A Numerical Investigation of Impact Severity Reduction in Football Heading

Keown, M., Shewchenko, N.
Biokinetics and Associates Ltd., Ottawa, Ontario, Canada

Dvorak, J.
F-MARC, FIFA Medical and Research Center, Schulthess Klinik, Zürich, Switzerland

ABSTRACT

While heading in football (soccer) has evolved into an essential part of play and a key skill for many players, recent media attention and research publications have raised questions about the long-term health aspects. The general lack of understanding of repeated head-to-ball impacts has raised much controversy over the effects and required countermeasures. The current study aims to provide a better understanding of the biomechanics of heading to address these issues.

The test methodology employs a hybrid approach of integrating standard biomechanical analysis techniques with a human body MADYMO model containing detailed representations of the head and neck. The modelling approach employed a single representative human subject for a specific heading scenario to validate the model and as a baseline for further exploration. The model was manipulated by forcing the motion of the torso to replicate that observed with the test subject. Activation of the neck muscles was subsequently adjusted, based on measured EMG responses, to provide the correct head kinematics. A parameter study was then conducted to investigate the effects of ball mass and compliance, muscle activation, and head and torso coupling on the biomechanics of heading.

The results have shown that decreases are possible for all head/neck responses when using the lightest ball while small changes in the pressure of the ball had a less significant effect. Increased muscle activation was shown to reduce head response but increase neck axial compression loads.

The forced manipulation of the MADYMO human model effectively simulated the observed heading technique and the execution of the parameter study was both simple and effective. The results of this study led to a number of options available for the desired reduction of heading severity to the head and neck.

INTRODUCTION

Heading in football (soccer) was first seen in the early 1870's and has since evolved to an essential part of the modern day game and a key skill for many players. Heading is also a unique component of football where the head is used as an active element to interact with and control the ball.

Over the past two years, media attention and research publications have raised questions about the long-term health aspects. The risk of mild traumatic brain injury (MTBI) resulting from a single impact is receiving significant attention from a diagnostic and treatment perspective, not only in football, but in other sports as well such as American football, ice hockey, and rugby.

There has been recent controversy on the effects of repeated impacts concerning injury risk after having already sustained a concussion and the effects of repetitive "low-level" impacts to the head thought to be associated with chronic traumatic brain injury (CTBI) [McCrory et al., 2003, Matser et al., 1998]. The controversy remains partially due to the difficulty in identifying the concussion and severity level, and due to the poor understanding of associated injury mechanisms.

Recommended practices and guidelines have been developed over the years to teach adults and youth in the *correct* methods of heading. These have been based on the experience of players and trainers alike but lack quantitative data on the physical effects and potential sequelae. Some recent biomechanical studies have provided insight into the impact severity of heading but additional work is required to better understand the biomechanics so that preventative measures can be taken.

The objective of the current study is to understand the biomechanics of heading and the effects of the impact environment on impact severity. This paper focuses on the numerical modelling and parametric study conducted towards this effort.

APPROACH

The approach employed in the study combined the measured kinematics and kinetics of human test subjects with numerical modelling to gather rudimentary information on heading biomechanics. The numerical model allows the exploration of many parameters that would otherwise be difficult to achieve and control with humans. Additionally, variability of the results is eliminated thereby providing clear insight into the effects of each parameter. This allows the effect of different heading techniques and ball characteristics to be investigated, and their implication on impact severity determined.

A total of seven human subjects experienced in football heading were selected for the initial biomechanical study. Subjects were approximately 50th percentile in stature and weight and had no physical or neuropsychological pre-conditions. Kinematic data was collected with high speed video (500 frames per second) to monitor head and torso trajectories and orientations. Kinetic data included bi-axial linear and angular accelerometers to obtain translational and rotational measurements in the midsagittal plane during impact. Neck muscle activity was measured by EMG surface electrodes placed on the major muscle groups responsible for flexion and translation of the head, the sternocleidomastoid and trapezius.

Implementation of the subject data with the numerical model required the selection of a single representative subject. This was essential for validation and implementation of the model since kinematic and kinetic responses were dependent on the individual's technique and physical characteristics. With this approach, relative changes in heading technique and response could be investigated and related back to the averaged response envelope.

