

SIMULATION OF REAR END IMPACT WITH A FULL BODY HUMAN MODEL WITH A DETAILED NECK: ROLE OF PASSIVE MUSCLE PROPERTIES AND INITIAL SEATING POSTURE

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ABSTRACT

To study the mechanics of the neck during rear end impact, in this paper an existing global human body model and an existing detailed submodel of the neck were combined into a new model. The combined model is validated with responses of volunteers and post mortem human subjects (PMHSs) subjected to rear end impacts of resp 5g and 12g. The volunteers (n=7, 7 tests) were seated on a standard car seat with head restraint, while the PMHSs (n=3, 6 tests) were placed on a rigid seat without head restraint. The model shows good agreement with the PMHS responses when muscle tensile stiffness is increased towards published PMHS tissue properties. For the volunteer simulations, initial seating posture and head restraint position were found to strongly influence the model response. More leaning forward (increasing of horizontal distance head head restraint) results in larger T1 and head motions. A correct vertical position of the head restraint (top of head in one line with top of head restraint) reduces the head extension angle. The model has the potential to study injury mechanisms.

INTRODUCTION

Neck injuries resulting from rear end impact rank among the top vehicle safety problems and have serious implications for society. In order to get more insight in human behaviour during impacts, mathematical models of the real human body can be used. They offer biofidelity for a wide range of conditions and allow the study of aspects like posture, body size, muscular activity, and injury mechanisms. In addition, detailed human body modelling allows analysis of injury mechanisms at tissue level. In the past, a large number of models of the entire human

body were published. Also, biomechanical models of the neck with varying complexity are presented in the literature. Two-pivot lumped mass models are the simplest models in which head and torso are modelled as rigid bodies connected by a rigid or extensible neck-link [1,2,3,4]. Multibody models can be regarded as an extension of lumped mass models, incorporating anatomical details [5,6,7,8]. The head and vertebrae are modelled as rigid bodies, whereas the soft tissues (e.g. ligaments, intervertebral discs, and muscles) are modelled as massless spring-damper elements. These models are detailed enough to describe the loads and deformations of the tissues and can be used to evaluate injury mechanisms. Finite element models, e.g. [9,10,11] allow even more detailed representations of geometry and material behaviour of the cervical spine, making it possible to study the stress strain behaviour of the tissues. Multibody models are computationally more efficient than finite element models, which will enhance their practical usefulness.

In this study a previously global human body model; was extended with a detailed submodel of the neck complete with detailed occipital condyles (OC), facet joints, intervertebral discs, ligaments and muscles. Ultimately, this model is to be used to study neck injury mechanisms, but first the model should be validated. Therefore, the main objective of this paper is:

- ◆ To validate a full body human model with a detailed multibody neck using existing rear end sled experiments performed on PMHSs and volunteers.

The utility of the cadaver response in representing the human response for high severity impact is governed

by the anthropometric similarity of the cadaver and human and the degree to which the constitutive properties of the cadaver tissues match those of human tissue. While the tissues of bone ligament, tendon and skin undergo small changes in mechanical properties post mortem, skeletal muscle stiffness is a source of uncertainty [12]. Results by van Ee [12] demonstrated that postmortem post-rigor handling of cadaveric tissue prior to testing greatly affects muscle properties. The immediate postmortem response was not found different from the live passive response. The post-rigor muscle response was unrepeatable but stiffer than the immediate postmortem or live passive response. After ‘preconditioning’ (repeating elongation tests on muscle tissue until the peak force varied by less than 2 percent), the response was repeatable but was significantly less stiff than perimortem and live passive muscle. Therefore, the second objective of this paper is:

- ◆ To study the effect of the postmortem change of passive muscle properties on neck response in rear end impact.

In simulating the volunteer experiments, both the role of muscle activation and exact seating posture are uncertain. Although the detailed neck model has the capability to simulate muscle activation and active muscle behaviour seems essential to describe accurately the human head neck responses of live volunteers [8] in this study only passive muscles are simulated. The role of active muscle behaviour in rear end impact will be published elsewhere [13].

From literature [14,15,16] it is known that initial seating posture and head restraint position are important parameters of the human head neck response. Therefore, the third objective is formulated:

- ◆ To study the influence of the initial seating posture and the vertical position of the head restraint for simulations in a standard car seat.

