Verification of Biomechanical Methods Employed in a Comprehensive Study of Mild Traumatic Brain Injury and the Effectiveness of American Football Helmets

J. A. Newman, M. C. Beusenberg, N. Shewchenko, C. Withnall, E. Fournier

Biokinetics and Associates Ltd.
2470 Don Reid Drive
Ottawa, Ontario, K1H 1E1
Canada

BIO/2002/002541
First Revision

Corresponding Author:

Christopher Withnall
Senior Engineer
Biokinetics and Associates Ltd.
2470 Don Reid Drive
Ottawa, Ontario, K1H 1E1
Canada

Phone: +1 (613) 736-0384 ext. 227
Fax: +1 (613) 736-0990

withnall@biokinetics.com
www.biokinetics.com

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Abstract

Concussion, or mild traumatic brain injury, occurs in many activities, mostly as a result of the head being accelerated. A comprehensive study has been conducted to understand better the mechanics of the impacts associated with concussion in American football. This study involves a sequence of techniques to analyse and reconstruct many different head impact scenarios. It is important to understand the validity and accuracy of these techniques in order to be able to use the results of the study to improve helmets and helmet standards.

Two major categories of potential errors have been investigated. The first category concerns error sources specific to the use of crash test dummy instrumentation (accelerometers) and associated data processing techniques. These are relied upon to establish both linear and angular head acceleration responses. The second category concerns the use of broadcast video data and crash test dummy head-neck-torso systems. These are used to replicate the complex head impact scenarios of whole body collisions that occur on the football field between two living human beings.

All acceleration measurement and processing techniques were based on well established practices and standards. These proved to be reliable and reproducible. Potential errors in the linear accelerations due to electrical or mechanical noise did not exceed 2% for the three different noise sources investigated. Potential errors in the angular accelerations due to noise could be as high as 6.7%, due to error accumulation of multiple linear acceleration measurements. The potential error in the relative impact velocity between colliding heads could be as high as 11%, and was found to be the largest error source in the sequence of techniques to reconstruct the game impacts. Full-scale experiments with complete crash test dummies in staged head impacts showed maximum errors of 17% for resultant linear accelerations and 25% for resultant angular accelerations.

Keywords: Concussion, Mild Traumatic Brain Injury, Sports, Reconstruction, Validation, Helmet, Kinematics, Video Analysis, Acceleration, Impact.

1. Introduction

Concussion can be regarded as a form of mild traumatic brain injury (MTBI), which can be induced mechanically as a result of the head being accelerated (Gennarelli 1981, Newman 1982). It is generally regarded that the motion of the brain lags that of the skull and the brain distorts. If this distortion is excessive, neurological dysfunction will subsequently be observed. Protective headgear, i.e. a helmet, reduces the potential for brain injury by reducing the extent to which the brain experiences mechanical loading within the skull. It accomplishes this by reducing the acceleration of the head upon impact.

Criteria by which one might develop human concussion tolerance limits and thereby assess the effectiveness of various helmet design features, have generally been based upon various measures of head acceleration (Newman 1998). To be completely general, both linear and angular accelerations are taken into account.

Most of what is known in this regard has been established from experiments with various surrogates including animals, volunteers, cadavers and occasionally from reconstructing motor vehicle accidents. The limitations of these methods are well known (Newman 1993, Viano et al. 1989).

Traumatically induced brain injury has not been studied extensively for the obvious reason that one cannot willingly expose a live human being to a potentially injurious blow to the head. However, there is an untapped database of information where such blows do occur unwillingly: professional American-style football. In this game, head impacts occur frequently and, though standardized helmets are always worn, athletes do occasionally sustain a concussion.

From 1997 to 2002, a database of 182 head impacts was generated, both with and without MTBI (see Table 1). Several categories of impact have been distinguished, including head-to-head, head-to-ground, and head-to-body part. Injury data was obtained through the National Football League (NFL) Injury Surveillance System. Of these 182 cases, 31 were selected for full-scale laboratory reconstruction.

The objective of the research, as identified by the NFL subcommittee on MTBI, is "to gain a better understanding of how and why concussion occurs, and to devise better means to measure concussion, such that ultimately improved protection can be offered." The work presented herein deals primarily with the means of measuring concussion, or more specifically the kinematic analysis of on-field incidents and the methods of re-enacting them in the laboratory.

2. Football Helmets

2.1 Helmet Design Features

A typical helmet worn by professional American football players is shown in Figure 1. The helmet includes a tough thermoplastic shell covering a series of interlocking pads made of slow recovery polymer foam. This interior padding is supplemented by a series of air inflation chambers for fit adjustment. A faceguard, usually made of coated and welded steel wire, attaches to the shell. The helmet is retained on the head by a Y-type chinstrap. Football helmets must meet specific performance requirements described by the National Operating Committee on Standards for Athletic Equipment (NOCSAE, 1996).

2.2 NOCSAE Football Helmet Standard

NOCSAE was formed in 1969 in response to a growing number of fatalities occurring in the game of American football. Their first football helmet standard was introduced in 1973, and while it has been revised since then, the scope has remained largely unchanged. The test method adopts a humanoid headform instrumented with triaxial accelerometers that is dropped onto a stiff rubber pad from a height up to 1.52 m yielding a direct impact speed of 5.5 m/s. The current failure limit is a severity index (SI) of 1200, where SI = $_T\int a(t)^{2.5}dt$, with a(t) the resultant linear acceleration at the head centre of gravity as a function of time, and T the duration of the acceleration pulse (Gadd, 1966). Variations in headform response from lab to lab are corrected by tuning the accelerometer outputs of bare headform hits onto a specific calibration reference pad. Six specific sites on the helmet and three on the facemask are targeted.

While the NOCSAE standard and improved football helmet designs have combined to almost eliminate severe head injuries and deaths in football, it is estimated that 100,000 concussions occur annually in the game (Kelley and Rosenberg, 1997). Such estimates indicate the need to investigate further the biomechanics of concussion in football.

3. Incident Reconstruction

3.1 Experimental Procedures

Detailed photogrammetric analyses (Newman et al. 1999) yielded the impact sites on, and the relative impact velocity of the players' helmeted heads. Impact velocity, contact points and orientation of the players were then carefully used to reconstruct the incident, using automotive crash test dummies to represent the players. The laboratory setting employs elements of two Hybrid-III adult male anthropomorphic test devices (ATDs or crash test dummies) (Backaitis and Mertz, 1994). A helmeted head/neck assembly is guided in freefall from a height sufficient to achieve the same velocity at impact as the relative velocity determined from the video kinematics analysis. Impact is against another helmeted head/neck assembly attached to a freely suspended ATD torso. In this manner the two colliding heads are free to rebound naturally on impact. In fact the similarity of the post-impact kinematics to the game video is used for verification. A typical test set-up is shown in Figure 2.

