Validation of a human model for frontal impacts

Validation of a Human Model for Frontal Impacts.

Ronald Ruimerman

Stage rapport WFW 98-004

Begeleider: Riender Happee, TNO Crash-Safety Research Centre

Abstract

It is important to know the kinematic behaviour of the human body during car accidents. If this behaviour is known, car interiors could be designed, for example, in a way that physical contact between occupant and environment can be avoided during a crash. For this reason mathematical models are used to predict the human motion. A. van den Kroonenberg developed a mathematical multibody model to predict the human behaviour during impact. To improve the conformity of model output and reality, the human spine was modeled with the help of 16 bodies. For the neck the global head-neck model of M. de Jager was used.

The objective of this study is to validate this human multibody model made by A. van den Kroonenberg with the help of frontal sled tests of different severity and a pendulum test. If necesary, changes will be made to improve the conformity of model output and experimental data. The model will also be validated for rear impact.

This study revealed that the stiffness parameters in the neck had to be modified to improve the model output. Stiffness for flexion-extension had to be increased. For tension-compression and anterior-posterior shear the parameters were slightly decreased. Also damping was increased. This resulted in a model that is appropriate to predict the global motion of a 50th percentile male. The pendulum test showed that the stiffness of the chest is too high.

Contents

Chapter 1

Introduction

1.1 Preface

To prevent injuries because of traffic accidents it is important to be able to predict the behaviour of the human body during impact. It is possible to predict the behaviour in different situations with the help of dummies and mathematical models. [see H. Ishikawa et.al. (1993) and Yue Huang et.al. (1994)] There are mathematical models of different complexity. Very complex FEM-models can give detailed information about the biomechanical injury mechanism. These models will have long calculation times.

If the motion of the human body must be predicted, then a mathematical multibody-system model will be sufficient. Calculation times for multibody models are much lower than for FEM-models.

The multibody model must have the same behaviour as the human body during impact. Because of the variety of the human body a 50th percentile male is chosen. The model must predict the behaviour of the 50th percentile male as realistic as possible.

To check a model a validation must be performed. This can be done by comparing the model output with experimental data from tests with volunteers and Post Mortem Human Subjects (PMHS). A disadvantage of volunteers is that impact cannot be high and that fixation of instrumentation is difficult. [see: J. Thunnissen et.al. (1995)] Also possible interference of instrumentation and seatback can be a problem. A disadvantage of PMHS's is that no muscle reaction is present. This will result in different behaviour between tests with PMHS's and human volunteers. [see J. Wismans (1987)]

1.2 Objective

The objective of this paper is to validate an existing multibody model of the human body [see: A. van den Kroonenberg et.al. (1997)] for frontal impact. This is done with the help of the multibody/finite element software package MADYMO (MAthematical DYnamical MOdels) developed by the TNO Crash Safety Research Centre. If necessary changes will be made to improve the conformity of model output and experimental data. The model has been validated for rear impact. Changes made must not negatively influence the results for rear impacts.

If the model has been validated it can be used to predict the motions of car occupants in frontal and rear loading. The model can be used for parametric studies to aid in the design process of car seats and cars to avoid contact of the human body and environment during impact. It can also be used for research of injury biomechanics. The global motion of the human body during frontal and rear loading can be determined with small calculation times. If more detailled information of a specific part of the body is requested, then this part can be replaced by a more detailled model, for instance using the FEM.

1.3 Approach

To validate the model the output will be compared with experimental data from volunteers for a 15G frontal impact. [see: C.L. Ewing et.al. (1968)] If the output of the human model is not conform the experimental data, then modifications will be made to improve the human model.

The model will also be validated for impacts lower then 15G and for a thorax pendulum test with two different impact velocities. For lower impacts, the model output must also be adequate to predict the motion of the human body. A validation for the pendulum test on the chest is important to predict behaviour of the chest during loading. This can be the interaction of the chest and belts or chest and airbag.

The changes of the model must not negatively influence the reliability for the rear impact. The model must still satisfy the requirements for low and mid severity rear impact. [see: M.L.C Hoofman et.al. (1997)].

Chapter 2

The model

To predict the motion of the human body during impact a new model is develloped by A. van den Kroonenberg [see: A. van den Kroonenberg et.al. (1997)]. The model is based upon the MADYMO model of the Hybrid III dummy with some modifications. The human model consists of three components: a thoracic+lumbar model representing the human spine, a head-neck model and the remaining bodyparts such as arms and legs. In the new model the human spine and the neck are more detailed.

The first component is the thoracic+lumbar model. To obtain an accurate estimation of the kinematics of T1 the spine is modelled with the help of 16 bodies representing the lumbar and thoracic vertebrae. The total mass of the lumbar+thoracic spine is 12.6 kg. This mass is equally divided among the 16 bodies. This results in a mass of 0.79 kg per vertebrae. The mass moments of inertia are chosen 0.005 kg/m^2. The location and orientation of the spine joints are chosen according to the anthropometry of a 50 percentile seated adult male. Although the spine model is in principle a three dimensional model, only two degrees of freedom are allowed in this model: flexion/extension and axial elongation/compression. Stiffness and damping parameter values are obtained from Prasad and King, [see: Prasad and King (1974)]. The range of motion (ROM) represents the total amount of displacement or rotation that the biomechanical structure can sustain without being damaged. The ROM for flexion cq. extension is 105 /60 degrees according to Kapandji [see: Kapandji (1974)]. These ROMs are equally devided between lumbar and thoracic spine and equally distributed among the vertebra in these regions.

The second component is a head-neck model. Different head-neck models of different complexity exist. [see: R. van Haaster (1996), M. de Jager et.al. (1994) and J. Brelin-Fornari et.al.] Here the "global model" developed by M. de Jager is used [see article: M. de Jager (1996)]. The model consists of 9 bodies. There are 8 bodies used to model the neck, including the first thoracic vertebra T1 and one body to model the head.
The mechanical characteristics of the cervical spine are difficult to obtain experimentally. Consequently large variations between different studies are found. Muscle tension can significantly decrease the neutral zones (NZ) and ROMs. Large preloads may stiffen the motion segment and result in smaller displacements.

In vitro mechanical characteristics are fairly complete for static and quasi static loading. Missing characteristics can be estimated through comparison with other segments, tissues and loading directions or magnitudes for which characteristics have been quantified.
The bodies are connected by free joints. Free joints have three translational and three rotational degrees of freedom. The total number of degrees of freedom is six. Behaviour of the intervertebral soft tissues is lumped into a single element representing motion segment behaviour. For simplicity also muscle behaviour was lumped into the intervertebral joint stiffnesses.
For the lower cervical joint stiffnesses the in vitro ROMs are much larger compared to in vivo ROMs. If values of the in vitro experiments are used, this will result in an unrealistic model that is much to flexible. Especially for rotation. Therefore data are modified for each joint separately by multiplying the rotations with the ratio of the in vivo ROM for that joint to the average in vitro ROM. Consequently the moment-rotation curves are now different for each joint and reflect the regional differences found in vivo. For translational ROMs no correction was applied because no in vivo data were available.
For the upper cervical joint stiffnesses static moment-rotation characteristics are available for C0-C1 and C1-C2. The values for the stiffness parameters for in vivo and in vitro agree very well so no correction for the in vivo ROMs were introduced. Because no force-translation data were available a relatively large stiffness of 500 $[N/mm]$ is chosen for tension/compression. This results in little translational motion under normal physiologic loads which agrees with observations for these joints. [see: M. de Jager (1996)]
The stiffness characteristics for flexion/extension are modified by A. van den Kroonenberg to improve the correlation for rearward with the experiments with volunteers. The stiffnesses are increased.

The third component are the remaining bodyparts such as arms and legs. These are modelled with the help of 25 bodies. The total model consists of 49 bodies.

2.1 Validation for frontal impact with a Sled test

2.1.1 The experiment

In the following the model will be validated for frontal impact. Experimental data of the dynamic response of the head and the neck from living human subjects to $-G_x$ sled acceleration is determined by Ewing et.al. at the Naval BioDynamics Laboratory in New Orleans. [see: Channing L. Ewing et.al. (1968)]. Sled tests with accelerations up to $15G$ have been done. The volunteers were selected to encompass the 5th to 95th percentile distribution of sitting height according to a selected reference.
In the design of the experiment certain simplifying assumptions were made. The first was that all accelerations acting on the head and neck of the seated pelvic- and torso restrained individual subjected to a $-G_x$ acceleration would be detectable at the first thoracic vertebra (T1). Additional accelerations transmitted by the soft tissues of the neck could be a source of error. The second assumption was that the head is a rigid body. The third is that all significant head and neck motions are in the mid-sagittal plane.
Precision inertial transducers were used to determine the linear and angular acceleration of the head and the first thoracic vertebra. Precision high-speed cameras were mounted on the sled to determine the displacements of the head and the neck of the subject.

The volunteer was sitting in a rigid seat with a horizontal seat cushion and a seat back perpendicular to the seat cushion. The model of the human body will be set in an environment that equals the environment of the volunteer in the test conditions. The neck is placed on the body and the model is placed in a sitting position. Then with the help of initial conditions the model is put in a position close to the position of the volunteer. The rotations defined under joint DOF (degrees of freedom) are chosen small because otherwise the model would start moving under an initial tension in its spine. Plane-Ellipsoid contact interactions between the seat and lower torso, upper torso and the bodies representing the vertebrae must be defined. Contact interaction between seat cushion and the hips are added.

The volunteer was wearing a belt consisting of six parts. Five belt segments and one rigid part. For the rigid part a second system existing of one body is required.
The initial position of the rigid part and the coordinates of the attachment points of the belt segments must be defined. Also functions for the elastic belt stiffness and hysteresis must be defined.

To avoid disturbances of the arms the arms are strapped down during the experiment. In the model this is simulated with the help of point-restraints. Also the legs are strapped down in the test. This is modelled with the help of point restraints between feet and floor.