METHODS

A detailed neck model is included in a full body human model. Both MADYMO models are described below, followed by a short description of the experiments used for validation. Finally the simulation method will be explained.

HUMAN BODY MODEL – A mathematical human body model representing a 50th percentile male has been developed (Figure 1). The human geometry was

obtained from RAMSIS anthropometric data, which provided a realistic surface description, in particular for seated automotive posture. A 50th percentile male model from RAMSIS with 1.74 m standing height and 75.7 kg weight has been chosen. Detailed descriptions of the model and frontal lateral and rear end impact validation can be found in [17,18,19,20,21].

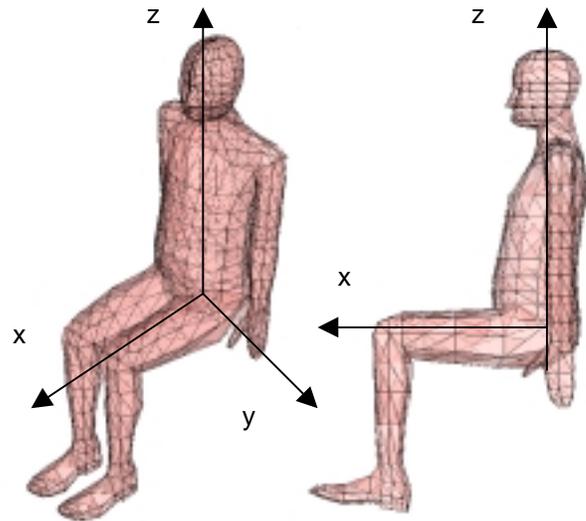


Figure 1. Human body model representing a 50th percentile male, erect seating position.

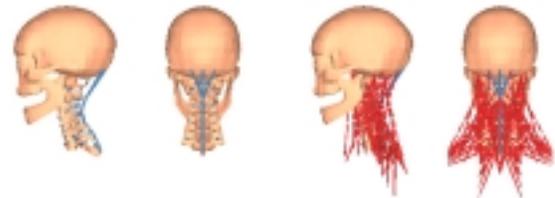


Figure 2. Detailed neck model. Lateral and rear view, muscles invisible in left view.

DETAILED SUBMODEL OF THE NECK – A detailed submodel, representing the human cervical spine from T1 until the skull has been developed (Figure 2). The model was integrated into the model of the entire human described above. Earlier versions of the neck model are published in [6,7,8]. Rigid bodies represent the skull and vertebrae. The geometry of the vertebrae and the skull is based on several studies reviewed by De Jager [6]. The surface description of skull and vertebrae is based on a PMHS scan and on anatomical textbooks and is implemented as “arbitrary surfaces”. These surfaces consist of triangular or quadrangular facets, which are

supported by nodes (vertices) on rigid bodies [25]. Contact can be simulated with other arbitrary surfaces, with ellipsoids, planes or with finite elements. The geometry is refined for the contact areas of the dens of C2 and the occipital condyles (OC). Translational and rotational degrees of freedom of adjacent vertebrae are coupled through nonlinear springs and dampers, representing the discs, two-dimensional nonlinear viscoelastic ligaments, frictionless facet joints (spring damper elements for compression) and frictionless contact for C2-dens-C1 and OC contact (stress strain contact) and Hill-type muscles.

Compared to the earlier neck model [6,7,8] the geometry has been refined. This resulted in new defined locations of the facet joints, muscles and ligaments. Also the ligament stiffness is updated and the disc stiffness for compression, flexion and extension has been improved using recent biomechanical data. The modelling of facet joints has been changed as well. Finally the muscle physiological cross sectional areas and the maximum isometric stress are updated. For a detailed model description and extended model validation the reader is referred to [13].

VALIDATION

Two test series performed within the European Community funded whiplash project [22] are used for model validation:

1. PMHS experiments performed at $\Delta V = 10$ km/h by the Laboratory of Accidentology and Biomechanics (LAB), France [10,23]
2. Volunteer tests performed at $\Delta V = 9.5$ km/h by Allianz Zentrum für Technik (AZT) in Ismaning, Germany [24]

Table 1 shows a summary of the properties of the tests used.