Each headform is equipped with nine linear accelerometers set up in a so-called "3-2-2-2 configuration" (Padgaonkar et al. 1975). Processing of the nine signals allows the determination of the complete three-dimensional motion of each head. Linear head centre of gravity acceleration components are measured directly by the three accelerometers located at this position. Angular acceleration components are deduced from the measures of linear accelerations at various sites within the headform. The components of angular accelerations for a system of nine orthogonal accelerometers are determined from the first principles of rigid body dynamics defined by Padgaonkar (1975):

$$\alpha_x = (A_{z1} - A_{z0})/2\rho_{y1} - (A_{y3} - A_{y0})/2\rho_{z3} \quad \text{Eq. 1}$$

$$\alpha_y = (A_{x3} - A_{x0})/2\rho_{z3} - (A_{z2} - A_{z0})/2\rho_{x2} \quad \text{Eq. 2}$$

$$\alpha_z = (A_{y2} - A_{y0})/2\rho_{x2} - (A_{x1} - A_{x0})/2\rho_{y1} \quad \text{Eq. 3}$$

where:

α_i = angular acceleration for component i (x, y, z);

A_{ij} = linear acceleration for component i (x, y, z) along orthogonal arm j (1, 2, 3);

ρ_{ij} = length of orthogonal arm j acting along component i.

The above equations are valid for accelerometers coincident with the origin of the system or coincident with one of the axes. Practical implementation of such a system and computation of the body kinematics are further defined by DiMasi (1995).

3.2 Summary Results of the Reconstructions

The head acceleration time histories for all 31 cases reconstructed are secured in a database. Note that for 27 cases the impact involved two players, so two datasets were obtained. The remaining 4 cases involved impact to the ground, so only one dataset was obtained. Table 2 provides a brief summary of the results of the reconstructions. The average maximum resultant linear and angular accelerations for those who sustained an MTBI were 959 m/s^2 and 6432 rad/s^2 respectively. For those not sustaining MTBI, the corresponding values were 557 m/s^2 and 4028 rad/s^2.

While the above data includes 31 cases of reconstruction, previous reporting of the first twelve cases applied the data to produce various injury risk functions relating to peak linear and rotational accelerations, HIC, SI and a newly introduced maximum head impact power index (HIPI) (Newman, 2000b). Such injury risk functions are based completely on measured acceleration data, and are of limited value if the incident reconstructions which form the basis of these kinds of statistical inferences, are inadequate in some ways. Therefore, possible error sources were studied in detail and are discussed in the remainder of this paper.

4. Discussion of Error Sources in the Reconstruction Process

Six potentially important sources of error are identified. The first three are associated with data acquisition and processing, the latter three with assumptions and simplifications made relative to the ATD representation of actual incidents involving living human beings.

The former include:

- Data anomalies due to transducer and data processing noise.
- Headform system response anomalies associated with signal frequency content and the effects of analog and digital filtering of raw and processed data.
- The inherent accuracy and stability of the specific 3-2-2-2 nine-accelerometer package employed to compute the angular accelerations due to the physical implementation and signal conditioning methods.

The latter include:

- Errors in establishing the relative velocity of colliding players from video recordings, and the accurate identification of the helmet impact sites. These matters relate to scaling factors, image resolution and unknown camera locations.
- The effects of neck coupling and effective body mass on head acceleration response.
- The fidelity of representing the whole body collisions that occur on the football field by the ATD head-neck-torso simplifications employed in the laboratory reconstructions.

4.1 Data Anomalies Due to Noise Sources

High frequency noise, inherent to any data acquisition system, was analyzed and quantified for its effect on acceleration magnitude and bias. The linear accelerometers used in this study (Endevco model 7264B-

2000) were energized and held stationary while sampling at 10 kHz, and filtered using a CFC 1000 anti-aliasing filter complying with the Society of Automotive Engineering standard J211 (SAE 1995).

In these stationary tests, the accelerometer signals show oscillations of $\pm$ 1.2 m/s^2 at 500 g full-scale setting. The oscillation frequency was above the filter low-pass cut-off settings, which indicates that it originated from the data acquisition card used to convert the analogue data to digital. When processed through the nine-accelerometer-package to calculate the rotational accelerations, the effect of this noise ranges from -50 rad/s^2 to 85 rad/s^2, again at 500 g full-scale setting of the accelerometers.

In the reconstructions, head responses typically far exceed 10g and 2000 rad/s^2. Noise effects on the linear and angular amplitudes are hence well under 1.2% for linear accelerations and well under 4% for angular accelerations.

Bias, or offset, in an acceleration signal can cause errors when signals need to be integrated over time. This is particularly relevant in this study where the calculation of angular acceleration through the nine-accelerometer-package uses angular velocity, which is established by integrating linear acceleration signals. Typical data acquisition practices, such as employed in this study, minimize this bias electronically and computationally with data processing.

Bias was not present in the individual linear accelerometers (noise oscillations are equally positive and negative around the mean excitation), however combined responses, such as resultant accelerations, possess bias. Integration of such responses over time can result in erroneous velocity calculations. In the static tests conducted for a full-scale setting of 500g, the bias magnitude for the resultant linear acceleration was found to be in the order of 0.17% to 2%, and 0.4% to 2% for the resultant angular acceleration. For linear velocity and angular velocity calculations based on integration of accelerations, maximum bias errors of up to 2% and 6.7% were observed. Head injury assessment functions that employ linear or angular velocity should take into account potential errors introduced by signal bias. Errors in such functions are not necessarily at the same level as that of the velocity, but depend on the calculation method.

4.2 System Vibration Response Anomalies

4.2.1 Hybrid-III Head-and-Neck System Responses

The Hybrid-III Anthropomorphic Test Device (ATD) was developed to help assess injury potential in automotive crash testing (Backaitis and Mertz, 1994). It typically does not "wear" a helmet and indeed this dummy was not intended for the application envisaged here. To examine possible error sources and artefacts associated with the use of the Hybrid-III headform in this type of application, the test set-up shown in Figure 3 was utilized.

The Hybrid-III head and neck assembly was mounted on an adjustable platform that allowed any point on the head to be presented to the impact face. The headform was fitted with two layers of nylon stocking material to provide a uniform surface friction and to facilitate helmet donning. A large Riddell® VSR-4 helmet, without a face cage, was placed on the headform. It was rotated until the lower edge of the brow pad was 80 mm above the tip of the nose. The chin cup was secured and any slack in the straps was removed.

