# HEAD, NECK, AND BODY COUPLING IN RECONSTRUCTIONS OF HELMETED HEAD IMPACTS

**Marc Beusenberg*, Nicholas Shewchenko*, James A. Newman*, Ronald de Lange◇, Hans Cappon◇**

*** Biokinetics and Associates Ltd. - Canada**
**◇ TNO-Automotive - the Netherlands**

## ABSTRACT

A mathematical modeling study is conducted to investigate neck coupling in helmeted head impacts. The main objective of the study is to provide direction for the experimental reconstruction of American football player impacts. Head responses are compared in MADYMO simulations of various impact scenarios without neck coupling, with an improved human neck model, and with a Hybrid-III neck model.

The human neck model is a continued development of an existing neck model, with improved multi-directional biofidelity. Also the helmet model has substantial effect on the interpretation of the simulation results, and a model of an American football helmet is developed.

In the simulated impact scenarios, linear head accelerations show limited change as a result of neck coupling, but angular head accelerations change drastically depending on whether or not a neck is included in the model. Effects of body mass on head accelerations are limited for the impact conditions simulated in this study. This study shows that neck coupling should be accounted for if head linear and angular accelerations obtained from accident reconstructions are to be used to establish human tolerance.

**KEYWORDS**: Accident Reconstruction, Coupling, Helmets, MADYMO, Neck, Simulation

RECONSTRUCTION OF IMPACT EVENTS usually relies on circumstantial evidence. Damage to collision partners (e.g. vehicle, helmet, road furniture), design data of these products, markings on road surface or objects hit, lay-out of the accident site, pre- and post impact position of the colliding objects, and not in the least the injury pattern of the persons involved, are used to re-enact a collision scenario. In very few cases, eyewitness material is available, let alone film recording of an event.

Many reconstruction techniques are based on the inverse dynamics principles, and start with estimating the kinematics of the impacting objects. The dynamics of the impact can then be simulated through re-enactment of the pre-, and post-impact kinematics of the colliding objects, provided the models of the colliding objects exhibit the relevant dynamic characteristics (stiffness, damping, inertia, etc.). Obviously, the better the models represent the actual objects involved in a collision, and the better the quality of the kinematics, the better the reconstruction will result in an accurate simulation of the dynamics of the impact.

In many professional sports, such as American football, broadcast cameras record impact events. This provides objective eyewitness material in the form of multiple camera views. To establish the kinematics of the objects involved in an impact, the broadcast video footage needs to be processed. With proper digitizing and scaling, we obtain a series of still images of the relative pre-, during- and post-impact position and orientation of the objects involved in the collision; see Figure 1.

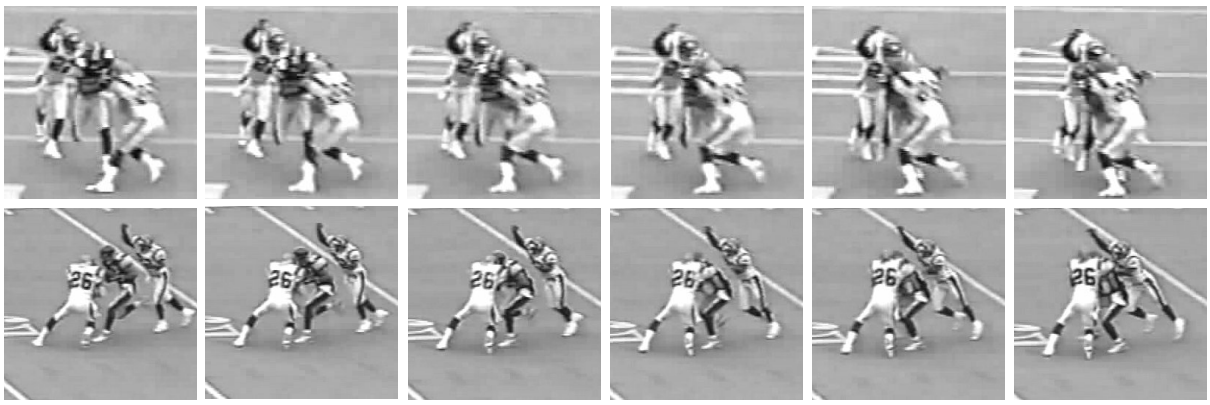


**Figure 1: Series of pre- and post-impact images of a helmet-to-helmet impact in two different views (f = 60 Hz; cropped and reduced in size for illustration purposes)**

From these images, the relative velocity of the colliding objects is established in each camera view. Using reference marks in the field of view, such as sidelines, yard lines, and hash marks, the orientation of each camera is determined with respect to inertial space. Combining the relative velocities in each camera view with the information of the camera orientations, enables the calculation of the relative three-dimensional impact velocity (Newman et al., 1999).

To establish the dynamics of such impact events, pre- and post- impact kinematics are replicated experimentally, with the relative three-dimensional impact velocity and the relative orientation of the helmeted heads as inputs. Helmets involved in professional American football are all of quite similar design, both in shape and materials, and do not exhibit permanent deformation after the studied impacts. Hybrid-III crash dummy heads represent the players' heads. The Hybrid-III head is chosen for its biofidelity in frontal and lateral contact impacts (Mertz, 1985; Howe et al., 1991), its inertial characteristics (Hubbard et al., 1974), and it allows the assessment of both linear and angular accelerations of the head centre of gravity (King et al., 1975; Denton et al., 1987). The role of coupling between the head and neck (and neck and the rest of the body) in the reconstructions of the game impacts is investigated before running the experimental reconstructions, and is the subject of this paper.

## OBJECTIVES

The main objective of the current study is to investigate the effects of head-neck-body coupling on the kinematics and dynamics of a helmeted head, in the reconstruction of concussion impacts in American football. MADYMO helmet-head-neck-body models of two football player are developed, and the neck and body models are changed to study coupling effects. Some modeling issues encountered in this study are presented, which provide direction for the use of mathematical modeling to support experimental set-ups.

## MODEL DEVELOPMENT

HELMET MODEL: A rigid body (ellipsoid) MADYMO helmet model is developed of an American football based on tests according to the helmet standard ASTM F717-89 (ASTM, 1989). Mass of the helmet model is 1.78 kg, and the mass moments of inertia are approximated at $J_{xx} = 4.06 \cdot 10^{-3}$ kgm$^2$ and $J_{yy} = J_{zz} = 4.42 \cdot 10^{-3}$ kgm$^2$.

Average head acceleration responses of three tests each, conducted on the front, front boss (oblique front at 45 degrees), top, and side of the helmet are used to establish force-deflection curves and damping specifications for each location. The resulting force-deflection curves for each location are presented in Figure 2. The damping coefficients for each location are given in Table 1. Appendix A includes the validation results of the helmet model, showing good match between the simulations and average experiments for each location.

