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Validation and Application of a Methodology to Calculate Head Accelerations and Neck Loading in Soccer Ball Impacts

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ABSTRACT

Calculating head accelerations and neck loading is essential for understanding and predicting head and neck injury. Most of the desired information cannot be directly measured in experiments with human volunteers. Achieving accurate results after applying the necessary transformations from remote measurements is difficult, particularly in the case of a head impact. The objective of this study was to develop a methodology for accurately calculating the accelerations at the center of gravity of the head and the loads and moments at the occipital condyles. To validate this methodology in a challenging test condition, twenty (20) human volunteers and a Hybrid III dummy were subjected to forehead impacts from a soccer ball traveling horizontally at speeds up to 11.5 m/s. The human subjects and the Hybrid III were instrumented with linear accelerometers and an angular rate sensor inside the mouth. The dummy was also equipped with accelerometers at the center of gravity of the head and load cells in the upper and lower neck. The force applied to the head by the soccer ball was calculated by double differentiating the ball displacement measured from high speed video. Standard mechanics equations were used to transform mouth accelerations to the head center of gravity and to calculate loads and moments at the occipital condyles. Accurate angular acceleration data were obtained by rigidly mounting a small angular rate sensor inside the mouth on a bite block. The neck loads calculated using inverse dynamics required filtering to a cutoff frequency of 50 Hz in order to reduce the noise to an acceptable level and achieve a good match with the neck load cell data. Noise was a particular problem in the calculated occipital condyle sagittal plane bending moment. Although some differences in the results of the human and dummy tests were observed, The Hybrid III dummy head and neck appeared to be reasonably biofidelic in this loading scenario.

INTRODUCTION

Researchers have been using head-mounted accelerometers in biomechanical studies for many decades. Detailed methods for transforming remotely mounted accelerometer readings to the center of gravity of the head and calculating loads and moments at the upper neck using standard physics equations were described by Mertz and Patrick [1] and Ewing and Thomas [2]. Since that time, hundreds of studies have reported on head kinematics in human volunteer testing, and some have also reported neck loads and moments [3-4].

Although straightforward in theory, the inverse dynamics method is difficult to implement in practice. When transforming remotely measured head accelerations to the center of gravity of the head, it is necessary to know the location and orientation of the head sensors relative to the center of gravity of the head. However, there is currently no practical method for determining its location in a living human, so its location must be estimated from measurements made on cadavers. Another practical issue is the mounting of sensors to the head of a living human. Ideally, sensors should be mounted as rigidly as possible to the skull to minimize vibration artifacts. Motion of the sensors relative to the head is usually not a problem when head accelerations are low. However, past studies have generally reported noisy signals from

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head accelerometers when the subjects experienced a head impact [1, 5-6]. Head impacts present an additional challenge when calculating neck loads, because the magnitude, direction, and point of application of the force on the head must be determined. To our knowledge, no studies since Mertz and Patrick [1] have attempted to calculate neck loads in live human volunteers subjected to a head impact. The testing analyzed in this study was previously described by Funk et al. [7-8]. The purpose of the present paper was to present a detailed analysis and validation of the methodology used to calculate head accelerations and neck loads in live humans in a challenging test condition involving a soccer ball impact to the forehead. This methodology was then applied to human volunteers to verify that reasonable results could be obtained and to evaluate the biofidelity of the Hybrid III head and neck in this test condition.

METHODS

Twenty (20) human volunteer subjects participated in the study, all of whom were employees of Biodynamic Research Corporation. Volunteers were selected to obtain a representative sample of the general population, and included twelve (12) males and eight (8) females spanning a wide range of ages (26 – 58 yrs, mean 44 yrs), heights (150 – 191 cm, mean 172 cm), and weights (54 – 99 kg, mean 80 kg). Plain film lateral lumbar, cervical, and head x-rays were taken of each subject to rule out significant pathology. Written consent was obtained from each participant and the study protocol was approved for human use by both an internal Research Review Board and an external Institutional Review Board (IRB).

Volunteers were instrumented with tri-axial accelerometers (Endevco 7596, 30 g) attached to custom-fit mounts that were securely strapped around the lumbar and upper thoracic regions. Bite blocks made from dental impressions of each volunteer were instrumented with two accelerometers (Endevco 7265-HS, 20 g) and one angular rate sensor. Initially, a magneto-hydrodynamic (MHD) (ATA ARS-01) angular rate sensor was mounted to the bite block and positioned outside the mouth. This arrangement caused unacceptable vibrations in the angular rate signal in initial tests involving a head impact, so the MHD was replaced with a smaller angular rate sensor (DTS ARS1500k) that was rigidly mounted to the bite block and fit inside the mouth (Fig. 1) and the tests were repeated. For hygienic reasons, the instrumented bite block was wrapped in a thin plastic bag before being placed in the subjects' mouths. Instrumentation data were collected using a TDAS-PRO 32-channel rack (DTS) at 10 kHz. All instrumentation data were digitally filtered to CFC180. The sagittal plane bending moment measured by the upper neck load cell of the Hybrid III dummy was transformed to the occipital condyles. High speed digital video (Phantom v7.1, Vision Research, Inc.) was collected at 3 kHz. Video data were analyzed by tracking various points using WINanalyze Tracking 3D Software (Mikromak, Inc.).

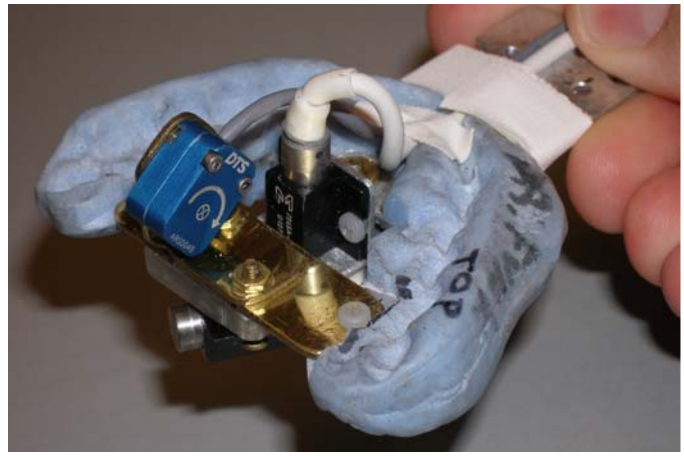

Figure 1. Photograph of an instrumented bite block.

The volunteers were subjected to a variety of test scenarios, some of which included a head impact and some of which did not [7]. All scenarios involved loading in the sagittal plane without significant off-axis components. The current paper focuses on the soccer ball head impact scenario. Participants stood near a custom-made apparatus that was aligned to shoot a regulation adult size 5 soccer ball inflated to 55 kPa horizontally such that it struck the subjects in the forehead. Up to four tests were conducted on each subject at increasing ball speeds: one low speed impact at 5 m/s, and three moderate speed impacts at 8.5, 10, and 11.5 m/s. The participants remained stationary and did not attempt to actively head the soccer ball. In fact, the ball was released and struck the subjects' foreheads before they had a chance to react. Testing began at the lowest speed and the test speed was progressively increased. After each test, participants were asked if they wished to continue to the next higher speed. Subjects reported any symptoms related to testing to a physician (CEB). Testing was also conducted in as identical a manner as possible on a seated Hybrid III 50th percentile male dummy. Three tests were conducted at each of the four test speeds. Accelerometers were mounted at the head CG as well as at a position approximating the inside of the mouth (Fig. 2).

