# Concussion in Professional Football: Biomechanics of the Striking Player—Part 8

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**OBJECTIVE:** Concussive impacts in professional football were simulated in laboratory tests to determine the collision mechanics resulting in injury to the struck player and the biomechanics of the striking players, who were not concussed or neck-injured in the tackle.

**METHODS:** Twenty-seven helmet-to-helmet collisions were reconstructed in laboratory tests using Hybrid III dummies. The head impact velocity, direction, and kinematics matched game video. Translational and rotational head accelerations and six-axis upper neck loads and moments were used to evaluate how the striking player delivered the concussive blow. The neck injury criterion, *Nij*, was calculated to assess neck injury risks in the striking player.

**RESULTS:** The time-averaged impact force reached 6372 ± 2486 N at 7.2 milliseconds because of 46.8 ± 21.7$g$ head acceleration and 3624 ± 1729 N neck compression force in the striking player. Fifty-seven percent of the load was contributed by neck compression. The striking players had their heads down and lined up the impact axis through their necks and torsos. This allowed momentum transfer with minimal neck bending and increased the effective mass of the striking player to 1.67 times that of the struck player at peak load. The impact caused 94.3 ± 27.5$g$ head acceleration in the concussed players and 67.9 ± 14.5$g$ without concussion ($t = 2.06$, $df = 25$, $P = 0.025$). The striking player's *Nij* was greater than tolerance in 9 of 27 cases by exceeding the 4000 N neck compression limit. For these cases, the average neck compression force was 6631 ± 977 N (range, 5210–8194 N). *Nij* was 1.25 ± 0.16 for eight cases above the tolerance $Nij = 1.0$.

**CONCLUSION:** In the NFL, striking players line up their heads, necks, and torsos to deliver maximum force to the other player in helmet-to-helmet impacts. The concussive force is from acceleration of the striking player's head and torso load through the neck. Even though neck responses exceeded tolerances, no striking player experienced neck injury or concussion. A head-up stance at impact would reduce the torso inertial load in the collision and the risk of concussion in the struck player.

**KEY WORDS:** Biomechanics, Concussion, Football, Impact tolerances, Neck injury, Spearing, Sport injury prevention

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Since the 1970s, there has been concern for head-down tackling, or spearing, which can result in catastrophic neck injuries in the striking player. This concern is generally related to head impacts into the tackled player's torso, in which the mass of the struck player's body increases the load in the striking player's neck. Neck flexion and lateral bending increase injury risks. This type of tackle can lead to compression-flexion or other compression-bending injuries in the striking player, with quadriplegia and death the most serious consequences (1, 7, 21, 25, 26, 49, 52).

The incidence of catastrophic neck injuries has been tracked for more than 30 years in a national registry of cervical spine injuries in football and other databases (4, 13, 20, 47, 51–55). These epidemiological and cinematographic analyses of neck injuries have shown that the majority of cervical fracture-dislocations are caused by axial loading. This has resulted in rule changes in high school, college, and professional football banning deliberate spearing and the use of the top of the helmet as the initial point of contact in a tackle (11, 12, 27, 28). The rule changes significantly reduced the incidence of cervical spine injuries by the late 1970s, with a continued decline until the present (14).

The injury statistics have also led to the use of isometric and resistance exercises to develop strong neck musculature and reduce injury risks. These exercises are part of preseason conditioning to prevent catastrophic head and neck injury. With stronger necks, more impact force can be delivered in tackles without injury; however, the greater tolerance of the striking player can have negative consequences in those struck. Furthermore, training in proper blocking and tackling techniques is given to reduce head-down spearing (9, 21).

Biomechanical studies have been conducted to assess neck loads causing fracture-dislocations during head-down impacts in tackling-dummy practice (24) and game collisions (3, 15, 16, 48). Neck compression forces greater than 4000 N are considered sufficient to seriously injure a player because of axial compression of the cervical spine. These and other studies have established injury tolerance criteria for neck loading that are used with test dummies to study injury risks in sports, automotive crashes, and other impacts (5, 18, 22, 43, 50).

This study is part of a larger series on concussion in professional football. The National Football League (NFL) has a Mild Traumatic Brain Injury (MTBI) Committee, which has undertaken research aimed at defining the biomechanics of concussive impacts in professional football (37). One aspect of the effort focused on the analysis of multiple views of concussive impacts from game video to determine the speed of impact. Laboratory reconstructions of the collisions were performed using instrumented test dummies to simulate the helmeted players.

The laboratory re-enactments closely matched the field situation. With transducers in the dummy, the translational and rotational accelerations of the head and neck loads in the striking player allowed an evaluation of biomechanical responses during concussive impacts. This article evaluates the impact biomechanics of the striking players who are not concussed or neck-injured in the collisions; it also describes the collision biomechanics resulting in concussion of the struck player.

This study points out a new concern with head-down tackling, which is concussion of the struck player. This type of tackle can lead to injury in the struck player with little risk of head or neck injury in the tackler, because only the head and neck of the struck player initially resist the impact, and the striking player lines up his head, neck, and torso. This study addresses the biomechanics of the striking and struck players with head-down tackling in NFL helmet-to-helmet collisions causing concussion.

## MATERIALS AND METHODS

### Video Analysis of NFL Game Impacts

Details of the game film selection and analysis can be found in the studies by Pellman et al. (38, 39). For this study, a short overview of the laboratory methods is provided. When an MTBI occurred on the field during an NFL game, it was reported to Biokinetics and Associates, Ltd. (2470 Don Reid Drive, Ottawa, ON K1H 1E1, Canada), the engineering group contracted to analyze and reconstruct game impacts. Network tape of games was obtained from the NFL and subsequently analyzed. In addition to concussion impacts, other cases of significant head impact were selected for analysis. These were determined by NFL films. During the period 1996 to 2001, 182 cases were obtained on video for analysis. The initial analysis determined the impact location on the helmet and the contact source (helmet, ground, shoulder, etc.); 61% of the collisions involved helmet-to-helmet contacts (39).

Biokinetics determined the feasibility of determining the three-dimensional impact velocity, orientation, and helmet kinematics. At least two clear views were necessary to make this analysis. For those videos in which the three-dimensional impact velocity could be analyzed, a laboratory setup with crash dummies was made to re-enact the game impact. Helmets were placed on the dummies in the laboratory reconstructions, and the velocity and orientation of impact were simulated along with the subsequent helmet kinematics. A number of significant impacts were also reconstructed in which MTBI did not occur to study nonconcussion impacts. In total, 27 NFL helmet-to-helmet collisions were reconstructed; 22 involved concussion of the struck player, and 5 involved no injury. There was no injury to the striking players.

### Laboratory Reconstruction Techniques

*Figure 1* shows the reconstruction setup, which involved two Hybrid III male dummies (2). A helmeted head-neck assembly representing the struck player was attached to a 7.1

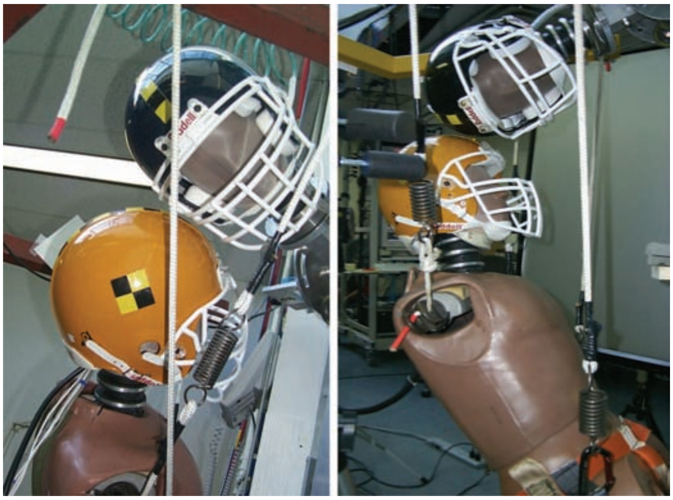

**FIGURE 1.** *Photographs showing reconstruction of game impacts in laboratory tests with instrumented dummies (*left*, Test 39, and* right*, Test 162). The torso and pelvis of the striking player were suspended from below, and the struck player was simulated with the head and neck attached to a 7.1 kg drop weight. Adjustments were made in the setup to duplicate the helmet kinematics in the game impact. The tests involved VSR-4 helmets by Riddell.*

kg mass simulating the struck player's torso and guided in free fall from a height to match the impact velocity determined from video analysis of the game collision. The Hybrid III head and neck weighed 4.38 kg with instrumentation. The helmet and face mask weighed 1.92 kg, and the falling mass was 15.1 kg. Impact was against another helmeted head-neck assembly attached to the torso and pelvis of the Hybrid III dummy. This dummy weighed 46.4 kg without arms and legs and was suspended by flexible cables.