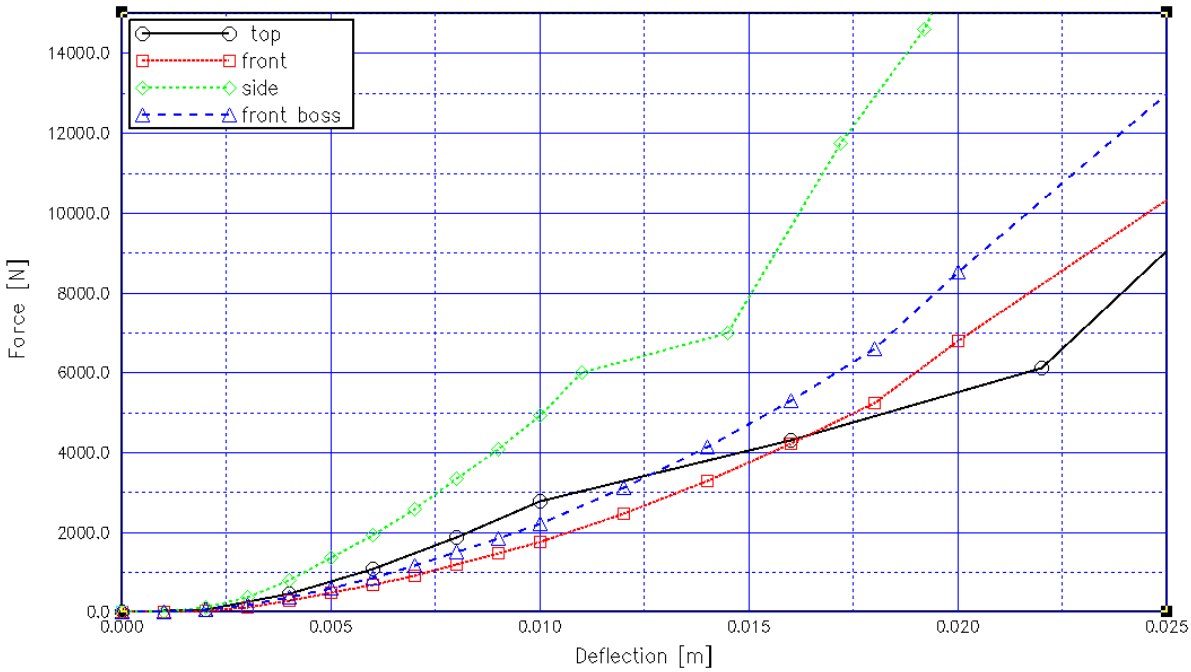



**Figure 2: F-δ for the football helmet in ASTM type tests (loading phase only)**

**Table 1: Helmet model II damping coefficients**

| Impact Direction | Damping Coefficient (Ns/m) |
|---|---|
| Top | 1000 |
| Front | 400 |
| Side | 800 |
| Oblique (Front Boss) | 100 |

HEAD MODEL: The MADYMO Hybrid-III head database is chosen as the head model, which is consistent with the experimental set-up for the reconstructions.

NECK MODEL: The human neck model applied in this study is based on the so called "global head-neck model" developed by de Jager (de Jager et al., 1996), and later further updated to meet human rear-end impact responses (van den Kroonenberg et al, 1997). The cervical and first thoracic vertebrae are represented by rigid bodies with proper inertial properties. These rigid bodies are connected through three-dimensional non-linear visco-

elastic intervertebral joints, representing the lumped mechanical behaviour of the intervertebral disks, ligaments, facet joints, muscles, and other soft tissue. The initial posture of the neck model elements includes the cervical lordosis as established from x-rays from male volunteers (Nissan et al., 1984). A passive muscle element is added at the back of the neck to improve head rotation responses. Figure 3 shows this model.

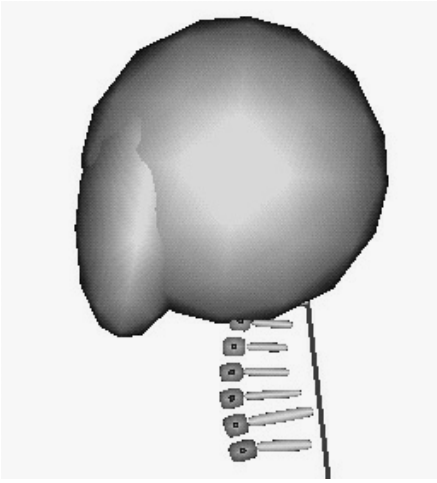


**Figure 3: Global head-neck model**

The choice of this model is based on three major considerations. First, the neck model includes sufficient detail to study head-neck-body coupling under the impact conditions considered. Individual contributions of the head-neck complex from muscles, ligaments, intervertebral disks, etc., are not required to simulate the head-neck kinematics, but sufficient segmentation of the neck is required. Second, the neck model is validated in several impact directions. Frontal, lateral, oblique, and rear (non-contact) impact validations with this neck model have been performed in the past (van den Kroonenberg, 1997). Third, the neck joint characteristics can be further tuned to meet human compression responses as defined by Pintar et al. (Pintar et al, 1995).

Preliminary simulations with the model showed unrealistic bottoming of the neck joints at their end of range of motion, causing sudden changes (discontinuities) in neck and head responses. Figure 4 illustrates the tuning applied to the moment-angle relationship of the neck joints (shown only for C0-C1 joint). Although this change could introduce larger ranges of motion in flexion, the total neck flexion in this study never exceeded normal human values.

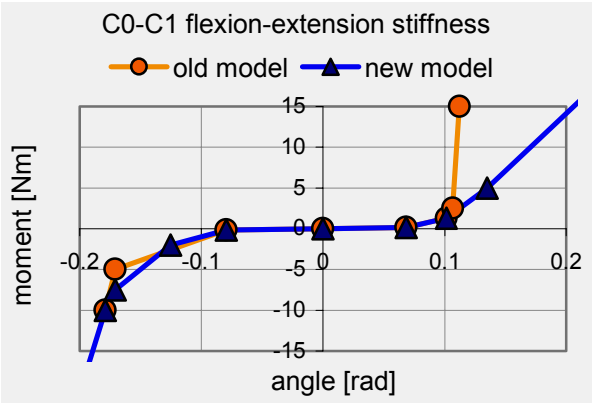



**Figure 4: Moment-angle relationship of C0-C1**

Validation results of the human neck model are included in Appendix B. The human neck model performs reasonably well compared to volunteer data in frontal (15g), lateral (7g), oblique (11g), and rear (7g) non-contact impacts, and better than previously reported models of this type (i.e. models based on the “de Jager global model”).

Simulations are also performed with the MADYMO Hybrid-III neck database (version 5.6) and without a neck attached to the head. This is described under the parameter study below.

BODY MODEL: Three representations of the player’s body are used: a rigid body of 50 kg with its centre of gravity located where the centre of gravity of the Hybrid-III database would be if it were attached to the head-neck model, a 5 kg point mass rigidly attached to the T1 vertebra, and an infinite mass attached to T1 (T1 rigidly attached to inertial space).