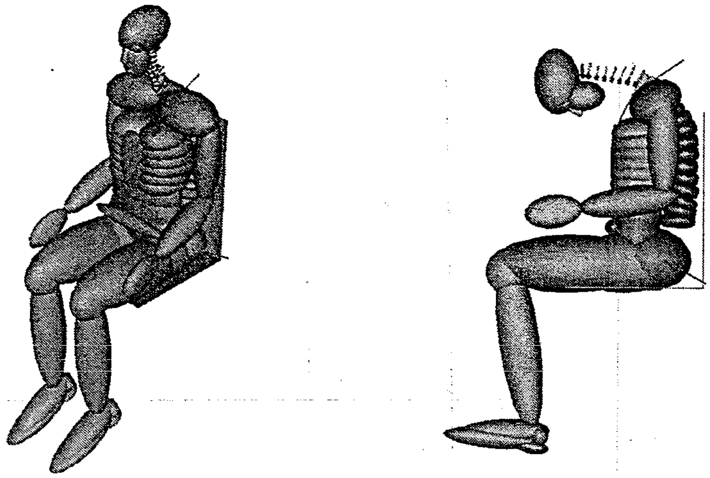


Figure 2.1: Initial and extreme intermediate position.

2.1.2 Validation and modification of the model.

To validate the model and to improve the model by changing parameters, a 15G sled test is used. In this test the acceleration vs time is according to figure [2.2]. The maximum occuring acceleration is 15 times the gravity.

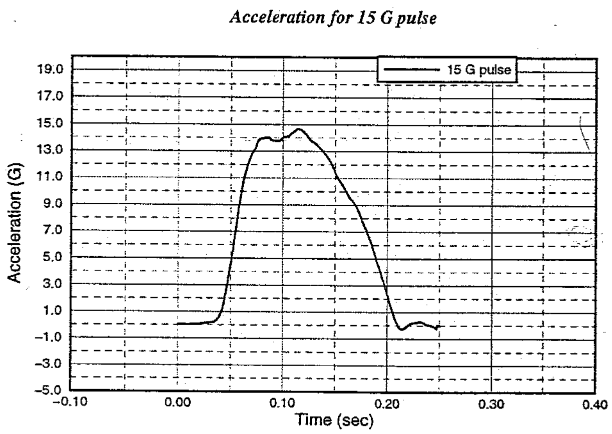


Figure 2.2: Sled acceleration for the 15G sled test.

The kinematics of T1 can be considered as the input for the head-neck model. For this reason first the dynamic behaviour of T1 must match with the experimental data. The behaviour is influenced by the stiffness parameters of the belt. Also the attachment points of the belt with the body have some influence. The kinematic behaviour of T1 will also be influenced by the mechanical properties of the neck. After changing the stiffness parameters in the neck, the behaviour of T1 will be checked again.

The best result for the T1 displacement in x-direction is found for the belt stiffness function in figure [2.3]. First a small range with low stiffness. In this region the volunteer is pulled tight in the belt. Then the belt will be tensioned. The stiffness in this region is much higher.

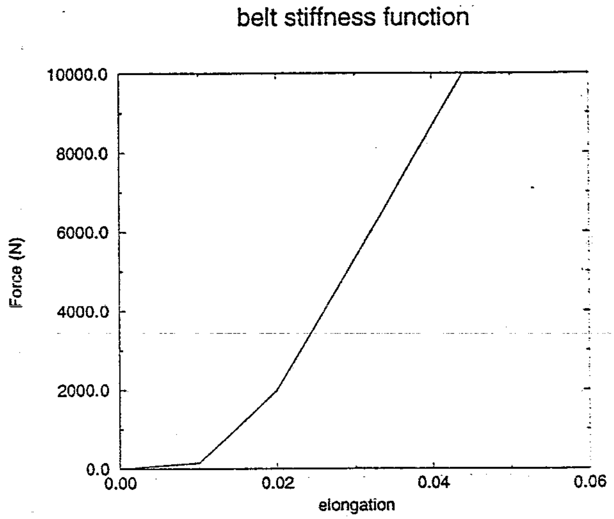


Figure 2.3: Belt Stiffness function

The T1 diplacement in X-direction and the T1 rotation are like the experimental data. The displacement in Z-direction is too high. Also the acceleration of T1 in X-direction is not conform the experiments. This can be the result of a poor fixation of the instrumentation. [See Thunnissen et.al.(1995)]

After the kinematics of T1 are as good as possible stiffness parameters for the neck joints for flexion/extension, tension/compression and anterior/posterior shear are modified. The signals that are used for changing the stiffness parameters are the displacement of the head centre of gravity in X-direction and Z-direction, the resulting linear acceleration of the head centre of gravity and the head rotation.

For the intervertebral joints the load-displacement curves exist of three regions:
1. Neutral Zone (NZ)
2. Elastic Zone (EZ)
3. Zone ouside the Range Of Motion (ROM)
In the NZ the forces and torques are close to zero. Then an elastic region EZ is observed. The neutral and elastic zone together form the Range Of Motion. Outside the range of motion the forces and torques become very large. The deformations will only exceed these boundaries very little.

The changes in the neck of the human body model are as follows.

- The tension/compression stiffness is modified. The NZ and EZ are made larger. The stiffnesses are lower.
- For flexion/extension the NZ and EZ are made smaller. (85 percent of the original values.) To avoid large changes in the stiffness an extra point is added. The values for c0c1 and c1c2 are not changed. Stiffnesses of these joints have little influence on the dynamical behaviour.

- The stiffness for anterior/posterior shear is changed. The NZ and EZ are made 1.5 times as large. Stiffness is the same.
- Rotational damping is increased. In the model of M. de Jager the damping coefficient did not have a fundamental basis because no reliable data are available. The coefficients were determined such that a satisfactory model response was obtained.

The influence of rotational damping on the calculation time is considerable. If damping is increased, then the calculation times become much larger. A compromise between calculation time and correlation between experimental data and model output is made. Increasing the chosen rotational damping leads to higher calculation times but it will hardly improve the correlation of test results and model output.

The changes are given in figure [2.4]. The dashed lines give the original stiffnesses of M. de Jager. The stiffness functions for flexion/extension were modified by A. v.d. Kroonenberg. Further modifications are made to improve the correlation of the output of the volunteertests and the model output. The functions that were found are given with the solid lines.

The following validation is done. The resulting linear acceleration of the head centre of gravity. The displacement of the head centre of gravity in X-direction and Z-direction and also the head rotation are determined and compared with the results of the volunteers. The head rotation is corrected for the rotation of T1. The experimental data of the Naval BioDynamics Laboratory are corrected for the errors in the T1 rotations that occur because the instrumentation for T1 is not mounted firmly to the spine. [see J.Thunnissen et.al. (1995)]
The experimental data and the model response are given in figure [2.5].

The accelerations are given in the local coordinate systems The translations and rotations are given in the sled coordinate system

The resulting linear acceleration of the head centre of gravity is conform the experimental data. Also the displacement of the head centre of gravity is good. This means that the trajectory of the head is good. The head rotation is too high. The reason for this can be that muscles must also be modelled.

2.1.3 Sensitivity analysis

To see the influence of the spine the model output is also determined for a model with a locked spine. It is clear that the quality of the kinematics of T1 is increased if the spine is modelled. The input for the head and also the kinematics of the head are much better.