The modelling approach for the study required accurate representation of the head, neck structure, and torso in terms of their mass properties. Realistic head to ball impact compliance, active and passive neck musculature, and a means to reproduce upper body heading motion were essential requirements that dictated the selection of the model.

Numerical models in football have been used by researchers to gain a better understanding of head injuries and their reduction [Ziejewski et al. (2000), Schneider et al. (1988)]. The model employed by Ziejewski consists of three-dimensional rigid bodies implemented in the ATB software code. A single body with upper and lower neck pivots represent the neck structure. A ball and socket joint attaches the head to the upper pivot while a more complex ball-socket and slip joint with a spring damper connect the lower pivot to the torso. The model was validated to reproduce the passive fore-aft acceleration response of the Hybrid III automotive test dummy head-neck structure subjected to football impacts with a padded headband. Ziejewski (1999) continued the modelling efforts with rigid body and finite element methods to study the efficacy of reducing head trauma with headbands. Peak linear and angular acceleration, HIC, and cumulative strain damage measure were used to assess injury risk.

Earlier modelling efforts by Schneider, consisted of a rigid body representing the head, seven bodies for the neck, and one for the torso. Springs and dampers interconnect the bodies to represent the overall characteristics of the discs, ligaments, and muscles. The model was validated for passive football impact conditions represented by force-time functions and with experimental data. The effects of head/ball mass ratio, impact velocity, and elasticity were investigated to establish safe playing conditions. Standard automotive research based linear and angular acceleration injury criteria were employed to assess the risk of injury (i.e. HIC, α_{max}).

While the above models enhanced the understanding of head impacts in football, improvements are required to study possible countermeasures for reducing impact severity including modified heading techniques and impact environment. The ability to investigate the effects of neck muscle activation, body segment alignment, directional sensitivity, and ball characteristics are required for the current study.

The selection of a particular numerical modelling approach depends on the objectives of the study, the fidelity required from the output, and from practical considerations and resources. The current study used a validated detailed human body model implemented in MADYMO. This was preferred over complex finite element approaches due to the required fidelity, and relative ease in validation and manipulation of the model.

MODEL DESCRIPTION AND IMPLEMENATION

HUMAN MODEL

The 50th percentile human model with the detailed neck, v.1.2.1.1, for MADYMO 6.0.1 was used to simulate the football player. A typical pre-impact configuration is depicted in Figure 1. The human model was originally developed to address frontal, rear, and lateral loading conditions and has been validated with human subjects. While the model has not been fully validated for loading in the axial direction or for impact conditions during heading, additional validation efforts were made with the detailed head-neck model.

The head-neck model possesses passive biofidelity when subjected to inertial and impact loads. Unlike the typical passive use of the model, the current study uses the model actively by forcing the muscle elements to control the motions of the head and neck system as well as the rebound kinematics of the ball. With correct representation of the head mass and neck constraints, the accelerations from the simulated heading scenario can be computed and related to some metric of injury.

The approach taken in the study to implement the MADYMO model was to obtain dynamic similitude between the model and subject by replicating the motion, or kinematics, of the player throughout the impact event. This included replicating the correct torso, neck, head, and ball motions just prior to ball impact and for a short time thereafter. Replicating the entire event from subject standstill to full follow-through was not practical due to large computational time and difficulty in configuring the model for extensive pre-impact motion and post-impact response.