The impact pendulum comprises a weighted hammer that is drawn back to a pre-determined height and released. It was designed primarily to mimic head-to-head football collisions. The pendulum head has a face diameter of 152 mm and a spherical radius of 130 mm. The impact face was covered with a layer of polycarbonate sectioned from a football helmet to provide the proper friction interaction with a helmeted headform. The pendulum has a moment of inertia of 86.69 kgm^2 about the pivot and an arm length of 2.46 m from the pivot to the centre of the impact face.

The forehead target was aligned with the mid-sagittal plane of the headform and 38.1mm above the head centre of gravity. The side impact was on the left side at 90 degrees to the mid-sagittal plane, at a point in-line with and 38.1mm above the head centre of gravity. The headform was equipped with nine accelerometers in the 3-2-2-2 configuration. For impacts of severity consistent with those associated with head injury in the field, the impact pendulum was drawn back to a height of 1.5 m to achieve an impact velocity of approximately 5 m/s.

The goal of this test program was to study the effect of data acquisition parameters on the measurement accuracy of the data in both magnitude and phase. To minimize the scope of this exercise, the sampling rate of the data was maximized to 10 kHz, to ensure accurate capture of temporal data in the frequency range of interest. The full-scale setting of the data acquisition system was set to the normal value (500 g) and at a minimum value (250 g) above the expected peak accelerations loads in the re-enactments.

Linear acceleration signals were analysed to study the frequency content of the signal. This provided insight into the response of the headform by indicating if specific frequencies are being attenuated or amplified as a result of anti-resonance or resonance modes of the accelerometers or headform. Typical headform accelerometer response and corresponding power spectral density plots are presented in Figure 4. Similar trends are noted for frontal and lateral impacts. The signals include limited high frequency components (>1000 Hz), and no specific resonance or anti-resonance modes are observed.

4.2.2 Filtering Effects

Impact response data was filtered with a CFC 1000 low pass anti-aliasing filter, appropriate for a sampling rate of 10 kHz according to SAE J211 (SAE 1995). In addition, the standard practice for the incident reconstructions was to further filter the data with a CFC 180 low pass filter in order to help minimize the influence of spurious mechanical noise on the angular acceleration calculations. The effect of such filtering was determined in actual tests.

Table 3 compares peak acceleration responses filtered at CFC 1000 and CFC 180 obtained in frontal pendulum impact tests with and without a helmet. In the helmeted headform tests, differences between the two filters are small: less than 2.2%. Because of higher frequency content in the headform only test, the effect of filtering is higher: up to 8.1%. This condition, however, is of less significance for the reconstruction of head impacts in football, and is included here to illustrate that filtering effects depend on the system considered. At the higher impact velocity, peak linear and angular acceleration are slightly higher after filtering at CFC 180, which is due to amplification of the low frequency signal content inherent to the filter. The increase is very small (less than 1%) and not considered significant for the current study.

4.3 Angular Acceleration Measurement Variables

4.3.1 Neck Pendulum Test Set-up and Impact Configurations

Verification of the angular acceleration measurement method was completed with the use of a standard Hybrid-III neck pendulum, shown in Figure 5. This is typically used for calibration of neck structures (FMVSS 208, 1999). Three types of impact were employed. The first involved releasing the pendulum from a fixed height and arresting the arm with Hexcell®, an aluminium honeycomb structure that provides a near constant pendulum deceleration. In this impact type, the bare head and neck assembly underwent pure inertial loading. In the second impact type, the bare headform impacted a padded surface at the same instance that the pendulum was arrested. In the third impact type, the helmeted headform struck a rigid surface at the same time that the pendulum was arrested.

The Hexcell® was of the same type used in standard crash dummy neck calibration tests but with block dimensions of 108 mm high and 162 mm wide, or 7x9 cells respectively. Padding for the impact surface was similar to that used in football helmets: a vinyl-nitrile foam rubber made by Rubatex, model R3953, 17.5mm thick. The rigid impact face was covered with polycarbonate sheet, similar to the material used for football helmet shells. Pendulum instrumentation included a uni-axial accelerometer mounted on the arm, and a velocity gate to determine the impact speed.

Three impact configurations were considered: frontal, lateral, and oblique (aligned with the head at the same elevation but rotated 45 degrees); see also Figure 5. The impact conditions for the pendulum were chosen to obtain the same impact severity as the re-enactments in terms of peak acceleration and velocity change.

For the test series, a large-sized football helmet, without a facemask, was installed onto the headform. All repeated impacts were conducted on the same helmet with a rest period to allow for helmet and neck recovery. The helmet and chin strap were properly fitted and adjusted prior to each test. The position of the headform was adjusted to ensure the pendulum struck the desired target. The forehead target was aligned with the mid-sagittal plane and 38.1 mm above the head centre of gravity. The side impact was on the left side of the headform at a point in-line with and 38.1 mm above the head centre of gravity, in the direction 90 degrees to the mid-sagittal plane.

The test matrix included both inertial and impact tests for non-helmet and helmeted headforms for frontal, side and oblique impacts. Only the high severity helmeted head impact series, which is most relevant for the reconstructions, is considered.

4.3.2 Angular Acceleration Methods

Verification of the angular acceleration methods was conducted by comparing the results of three independent systems: the 3-2-2-2 method, angular accelerometers, and the in-line method. The 3-2-2-2 method consisted of nine uni-axial Endevco® 7264B-2000 accelerometers at known positions in an orthogonal arrangement. Three sensors were mounted near the headform centre of gravity (CG), two on the anterior surface of the skull, two on the lateral surface, and another two on the superior surface (R.A. Denton headform model 3623). The nine-accelerometer configuration is implemented in the Hybrid-III headform as shown in Figure 6. The angular acceleration computations assumed a zero initial condition just prior to impact.

The direct angular accelerometer method utilized transducers capable of directly measuring angular accelerations (Endevco® 7302B-M4). These were intended to meet the frequency response specifications in SAE J211 (SAE 1995) but incorporate oil damping to suppress resonance modes in the frequency range of interest. A mounting block was designed (see Figure 6, middle) to allow the addition of three angular accelerometers to the standard Hybrid-III headform with the 3-2-2-2 accelerometer configuration. The block fitted over the linear tri-axial accelerometer cluster at the head CG. Once in position, the angular accelerometers were aligned with the three principal axes of the head.