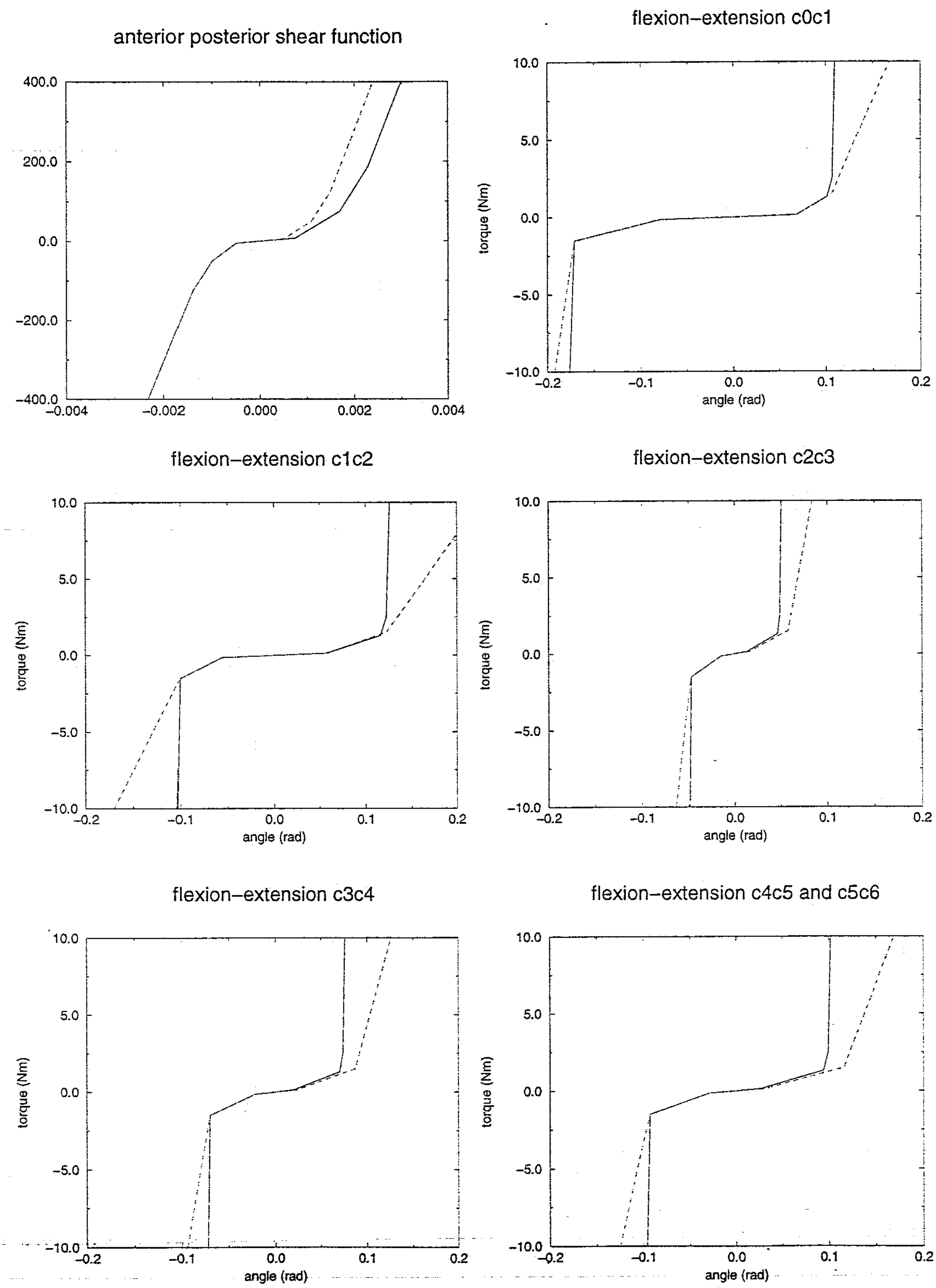


Figure 2.4a: Biomechanical functions for anterior/posterior shear, flexion/extension and tension/compression for cervical joints.

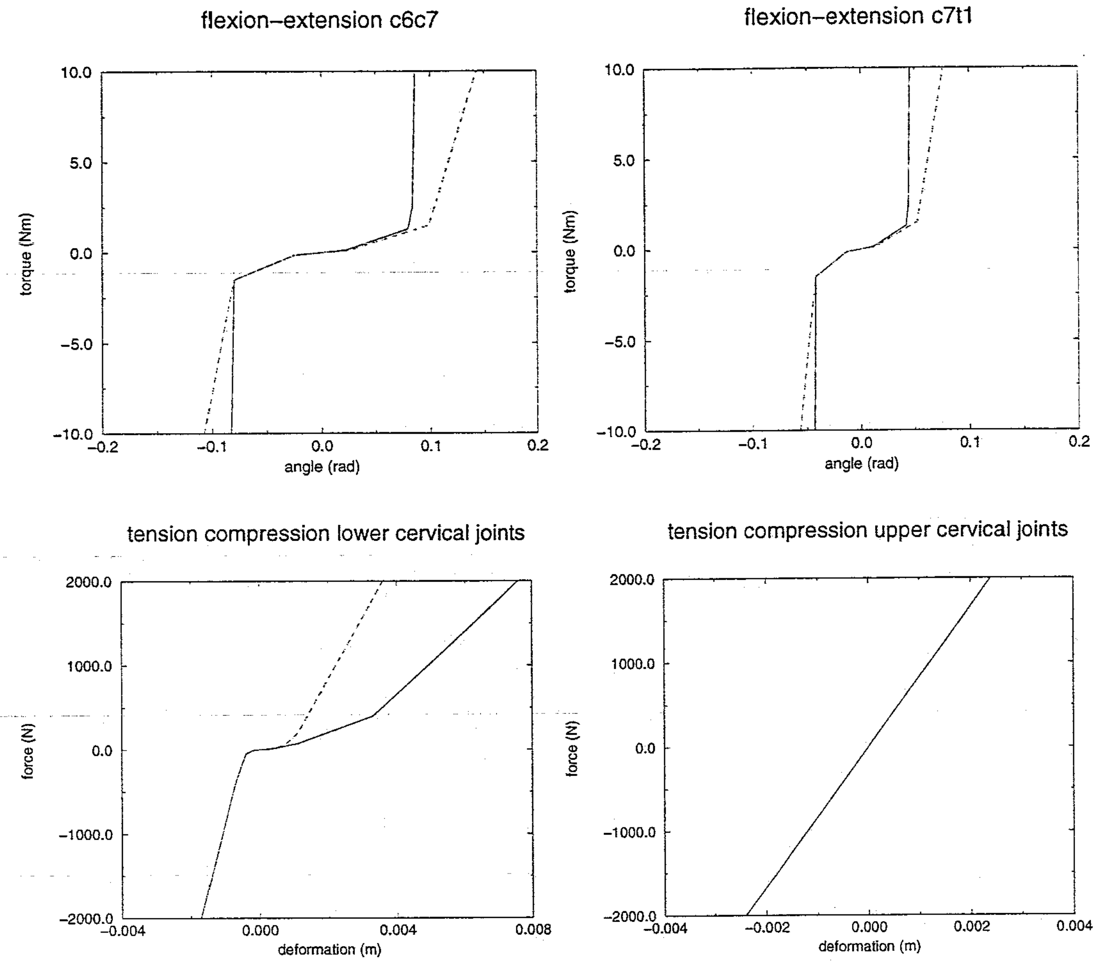


Figure 2.4b: Biomechanical functions for anterior/posterior shear, flexion/extension and tension/compression for cervical joints.

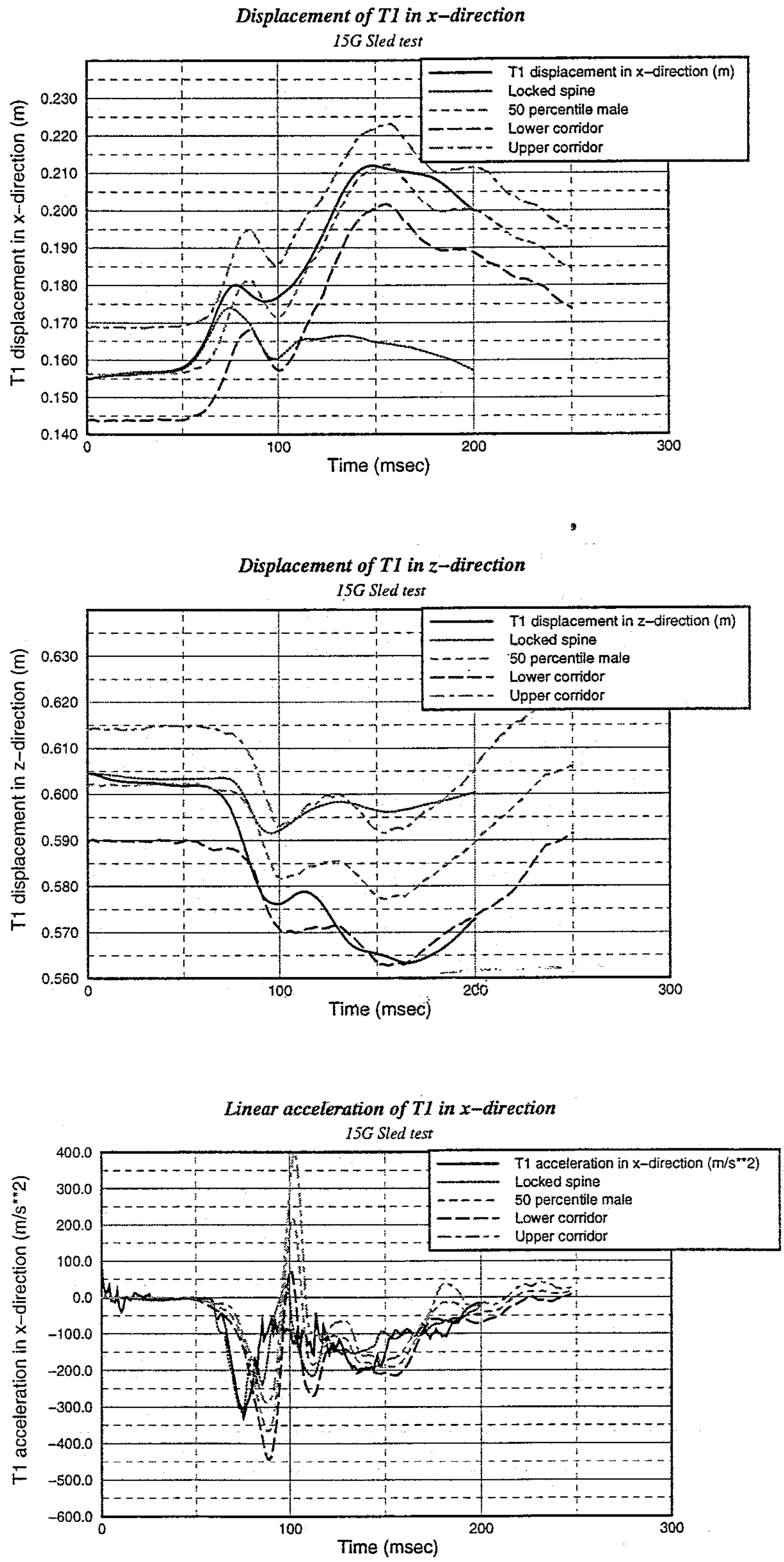


Figure 2.5a: Model response for 15G impact.