LAB PMHS TEST SETUP – The experimental set-up was designed to be both simple and reproducible, using a rigid seat. The seat back and seat panel had an inclination of 25° and 10° respectively. The subjects were restrained to the seat by three belts, restraining the thighs, pelvis and thorax tightly to the seat. Since the head-neck position was not stable, an electromagnet was used to keep the head in the initial position (i.e. Frankfort plane horizontal, see Figure 3). The rigid seat had no head restraint. The conditions in which these tests were performed were highly reproducible, resulting in quite consistent

responses of the PMHSs. The reproducibility of the test was demonstrated by performing two tests on each subject. Furthermore substantial similarities were observed in the responses of all three subjects.

	LAB [10,23]	AZT [24]
Subjects	PMHS	Volunteer
Seat type	Rigid	Standard
Head restraint	No	Yes
Belt system	single belts over limbs, pelvis and thorax	3-point belt with retractor
ΔV (km/h)	10	9.5
Max sled acc (g) $1g=9.81m/s^2$	12	5
Total number of subjects	3	7
Total number of tests	6	7
Av. subject mass (kg)	50	75
Av. subject height (m)	1.64	1.80

Table 1 Properties of the tests used.

AZT VOLUNTEER TEST SETUP- In the AZT experiments a standard car seat has been used, which was selected in the European whiplash project. The car seat was mounted on a sled. The seat back angle was set to 25 degrees using an H-point manikin according to regulation SAE J826 §4.3. The head restraint was positioned so that the top of the head and the head restraint are aligned. If this was not possible due to the subject's height, the maximum head restraint height was taken. The volunteers were asked to take a normal automotive passenger posture (see Figure 4). The Frankfort plane was initially horizontal, which implies an initial head angle of 0 degrees.

SIMULATION – In the simulations of the experiments the outer surface of seat, floor and foot planes were implemented as “arbitrary surfaces”. The load deflection curves and joint characteristics of the standard seat were based on quasi-static experiments of the seat performed within the European whiplash project. The position of the seat, head restraint and the belt attachment points were derived from photos of the experimental setups. The belts in the LAB experiment were modelled as single spring elements, while for the AZT test set-up a three point finite element belt model [25] was included to restrict the rebound of the model.

In the simulations the human body model was positioned based on photos of the experiments. The simulations were organized such that before the beginning of the simulated experiment the human body model was allowed to sink into the seat (rigid or standard soft seat) to find an equilibrium position from a position just above it. The electromagnet used in the LAB experiments to keep the head of the PMHS horizontal was simulated by a stiff translational spring in this presimulation stage. In case of volunteer experiments, the muscles are slightly activated to maintain the initial position of the body, while the body settles into the seat. To incorporate this into the model would require neural excitation of muscles utilizing complex feedback mechanisms, which is beyond the scope of this project. This muscular activity was approximated using a translational spring to keep the head horizontal in the presimulation stage. The spring has been released in the final simulations.

Both, PMHS and volunteer simulations are performed with passive muscle behaviour based on the tensile properties of a sternocleidomastoid [5,13,26]. Since the mechanical properties of the PMHS muscles varied significantly over the postmortem period [12] an additional simulation with stiffer passive muscles has been performed representing post-rigor PMHS tensile muscle properties. A parametric study on active muscle behaviour of volunteer simulations will be published elsewhere [13].

Postural variability has been studied for the volunteer experiments. The influence of the initial seating posture as well as of the vertical position of the head restraint is studied. Defining initial positions of the model is done in a similar way as described above. The human model is positioned just above the standard seat in three different positions. One close to the head-restraint (P3), one with a large distance between head and head restraint (P1) and one in between (P2). In all three situations the model was allowed to sink into the seat to find equilibrium. The final initial positions of the model for the PMHS and volunteer simulations are shown in Figure 3 and Figure 4. The horizontal distance between head and head restraint (Δx) and the T1 angle of the different initial postures is presented in Figure 4. For the volunteer simulations, the final seating postures of the model and the experiments were compared to a driver's posture predicted by RAMSIS. Posture P3 showed the best similarity with most experiments and with a posture predicted for this seat using RAMSIS version 3.4.1. RAMSIS predicts a posture of a driver, resulting in arm positioning on the steering wheel.

