Linear and Angular Head Accelerations during Heading of a Soccer Ball

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ABSTRACT

NAUNHEIM, R. S., P. V. BAYLY, J. STANDEVEN, J. S. NEUBAUER, L. M. LEWIS, and G. M. GENIN. Linear and Angular Head Accelerations during Heading of a Soccer Ball. *Med. Sci. Sports Exerc.*, Vol. 35, No. 8, pp. 1406–1412, 2003. **Purpose:** Cognitive deficits observed in professional soccer players may be related to heading of a soccer ball. To assess the severity of a single instance of heading a soccer ball, this study experimentally and theoretically evaluated the linear and angular accelerations experienced by the human head during a frontal heading maneuver. **Methods:** Accelerations were measured using a set of three triaxial accelerometers mounted to the head of each of four adult male subjects. These measurements (nine signals) were used to estimate the linear acceleration of the mass center and the angular acceleration of the head. Results were obtained for ball speeds of 9 and 12 $m \cdot s^{-1}$ (approximately 20 and 26 mph). A simple mathematical model was derived for comparison. **Results:** At 9 $m \cdot s^{-1}$, peak linear acceleration of the head was 158 ± 19 $m \cdot s^{-2}$ (mean $\pm$ standard deviation) and peak angular acceleration was 1302 ± 324 $rad \cdot s^{-2}$; at 12 $m \cdot s^{-1}$, the values were 199 ± 27 $m \cdot s^{-2}$ and 1457 ± 297 $rad \cdot s^{-2}$, respectively. The initial acceleration pulses lasted approximately 25 ms. Measured head accelerations confirmed laboratory headform measurements reported in the literature and fell within the ranges predicted by the theoretical model. **Conclusions:** Linear and angular acceleration levels for a single heading maneuver were well below those thought to be associated with traumatic brain injury, as were computed values of the Gadd Severity Index and the Head Injury Criterion. However, the effect of repeated acceleration at this relatively low level is unknown. **Key Words:** KINEMATIC MEASUREMENTS, KINEMATIC MODELING, BRAIN INJURY, IMPACT MODELING

Heading of the ball is an integral part of the game of soccer. Recent papers by Tysvaer et al. (27), Jordan and Green (14), and Matser et al. (20) have focused attention on the potential for long-term cognitive deficits caused by heading. Longstanding debate continues over whether the long-term cognitive deficits observed in soccer players is due to repeated subconcussive head impact or to occasional concussive head impact due to collisions with elbows and heads (e.g., 2,5–7,12–14,16,20,25–27). Some advocate usage of protective headgear for youth soccer players; however, this headgear has not been proven to reduce cognitive deficits, and several commercial products have recently been shown to be relatively ineffective in reducing the linear acceleration of the head during heading (21).

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When a ball strikes the head, the skull accelerates, causing the brain to stretch and twist. The magnitude of this stretching and twisting is unknown, as are injury thresholds for concussion and for cognitive deficits. What is known is that sufficient skull acceleration can lead to brain injury. Empirical formulations such as the Gadd Severity Index (GSI) and the Head Injury Criterion (HIC) predict when a single uniform, linear acceleration of the head may lead to brain injury (8,22,24). However, the cumulative effects of repeated accelerations remain unknown (e.g., (9)).

Heading of a soccer ball is a complicated process involving a combination of linear and angular acceleration of the skull. Angular acceleration of the head may be far more damaging to the brain than linear acceleration (1,10,23). The fluid-like character of the brain causes it to be nearly incompressible; the brain is more susceptible to injury from the rotational distortion of groups of cells than from cells being forced together or pulled apart. This distortion, called shear deformation, can lead to physiological change (4) and can be amplified at interfaces between substructures in the brain, such as the junctions between the hemispheres and brainstem. Bailey and Gudeman (3) describe shear strain caused by rotation as the most common source of injury in mild brain trauma. Rotation of the brain has been associated with concussion by Gennarelli (10), who found that monkeys wearing cervical collars which prevented the head from rotating were much less likely to receive concussions than were animals whose heads were allowed to rotate freely. Several studies (e.g., 18) have shown that diffuse axonal injury is associated with large angular accelerations.