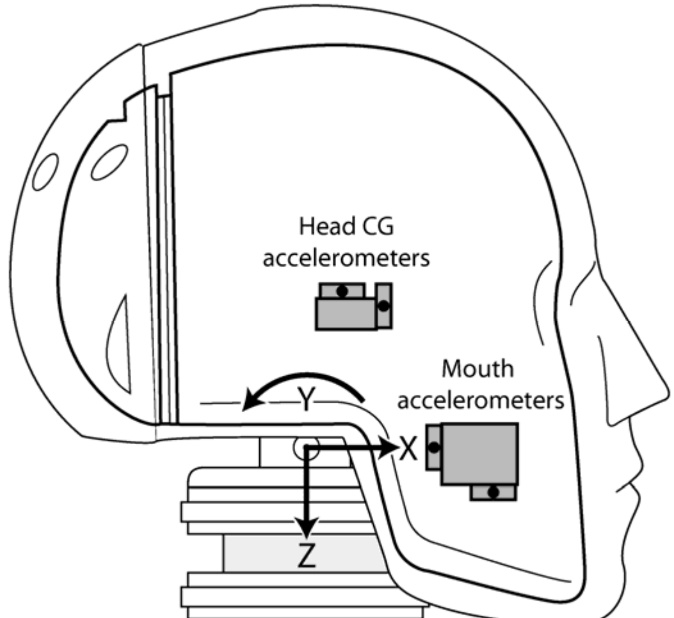

Figure 2. Accelerometer locations in the Hybrid III dummy head.

Accelerations measured at the bite block ($\mathbf{a}_{BB}$) were transformed to the center of gravity of the head ($\mathbf{a}_{CG}$) using standard rigid body dynamics equations:

$$\vec{a}_{CG} = \vec{a}_{BB} + \vec{\alpha} \times \vec{d} - \vec{\omega} \times \left(\vec{\omega} \times \vec{d}\right) \qquad (1)$$

where α was angular acceleration, ω was angular velocity, and **d** was the distance from the bite block accelerometers to the head CG. In order to obtain angular acceleration, the angular rate signal was filtered to CFC180, differentiated, then filtered to CFC180 again. The location of the center of gravity of the head, the bite block accelerometers, and various anatomical landmarks were measured precisely on scaled images in which lateral head x-rays and photographs of each volunteer were superimposed (Fig. 3). The coordinate system of the head was defined using the top of the external auditory meatus (in x-rays) or the tragus (in photographs) as the origin. The x-axis pointed anteriorly through the inferior orbital rim along the Frankfort plane. The z-axis pointed inferiorly, perpendicular to the Frankfort plane, and the y-axis pointed to the right, in accordance with the SAE sign convention. The location of the center of gravity of the head in all volunteers was assumed to be 8.4 mm anterior and 31 mm superior to the origin of the head coordinate system, which is the mean value for a 50th percentile male [9]. The bite block accelerometers were oriented at right angles in the sagittal plane, with the two accelerometers approximating the x- and z-axes of the head coordinate frame. The downward angle of the bite block x-axis relative to the x-axis of the head frame (θ_{BB}) was defined as positive, and averaged 14 degrees in the volunteers. The bite block accelerometers were located 34 – 56 mm (mean 45 mm) anterior and 79 – 96 mm (mean 87 mm) inferior to the head CG in the volunteers, and 51 mm anterior and 48 mm inferior to the head CG in the Hybrid III dummy.

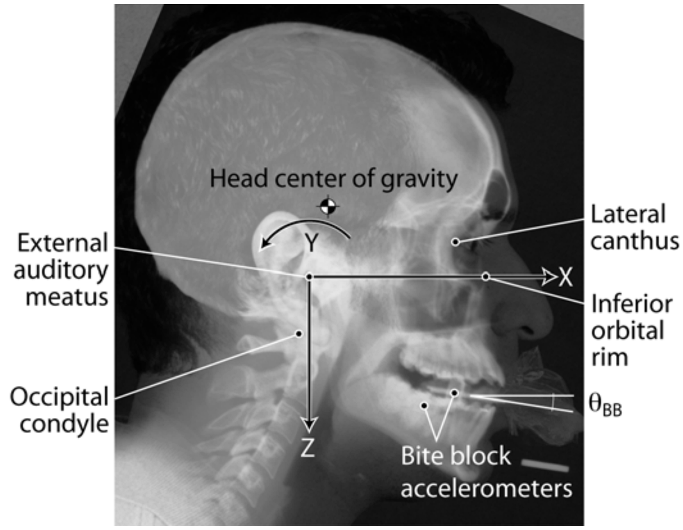

Figure 3. Example of x-ray and photograph overlay used to locate anatomical landmarks in the head coordinate frame.

Transforming bite block accelerations to the center of gravity of the head required several steps. First, the bite block acceleration and angular rate signals were debiased based on the pre-impact readings to remove the effects of signal drift and gravity. Next, the accelerations measured at the bite block (a_{BBx} and a_{BBz}) were transformed to the center of gravity of the head while still in the rotated bite block reference frame (a_{CGxBB} and a_{CGzBB}) according to equation (1):

$$\begin{bmatrix} a_{CGx} \\ a_{CGz} \end{bmatrix}_{BB} = \begin{bmatrix} a_{BBx} \\ a_{BBz} \end{bmatrix} + \begin{bmatrix} -d_z^{BBx} \\ d_x^{BBz} \end{bmatrix} \alpha + \begin{bmatrix} d_x^{BBx} \\ d_z^{BBz} \end{bmatrix} \omega^2 \qquad (2)$$

where the d terms refer to the distances between the bite block x- or z-axis accelerometers (BBx or BBz in the superscript) and the center of gravity of the head in the x or z directions in the *bite block* reference frame (x or z in the subscript). These distances were calculated based on the coordinates of the center of gravity of the head (x_{CG}, z_{CG}) and the bite block x-axis (x_{BBx}, z_{BBx}) or z-axis (x_{BBz}, z_{BBz}) accelerometers in the *head* reference frame:

$$d_z^{BBx} = \left(z_{BBx} - z_{CG}\right)\cos\theta_{BB} - \left(x_{BBx} - x_{CG}\right)\sin\theta_{BB} \qquad (3a)$$

$$d_x^{BBx} = \left(z_{BBx} - z_{CG}\right)\sin\theta_{BB} + \left(x_{BBx} - x_{CG}\right)\cos\theta_{BB} \qquad (3b)$$

$$d_x^{BBz} = \left(z_{BBz} - z_{CG}\right)\sin\theta_{BB} + \left(x_{BBz} - x_{CG}\right)\cos\theta_{BB} \qquad (3c)$$

$$d_z^{BBz} = \left(z_{BBz} - z_{CG}\right)\cos\theta_{BB} - \left(x_{BBz} - x_{CG}\right)\sin\theta_{BB} \qquad (3d)$$

The accelerations at the center of gravity of the head were then transformed from the bite block frame to the head frame:

$$\begin{bmatrix} a_{CGx} \\ a_{CGz} \end{bmatrix}_{head} = \begin{bmatrix} \cos\theta_{BB} & -\sin\theta_{BB} \\ \sin\theta_{BB} & \cos\theta_{BB} \end{bmatrix} \begin{bmatrix} a_{CGx} \\ a_{CGz} \end{bmatrix}_{BB} \qquad (4)$$

Finally, the effect of gravity was added back in to the head accelerations to implicitly account for the mass of the head during the subsequent neck load calculations:

$$\begin{bmatrix} a_{CGx} \\ a_{CGz} \end{bmatrix}_{g} = \begin{bmatrix} a_{CGx} \\ a_{CGz} \end{bmatrix} + 1\,g \begin{bmatrix} \sin\theta_{head} \\ -\cos\theta_{head} \end{bmatrix} \qquad (5)$$

where θ_{head} was defined as positive when the x-axis of the head frame was angled upward relative to the horizontal (Figure 4).