## MODEL SET-UPS

Football players are involved in a variety of impact scenarios. Helmet-to-helmet collisions account for the majority of impacts resulting in mild traumatic brain injury. The current study focuses on the effects of neck coupling in this impact scenario.

The basic simulation set-up includes representations of two football players, A and B, impacting head-to-head (helmet-to-helmet). At $t_0$, player A is stationary and impacted by player B. Player B initially moves at the relative impact velocity ($v_i$), which for this study is set at 10 m/s. This velocity is chosen somewhat arbitrarily, but is within the range of relative impact velocities obtained from kinematics analyses of more than 30 actual professional American football game incidents (relative impact velocities of up to 13.5 m/s are observed).

In all simulations, player A is impacted at the crown (top) of the helmet. The initial posture of player A is aligned with the direction of impact, which is perpendicular to the transverse plane of head A. The helmet-head-neck-body model of player B is oriented in space such that four contacts are simulated:

- front, with the direction of impact going through the head CG of player B;
- top, with the direction of the impact going through the head CG of player B;
- side, in the transverse plane and directed through the head CG of player B;
- oblique front, in a transverse plane and parallel to the y-axis of the head of player B.

Figure 5 shows the basic set-up of the simulations.

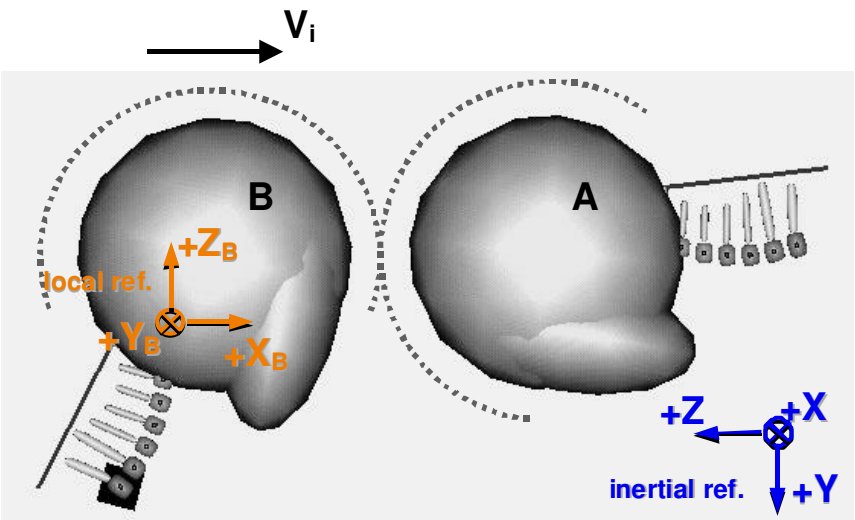



**Figure 5: Basic model set-up**

Several combinations of neck models and body models for both players A and B are applied. Table 2 shows the different models of players A and B included in this study. Combining the five different models for each player A and B and the four contacts, results in a matrix of 100 simulations. For practical reasons, however, not all combinations of models and contacts are simulated. The most important ones are discussed below.

**Table 2: Helmet-head-neck-body models for player A and B**

| | Helmet model | Head model | Neck Model | Body Model |
|---|---|---|---|---|
| **Player A** | Football Helmet | Hybrid-III | None | None |
| | ,, (same) | ,, (same) | Hybrid-III | 50 kg |
| | ,, (same) | ,, (same) | Hybrid-III | Infinite Mass |
| | ,, (same) | ,, (same) | Human Model | 50 kg |
| | ,, (same) | ,, (same) | Human Model | Infinite Mass |
| **Player B** | Football Helmet | Hybrid-III | None | None |
| | ,, (same) | ,, (same) | Hybrid-III | 5 kg |
| | ,, (same) | ,, (same) | Hybrid-III | 50 kg |
| | ,, (same) | ,, (same) | Human Model | 5 kg |
| | ,, (same) | ,, (same) | Human Model | 50kg |

## PARAMETER STUDY: RESULTS AND DISCUSSIONS

Because of the specific model set-up chosen for this study, player A's neck is predominantly loaded in compression. Although this loading condition hardly occurs in football tackles (and never in the incidents included in our database), it was chosen to provide a repeatable impact scenario. Unfortunately, the human neck model is unstable in the impacts simulated in this study, showing excessive compression and collapses onto itself. Figure 6 illustrates this behaviour.

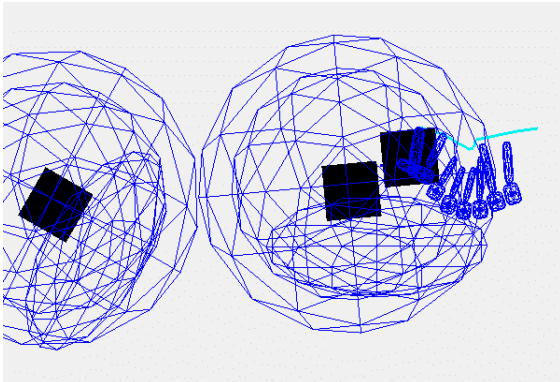


**Figure 6: Instability of human neck model in compression (player A at 25 ms)**

Figure 7 shows that the player B head accelerations (expressed in the inertial reference frame) are all very similar for player A modeled without neck, with a human neck attached to a 50 kg body mass, or with a human neck with T1 rigidly attached to inertial space. Axial stiffness provided by the human neck model is relatively low and hence the head of player A reacts as if no neck is attached to it. This is observed for simulations of impacts in different directions (only responses to oblique impact are included in Figure 7).

As a result, all conditions simulated with player's A neck represented by the human neck model can be considered as if no neck is attached to player A's head model. Unfortunately, this limits the use of many of the performed simulations, and neck coupling effects cannot be addressed in the current study using responses of player A.