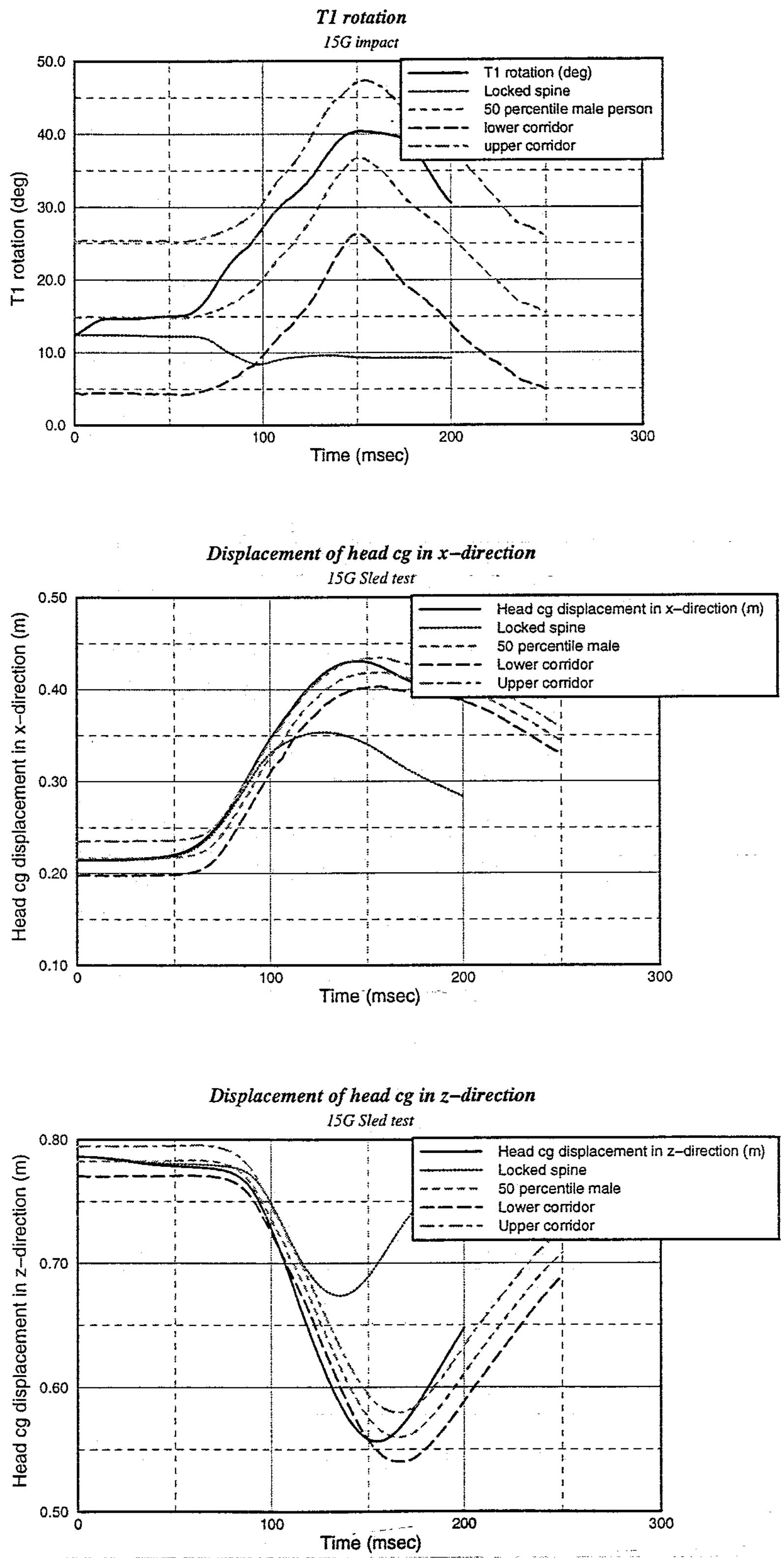


Figure 2.5b: Model response for 15G impact.

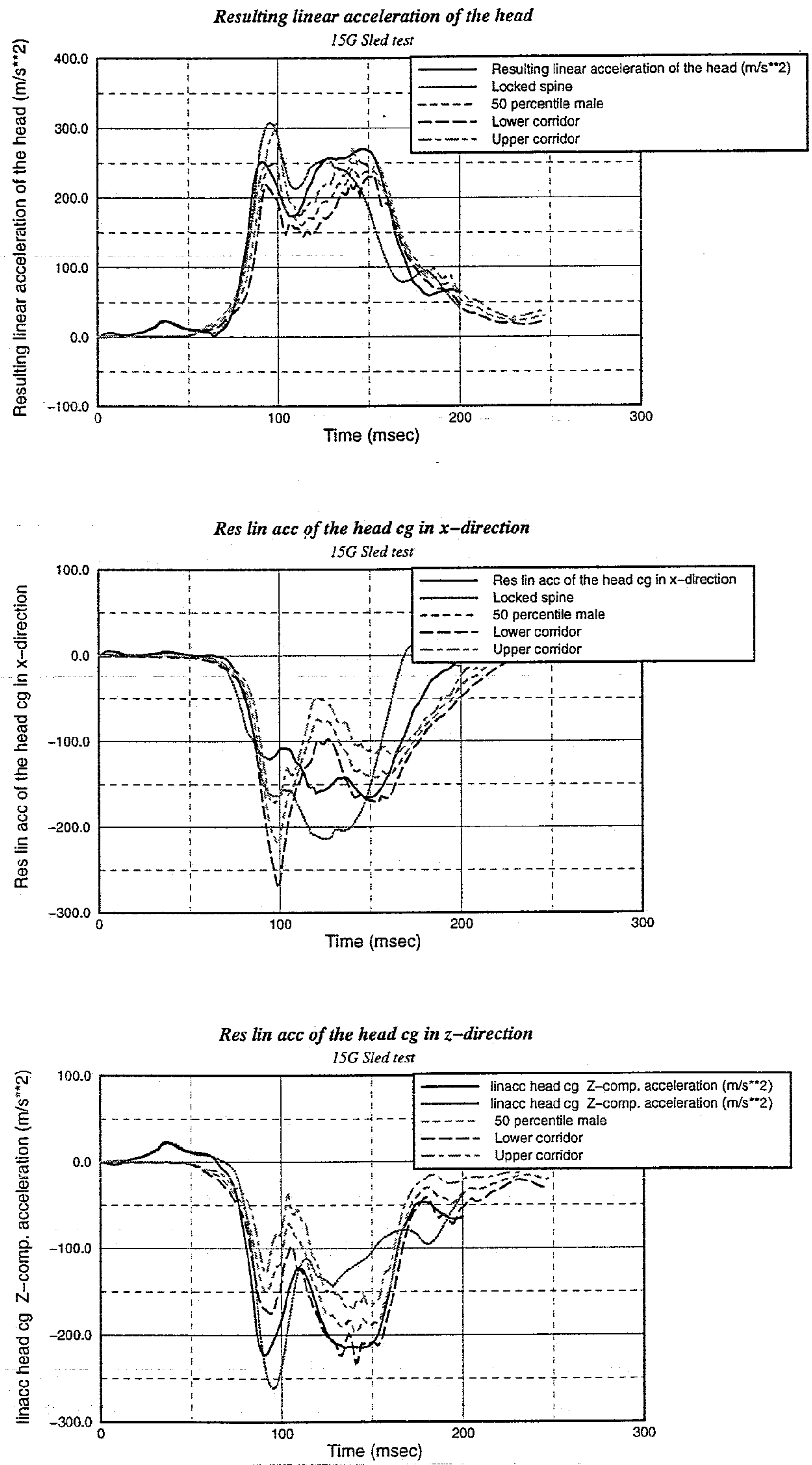
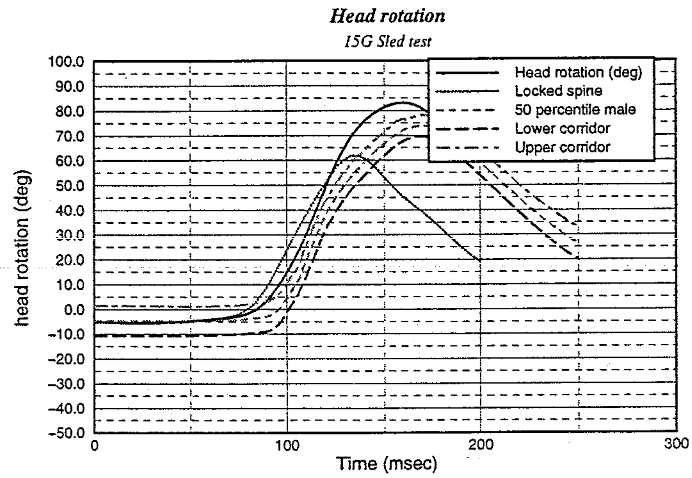


Figure 2.5c: Model response for 15G impact.



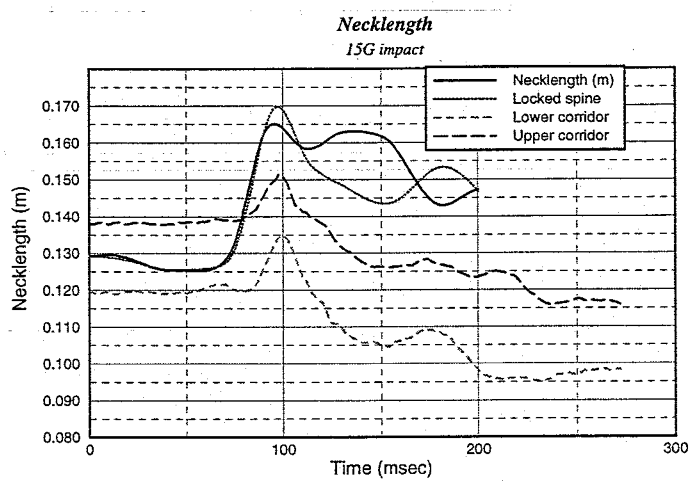


Figure 2.5d: Model response for 15G impact.

2.1.4 Lower severity frontal impacts

The human body model is also validated for lower severity frontal impacts. The test environment is similar to the test environment of the 15G sled test [see C.L. Ewing et.al. (1968)]. Only the impact velocities and the occuring accelerations are lower. The signals for which experimental data are available are the resultant linear acceleration of the head, the angular acceleration of the head and the head rotation. These signals have been validated for impacts with maximum occuring accelerations of 3G, 6G, 8G, 10G and 12G. The signal for the head rotation is not corrected for the rotation of the first vertebra T1.