The in-line method used five Endevco® 7264B-2000 accelerometers mounted to a machined surface on the inside of the skull cap with their sensitive axes aligned with the mid-sagittal plane of the headform. The vertical positions of the accelerometers included one at the height of the centre of gravity of the head and the others above and below (see Figure 6, right). This was similar to the method employed by General Motors Corporation in the early 1980's (Viano, 1986), where three sets of five accelerometers were used for triaxial angular acceleration measurement. However, the current set-up was sensitive only to angular acceleration about the Y-axis. It has nevertheless been dubbed the 2D in-line method due to its sensitivity to rotational motion in the two-dimensional mid-sagittal plane. Calculation of the angular accelerations was performed using a least squares approximation of the angular accelerations determined from the five accelerometers and corresponding moment arms about the centre of gravity.

4.3.3 Comparison Based on Frequency Content of the Angular Acceleration

Comparison of the high severity impact data was completed in the frequency domain with a typical plot as shown in Figure 7. It can be seen that there are no significant differences for the main portion of the signal content (below 1000 Hz). However, some systems did exhibit artefacts of sufficient magnitude above this frequency to modify the true signal (i.e. the angular accelerometer system had resonance modes around 1875 Hz - 3200 Hz).

4.3.4 Comparison Based on Peak Magnitude of Angular Acceleration

For inertial-only tests, where no direct headform contact occurred, the differences between the three systems were nearly identical and unremarkable. However, where padded and/or helmeted headform impact occurred, there were notable differences. Typical angular acceleration responses for such tests are provided in Figure 8 where the angular accelerometers are shown to resonate. This is likely related to the inherent design of the unit and its susceptibility to resonance (see also Figure 7). For the frontal impact cases which excite the headform in the mid-sagittal plane only, the 3-2-2-2 and 2D in-line systems were observed to compare favourably, with peak angular response in frontal impact tests differing by no more than 6%. The in-line method was not suitable for evaluation under the lateral and oblique impact conditions and could therefore not be evaluated.

4.3.5 Selection of Angular Acceleration Method

The angular accelerometer approach is attractive in its simplicity and ability to measure data directly with no complex mathematical calculations. However, its inherent ringing made it unsuitable for direct headform impact situations. The 3-2-2-2 and 2D in-line methods appeared to both be suitable for direct headform impact. But for the in-line method, to achieve complete 3D angular acceleration measurement, fifteen accelerometers would be required, plus three at the centre of gravity for a total of eighteen, which is twice as many units compared to the 3-2-2-2 system. Additionally, the computational algorithms for the 3-2-2-2 system are accurate, robust, based on physical principles, are publicly available, and the machined headform is commercially available. The tests also showed that accuracy was acceptable with the current system in the presence of headform vibrations, accelerometer mounting errors, accelerometer frequency response, signal noise, bias, and conditioning methods for the impact environments evaluated, For these reasons, the 3-2-2-2 system is considered the most suitable for the NFL laboratory reconstructions.

4.4 *Kinematic Analysis of Broadcast Video Recordings –Velocity Determination*

In order to conduct full-scale laboratory reconstructions of head impacts in American football, the speed of collision needed to be determined. Video clips of the game incident were available, but they were taken from some unknown angle relative to the true impact velocity vector. To resolve this, a videographic analysis method was developed (Newman, 1999). This process involved establishing the observed velocity vector of one player's head relative to the other, at the moment just prior to collision, in the plane of the camera view. This was repeated for a second camera view. Speed was calculated by measuring the successive distance moved by a player's head from one video frame to the next. The helmet was used as a scaling reference to convert pixels to metres, and also to correct for camera zooming between frames. Using the gridlines on the field for reference, a CAD representation of the field was used to determine the vantage point of each camera, and then using the elevation and skew angles, these two velocity vectors were projected back onto the field. The intersection of the two vector heads and tails generated the true velocity vector of the player on the field. This method is illustrated in Figure 9.

To confirm that this method of camera angle and speed estimates was valid, an event was staged at a NFL football stadium. Players were represented by helmeted volunteers driving motorized utility carts. A broadcast video crew set up cameras in various locations, and recorded a series of trials where the volunteers passed each other at speeds similar to those of running players. The cart speeds were measured directly using surface-mounted tape switches a known distance apart that triggered an electronic timer. The actual 3D coordinates of the camera positions were surveyed relative to the football field. Three scenarios, each at different locations on the field, were conducted, as shown in Figure 10.

Following the established protocol, the video of the three scenarios was analyzed and the camera angles relative to the field were calculated to obtain the true relative velocity vectors. The calculated camera angles compared to true surveyed angles, and the velocity calculations relative to the true measured cart speeds are provided in Table 4.

Camera angle error was shown to be very low, with only 1.6 degrees maximum error. The speed calculation error for the first scenario was also very low, at only 1.2%. For the second and third cases, errors were larger, at 10.6% and 11.3% respectively, and can largely be attributed to errors in scaling and digitizing the video images. In the second case, the motion was diagonal to the camera view, so only a skewed view of the helmet was seen. This made it harder to scale the observed helmet width to an actual helmet. In the third scenario, the test was at the far end of the field, so the images were smaller. This meant that there were fewer pixels measured across the helmet image, offering poorer scaling resolution. These two latter cases could be considered worst case situations in comparison to the several game impacts that were analyzed.

4.5 *Head-Neck-Torso Coupling*

In the laboratory re-enactment set-up (see Figure 2) for head-to-head collisions the striking player was represented by a full Hybrid III head, neck and torso. The struck player was represented by only a head and neck, with the neck being rigidly connected to a rolling carriage. To investigate the appropriateness of removing the torso mass from this second dummy an investigation was conducted using computational modelling methods with MADYMO (Beusenberg, 2001).

A rigid body mathematical helmet model was created using laboratory impact test data. The neck model was the so-called "global head-neck model" developed by de Jager (de Jager, 1996). Various situations of head-neck-torso coupling were investigated, using the human neck model, a Hybrid III neck model, and various body mass values.

In neck compression, the simulations revealed that the human neck model simply collapsed, and the Hybrid III neck was effectively infinitely stiff, and the results were of limited value. However, in lateral impact, which tends to represent the struck, or typically injured head, the simulations revealed that neck coupling has limited effect on linear head acceleration, but considerably affects the rotational head response. The effect of torso mass was found to be insignificant, suggesting that the current laboratory re-enactment practise, where there is no torso for the dummy representing the struck player, is suitable.

4.6 *Kinematic Replications*

Despite the good confidence in the input parameters for re-enacting the on-field collision, it was necessary to investigate how accurately the overall laboratory re-enactment set-up worked. This included not only the video speed estimates, but the orientation of the lab dummies and the contact point. It was decided to conduct an "on-field" player impact using complete instrumented test dummies, then to replicate this incident in the lab using traditional techniques.