Are the skull accelerations from heading of a soccer ball sufficient to cause brain injury? A soccer ball weighs roughly 1 lb (0.454 kg) and may travel at over 40 mph (17.8 $m \cdot s^{-1}$) in a typical high school game (17). During the first few milliseconds of a heading maneuver, before the neck muscles stretch significantly, most of the impulse for redirection of the ball accelerates the head; the impulse absorbed by the neck is significant only after the neck has stretched sufficiently. A first step toward understanding potential brain injury due to heading a soccer ball is the characterization of the resulting linear and angular accelerations of the head. This paper describes the linear acceleration of the head's mass center, and the angular acceleration of the head, during heading of a soccer ball by human subjects. Additionally, these measurements are checked by comparison with a mathematical model and by comparison to published laboratory headform measurements.

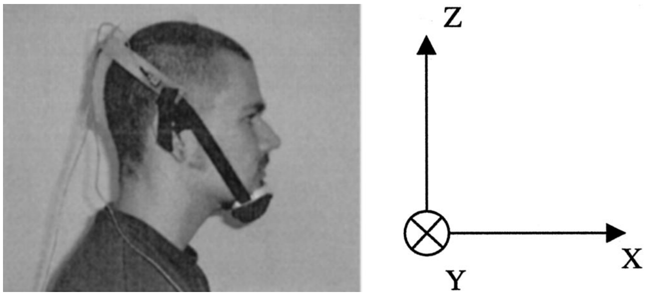

FIGURE 1—Photograph of headgear and accelerometer array. Not shown are the latex rubber bands stretched over the headpiece and under the chin and nose to secure and preload the headpiece. Accelerometers are attached with screws to the headgear, and aligned so that the X-Z plane of the accelerometers is the sagittal plane, and the Y-axis points from the right to left ear. The X-Y-Z coordinate frame used in the study is illustrated on the right; the axes move with the subject's head.

METHODS AND MATERIALS

Subjects

Four subjects performed a series of heading tests while wearing a headpiece instrumented to measure linear and angular accelerations. The subjects were all adult males between 25 and 36 yr old who had played soccer in high school. Before participation, each subject was informed of the study's objectives and methods, and was asked to sign an informed consent form approved by Washington University's Institutional Review Board. Subjects were interviewed regarding soccer experience and concussion history. Their heading technique was viewed by an experienced soccer coach and judged to be typical of competent high school players. No subjects uncomfortable with heading or with history of concussion were included the study.

Protocol

Three triaxial accelerometers (PCB Model 356B11, PCB Piezotronics, Inc., Depew, NY) were mounted at known locations on the fairly rigid polyethylene headpiece (Fig. 1). The accelerometers were aligned in identical directions relative to the headpiece. The headpiece was fitted to the head of each subject, so that the sagittal plane of the subject's head and the X-Z plane of the accelerometer array (refer to Fig. 1) were coincident. Folded high-friction (static friction coefficient of 2.5–3.5) material (Dycem™, Dycem Technologies, Bristol, UK) was used to fill any gaps between the skull and the headpiece. The thickness of Dycem™ was routinely between 0.5 and 3 mm before it was compressed against the skull; this introduces a small compliance between the head and the headpiece, which was minimized as much as possible. Two thick latex rubber bands were then stretched over the headpiece. One band was stretched under the chin and the second over the upper lip to firmly preload the headpiece against the skull. A fourth triaxial accelerometer was glued to a plastic mouthpiece inserted in the subject's mouth. The position of the head's center of mass relative to these triaxial accelerometers was estimated using a three-dimensional video measurement system (Motion Analysis Corp., Santa Rosa, CA).

Standard 450-g soccer balls inflated to 55,300 Pa (8 psi) were projected at the subject from a distance of 6 m (~20 ft) using a mechanical soccer ball driver (Soccer Tutor, Burbank, CA) mounted 1.2 m from the ground, as shown in Figure 2. Each subject headed three balls projected at 9 $m \cdot s^{-1}$ and three balls projected at 12 $m \cdot s^{-1}$ (approximately 20 and 26 mph.) Subjects were instructed to head the ball directly back toward the machine. Signals from the accelerometers were low-pass filtered (800 Hz, Butterworth), digitized (12-bit), and sampled (4800/s) using a National Instruments AT-MIO-64F-5 data acquisition system. Video data were acquired at 240 $frames \cdot s^{-1}$. Data thus consisted of nine acceleration channels (the X-, Y-, and Z-axes of three triaxial accelerometers) and up to 18 channels of position data (X-, Y-, and Z-locations of each of five markers on the head and a marker on the ball).