The results for 3G and 10G impacts can be found in figure [2.6] respectively figure [2.7].

The results for 3G are not very accurate. The linear acceleration of the head of the model shows a peak at 40 ms. This peak is not observed at the volunteers. The peak also occurs if there is no acceleration pulse on the human model in x-direction. Also the angular acceleration of the head and the head rotation are not conform the experimental data. The model response is too high.
If there is no pulse, then the resulting linear acceleration of the head centre of gravity and the angular acceleration near the start are high and do not go to zero during time. The head drops forward. The head starts to move and to rotate. The translations and rotations increase. This happens rather slowly. See figure [2.8] for the results. The influences will be noticed for lower severity impacts. This can be clearly seen if the linear acceleration of the head center of gravity and the angular acceleration of the model will be compared for 3G impact and no impact. For higher impacts the influences are relatively small. The influences on the model response can be neglected.

For 10G the results are good to predict the response of the volunteer to the impact. Only the head rotation is too high. The head rotation predicted by the model is 83 degrees. The head rotation should be less than 74 degrees. This can be the result of the influence of the gravity on the human model.

For lower impact severities the model output is not good to predict the response of the volunteer. For higher impacts with occuring accelerations of 10G the output is like the experiments. A reason for this is that the model is not in a state of equilibrium. The model is not in a balanced position and will start moving because of the gravity and initial tension.

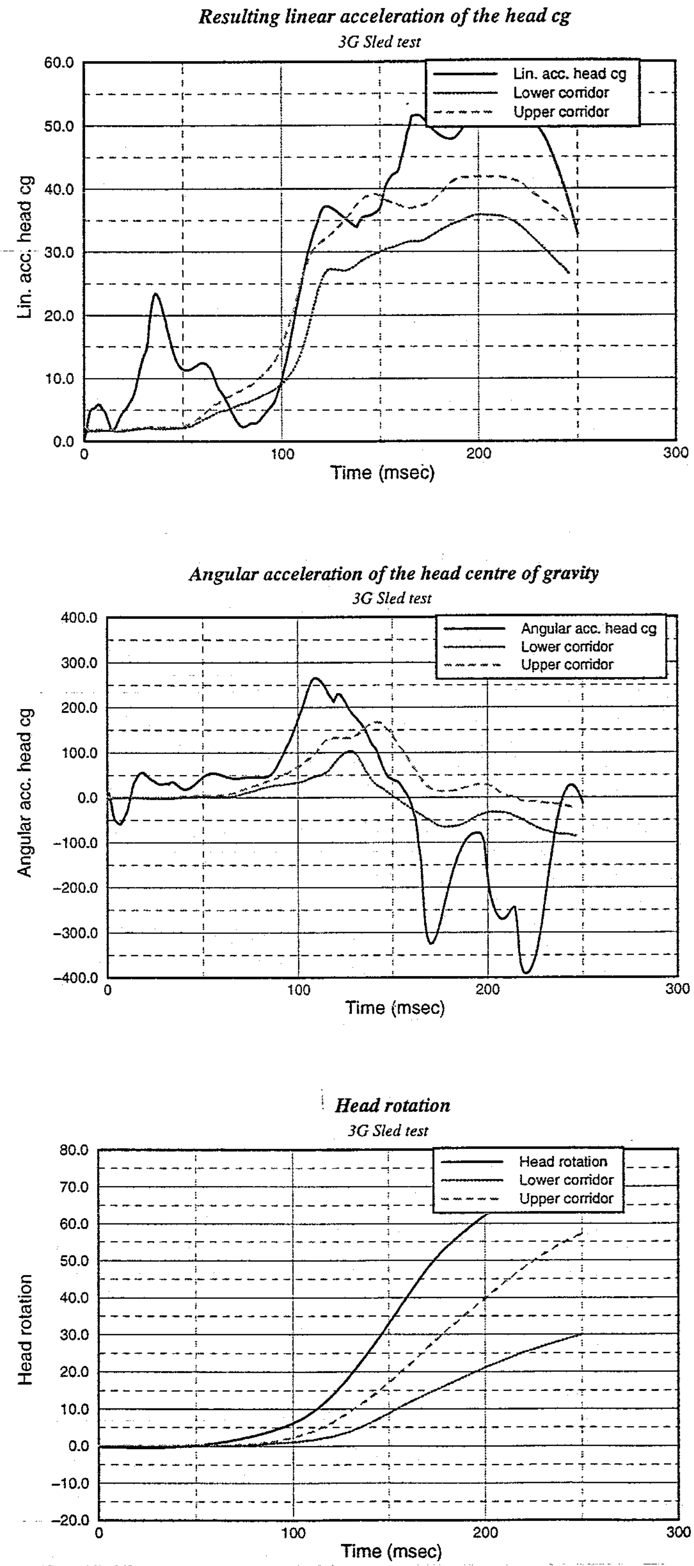


Figure 2.6: Response for 3G impact.

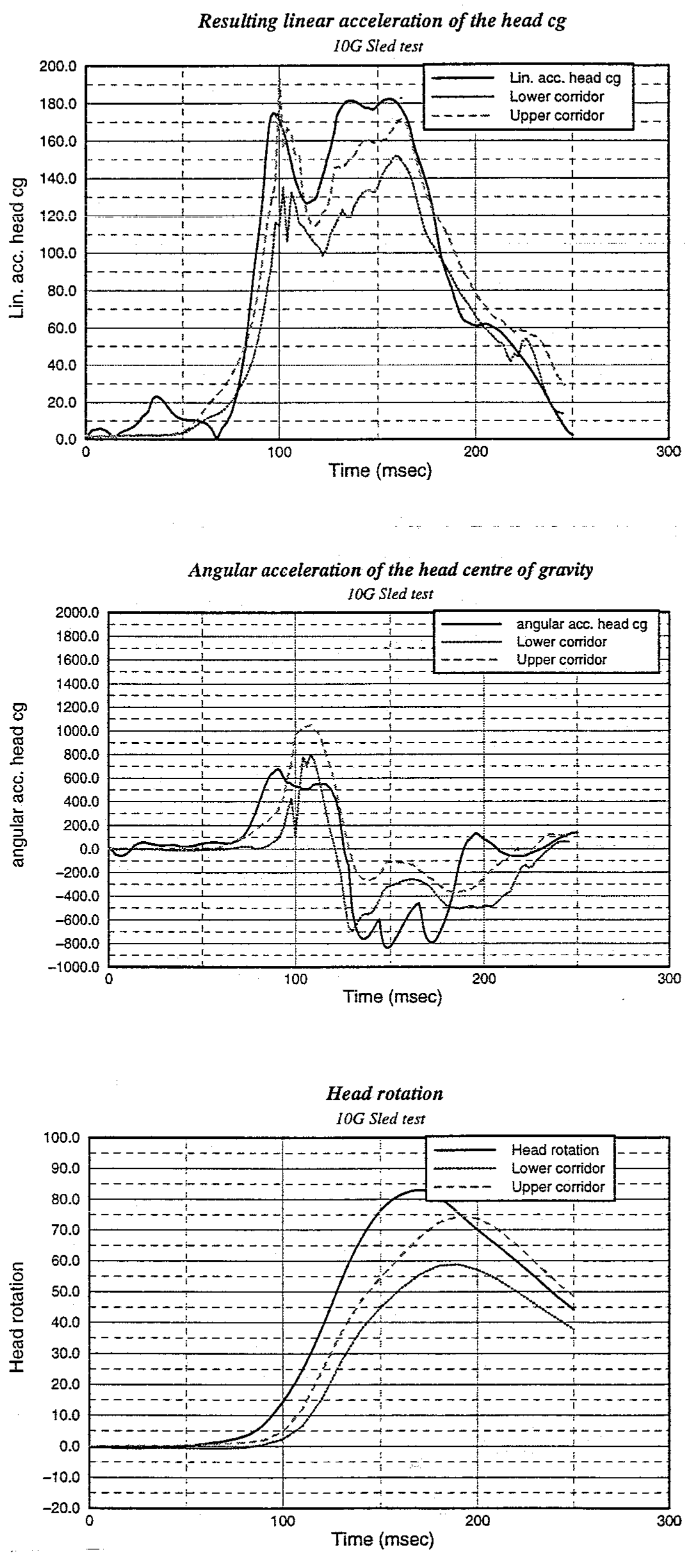


Figure 2.7: Response for 10G impact.