Acceleration was measured in both dummy heads. The center of gravity (cg) of the head is a reference point, which is defined by its position in three orthogonal axes. The motion of the head cg is defined by three orthogonal components of velocity and acceleration. The acceleration is translational, even though the trajectory is curvilinear. As the head cg moves in space under translational acceleration, it can also rotate about the head cg. This involves rotational acceleration, and there are three orthogonal axes for rotational acceleration and velocity. When the head is assumed to be rigid, as in the dummy, the three axes of translational and rotational acceleration define the motion sequence of the head during impact. The sign convention used in this study has neck compression as $-Fz$ and neck tension as $+Fz$, because the positive $z$ axis is from the neck upward through the top of the head (46). The positive $x$ axis is forward, and the positive $y$ axis is through the left ear. Neck extension is $-My$, and flexion is $+My$.

Each head form was equipped with standard accelerometers at the head cg and nine linear accelerometers set up in a so-called "3-2-2-2 configuration" to determine rotational acceleration (36). The analysis is valid for accelerometers coincident with the origin of head cg or coincident with one of the axes. Deviations from this were required in the head-form configuration used in these tests, and a correction for centripetal and Coriolis acceleration was made (6).

The dummy representing the striking player had a six-axis neck transducer installed between the head and the top of the neck. The transducer measured three axes of neck force ($Fz$, compression-tension; $Fx$, fore-and-aft shear; and $Fy$, left-to-right shear) and three axes of neck moment ($My$, flexion-extension; $Mx$, lateral bending; and $Mz$, rotation about the $z$ axis).

High-speed video recorded head kinematics in the reconstruction. The camera was positioned similarly to one of the views from the game video. This allowed a one-to-one comparison of the game and reconstruction kinematics and facilitated fine adjustments in the impact orientation and alignment of the laboratory impacts to closely match the helmet kinematics in the game (39).

Extensive testing was conducted to isolate and quantify sources of error and variability in the reconstructions (30). This work showed the reconstructions to be repeatable and with minimal error for this type of testing. In the laboratory reconstructions, every effort was made to reduce potential sources of error.

## Game Impacts

*Figure 2* shows the initial helmet contact points determined from game video. The *top* shows the location for the struck players, who were either concussed or not concussed, and the *bottom* shows the striking players, none of whom were injured in the collisions. The impacts are shown on the right side of the helmet, although contacts occurred on both sides. More than half of the impacts involved the face mask or area in which the face mask attaches to the helmet shell of the struck player, whereas virtually all striking players (26 of 27 cases) involved the front crown or top portion of the helmet, because their head was down and the axis of impact was through their head


**FIGURE 2.** *Photographs showing location of initial helmet contacts for the struck players* (top, *both concussive and nonconcussive impacts) and striking players (*bottom, *none of whom were concussed). All of the strikes in the plot are shown on the right side of the helmet to visualize the impacts, although the game impacts occurred on both sides of the helmet.* MTBI (H-H) *indicates a concussed player involved in a helmet-to-helmet impact (modified* from, *Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF: Concussion in professional football: Reconstruction of game impacts and injuries.* ***Neurosurgery*** *53:799–814, 2003 [39] by removing the helmet-to-ground impacts).*

cg, neck, and torso. Deviations from this alignment in game impacts and laboratory tests caused the helmets to slide off because of the smooth plastic shell of the helmets. This dramatically lowered the impact responses.

## Collision Biomechanics

Impact force ($F$) from the striking player was determined by adding the head inertia force of the striking player and neck compression force:

$$F = m_{\text{Striking}} a_{\text{Striking}} + F_{\text{N}} \quad (1)$$

where $a_{\text{Striking}}$ is the resultant acceleration of the striking player's head, $F_{\text{N}}$ is the resultant neck compression force, and $m_{\text{Striking}}$ is the mass of the striking player's head above the neck load cell. The mass was $m_{\text{Striking}} = 5.90$ kg and included the Hybrid III head (3.64 kg), the load cell above the sensing element (0.34 kg), and the helmet with face mask (1.92 kg). Mass below the sensing element is not included in Equation 1, because the striking player's neck load was measured. In the collision, the striking player used the top or crown portion of the helmet. This area is substantially stiffer than the side or face mask region of the helmet.

The impact force is equilibrated by the struck player. His head, neck, helmet, and a portion of the torso are involved:

$$F = m_{\text{Struck}} a_{\text{Struck}} \quad (2)$$

where $a_{\text{Struck}}$ is the resultant acceleration of the struck player's head. The mass of the struck player is $m_{\text{Struck}} = 8.40$ kg and includes the head (4.38 kg), neck (1.06 kg), helmet and face mask (1.92 kg), and a portion of the torso mass (1.04 kg). The difference in the Hybrid III head mass between the striking and struck players reflects the full weight of a bracket that is used in place of the neck load cell.

Head acceleration of the striking player is lower than that of the struck player, so the effective mass of the striking player is greater than that of the struck player. The neck load cell measures the contribution from the torso mass in the collision, which adds to the impact force. The effective mass of the striking player is

$$m_{\text{Eff.Striking}} = F / a_{\text{Striking}} \quad (3)$$

On the basis of the average head acceleration and impact force, $m_{\text{Eff.Striking}} = 14.0$ kg, indicating a mass ratio of $m_{\text{Eff.Striking}} / m_{\text{Struck}} = 1.67$, or a 67% greater effective mass of the striking player than that of the struck player during peak force. The mass ratio equals the ratio of head accelerations:

$$a_{\text{Struck}} / a_{\text{Striking}} = m_{\text{Eff.Striking}} / m_{\text{Struck}} \quad (4)$$

where the relationship assumes that a single mass is involved in the head impact for each player. The impact force and other biomechanical responses, including head accelerations and changes in velocity ($\Delta V$), describe the collision mechanics leading to concussion in the struck player.

## Head Injury Tolerances

The primary response of the head is the resultant translational acceleration of the head cg. This was determined from three orthogonal accelerations measured in the dummy. Although translational acceleration is measured in units of m/s$^2$, it is reported in units of gravity ($g$), where the measured acceleration is normalized by the acceleration of gravity ($1g = 9.8$ m/s$^2$). Integration of the resultant acceleration gave the change in head velocity, or $\Delta V$, during impact.

For the head impacts, the resultant acceleration is used to calculate two head injury criteria. The National Operating Committee on Standards for Athletic Equipment (29) football helmet standard uses the severity index (SI), which is determined by the method of Gadd (10):

$$SI = \int^{T} a(t)^{2.5} dt \quad (5)$$

where $a(t)$ is the resultant translational acceleration at the head cg and $T$ is the duration of the acceleration. The National Highway Traffic Safety Administration (NHTSA) uses a variation of SI to assess head injury risks in car crashes. The head injury criterion (HIC) is determined by

$$HIC = \{(t_2 - t_1)[\int_{t_1}^{t_2} a(t)dt/(t_2 - t_1)]^{2.5}\}_{\max} \quad (6)$$

where $t_1$ and $t_2$ are determined to maximize the HIC function and $a(t)$ is the resultant translational acceleration of the head cg. In practice, a maximum limit of $T = t_2 - t_1 = 15$ milliseconds is used.

The second type of biomechanical response of the head involves rotational acceleration and rotational velocity. Many researchers have speculated that rotational acceleration is a key response associated with head injury (35).

## Neck Injury Tolerances

The early neck tolerance for axial compression was estimated by using a Hybrid III dummy to measure neck loads when struck by a tackling block that had produced serious head and neck injuries in football players (24). The compression tolerance varied with load duration but was estimated to be 4000 N. Neck tension and shear load tolerances were estimated using the Hybrid III dummy in reconstructions of three-point belted occupant injuries in frontal car crashes (34). The limits for tension and shear force were 3300 and 3000 N, respectively.

