




Laboratory Reconstructions of Concussive Helmet-to-Helmet Impacts in the National Football League

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Abstract—Seventeen concussive helmet-to-helmet impacts occurring in National Football League (NFL) games were analyzed using video footage and reconstructed by launching helmeted crash test dummies into each other in a laboratory. Helmet motion on-field and in the laboratory was tracked in 3D before, during, and after impact in multiple high frame rate video views. Multiple (3–10) tests were conducted for each of the 17 concussive cases (100 tests total) with slight variations in input conditions. Repeatability was assessed by duplicating one or two tests per case. The accuracy of the input conditions in each reconstruction was assessed based on how well the closing velocity, impact locations, and the path eccentricity of the dummy heads matched the video analysis. The accuracy of the reconstruction output was assessed based on how well the changes in helmet velocity (translational and rotational) from the impact matched the video analysis. The average absolute error in helmet velocity changes was 24% in the first test, 20% in the tests with the most accurate input configuration, and 14% in the tests with minimal error. Coefficients of variation in 22 repeated test conditions (1–2 per case) averaged 3% for closing velocity, 7% for helmet velocity changes, and 8% for peak head accelerations. Iterative testing was helpful in reducing error. A combination of sophisticated video analysis, articulated physical surrogates, and iterative testing was required to reduce the error to within half of the effect size of concussion.

Keywords—Biomechanics, Brain injury, Crash test dummy, Hybrid III.

INTRODUCTION

The reenactment of sport-related concussions using crash test dummies is becoming an increasingly common method for studying the biomechanics of head injury.^{5,6,12,13,20,22,25,26,32} Sport provides an environment in which concussions occur regularly and are frequently captured on high-quality video. In helmeted sports such as American football and ice hockey, video analysis can provide detailed kinematics for the players' helmets, but not necessarily their heads. However, video analysis can be supplemented by a laboratory reconstruction with helmeted crash test dummies to simulate the head response in a helmeted impact. The kinematic response of the crash test dummy head (acceleration time histories) can then be used as an input to a finite element model of a human head to estimate stresses and strains within the brain.

The combination of video analysis and laboratory reconstruction for football impacts was pioneered in 1999–2000 by Newman *et al.*^{21,22} as part of a large study on concussions in the National Football League (NFL).²⁵ In that study, video analysis of two camera views was used to determine the closing velocity between players in on-field in-game helmet-to-helmet impacts. Laboratory reconstructions of 25 helmet-to-helmet impacts and 6 ground impacts were conducted by mounting helmeted Hybrid III dummy heads attached to flexible necks to a drop tower or dummy body. The height of the drop was chosen to match the closing speed determined from video, and the orientation of the dummy heads was adjusted by the researchers to visually match the video. Tests were sometimes iterated by

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adjusting the orientations of the heads until the post-impact movement of the dummy heads subjectively matched the game video. The accuracy of the method was assessed by conducting four full-body dummy impact tests on a football field and reconstructing the impacts in the laboratory based on video analysis.²⁰ The absolute error averaged 11% for peak translational head accelerations and 15% for peak rotational head accelerations. These values reflect errors in the video analysis and dummy reconstruction procedure, but do not address errors caused by a lack of dummy biofidelity. The repeatability of the method was not investigated.

Subsequent laboratory reconstruction work has proliferated, but questions about the accuracy and repeatability of the method remain. Although many laboratory reenactments have utilized two on-field camera views in their video analysis,^{12,13,26} many others have relied on an analysis of a single low-frame rate video view.^{5,6,23,32} Some researchers have opted to use a linear impactor instead of a full or partial dummy as a surrogate for a striking player.^{6,26,32} The accuracy of this method has never been fully investigated, but is presumably worse than a crash test dummy due to the impactor arm being constrained to a single degree of freedom. Reported repeatability using a linear impactor as a reconstruction tool has varied widely, with coefficients of variation ranging from 3%²⁶ to 14%³² for peak translational head acceleration and 7%²⁶ to 20%³² for peak rotational head acceleration.

The purpose of the present study was to apply advanced video analysis and laboratory reconstruction methods to determine the biomechanical parameters associated with concussive head impacts in NFL games as accurately as possible. This study focused on reconstructing helmet-to-helmet impacts in which one player was injured and the other was not in order to create a matched case-control study design. Multiple views of high-frame rate video were analyzed using model-based image matching to determine the velocity and orientation of both helmets in six degrees of freedom before, during, and after impact. Iterative testing was conducted by launching helmeted partial Hybrid III crash test dummies into one another using slightly varied input conditions. A secondary purpose of this study was to evaluate the accuracy, repeatability, and sensitivity of the reconstruction process in terms of the benefit gained by applying these more advanced experimental techniques.

MATERIALS AND METHODS

Case Selection

Concussions sustained by NFL players during games are diagnosed with the NFL Game Day Con-

cussion Diagnosis and Management Protocol and recorded in the NFL's league-wide electronic medical record (EMR) system.¹⁷ A player is considered to have sustained a concussion if, after examination and sideline video review by the team physician and the unaffiliated neurotrauma consultant, the diagnosis is made using protocol criteria.⁸ Permission to use these data for research is provided by NFL players for studies jointly approved by the NFL and the National Football League Players Association. This study obtained that approval.

Video footage of professional football games was obtained for all 367 concussive impacts occurring in the 2015-2017 NFL seasons that could be identified on video.¹⁷ Of those, 57 events were selected for in-depth video analysis based primarily on the quality of the video.¹ Of the 57 video-analyzed events, 17 impacts were selected for laboratory reconstruction based on the following criteria:

- The primary impact was a helmet-to-helmet collision between two players
- There was no significant contact to either player immediately before the helmet impact
- There was no significant pre-impact rotational velocity of either helmet
- The helmet impact was clearly visible in at least two distinct camera views
- The video quality was high (resolution > 3000 pixels/helmet and frame rate > 120 fps in at least one camera view)

The present study utilized a case-control plus case series design. In 16 of the 17 cases selected for reconstruction, only one of the two players involved in the impact sustained a concussion. The concussed player was designated "P1" and the uninjured player was designated "P2." In one case (case 2), both players were concussed.