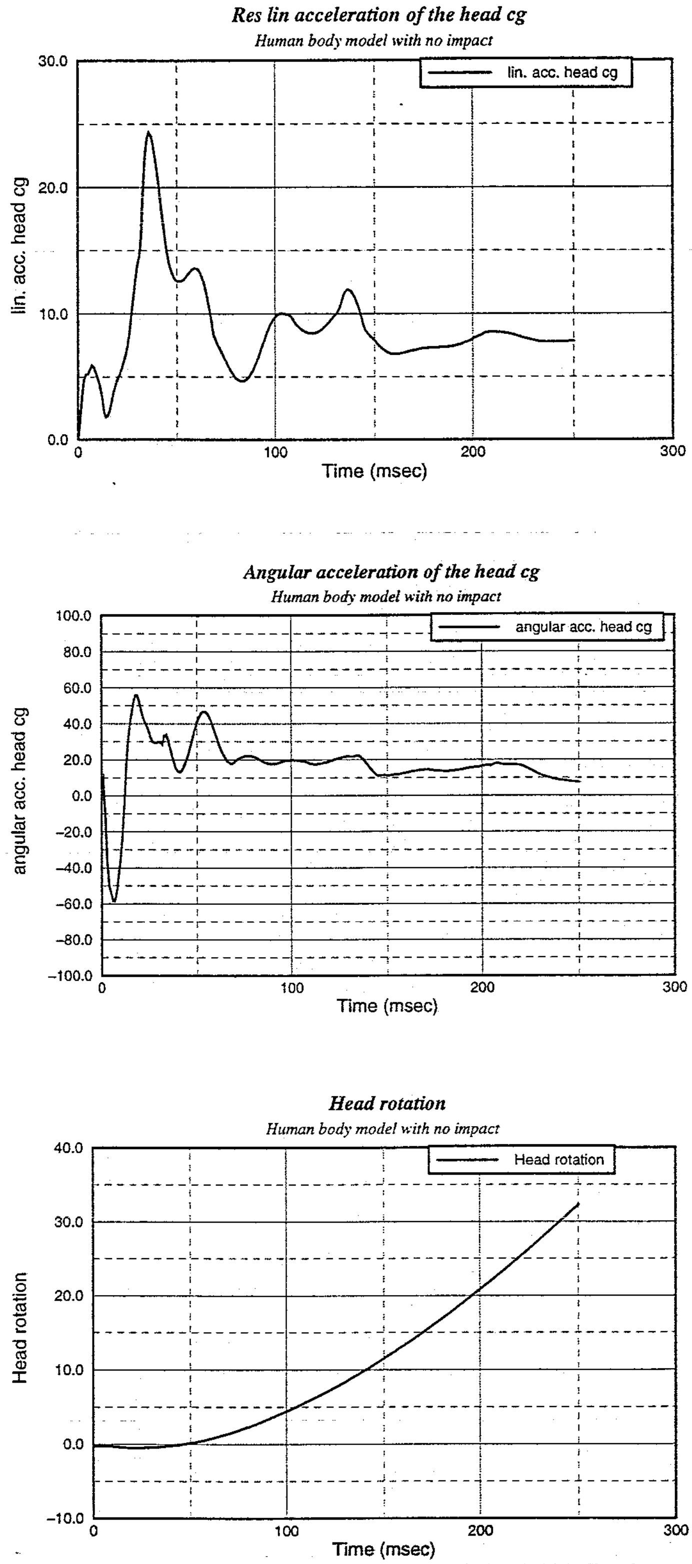


Figure 2.8: Response with no impact.

2.2 Validation for pendulum impact on the thorax

To assure biomechanical fidelity of the chest to blunt-frontal midsagittal impacts, performance guidelines have been introduced, see [article Motor Vehicle Safety Systems Testing Committee]. Meeting these guidelines does not assure biofidelity for other loading conditions. Specifically, it has not been demonstrated that these performance criteria are appropriate for frontal loading due to belts or airbags. It is however assumed that if the requirements following from these guidelines are satisfied, then there is some potential for extrapolation of the results.

2.2.1 The experiment

The human body model is placed in a sitting position on a flat, horizontal surface without back support. The arms and legs are extended horizontally forward and parallel to the midsagittal plane. The subject is placed in a position such that the surface of the thorax on the centerline of the impactor is vertical. The longitudinal centerline of the impactor has the same vertical height as the mid-sternum and lies in the midsagittal plane of the subject.
The impactor has got a cylindrical end of 152 mm in diameter, a flat face perpendicular to the longitudinal axis, and an edge radius of 12.7 mm. The mass including the instrumentation equals 23.4 kg. See figure [2.9].

The impact velocities for which requirements are given are 4.27 and 6.71 $[m/s]$ in the x-direction. Corridors for the resulting acceleration of the pendulum against the chest deflection are given. The chest deflection is the relative displacement of the sternum to the spine. The corridors are determined with the help of tests with human unembalmed cadavers. The response definition recommendations are adjusted upward to account for the lack of muscle tone in the cadaver subjects. These adjustments are based on limited volunteer testing.

The thoracic response and corridors for the 50th percentile male for pendulum impact velocities of 4.27 and for 6.71 $[m/s]$ are given in figure [2.10] respectively figure [2.11]. Also the response of the model of the Hybrid III dummy is given. The model output of the Hybrid III dummy model is not conform the requirements. The changes of model of the Hybrid III dummy that lead to the new human model further decrease the biofidelity for the mechanical response of the chest.

If the values of the stiffness parameters in the back are increased, then the resulting pendulum acceleration will be increased. Also the chest deflection will be increased. Also the model output for a locked back is plotted to see the influences of the back. The resulting pendulum acceleration and the chest deflection increase like expected.

If the stiffness of the chest will be decreased, then the resulting pendulum acceleration will be lower and the chest deflection will be higher. If the model is compared to the Hybrid III dummy, then it can be concluded that the stiffness of the chest of the new model is too high.

In the model of the Hybrid III dummy the sternum was connected to the ribs with the help of a translational joint. The ribs are connected to the upper torso with the help of a cardan restraint and a point restraint.
In the new model the ribs are connected to the back with the help of a cardan restraint and two point restraints. The cardan restraint is placed between the ribs and T10. The point restraints are placed between the ribs and L1 respectively the ribs and T6. The stiffnesses, damping and hysteresis of these point restraints are half the stiffnesses, damping and hysteresis of the point restraint in the Hybrid III dummy. The total mechanical behaviour should hardly be influenced by these modifications.
There are also point restraints attached between clavicles and T1 and between clavicles and ribs. These point restraints will influence the stiffness of the chest. The total stiffness of the chest is increased. Modifications are necessary to improve the biofidelity of the chest.

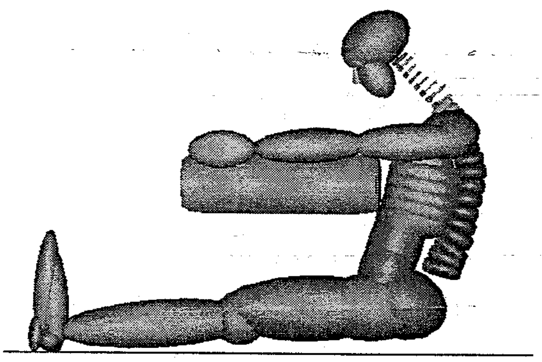


Figure 2.9: intermediate position.

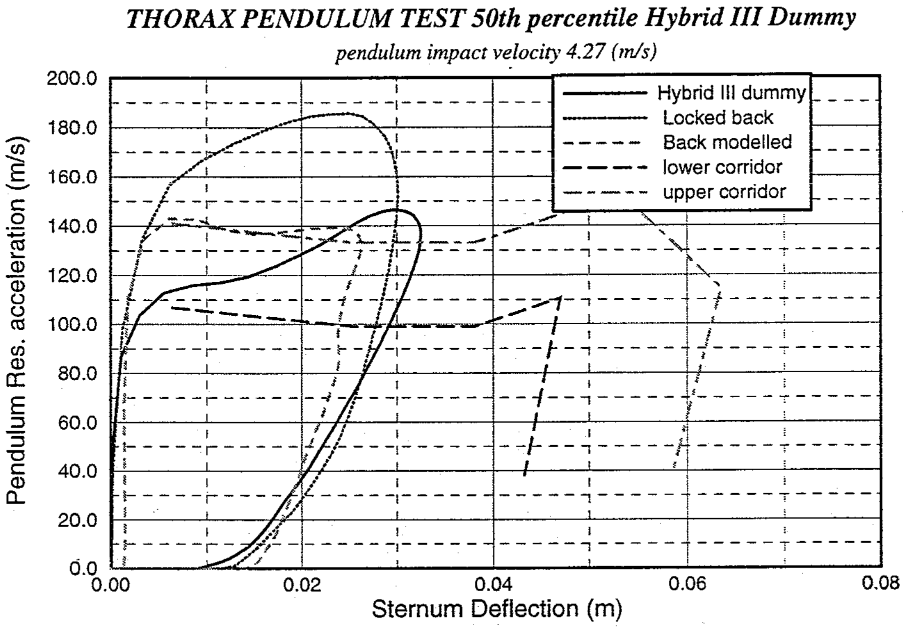


Figure 2.10: Pendulum test and corridors for impact velocity $4.27[m/s]$.

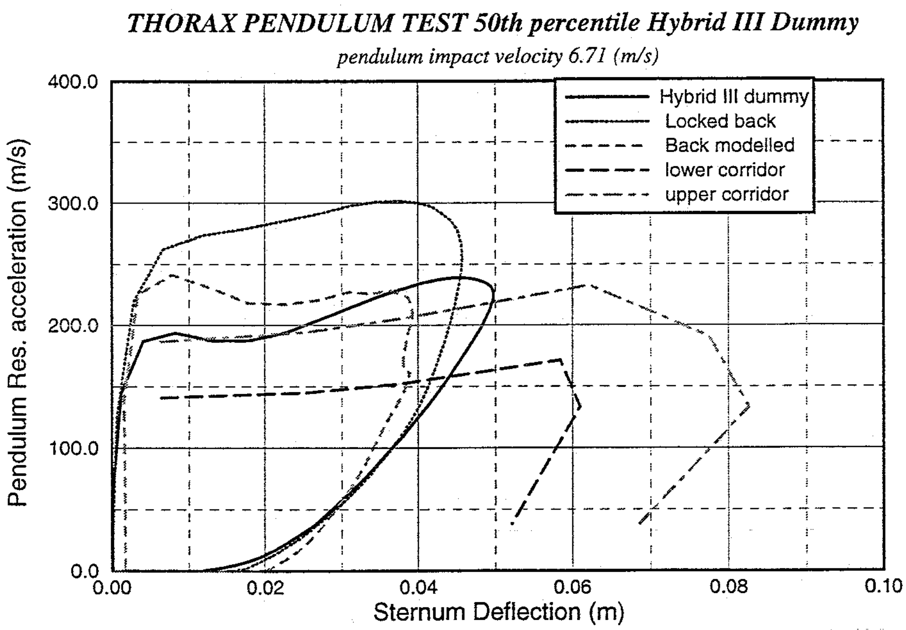


Figure 2.11: Pendulum test and corridors for impact velocity $6.71[m/s]$.

