player's height.[9] All positions and orientations were referenced back to a global field reference frame with the origin located in the corner of one of the end zones, x aligned with the longitudinal axis of the field, and z oriented into the turf (Fig. 2).

The local coordinate system for each helmet was aligned with the head coordinate system defined by the Society of Automotive Engineers document J211a for which the x–y plane is parallel to the Frankfort plane and the origin is located at the estimated center of gravity of the head.[43] A transformation matrix from points on the exterior of the helmet to the head coordinate system was generated for each helmet model by donning each helmet model on a Hybrid III headform,

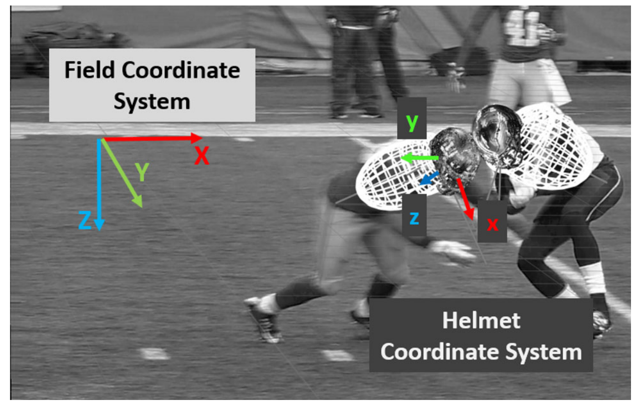

FIGURE 2. Helmet and field coordinate systems used for video reconstruction data processing. Lower-case subscripts denote a local coordinate system while the upper-case subscripts denote the global (inertial) coordinate system. Ellipsoids for quantifying torso orientation are also shown.

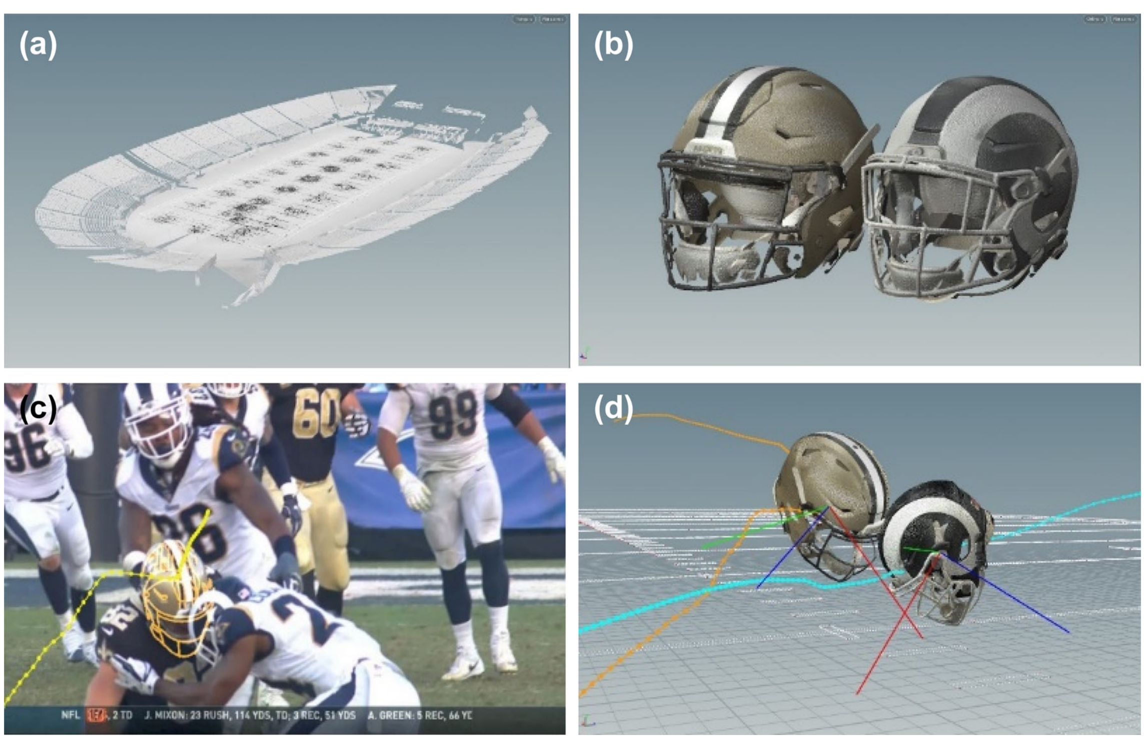

FIGURE 1. Three-dimensional scans of the stadium (a) and player helmets (b) that were used for the videogrammetry calibration process. (c) Three-dimensional helmet scan overlaid in the video frame that was used to track helmet position and orientation with respect to the field throughout the impact time history (d).

positioning the brim of the helmet 75 mm from the tip of the nose using a nose gauge, and measuring the helmet points relative to the head coordinate system using a Romer Absolute Arm-6Axis (Exact Metrology, Cincinnati, OH). These transformation matrices were used to output helmet rotations and translations in a body-centric coordinate system that could be easily related to data generated by previous studies.