On-Field Video Analysis

Once the camera views were selected for analysis, the videos were analyzed to track the location and orientation of each player's helmet relative to the field. The process was described in detail by Bailey *et al.*² and is summarized here. First, the background in each camera view was "stabilized" to remove the effects of camera movement such as panning and zooming (Nuke X 10.0v, Foundry, London, UK). Next, a technique called "camera matching" (3ds Max 2018, PF-Track 2015.1.1, Pixel Farm, Kent, UK) was used to determine the location of each camera based on information from a laser scan (Faro Focus X330, FARO, Lake Mary, FL, USA) that was taken of each

stadium in which a concussion case occurred. The position and orientation of each player's helmet in all six degrees of freedom was determined for each frame using a model-based image matching technique. The actual helmets worn by the concussed players were obtained and optically scanned (Faro Focus, FARO, Lake Mary, FL, USA) to create the helmet models with outlines of the facemask, logos, and other landmarks. The helmet model was superimposed on each video frame and manually adjusted in all six degrees of freedom until the model matched the video (Houdini, Side Effects Software, Toronto, Canada; 3ds Max, Autodesk, San Rafael, CA, USA). The velocity of each helmet was calculated for 200 ms before, during, and 200 ms after the impact by filtering (4-pole Butterworth filter with a cut-off frequency of 50 Hz) and differentiating the helmet position and orientation data. In addition, a simple torso model in the shape of conical frustum was superimposed on the video views to estimate the angle of the torso relative to the head at impact in flexion, axial rotation, and lateral bending.

Test Apparatus

Helmeted crash test dummies were launched into each other using two custom-made electric-powered belt-driven sleds (Fig. 1). In 15 of the 17 cases, reconstructions were performed by propelling two moving dummies into each other. In the remaining two cases (cases 5 and 18), one dummy was stationary and the other was moving at the time of impact. In all cases, the dummies were equipped with the same model of football helmet as the corresponding player involved in the collision. The size of the helmet was chosen to properly fit on the dummy's head. The Riddell SpeedFlex Precision used in this study was custom-

made based on a scan of the dummy's head. For all other helmets, the large size best fit the dummy's head. The crash test dummies were partial Hybrid III 50th percentile male anthropomorphic test devices (Humanetics Innovative Solutions, Farmington Hills, MI, USA) consisting of a head, neck, torso, and pedestrian pelvis. The upper and lower extremities were removed from the dummies out of concern that their effect on head kinematics would be unrealistic. In some tests, the lower neck mount of the Hybrid III dummy was modified to incorporate a spherical ball joint that allowed for lateral flexion and twist of the neck. Each dummy was hung on a forward-facing steel pin that was connected to the sled carriage via an adjustable frame of T-slot extruded aluminum profiles. The pelvis was supported as well to maintain the position of the dummies as they were accelerated. The dummies were launched into each other by accelerating each carriage up to the desired pre-impact speed, then rapidly stopping it at the end of the sled track. As the carriage decelerated, the dummies slid off their supports and struck each other helmet-first while airborne and under no external forces. Gymnastic mats were placed below the impact area to cushion the falls of the dummies and prevent damage to the equipment.

The desired velocity, position, and orientation of each dummy at impact was calculated from the results of the video analysis. The pre-impact head positions and orientations determined from video analysis were coordinate transformed from the field frame to the impact plane, which was defined by the cross-product of the pre-impact helmet velocity vectors. The sleds, which were portable, were oriented to achieve the angle between the pre-impact velocity vectors of each player's head in the impact plane. The dummies could be positioned and oriented in a variety of ways by adjusting the arms of the extruded aluminum frame. Proper orientation of the dummy head was verified by bolting a piece of angle square to the dummy's vertex, measuring the elevation angles of the head axes with an inclinometer, and adjusting the dummy position until the measured angles matched the desired angles calculated from video analysis. A similar process was followed for the torso by matching two inclinometer measurements referencing the spine box. The modified lower neck ball joint allowed both the head and torso of the dummies to be positioned at the same angles as the human players. The height of the top of each dummy's helmet above the ground and the left/right position of the carriage on the sled were adjusted based on coordinate transformed values calculated from video analysis. Adjustment of the dummy position in the fore-aft direction was achieved by altering the position of the sleds or the relative timing of the two sled acceleration pulses.

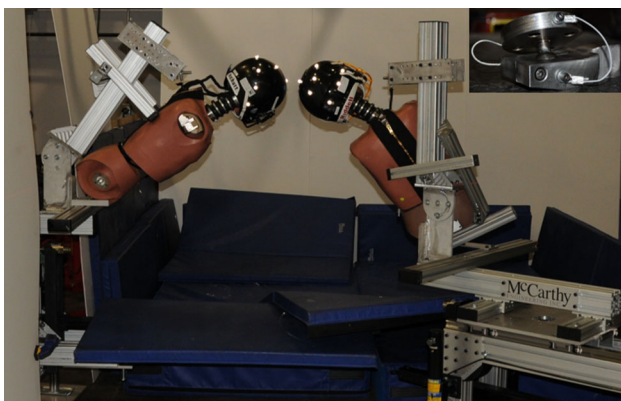


FIGURE 1. Photograph of the laboratory reconstruction test setup. Inset shows the ball joint added to the dummy neck in some tests to achieve a wider variety of positions.

Data Processing

The dummy responses were quantified by dummy instrumentation and video analysis. The dummy heads were instrumented with 6DX-Pro 2000-8 k sensors (Diversified Technical Systems, Seal Beach, CA, USA) at their centers of gravity and 3 single-axis accelerometers (Endevco 7264B-2000, Meggitt, Irvine, CA) at remote locations within the dummy skull. Data were acquired at a sample rate of 10 kHz using a SLICE data acquisition system mounted inside the dummy. Head accelerometer data were filtered to CFC1000 and head angular rate data were filtered to CFC180 in accordance with SAE J211.²⁸ Head sensor array data were used to calculate six degree-of-freedom head kinematics at the head center of gravity.⁷ The Head Injury Criterion (HIC) was calculated from the accelerometer data.³⁰ A kinematic rotation-based severity metric, diffuse axonal multi-axis general evaluation (DAMAGE) was calculated from the angular acceleration data.¹¹

The motion of the helmets was measured using 3D motion tracking video analysis. Retroreflective fiducial markers (Natural Point Inc., OR, USA) were mounted on M4 nylon threaded rods by tapping holes into the helmet shell at locations that would not interfere with the helmet contact. The positions of each marker relative to the head center of gravity were determined prior to each test by placing the helmeted head in a calibration jig. Each test was captured on digital video at 500 Hz by three Edgertronic SCI high speed cameras positioned at widely separated angles in the laboratory. Using a commercial software package (ProAnalyst3D, Xcitex, MA, USA), the fiducial markers were tracked in multiple camera views and the motion of both helmets in all six degrees of freedom was calculated. The change in translational and rotational velocity of each helmet was calculated by differentiating the displacement data. In some cases, the beginning and ending times used to calculate helmet velocity changes were truncated to exclude pre-impact rotational helmet velocity or closely spaced secondary impacts that could not be duplicated with our test apparatus. For that reason, some of the helmet velocity changes reported here differ from the corresponding video analysis results of Bailey *et al.*,¹ which included the full peak-to-peak velocity change of the helmet.