2.3 Validation for rear impact

There have been done a lot of human experiments for rear impacts. [See: J. Thunnissen (1997)]. With the help of existing human experiments response requirements for rear impacts have been derived [see: M.L.C. Hoofman et al. (not published)]. Experiments with head restraints are poorly documented. For this reason only experiments without head-restraints are used.
To obtain insight in the dynamic behaviour during rear-end impacts the model response is determined. The model is placed in the same environment and placed in the same position as during experiments. The model was seated on a seat with a known loading and unloading function for the seat back and seat cushion. There is no head restraint.

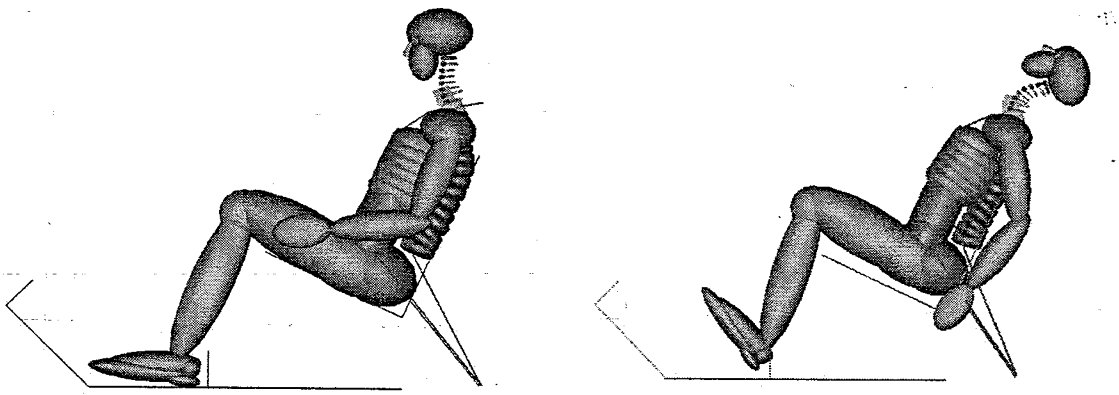


Figure 2.12: Initial position and extreme intermediate position.

To validate the model there are corridors for a few different signals. There are corridors for the head rotation vs time for two different acceleration pulses. A low severity acceleration pulse with an average acceleration of 32 $[m/s^2]$ and a high severity pulse with an average acceleration of 50 $[m/s^2]$. [see: M.L.C. Hoofman et al. (not published).] Also requirements for the maximum head rotation vs the average acceleration and the Mertzmoment vs head rotation are given by Hoofman et al.. The Mertzmoment is the moment of force at the occipital joint.

The model response is determined for the high and low severity acceleration pulses. The model output is given in the figures (2.13) to (2.17).

The results must satisfy the requirements for low and high severity rear impact. The modifications made in chapter 2.1.2 do not influence the model output for rear impact. The head rotation vs the average sled acceleration is good according to the corridors given by Hoofman et al.. The head rotation against time is according the corridors. The moment of force at the occipital joint against the head rotation is not conform the requirements. For the high severity rear impact the output is outside the corridors. This is not the result of the changes in stiffness parameters in the neck.

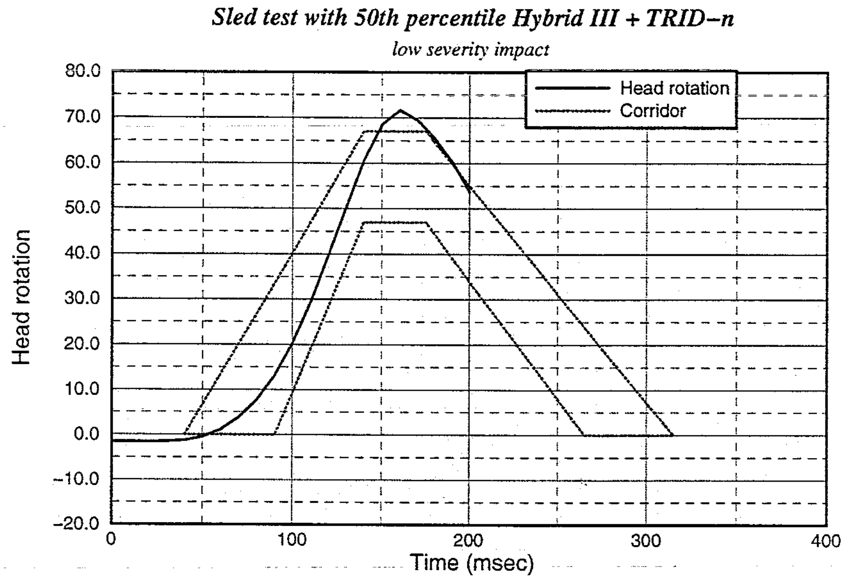


Figure 2.13: Head rotation for low severity rear impact.

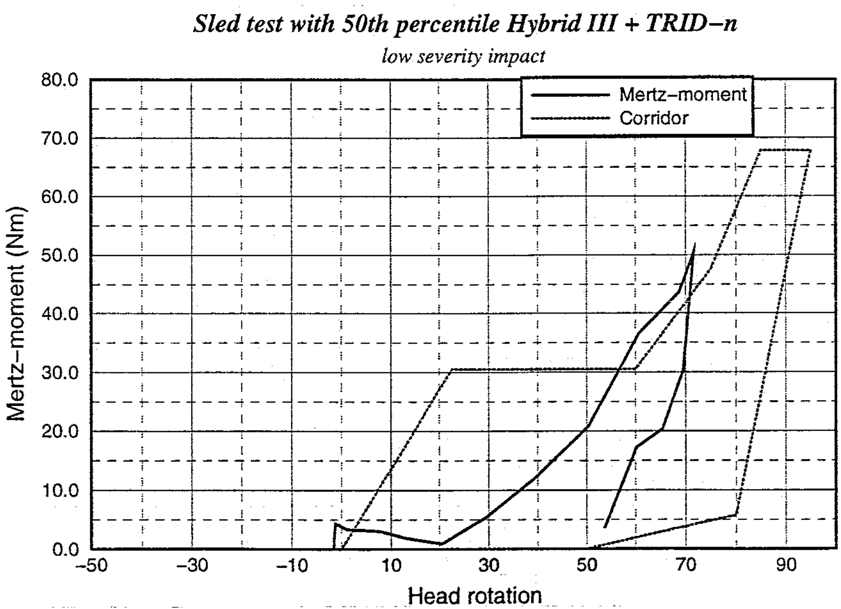


Figure 2.14: Mertz-moment vs head rotation for low severity rear impact.

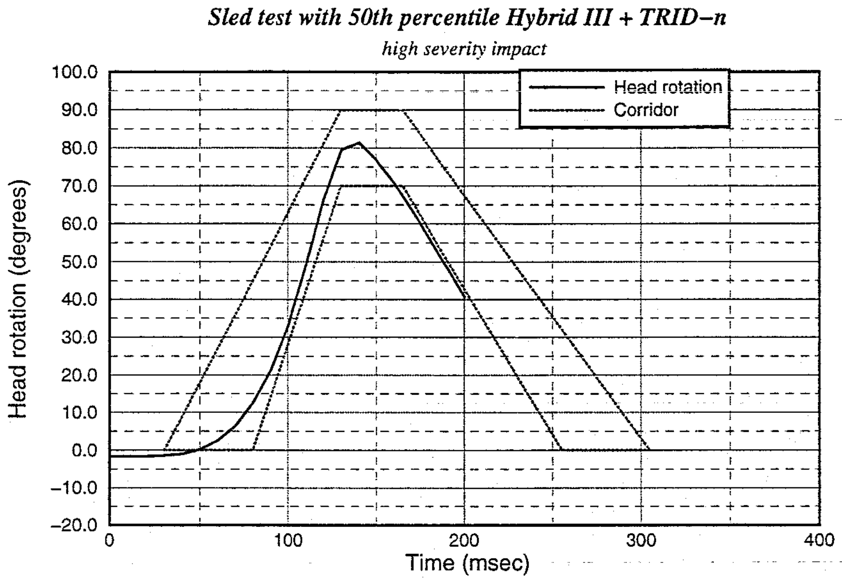


Figure 2.15: Head rotation for high severity rear impact.

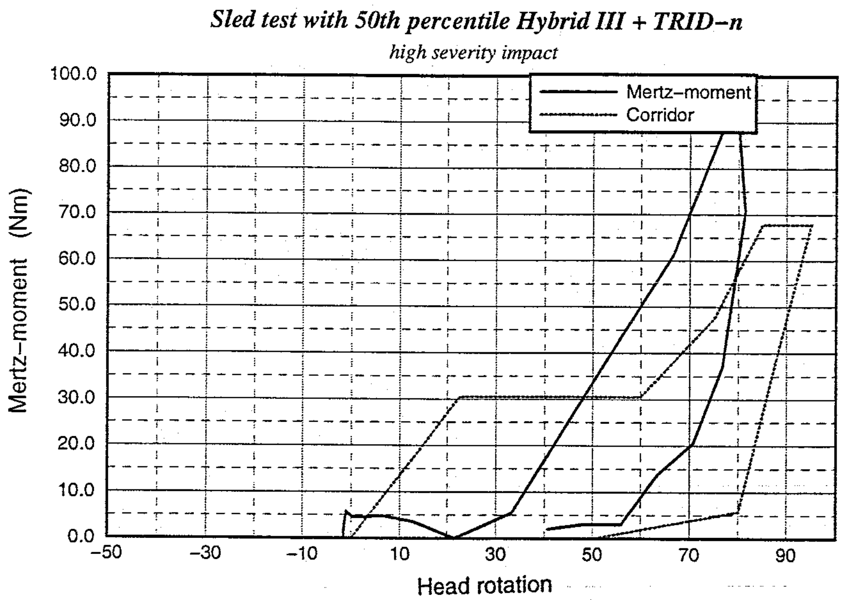


Figure 2.16: Mertz-moment vs head rotation for high severity rear impact.