Data Processing

Translational and rotational kinematics were calculated for each helmet throughout the impact. For each time step, data were exported as direction cosine matrices that related the local helmet coordinate systems to the field coordinate system. From this data, six degree-of-freedom kinematics were calculated. Position time-histories were filtered using a 30 Hz, 4-pole Butterworth filter. Velocities were calculated using the central difference method and then filtered using a 50 Hz Butterworth filter. Filter selection was based upon the findings of a previous validation study.[4] Rotational and translational velocities were calculated using Eqs. (1) and (2), respectively.

$$\begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix} = \frac{\mathrm{d}T(t)}{\mathrm{d}t}[T(t)]^{-1}, \tag{1}$$

$$(V_i)_j = \frac{(r_i)_{j+1} - (r_i)_{j-1}}{t_{j+1} - t_{j-1}}, \tag{2}$$

where ω_i are components of local helmet rotational velocity, T is the 3×3 direction cosine transformation matrix, and t is time. V_i are the components of translational velocity in the field coordinate system, r is the position relative to the field coordinate system, i is the vector component index, and j is the timestep index.

Changes in the resultant translational and rotational velocities (ΔV, $\Delta\omega$) were calculated separately. Start and end times of the impact were manually selected based on a visual inspection of the slope of the translational velocity time-histories alongside the game video footage. Next, the resultant change in velocity between each time step from 10 ms prior to the selected start time (t_S) to 50 ms after the selected end time (t_E) were calculated; this resulted in a square matrix of changes in velocities. The maximum was selected from this matrix and is reported as the change in velocity [Eq. (3)]. Start and end times for the change in velocity calculations were allowed to float in order to (1) estimate an objectively-selected time of impact (t_0) based on translational velocity and (2) to capture peak-to-peak velocity changes during the impact event, particularly for rotational velocities for which individual components tended to change polarity during the impact. In all but four cases, the change in translational velocity time range started within 5 ms of the manually selected start time. The manually selected start time for the remaining four cases was between 5 and 15 ms earlier than the matrix-selected start time. In many cases, the peak changes in translational and rotational velocity did not occur over the same time range. The reported translational and rotational velocities at time of impact (V_0 and ω_0) are reported as the resultant of those vectors at time t_0. Components of the change in rotational velocity vector ($\Delta\omega_x$, $\Delta\omega_y$, $\Delta\omega_z$) were calculated by taking the difference for each component between the start and end times associated with the maximum resultant change.

$$\Delta V = \max \begin{bmatrix} |\mathbf{V}(t_S - 10\,\mathrm{ms}) - \mathbf{V}(t_S - 10\,\mathrm{ms})| & \cdots & |\mathbf{V}(t_E + 50\,\mathrm{ms}) - \mathbf{V}(t_S - 10\,\mathrm{ms})| \\ \vdots & \ddots & \vdots \\ |\mathbf{V}(t_S - 10\,\mathrm{ms}) - \mathbf{V}(t_E + 50\,\mathrm{ms})| & \cdots & |\mathbf{V}(t_E + 50\,\mathrm{ms}) - \mathbf{V}(t_E + 50\,\mathrm{ms})| \end{bmatrix}, \tag{3}$$

where $\mathbf{V}$ is a velocity vector, t_S and t_E are the manually selected start and end times for the impact, respectively.

Closing velocities (V_c) were calculated using vector subtraction of the translational velocity of the player's helmet from the velocity of the impact source at t_0 (i.e. helmet or shoulder). For shoulder impacts, velocities from NFL Next Generation Stats radio-frequency identification sensors in the shoulder pads of the collision partner were used to calculate closing velocities between the helmet of the concussed player and shoulder of the collision partner (NFL Enterprise, LLC, New York, NY). A sample time-history is provided along with still images from the processed video with the 3D helmet model and local helmet coordinate system overlaid (Fig. 3).

Helmet impact locations were established using the three-dimensional (3D) MBIM results to determine the first point of contact between the two overlaid helmet models with respect to the local helmet coordinate systems. Heat maps for impact location density were generated by counting the number of impact points within 25° elevation and 25° azimuth of each point on the surface of a 3D helmet model and coloring that point on the helmet accordingly.

In order to quantify each players' alignment with the impact vector, a torso alignment vector was established. At the time of impact, the angle between the long axis of the torso (z_{torso}) and the closing velocity (V_c) was calculated using the vectors expressed in the field coordinate system [Eq. (4)]. Changes in translational and rotational velocity of the helmet were