Test Protocol

Iterative testing was conducted for all cases. The first test was always an attempt to match the video-derived input conditions (i.e. impact velocity, location, and path eccentricity) as closely as possible. Additional

tests were conducted when the post-impact motion of the dummy heads in the reconstruction did not match the game video. The process was often repeated several times to improve the match between the reconstruction and the video. In several cases, sensitivity tests were conducted in which the initial position of one of the dummies was systematically altered by moving it up, down, left, and right of the initial position by 25–50 mm. Over the course of testing, several test conditions (1–2 per case) were repeated with the same initial conditions.

Data Analysis

For most parameters, repeatability was quantified by the coefficient of variation (standard deviation divided by the mean). The precision and repeatability of the dummy paths and contact points were quantified by standard deviation and range.

The accuracy of the reconstruction inputs and outputs were evaluated separately. Our test apparatus was not capable of imparting pre-impact rotational velocity to the dummies. Given that limitation, a complete kinematic description of the helmet interaction at impact could be obtained from the translational closing speed, path eccentricity, and impact location on each helmet (Fig. 2). The closing speed was determined based on the motion of the centers of gravity of the dummy heads immediately prior to first contact. The path eccentricity was calculated as the perpendicular distance between the center of gravity of one dummy head and the closing velocity vector projected from the center of gravity of the other dummy head. The resultant path eccentricity was the same for both dummies, but the components differed depending on the orientation of each head.

For the purpose of quantifying the input accuracy of each reconstruction test, the contact point was assumed to be the midpoint between the centers of gravity of the two dummy heads in the frame immediately preceding first contact. This method was simple and sufficient to ensure that each helmet was correctly oriented in yaw, pitch, and roll at the time of impact. However, the method may have had some error in identifying the exact location of contact on the helmet because it effectively assumed that the two helmets were spheres of equal size. The impact location was characterized in terms of the azimuth and elevation angles of the contact point in each player's head coordinate frame. Impact locations quantified from video analysis and reconstruction tests were compared to the subjective assessments of helmet impact region in these same cases that were made in the previous video review study.¹⁷

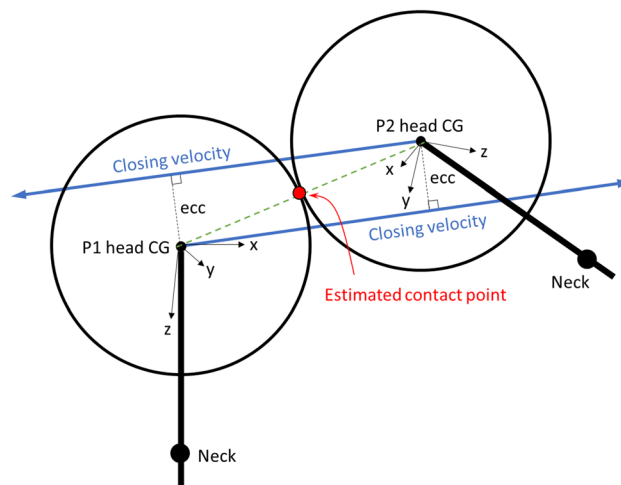


FIGURE 2. Schematic showing how path eccentricity and helmet impact location were calculated.

To characterize accuracy of the input configuration in each reconstruction with a single parameter, a composite input error was developed. The composite input error was calculated by averaging the error in the path eccentricity and the impact location for each helmet:

$$e_{\text{composite input}} = \frac{e_{\text{path ecc}} + e_{\text{P1 contact}} + e_{\text{P2 contact}}}{3} \quad (1)$$

The error in path eccentricity was calculated as the distance between the tips of the path eccentricity vectors in the reconstruction and game video (both in the player's head coordinate frame). The error in helmet impact location for both P1 and P2 was calculated as the distance between the estimated contact points in the reconstruction and game video in the head coordinate frame of each player. Error in closing velocity was not included in the composite input error because it had different units and could not be incorporated in a straightforward fashion.

The accuracy of the output of each reconstruction was assessed by comparing the translational and rotational velocity changes of both players' helmets relative to the video analysis results. Resultant translational helmet velocity changes were compared directly:

$$e_{\Delta V} = \frac{\Delta V_{\text{rec}} - \Delta V_{\text{vid}}}{\Delta V_{\text{vid}}} \quad (2)$$

When calculating the error in the rotational helmet velocity change, we wished to account for the fact that certain directions of rotation are more damaging to the brain than others. Therefore, the directional components of rotational helmet velocity change were weighted according to the intercepts for the Brain Injury Criterion (BrIC)²⁹:

$$e_{\Delta\omega} = \frac{\sqrt{\left(\frac{\Delta\omega_{x-\text{rec}}}{66.25}\right)^2 + \left(\frac{\Delta\omega_{y-\text{rec}}}{56.45}\right)^2 + \left(\frac{\Delta\omega_{z-\text{rec}}}{42.87}\right)^2} - \sqrt{\left(\frac{\Delta\omega_{x-\text{vid}}}{66.25}\right)^2 + \left(\frac{\Delta\omega_{y-\text{vid}}}{56.45}\right)^2 + \left(\frac{\Delta\omega_{z-\text{vid}}}{42.87}\right)^2}}{\sqrt{\left(\frac{\Delta\omega_{x-\text{vid}}}{66.25}\right)^2 + \left(\frac{\Delta\omega_{y-\text{vid}}}{56.45}\right)^2 + \left(\frac{\Delta\omega_{z-\text{vid}}}{42.87}\right)^2}} \quad (3)$$

To characterize accuracy of the output of each reconstructed impact with a single parameter, a composite output error was developed. Using a standard propagation of errors approach and assuming the errors to be independent, the composite output error of each reconstruction was calculated as the resultant of the four helmet velocity change errors:

$$e_{\text{composite output}} = \sqrt{e_{P1\Delta V}^2 + e_{P1\Delta\omega}^2 + e_{P2\Delta V}^2 + e_{P2\Delta\omega}^2} \quad (4)$$

One feature of this approach is that the resultant error is two to four times larger than the average absolute error, depending on the spread in the four errors. The average absolute output error of the four helmet velocity change errors was also calculated:

$$e_{\text{average output}} = \frac{|e_{P1\Delta V}| + |e_{P1\Delta\omega}| + |e_{P2\Delta V}| + |e_{P2\Delta\omega}|}{4} \quad (5)$$