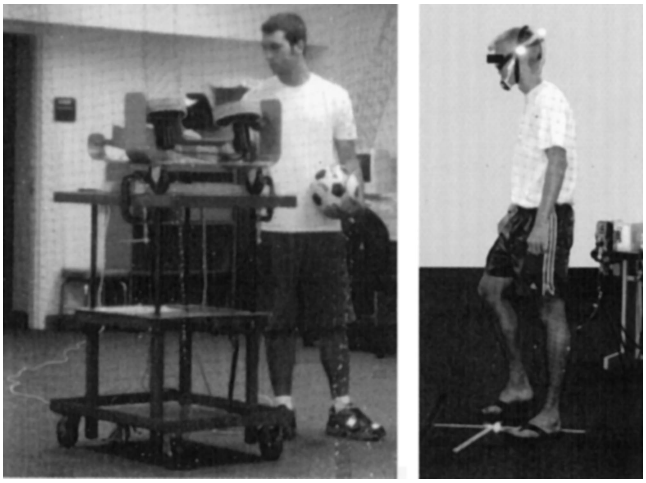

FIGURE 2—Experimental environment. Left, soccer ball pitching machine before shooting a ball. Right, instrumented test subject preparing to head a ball.

Data Analysis

The motion of the headpiece is approximately governed by the well-known three-dimensional kinematic equations for acceleration of a rigid body shown as equations 1a–c (for example, reference 11, p. 759). These equations show that to completely characterize the acceleration of the skull, one must determine: 1) the skull's angular velocity vector, $\bar{\omega}$; 2) the skull's angular acceleration vector, $\bar{\alpha}$; and 3) the acceleration vector of the head's center of mass, $\bar{a}^{CM}$. Each of these vectors consists of three components, one in each of the anterior, left, and superior directions, which are denoted x, y, and z, respectively, in equation 1. Thus, (a_x^{CM}, a_y^{CM}, a_z^{CM}) are, respectively, the anterior, left, and superior accelerations of the head's center of mass; (ω_x, ω_y, ω_z) are rates of spin around axes parallel to the anterior, left, and superior directions, respectively; and (α_x, α_y, α_z) are the rates of change of these spin rates. (a_x, a_y, a_z) are the linear acceleration of a point whose position, measured from the head's center of mass, is (x, y, z).

$$a_x = a_x^{CM} + (\alpha_y z - \alpha_z y) + (\omega_x(\omega_y y + \omega_z z) - (\omega_y^2 + \omega_z^2)x) \quad (1a)$$

$$a_y = a_y^{CM} + (\alpha_z x - \alpha_x z) + (\omega_y(\omega_z z + \omega_x x) - (\omega_x^2 + \omega_z^2)y) \quad (1b)$$

$$a_z = a_z^{CM} + (\alpha_x y - \alpha_y x) + (\omega_z(\omega_x x + \omega_y y) - (\omega_x^2 + \omega_y^2)z) \quad (1c)$$

Because the right hand sides of the equations contain nine unknowns, all nine acceleration measurements from the three triaxial accelerometers (of known position) may be used to estimate the unknown quantities (ω_x, ω_y, ω_z, α_x, α_y, α_z, and a_x^{CM}, a_y^{CM}, a_z^{CM}).