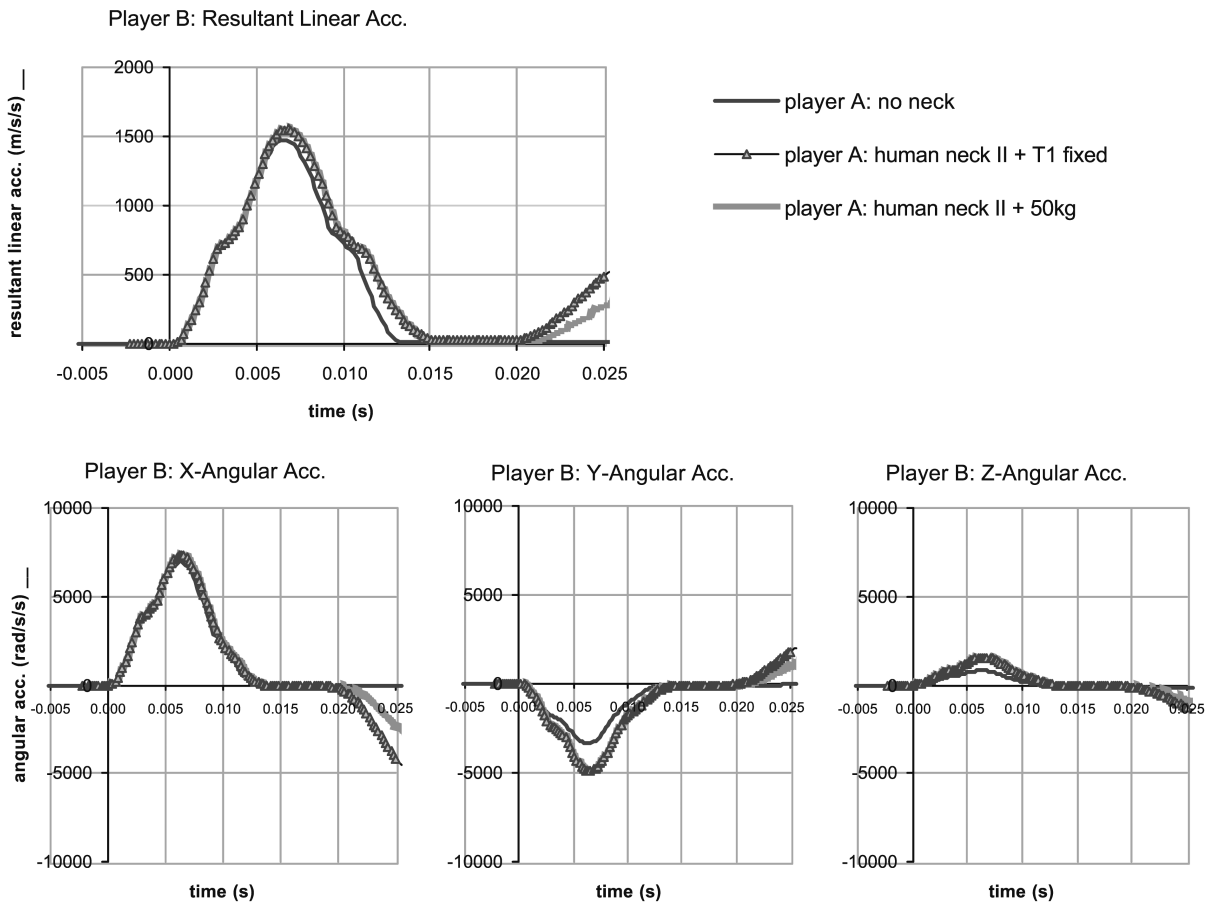



**Figure 7: Oblique impact responses (in inertial reference) of player B (helmet-head only) using different models of player A.**

Four different neck-body model combinations are applied for player B, with player A's neck not included or represented by the human neck model: 1. no neck; 2. Hybrid-III neck model with 5 kg body mass; 3. human neck model with 5 kg body mass; and 4. human neck model with 50 kg body mass. Figure 8 shows linear and angular head acceleration responses of player B for these four combinations in oblique loading. All accelerations are expressed with respect to the inertial reference frame, which axes +X, +Y, and +Z line up with the local head axes $-Z_B$, $+X_B$, and $-Y_B$ respectively of player B at $t_0$ (see also Figure 5).

Linear head accelerations of player B do not change much as a result of applying a different neck model for player B, despite quite different bending characteristics of the applied neck models. The angular accelerations of player B, however, change considerably with different neck models.

For all neck models of player B, angular X-accelerations ($-Z_B$ at $t_0$) are largest and are reasonably equal up to about 20 ms. Differences in angular accelerations in Y direction are substantial and can be explained by neck coupling. The "no neck" simulation shows a negative response, indicating rotation around the head CG, while with the Hybrid-III neck or human neck, rotation is initiated around OC and in opposite direction due to the location of contact. A similar effect is seen for the Z-angular acceleration, although at much lower amplitude.

Furthermore, and also observed in preliminary simulations, Figure 8 shows that the changes in body mass (5 kg versus 50 kg) have very limited effect on the head accelerations of player B with the human neck model.