For each case, the test with lowest average absolute output error was chosen to be the optimized test representing that particular case. Average absolute output errors among the optimized tests were compared to the effect size of concussion (d), which was calculated in a similar manner as the errors above, but using average helmet velocity changes calculated from video analysis:

$$d_{P1\Delta V} = \frac{\Delta V_{P1\text{vid}} - \Delta V_{P2\text{vid}}}{\Delta V_{P1\text{vid}}} \quad (6a)$$

$$d_{P1\Delta\omega} = \frac{\Delta\omega_{P1vid} - \Delta\omega_{P2vid}}{\Delta\omega_{P1vid}} \quad (6b)$$

$$d_{P2\Delta V} = \frac{\Delta V_{P2vid} - \Delta V_{P1vid}}{\Delta V_{P2vid}} \quad (6c)$$

$$d_{P2\Delta\omega} = \frac{\Delta\omega_{P2vid} - \Delta\omega_{P1vid}}{\Delta\omega_{P2vid}} \quad (6d)$$

$$d_{\text{average}} = \frac{1}{4}[d_{P1\Delta V} + d_{P1\Delta\omega} + d_{P2\Delta V} + d_{P2\Delta\omega}] \quad (6e)$$

The relationship between peak translational and rotational velocity changes was investigated using the Pearson correlation coefficient and by calculating an effective radius of rotation, which was defined as the ratio of peak translational to peak rotational velocity change:

$$r_{\text{effective}} = \frac{\Delta V}{\Delta\omega} \quad (7)$$

RESULTS

Case Selection

Of the 17 cases selected, 13 (76%) were passing plays, 3 (18%) were rushing plays, and 1 (6%) was a kickoff return (Table 1). The most common positions in the concussed group of players were wide receiver (39%), safety (22%), and tight end (17%), while their uninjured collision partners consisted mostly of safeties (38%), linebackers (25%), and running backs (19%). Most players in the cases selected wore a large helmet (68%), with the remainder wearing medium (18%), extra-large (6%), or custom Riddell SpeedFlex Precision (9%) helmets.

Video Analysis

According to video analysis, the average closing velocity between helmets in the 17 concussive impacts studied was 9.6 ± 1.5 m/s (Table 2). The impact with the highest closing velocity (12.4 m/s) was case 2, the only case studied in which both players sustained a concussion. Concussed players experienced helmet velocity changes of 6.3 ± 1.5 m/s and 34 ± 10 rad/s, while the uninjured players experienced helmet velocity changes of 4.7 ± 1.6 m/s and 23 ± 10 rad/s. The estimated contact point (Fig. 1) was significantly lower on the helmet in the concussed players than their uninjured collision partners (elevation angle = $14^\circ \pm 32^\circ$ vs. $42^\circ \pm 20^\circ$, $p = 0.005$ in a two-tailed paired t-test). Path eccentricity was highly variable, with some im-

pacts being nearly centric and some quite glancing. The average path eccentricity (124 ± 63 mm) was approximately half the diameter of a typical football helmet shell.

Repeatability

134 reconstruction tests were performed in total. Due to video tracking errors (17 tests), dummy instrumentation data collection errors (16 tests), and hardware failures (2 tests), 34 tests had to be excluded, leaving 100 tests available for analysis. For each case, 1–2 test conditions were repeated 2–3 times. In total, there were 22 test conditions that were repeated, involving 59 of the 100 tests. Among the 22 repeated test conditions, the coefficient of variation was 3% for closing speed. The standard deviations in path eccentricity and contact points averaged 15 mm and 6 degrees (~ 14 mm), respectively. The range in distance between estimated helmet contact points in repeated tests was 9 degrees (~ 21 mm) on average. Among the output parameters, head velocity changes were most repeatable (COVs of 4% and 6% for translation and rotation), followed by helmet velocity changes (COVs of 6% and 8% for translation and rotation), head accelerations (COVs of 7% and 9% for translation and rotation), and associated injury criteria (COVs of 11% for HIC and 7% for DAMAGE).

Optimization

The iterative testing approach exposed surprising tradeoffs that prevented a good match of all parameters in all cases (Table 3). In many cases, adjustments in initial conditions would succeed in improving the match in helmet velocity change for one player but degraded the match for the other player. Test results were also highly sensitive to small changes in impact location and path eccentricity, as would be expected in a collision between two spheroidal objects. This sensitivity was generally higher for players struck near the “equator” of the helmet as opposed to the top of the helmet. Particularly concerning was the observation in many cases of a tradeoff between input accuracy (Eq. 1) and output accuracy (Eq. 4). Optimized tests were chosen based solely on output accuracy. Interestingly, the optimized tests often proved to be exploratory tests that were not run explicitly to improve the match with video, but rather to assess repeatability or sensitivity. All of these observations apply to the reconstruction testing for case 10, which is described in depth in the Online Appendix.

In spite of its difficulties, the iterative testing method resulted in a substantial improvement in the accuracy of the reconstructions. The first test conducted for each

TABLE 1. Information for the 17 selected cases (case 3 was excluded).

Case	P1 position	P2 position	Play type	P1 helmet	P2 helmet	# Tests
1	S	RB	Rush	L Schutt Air XP Pro	L Riddell SpeedFlex	9
2	S	WR	Pass	L Schutt Air XP Pro	M Riddell Speed Classic	9
4	CB	WR	Pass	M Rawlings NRG Quantum	L Schutt Air XP Pro	5
5	WR	LB	Pass	L Schutt Vengeance Z10	XL Schutt Air XP	7
6	LB	RB	Rush	L Schutt Air XP Pro	L Riddell Speed Classic	9
7	TE	LB	Pass	XL Riddell Revolution Speed	L Riddell SpeedFlex	4
8	WR	S	Pass	L Rawlings Impulse	M Xenith X2E	3
9	S	RB	Kickoff Return	M Riddell SpeedFlex	L Riddell Revolution Speed	6
10	QB	CB	Rush	L Riddell Revolution Speed	L Riddell Speed Classic	7
11	TE	CB	Pass	L Riddell Revolution Speed	L Schutt Air XP Pro	4
12	WR	S	Pass	L Schutt Air XP Pro	L Rawlings Impulse +	6
13	S	LB	Pass	L Riddell SpeedFlex	L Riddell SpeedFlex	7
14	WR	S	Pass	L Schutt Air XP Pro	L Riddell Revolution Speed	5
15	WR	LB	Pass	L Schutt Air XP Pro Q10	Riddell SpeedFlex Precision	5
16	WR	S	Pass	M Schutt Air XP Pro	Riddell SpeedFlex Precision	5
17	TE	S	Pass	Riddell SpeedFlex Precision	M Riddell SpeedFlex	4
18	CB	S	Pass	L Schutt Air XP Pro	L Schutt Air XP Pro Q10	5

TABLE 2. Video analysis results.