The loads and bending moment in the sagittal plane at the occipital condyles were calculated using standard equations of dynamic equilibrium. The force applied to the head by the soccer ball was determined by analysis of the high speed video. The displacement of the center of the ball at each time point was first measured in the earth-based video reference frame. The displacement data were differentiated and filtered to 100 Hz (CFC60) to calculate the velocity of the ball, then differentiated again (but not filtered) to calculate the acceleration of the ball. The appropriate cutoff frequency for filtering was determined subjectively through trial and error. 1 g was added to the calculated vertical acceleration of the ball in order to implicitly account for the mass of the ball in the subsequent ball force calculations. The acceleration of the ball was multiplied by its mass (434 g) to obtain the force applied to the ball in the earth-based video reference frame. By Newton's Third Law, this was also the force applied to the head. The resultant ball force was assumed to be directed through a point at the center of the contact patch between the ball and the forehead, which was also determined from video analysis (Fig. 4).

The ball force vector was then located in terms of the head coordinate frame. The origin of the head reference frame was obtained by tracking the external auditory meatus (EAM) on video. The orientation of the head frame (θ_{head}) was obtained by tracking the lateral canthus on video, which had a known relationship to the Frankfort plane that defined the x-axis (Fig. 3). The ball force vector (F_{ball}) was transformed to the moving head coordinate frame at each time point:

$$\begin{bmatrix} F_{ball\,x} \\ F_{ball\,z} \end{bmatrix}_{head} = -m_{ball} \begin{bmatrix} \cos\theta_{head} & \sin\theta_{head} \\ \sin\theta_{head} & -\cos\theta_{head} \end{bmatrix} \begin{bmatrix} a_{ball\,x} \\ a_{ball\,z} + 1\,g \end{bmatrix}_{video}$$
(6)

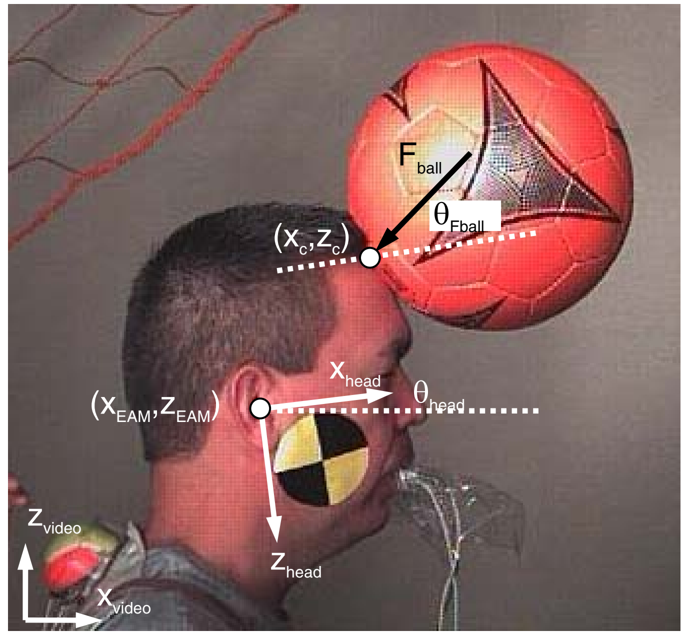

Figure 4. Video capture of a soccer ball impact.

The resultant ball force (F_{ball}) was then calculated:

$$F_{ball} = \sqrt{F_{ball\,x}^2 + F_{ball\,z}^2} \qquad (7)$$

The angle of the ball force vector (θ_{Fball}) relative to the head coordinate frame was defined to be positive when the ball force vector had a downward component:

$$\theta_{Fball} = \tan^{-1}\left(\frac{-F_{ball\,x}}{F_{ball\,z}}\right) \qquad (8)$$

The location of the ball contact point (x_c, z_c) on the forehead was likewise transformed from the video reference frame to the head reference frame:

$$\begin{bmatrix} x_c \\ z_c \end{bmatrix}_{head} = \begin{bmatrix} \cos\theta_{head} & \sin\theta_{head} \\ \sin\theta_{head} & -\cos\theta_{head} \end{bmatrix} \begin{bmatrix} x_c - x_{EAM} \\ z_c - z_{EAM} \end{bmatrix}_{video} \qquad (9)$$

The perpendicular distance between the ball force vector and the center of gravity of the head was determined geometrically at each time point:

$$r_{CG} = (z_{CG} - z_C)\cos\theta_{Fball} + (x_{CG} - x_C)\sin\theta_{Fball} \qquad (10)$$

where all coordinates were in the head frame. Having determined the accelerations at the center of gravity of the head and the force applied to the head by the soccer ball force, it was then possible to calculate the upper neck forces at the occipital condyles using standard equations of dynamic equilibrium:

$$F_{OCx} = m_{head} a_{CGx} - F_{ball\,x} \quad (11)$$

$$F_{OCz} = m_{head} a_{CGz} - F_{ball\,z} \quad (12)$$

$$M_{OCy} = I_{head}\alpha - F_{ball} r_{CG} - F_{OCx}(z_{OC} - z_{CG}) - F_{OCz}(x_{CG} - x_{OC}) \quad (13)$$

where all coordinates were in the head frame. Positive F_{OCx} represented rearward force on the head relative to the body, positive F_{OCz} represented neck tension, and positive M_{OCy} represented a neck flexion moment, in accordance with the SAE sign convention. The location of the occipital condyles in the head coordinate frame (x_{OC}, z_{OC}) was identified radiographically for each volunteer, and ranged from 8 – 22 mm (mean 13 mm) posterior and 49 – 66 mm (mean 58 mm) inferior to the estimated location of the head center of gravity. The location of the head CG in the Hybrid III dummy was measured directly and found to be 22 mm anterior and 51 mm superior to the occipital condyles. The head mass (m_{head}) of each volunteer was estimated using the regression equations of Clauser et al. [10], and ranged from 4.3 – 5.5 kg (mean 5.0 kg). The mass of the Hybrid III dummy head was measured to be 4.4 kg. The sagittal plane moment of inertia for the head (I_{head}) was estimated from the head mass using the data from Beier et al.[11], and ranged from 217 – 327 kg-cm^2 (mean 279 kg-cm^2). The sagittal plane moment of inertia of the Hybrid III dummy head was measured directly and found to be 204 kg-cm^2.