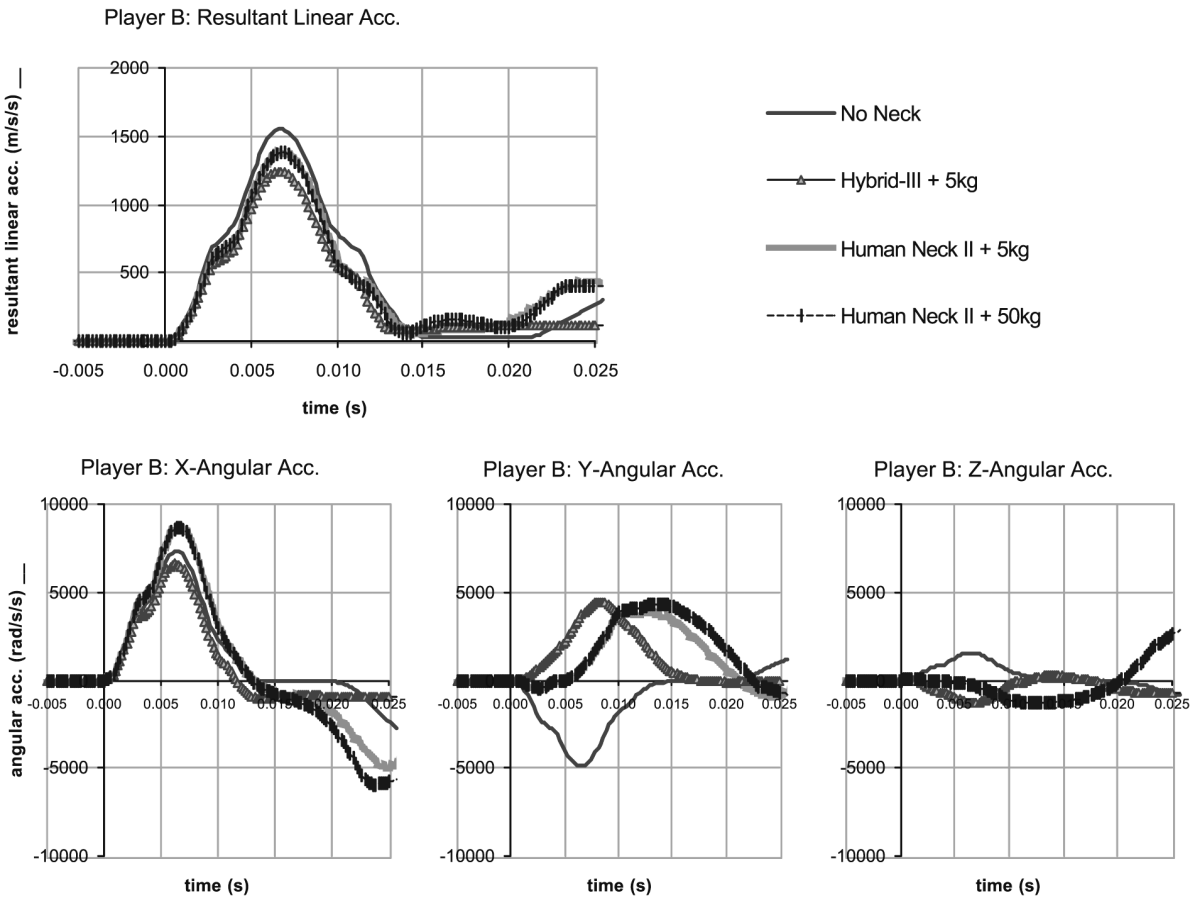



**Figure 8: Oblique impact responses (in inertial reference) of player B using different neck-body models for player B**

Some studies have been conducted in the past concerning the effects of neck coupling on head responses in helmeted head impacts, but with different objectives than the current study. As part of the European COST 327 project (Gillingham, 2001), Hering et al. (2000) compare linear and angular head acceleration responses of motorcycle helmet tests conducted with a full Hybrid-III dummy to those obtained with a head only. Whereas that study concludes that assessment of helmet performance may not require more complex representation of the human than a headform only, it also indicates that neck coupling needs to be accounted for if rotational head accelerations are to be indicative of the risk of head injury.

The current study goes further, in that it indicates that replication of helmet-head motion is difficult without neck coupling, and that the experimental reconstruction of football player impacts require a neck, coupling the head to a body mass, in order to correctly study brain tolerance to concussion on the basis of linear and angular accelerations.

Chinn et al. (1999), also as part of the European COST 327 project, report linear and angular head responses from reconstructions of motorcycle accidents. Although not the helmet-head motions are reconstructed, but rather the helmet damage, the fact that a head-only system was used to represent the human so no neck coupling was accounted for, may limit the use of this dataset for establishing head injury tolerance.

In contrast to the findings of Hering et al. (2000), the current study shows that changes in body mass have limited effect on head linear and angular accelerations, and accident reconstructions may not require the human body to be represented by a full crash test dummy. Different conclusions of the two studies may be a result of different impact configurations simulated, and require further study.

## CONCLUSIONS

A mathematical modeling study is conducted to investigate coupling effects of the neck and body in helmeted head impacts. Main objective is not to establish whether similar acceleration responses can be obtained with or without neck coupling, such as is the case in other research, but rather to ensure that the head motion can be replicated as established from video analysis of impacting football players.

Preliminary simulations were run to verify the appropriateness of the helmet-head-neck-body model. Not only the human neck model required updates, but also the model of the helmet needed to be improved, in order to be able to assess the effects of neck coupling on linear and angular head accelerations.

The simulations reported here compare head accelerations responses without neck coupling, with a human neck model, and with a Hybrid-III neck model. The simulated impact scenarios load one of the neck models (player A) in compression. In this condition, the human neck model appears unstable and provides very little coupling, while the Hybrid-III neck model appears infinitely stiff. Both these conditions are not considered realistic, and the analysis is further limited to head responses of the other player (loaded predominantly perpendicular to the central axis of the neck).

Analysis of head responses in simulated oblique impacts (player B) show that neck coupling has limited effect on the linear head accelerations, but considerably changes the rotational head response. Head kinematics can be quite different depending on whether or not a neck is included to represent the human. In simulations where a human neck model is included, the head accelerations show little sensitivity to changes in body mass.

The current study therefore shows that neck coupling should be accounted for if head linear and angular accelerations obtained from accident reconstructions are to be used to establish human tolerance. However, a complex representation of the human body does not seem to be required for the impact conditions studied here, since the effects of body mass are limited. These findings have supported the experimental set-up applied to reconstruct head impacts occurring in professional American football.

## ACKNOWLEDGEMENT AND DISCLAIMER

This study is conducted under a grant to Biokinetics from the National Football League for the study of Mild Traumatic Brain Injury. The opinions expressed here are those of the authors and do not necessarily reflect those of the National Football League.

## REFERENCES

American Society for Testing and Materials (ASTM) standard F717-89 “Standard Specifications for Football Helmets”, 1989.

Chinn B.P., D. Doyle, D. Otte, E. Schuller: “*Motorcyclist Head Injuries: Mechanisms Identified From Accident Reconstruction and Helmet Damage Replication*”, in Proceedings of the 1999 Int. IRCOBI Conference, pp. 53-71; Bron, France; 1999.

Denton R., C. Morgan: “*An Overview of Existing Sensors for the Hybrid-III Anthropomorphic Dummy*”, SAE Paper No. 876048, in “Hybrid-III: The First Human-Like Crash Test Dummy”, SAE PT-44, eds. S. Backaitis, H.J. Mertz; Society of Automotive Engineers; 1994.

Gillingham S., et al.: "COST 327 Motorcycle Safety Helmets", European Community Action COST-Transport; www.cordis.lu/cost-transports/src/cost-327.htm; 2001.

Hering A.M., S. Derler: "*Motorcycle Helmet Drop Tests Using A Hybrid-III Dummy*", in Proceedings of the 2000 Int. IRCOBI Conference, pp. 307-321; Bron, France; 2000.

Howe J.G., D.T. Willke, J.A. Collins: "*Development of a Featureless Free-Motion Headform*", SAE Paper No. 912909, in Proceedings of the 35th Stapp Car Crash Conference, pp. 289-301; Society of Automotive Engineers; 1991.

Hubbard R.P., D.G. McLeod: "*Definition and Development of a Crash Dummy Head*", SAE Paper No. 741193, in Proceedings of the 18th Stapp Car Crash Conference, pp. 599-628; Society of Automotive Engineers; 1974.

Jager M. de, A. Sauren, J. Thunnissen, J. Wismans: "*A Global and a Detailed Mathematical Model for Head-Neck Dynamics*", SAE Paper No. 962430 in Proceedings of the 40th Stapp Car Crash Conference, pp. 269-281; Society of Automotive Engineers; 1996.

King A.I., K.W. Krieger, A.J. Padgaonkar: "*Measurement of Angular Acceleration of a Rigid Body Using Linear Accelerometers*" Journal of Applied Mechanics, Vol 42, No. 3, pp. 552-556; 1975.

Kroonenberg A. van den, J. Thunnissen, J. Wismans: "*A Human Model for Low-Severity Rear-Impacts*", in Proceedings of the 1997 Int. IRCOBI Conference, pp. 117-132; Bron, France; 1999.

Mertz H.J.: "*Biofidelity of the Hybrid-III Head*"; SAE Paper No. 851245, in "Hybrid-III: The First Human-Like Crash Test Dummy", SAE PT-44, eds. S. Backaitis, H.J. Mertz; Society of Automotive Engineers; 1994.