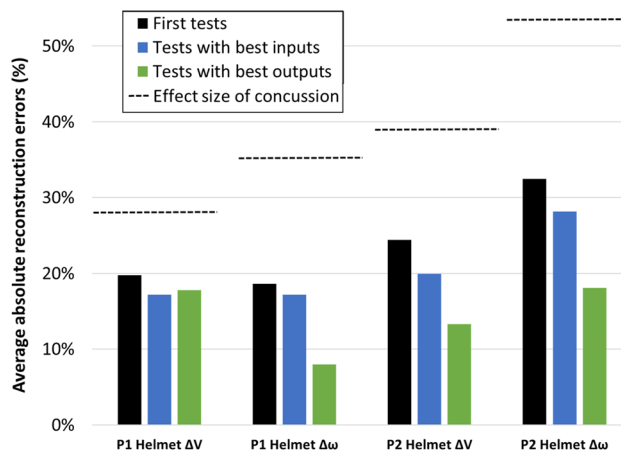
Case	Helmet velocity changes					Helmet impact locations				Path Eccentricity (mm)
	Closing velocity (m/s)	Concussed		Uninjured		Concussed		Uninjured		
		Translational (m/s)	Rotational (rad/s)	Translational (m/s)	Rotational (rad/s)	Azimuth (deg)	Elevation (deg)	Azimuth (deg)	Elevation (deg)	
1	9.5	5.6	29	5.1	29	- 37	36	104	51	107
2	12.4	7.3	49			71	53			49
2		10.7	45			- 73	51			
4	11.8	6.1	46	6.9	52	- 103	51	56	38	114
5	5.8	5.6	14	1.6	12	- 113	- 16	25	31	138
6	9.8	4.0	31	5.0	12	- 46	56	- 32	61	10
7	10.1	5.2	33	4.7	17	129	34	- 6	1	146
8	10.1	6.0	30	7.9	21	- 53	- 54	103	63	227
9	9.9	5.5	32	4.4	17	29	9	- 109	37	136
10	9.9	5.8	29	3.0	25	47	- 1	- 121	11	203
11	9.9	5.6	25	5.4	26	- 43	- 8	77	38	161
12	8.7	5.1	41	3.4	32	111	12	- 102	48	143
13	12.3	7.3	51	6.2	18	- 81	60	36	36	180
14	8.2	6.6	22	6.1	14	- 71	- 10	53	70	151
15	9.0	6.3	36	4.6	24	- 65	- 12	47	50	41
16	9.5	5.8	32	3.5	22	- 48	- 11	114	32	166
17	8.7	6.7	44	3.7	18	- 25	20	70	73	126
18	10.1	8.4	31	3.7	26	20	- 9	- 87	33	15
Mean	9.6	6.3	34	4.7	23		14		42	124
SD	1.5	1.5	10	1.6	10		32		20	63

case resulted in an average absolute error of $24\% \pm 14\%$ in helmet velocity changes (Fig. 3). This was two thirds of the average absolute effect size of concussion (which was 36%). Overall, the first test conducted for each reconstruction was also the test with lowest composite output error in 5 cases, meaning that iteration improved the outcome of the reconstruction in the other 12 of 17 (72%) cases. Choosing tests with the

most accurate input conditions reduced the input error substantially (-35% relative to the first test) but did not significantly improve the output error (- 7% relative to the first test). In fact, in 82% of cases (14/17), the test with the most accurate inputs did not produce the most accurate outputs. Interestingly, the tests with the best outputs had composite input errors that were only slightly (14%) lower than the first test conducted for

TABLE 3. Optimized laboratory reconstruction tests having the lowest composite output error for each case.

Optimized test	Closing velocity (m/s)	Path eccentricity (mm)	Path eccentricity error (mm)	Composite input error (mm)	Composite output error	Average absolute output error
1-6	8.9	45	88	57	14%	6%
2-18	11.7	59	59	36	35%	14%
4-12	10.7	74	41	31	19%	9%
5-5	5.4	162	30	21	28%	12%
6-5	9.7	96	89	66	47%	18%
7-2	10.7	205	59	39	19%	7%
8-5	9.7	199	38	34	20%	8%
9-6	9.9	362	226	155	65%	26%
10-6	10.0	307	105	81	65%	30%
11-1	7.6	162	86	116	30%	14%
12-4	8.2	192	54	80	13%	5%
13-5	13.0	124	62	39	59%	22%
14-1	7.8	200	76	43	37%	13%
15-5	8.7	78	37	27	27%	11%
16-1	9.4	159	28	26	46%	21%
17-5	7.8	114	15	16	24%	11%
18-1	8.6	59	55	57	33%	14%
Mean	9.3	153	68	54	34%	14%
SD	1.8	88	48	37	17%	7%

**FIGURE 3. Average absolute errors in various classes of reconstruction tests relative to the effect size of concussion.**

each case and substantially (32%) lower than the tests with the best inputs. Choosing tests with the lowest composite output error reduced the average absolute error to $14\% \pm 7\%$ (39% lower than the first tests), which was less than half (39%) of the effect size of concussion.

Accuracy

Overall, the optimized dummy reconstructions recreated the helmet velocity changes with a low average errors (-6% for P1 ΔV , 1% for BRIC-weighted P1 $\Delta\omega$, 6% for P2 ΔV , and 12% for BRIC-weighted P2 $\Delta\omega$). However, there was significant scatter in the

results (Fig. 4), so the average absolute errors were relatively high (18% for P1 ΔV , 8% for BRIC-weighted P1 $\Delta\omega$, 13% for P2 ΔV , and 18% for BRIC-weighted P2 $\Delta\omega$) (Fig. 3). The closing velocity between the dummy heads was $-5\% \pm 7\%$ (average absolute error = 6%) of the target closing velocity determined from video analysis. These reconstruction tests missed the target path eccentricity by 64 ± 49 mm and the target helmet contact points by 45 ± 43 mm. The contact points in the reconstructed cases were widely distributed over the helmets with a concentration on the upper side and facemask areas (Fig. 5). The contact points were significantly higher on the helmets of the uninjured players than the concussed players (elevation

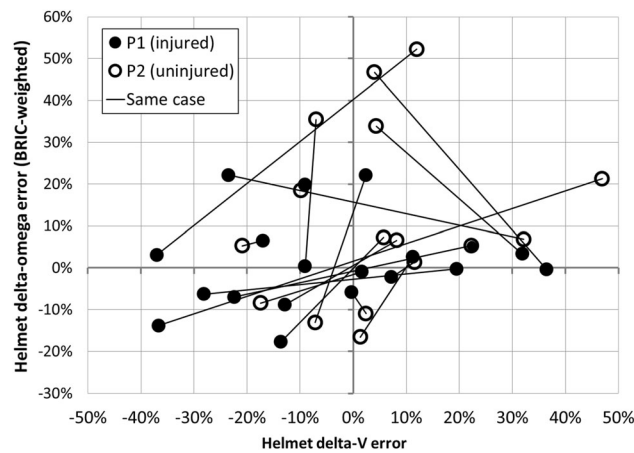


FIGURE 4. Error in helmet velocity changes for the optimized reconstruction tests with the lowest composite output error.