When calculating upper neck loads, it was observed that the addition of large and opposing inertial and ball force terms created high-frequency noise that had a relatively high amplitude, especially in the case of the occipital condyle bending moment. This noise was removed by digitally filtering the occipital condyle forces and moment calculated from inverse dynamics to 50 Hz (CFC30). The appropriate cutoff frequency for filtering was determined by matching the occipital condyle loads and moment calculated by inverse dynamics in the Hybrid III dummy tests to the actual occipital condyle loads and moment measured by the upper neck load cell in the same test. The biofidelity of the Hybrid III head and neck was evaluated by comparing the time histories and peak magnitudes of the head accelerations and neck loads to the results of comparable human subject tests.

A sensitivity analysis was conducted to investigate whether potential errors in the estimated mass, moment of inertia, and location of the center of gravity of the head would significantly affect the results. Errors of up to 10% in the estimated mass and moment of inertia of the head were investigated in combination with errors of up to 10 mm in the estimated location of the center of gravity of the head. Moment of inertia was assumed to vary proportionately with head mass. These error ranges were chosen for convenience and because they encompassed a large majority of the cadaver data [9-11]. All possible combinations of these variations were evaluated for the 11.5 m/s tests to determine the worst-case potential errors in the calculated head accelerations and neck loads.

RESULTS

64 tests were conducted on the 20 human volunteers and 12 tests were conducted using the Hybrid III dummy. The soccer ball approached the subjects' foreheads at a nearly horizontal angle, contacted the subjects' heads for a duration of about 15 – 20 ms, then rebounded off the subjects' heads in a roughly vertical direction. The head moved very little during the brief contact with the soccer ball (~ 3 mm translation and ~ 3° rotation). The soccer ball impacts were generally well-tolerated, with most of volunteers reporting no symptoms. Seven people did report symptoms, primarily minor headaches, all of which lasted less than a day [7].

The methodology for transforming accelerations measured remotely at the bite block to the center of gravity of the head was validated through testing on the Hybrid III dummy. Accelerations were measured in the mouth area of the dummy at approximately the location where the bite block would be in a human (Fig. 2). The mouth area of the dummy experienced an upward and rearward acceleration. The translational and rotational accelerations all peaked at about the same time (Fig. 5). When these accelerations were transformed to the center of gravity of the head (eq. 2), the magnitude and shape of the acceleration traces changed dramatically (Fig. 6). The large difference between the acceleration at the mouth and the center of gravity of the head was due almost entirely to the substantial angular acceleration of the head. The centripetal acceleration term ($\mathbf{d}\omega^2$) in eq. (2) was negligible. The validity of the transformation process was confirmed by the fact that in every test, the time histories of the transformed mouth accelerations were almost identical to the accelerometer readings at the center of gravity of the head (Fig. 6).

The initial human tests with the MHD mounted outside the mouth produced unacceptable ringing in the angular rate signal (Fig. 7). The effect of the ringing was greatly magnified when the angular rate signal was differentiated, so much so that the peak angular acceleration was three times higher when calculated from a sensor outside the mouth compared to a sensor inside the mouth (Fig. 8). The shape of the angular acceleration pulse when calculated from a sensor inside the mouth in a human generally matched the results from the Hybrid III dummy tests (Figs. 5 and 8). Likewise, the linear bite block acceleration pulses showed a bimodal shape in the human tests (Fig. 9), similar to the results seen in the Hybrid III tests (Fig 5). When the bite block accelerations in the human tests were transformed to the center of gravity of the head, their shape changed from biphasic to monophasic (Fig. 10), which was the expected result and also one that matched the results of the Hybrid III tests (Figs. 5 and 6).

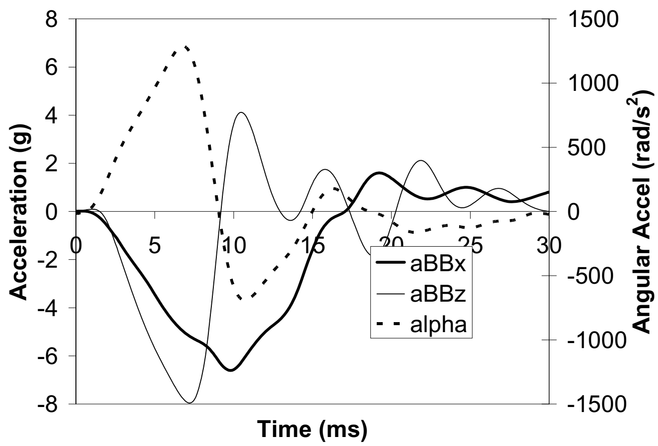

Figure 5. Sensor readings from the mouth area simulating a bite block in a 8.5 m/s Hybrid III dummy test.

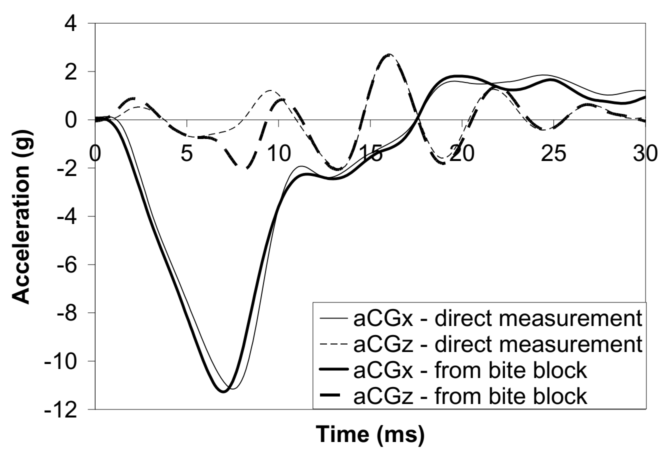

Figure 6. Comparison of measured and calculated head CG accelerations in the Hybrid III test shown in Fig. 5.

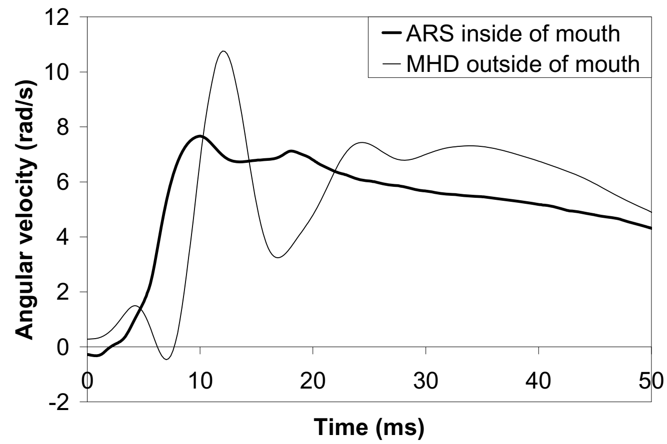

Figure 7. Head angular velocity measured in two similar 11.5 m/s soccer ball head impacts on the lead author (JRF).