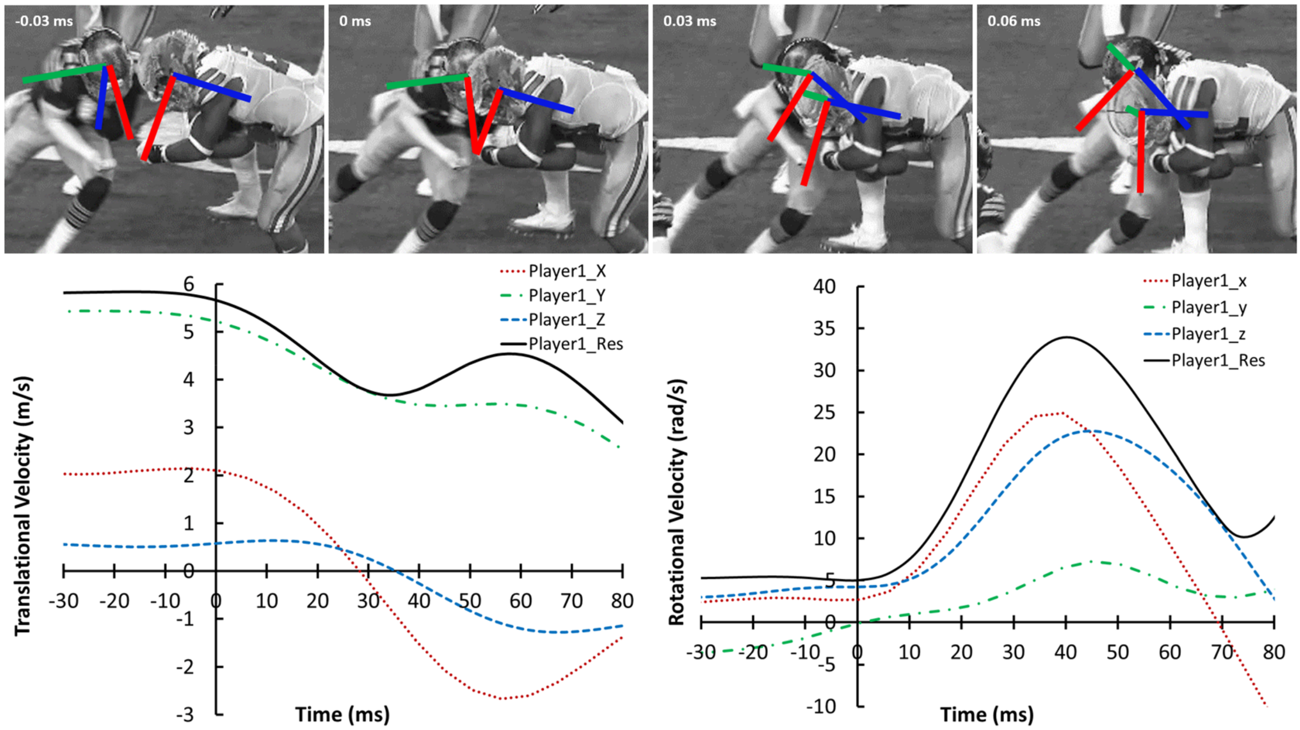

FIGURE 3. Sample translational (field coordinate system) and rotational (helmet coordinate system) velocity time-histories for Player1 for Case 06 helmet-to-helmet impact, where $t = 0$ is t_0. Snapshots from the video with overlaid 3D images of the helmets are provided, with Player1 on the left, and Player2 on the right.

also calculated with respect to the helmet coordinate system referenced to the time of impact (t_0).

$$\text{Torso Alignment} = \cos^{-1}\left(\frac{V_c \cdot z_{\text{torso}}}{|V_c||z_{\text{torso}}|}\right). \quad (4)$$

Statistical Analysis

Results from impacts with different sources were compared using unpaired Student's t tests using a 95% confidence interval. Peak kinematics for concussed and uninjured players from helmet-to-helmet impacts were compared using paired Student's t tests with a 95% confidence interval.

RESULTS

At least two unobstructed views of the concussive impact were available in all the selected cases. The quality of the video footage based on frame rate was rated as 'intermediate' in the vast majority of the selected concussion cases (52 out of 57, 91%) using criteria from a previous validation study that estimated videogrammetry error for translational and rotational velocities compared to those from a motion capture system.[4] In that study, frame rate was defined by the number of images captured per second to avoid confusion caused by interlacing of images. Three groups were defined based on video analysis performed with different frame rate footage: two videos with 60 ips ('low'); one video at 60 ips and one at a higher frame rate of 240 ips ('intermediate'); and at least two videos at 240 ips ('high').[4] The expected average absolute errors based on the videogrammetry validation study[4] for the 'intermediate' group were 9% for the translational velocity change of the helmet and 17% for the rotational velocity change of the helmet. In 4 (7%) of the selected concussion cases, the video quality was rated as 'low,' and the expected average absolute errors were 19% for the translational velocity change of the helmet and 24% for the rotational velocity change of the helmet.[4]

The average closing velocity associated with the concussive impacts in this study was 8.9 ± 2.0 m s^{-1}. The closing velocity did not vary significantly by impact source (Fig. 4). The average translational velocity change of the helmet was significantly higher ($p < 0.001$) when the concussion was caused by an impact with the ground (8.1 ± 1.7 m s^{-1}) compared to another player (5.9 ± 2.0 m s^{-1}). Ground-induced concussions were also associated with higher pre-impact rotational velocities (14 ± 12 rad s^{-1}) than concussions caused by collisions with other players

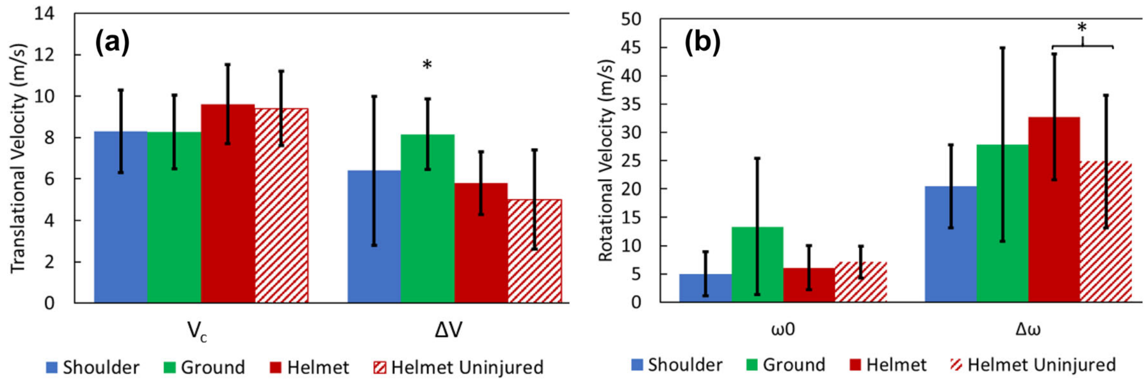

FIGURE 4. (a) Summary of initial closing and change in translational helmet velocity (V_c, ΔV); and (b) initial and change in rotational velocity (ω_0, $\Delta\omega$). *Indicates statistical significance (p < 0.05) to a 95% confidence level.