In practice, the three rigid body velocity equations, Equations 2a–c (11), can provide additional information for estimation of angular velocity. Angular velocity data were estimated from application of these equations at each sensor location; the velocity vector at each site (ν_x, ν_y, ν_z) was estimated by integration of the acceleration signals. The unknowns in Equations 2a–c were the components of angular velocity (ω_x, ω_y, ω_z), and the components of the linear velocity of the mass center, (ν_x^{CM}, ν_y^{CM}, ν_z^{CM}). These were estimated by applying Equations 2a–c to all three sensors. The six unknown components of angular acceleration and linear acceleration were then estimated by numerically solving Equations 1a–c simultaneously for all sensors.

$$\nu_x = \nu_x^{CM} + (\omega_y z - \omega_z y) \quad (2a)$$

$$\nu_y = \nu_y^{CM} + (\omega_z x - \omega_x z) \quad (2b)$$

$$\nu_z = \nu_z^{CM} + (\nu_x y - \nu_y x) \quad (2c)$$

A Gauss-Newton method (MATLAB® software) was used to solve all equations. Because the velocity and acceleration equations were over-constrained (nine equations, six unknowns in each case), the equations were solved in a least-squares sense.

Traditionally, severity of head impact has been described by one of two indices: Gadd's Severity Index (GSI) (8,24) or the Head Injury Criterion (HIC) (22,24). These parameters were originally proposed to provide an empirical number that captured both the magnitude and duration T of acceleration during head trauma. The definitions are given as:

$$\text{GSI} = \int_0^T a(t)^{5/2}\, dt \quad (3)$$

and

$$\text{HIC} = \left\{ (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_0^T a(t)\, dt \right]^{5/2} \right\}_{max} \quad (4)$$

where t_1 and t_2 are the bounds (in seconds) of the time interval during which the HIC is maximal. In both definitions, the acceleration is defined in units of "g" (g = 9.8 m·s^{-2}). Note that both HIC and GSI have units of seconds according to these formulas. These indices were both designed to have a value of 1000 for a fatal head injury (24). Although not a focus of the present study, the HIC and GSI are reported to facilitate rough comparison of heading with other events.

A primary concern was ensuring the headgear did not slip relative to the skull during heading. In addition to qualitative visual checks for slippage, two evaluative measurements were made: one using a fourth mouthpiece-mounted triaxial accelerometer, and a second using the three-dimensional video measurement system described above. Measurements from the mouthpiece accelerometer provided a rough gauge to assess the peak values of acceleration. Although uncertainty in the precise location of the mouthpiece accelerometer prevented exact comparison, measurements from the mouthpiece were checked to ensure that they were (1) of the same order as and (2) of lower magnitude than the estimated acceleration of the center of mass. Because the mouthpiece accelerometer was anterior and inferior to the center of mass, a lower acceleration was expected for the combination of backward translation and rotation observed in these tests.

Output from the video measurement system confirmed that the markers on the headgear did not slip measurably relative to markers on the subject's skin after each heading maneuver (within the 5-mm resolution of the motion analysis system.) Also, the subjects did not need to adjust or correct the position of the headgear during the tests. These observations indicate that acceleration data from the sensors mounted on this headpiece can be used to describe the motion of the head accurately.

Experimental Results

Raw acceleration measurements. Raw acceleration signals from the 12 accelerometer channels are shown in Figure 3 for one typical impact. A rise or dip in each of these channels occurs at a time of approximately 200 ms. This rise or dip corresponds to the impact with the soccer ball. The first dip or rise corresponds to acceleration of the head in a particular direction; the subsequent rise or dip corresponds to deceleration of the head in that direction. In each channel, the most significant acceleration of the head