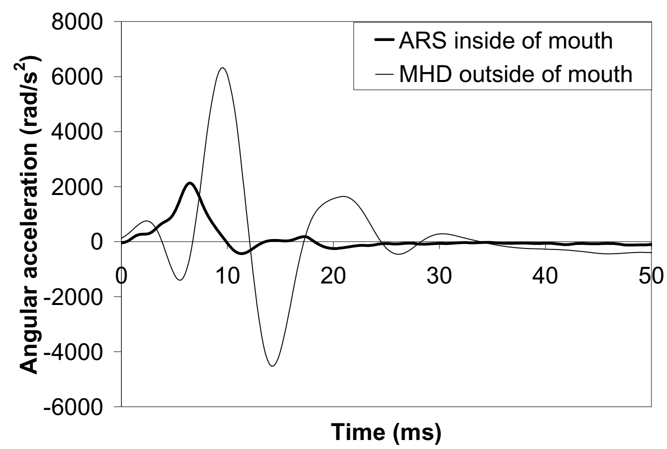

Figure 8. Head angular acceleration measured in the two tests shown in Fig. 7.

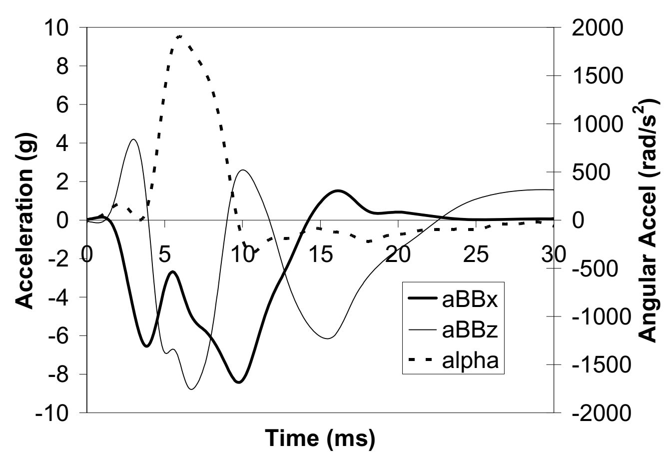

Figure 9. Sensor readings from the bite block in an 11.5 m/s human test.

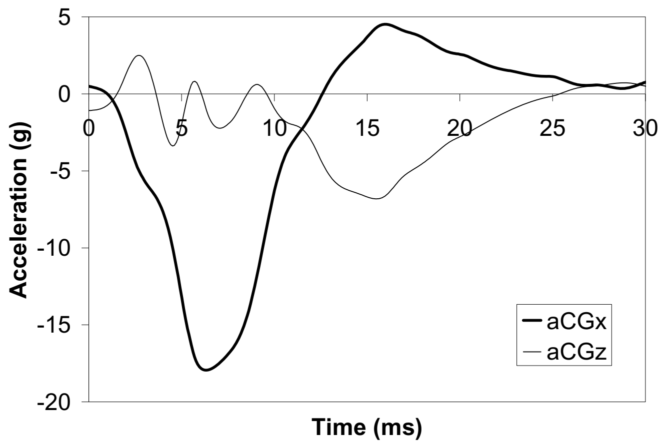

Figure 10. Accelerations at the head center of gravity transformed from the bite block readings shown in Fig. 9.

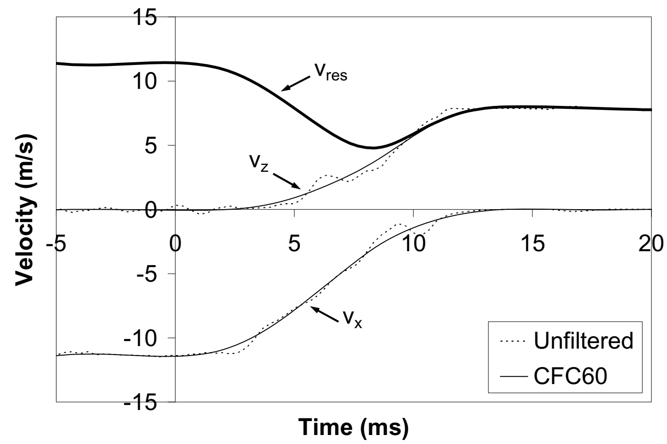

Figure 11. Soccer ball velocity components in the video frame (Fig. 4) for the test shown in Figs. 9 and 10.

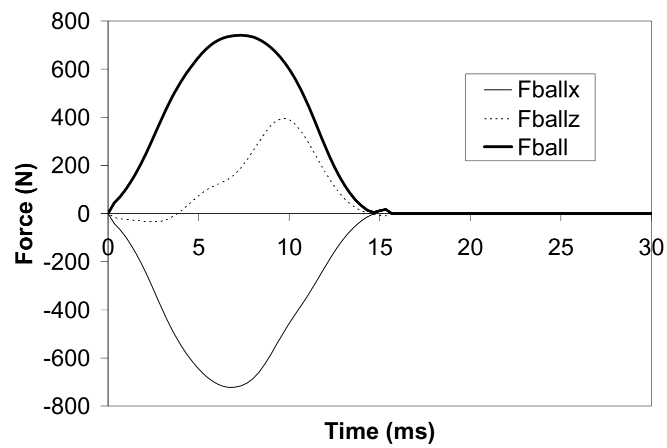

Figure 12. Soccer ball force components obtained by differentiating the data in Fig. 11.

In order to calculate neck loads, the force exerted on the head by the soccer ball had to be determined. The force was calculated from the ball acceleration, which was determined by analysis of the high speed video. The video was sampled at 3000 frames per second, which yielded approximately 50 frames in which the soccer ball was visibly in contact with the head in each test. The displacement of the center of the ball was also tracked for approximately 50 frames before and after the head contact in order to determine the speed and trajectory of the ball as it approached and departed the head. Differentiating the ball displacement data produced good-quality ball velocity traces (Fig. 11). However, the differentiation process amplified the high frequency noise too much for the curves to be suitable for a second differentiation without first filtering them. Therefore, the ball velocity curves were filtered to 100 Hz (CFC60), which effectively removed the noise without altering the essence of the signal. The ball acceleration curves obtained by differentiating the filtered velocity traces contained no high-frequency noise, and therefore did not require any additional filtering. The ball force was then obtained by multiplying the ball acceleration by the ball mass (434 g) and transforming to the head frame (eq. 6) (Fig. 12).

When calculating upper neck loads, it was observed that the ball force and head inertial terms in the dynamic equilibrium equations (eqs. 11 – 13) were large and opposing. The calculated neck force or moment was small compared to these terms, and the addition of these large and opposing terms created high-frequency noise that had a fairly high amplitude relative to the sum (Figs. 13 and 14). This problem was particularly severe in the calculation of the occipital condyle bending moment (Fig. 15). The noise in the signal was not felt to represent a realistic neck response. Therefore, it was removed by digitally filtering the occipital condyle forces and moment calculated from inverse dynamics to 50 Hz (CFC30). This level of filtering dramatically reduced the peak values of the calculated neck forces (Fig. 16). However, after filtering, the calculated occipital condyle loads and moment matched the upper neck load cell readings in the Hybrid III dummy tests surprising well both in terms of peak magnitude (Fig. 16) and pulse shape (Fig. 17).