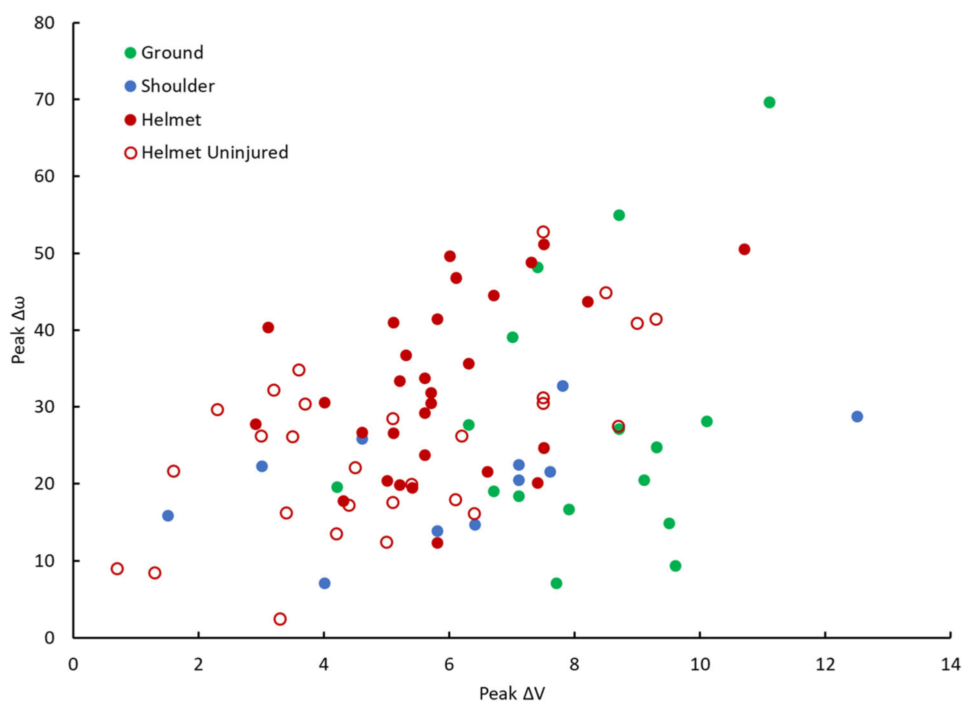

FIGURE 5. Change in translational helmet velocity vs. change in rotational helmet velocity by impact.

(5.8 ± 3.9 rad s^{-1}). The average rotational velocity change of the helmet in the concussive impacts reviewed was 29 ± 13 rad s^{-1}, with helmet impacts resulting in a significantly larger change in rotational velocity than shoulder and ground impacts for the concussed player (p = 0.015) (Appendix Tables A1, A2 and A3). In concussive helmet-to-helmet impacts, the concussed player experienced a higher rotational helmet velocity change than the uninjured player (33 ± 11 vs. 25 ± 12 rad s^{-1}, p = 0.005), but the translational helmet velocity change was similar (5.9 ± 1.5 vs. 5.0 ± 2.4 m s^{-1}, p = 0.119) (Fig. 5).

Impact locations on the concussed player's helmet clustered around different regions depending on the impact source (Figs. 6 and 7). Concussions caused by shoulder impacts were concentrated on the frontal region of the helmet. Concussive ground impacts were primarily against the rear and side of the helmet. Helmet-to-helmet impacts occurred over the front, side, and facemask regions of the helmet for the concussed player but were concentrated in the upper front and side portions of the helmet for the uninjured collision partner.

To study body alignment, the angle between the long axis of the torso and the closing velocity vector