Also the RAMSIS model showed a larger horizontal head head restraint distance ($\Delta x = 7$ cm), which could be explained by simulating a driver sitting in a more observant posture than a passenger. However, the spinal posture of the RAMSIS model and the MADYMO P3 model were similar.

To study the influence of the vertical position of the head restraint as well as the absence of the head restraint additional simulations with a low head restraint (P3-lhd) as well as without head restraint (P3-nhd) were performed.

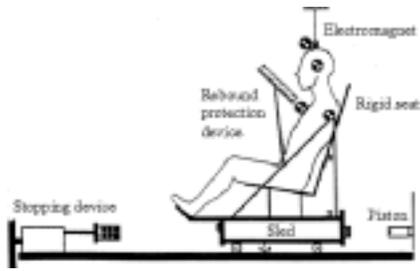
Impact simulations were performed using the sled acceleration (average of all experiments per test series) as input to the models. Furthermore, a vertical acceleration field simulating the gravity is added. In contrast to other studies [7,10,11] the model with detailed neck has been validated as one piece instead of validating the sub model separately.

RESULTS

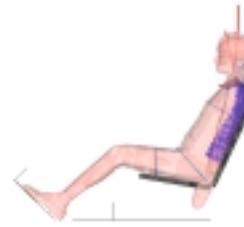
The following terms are introduced helping to describe the results: *good* means within the envelope of the experimental data, *reasonable* means close to this envelope, and 25% deviation allowed, while a *poor* correlation stands for more than 25% deviation from the envelope.

LAB SIMULATION AND PASSIVE MUSCLE PROPERTY VARIATION

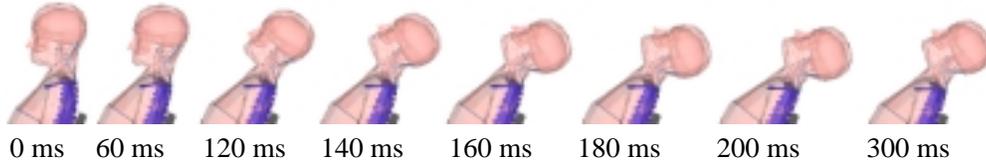
The overall response of the model with stiff passive muscles is shown in Figure 3c. In the initial part of the response, the head only translates. The rebound of the model starts 200 ms after the beginning of the input pulse. This is also observed in the PMHS experiments on films. However the experimental data was only analysed until 200 ms.



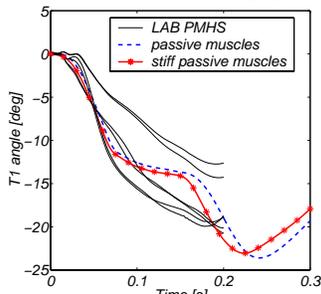
(3a) LAB experiment [10,23]



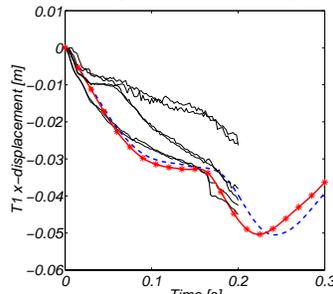
(3b) model



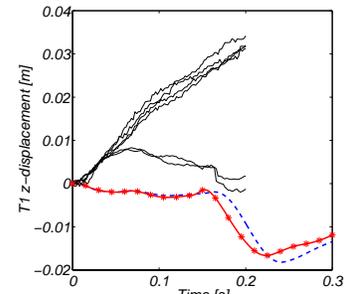
(3c) model response (stiff passive muscles) to 12g rear impact



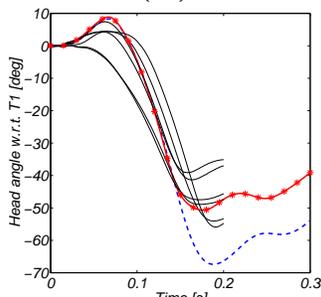
(3d)



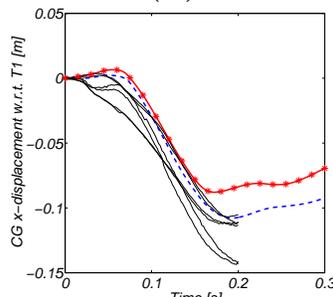
(3e)