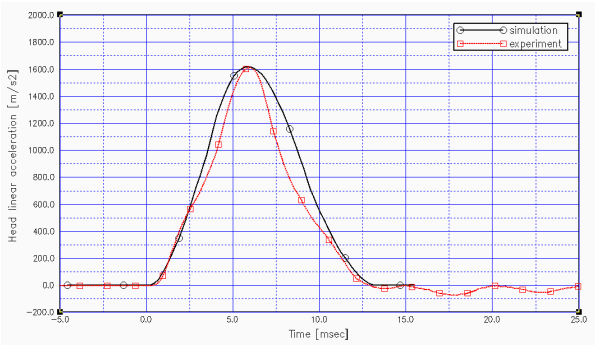
Newman J.A., M. Beusenberg, E. Fournier, N. Shewchenko, C. Withnall, A. King, K, Yang, L. Zhang, J. McElhaney, L. Thibault, G. McGinnes: "*A New Biomechanical Assessment of Mild Traumatic Brain Injury, Part 1: Methodology*", in Proceedings of the 1999 Int. IRCOBI Conference, pp. 17-36; Bron, France; 1999.

Nissan M., I. Gilad: "*The Cervical and Lumbar Vertebrae-Anthropometric Model*", Engineering in Medicine, Vol. 13: pp. 11-114; 1984.

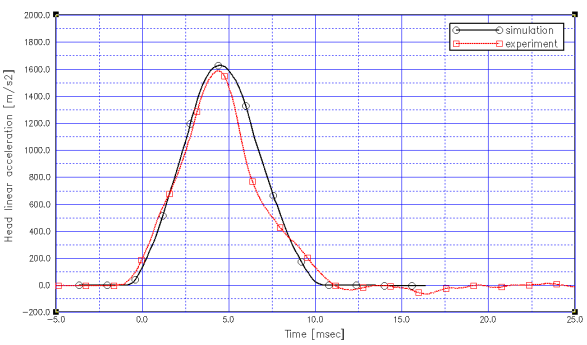
Pintar F., N. Yoganandan, L. Voo, J.F. Cusick, D.J. Maiman, A. Sances Jr.: "*Dynamic Characteristics of the Human Cervical Spine*", SAE Paper No. 952722 in Proceedings of the 39th Stapp Car Crash Conference, pp. 195-202; Society of Automotive Engineers; 1995.