Tolerance levels for neck flexion and extension were estimated by use of sled tests of volunteers and cadavers (23, 45). Volunteer tests provided data up to the pain threshold for extension bending moments, and cadaver tests estimated tolerance limits for serious injuries at 57 Nm. The maximum voluntary flexion moment of 190 Nm was set as the tolerance limit. The bending moments were based on human responses, rather than dummy measurements. Cadaver tests on neck tension showed failure at 3373 N (56); however, lower forces were found for combined loading conditions of tension-extension (44).

Kleinberger et al. (19), working with the NHTSA, reviewed earlier studies and developed a neck injury criterion, *Nij*, in which the "*ij*" indices represent four injury mechanisms: tension-extension (*Nte*), tension-flexion (*Ntf*), compression-extension (*Nce*), and compression-flexion (*Ncf*). The criterion emphasizes injury risks for sagittal plane motion. Crash tests using a six-axis upper neck load cell in the Hybrid III dummy in frontal crashes established limits for flexion-extension bending (*My*) and tension-compression force (*Fz*). Shear load (*Fx*) was used to calculate the effective moment at the occipital condyles by multiplying the shear load by the height of the load cell above the condyles and subtracting this value from the measured *My*.

*Nij* is calculated as a function of time by normalizing *My* and *Fz* with intercept tolerances for extension, flexion, tension, and compression. The normalized flexion-extension moments are added to the normalized axial loads to give *Nij*:

$$Nij = (Fz/Fzc) + (My/Myc) \qquad (7)$$

where *Fzc* is the critical intercept for axial neck loading and *Myc* the critical intercept for flexion-extension bending moment at the occipital condyles. The critical intercepts are $Fzc =$ 6806 N for tension, $Fzc = 6160$ N for compression, $Myc = 310$ Nm for flexion, and $Myc = 135$ Nm for extension. The neck extension intercept is substantially higher than the earlier 57 Nm estimate by Mertz and Patrick (23).

During an impact, all four combinations of neck response need to be below $Nij = 1.0$. In addition, peak neck tension cannot exceed 4170 N and compression 4000 N. The *Nij* criterion is consistent with information from experimental and laboratory studies (17, 31–33, 40–42).

### Establishing Time Zero to Align the Data

The following procedure was used to align time zero for the individual cases, because the orientations of the collisions and timing varied between tests. A "soft trigger" was used to determine the start of head acceleration. For most cases, a 1*g* trigger was used to determine the start of the impact; however, some tests had noise on the responses requiring a 3*g* (Tests 7, 38, 39, 48, 59, 69, 84, and 92) or 5*g* trigger (Test 77). The responses presented here were based on the time-zero adjustments to align the impact responses by assuming that time zero occurs when the head acceleration surpasses the soft trigger.

### Statistical Analyses

The significance of differences in responses for the striking player causing concussion and no concussion were determined using *t* tests assuming unequal variance and a single-sided tail distribution. If Levine's test suggested inequality of variances, a *t* test with adjustment for unequal variances was used. The *t* tests were performed using SPSS 11.5 for Windows (SPSS, Inc., Chicago, IL). The regression analysis was also used from Excel (Microsoft, Seattle, WA), which determined the average and 95% confidence interval for a linear fit between response data.

## RESULTS

### Biomechanics of the Striking Player

*Table 1* shows the peak responses for the 27 NFL helmet-to-helmet collisions reconstructed in laboratory tests. The average impact speed for these collisions was 9.3 ± 1.9 m/s. The peak resultant head acceleration for the striking player was 56.1 ± 22.1*g*, and impact resulted in a 4.1 ± 1.2 m/s change in head velocity. HIC was 117 ± 101 in the striking player. More momentum was transferred to the struck player than the striking player in the collision, because the $\Delta V$ of the struck player was 6.8 ± 1.8 m/s. The mass ratio based on $\Delta V$ was $m_{\text{Eff.Striking}}/m_{\text{Struck}} = 1.67$ (6.8/4.1). The average peak impact force was 7191 ± 2352 N. The calculated impact force and neck responses for the striking player, including *Nij*, are new information, whereas the head accelerations of the players and collision speeds are re-reported from Pellman et al. (39) and shown in *Table 2* for the struck players. This study excludes the helmet-to-ground impacts.

The NFL game impacts involved the striking player hitting either the right or the left side of the opponent's helmet. A notation is included in *Table 1* about the side of helmet impact. This is one factor in the direction of neck forces and moments. Head kinematics was complex, and variations in the direction of neck responses occurred during impact. The primary neck load was axial compression (*Fz*) in the striking players. The average peak neck compression force was 4227 ± 1888 N in the striking player. *Table 1* also includes the average and standard deviation in positive and negative neck responses. The average positive fore-and-aft neck shear (*Fx*) was 767 ± 327 N, and lateral shear (*Fy*) was 504 ± 217 N. The average peak neck bending moment (*My*) was 47.2 ± 38.7 Nm in flexion and 35.7 ± 20.5 Nm in extension.

The average *Nij* was 0.79 ± 0.33. However, eight of the tests involved *Nij* greater than 1.0, which is the NHTSA human tolerance level for neck loading. For these tests, *Nij* was 1.25 ± 0.16. Also shown is the type of *Nij* associated with the peak value. All cases involved neck compression, but there were 23 cases of compression-flexion (*Ncf*) and 4 cases of compression-extension (*Nce*). Nine cases exceeded the neck compression force of 4000 N and averaged 6631 ± 977 N (range, 5210–8194 N).

### Head-Neck Impact Kinematics

*Figure 3* shows the kinematic sequence of the helmet impacts from two reconstructions (Cases 38 and 39) out of the 27 cases in this study. Both struck players were concussed in NFL games. The sequence from the high-speed video progresses from the top down and shows the striking player on the *left* and the struck player on the *right*. The laboratory tests re-

**TABLE 1. Peak responses from the laboratory reconstruction of the striking player in National Football League collisions[a]**