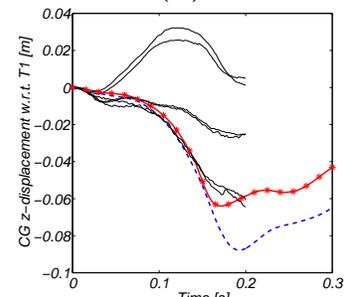
(3f)



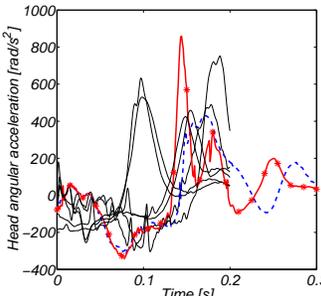
(3g)



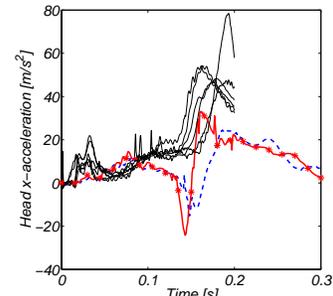
(3h)



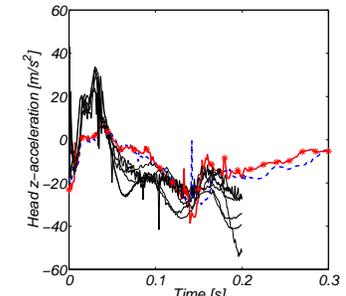
(3i)



(3j)



(3k)



(3l)

Figure 3. Response to 12g rear end impact of the human body model with passive muscles compared to LAB PMHS response. Head and T1 kinematics versus time. (+x is forward, +y is to the left, +z is upward; thus, flexion is positive and extension negative).