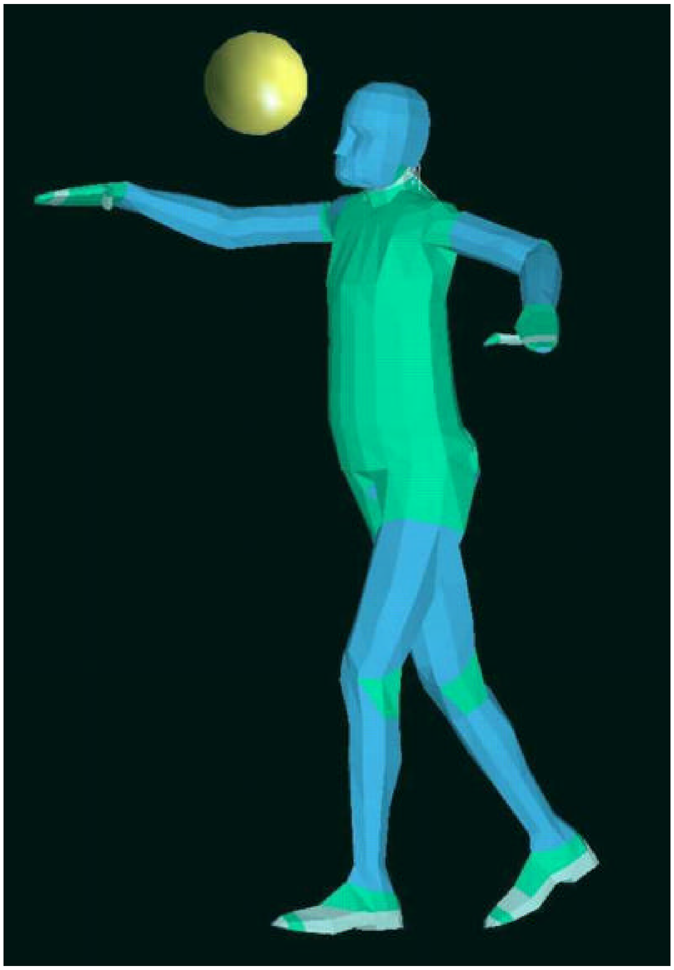


Figure 1: Human model.

To employ the human model as previously described, some specific set-up was required. The lumbar and thoracic vertebrae were locked to prevent translation or rotation between any two vertebrae. This simplification was done to ease manipulation of the torso and was confirmed by the limited torso motion observed in the subject tests. The *human_jnt* joint defined in the model was used to connect the model to the reference space of the simulation. The joint, approximating the hip joint centre, was then used to define the position and movement of the torso of the model. Additionally, several different elements were added to the model to output specific data such as the location of anatomical landmarks and the absolute angle of the head vs. time for comparison with the experimental data.

BALL MODEL

The football was created using a multi-body model with an ellipsoid outer surface. Although, the shape or contact area will not affect the interaction between the head and football for this type of model, the additional fidelity obtained with the use of a more complex model, e.g. finite element model, was not warranted for this study. This was justified as the modelling efforts are representative of a single player and it is the trends in head and neck response based on various parameters that are of interest.

The football model was based on the adidas® Fevernova Tri-Lance™, size 5 with a pressure of 80 kPa. Properties such as the size and mass of the ball were measured directly. The force-deflection characteristics were measured by impacting the football with a steel hemispherical anvil approximating the forehead shape that was mounted to a piezo-electric force transducer. Impact speeds of 7 m/s were used to approximate the ball test speed of the human trials. The ball deflection during impact was measured using high-speed video. Impact hysteresis was small and a linear approximation was used to represent both the loading and unloading phases. Different footballs with variations such as pressure and weight were tested to provide suitable characteristics for the parameter study.

POSITIONING PROCEDURE

TORSO, HEAD AND NECK

Motion of the thoracic vertebrae was locked so the only torso positioning required was the hip joint centre or h-point position and back angle. The h-point position was set to approximate the gross position of the subject because later position manipulation was to be required. The back angle was accurately set using data measured from the subject video and digital overlays of the subject images with the numerical model images.

The positioning of the numerical model neck required small rotations to be defined for the relative rotation between each cervical vertebrae. As this data was impossible to measure for the subject video, an approximation for the numerical model was required. Each value was estimated, within acceptable tolerances, until the overall shape of the neck closely resembled that of the subject data, again using digital image overlays.

Pre-simulations were executed with the numerical model that lasted only a few milliseconds and did not include any ball contact or other interactions. This was done in order to output the head angle of the model. Using this data, the upper cervical vertebrae were minutely adjusted until the head angles of the subject and the models were equal.