| Case no. | Side (L or R) | Impact velocity (m/s) | SI | HIC | Head responses: Transl. accel (*g*) | ΔV (m/s) | Rotat. accel (rad/s²) | Rotation velocity (r/s) | Neck response: *Fx* (N) | *Fy* (N) | *Fz* (N) | *Mx* (Nm) | *My* (Nm) | *Mz* (Nm) | *Nij*: Value | Type | Impact force (N) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| 7 | L | 6.9 | 65 | 51 | 50 | 2.2 | 2832 | 9.8 | −750 | 285 | −3822 | 12 | 42 | 2 | 0.73 | *Ncf* | 6030 |
| 9 | R | 10.3 | 275 | 217 | 79 | 5.2 | 6719 | 18.7 | 643 | 703 | −7657 | 53 | −66 | 13 | 1.64 | *Nce* | 11680 |
| 38 | L | 9.5 | 157 | 127 | 60 | 4.0 | 5205 | 28.2 | 662 | 270 | −6406 | −18 | −38 | −8 | 1.09 | *Nce* | 9776 |
| 39 | R | 10.9 | 60 | 43 | 44 | 2.3 | 4487 | 10.4 | −707 | −275 | −6660 | 54 | 50 | −9 | 1.21 | *Ncf* | 7889 |
| 48 | R | 9.7 | 44 | 37 | 32 | 3.2 | 2939 | 28.0 | −690 | 347 | −2327 | −18 | 29 | 4 | 0.46 | *Ncf* | 4108 |
| 57 | R | 8.8 | 48 | 38 | 32 | 4.1 | 4151 | 33.2 | −521 | 444 | −3558 | −50 | 33 | 14 | 0.67 | *Ncf* | 5333 |
| 59 | L | 5.3 | 32 | 26 | 32 | 2.3 | 2087 | 13.1 | −741 | −70 | −3784 | 8 | 33 | −3 | 0.72 | *Ncf* | 4913 |
| 69 | R | 10.3 | 55 | 50 | 38 | 3.1 | 2620 | 23.0 | 347 | 419 | −3243 | −14 | −20 | −5 | 0.56 | *Ncf* | 4796 |
| 71 | R | 10.3 | 512 | 434 | 102 | 6.6 | 5541 | 32.4 | −1212 | 755 | −2079 | 43 | 80 | 17 | 0.53 | *Ncf* | 8258 |
| 77 | R | 9.9 | 65 | 53 | 35 | 4.2 | 2714 | 25.5 | −1027 | 440 | −3164 | −9 | 58 | 12 | 0.68 | *Ncf* | 5612 |
| 84 | R | 9.4 | 96 | 78 | 45 | 4.4 | 3169 | 26.5 | −663 | 520 | −3875 | −19 | 25 | 4 | 0.71 | *Ncf* | 6431 |
| 92 | R | 11.1 | 204 | 164 | 60 | 5.6 | 6070 | 43.8 | −392 | 889 | −8194 | −53 | 45 | 5 | 1.45 | *Ncf* | 11510 |
| 98 | L | 9.6 | 241 | 187 | 84 | 4.8 | 4487 | 38.5 | 140 | −851 | −3633 | 50 | 14 | −6 | 0.63 | *Ncf* | 7953 |
| 113 | R | 7.0 | 101 | 75 | 61 | 3.7 | 3700 | 31.2 | 152 | 575 | −3343 | 42 | 5 | −4 | 0.56 | *Nce* | 6323 |
| 118 | R | 10.7 | 122 | 73 | 56 | 3.7 | 3687 | 23.4 | 1351 | −303 | −6269 | −10 | −59 | 7 | 1.23 | *Nce* | 8937 |
| 124 | R | 11.4 | 105 | 73 | 56 | 3.1 | 4086 | 16.1 | −651 | −306 | −5056 | −22 | 47 | 7 | 0.84 | *Ncf* | 7959 |
| 125 | R | 11.7 | 132 | 111 | 47 | 4.2 | 3366 | 28.1 | −584 | 746 | −6391 | −42 | 49 | −10 | 1.06 | *Ncf* | 9015 |
| 135 | L | 10.0 | 230 | 179 | 81 | 3.8 | 5005 | 29.3 | 1244 | −463 | −7395 | −53 | −97 | −16 | 1.72 | *Nce* | 11490 |
| 148 | R | 6.6 | 47 | 37 | 33 | 3.9 | 2466 | 26.5 | 470 | −465 | −2193 | 21 | −24 | −6 | 0.53 | *Nce* | 4065 |
| 154 | R | 6.6 | 35 | 31 | 29 | 3.1 | 3159 | 23.1 | 767 | −169 | −2099 | 12 | −37 | −6 | 0.60 | *Nce* | 3774 |
| 155 | R | 9.1 | 76 | 61 | 45 | 4.2 | 4217 | 29.5 | 943 | −396 | −3591 | 30 | −41 | −11 | 0.87 | *Nce* | 6247 |
| 157 | R | 10.8 | 215 | 180 | 79 | 5.0 | 4662 | 15.7 | 1317 | −286 | −5494 | −28 | −117 | −4 | 1.53 | *Nce* | 9568 |
| 162 | R | 5.5 | 34 | 30 | 29 | 3.2 | 1672 | 17.2 | 645 | −242 | −2894 | −17 | −40 | −2 | 0.74 | *Nce* | 4505 |
| 164 | R | 10.8 | 243 | 202 | 89 | 5.1 | 6136 | 30.8 | 373 | −682 | −2629 | 38 | −25 | −12 | 0.61 | *Nce* | 7872 |
| 175 | R | 9.6 | 81 | 62 | 47 | 3.9 | 2535 | 19.3 | 501 | −819 | −2518 | −50 | −35 | −12 | 0.61 | *Nce* | 5011 |
| 181 | L | 11.7 | 402 | 333 | 85 | 7.3 | 6613 | 55.8 | −632 | 905 | −2312 | −56 | 25 | −8 | 0.38 | *Ncf* | 6877 |
| 182 | R | 8.1 | 272 | 213 | 87 | 4.7 | 3206 | 27.2 | 977 | −717 | −3386 | −39 | −60 | −12 | 0.96 | *Nce* | 8239 |
| Average | | 9.3 | 146 | 117 | 56.1 | 4.1 | 3983 | 26.1 | 702 | 561 | | 33.0 | 38.2 | 8.5 | 0.86 | | 7191 |
| SD | | 1.9 | 121 | 101 | 22.1 | 1.2 | 1402 | 10.0 | 395 | 219 | | 17.3 | 19.0 | 5.1 | 0.38 | | 2352 |
| | | | | | | | | −Average | −714 | −432 | −4221 | −31.1 | −50.7 | −7.9 | | | |
| | | | | | | | | SD | 227 | 246 | 1885 | 17.3 | 28.9 | 3.9 | | | |

[a] L, left; R, right; SI, severity index; HIC, head injury criterion; Transl., translational; accel, acceleration; Rotat., rotational; *F*, neck force; *M*, neck movement; *x*, *y*, *z*, axes; *Nij*, neck injury criterion; SD, standard deviation; *Ncf*, compression-flexion; *Nce*, compression-extension.

**TABLE 2. Peak responses from the laboratory reconstruction of the struck player in National Football League collisions[a]**

| Case no. | MTBI | Velocity (m/s) | SI | HIC | Peak transl. accel (*g*) | Peak $\Delta V$ (m/s) | Peak rotat. accel (rad/s$^2$) | Peak rotation velocity (r/s) |
|---|---|---|---|---|---|---|---|---|
| 7 | Yes | 6.9 | 120 | 93 | 61 | 4.6 | 6266 | 28.1 |
| 9 | Yes | 10.3 | 848 | 600 | 134 | 10.1 | 7428 | 27.4 |
| 38 | Yes | 9.5 | 736 | 554 | 118 | 9.7 | 9678 | 50.8 |
| 39 | Yes | 10.9 | 656 | 522 | 129 | 8.4 | 5921 | 36.1 |
| 48 | No | 9.7 | 155 | 130 | 57 | 4.7 | 5617 | 42.4 |
| 57 | Yes | 8.8 | 253 | 206 | 77 | 6.0 | 6514 | 37.0 |
| 59 | No | 5.3 | 205 | 138 | 82 | 5.6 | 5387 | 26.9 |
| 69 | Yes | 10.3 | 177 | 153 | 61 | 5.0 | 4381 | 19.9 |
| 71 | Yes | 10.3 | 658 | 510 | 123 | 7.3 | 5400 | 35.0 |
| 77 | Yes | 9.9 | 226 | 185 | 80 | 5.2 | 5148 | 36.4 |
| 84 | Yes | 9.4 | 276 | 222 | 82 | 6.3 | 9193 | 80.9 |
| 92 | Yes | 11.1 | 630 | 508 | 107 | 10.0 | 6878 | 44.2 |
| 98 | Yes | 9.6 | 351 | 301 | 91 | 6.2 | 7548 | 43.4 |
| 113 | Yes | 7.0 | 163 | 140 | 59 | 5.1 | 3965 | 12.8 |
| 118 | Yes | 10.7 | 492 | 378 | 101 | 9.6 | 7017 | 42.9 |
| 124 | Yes | 11.4 | 380 | 282 | 81 | 7.5 | 7138 | 34.8 |
| 125 | Yes | 11.7 | 817 | 633 | 113 | 9.1 | 7716 | 63.3 |
| 135 | Yes | 10.0 | 751 | 566 | 138 | 8.6 | 7540 | 41.0 |
| 148 | Yes | 6.6 | 117 | 99 | 48 | 5.1 | 3476 | 23.9 |
| 154 | No | 6.6 | 136 | 114 | 53 | 5.1 | 4167 | 24.0 |
| 155 | Yes | 9.1 | 418 | 341 | 100 | 6.6 | 6940 | 37.0 |
| 157 | Yes | 10.8 | 545 | 472 | 103 | 8.1 | 6750 | 33.5 |
| 162 | Yes | 5.5 | 94 | 77 | 52 | 4.2 | 2615 | 18.4 |
| 164 | Yes | 10.8 | 451 | 370 | 124 | 6.0 | 9590 | 26.6 |
| 175 | No | 9.6 | 158 | 125 | 62 | 5.6 | 3555 | 39.2 |
| 181 | Yes | 11.7 | 423 | 382 | 93 | 7.1 | 8011 | 36.5 |
| 182 | No | 8.1 | 256 | 208 | 85 | 5.9 | 5512 | 17.8 |
| | Average | 9.3 | 389 | 308 | 89.4 | 6.8 | 6272 | 35.6 |
| | SD | 1.9 | 240 | 182 | 27.5 | 1.8 | 1851 | 14.2 |

[a] MTBI, mild traumatic brain injury; SI, severity index; HIC, head injury criterion; transl., translational; accel, acceleration; rotat., rotational; SD, standard deviation.

versed the collision by giving all of the impact velocity to the struck player. The collision was reconstructed by dropping the head, neck, and torso mass of the struck player into the suspended Hybrid III dummy simulating the striking player.