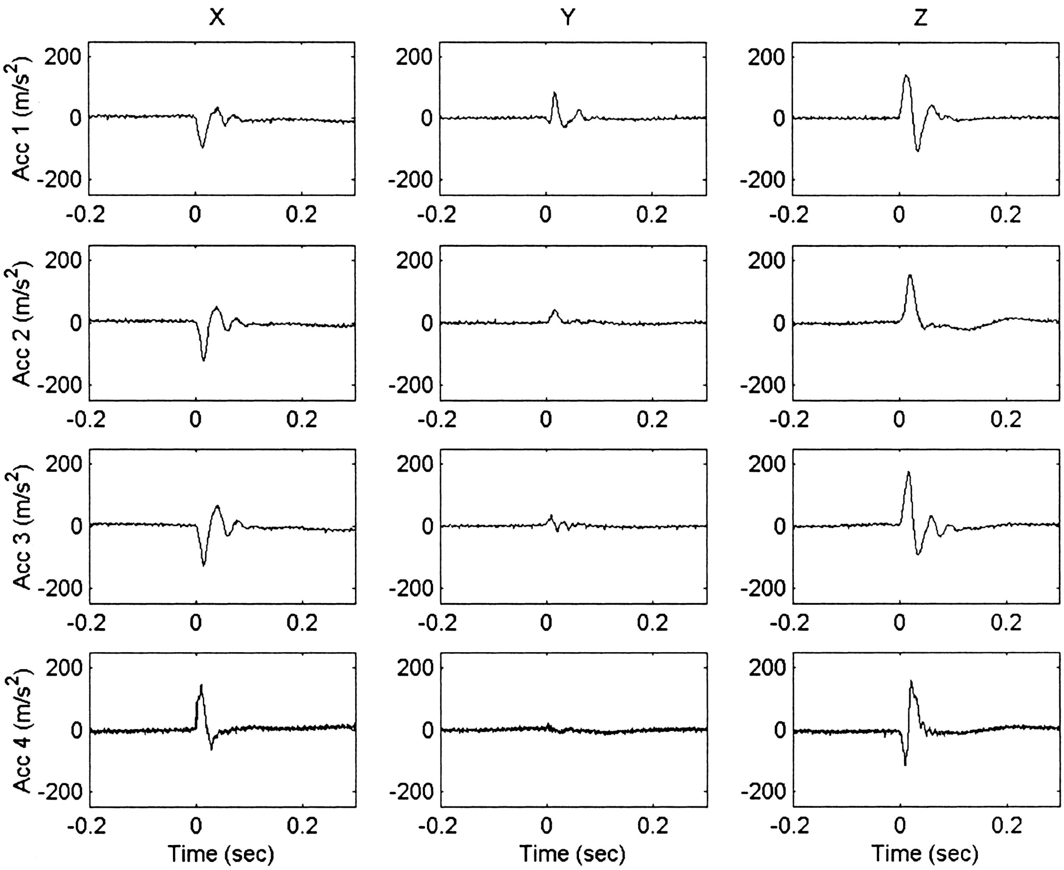

FIGURE 3—Signals from three headpiece accelerometers (Acc. 1–3) and one mouthpiece accelerometer (Acc. 4) during one heading trial. Ball speed was 9 m·s^{-1}; ball pressure was 55,300 Pa.

occurred during the first 15 ms after impact with the soccer ball.

Analyzed data. The time histories of the linear acceleration of the mass center and the angular acceleration of the head are shown in Figures 4 and 5 for one typical heading maneuver. These were computed from the data shown in Figure 3 using equations 1a–c. The first three traces in each of these figures are the components in the anterior, left, and superior (X-, Y-, and Z-) directions; the fourth trace in each figure is the magnitudes, defined as the square root of the sum of the squares of the vectors' components. As was visible from the raw acceleration data, the peak linear acceleration occurs during the impact acceleration pulse, within 15 ms of impact. However, the peak angular acceleration magnitude occurs during the deceleration pulse, 20–40 ms after impact. Mean ($\pm$ SD) linear and angular accelerations for all subjects are listed in Table 1.

Measures of slippage. The two quantitative estimates of slippage were 1) a comparison between the estimate of acceleration at the center of mass to the measurement from the mouthpiece accelerometer, and 2) an estimate of the relative displacement of headgear-mounted optical markers using the video measurement system. Mouthpiece accelerations averaged 15% below center of mass acceleration. A simple calculation shows that part of this deviation is explained by the fact that the mouthpiece acceleration is due to combined linear and angular acceleration of the head. For the tests in which the mean peak linear acceleration was 199 m·s^{-2}, the mean peak angular acceleration of 1457 rad·s^{-2} multiplied by the 0.02- to 0.03-m inferior distance between the center of mass and the mouthpiece yields a deviation of 29–44 m·s^{-2}, or 14.6–22.0%.

No relative motion was observed between the video markers on the headgear and markers on the head. Additionally, visual inspection of the headgear uncovered no observable changes in the position of the headgear on the subjects during the series of tests.

Mathematical Model

Formulae for head acceleration. The measurements were verified to assure that they were of the correct order by using a simple mathematical model. As pictured in Figure 6, the model considered a ball impacting the head, then returning back along its original path.