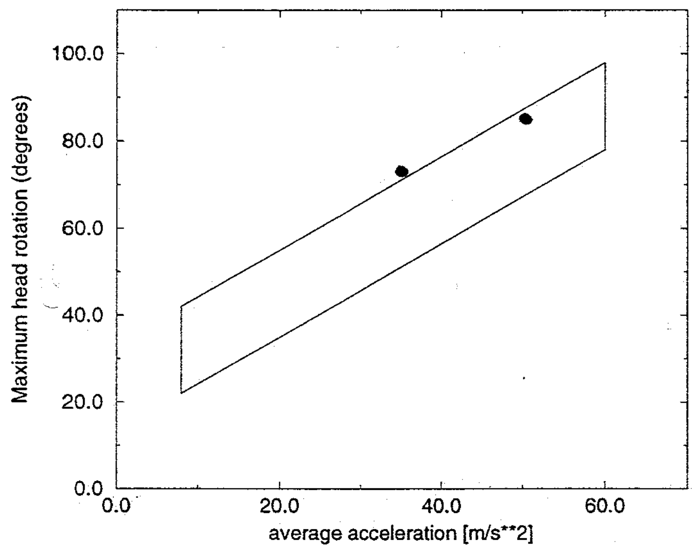


Figure 2.17: Maximum head rotation against average acceleration.

Chapter 3

Discussion

The total human model has been validated for 15g frontal impact. The behaviour is determined by three important parts. First of all by the belt. The second component is the model of the human spine and the third component is the head neck model. There is a lot of interaction between these parts and it is difficult to determine the influences of one part. For this reason the model response is also determined for the model with a locked spine. The influences of a flexible spine compared to a locked spine can be clearly seen in section 2.1. Figure 2.5.

The mechanical properties of the individual components must be handled with care. If the parameters in one component are changed, then the modification of another component can lead to a model output that still satisfies the requirements. The model response can be adequate to predict the dynamic behaviour of a 50th percentile male, but the mechanical properties are not neccesary like reality.

The mechanical behaviour of the in vitro spine is different from the mechanical behaviour in vivo. For this reason the stiffness parameters that result from experiments must be handled with care.

The mechanical stiffnesses for the head-neck model that are used are determined with quasistatic tests. During impact the deformations take place in a short time. It is reasonable to assume that the mechanical properties of the intervertebral joints are different under these dynamical conditions.

Another reason why the mechanical properties of the joints could be different is that it is assumed that the joint displacements and rotations do not interact. Complex displacements are supposed to be a superposition of displacements in each degree of freedom. The behaviour of one degree of freedom does not depend on displacements in other degrees of freedom. As shown by Panjabi for the upper cervical joints [see: Panjabi et.al. (1993)] this is unrealistic. The joint stiffness for a single degree of freedom also depends on the other degrees of freedom. Sufficient experimental data, however, is not available.

Also coupling behaviour is not present in the model. Main displacements in one degree of freedom do not result in coupled displacements in the other degrees of freedom. In reality two characteristic couplings are present in the lower cervical spine. Flexion/extension is coupled with anterior/posterior translation and lateral bending is coupled with axial rotation. In the upper cervical spine axial rotation of the atlas is coupled with vertical translation of the atlas.

Another simplification in the model is that the mechanical behaviour of a few joints are supposed to be a superposition of the behaviour of the seperate joints. Muscles and ligaments have influences on the ROM and the mechanical behaviour of combination of joints. A preload of the muscles also lead to other dynamical behaviour.

In the begin position the model is not in a state of equilibrium. The dynamical behaviour is influenced by the gravity and the model will start moving even if there is no pulse in x-direction. This will influence the model output, especially for low impacts. One way to solve the problem of moving under lower loading conditions is to model the muscles and keep the model in its original position. The model is then more like reality but higher calculation times will occur. Modelling of the muscles would lead to a more realistic model [see: M.J. van der Horst (1997)]. Calculation times however will be higher.

Because of the changes that were made in the mechanical properties of the global neck model of M. de Jager the biofidelity of the model is improved compared to the model of A. van den Kroonenberg. The modelling of the spine leads to a good input for the head-neck model. For this reason it can be assumed that the modelling of the spine is usefull to improve the model. The biofidelity of the resulting human model for impacts on the chest is poor according to the corridors given in section 2.2. The stiffness of the chest is too high. The resulting human model is good to predict the global motions of a 50th percentile male in frontal and rear impacts. The model response correlates good with the volunteer tests.

Bibliography

Janet Brelin-Fornari, Yih-Charng Deng and Ara Arabyan.: "Validation of a Human Head/Neck Model Using MADYMO." North American MADYMO User's Meeting. General Motors Research and Development Center and University of Arizona.

Channing L. Ewing, D.J. Thomas, G.W. Beeler, L.M. Patrick and D.B. Gillis (1968): "Dynamic Response of the Head and Neck of the Living Human to -Gx Impact Acceleration." Proc. 12th Stapp Car Crash Conference. SAE-paper 680792.

Rene van Haaster, (August 1996): "Modelling of the Human Head-Neck System in Impact Conditions." Master's Thesis, TNO Crash Safety Research Centre Delft, Eindhoven University of Technology.

M.L.C. Hoofman, J.G.M. Thunnissen, M.R. van Ratingen and J.S.H.M. Wismans, (June 1997): "Requirements for a Low and Mid Severity Rear Impact Dummy (RID)." TNO Crash-Safety Research Centre.

M.J. van der Horst, J.G.M. Thunnissen, R. Happee, R.M.H.P. van Haaster and J.S.H.M. Wismans, (1997): "The Influence of Muscle Activity on Head-Neck Response During Impact." draft STAPP, Eindhoven University of Technology and TNO Crash Safety Research Centre Delft.

Yue Huang, Albert I. King and John M. Cavanaugh, (May 1994): "A MADYMO Model of Near-Side Human Occupants in Side Impacts." Journal of Biomechanical Engineering, Bioengineering Center Wayne State University, Detroit.

Hirotoshi Ishikawa, Janusz Kajzer and Gunter Schroeder, (1993): "Computer Simulation of Impact Response of the Human Body in Car-Pedestrian Accidents." Japan Automobile Research Institute, Chalmers University of Technology and Hannover Medical Univ..

Marko de Jager, (May 1996): "Mathematical Head-Neck Models for Acceleration Impacts." PhD thesis, Eindhoven University of Technology.

M. de Jager, A. Sauren, J. Thunnissen and J. Wismans, (Nov 1994): "A Three-Dimensional Head-Neck Model: Validation for Frontal and Lateral Impacts." 38th STAPP Car Crash Conference. Eindhoven University of Technology and TNO Crash Safety Research Centre.

I.A. Kapandji (1974): "The physiology of the joints. Volume 3." Churchill Livingstone, Edinburgh London and New York.

Charles K. Kroell, Dennis C. Schneider and Alan M. Nahum.: "Impact Tolerance and Response of the Human Thorax II." STAPP 74 Biomedical Science Dept. General Motors Research Laboratories and University Hospital, San Diego.

A. van den Kroonenberg, J. Thunnissen and J. Wismans, (June 1997): "A human model for low-severity rear-impacts." IRCOBI Conference, TNO Crash-Safety Research Centre.

Deren Ma, Louise A. Obergefell, Annette L. Rizer, (1995): "Development of Human Articulating Joint Model Parameters for Crash Dynamics Simulations." Stapp-Conference 1995, Systems Research Laboratories Inc. and Department of the Air Force.

Raymond F. Neathery.: "Analysis of Chest Impact Response Data and Scaled Performance Recommendations." STAPP 74, Biomedical Science Dept. General Motors Research Laboratories.

P. Prasad and A.I. King, (1974): "An experimentally validated dynamic model of the spine." J. Appl. Mech. pp. 546-550.

Panjabi, M.M., T. Oda, J.J. Crisco III, J. Dvorak and D. Grob, (1993): "Posture affects motion coupling patterns of the upper cervical spine." Journal of Orthopaedic Research, 11:525-536.

A. Shirazi-Adl and M. Parnianpour, (Februari 1996): "Stabilizing Role of Moments and Pelvic Rotation on the Human Spine in Compression." Journal of Biomechanical Engineering, Ecole Polytechnique in Montreal, Quebec and Ohio State University in Columbus, Ohio.

J. Thunnissen, (1997): "Human Head-Neck response in extension: a literature review." TNO report 97.BV.OR.021.1/JTH, EC Whiplash Project, Brite Euram III RTD Project, Brussel

Jan G.M. Thunnissen, Michiel R. van Ratingen, Marc C. Beusenberg and Edgar G. Janssen, (May 1996): "A dummy neck for low severity rear impacts." 15^{th} International Technical Conference on the Enhanced Safety of Vehicles, Melbourne, Australia.

J. Thunnissen, J. Wismans, C.L. Ewing and D.J. Thomas, (1995): "Human Volunteer Head-Neck Response in Frontal Flexion: A New Analysis." TNO Crash Safety Research Centre Delft, Eindhoven University of Technology.

TNO Crash Safety Research Centre, Delft, The Netherlands. MADYMO User's Manual, Version 5.2 1996

J. Wismans, M. Philippens, E.v. Oorschot, D. Kallieris and R. Mattern, (1987): "Comparison of Human Volunteer and Cadaver Head-Neck Response in Frontal Flexion." Stap

1987, TNO Road Vehicles Research Institute Delft, Institute for Forensic Medicine University of Heidelberg, West Germany.

Report of the Motor Vehicle Safety Systems Testing Committee, (March 1985): "Human Mechenical Response Characteristics." SAE J1460 Mar85.