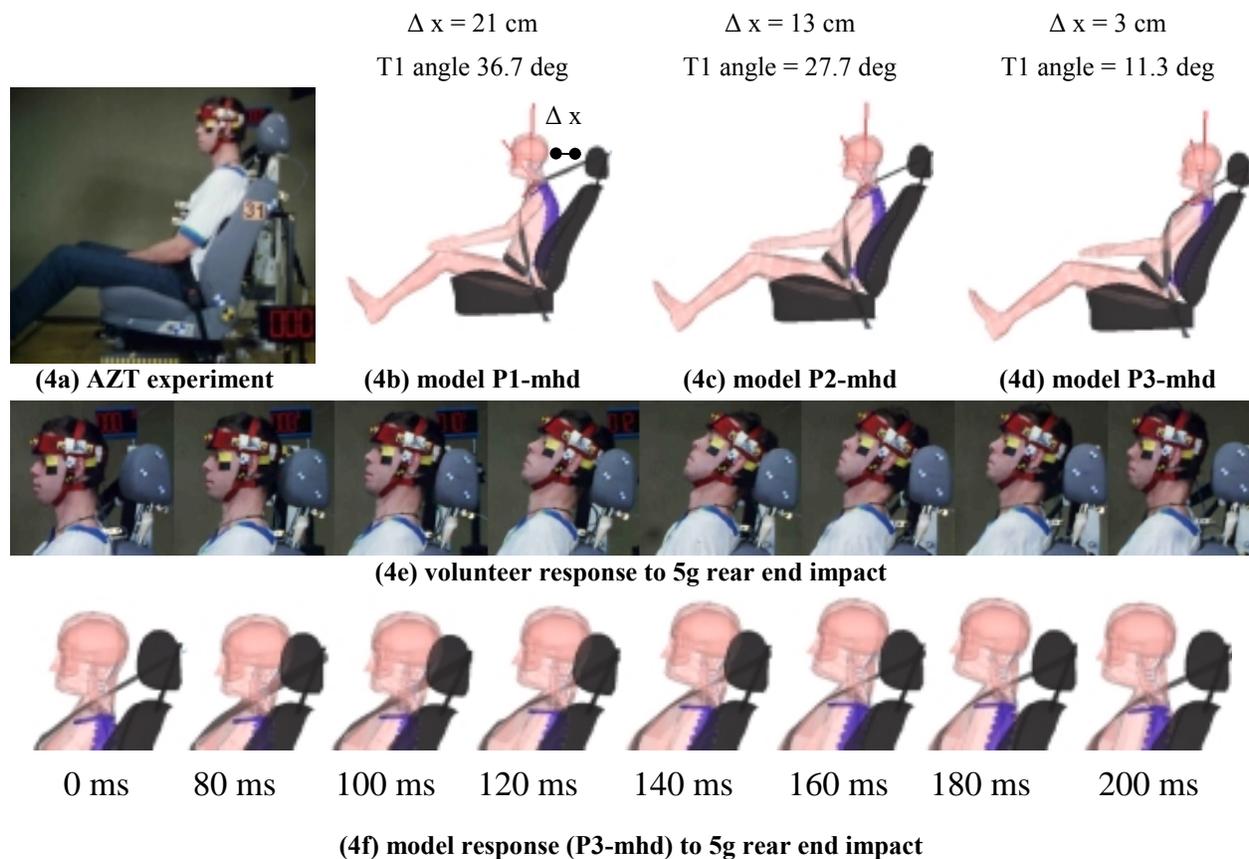


Figure 4. AZT setup.

Head and T1 kinematics of the experiments and the simulation of the PMHS are shown in Figure 3d-3l. For each of the three subjects, results of two experiments are shown. The repeatability of the experiment is apparent from the close match between the two tracings for each subject. A consistent rearward motion of T1 of the PMHS is seen. Note that one subject behaved differently and hardly showed any ramping-up with very little z -displacement for T1. Also the T1 rotation is smaller for this subject compared to the other two. The human body model shows reasonable agreement with the experimental T1 responses in rotation and x -displacement, but the T1 z -displacement remains outside the envelope of the experimental results. The model simulation response shows a sudden increase of T1 rotation and displacement at about 160 ms. This increase occurs earlier for the model with stiff muscles. In general the head response of the model with muscles based on PMHS tissue properties (stiff passive muscles) is more realistic than the response with normal muscles.

The head rotation is shown with respect to T1 (Figure 3g). Again the consistency of the responses of each subject is clearly visible. The head rotation illustrates that the head starts its backward rotation after T1 does, resulting in a small forward rotation of the head relative to T1. Comparison with the model shows that the timing of the head rotation of both models is acceptable. However the maximum head rotation is too large for the model with normal passive muscles, but is within the response envelope for the stiff muscles.

The position of the head CG with respect to T1 versus time is shown in Figure 3h-3i. The PMHS, who showed a rather small ramping up, shows a positive CG z -displacement with respect to T1, while the CG x -displacement is consistent for all the PMHSs. The model with normal passive muscles falls well within the experimental envelopes of the CG x -displacement. The stiff passive muscle model falls within the envelope during the first 160 ms, but finally shows a smaller CG x -displacement. The CG

z -displacements of both models are similar and close to the response of one PMHS until 150 ms. Then, the model with stiff passive muscles almost reached the maximum CG z -displacement while the other model reached a larger maximum at 190 ms.

The z -acceleration (Figure 3l) and angular acceleration (Figure 3j) of the head CG falls within the envelope of the PMHSs for both models. The head x -acceleration (Figure 3k) is close to the PMHS response for the first 100 ms, but later the head x -acceleration of both models shows poor correlation with the PMHS responses.

AZT SIMULATION AND POSTURE VARIATION

The overall response of the volunteers and the model is shown in Figure 4e-f. The model response for three different initial postures is presented in Figure 5. A large influence of posture is seen for all signals. Posture P3 shows the best correlation compared to the volunteer responses. A good correlation of the T1 response is seen (Figure 5a-c). The T1 z -displacement starts a little bit later compared to the volunteers. The head rotation and the CG displacement is shown in Figure 5d-f. The head angle with respect to T1 shows initially flexion as is also seen for the volunteers. However, the flexion is too large and the extension phase starts too late compared to the volunteers, and also the maximum head angle is too small. The T1 motion for P1 and P2 is much larger than for P3, except for the T1 z -displacement, which is smaller. The sudden increase for the T1 angle at about 225 ms (Figure 5a and in Figure 3d), becomes smaller for the more forward placed model (P1).

Studying the head angle with respect to T1 shows head flexion followed by extension (Figure 5d). In case P1 and P2 the T1 angle is much larger than for P3. During the first part of the impact the head translates with respect to the sled with a bit of flexion, together with a large T1 angle for P1 and P2, this results in head flexion with respect to T1. Head contact occurs first for model P3. For model P1 the head never reaches extension.