# APPENDIX A: VALIDATION DATA OF THE FOOTBALL HELMET

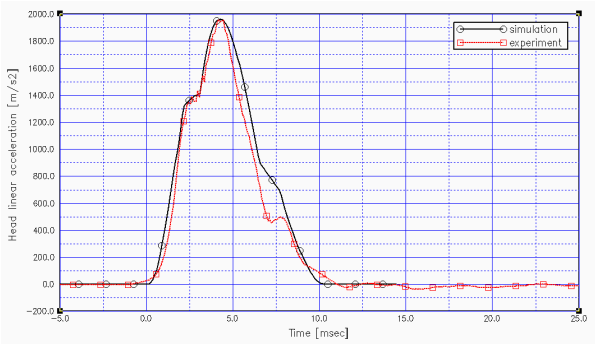
FRONT



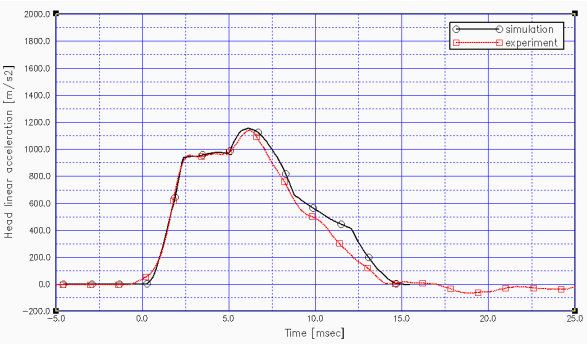

FRONT BOSS



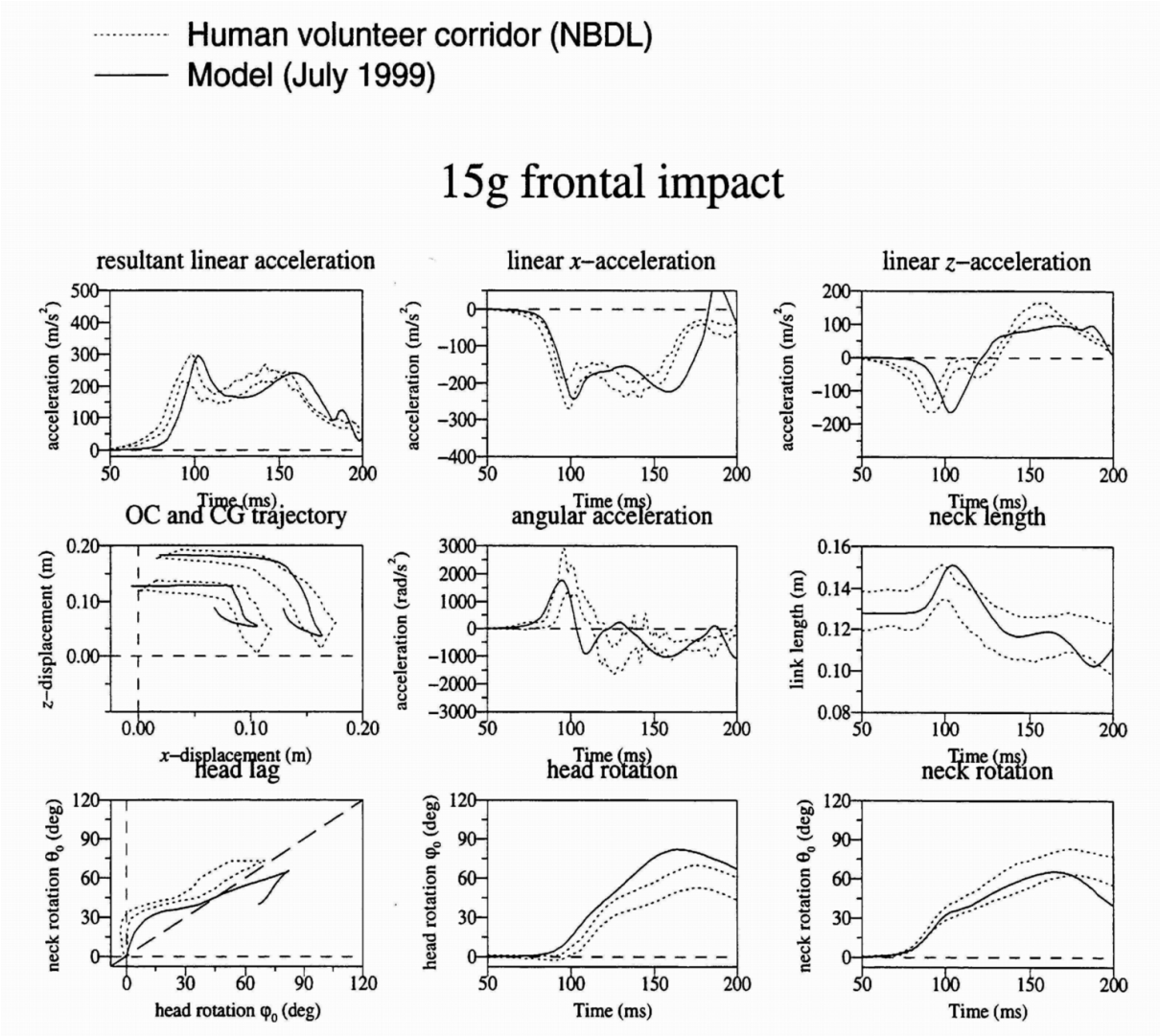

SIDE




TOP

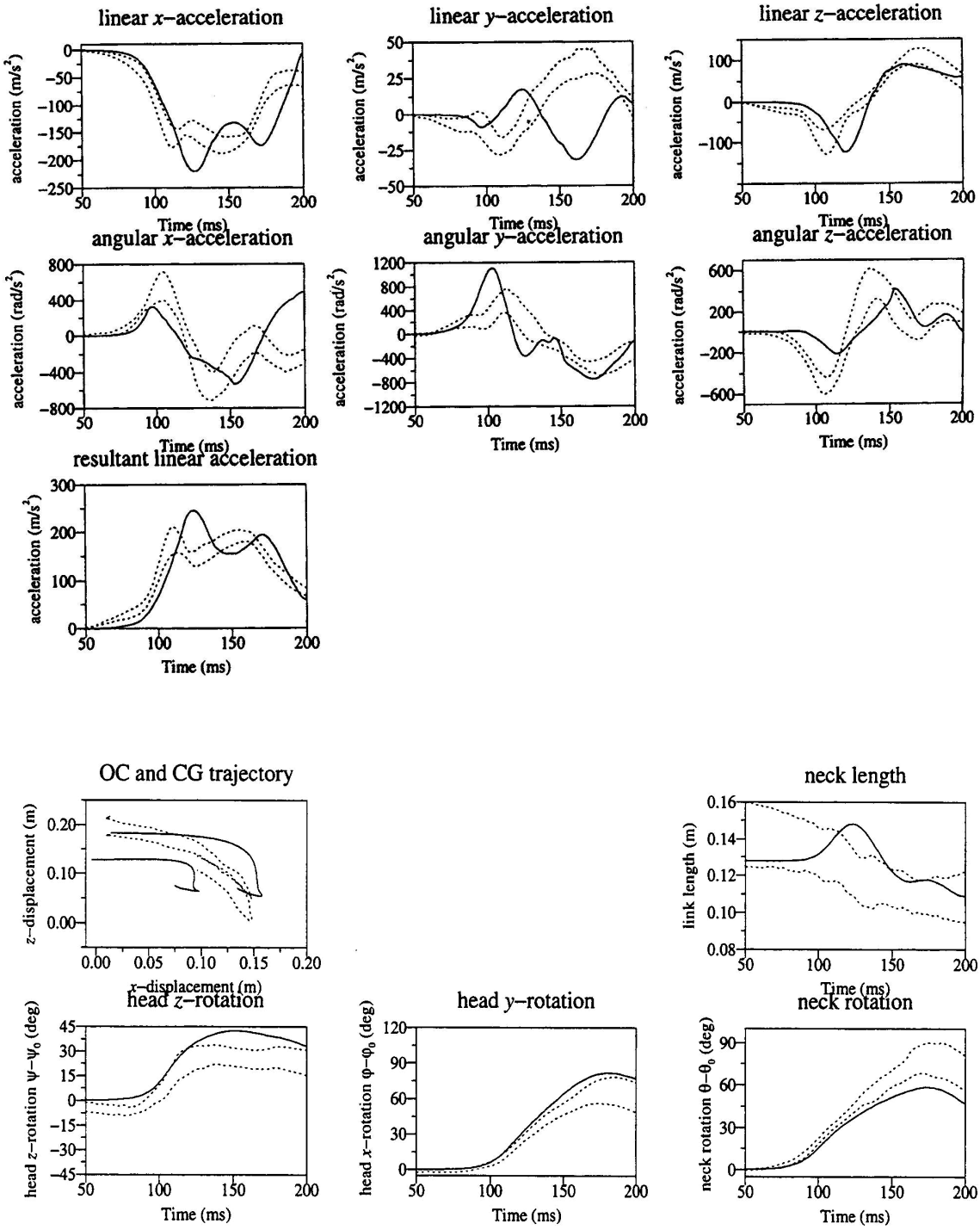


## APPENDIX B: VALIDATION DATA OF NECK MODEL



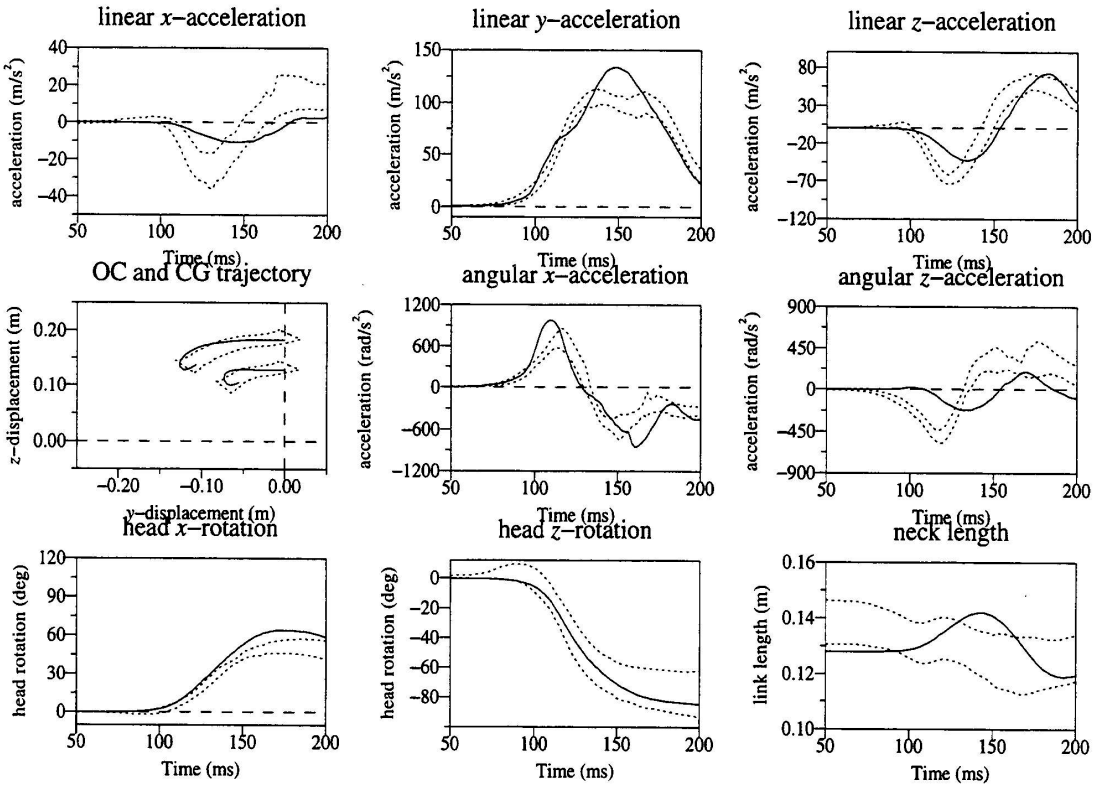
—— Model (July 1999)
····· Human volunteer corridor