The specific case of an $M = 4.5$ kg head striking an $m = 0.45$ kg ball at $v = 10$ m·s^{-1} was considered (15). The moment of inertia I for rotation in a sagittal plane was taken as $I = 0.02$ kg m^2, and the impact point was taken to be r $= 0.05$ m above the center of mass of the head. The coefficient of restitution, e, may range from 0 when the maximum possible energy loss occurs during impact, to 1 when no energy loss occurs. A realistic range, based on drop tests of soccer balls in which the ball bounced to 60–70% of

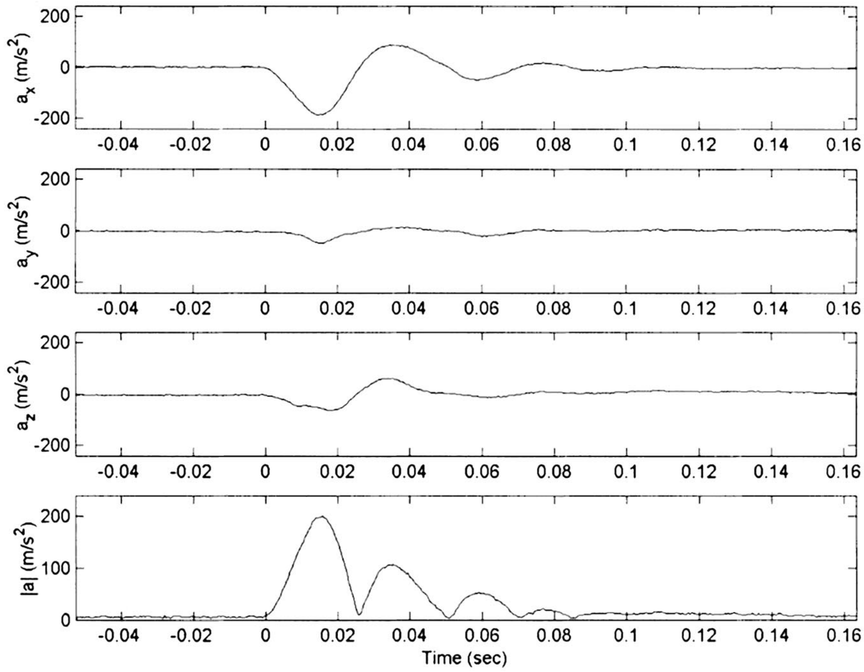

FIGURE 4—Estimates of the components of linear acceleration of the center of mass of the head for the heading trial shown in Figure 3. The time axis is expanded to show the impact. Time is shown relative to the moment of impact.

the drop height, is $0.4 < e < 0.7$. Using the principle of conservation of momentum, one can easily arrive at the formulae for the changes in linear and angular velocity of the head resulting from heading, following procedures outlined in a standard dynamics text (e.g., reference 11, pp. 727–736). Assuming that the impact pulse is sinusoidal and occurs in a time $t_0 = 0.015$ s, one arrives at the following formulae for the linear acceleration a^{CM} of the center of mass of the head and the angular acceleration α of the skull during heading:

$$a^{CM} = \frac{\pi}{2} \frac{(1+e)}{\left(1 + \frac{M}{m} + \frac{Mr^2}{I}\right)} \frac{\nu}{t_0} \tag{5}$$

$$\alpha = \frac{\pi}{2}\left(\frac{Mr^2}{I}\right) \frac{(1+e)}{\left(1 + \frac{M}{m} + \frac{Mr^2}{I}\right)} \frac{\nu}{rt_0} \tag{6}$$

The dimensionless parameters (M/m) and ($M\,r^2/I$) govern the severity of head acceleration received during heading. By inserting the parameters into equations 5 and 6, one arrives at the values of acceleration of the head that are listed in Table 2.