The set-up for the complete dummy tests consisted of two gantries that guided suspended full-size Hybrid-III dummies to collide with each other at a predetermined impact site and velocity. The dummies were suspended over a mock-up football field which included simulated gridline and yard-line markings as shown in Figure 11.

The simulated scenarios included: a straight on collision with both players moving, a straight on collision with one player stationary, and an oblique collision with both players moving. An additional head-to-ground impact was also simulated. In each simulation the dummy heads were instrumented to measure the linear and angular head accelerations. Each dummy was fitted with a football helmet and protective shoulder padding.

A professional film crew was hired to video tape the tests from two different vantage points. Kinematic analysis of the resulting video determined the camera angles and the relative impact velocities. These results were compared to the actual recorded velocities as shown in Table 5.

The "actual" values indicated in Table 6 were obtained from direct measurement of the test set-up and confirmed with supplemental high-speed video documentation The largest velocity error was 8.3% in the head-to-ground test.

A laboratory reconstruction of each full-scale simulation was conducted using the results from the kinematic analyses. The reconstruction method was done by the established methods (Newman, 1999). Head responses obtained from the laboratory reconstructions are compared to the responses obtained during the full-scale simulations and are given in Table 6.

In this testing, the blue helmeted dummy was set up to represent typically the striking player and the yellow helmeted dummy was set up to represent typically the struck (i.e. injured) player. Differences in maximum resultant linear accelerations between the full-scale simulations and the laboratory reconstructions range from 6% to 17% for the struck dummy and 11% to 12% for the striking dummy. Differences in maximum resultant angular accelerations range from 4% to 14% for the struck dummy and 13% to 25% for the striking dummy. These findings confirm that the laboratory reconstruction methodology is sufficiently representative of the on-field player collisions within the expected bounds of repeatability of the experimental setup.

Conclusions

Many issues surrounding the prevention, management, and treatment of mild traumatic brain injury (concussion) remain under study. The biomechanics research on concussion occurring in professional American football that has continued for the past 5 years, provides a new foundation for understanding the biomechanical conditions for concussion, and hence for improving the effectiveness of football helmets. Through a comprehensive analysis and reconstruction of head impacts sustained by football athletes with and without concussion, linear and angular head acceleration data related to injury outcome has become available. This will provide the opportunity to devise injury risk functions, and further improve the protection offered to athletes in sports where concussion is a risk.

The reliability of such data, and its potential future contribution to the advancement of helmet standards, depends on the reliability of the methods used to obtain it. These have been investigated in detail and major error sources of the game impact reconstruction methodology have been identified and quantified, and the findings are summarized as follows:

- Inaccuracies can be introduced due to noise or bias errors inherent to the data acquisition system. In most cases, these are negligible but in worst cases will not exceed 7%.
- The head-neck system used in the study (Hybrid-III) does not exhibit resonant frequencies in the range of interest when assessing helmet performance.
- Potential loss of signal content (i.e. amplitude) due to CFC 180 filtering is small and does not play a significant role in the assessment of angular head acceleration.

- The 3-2-2-2 method used to establish angular accelerations in the reconstruction of actual game impacts, does not exhibit signal anomalies due to the implementation methods. Also, this method was found to be accurate, stable, and repeatable. When tested with two other methods under identical impact conditions, comparative differences in angular acceleration amplitude were less than 6%.
- The largest error source in the reconstruction technique is the determination of the relative impact velocity of a helmet colliding with another object. Due to the unknown absolute camera positions with respect to the location of the impact in space, the calculated velocity could deviate as much as 11% from the actual velocity.
- The issue of head-neck-torso coupling was investigated through MADYMO modelling using a human neck model, a Hybrid III neck model, a football helmet model and various torso masses. The results indicate that neck stiffness has minimal effect on linear accelerations, but does affect angular head accelerations. Torso mass has minimal effect on head accelerations, supporting the current laboratory reconstruction techniques.
- The overall accuracy of the reconstruction method was established in four different staged impact scenarios. Based on comparisons of peak linear and angular accelerations, the error for a typical reconstruction is less than 17% for peak linear head acceleration, but can occasionally be as high as 25% for the angular accelerations.

Despite the complexity of the complete reconstruction methodology, from establishing impact location, direction, and velocity from broadcast video recordings, measuring linear accelerations of crash test dummies representing the athletes in the re-enacted game incidents, to post-processing the data in combinations of linear and angular head accelerations, the overall accuracy is well within the expected bounds of repeatability of the experimental set-up. This provides a good foundation for the use of the experimental data in the development of new injury risk functions related to concussion, as well as the evolution of helmet performance standards.

Acknowledgements

Biokinetics would like to thank the NFL and NFL Charities who made this research possible. The NFL Subcommittee on MTBI, chaired by Dr. E. Pellman, is thanked for their direction and support. Mr. D. Blandino is thanked for providing the video data. Dr. J. Powell is thanked for providing injury confirmation from the NFL injury surveillance system. Dr. D. Viano is thanked for his continued guidance. Ms. D. Mitchell is thanked for her enthusiasm and support for this research.

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Figure Captions

Figure 1: American style football helmet.

Figure 2: Typical set-up of a reconstruction of a "helmet-to-helmet" impact of American football athletes. Note: the picture is rotated 90 degrees (G=gravity).

Figure 3: Impact pendulum set-up for frontal (left) and lateral (right) strikes.

Figure 4: Accelerations (left) in front and side impact, and power spectral density graphs (right).

Figure 5: Neck pendulum set-up (right) and detail of oblique test with helmet and striker plate (left).

Figure 6: Angular acceleration assessment methods: 3-2-2-2 cluster (left); direct method (middle); 2D in-line method (right)

Figure 7: Typical power spectral density plot for all three systems in a contact event.

Figure 8: Typical rotational acceleration responses in neck-pendulum tests (CFC 1000 for all systems).

Figure 9: Principle of reconstruction of the 3-dimensional relative impact velocity from two camera views.

Figure 10: Illustration of camera orientation assessment method.

Figure 11: Full-scale simulation of an actual game collision.

Figure 1: American style football helmet.