angles determined from video analysis were $42^\circ \pm 20^\circ$ for uninjured players vs. $16^\circ \pm 32^\circ$ for injured players, $p < 0.01$). The impact location on the helmet of the concussed players determined using the spherical helmet assumption in this study (Fig. 2) matched the subjective assessment of these same cases in the previous video review study¹⁷ quite well. The contact point determined from video analysis clearly matched the region(s) of the helmet identified as the contact point in the video review study in 28 out of 34 (82%) comparisons, was very near the boundary (within about 10 mm) in another 5 comparisons (15%), and was only clearly different for one (3%) player. The helmet contact points in the reconstruction tests did not match the video review study as closely due to impact location errors. There was a clear match in 22 comparisons (65%), a near match in 6 comparisons (18%), and a clear mismatch in 6 comparisons (18%).

The calculated change in velocity of the head did not always match the helmet in the dummy reconstructions (Tables 4 and 5). It should be noted that the head and helmet velocity changes are not directly comparable in this study because they were calculated differently. Head velocity changes in the dummies were measured by dummy sensors that were sampled at a high rate (10,000 Hz), while the helmet velocity changes were calculated by differentiating position data from much lower frame rate video (60–500 Hz). Overall, the calculated helmet velocity change was $7\% \pm 15\%$ lower than the head velocity change in translation and $12\% \pm 21\%$ lower than the head velocity change in rotation. The average absolute differences in velocity change between the helmet and the head were $12\% \pm 11\%$ in translation and $19\% \pm 14\%$ in rotation ($16\% \pm 13\%$ combined).

Video Analysis vs. Laboratory Reconstructions

The dummy reconstructions tended to reduce the differences between the concussed and uninjured players. For example, the change in velocity of the helmet was consistently higher in the concussed players ($66\% \pm 15\%$ of closing velocity) than their uninjured counterparts ($49\% \pm 14\%$ of closing velocity) in the video analysis ($p < 0.05$). However, comparing concussed players to their uninjured collision partners (excluding case 2 where both players were injured), helmet translational velocity change was $45\% \pm 69\%$ higher in the video analysis but only $29\% \pm 71\%$ higher in the optimized laboratory reconstructions. Likewise, helmet rotational velocity change was $58\% \pm 59\%$ higher for concussed players in the video analysis but only $41\% \pm 40\%$ higher in the reconstructions (Fig. 6).

In addition, the helmet behaved as if it were more tightly coupled to the body in the reconstruction tests than in the game impacts. There was a stronger relationship between translational and rotational velocity changes in the dummies compared to the humans. The correlation between the translational and rotational helmet velocity changes as determined from video analysis was very low ($R^2 = 0.18$). This correlation was much higher in the reconstruction tests, both for the helmet ($R^2 = 0.35$) and especially for the head ($R^2 = 0.53$). The average effective radius of helmet rotation was 11% lower for all players in the reconstruction tests (190 ± 67 mm) compared to the video analysis (213 ± 91 mm) ($p = 0.006$), indicating relatively more helmet rotation for the same translation (or alternatively, that there was less translation for the same rotation) in the reconstruction tests than the

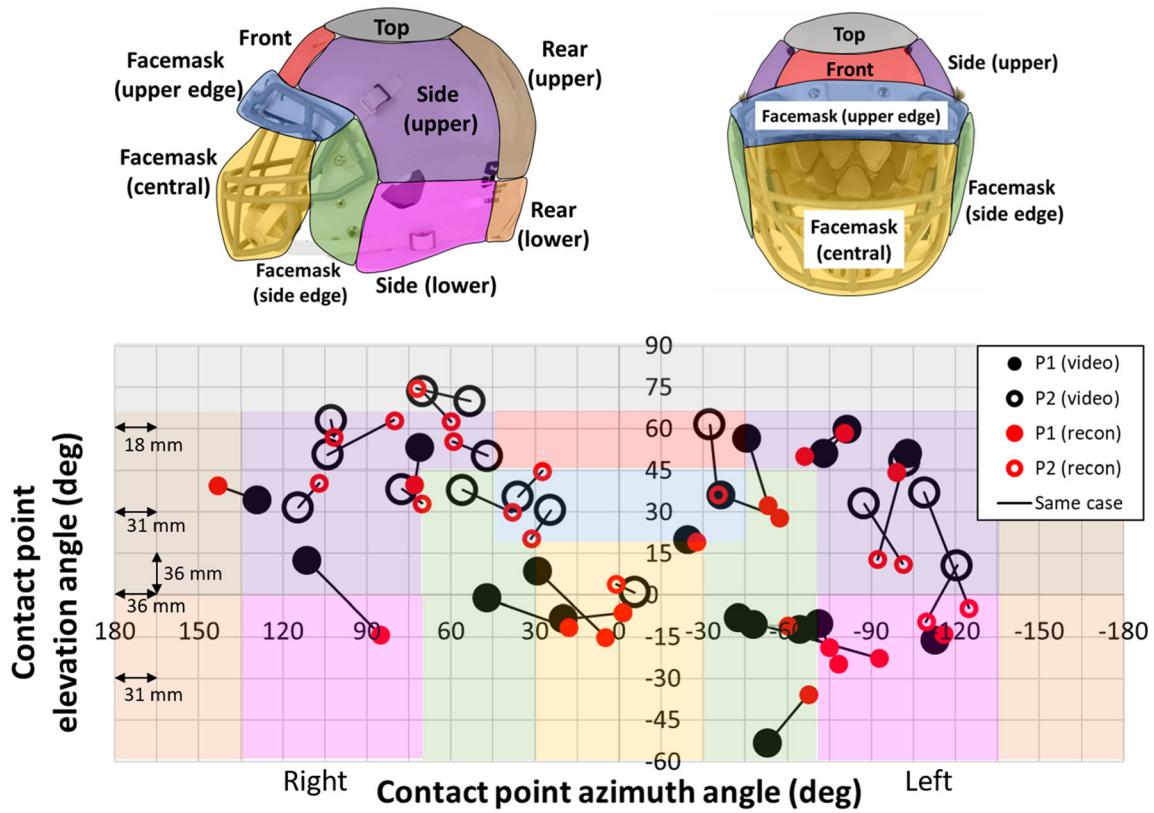


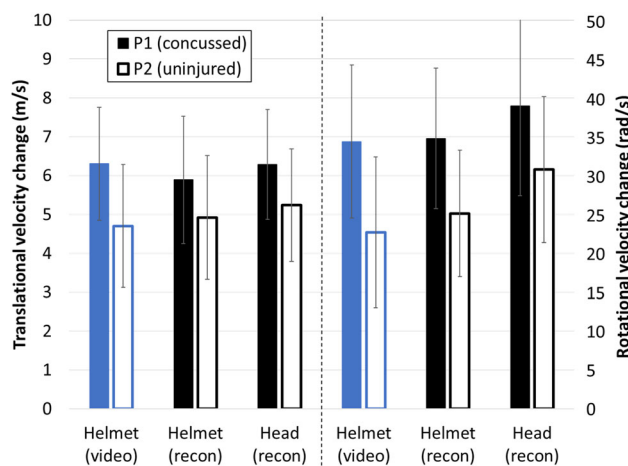
FIGURE 5. Estimated contact points in the video analysis and reconstructions plotted as a Mercator projections with regions of the helmet identified by color coding.