A *white line* has been added to the photographic sequence to help visualize the motion and compression of the striking player's head-helmet and neck in the impact and eventual neck bending as the helmets separate. In the top three photographs of the sequence, the striking player's head and helmet are pushed to the left by compressing the head-helmet interface and then the neck. The last three photos in each sequence show neck bending and eventual release after impact.

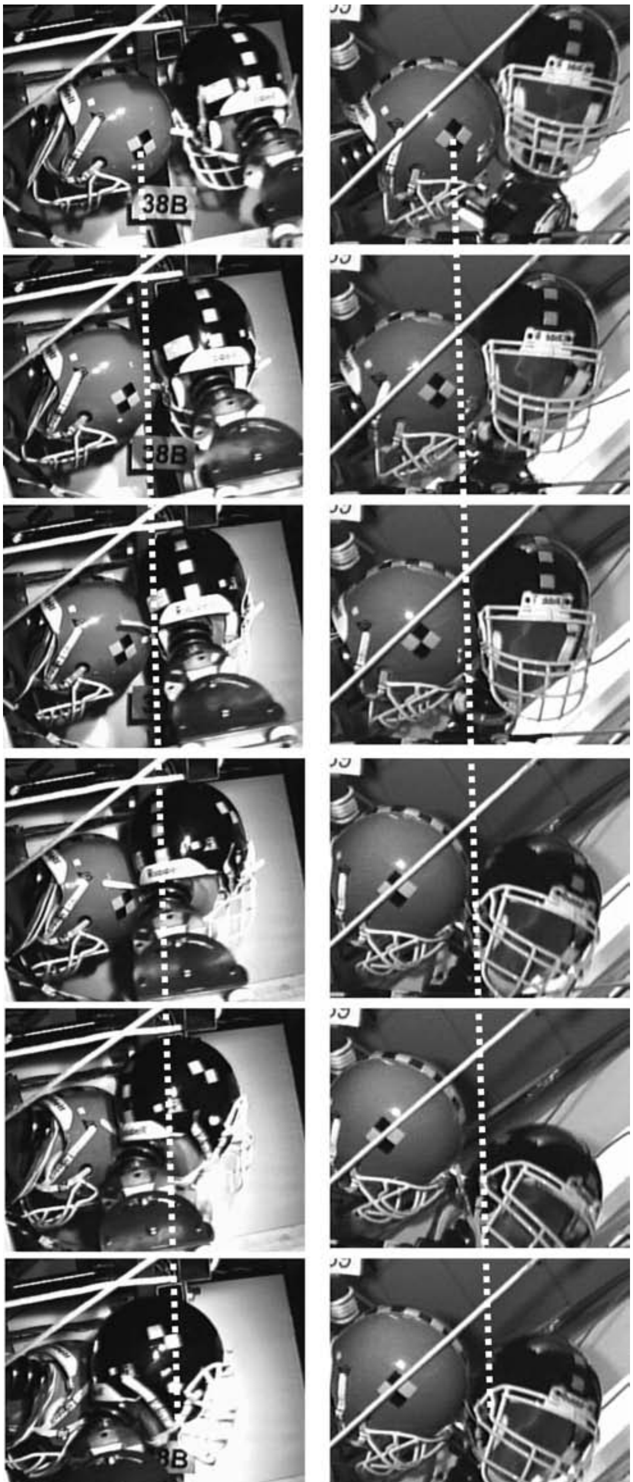


**FIGURE 3.** *Sequence from high-speed video of laboratory reconstruction of NFL concussion cases (*left*, Case 38, and* right*, Case 39). The sequence is from* top *to* bottom*, with the striking player on the* left *and struck player on the* right*. The* vertical white line *helps visualize the movement of the helmet and neck of the striking player.*

*Figure 4* shows the time-average head acceleration and impact force of the striking players for the 27 NFL collisions reconstructed in laboratory tests with the Hybrid III dummies. The peak values in these plots are lower than the values in *Table 1* because the average response was determined as a function of time. Because peak values occur at different times for each reconstruction, there are lower values in the time histories, which smooth the peaks. At impact, there was an increase the striking player's head acceleration, which reached an average of 46.8 ± 21.7*g* at 7.2 milliseconds. The struck player's head acceleration reached 76.9 ± 26.2*g*. The peak impact force was 6372 ± 2486 N acting on the struck player. The neck compression force was 3624 ± 1729 N and contributed 57% of the impact load. The peak head rotational acceleration was 4289 ± 2156 $r/s^2$. Double integration of the struck player's head acceleration indicated a 48 mm displacement at peak force.

The impact biomechanics was consistent for the cases reconstructed from the NFL. The biomechanical responses of the striking and struck players demonstrated a 67% higher effective mass of the striking player than the struck player at peak force. There were also high neck compression forces in the striking player, causing concussion. Neck shear forces were considerably lower in amplitude than the neck compression force, and the bending moments were moderate and primarily flexion and lateral bending. This is consistent with the alignment of the impact axis through the head cg, neck, and torso of the striking player to effect a solid blow on the struck player.

## Head-Neck Impact Biomechanics

*Figure 5* shows the neck shear and compression responses, which occurred during the loading. In the reconstructions, the forces are either positive or negative, depending on the orientation of the heads at impact. The average and standard deviation in fore-and-aft and lateral neck shear are shown for the cases with positive values along with the identification of the particular tests making up the response. The neck shear forces were less than one-fifth the level of the neck compression force in the collisions. Also shown are the average and standard deviation in neck compression force for the nine highest responses. This gives an indication of the most severe impacts by the striking players and levels of neck loading tolerated by the striking players without head or neck injury.

*Figure 6* shows the neck bending responses, which peak somewhat later than the impact force, head accelerations, and neck shear forces. Neck bending can be seen occurring later in the impact sequences shown in *Figure 3*. Depending on the orientation of the heads at impact, the bending moments vary from the positive to the negative in value. The average and standard deviation in responses for the positive-moment cases are shown, along with the identification of the particular tests making up the responses. The *Mz* bending moment was much lower in amplitude.

*Figure 7* shows the neck compression force and flexion-extension bending moment at peak *Nij* for the reconstructions. Superimposed on the plot is the NHTSA neck tolerance criterion. *Nij* combines the normalized neck compression force and flexion-extension moment as a function of time to estimate the significance of neck loading for serious cervical injury. The *lines*

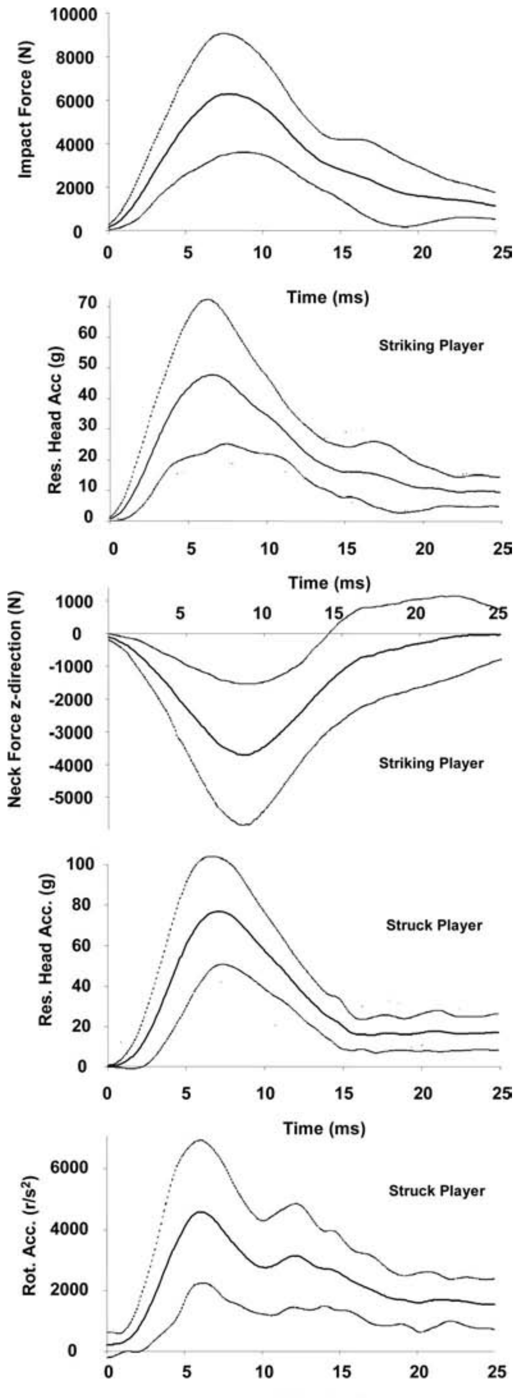


**FIGURE 4.** Top trace, *graph showing average and* ±1 *standard deviation in the impact force of the striking player. The lower traces show the resultant head acceleration* (second trace*) and the neck compression force (*third trace*) of the striking player (the sign of the neck compression force is reversed and added to the head inertial load to obtain the impact force in the* top trace*). The* bottom traces *are the resultant head translational and rotational accelerations of the struck player.*