A key approximation in this model is that the neck plays a minor role in the peak acceleration of the head during heading. The justification for this is that the muscles in the neck apply forces that are expected to be small compared with the approxi-

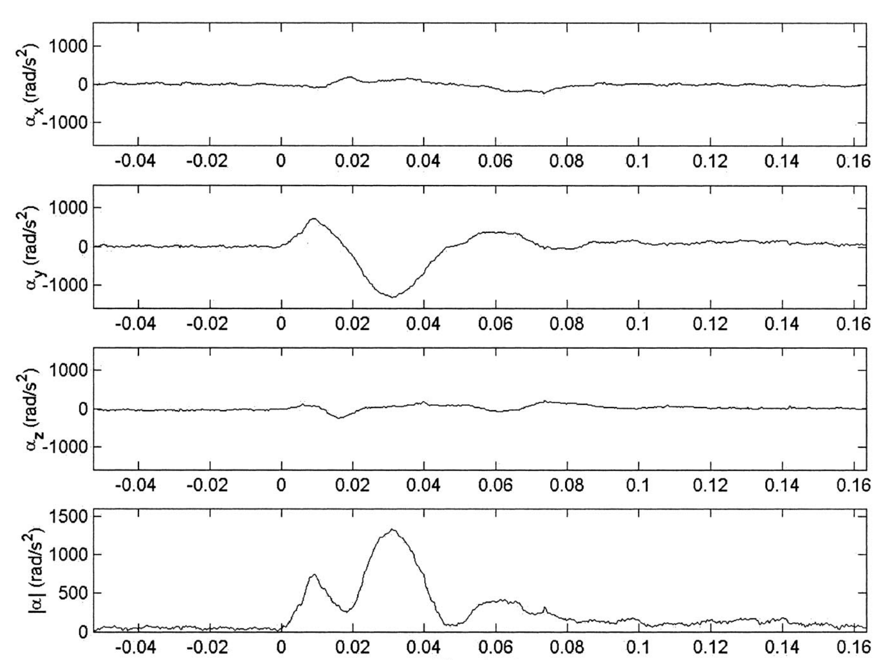

FIGURE 5—Estimates of the components of angular acceleration of the head for the heading trial shown in Figure 3. Time is shown relative to the moment of impact.

TABLE 1. Measured mean (± SD) linear and angular accelerations of the head during soccer heading.

Ball Speed	Linear Acceleration of Center of Mass ($m \cdot s^{-2}$)	Angular Acceleration of Skull ($rad \cdot s^{-2}$)
9 $m \cdot s^{-1}$ (~20 mph)	158 ± 19	1302 ± 324
12 $m \cdot s^{-1}$ (~26 mph)	199 ± 27	1457 ± 297

TABLE 2. Theoretically predicted linear and angular accelerations of the head during soccer heading, for restitution coefficients $0.4 < e < 0.7$.

Ball Speed	Linear Acceleration of Center of Mass ($m \cdot s^{-2}$)	Angular Acceleration of Skull ($rad \cdot s^{-2}$)
9 $m \cdot s^{-1}$ (~20 mph)	$114 \le a^{CM} \le 139$	$1284 \le \alpha \le 1559$
12 $m \cdot s^{-1}$ (~26 mph)	$153 \le a^{CM} \le 185$	$1722 \le \alpha \le 2091$

mately 400–1000 N (~100–250 lb) impact forces, especially given the very short impact duration (15). This approximation appears to be supported by relatively good agreement between predicted and measured accelerations.

DISCUSSION

The head of a soccer player is subjected to linear accelerations of approximately 15–20 g ($g = 9.8\ m \cdot s^{-2}$), as well as to angular accelerations of 1000–2000 $rad \cdot s^{-2}$, during heading of a soccer ball traveling at 9–12 $m \cdot s^{-1}$. Accelerations seen in actual play will vary much more depending on the direction of the ball, its spin, and the location of impact. Although defenders sometimes head a long clearing pass back along its flight path, the majority of heading events involve redirection of the ball at an angle. Although the threshold for mild traumatic brain injury (concussion) is not well established, these values are thought to be well below those required for an acute traumatic brain injury (19). The cumulative effect of repeated subconcussive threshold acceleration on the brain is unknown and may be an important factor in long-term cognitive impairment.