# 11g oblique impact



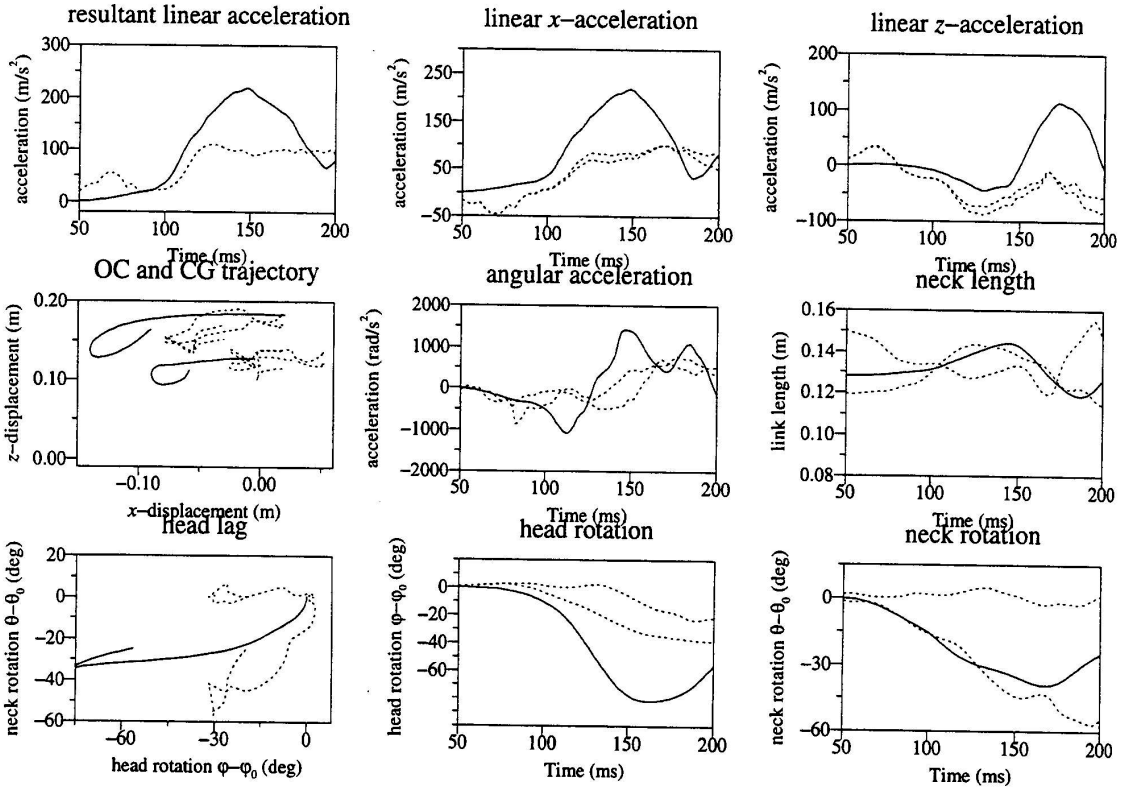
Model (July 1999)
Human volunteer corridor (NBDL)

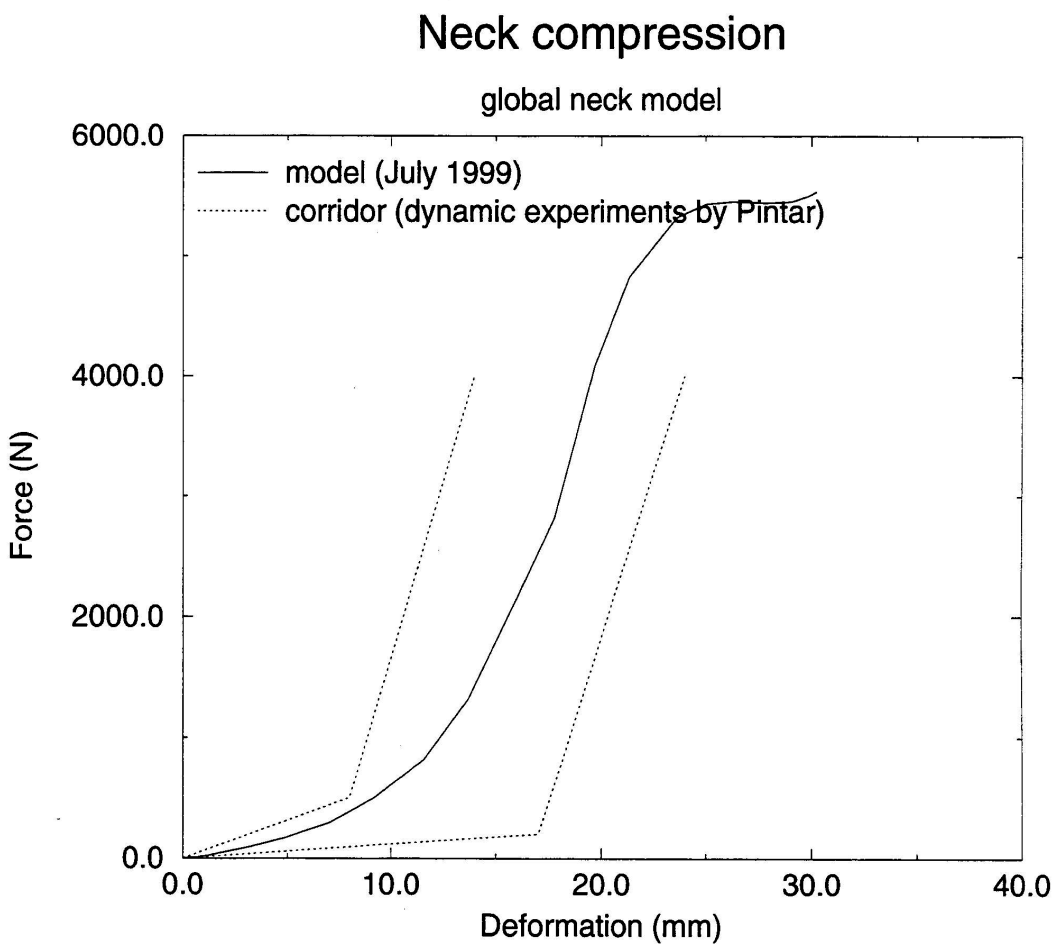
# 7g lateral impact



PMHS corridor (2 experiments)
Model (July 1999)

# 7g rear end impact




Neck compression
global neck model
model (July 1999)
corridor (dynamic experiments by Pintar)
Force (N)
6000.0
4000.0
2000.0
0.0
0.0
10.0
20.0
30.0
40.0
Deformation (mm)