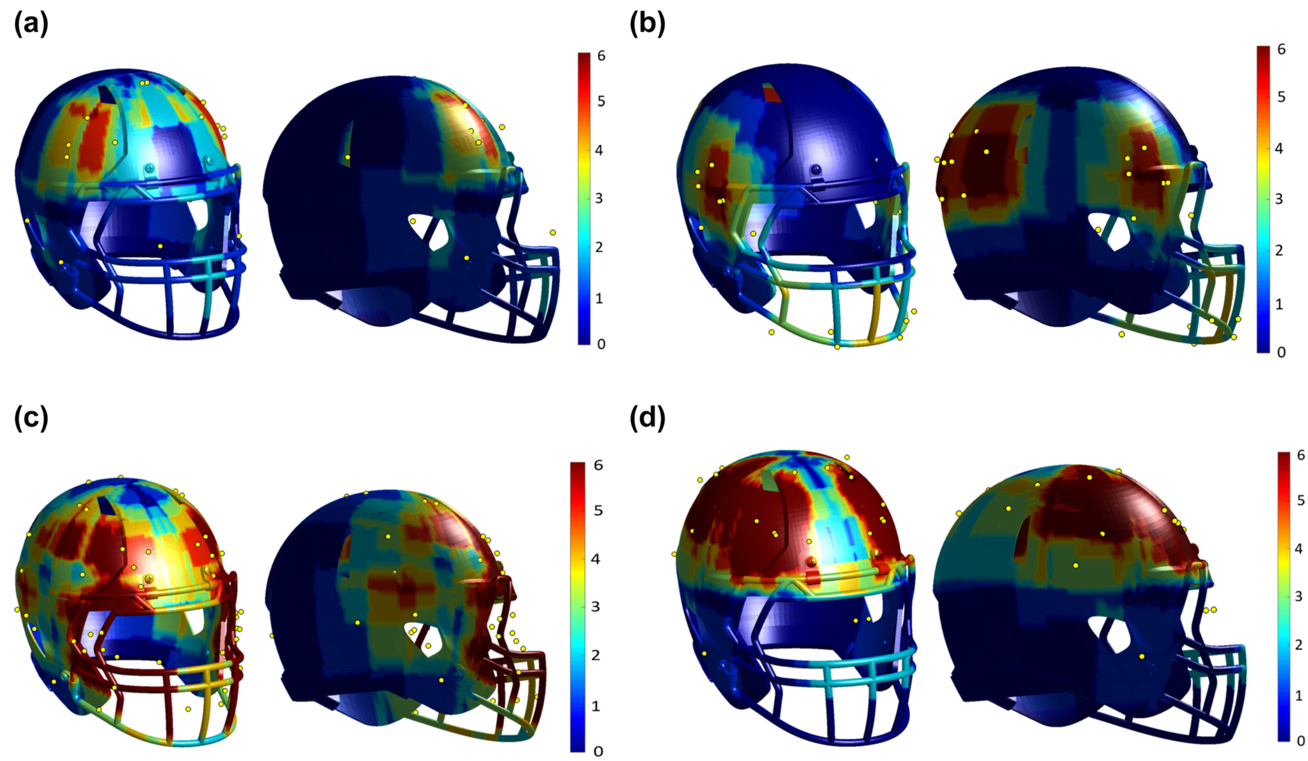

FIGURE 6. Heat maps representing density of impact locations for concussions caused by impact to the shoulder (a); ground (b); and another helmet (c); and for the non-concussed other helmet (d). The legend indicates the number of impacts within 25° of azimuth or elevation on the helmet surface. Each impact location was reflected about the *x–z* plane of the helmet to provide symmetric images.

was calculated for concussed players and collision partners in helmet-to-helmet impacts (Fig. 8). In 21 of the 28 helmet-to-helmet cases (75%), the uninjured collision partner's torso angle was greater than the torso angle of the concussed player, indicating that the uninjured collision partners were more aligned with the impact velocity vector than their concussed counterparts. Furthermore, the average torso angle for the concussed players (120° ± 23°) was significantly smaller (torso less aligned with the closing velocity vector) than the torso angle for the uninjured collision partners (136° ± 19°, $p = 0.002$). This may suggest that the uninjured player may have been more aware of or prepared for the impending impact than the concussed player in some of these cases. The angle between the long axis of the two players' torsos was 87° ± 28° on average (see Supplemental Material).

For 36 out of the 57 concussion cases, video review identified multiple helmet impacts for the concussed player during the identified play. The designation of 'primary' was given to the impact exposures identified subjectively as the most severe of these multiple impacts, while 'secondary' exposures were deemed less severe.[26] Whenever possible (25 cases), helmet kinematics were tracked through the secondary exposures, as well. Secondary exposures with either a helmet ΔV greater than 5 m s^{-1} or a helmet $\Delta\omega$ greater than 20 rad s^{-1} were observed for 21 cases. In 8 cases, the helmet velocity change was higher in the secondary exposure than the primary exposure with respect to translation (4 cases), rotation (3 cases), or both (1 case). The average absolute time between a primary concussive exposure with another player and a secondary impact with the ground was almost half a second (446 ± 144 ms, $n = 8$). When both the primary and secondary exposures both involved contact with another player, the time between impacts was much closer on average (265 ± 285 ms, $n = 10$) (Table 1).

DISCUSSION

This study used high-quality camera footage and advanced techniques in videogrammetry (i.e., model-based image matching, camera stabilization, camera matching, lens distortion correction, and three-dimensional scanning of helmets and fields) to analyze helmet motion in 57 selected concussion cases occurring in NFL games from the 2016–2017 and to 2017–2018 seasons.[4,29,30] The average closing velocity associated with the concussive impacts in this study