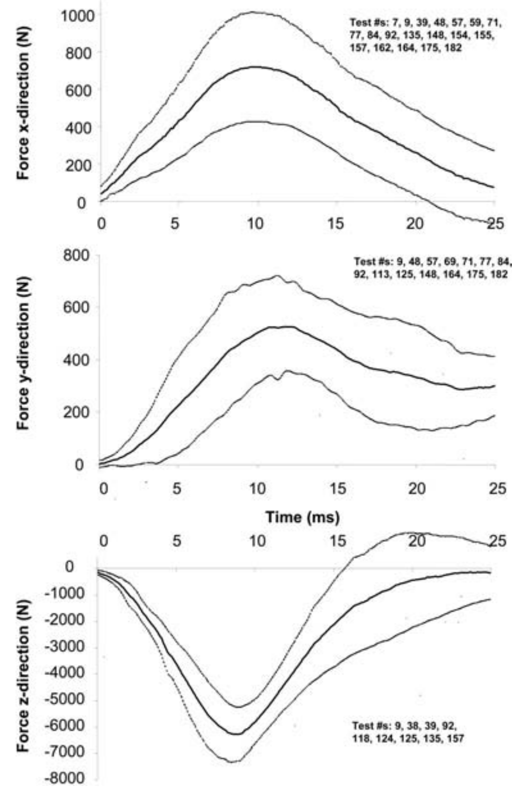


**FIGURE 5.** *Graphs showing average and* ±1 *standard deviation in the upper neck forces for the striking player. Because some impacts involved positive and others negative responses, depending on the impact orientation, the averages are shown for the group with positive neck* x *and* y *responses.*

in *Figure 7* show $Nij = 1.0$ and a 4000 N limit on peak neck compression force. These values are the current human tolerance limits for neck loading. The NFL reconstructions show nine cases outside the tolerable limits. In all nine cases, the neck compression force exceeded the tolerance limit of 4000 N, and in a few cases, it exceeded it by almost a factor of two. For these cases, the average neck compression force was 6631 ± 977 N (range, 5210–8194 N), and *Nij* was 1.25 ± 0.16 for those greater than 1.0. It is also interesting that there was a relatively moderate level of neck flexion-extension moment. The primary impact response was neck compression.

*Figure 8* shows the peak neck compression force versus the initial impact speed and the computed change in head velocity at peak *Nij*. These data give an impression of the speed of neck compression in the striking player in the NFL impacts. The average collision speed for the reconstructions was 9.3 ± 1.9 m/s, and the average head velocity change was 2.8 ± 0.9 m/s at peak *Nij*, which is 30% of the initial collision speed. Most of the impact velocity (73% = 6.8/9.3) was transferred to the struck player in the collision.

## Impact Conditions with Concussion

Five of the reconstructions involved hard hits in the game but no concussion to the struck player. The remaining 22 collisions

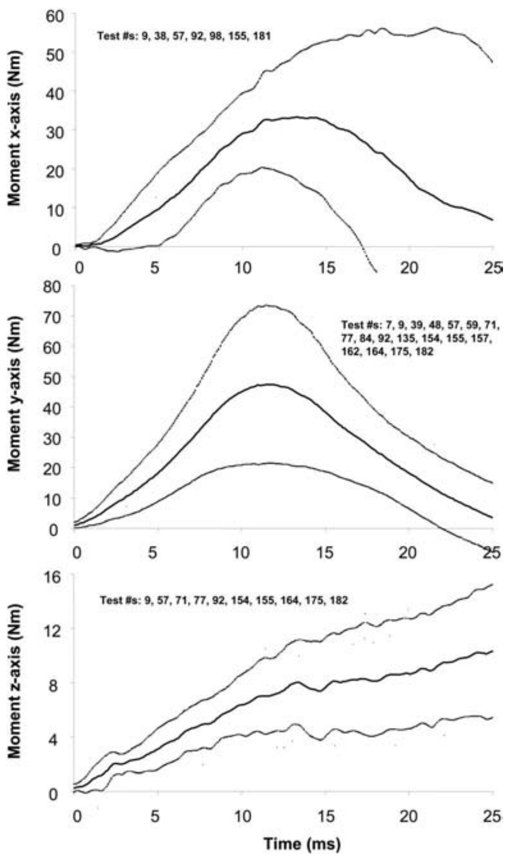


**FIGURE 6.** *Graphs showing average and ±1 standard deviation in the upper neck moments for the striking player. Because some impacts involved positive and others negative responses, depending on the impact orientation, the averages are shown for the group with positive responses.*

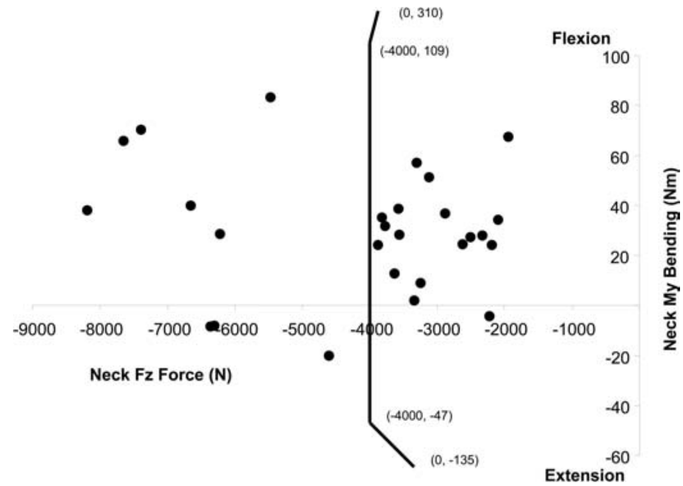


**FIGURE 7.** *Scatterplot of neck compression force and flexion-extension moment at the time of peak* Nij. *The* lines *show the tolerance criterion of* Nij *= 1.0 and limit of 4000 N on neck compression force.*

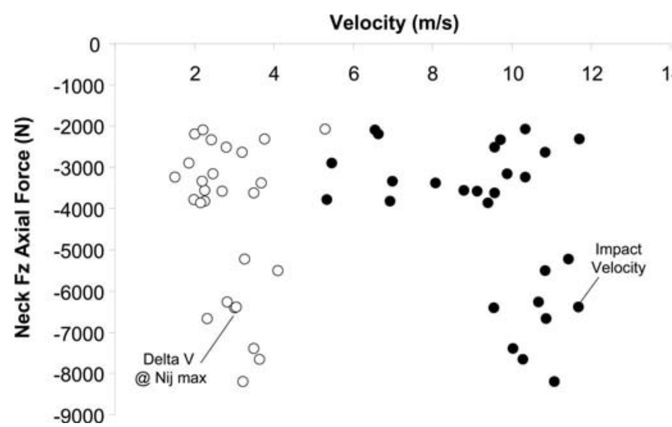


**FIGURE 8.** *Scatterplot of neck compression force for the 27 reconstructed collisions in the NFL showing the impact velocity of the helmets and the change in velocity of the striking player's head at peak* Nij.

involved concussions. When the data were analyzed for impact conditions that caused concussion, the impact biomechanics of the struck player causing concussion was higher than for the players without injury. Peak head acceleration was 94.3 ± 27.5$g$ with concussion and 67.9 ± 14.5$g$ without ($t$ = 2.06, $df$ = 25, $P$ = 0.025). The peak impact force averaged 7642 ± 2259 N with concussion and 5209 ± 1774 N without injury ($t$ = 2.24, $df$ = 25, $P$ = 0.017). The head $\Delta V$ was 7.08 ± 1.88 m/s with concussion and 5.38 ± 0.48 m/s without ($t$ = 3.75, $df$ = 24, $P$ = 0.0005). The striking player experienced a 4.26 ± 1.23 m/s head $\Delta V$ with concussion and 3.44 ± 0.90 m/s without injury ($t$ = 1.39, $df$ = 25, $P$ = 0.088). These differences are statistically significant. The average peak neck compression force in the striking player was 4539 ± 1931 N with concussion compared with 2823 ± 725 N without concussion ($t$ = 3.27, $df$ = 18, $P$ = 0.002).