Even with the torso set at the correct angle, and the head and neck in approximately the same orientation to the subject, the positions of the head CG's were not aligned due to size variations between the model and the subject. The measured data from the subject was actually the tragion and a similar output marker on the model was therefore used for a direct comparison. The tragion was defined within the human model using data from the Hybrid III crash test dummy [Kaleps, 1994].

Again, pre-simulations were needed to conduct iterations and output the model's tragion position. The measured discrepancy between the tragion position from the subject to the model was used as a translation applied to the h-point position. Translating the h-point allowed the entire model to be shifted without affecting previous positioning efforts.

ARMS AND LEGS

The position of the arms and legs of the human model were approximated using the subject video but were done so only for visualization purposes. Due to the method of driving the entire torso about a single joint, the extremities do not influence the upper body kinematics. Many of the joints in the arms and legs were locked to prevent flailing while the torso moved. Similarly, to make the simulation look realistic, the feet were constrained to their initial position such that the model did not appear to jump.

BALL

The h-point of the model was also used as a reference to define the position of the soccer ball. This was simply a matter of measuring the position of the ball in the subject video and transferring that data into the numerical simulation. The position was verified using digital overlays of the video and simulation.

CONDITIONS OF SIMULATIONS

HEADING SCENARIO

From the results of the human subject trials, different heading techniques were observed for each of the subjects. This entailed differences in body stance, initial position, and the kinematics of the head, neck, torso, arms, and legs, even though the incoming and outgoing ball trajectories were similar. To obtain simulations that are representative of the subjects in general, it was decided to select a single subject for a specific heading scenario so that the experimental data could be used to validate the numerical model, and more importantly, to use this as a baseline upon which exploration of the heading techniques would be done. Data from the remaining test subjects was used to define the range of acceptable responses and parameters for the exploration studies.

The subjects were instructed to head the football in a manner representative of several typical game situations, including passing, clearing and scoring. The football was launched at the subjects at either a high speed or a low speed and targets situated at various distances from the subject were used. The scenario chosen as the reference test condition was labeled LS-2, which had the low-speed football with a passing maneuver to a target approximately 5.5 m directly in front of the subject. This scenario was considered to provide averaged head kinematics and kinetics among the heading scenarios selected. Subject 12 was chosen as the reference because he had a typical heading technique. The LS-2 scenario was used as the baseline for the remainder of the study.

TORSO MOTION

Simple motion, be it displacement or acceleration, is controlled in MADYMO through the use of the joints that connect different bodies and interactions between them. The human joint, which here represents the hip joint (h-point), is the means by which the numerical model is connected to reference space. It is the joint that will drive the model to match the observed subject kinematics. With the hip joint, there are many different ways to induce motion.

It was previously anticipated that the human joint could precisely position the torso of the human model throughout the entire heading event. For each time step measured in the high-speed video, a position vs. time plot could be used to specify the exact position and angle of the human joint of the model. This method had appeal because the torso position would not be affected by the contact of the ball. Trials with this method showed that the resolution of the subject movement was too coarse. Small irregularities in the input caused small accelerations between each time step. These accelerations, originating near the H-point, combined with the inertial properties of the head, resulted in large oscillations in the neck forces. This artifact measured beyond injurious neck load levels and hence precluded this method of torso manipulation.

Instead, an initial velocity about the human joint was defined, based on the average velocity measured from the subject video. The difference lay in the fact that the torso was now free to move but also would be affected by external forces from contact. The large mass difference between the ball and the human model meant that the model was not greatly affected by the contact and the human model followed through after ball contact in a realistic manner.

BALL SPEED

The ball speed was measured from the subject video over the course of the impact event. An average incoming speed towards the head was computed and used at the initial joint velocities (x and z directions) for the ball model in the simulation. Similarly, the average outgoing speed from the head was determined and could later be used for validation purposes. Since the position and speed was recorded as the ball approached the head, the effects of gravity were already taken into account and therefore the addition of gravity in the simulation was not required.

MUSCLE ACTIVATION

The detailed neck model contains 68 pairs of muscles organized into 3 groups based on their primary function in head and neck movement. The groups are named flexor, extensor and sterno and each group is subdivided into left and right sides. The manipulation of each muscle group through activation time histories can achieve many common head movements. Since muscles within a given group may have very different maximum force allowances, the software user specifies only a level of activation (i.e. 80% of maximal effort) and each muscle will be activated accordingly. While the active muscles are controlled by input to the model, the passive muscles react to elongation due to external loads.