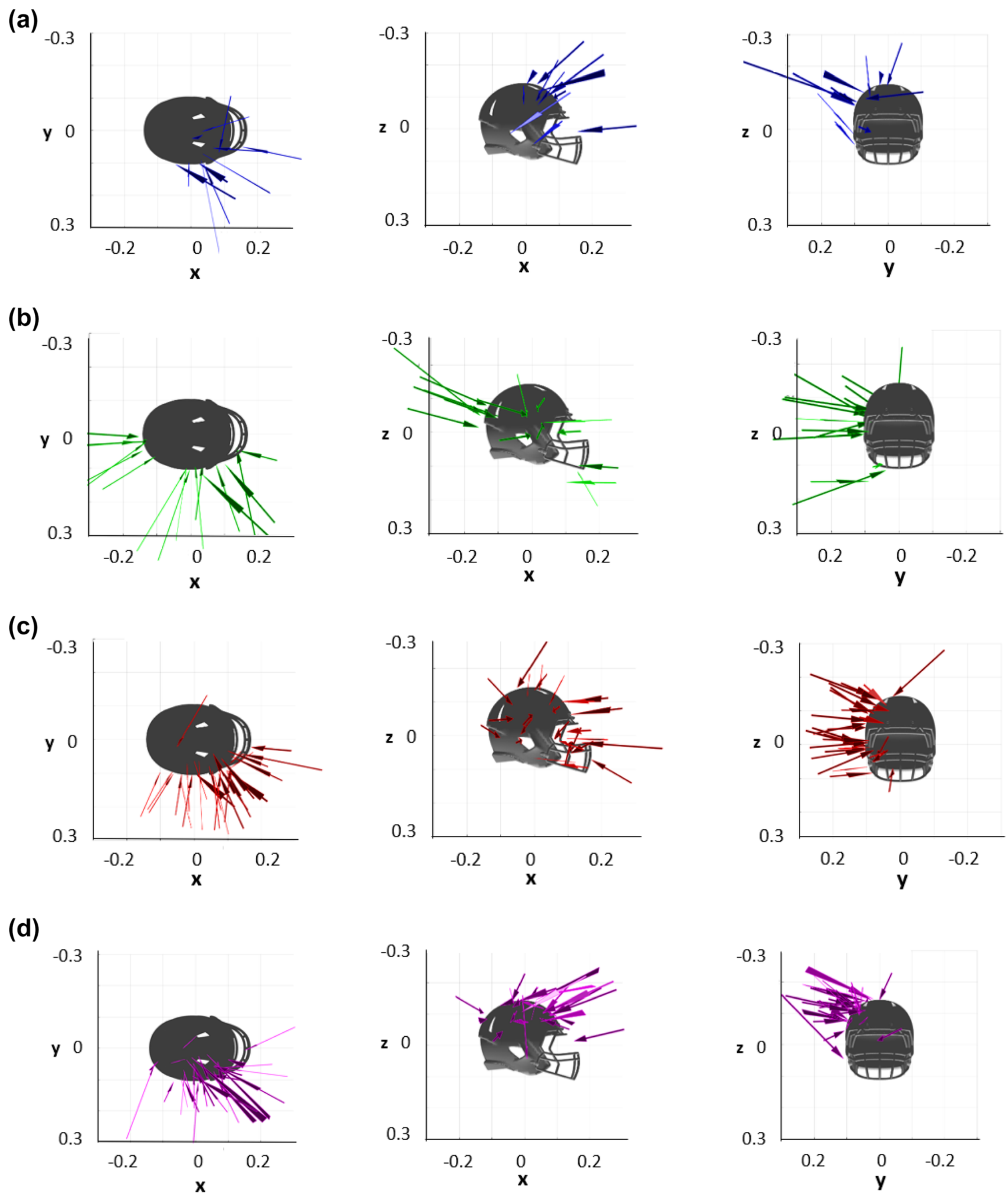

FIGURE 7. Change in velocity vectors with respect to the helmet coordinate system for players concussed in shoulder (a); and ground (b); helmet (c) impacts, and for the uninjured collision partners (d). Note that the head of each arrow is located at the impact location and the length is proportional to the magnitude of the ΔV vector. Impact locations and vectors have been reflected about the *x–z* plane for impacts on the left side of the helmet. Axis scale is in m for helmet dimensions. Velocity vector scale is 30 m s^{-1} to 1 m.

(8.9 ± 2.0 m s^{-1}) was not significantly different ($p = 0.39$) from those observed in the previous study by Pellman *et al.* (9.3 ± 1.9 m s^{-1}).[26] Trends in impact locations for helmet-to-helmet impacts were also similar in that the impact locations for the uninjured collision partners tended to be concentrated in the front and top regions of the helmet, while impact locations for the concussed players were generally lower on the helmet (40% of impact locations below the helmet *x–y* plane). Changes in translational and rotational velocity for the concussed players' helmets measured by video analysis (6.6 ± 2.1 m s^{-1} and

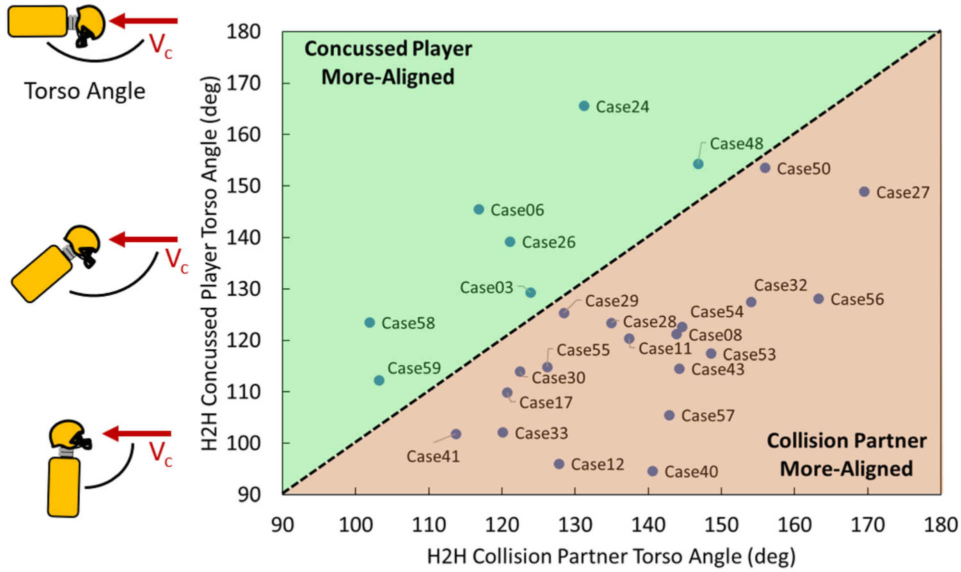

FIGURE 8. Comparison of the angle between the long axis of the torso and the closing velocity vector for single-concussion helmet-to-helmet impacts. Larger torso angles indicate more alignment of the torso with the velocity vector. Diagrams to the left of the chart demonstrate the direction of the closing velocity vector relative to the torso alignment for various angles.