## DISCUSSION

In these helmet-to-helmet impacts, the striking player lowers his head and lines up his head, neck, and torso to deliver maximum force to the struck player, whose head and neck resist the impact. This is the typical situation when the struck player does not see the tackle coming and does not prepare for the collision. *Figure 9* shows an example (Player 38) of this tackling technique in a helmet-to-helmet impact causing concussion in an NFL game. With greater inertia of the striking player behind the impact, the average peak acceleration of the struck player's head reached 94.3 ± 27.5$g$, causing concussion. This acceleration was significantly higher than the 67.9 ± 14.5$g$ in the nonconcussed players ($t$ = 2.06, $df$ = 25, $P$ = 0.025). The striking player had even lower peak head accelerations of 56.1 ± 22.1$g$ because of the added mass through neck compression. HIC and SI in the striking player were very low, so there was minimal risk of concussion in the striking player.

The key to the concussive blow is the head-down position, which involves a 67% greater mass of the striking player by coupling his torso into the collision. This kinematic transfers more momentum to the struck player. In the situation in which the struck players see the impending tackle, they have a chance to line up their bodies and prepare for the collision. In this case, they have a greater effective body mass and are better prepared to resist the momentum transfer, particularly if they can lean into the tackle and line up their body.

Even though there are high compressive forces in the struck player's neck in the collisions, no NFL player, to the best of our knowledge, has experienced serious neck injury or concussion in this type of tackle. In fact, in 9 of the 27 NFL collisions reconstructed in this study, the compressive neck force exceeded current tolerance criteria for serious neck injury. The avoidance of neck injury is primarily by maintaining an axial alignment of the impact-force vector through the neck and torso; by minimizing neck extension and lateral bending, which increase the risk of injury; and by engaging the helmet of the struck player. This delivers more momentum to the struck player.

In this study, neck compressive forces were recorded above the current tolerance for serious neck injury. Using the neck injury criterion of the NHTSA, the *Nij* averaged 1.25 ± 0.16 in eight cases above the tolerance limit of *Nij* = 1.0. The average neck compression force was 6631 ± 977 N (range, 5210–8194 N) for the nine cases above the tolerance limit of 4000 N neck compression force. The neck tolerance criterion represents a risk of serious injury in 30- to 35-year-old men. However, the NFL data are from a population of players with superior physical conditioning and training. Strengthening exercises for neck muscles give these players far greater tolerance to neck compression, and the players have an ability to maintain an axial alignment of their cervical spine during head-down tackle.

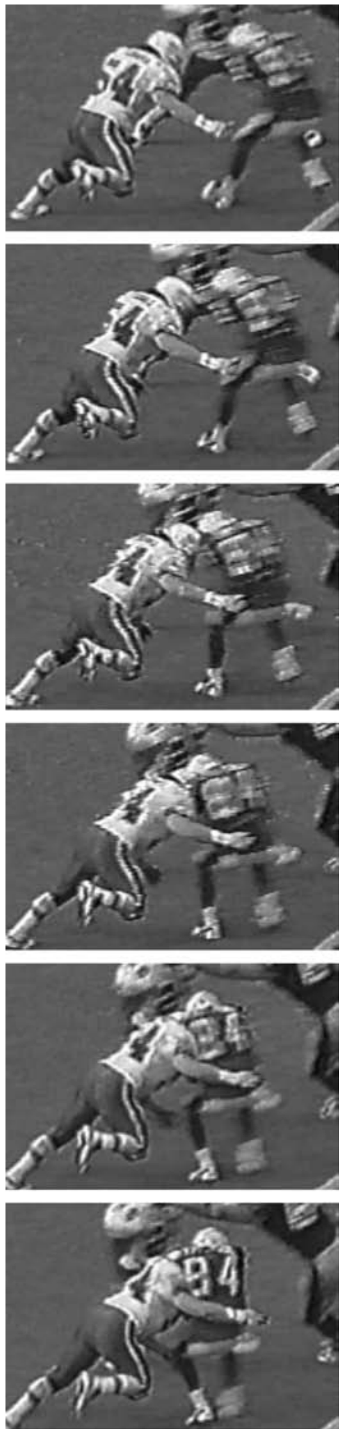

**FIGURE 9.** *Sequence from a game tackle showing the head-down position of the striking player as his helmet impacts the other player's head and his body drives forward. This is Case 38. Sequences from the laboratory reconstruction of this case are shown in Figure 3* (left sequence).

There is another reason why the NFL players are not experiencing neck injuries in concussive impacts. NFL players are typically bigger than the 50th percentile dummy used in the game reconstructions. It is known that the larger the player, the greater the tolerance to impact force. By using established scaling procedures, the NHTSA has found that the tolerable neck compression force for the 95th percentile man is 5440 N, compared with 4500 N for the 50th percentile man (8). However, even if the 95th percentile level is used, 8 of 27 NFL reconstructions exceeded the higher tolerance limit. Obviously, NFL players have a unique ability to sustain impact forces. Nonetheless, the NFL experience reported here provides new tolerance information relevant to a wide range of safety assessments.

Only 4 of 27 of the NFL tackles involved neck compression-extension (*Nce*), with relatively moderate extension moments of 35.7 ± 20.5 Nm. For the majority of cases with neck compression-flexion (*Ncf*), the peak flexion moment was 47.2 ± 25.7 Nm. These values are low in comparison to human tolerance levels. This indicates that the striking players control the impact alignment to limit bending moments and shear forces in their tackling technique. Experience has probably taught this lesson to the players.

The collision mechanics indicate that concussion occurs during the peak load when the highest head accelerations occur in the struck player. Head accelerations in the concussed players are statistically higher than in those not injured or in the striking players. Because the striking players have lower head accelerations, they have more mass in the impact. Their effective mass is 67% greater than that of the struck player. Because neck loads are high at this time, the mass includes torso inertia. Neck forces couple torso mass into the collision, which contribute to the higher effective mass of the striking player.

## Reducing Concussion Risks

There are several ways to potentially lower the risk of concussion in helmet-to-helmet collisions. The primary means would be to enforce head-up tackling techniques. This would reduce the torso inertia involved in the striking player's collision and reduce the impact force. Helmet impacts are 61% of the concussive collisions in the NFL. In this tackle, the striking player delivers more of his momentum to the struck player by impacting the helmet. This lowers the deformation of the striking player's neck, because he initially loads only the struck player's head. A head-up tackling position would reduce the torso mass and lower the force on the struck player. This gives a new reason to reinforce antispearing rules and thereby decrease the risk of concussion in struck players.

The prevention of concussion in the struck player provides another reason to enforce rules against head-down tackling or spearing in football. The most commonly reported reason for this rule has been the risk of catastrophic neck injuries in the striking player when the players tackle or block the torso of an opponent. The biomechanics of that type of tackle involve initial head acceleration in the striking player and a buildup of compressive forces in the neck; but the mass of the struck player's torso is substantial, so the striking player's neck eventually buckles in flexion, lateral bending, or another mode, leading to cervical fracture-dislocations and spinal cord injury.