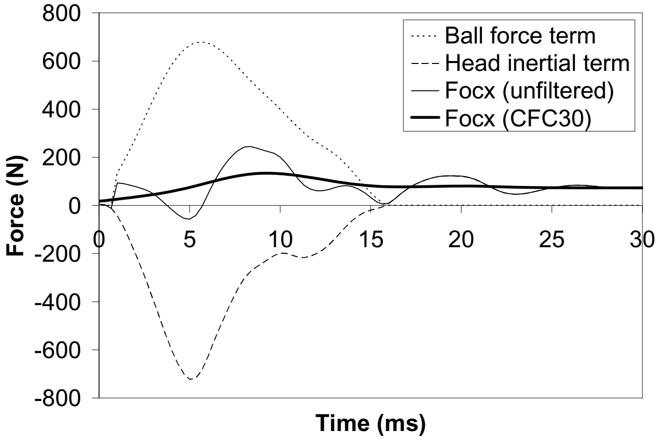

Fig. 13. Calculation of the occipital condyle shear force (Focx) in an 11.5 m/s test on the Hybrid III dummy.

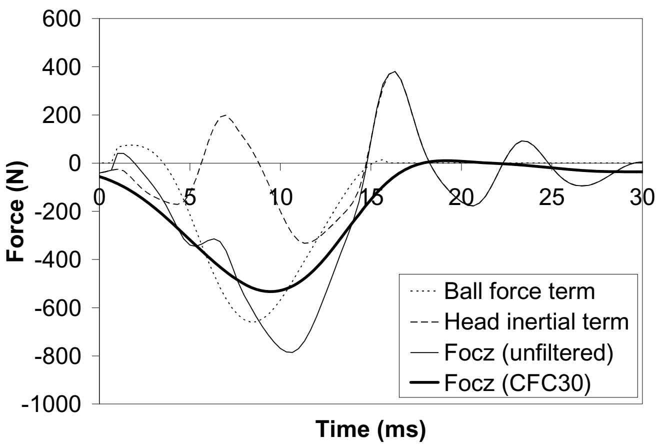

Fig. 14. Calculation of the occipital condyle axial force (Focz) in the 11.5 m/s Hybrid III test shown in Fig. 13.

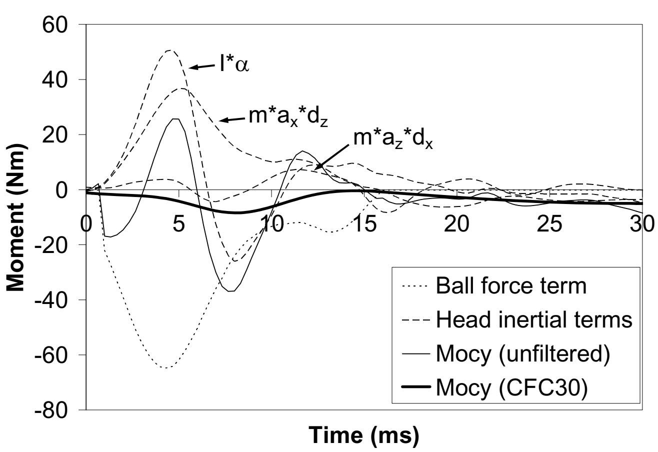

Fig. 15. Calculation of the occipital condyle bending moment (Mocy) in the 11.5 m/s test on the Hybrid III shown in Figs. 13 and 14.

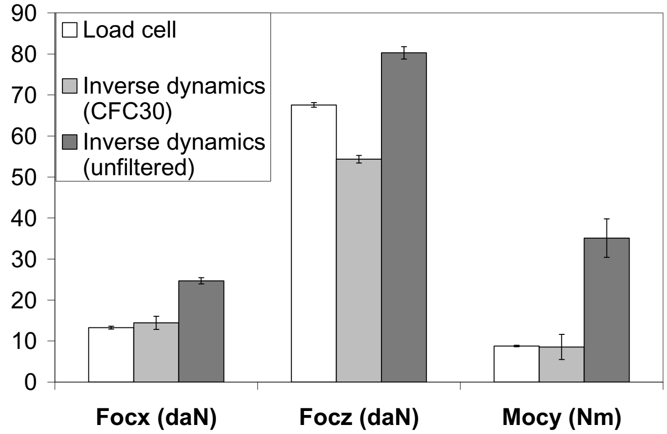

Fig. 16. Comparison of measured and calculated peak neck loads in the 11.5 m/s tests on the Hybrid III dummy. Mean absolute peak values are shown ± 1 s.d.

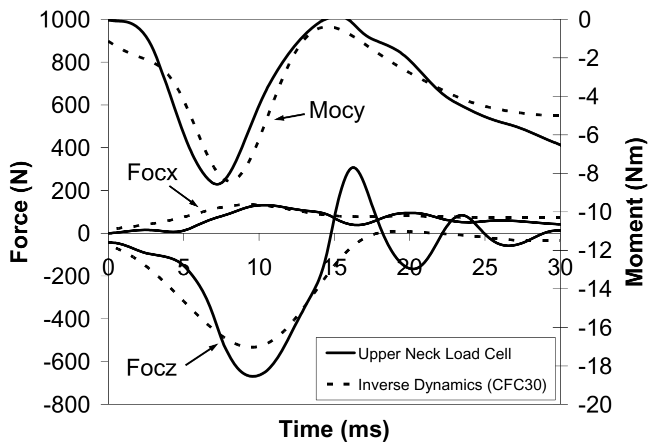

Fig. 17. Comparison of measured and calculated neck loads and moment in the 11.5 m/s test on the Hybrid III shown in Figs. 13 – 15.

Equivalent soccer ball impacts to the forehead generally produced similar peak head accelerations and neck loads in the Hybrid III dummy and the human volunteers (Figure 18). Due to the Hybrid III's stiffer neck, it experienced slightly lower peak head accelerations than the human volunteers, and much higher compressive neck loading. The shape of curves was also different in the Hybrid III and humans, with the peak forces and accelerations typically occurring earlier in the Hybrid III. The amount of time the soccer ball was in contact with the head was about the same in the Hybrid III (~15 ms) and human volunteers (~17 ms). The shape of the Hybrid III head was apparently similar to the humans, because the departure angle of the soccer ball as it rebounded off the head was similar in both groups (~10° rearward of vertical). However, the soccer ball experienced more restitution as it caromed off the head of the Hybrid III, with a departure speed of just under 80% of the approach speed in the Hybrid III and just under 70% in the humans.

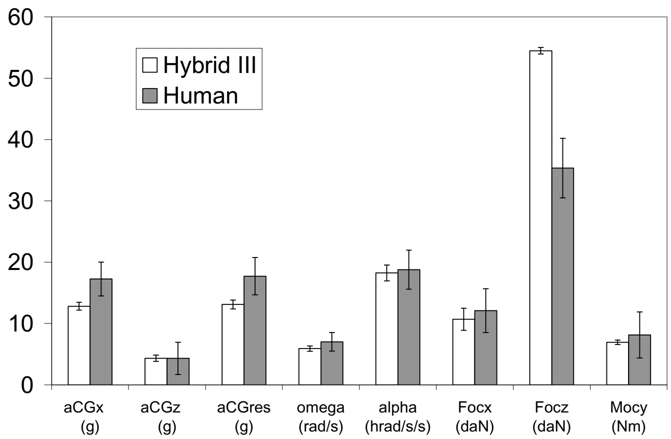

Fig. 18. Comparison of human and Hybrid III head and neck responses to soccer ball head impacts at 10 m/s. Mean absolute peak values are shown ± 1 s.d.