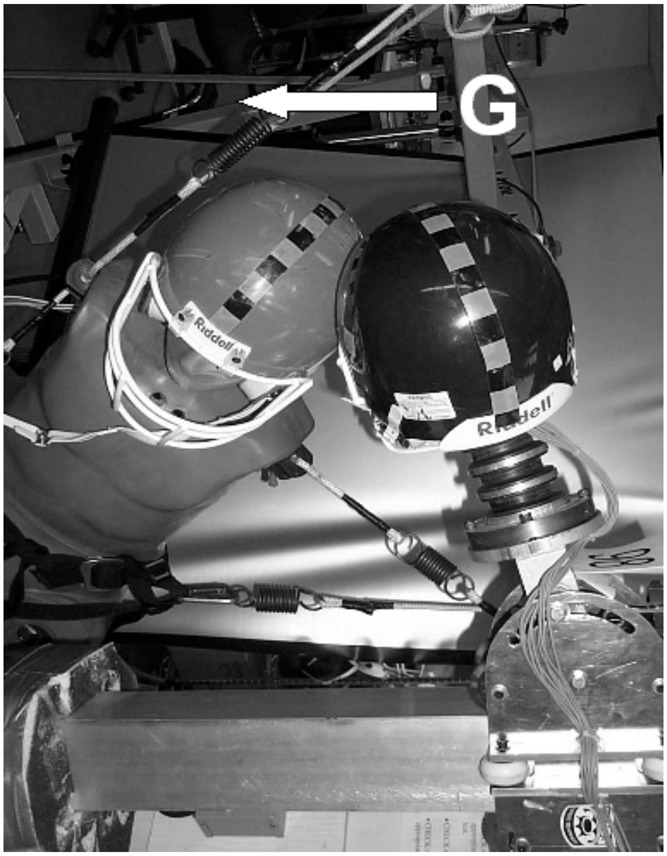


Figure 2: Typical set-up of a reconstruction of a “helmet-to-helmet” impact of American football athletes. Note: the picture is rotated 90 degrees (g=gravity).

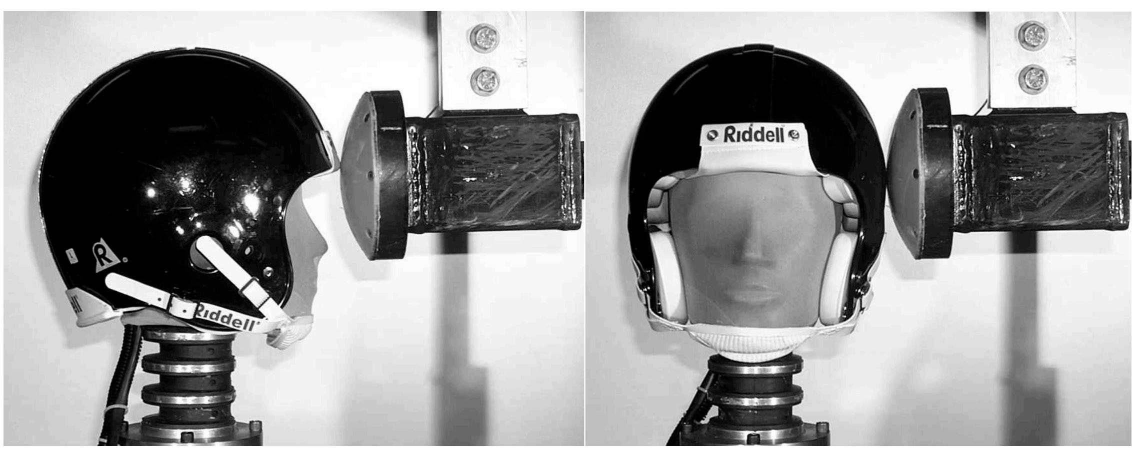


Figure 3: Impact pendulum set-up for frontal (left) and lateral (right) strikes.

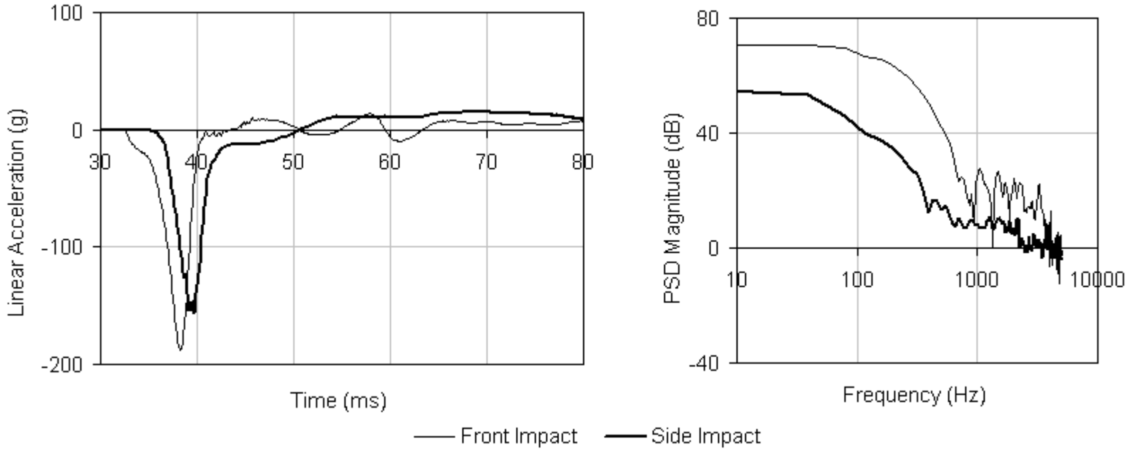


Figure 4: Accelerations (left) in front and side impact, and power spectral density graphs (right).

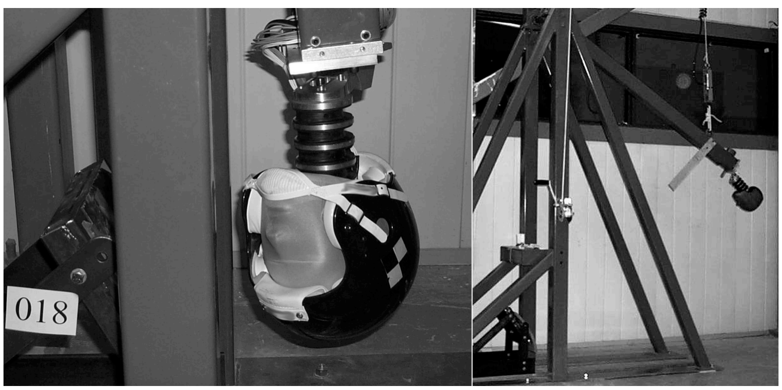


Figure 5: Neck pendulum set-up (right) and detail of oblique test with helmet and striker plate (left).