Approximating the initial linear acceleration pulses as sinusoidal and following the procedures outlined by Onusic (24), values for the empirical GSI and HIC measures may be found from the experimental data. For the case of a ball speed of 9 $m \cdot s^{-1}$, the average heading maneuver resulted in HIC = 10 s and GSI = 12 s; for a ball speed of 12 $m \cdot s^{-1}$, the average maneuver resulted in HIC = 18 and GSI = 21. As a reference value, head accelerations yielding HIC and GSI scores of approximately 1000 s are considered capable of causing death (24). However, the standard HIC and GSI scores are inappropriate measures of any injury that may be caused from heading of soccer balls, for two reasons. First, the empirical fit upon which the scores are based consider only severe head injury. Second, in the same way that angular acceleration lowers the linear acceleration measured in the mouthpiece by about 15% relative to the center of mass, it can raise or lower the GSI and HIC scores at points superior and inferior to the head's center of mass; however, the GSI and HIC scores represent only the average motion of the head. It is expected that magnitude and duration of angular acceleration are more physically relevant to head injury. An angular acceleration threshold of 107,000 $rad \cdot s^{-2}$ was observed for diffuse axonal injury in baboons (18); scaling by the mass ratio of the human brain leads to predicted critical values near 10,000 $rad \cdot s^{-2}$ (18). Direct evidence of such an injury threshold in humans is not available, however.

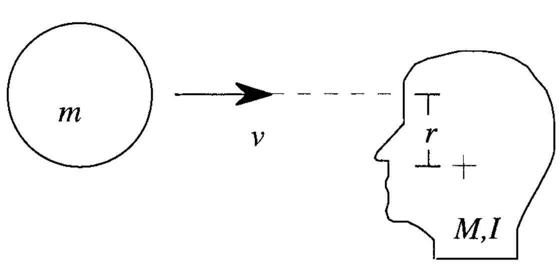

FIGURE 6—Schematic diagram and parameters of the mathematical model. The configuration shown is just before impact.

The acceleration measurements reported in this paper are based on assumptions that the headpiece is rigid, and rigidly attached to the skull. These assumptions are not true but are reasonable approximations of reality. The headpiece and procedures were designed to minimize motion between the skull and accelerometers. The low-frequency accelerations due to voluntary motion of the head were not removed (i.e., by a band-pass or high-pass filter), because these contribute to the total acceleration of the head during heading. These low-frequency accelerations turned out to be small in comparison to the maximum acceleration during impact.

The observed behavior is qualitatively accurate. The X- and Z- (anterior and superior) components of linear acceleration are much larger than the Y- (left) component, and the angular acceleration about the Y-axis is dominant, which is expected for motion in the sagittal plane.

The measured linear and angular accelerations reported in this paper are reinforced by the simplified theoretical modeling, as well as by results from the literature of tests using headforms (21). The comparison between the three is shown in Table 3. The good agreement in Table 3 lends support to the measurements presented in this article, and also serves to validate the use of a headform for prediction of acceleration in frontal heading of a soccer ball. Additionally, equation 5 appears to reasonably estimate the range of possible head accelerations in frontal heading.

The acceleration traces in Figures 4 and 5 show that the peak linear acceleration occurs within the first 15 ms of a heading maneuver. However, as can be seen in Figure 5, the peak angular acceleration magnitude reported in Table 1 occurs during the deceleration phase, perhaps

TABLE 3. Comparison of measured and predicted linear acceleration seconds with headform data from the literature (21). The tests are presented as (mean ± SD); the range on the theoretical model represents estimated upper and lower bounds on restitution coefficient $0.4 < e < 0.7$. $g = 9.81\ m \cdot s^{-2}$.

	Ball Speed = 9 $m \cdot s^{-1}$	Ball Speed = 12 $m \cdot s^{-1}$
Human heading test	16.1 ± 1.94 g	20.3 ± 2.75 g
Headform test (21)	15.1 ± 0.70 g	21.3 ± 1.78 g
Theoretical model	12.9 ± 1.25 g	17.3 ± 1.63 g

corresponding to interaction of the head and neck at the end of the heading maneuver. If this is the case, then the formula for peak angular accelerations from heading, equation 6, should not be applied.

Although the measured accelerations are well below commonly accepted injury thresholds, more study is needed to determine whether heading contributes to long-term cognitive deficits observed in professional soccer players.

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