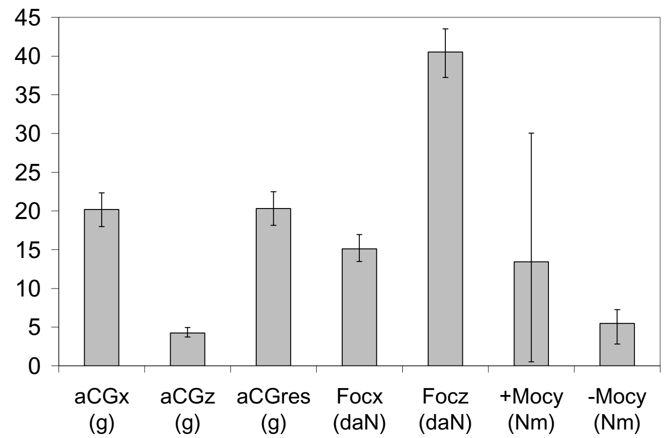

Fig. 19. Typical sensitivity analysis results for a sample 11.5 m/s test with error bars depicting ranges for worst-case combinations of varying the head mass, moment of inertia, and center of gravity location.

The results of the sensitivity analysis show that worst-case combinations of errors of 10 mm in the estimated head CG location and 10% in the estimated head mass and moment of inertia generally result in small (~10%) errors in the calculated head accelerations and neck loads (Figure 19). The notable exception is in the calculated neck moment (Mocy), which can be subject to large errors.

DISCUSSION

The first step in calculating head center of gravity accelerations or neck loads is to obtain accurate measurements of head accelerations at a remote location. In this study, we chose to directly measure linear accelerations and angular rate inside the mouth. Other techniques involving different instrumentation mounted at different locations on the head are theoretically possible. Attaining a rigid connection between the instrumentation and a skull is a challenge in tests on live humans. In cadaver tests, this problem can be overcome by screwing the instrumentation directly into the skull. We found that mounting sensors to a custom-fit bite block made with dental impressions provided a sufficiently rigid connection to the skulls of our volunteers. The most challenging parameter to measure accurately was the angular acceleration of the head. Initial testing with an MHD mounted outside the mouth to the bite block via a cantilevered beam resulted in unacceptable vibrations even after several efforts to rigidify the connection between the sensor and the bite block (Figs. 8 and 9). We were only able to correct this problem using instrumentation technology that first became available in 2006 with the development of an angular rate sensor (DTS ARS1500k) small enough to fit inside the mouth.

It should be noted that in other test scenarios (not reported here) that did not involve a head impact, vibration artifacts were not observed in the angular rate or angular acceleration data obtained from the MHD sensor mounted outside the mouth [7]. The issue of sensor coupling and vibration becomes an acute problem primarily in head impact scenarios, when the frequency content of the head acceleration more readily approaches or exceeds the natural frequency of the sensor mounting system. This is why a soccer ball impact to the forehead was considered a challenging test condition and was chosen to be the subject of analysis in this paper.

The ability to accurately measure the angular acceleration of the head was key to the success of the entire methodology. Accurate angular acceleration data were required to transform the remotely measured head accelerations to the center of gravity of the head, which were in turn required to calculate the neck loads. Furthermore, head acceleration must be transformed to the center of gravity of the head in order to evaluate head injury potential and provide a meaningful comparison to other studies. Accelerations measured remotely on the head, either on a bite block [6] or inside a helmet [12-13], cannot be easily compared between studies, because they may vary substantially and idiosyncratically from the accelerations at the center of gravity of the head, depending on the test condition and the sensor locations. In this study, the accelerations at the center of gravity of the head differed markedly from the accelerations inside the mouth (Figs. 9 and 10). The accuracy of the transformation methodology was confirmed by testing with the Hybrid III dummy (Figs. 5 and 6). Although it was not possible to compare the calculated accelerations at the center of gravity of the head to direct measurements in the human testing, the similarity in the magnitude and shape of the linear and angular acceleration traces in the human and Hybrid III tests indicated that the measurements in the human subjects were accurate. The transformation of the bite block accelerations to the center of gravity of the head changed the shape of the curve from biphasic to the expected monophasic shape with a pulse duration corresponding the time in which the soccer ball was in contact with the head. Therefore, high confidence can be placed in the accuracy of the calculated head center of gravity accelerations.

The methodological difficulties inherent in measuring head accelerations in a live human subjected to a head impact can be observed by comparing the results of this study to two other biomechanical studies involving soccer ball heading [5-6]. Naunheim et al. [5] instrumented volunteers with three triaxial accelerometers mounted to a polyethylene headpiece. The subjects actively headed soccer balls that were launched at 9 and 12 m/s. Although the peak values for linear and angular head accelerations reported in [5] were similar to the current study, the shapes of the curves were very different. Notably, the duration of the acceleration pulse appeared to exceed 50 ms [5]. In the present study, the acceleration pulse duration was 15 – 20 ms, which is consistent with the amount of time the soccer ball was observed to be in contact with the head on high-speed video. Furthermore, the acceleration traces reported in [5] show considerably more oscillation than the acceleration traces reported here, particularly for angular acceleration. These results suggest that the sensors were not rigidly coupled to the head in the study by Naunheim et al. [5].

Shewchenko et al. [6] instrumented volunteers with a bite plate with linear accelerometers and an angular accelerometer located outside the mouth. The subjects actively headed soccer balls that were launched at speeds of 6 and 8 m/s. Bite block accelerations were reported without transforming them to the center of gravity of the head. Shewchenko et al. [6] reported a biphasic linear mouth acceleration pulse lasting approximately 15 – 20 ms, which is consistent with our results. However, the head angular accelerations in [6] showed several oscillations within this pulse duration, which suggests that the sensors were vibrating due to their cantilevered connection to the bite plate. We believe this is the same problem that we observed in our initial testing with the MHD mounted to the bite block outside of the mouth (Figures 8 and 9). Given the high level of noise in the angular acceleration measurement, transforming the bite block accelerations to the center of gravity of the head may have been problematic if it had been attempted.

There were further methodological challenges associated with calculating the sagittal plane loads and moment at the occipital condyles using equations of dynamic equilibrium (eqs. 11 – 13). The first challenge was calculating the force between the soccer ball and the head. This task was accomplished by double differentiating the ball displacement as determined from analysis of the high-speed video. The resolution and sampling rate of the video were sufficient to obtain ball velocity data with minimal noise (Fig. 11). The ball velocity data showed the expected low frequency response (< 50 Hz), and filtering to 100 Hz did not alter the basic shape of the curves. Likewise, the ball force curves calculated from the filtered and differentiated ball velocity data were reasonable and smooth (Fig. 12).