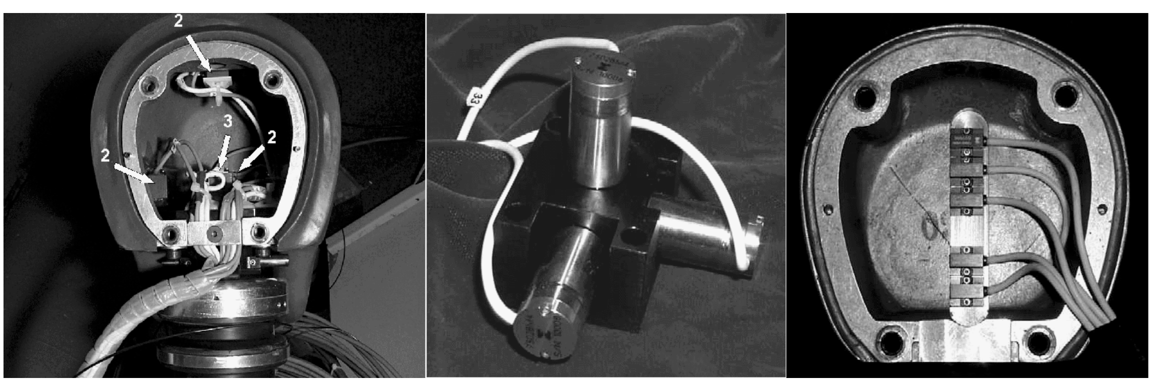


Figure 6: Angular acceleration assessment methods: 3-2-2-2 cluster (left); direct method (middle); 2D in-line method (right)

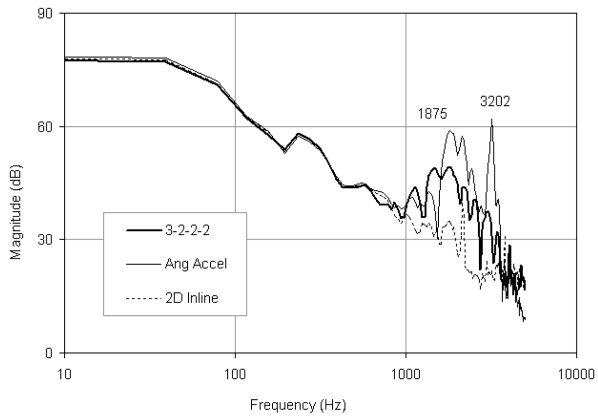


Figure 7: Typical Power Spectral Density plot for all three systems in a contact event.

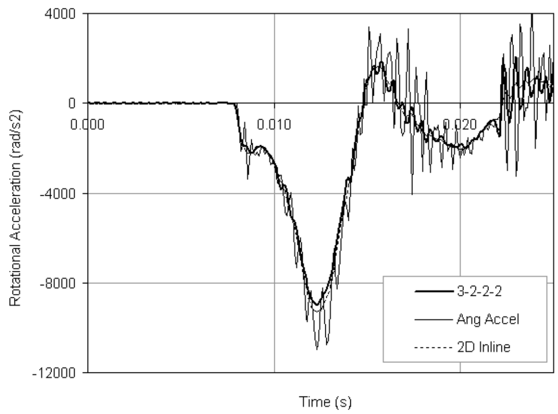
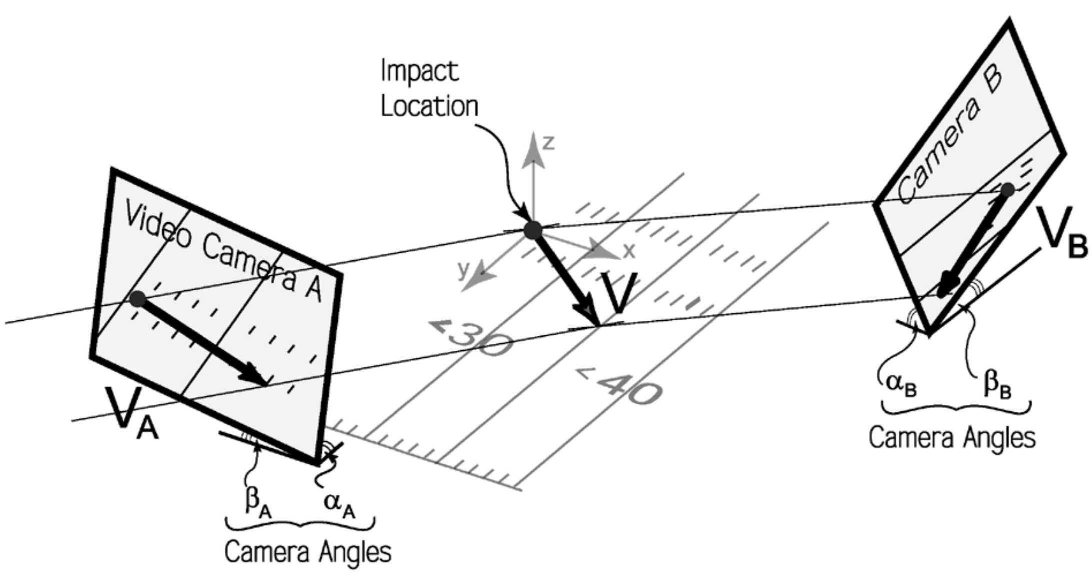


Figure 8: Typical rotational acceleration responses in neck-pendulum tests (CFC 1000 for all systems).



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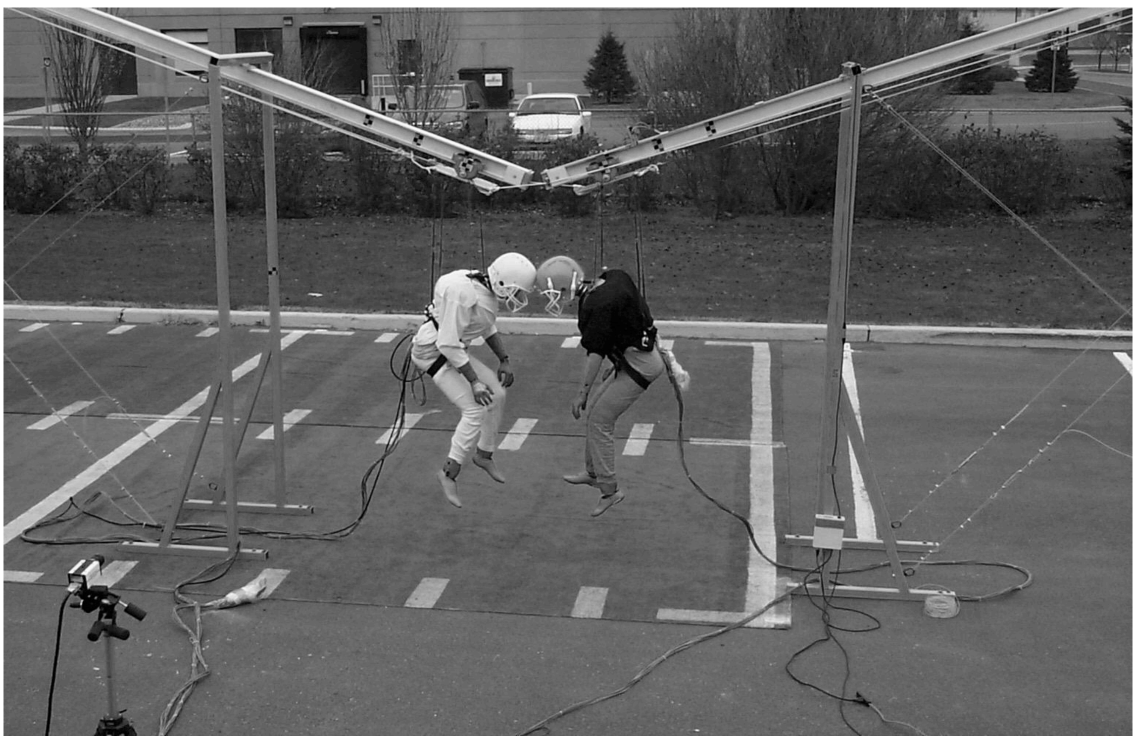