TABLE 1. Summary of multiple impact events.

Case ID	Primary exposure source	Secondary exposure source	$t_{0s} - t_{0p}$	ΔV_p	ΔV_s	$\Delta\omega_p$	$\Delta\omega_s$
8	Helmet	Helmet	8	3.1	1.8	39.5	26.9
10	Ground	Arm	− 50	8.3	5.6	51.1	21.3
13	Helmet	Ground	389	5.0	2.7	20.4	25.5
14	Helmet	Ground	699	7.2	5.5	49.0	39.9
16	Shoulder	Helmet	25	7.1	1.3	20.5	12.5
17	Helmet	Helmet	579	5.6	2.5	33.5	17.8
25	Helmet	Hand	64	5.6	5.3	23.8	22.6
26	Helmet	Shoulder	356	7.4	1	21.6	8.8
28	Helmet	Chest	94	5.3	1.2	36.7	11.9
32	Helmet	Back	− 201	5.8	2.3	41.1	20.1
36	Ground	Ground	508	9.2	1.1	24.8	9.5
39	Ground	Knee	344	4.0	3.6	19.6	23.7
46	Shoulder	Helmet	− 33	4.6	6.4	25.9	19
47	Shoulder	Ground	423	7.8	3.4	32.8	23.1
48	Helmet	Helmet	444	2.9	2.3	27.9	10.3
53	Helmet	Helmet	850	5.1	2.1	41.2	13.9
54	Helmet	Ground	394	7.5	3.5	51.1	22.8
55	Helmet	Ground	583	6.6	5.6	21.4	24.5
56	Helmet	Ground	494	5.7	8.6	31.5	26.9
58	Helmet	Ground	350	8.4	1.9	43.4	31.4
59	Helmet	Ground	236	6.7	7.5	44.9	14.9

Note that the primary exposure was defined subjectively as the most severe impact through a video review process,[19] and that the secondary exposure occurred before the primary exposure in some cases. t_0 refers to time of impact, with p referring to the primary exposure and s referring to the secondary exposure..

29 ± 13 rad s^{-1}) were lower than those measured by the Hybrid III dummy head (7.2 ± 1.8 m s^{-1} and 43 ± 11 rad s^{-1}) in the laboratory reconstructions performed by Pellman *et al.* and reanalyzed by Sanchez *et al.*[26,41] It should be noted, however, that the head and helmet may exhibit different kinematics during the same impact because of relative motion of the head and helmet.[6,21,22,36]

Concussions caused by the ground produced a 50% higher change in translational helmet velocity than concussions caused by contact with another player (8.1 ± 1.7 vs. 5.9 ± 2.0 m s^{-1}, $p < 0.001$). Helmet-to-ground impacts occurred in three main regions: central side, rear upper, and the facemask region. According to a comprehensive video review of concussion-causing impacts in the NFL,[26] a higher proportion of rear impacts were associated with ground contacts than in impacts with other players, and that trend was reflected in this study population. Notable differences in helmet kinematics were observed in ground impacts based on impact direction, largely a result of the body striking the ground prior to helmet impact. In rear impacts, the buttocks and lower torso often made first contact with the ground and produced a 'whipping' motion as the torso rotated in the sagittal plane, often imparting a substantial rotational velocity prior to impact. In the cases where concussion was caused by the side of the helmet striking the ground, the shoulder and helmet often hit the ground almost simultaneously, producing a smaller pre-impact angular velocity of the helmet and an impact velocity vector that was nearly parallel to the x–y plane of the head. The concussive facemask impacts into the ground included in this study showed no pattern, due to the wide variety of body contacts with the ground prior to impact.

Concussions caused by shoulder impacts were similar to helmet-to-helmet impacts in terms of the translational velocity change of the helmet (6.1 ± 2.9 and 5.9 ± 1.5 m s^{-1}, $p = 0.76$) but significantly lower in terms of the rotational velocity change of the helmet (21 ± 7 and 33 ± 11 rad s^{-1}, $p < 0.001$). The impact location regions for these two groups were similar in that they were dominated by impacts to the upper side of the helmet. This is consistent with the distribution observed by a larger video review study and summarized in the supplemental material.[26]

For helmet-to-helmet cases, there was no significant difference between the concussed player and collision partner's average change in translational helmet velocity (5.9 ± 1.5 and 5.0 ± 2.4 m s^{-1}, $p = 0.11$), though the difference in the average change in rotational helmet velocity was significant (33 ± 11 and 25 ± 12 rad s^{-1}, $p = 0.01$), which reinforces previous findings and hypotheses that the incidence of concussion may be more closely related to rotational than translational head kinematics.[14,19,45] The distribution of impact locations differed between the concussed player and uninjured collision partner for these impacts. While the concussed player experienced impacts over the entire front, side, and facemask regions of the helmet, the impact locations for the collision partner were dominated by the front and top regions of the helmet. This was consistent with a previous review of concussive impacts in the NFL.[26,35] In addition, the long axis of the torso was found to be more aligned with closing velocity vector in the uninjured collision partners, with 75% of the uninjured collision partners being more aligned with the closing velocity vector than their concussed counterparts. Consequently, the closing velocity vector was more aligned with the spine in the uninjured collision partners, which may have played a protective role by increasing their effective mass during impact and reducing the change in rotational velocity.