TABLE 4. Biomechanical data for the concussed player (P1) in each optimized reconstruction.

Optimized test	Helmet velocity change		Head velocity change		Peak head acceleration		Injury criteria	
	Translational (m/s)	Rotational (rad/s)	Translational (m/s)	Rotational (rad/s)	Translational (g)	Rotational (rad/s ²)	HIC	Damage ϵ
1-6	6.0	31	7.2	39	74	5822	237	0.29
2-18	8.7	51	9.7	58	143	8453	795	0.41
2-18	7.7	41	8.0	39	105	7331	592	0.30
4-12	5.3	43	7.0	55	104	5514	372	0.41
5-5	4.7	19	4.3	23	33	3750	42	0.23
6-5	5.3	31	7.1	46	115	6054	473	0.35
7-2	5.3	32	6.0	32	79	4146	217	0.28
8-5	6.7	31	6.0	48	60	7167	164	0.61
9-6	3.5	34	3.9	21	43	5542	71	0.25
10-6	3.7	25	4.1	27	55	3693	87	0.31
11-1	5.1	33	5.6	30	46	3531	125	0.31
12-4	5.1	40	5.8	38	66	5731	180	0.30
13-5	10.0	49	7.7	49	134	11798	637	0.42
14-1	6.0	19	6.4	30	76	4257	246	0.33
15-5	6.4	46	6.1	53	103	8348	401	0.60
16-1	4.4	38	6.1	51	84	7703	262	0.58
17-5	5.8	34	5.8	27	57	2413	160	0.24
18-1	6.5	30	6.3	35	69	4161	200	0.27
Mean	5.6	33	6.3	39	80	5857	292	0.36
SD	1.5	8	1.4	12	31	2345	212	0.12

TABLE 5. Biomechanical data for the uninjured player (P2) in each optimized reconstruction.

Optimized test	Helmet velocity change		Head velocity change		Peak head acceleration		Injury criteria	
	Translational (m/s)	Rotational (rad/s)	Translational (m/s)	Rotational (rad/s)	Translational (g)	Rotational (rad/s ²)	HIC	Damage ϵ
1–6	5.7	31	5.7	28	55	3958	147	0.22
4–12	7.4	43	8.3	51	120	7017	519	0.41
5–5	1.3	10	2.2	10	37	1815	28	0.10
6–5	5.2	17	5.3	30	68	4787	192	0.28
7–2	3.9	19	6.1	24	64	3114	187	0.29
8–5	8.0	20	6.7	37	50	3972	87	0.27
9–6	5.0	24	5.1	33	51	3700	65	0.28
10–6	4.5	32	4.4	34	62	4964	106	0.34
11–1	4.9	34	4.7	36	59	3442	100	0.25
12–4	3.4	28	3.7	33	44	2876	53	0.24
13–5	6.4	28	6.3	42	96	5444	332	0.33
14–1	5.7	17	6.0	21	43	3206	41	0.22
15–5	4.3	20	3.9	22	54	3072	78	0.18
16–1	4.6	31	4.9	35	56	3683	124	0.27
17–5	3.9	23	4.3	28	47	3684	77	0.22
18–1	4.6	24	6.3	32	65	3943	161	0.28
Mean	4.9	25	5.2	31	61	3917	143	0.26
SD	1.6	8	1.4	9	21	1198	125	0.07

**FIGURE 6. Translational (left) and rotational (right) velocity changes of the helmet and head.**

game impacts. When broken down by injury status, the average effective radius of helmet rotation was 17% lower in the concussed players relative to the uninjured players in both the video analysis (197 ± 72 mm vs. 231 ± 109 mm) ($p = 0.16$) and the optimized reconstruction tests (175 ± 54 mm vs. 206 ± 81 mm) ($p = 0.06$). This result indicates that the rotational component of helmet motion was more pronounced in the concussed players than their uninjured counterparts. Compared to the helmet, the average effective radius of rotation of the dummy head was lower and more uniform in the reconstruction tests (165 ± 31 mm for concussed players vs. 178 ± 48 mm for uninjured players).

DISCUSSION

The purpose of the present study was to determine the biomechanical parameters associated with concussive helmet-to-helmet impacts in NFL games through advanced video analysis and laboratory reconstruction methods. The current study was very similar to the previous NFL work^{22,25} in terms of the average closing velocity between players in concussive helmet-to-helmet impacts (9.6 m/s vs. 9.3 m/s), but differed somewhat in the reconstructed head kinematics of concussed players. Dummies representing concussed players experienced lower translational head kinematics in the present study compared to the pre-

vious NFL study^{22,25} (6.3 m/s vs. 7.2 m/s head velocity change, 80 g vs. 98 g peak head acceleration, and HIC of 292 vs. 381), but fairly similar rotational head kinematics (39 rad/s vs. 35 rad/s head velocity change and 5857 rad/s² vs. 6432 rad/s² peak head acceleration). Relative to the previous NFL study,^{22,25} the present work incorporated several key methodological advances and a more rigorous error analysis. A review of the methodology was warranted because subsequent analyses have raised critical questions about the accuracy of the original results.^{9,10,27} In addition, recent dummy reconstruction work related to sports concussions^{5,6,23,26,32} has tended to rely on more simplified video analysis and reconstruction methods than the earlier NFL study, so it is important to quantify the differences in error related to methodology.