A second means to lower concussion risks may be to reduce the stiffness of the top-crown region of the helmet. For reasons of durability and to gain performance in National Operating Committee on Standards for Athletic Equipment testing, the top of the helmet is the stiffest part of the plastic shell, much stiffer than the side of the helmet. The striking player uses the top of his helmet to strike the more flexible side of the struck player's helmet, leading to an incompatibility in the deformations of the two shells. It may be possible to include a load-limiting capability in the top of the shell, which would limit the impact force, lower head accelerations, and lengthen the duration of impact if the top of the helmet is used in a tackle. The impact force averaged 7642 ± 2259 N with concussion, so a load limit at the average minus 1 standard deviation would be 5383 N, or approximately 1200 lb. This level would limit the load close to the average for nonconcussion impacts in the NFL reconstructions. A local load-limiting function in the shell may also decouple the helmet mass in the collision. This approach is hypothetical and would require development to ensure an overall performance of any new helmet design to ensure comparable play in all situations and durability.

A third but potentially much less effective means would be to reduce the mass of the helmet, because this would lower the inertia of the striking player in the impact. Football helmets weigh 1.9 kg, compared with the 4.38 kg mass of the head and neck. If the helmet were reduced 20% in weight, there would be a 6% reduction in mass of the striking player's head and an even smaller reduction in the collision force if a head-down impact were used in the tackle. The determination of collision mechanics causing concussion may offer insights for innovators to consider in the development of new safety equipment. In the meantime, enforcement of head-up tackling offers the best means of reducing concussions in helmet-to-helmet collisions.

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## COMMENTS

Football and the National Football League (NFL) have arguably become the most popular sport and professional league, respectively, in the United States. Among 32 professional teams in the NFL and 117 Division IA teams in the National Collegiate Athletic Association ranks, there are thousands of individuals participating every autumn in football at or near the highest level. This number balloons when one includes other professional leagues (e.g., Canadian Football League, Arena Football League, NFL Europe) and non-Division IA collegiate teams.

The number of mild traumatic brain injuries (MTBIs) occurring during participation in football every year is likely underreported. The NFL, as the premier league, has taken the responsibility to fund important research in the biomechanics of injuries leading to MTBI. Viano and Pellman reconstructed from actual game footage 27 helmet-to-helmet collisions that occurred within a 5-year period. Twenty-two of the 27 collisions resulted in concussions to the struck player. Hybrid III male dummies were equipped with standard and linear accelerometers to measure the acceleration and calculate the force occurring during impact. Transducers were placed in the neck to measure the force received by the striking model during collisions. The data provided in this article are necessary to help guide the development of future helmets and to assist the NFL in making and enforcing new rules governing play. Care should be taken, however, when attempting to transfer the data to clinical situations. The research used uniform models and situations during the reconstruction of the impacts. The mass of the striking and struck models was idealized in the authors' experiments. In the NFL, the mass of players can range from 200 lb for defensive backs to 300 lb or more for linemen. Because mass is an important component of the calculation for force, this makes it more difficult to translate the laboratory data to clinical situations. The medical history of the players involved in the original collisions was not examined.

These individuals may have had previous MTBI, which may have made them more susceptible to future MTBI. The authors also state that only helmet-to-helmet collisions were studied.

They did not examine the force sustained during helmet-to-ground collisions, which may contribute to MTBI in the NFL.

This article by Viano and Pellman is an important addition to the series on the biomechanics of MTBI in professional football. This ongoing research is likely to contribute to increased safety in football at all levels.

**Min S. Park**
**Michael L. Levy**
*San Diego, California*

In Part 8 of the NFL concussion series, Viano and Pellman describe the biomechanics of the striking player in 27 helmet-to-helmet collisions in which the struck player sustained a concussion in 22 of the collisions. Videotape analysis was used to recreate the collisions for biomechanical analysis in the laboratory. The authors describe how the striking player uses a head-down position to transfer momentum maximally, which causes MTBI to the struck player. Although none of the striking players in the study sustained a head or neck injury, one-third of the collisions that were reconstructed in the laboratory produced excessive compression forces in the cervical spine that exceeded current tolerance criteria for serious neck injury. Viano and Pellman recommend a head-up tackling position and advocate the enforcement of antispearing rules in football. Since 2001, Bob Watkins has advocated a "see what you hit" approach to tackling in football. Dr. Watkins has made his videotape available free of charge through his foundation (available at: http://www.spineinsports.org/programs.htm). Football helmet companies, such as Riddell with the Revolution helmet and Schutt with the DNA helmet, have made changes to the size, weight, and face masks of helmets. Even with these changes, collisions in football should be initiated with proper technique so as to reduce injury to the striking player and the one being struck.

**Russ Romano**
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Viano and Pellman have provided objective data regarding football head and neck impact biomechanical and loading parameters. These data were acquired by using dummy models to recreate actual game videotaped scenarios. In so doing, the authors observed that supramaximal loads are occasionally encountered. From this important information, they derived recommendations for injury prediction and prevention. This study model may become the "gold standard" in the future. The authors have presented an objective scientific approach to football injury and injury prevention.

**Edward C. Benzel**
*Cleveland, Ohio*

The authors have addressed the biomechanics of the "striking and struck players" with head-down tackling in NFL helmet-to-helmet collisions causing concussion. In this study, 27 helmet-to-helmet collisions were reconstructed in the laboratory using dummies. Although limited to American football, this study does have implications in other football codes that have tackling as a major component of the game, including Australian Rules Football, Rugby Union, and Rugby League codes. However, it should be noted that the descriptions of the biomechanics described apply to the situation in which the player being "hit" does not see the tackle approaching.

This type of study does provide important data to sports administrators that would lower the risk of injury from helmet-to-helmet collisions. The authors suggest enforcing "head-up tackling" techniques and reducing the "stiffness" of the top crown region of the helmet as well as a "potentially less effective" means by reducing the mass of the helmets, because this would lower the inertia of the striking player in the impact. It is noted that helmet impacts comprise 61% of concussive collisions in the NFL, and it would be of interest to compare this study with those that are undertaken in other football leagues that do not use helmets. Devotees of other football codes often wonder whether what seems to be the excessive padding and armor worn by American footballers paradoxically might increase the risk of serious injury by providing a larger target and allowing the player to use the "armor" as a weapon.

The authors have provided an excellent study that continues the series dealing with neurosurgical injuries in professional football. These types of studies can only help us to understand the game better and provide a rational basis for the introduction of rule changes to prevent devastating injuries in the sport.

**Andrew H. Kaye**
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This Part 8 study by the NFL Committee on Mild Traumatic Brain Injury focuses on the important issue of examining the impact biomechanics of players involving the generation of concussive injuries. Specifically, Viano and Pellman have analyzed the factors implicated in imparting sufficient forces to the struck player such that he sustains a concussion, with special attention to the characteristics related to the potential for serious or catastrophic cervical spine injury. Using game videotape analysis, 27 helmet-to-helmet collisions were reconstructed using laboratory test mannequins, in which 22 struck players incurred concussions and 5 did not, although there were no injuries to the striking players (tacklers).

Their findings included that the majority of the impacts to the struck players occurred to the face mask or its area of attachment, whereas in 93% of instances, the striking players made contact with the top or crown of their helmet. This positioning allowed the latter to align their torso with their head and neck, not only striking with the hardest portion of the headgear but generating more mass into the collision. The struck player, in contrast, characteristically did not have sufficient time or warning to bring his body mass into the crash, leading to a 67% greater effective mass of the striking player at the moment of peak force. Regarding concussion, peak head acceleration, peak impact force, and the change of head velocity were all greater in those players sustaining such injuries than in those who did not. The average peak neck compressive force was also greater in the striking players than in the struck players who incurred concussions.

There are several characteristics of NFL players that make them unique and resistant to neck injuries, but this relative safety does not translate to lower of levels of play for the scholar athlete. Once again, rules, customs, and methods of play are usually admired and emulated by younger football players and coaches. The NFL player is ordinarily larger and stronger and has had additional years in which to develop hypertrophied and conditioned neck musculature and supporting ligamentous structures. Professional players have also learned to align their head center of gravity and control impact alignment in their favor to limit cervical flexion or extreme hyperextension movements, as the authors' data demonstrate. Because of the extreme influence that the NFL has on the lower levels of play, it would seem that their changing or evoking greater enforcement of the rules, which are supposed to limit or prohibit initial contact with the top or crown of the helmet, would be most beneficial to scholar athletes, who are at greater risk of sustaining catastrophic cervical spinal injuries. As the authors mention, there are numerous issues involved with helmet redesign that make it a less attractive or viable option to prevent such catastrophic injuries. This study is a valuable analysis of the biomechanical forces and issues involved in cervical spine injuries and should assist in future efforts to limit such occurrences.

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