MODEL VALIDATION

HEAD AND BALL TRAJECTORY

The high-speed video recording of the subject heading the soccer ball provided detailed trajectories of the ball, torso, and head. A comparison between the numerical simulation and experimental trial for the body and ball kinematics in the low-speed heading scenario is depicted in Figure 2. Qualitatively, it is shown that a good match was obtained before, during, and after impact with the ball.

A more quantitative measure was obtained from measuring the torso and head kinematics obtained from the high-speed video with corresponding data from the simulations. For the subject trials, the body kinematics

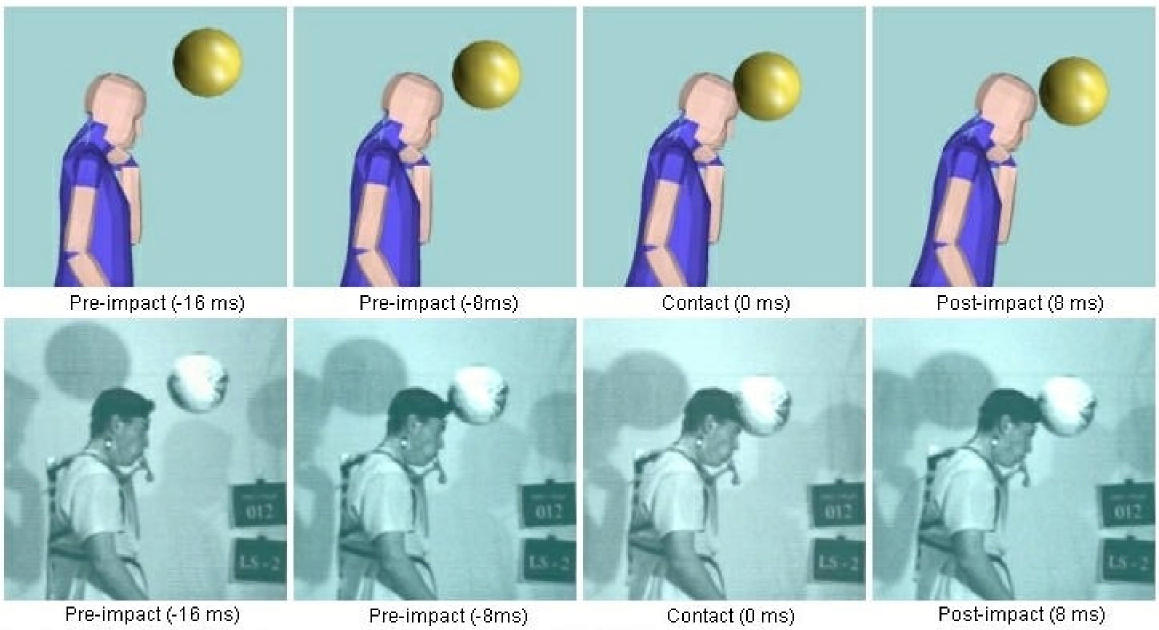


Figure 2: Qualitative comparison of model and subject kinematics.

During the subject testing, the muscle activity was measured with electromyography (EMG) on several muscles to represent the available muscle groups on the MADYMO model. This information provided insight into the activation timing of the muscle groups during the simulation but the required level of muscle activation could only be achieved through iterations. Because of the initial forward velocity of the model, the activation of the flexor muscles was of particular importance. This muscle group, applied evenly to the left and right sides of the neck, was activated in order to flex the neck and head towards the ball prior to impact. By comparing the head angle of the model with the high-speed video of the subject, activation of the remaining muscle groups was adjusted until suitable head orientation at the time of ball impact was achieved.

were measured from photographic targets mounted on the torso and head. Similarly, reference points were used to provide translations and rotations of the MADYMO model. Other data such as the head angle, ball position and ball speed was also determined. The comparison of these points at discrete time intervals of 10 milliseconds is shown in Figure 3. The torso motion of the model was prescribed by the subject data so a perfect match is observed. The kinematics of the model head is a result of the initial conditions and muscle activation and does not match as well. The ball trajectory is dependant on the head motion, the contact characteristics and the ball properties so larger deviations for the real football are expected.