Figure 11: Full-scale simulation of an actual game collision.

Table Captions

Table 1: Overview of incidents in the NFL-MTBI reconstruction database.

Table 2: Summary results of the reconstructions.

Table 3: Variations in peak acceleration responses between CFC 1000 and CFC 180 filtering.

Table 4: Calculated camera angle offset versus actual surveyed measurements and calculated speed estimates versus actual recorded values.

Table 5: Comparison of the measured and the actual relative impact velocities.

Table 6: Comparison of full-scale simulation and laboratory reconstruction results.

Table 1: Overview of incidents in the NFL-MTBI reconstruction database.

Impact Configuration	Incidents on Video	Reconstructions		
		Subtotal	MTBI	Non-MTBI
Head-to-Head	92	27	22	5
Head-to-Ground	31	4	3	1
Head-to-Body Part	44	-	-	-
Unknown Contact	15	-	-	-
Totals	182	31	25	6

Table 2: Summary results of the reconstructions.

		Struck Player			Striking Player		
case	velocity (m/s)	peak res. lin. accel. (m/s^2)	peak res. ang. accel. (rad/s^2)	MTBI? (yes/no)	peak res. lin. accel. (m/s^2)	peak res. ang. accel. (rad/s^2)	MTBI? (yes/no)
7	6.9	597	6266	Yes	488	2832	No
9	10.3	1317	7428	Yes	778	6719	No
38	9.5	1162	9678	Yes	587	5205	No
39	10.9	1262	5921	Yes	431	4487	No
57	8.8	757	6514	Yes	317	4151	No
69	10.3	595	4381	Yes	371	2620	No
71	10.3	1211	5400	Yes	1005	5541	No
77	9.9	788	5148	Yes	343	2714	No
84	9.4	803	9193	Yes	443	3169	No
92	11.1	1053	6878	Yes	585	6070	No
98	9.6	893	7548	Yes	827	4487	No
113	7.0	575	3965	Yes	597	3700	No
118	10.7	987	7017	Yes	545	3687	No
124	11.4	799	7138	Yes	545	4086	No
125	11.7	1109	7716	Yes	457	3366	No
135	10.0	1352	7540	Yes	790	5005	No
148	6.6	470	3476	Yes	323	2466	No
155	9.1	984	6940	Yes	437	4217	No
157	10.8	1007	6750	Yes	774	4662	No
162	5.5	511	2615	Yes	283	1672	No
164	10.8	1213	9590	Yes	870	6136	No
181	11.7	910	8011	Yes	832	6613	No
48	9.7	562	5617	No	310	2939	No
59	5.3	807	5387	No	314	2087	No
154	6.6	524	4167	No	285	3159	No
175	9.6	605	3555	No	464	2535	No
182	8.1	830	5512	No	857	3206	No
		head-to-ground cases					
67	8.1	1328	5957	Yes			
123	6.3	1188	4727	Yes			
133	14.6	1109	5012	Yes			
142	3.1	185	1170	No			

Table 3: Variations in peak acceleration responses between CFC 1000 and CFC 180 filtering.

Test Conditions				Responses at CFC 1000		Responses at CFC 180	
Site	Helmet	Drop Height (m)	Measured Velocity (m/s)	$a_{res,max}$ (m/s^2)	$\alpha_{res,max}$ (rad/s^2)	$a_{res,max}$ (m/s^2)	$\alpha_{res,max}$ (rad/s^2)
Front	VSR-4	2.486	6.32	1836	5985	1853	6033
Front	VSR-4	1.165	3.22	391	2254	389	2205
Front	-none-	1.165	3.21	1242	3862	1142	3637

Table 4: Calculated camera angle offset versus actual surveyed measurements and calculated speed estimates versus actual recorded values.

Subject 1	Subject 2	Location	Line of sight: offset from actual (degrees)				Velocity		
			50 yard-line camera		End zone camera		Actual	Calculated	Diff.
			Planar	Elevation	Planar	Elevation	(m/s)	(m/s)	(%)
Moving	Stationary	East 20 yard, along field	0.0	0.2	0.3	1.6	8.1	8.0	1.2
Moving	Moving	50 yard, diagonal	0.2	0.1	0.0	0.1	12.3	11.0	10.6
Moving	Moving	West 45 yard, across field	0.3	0.4	0.3	0.3	12.4	11.0	11.3

Table 5: Comparison of the measured and the actual relative impact velocities.

Test #	Simulated scenario	Relative velocity (m/s)	
		Measured	Actual
4	Straight-on, both players moving	8.2	8.2
5	Straight-on, one player stationary	5.2	5.1
7	Oblique, both players moving	7.6	7.2
10	Head-to-ground	4.4	4.8

Table 6: Comparison of full-scale simulation and laboratory reconstruction results.

Test #	Test Velocity (m/s)	Helmet	Full-scale simulations		Laboratory reconstruction	
			Res. linear (G)	Res.angular (rad/s^2)	Res. linear (G)	Res.angular (rad/s^2)
4g	8.39	Blue	76.8	3834	86.1	3047
		Yellow	70.1	2705	63.1	2800
5g	4.99	Blue	29.4	2192	32.9	1638
		Yellow	38.1	1784	33.8	1988
7f	7.68	Blue	44.6	3572	49.5	3097
		Yellow	52.3	4197	61.4	4775
10b	4.46	Yellow	87.6	3309	92.5	3783