In spite of the fact that both the acceleration curves and the ball force curves appeared reasonable and smooth, the addition of the opposing inertial and ball force terms created high-frequency, high-amplitude noise relative to the calculated neck loads. The fundamental reason for this difficulty was the loose coupling of the head to the body, which meant that the ball force was resisted primarily by the inertia of the head, rather than by force in the neck. As a result, the calculated neck loads were small relative to the inertial and ball force terms. Errors that were small relative to either the inertial or ball force terms became large relative to the calculated neck load.

To eliminate the unwanted noise in the calculated neck loads and moment, the data were filtered to a cutoff frequency of 50 Hz (CFC30), which is far lower than the signal filtering recommendations in SAEJ211. Considering the frequency content of the head and ball accelerations, as well as the compliance of the human neck, it was our judgment that the frequency content of any realistic human neck response in these test conditions would lie almost entirely below 50 Hz. This hypothesis was supported by the finding that after being digitally filtered to a cutoff frequency of 50 Hz (CFC30), the calculated neck loads in the Hybrid III dummy tests matched the upper neck load cell measurements quite well, both in terms of magnitude (Fig. 16) and pulse shape (Fig. 17). Nonetheless, such large changes in peak values and pulse shapes (Figs. 13 – 15) due to filtering to 50 Hz raise concerns, particularly for the calculated bending moment at the occipital condyles (Mocy). Certainly the effect of filtering to 50 Hz dwarfed any potential errors associated with an inaccurate estimate of a subject's head mass, moment of inertia, and center of gravity location (Fig. 19).

The Hybrid III dummy head and neck appeared to match the response of the human head and neck reasonably well in the test condition studied. Peak x-axis head accelerations were somewhat lower in the Hybrid III tests than in comparable human tests (Fig. 18). In addition, there was evidence of a stiffer response from the Hybrid III neck compared to a human neck, particularly in the axial direction. Peak compressive forces in the upper neck (Focz) were notably higher in the Hybrid III than in the human subjects in the same test conditions (Fig. 18), and the signals exhibited oscillations at approximately 125 Hz (Fig. 17). This high frequency ringing was felt to be an artifact of the construction of the Hybrid III neck, which is much stiffer than the human neck in axial compression at low levels of force [14]. It was felt that this vibration was not a biofidelic response that needed to be captured in the human subjects, so the reduction in the peak calculated neck compression values to a level below the peak measured values (Fig. 16) as a result of filtering to 50 Hz (Fig. 17) was not a cause for concern. In all other respects, the Hybrid III response was within the range of responses observed in the human subjects (Fig. 18).

CONCLUSIONS

Accurately calculating head accelerations and neck loads in a biomechanical test of a live human subjected to a head impact presents many methodological challenges. In this study, sagittal plane head accelerations were measured using two accelerometers and an angular rate sensor. By positioning the sensors inside the mouth on a bite block with custom-fit dental impressions, rigid coupling of the sensors to the skull was achieved with minimal vibration of the sensors. Calculating the acceleration of the center of gravity of the head and loads at the occipital condyles required detailed geometric and anthropometric measurements using photographs and x-rays. The force between the head and the soccer ball could be accurately determined from high speed video analysis. Data filtering far below the cutoff frequencies recommended in SAE J211 was required to eliminate unwanted noise, particularly when signals were processed by differentiation or addition. The methods presented here for transforming bite block accelerations to the center of gravity of the head and calculating loads at the occipital condyles were validated by their good agreement with direct measurements in Hybrid III dummy tests. Furthermore, the Hybrid III dummy head and neck were shown to be reasonably biofidelic in this particular test scenario of a horizontal soccer ball impact to the forehead.

REFERENCES

1. Mertz, H.J. and Patrick, L.M., "Investigation of the Kinematics and Kinetics of Whiplash," Paper 670919, *Proc 11th Stapp Car Crash Conf*, pp. 175 – 206, 1967.
2. Ewing, C.L. and Thomas, D.J., "Torque versus Angular Displacement of Human Head to –Gx Impact Acceleration," Paper 730976, *Proc 17th Stapp Car Crash Conf*, pp. 309 – 342, 1973.
3. Ono, K., Kaneoka, K., Wittek, A., Kajzer, J., "Cervical Injury Mechanism Based on the Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics During Low Speed Rear Impacts," Paper 973340, *Proc 41st Stapp Car Crash Conf*, pp. 339 – 356, 1997.
4. Vijayakumar, V., Scher, I., Gloeckner, D.C., Pierce, J., Bove, R., Young, D., Cargill, R., "Head Kinematics and Upper Neck Loading During Simulated Low-Speed Rear-End Collisions: A Comparison with Vigorous Activities of Daily Living," Paper 2006-01-0247, *SAE World Congress*, pp. 49 – 60, 2006.
5. Naunheim, R.S., Bayly, P.V., Standeven, J., Neubauer, J.S., Lewis, L.M., Genin, G.M., "Linear and Angular Head Accelerations during Heading of a Soccer Ball," *Med Sci Sports Exercise*, 35(8): 1406 – 1412, 2003.
6. Schewchenko, N., Withnall, C., Keown, M., Gittens, R., Dvorak, J., "Heading in Football. Part I: Development of Biomechanical Methods to Investigate Head Response," *Br J Sports Med*, 39(Suppl 1): i10 – i25, 2005.
7. Funk, J.R., Cormier, J.M., Bain, C.E., Guzman, H., Bonugli, E., "An Evaluation of Various Neck Injury Criteria in Vigorous Activities," *IRCOBI*, pp. 233 – 248, 2007.
8. Funk, J.R., Cormier, J.M., Bain, C.E., Guzman, H., Bonugli, E., "Relationship Between Linear and Rotational Head Acceleration in Various Activities," *Biomed Sci Instrum*, 44: 207 – 212, 2008.
9. Schneider, L.W., Robbins, D.H., Pflug, M.A., Snyder, R.G. Anthropometry of Motor Vehicle Occupants. Washington, DC, National Highway Traffic Safety Administration, UMTRI-83-53-1, 1983.
10. Clauser, C.E., McConville, J.T., Young, J.W., "Weight, Volume, and Center of Mass of Segments of the Human Body." Wright-patterson AFB, Air Force Systems Command, AMRL-TR-69-70, pp. 1 – 100, 1969.
11. Beier, G., Schuller, E., Schuck, M., Ewing, C.L., Becker, E.D., Thomas, D.J., "Center of Gravity and Moments of Inertia of Human Heads," *IRCOBI*, pp. 218 – 228, 1980.
12. Allen, M.E., Weir-Jones, I., Motiuk, D.R., Flewin, K.R., Goring, R.D., Kobetitch, R., Broadhurst, A., "Acceleration Perturbations of Daily Living. A Comparison to 'Whiplash'," *Spine*, 19(11): 1285 – 1290, 1994.
13. Naunheim, R.S., Standeven, J., Richter, C., Lewis, L.M., "Comparison of Impact Data in Hockey, Football, and Soccer," J Trauma, 48(5): 938 – 941, 2000.
14. Yoganandan, N., Sances, A., Pintar, F., "Biomechanical Evaluation of the Axial Compressive Responses of the Human Cadaveric and Manikin Necks," *J Biomech Eng*, 111: 250 – 255, 1989.

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