Dynamic validation of the simulation was carried out by a comparison of the head translational velocity. This was based on the mathematical differentiation of the kinematics obtained from the experimental and simulation data. With comparable velocities, it is assured that system energies are similar given that internal neck muscle activities are also similar. This is shown in Figure 4 where a second order polynomial representation of the velocity is used for comparison purposes. Typical velocity differences of less than 10% were observed at the time of ball impact confirming satisfactory representation of the test subjects.

data at a point in the same location as the physical accelerometer. The close correlation between the model and subject data suggested that the model could realistically predict the head accelerations measured at the CG during the simulations.

A validation of the neck load output from the numerical model was difficult because comparable data was not readily available from the test subjects. Instead, the

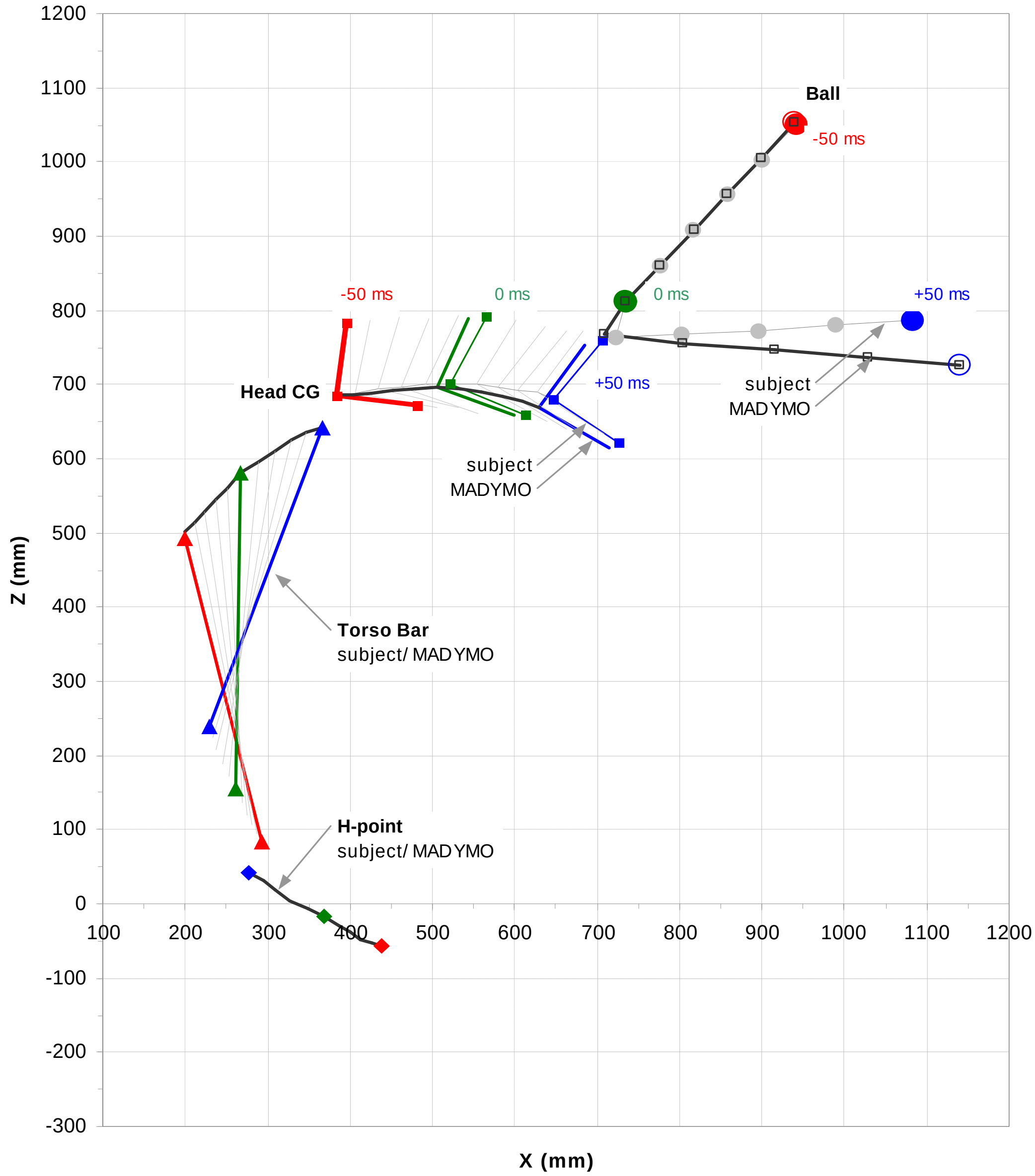
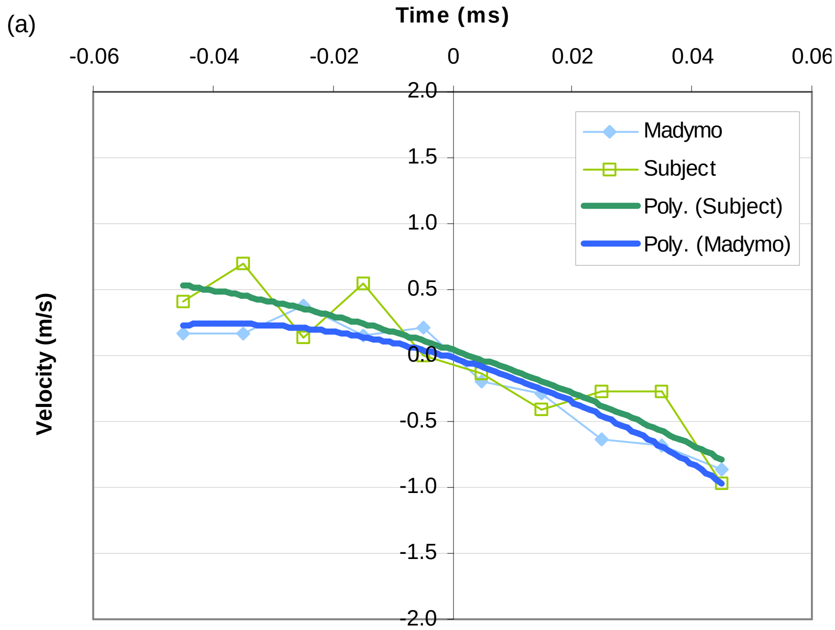