While components of the change in rotational velocity vector (ω_x, ω_y, ω_z) were calculated and are provided in Appendix Tables A1, A2 and A3, it is important to note that these data are based on helmet kinematics rather than head kinematics. Joodaki *et al.*[22] quantified the relative motion of the helmet on the head for a Hybrid-III dummy and found that the helmet can move up to 41 mm and rotate 37° with respect to the head in impacts with closing velocities comparable to those from the current data set. Furthermore, the rotational velocity of the helmet could be greater than or less than the head by as much as 40%.[22] Therefore, the helmet kinematics reported in this study should not be interpreted directly as biomechanical data, but rather as indirect measurements of impact severity.

Some (63%) of the concussion cases were complicated by the presence of multiple head impacts in a single play, though kinematics for those secondary exposures could only be analyzed for 37% of the cases due to visibility in available camera views. None of the multi-exposure events were found to have a secondary exposure with larger changes in both translational and rotational helmet kinematics than the impact designated as the primary exposure through the video review process. Considering limitations related to differences between head and helmet motion,[22] directional-dependence of brain injury,[17,38] and error associated with videogrammetry, the authors view the relative magnitudes of velocity for primary and secondary impacts to be generally consistent with the previously-published video review study.[26]

For multiple impact events, the time between the primary and secondary exposures ranged between 8 and 850 ms. In many cases, the secondary exposure was of comparable severity to the primary exposure. Since the brain behaves as a viscoelastic material, in some cases the primary and secondary exposures may have been close enough temporally that residual strain remained from the first exposure. The findings of this study indicate that this phenomenon could be important in the causation of football concussions, and future efforts should study the effects and frequency of multiple impacts spaced closely in time.

The methodology used in this study had various strengths and limitations relative to other methods of quantifying head impact severity in helmeted sports, such as sensor systems. One strength of video analysis is that each impact can be visually verified, whereas many sensor systems have difficulty distinguishing true impacts from other events.[7,25] Video analysis can also be used to calculate the closing speed between helmets, which is an important parameter that cannot be measured by current head or helmet sensors without additional supplemental information. The present study was an improvement on previous video analysis studies due to the availability of high-quality video data and a model-based image matching methodology that incorporated six degree-of-freedom motion tracking before, during, and after the impact event.

This video analysis was subject to important limitations. The videogrammetric method utilized in this study involved a labor-intensive and time-consuming process. The analysis required laser scans of each stadium and helmet studied and manual frame-by-frame matching of the models to each of the images in every video. The analysis was retrospective and not able to provide real-time impact data for diagnostic purposes, as sensor systems could conceivably do. The number of cases that could be analyzed was limited, so concussive events were intentionally selected. The results are therefore a reasonable characterization of the population of concussive impacts, but do not characterize the population of non-injurious events. As a result, this analysis is subject to selection bias and cannot be used to derive injury risk functions without performing a similar analysis on a representative sample of non-injurious impact events. Additional selection bias may exist within the concussion cases selected for the study since some impacts could not be studied due to obstructed views or limitations in camera resolution. Wearable sensor data are typically collected in all events regardless of injury, so they are not subject to selection bias based on injury outcome. The sampling frequency of video is usually much lower than sensors and is typically inadequate for calculating accelerations.

Other limitations exist in that this video analysis is only able to track helmet motion, that may not be representative of the underlying head motion. A previous study using Hybrid III dummies to simulate on-field football impacts showed that while head and helmet changes in translational velocities were similar, changes in rotational velocities of the head ranged from 71% less than to 37% more than helmet rotational velocities depending on the impact parameters.[22] Thus, the rotational velocities reported for the helmet in this study may deviate from those of the players' heads to a greater degree than those for translational velocities. Since there is likely minimal pre-impact relative motion between the head and helmet, pre-impact helmet closing velocities are assumed to be representative of head closing velocities. It is hypothesized that helmet fit affects the degree of helmet motion relative to the head. While no assessment of helmet fit was performed for this study, proper fit was assumed based on players' helmets being fit by professional equipment managers.

Despite the limitations, this study provides a summary of the conditions under which concussions are sustained in professional football. This information is useful for the development of new helmet testing methodologies that have the goal of mimicking the on-field environment. Future work will utilize the impact locations and closing velocity vectors from this study to define test conditions focused upon evaluating helmet performance under injurious test conditions. Further, this data set can be used to evaluate that test methodology's ability to replicate the change in translational and rotational velocities of the helmet using a laboratory test fixture. Finally, this data set provides a useful tool for assessing whether impacts from different sources (i.e., shoulder, helmet, or ground) can be simulated in the laboratory environment using a single test fixture.

The quantitative impact locations and velocity vector components with respect to the head and helmet during and after impact provide more detailed information that can improve the accuracy of analytical, computational, and physical reconstructions of the motion of the head inside the helmet. This data set also includes several helmet-to-helmet impacts in which one player was concussed and the other was not in the same impact event. The finding that the rotational helmet velocity change was higher in the concussed players than their non-concussed collision partners, while translational helmet velocity change was not different, reinforces the biomechanical consensus that rotational head kinematics play a dominant role in the causation of sports concussions.

ELECTRONIC SUPPLEMENTARY MATERIAL

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