The current study utilized several techniques to reduce the error in both the video analysis and the laboratory testing. Case selection criteria required multiple unobstructed camera views with at least one recording at a high frame rate (120–480 images/s). Model-based image matching using a scanned image of the same helmet that the player was wearing allowed for precise measurement of the helmet position in each frame in all six degrees of freedom, which is necessary for a full kinematic description of the impact conditions. Accurately tracking the rotational motion of the helmets was particularly important, because concussion has been biomechanically linked primarily to rotational head motion.²⁹ In the present study, the path eccentricity of the heads and impact locations on each helmet were determined quantitatively. Previous studies have not tracked path eccentricity at all and have only determined the helmet impact locations qualitatively by visually determining which region of the helmet was struck. Based on the observation that the results of the reconstruction tests can be extremely sensitive to small changes in impact location and path eccentricity ($> 1\%$ error per mm), those parameters must be tracked independently and quantitatively in order to ensure accuracy. This sensitivity is especially high in more glancing impacts, where a small change of eccentricity can greatly change the angle of contact due to both helmets being spheroidal in shape.

Perhaps most importantly, video tracking of the helmet motion was performed throughout the duration of the impact. To our knowledge, all previous researchers have only tracked the translational velocity of the helmet before the impact and relied on the laboratory testing to accurately simulate the helmet (and head) motion during and after impact. We based our error calculations and optimization approach in laboratory testing on matching helmet velocity changes observed on video because helmet velocity change proved to be a more biomechanically relevant param-

eter than closing velocity. In this case-control study, the closing velocity was identical (by design) in concussed and uninjured players, whereas translational and rotational helmet velocity changes were significantly higher in concussed players compared to their uninjured collision partners. Unfortunately, because previous studies have only tracked pre-impact helmet velocity on video instead of velocity change, the reconstruction output errors in this study cannot be compared to other studies. We chose to weight the errors in helmet rotational velocity change by directional component using the weights in the BrIC formulation²⁹ in an effort to optimize our reconstructions in the most biomechanically relevant directions of head rotation. Had we not weighted helmet angular velocity error in this way, we would still have chosen the same test as optimal in 13 out of 17 cases (cases 5, 11, 15, and 17 would have been different).

Advanced laboratory techniques helped better replicate the concussive impacts captured on video. The repeatability of the test apparatus was as good or better than previous reconstruction studies.^{26,32} The spread in closing speed, impact location, and path eccentricity among repeated tests (as quantified by standard deviation) was only 3%, 6 degrees (~ 14 mm), and 15 mm, respectively. Variation was higher in repeated tests for output parameters such as helmet velocity change (6%–8%), head velocity change (4%–6%), peak dummy head acceleration (7%–9%), HIC (11%), and DAMAGE (7%). Advanced physical surrogates were employed in the present study: namely, two helmeted partial (head, neck, and torso) 50th percentile male Hybrid III crash test dummies. The dummies were launched from sleds and struck each other in free flight to ensure that the impact forces were solely the result of specimen inertia. Previous reconstruction studies have utilized simpler physical surrogates, such as a helmeted dummy head and neck mounted to a sliding track²² or a padded metal cylinder.^{6,26,32} These simplified surrogates do not always match the effective torso mass of the more complete dummy and introduce unrealistic constraining forces at the boundary conditions. Even the complete (but limbless) dummies in the current study proved problematic due to problems with biofidelity, as evidenced by the fact that the tests with the most accurate inputs did not produce the most accurate outputs (Fig. 5). The use of limbless dummies and a lack of ground reaction forces in the present study are limitations of this reconstruction method that need to be investigated further. These limitations reinforce the need for iterative testing and video tracking during and after impact to verify and optimize the fidelity of the reconstruction.

In spite of the state-of-the-art methodology employed in the present study, reconstruction errors were

still high in many cases. The average absolute errors in the reconstruction process were 16% and 13% for the changes in helmet translational and rotational velocity, respectively (14% overall). Unfortunately, the problems of dummy biofidelity and sensitivity to initial conditions are fundamental and unlikely to be amenable to significant improvement, barring significant technological advances. The reconstruction test errors were compounded by the video analysis errors, which were estimated to be about 9% and 17% in terms of average absolute errors in the changes in helmet translational and rotational velocity, respectively.² Furthermore, even if a reconstruction matched the helmet motion precisely, the movement of the head within the helmet may differ between a dummy and human. This error is unknown, but it may be substantial given the high degree of movement that can occur between the helmet and the head in a football impact.¹⁴ Furthermore, the helmet may be more tightly coupled to the head of some players than others depending on differences in skull shape and hair style. In the current study, there was also evidence that the constraining forces of the dummy necks altered both the head and helmet motion in ways that reduced the differences between concussed and uninjured players.

Another important and often unrecognized limitation of the laboratory reconstruction method relates to experimental design. Conducting physical tests in a laboratory is time-consuming and labor-intensive. Because injuries are rare events, even in contact sports, an unbiased cohort study design is not feasible in laboratory reconstruction studies. Researchers try to maximize their limited resources by selecting injurious impacts for analysis either exclusively or at a disproportionately high rate.^{5,6,23,25,26} The resulting data structures reflect either a case series or case-control study design. A significant drawback of the reconstruction method is that selection bias generally makes it impossible to draw inferences about absolute concussion risk from case-control data.⁴ Although many researchers have attempted to determine absolute concussion risk as a function of head acceleration or some other biomechanical parameter using reconstruction data,^{3,15,16,18,19,21,24,25,31,33} the results of those particular analyses should be discounted.

In summary, the current study utilized a primarily case-control study design with enhanced video analysis and laboratory reconstruction methods to generate data that will help elucidate which biomechanical parameters best distinguish concussed players from uninjured players in equivalent-force helmet impacts. The accuracy, repeatability, and sensitivity of the reconstruction results were quantitatively evaluated.

Through judicious case selection, model-based image matching, and iterative laboratory testing with limbless helmeted dummies, we were able to reconstruct the pre- and post-impact helmet motion of both players with good repeatability and an average absolute error in the helmet velocity changes of 14%. To place this error in context, the effect of concussion on average helmet velocity change (determined from video analysis of game footage) was 36%. The finding that the reconstruction process introduced average absolute errors that were over one third as high as the effect size of concussion is concerning, especially since the reconstruction errors are compounded by video analysis errors. It is likely that other, less rigorous reconstruction methods introduce errors that are considerably higher. There are many technical difficulties in matching the response of a crash test dummy to a football player. Crash tests dummies suitable for reconstructing concussive sports impacts have effectively one size (50th percentile male) and posture, although the addition of a ball joint to the base of the neck in this study expanded the range of possible neck positions. Football players, on the other hand, have a wide range of body sizes, head shapes, and hair styles, as well as widely varying neck postures, limb positions, and muscle forces acting at the time of impact. Given the inherent limitations and relatively high errors associated with laboratory reconstruction methods, wearable sensors and other complementary methods for studying sports concussion biomechanics should also be pursued.

ELECTRONIC SUPPLEMENTARY MATERIAL

The online version of this article (<https://doi.org/10.1007/s10439-020-02632-8>) contains supplementary material, which is available to authorized users.

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