Figure 3: Comparison of model and subject markers.

ACCELERATIONS AND LOADS

During the subject trials, linear head accelerations were measured directly via accelerometers mounted to an intra-oral system that could be secured between the subject's teeth using two boil-and-bite mouth guards. The nearest accelerometer to the head centre of gravity was positioned just outside the lips of the subject. For comparison of this data with the numerical model, the model was first modified to output the head acceleration

neck load time histories were evaluated to assess whether the data was reasonable. First, the shape of the muscle force time history was considered. A smooth curve with no unnatural oscillations or severe spikes indicated good response to the model motion. Second, the magnitudes of the maximum loads were found to be well below suggested injurious levels for a human neck. This was of course supported by the lack of injury in the test subject.



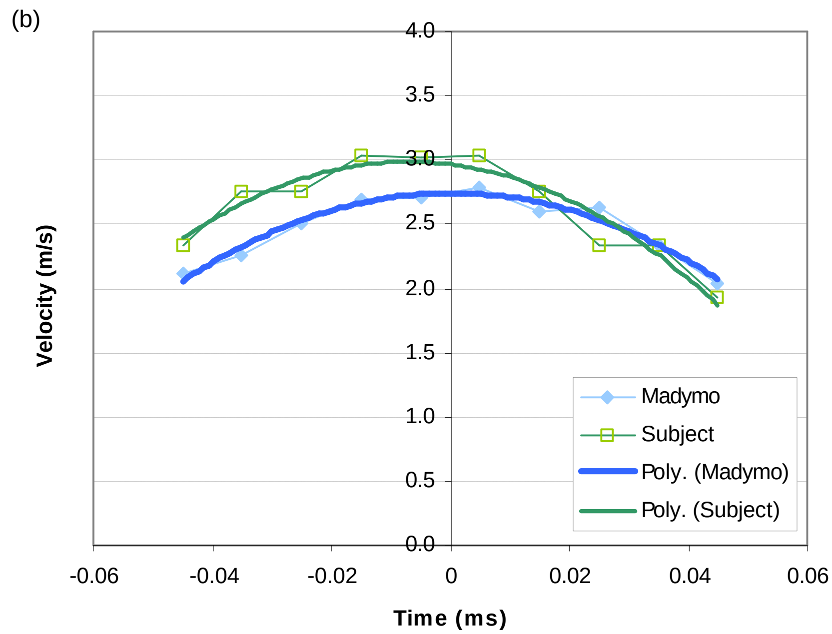


Figure 4: Head velocity comparisons in the vertical (a) and horizontal components (b).

PARAMETER STUDY

A parameter study was conducted to investigate the effects of ball properties, muscle activation and torso coupling on head biomechanics. Based on the tests of actual footballs, the force-deflection characteristics were changed to reflect different ball pressures. This property was then returned to the baseline value and the mass was changed to match that of two lightweight soccer balls. An additional effort examined the characteristics of several older generation soccer balls by simulating their mass and force-deflection properties.

The neck muscle activation levels were modified to evaluate the effects of increased coupling between the head and torso without modifying the overall pre-impact heading kinematics. The baseline set-up was modified by first increasing the flexor muscle group to a pre-defined level and then adjusting the activation of the extensor and sterno muscle groups until the resulting kinematics were reasonable similar to the baseline test. This process was completed for two cases where the flexor muscle activation levels were higher than the baseline.

To evaluate the effects of body alignment, the baseline body position was modified in an attempt to align the cervical spine and torso with the impact point on the head. This alignment was chosen to provide a more rigid support structure for the head, but it does not necessarily reflect an actual heading technique. With the neck set to its neutral position, the translation and rotation of the human joint was modified through several iterations until the resulting head motion passed through the required position and at the right angle at the time of ball impact.

The effect of these parameters was evaluated by comparing the model output including the peak head linear and rotational accelerations, upper neck (OC-C1) peak forces and moments. Other injury metrics that specifically address MTBI, or concussions, such as the Power Index developed for the American National Football League [Newman et al., 2000] were also used. The output was evaluated on the basis of percent change, either positive or negative, as compared to the baseline test.

Results of the parameter study identified several trends in head response as a function of ball properties as well as muscle activation and torso alignment. A follow-on study saw the return of several human subjects to evaluate whether the favorable changes in ball properties or heading techniques could be realized without significantly changing the nature of play. Similar trends were observed in results of this subject testing indicating that the MADYMO model was successful in predicting the head response of the football player.

CONCLUSION

The results of the study indicate that numerical modelling can provide valuable insight into the biomechanics of heading. Investigation of the player and ball parameters has provided a basis for introducing strategies to reduce ball impact severity. The validity and applicability of the results, however, depends greatly on the selection of the injury criteria, the physical parameters for the model, and the configurations employed in the analysis. A better understanding of the injury mechanisms, tolerance levels and directional impact sensitivity will allow for the continued improvement of the game and benefit to the player.

Current efforts are being directed towards advancing the understanding of head injuries including CTBI, MTBI, and TBI, and developing measures to maintain the level of safety in the sport. This will provide additional benefit to game play regulations, coaching and training methods, triage and treatment of injuries, as well as public education and enjoyment of the game. Prospective studies of head injuries in monitored game play and the improvement of ball impact response comprise some of the future efforts.

ACKNOWLEDGMENTS

The study was conducted under contract with the FIFA-Medical Assessment and Research Centre (F-MARC).

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CONTACT

Matthew Keown, a biomechanical engineer at Biokinetics, has been working with human safety issues in the areas of sports, transportation, and defence. He is experienced in all aspects of impact biomechanics comprising human experiments, full scale testing, injury analysis, and numerical modelling. Matthew has provided valuable contributions to the scientific community resulting in improvements to safety standards and protective equipment.