The head CG displacement shows a wide range of experimental data. The data is not corrected for the initial position, showing the different initial positions of the volunteers. The CG z -displacement (Figure 5f) of model P3 agrees with the volunteer response. The same trend for the CG x -displacement (Figure 5e) is seen for the volunteers and the model, although the initial position with respect to T1 differed.

The acceleration of the head is not presented here. The head x -acceleration shows good agreement compared to the volunteer envelope. The peak shows the moment of head restraint contact, occurring at about 100 ms. The head z -acceleration and the head angular acceleration shows poor correlation with the volunteers [13].

The head neck kinematics for varying head restraint position (mhd= normal, lhd= low, nhd= no head restraint) is presented in Figure 6. A large difference is seen for the cases with head restraint (P3-mhd and P3-lhd) and the one without head restraint (P3-nhd). Case P3-lhd shows the best correlation compared to the volunteer responses. Although the T1 kinematics is hardly influenced by varying the head restraint height, when simulating an impact without head restraint the T1 x -displacement and T1 angle show larger motions. The head motion is more limited for higher position of head restraint.

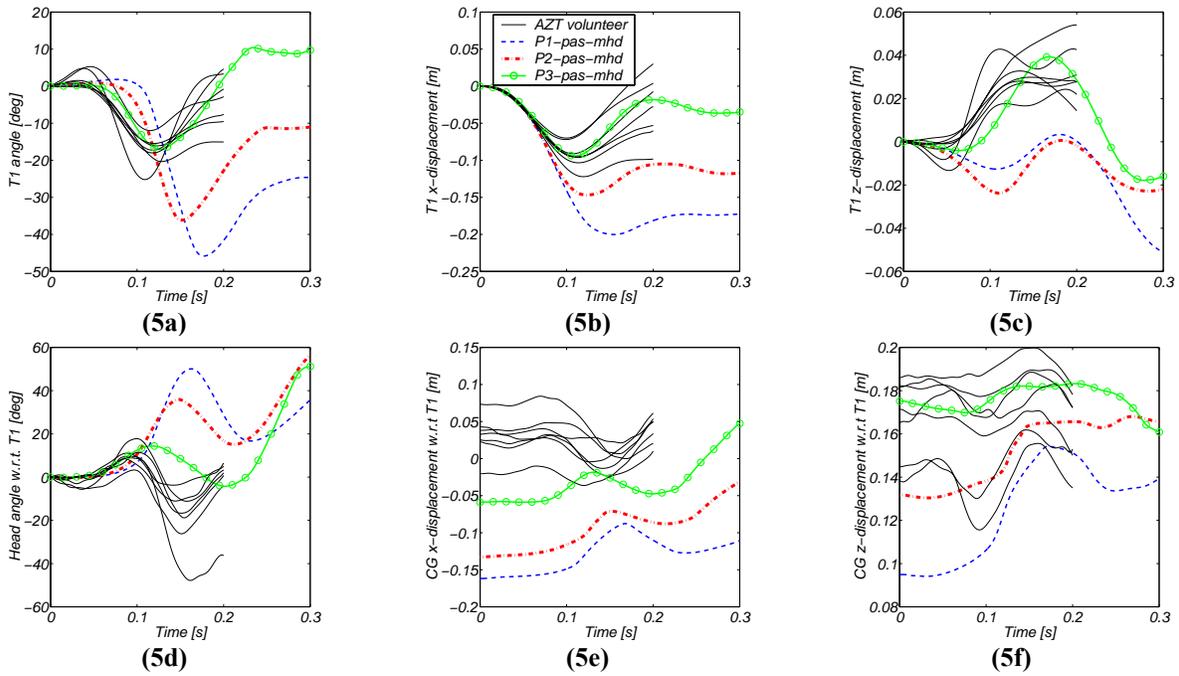


Figure 5. Variation in initial seating posture. Head and T1 kinematics versus time. Response to 5g rear end impact of the human body model with initial seating posture (P1, P2, P3), passive muscles (pas) and head restraint position normal (mhd) compared to AZT volunteer response. (+x is forward, +y is to the left, +z is upward; thus, flexion is positive and extension negative).

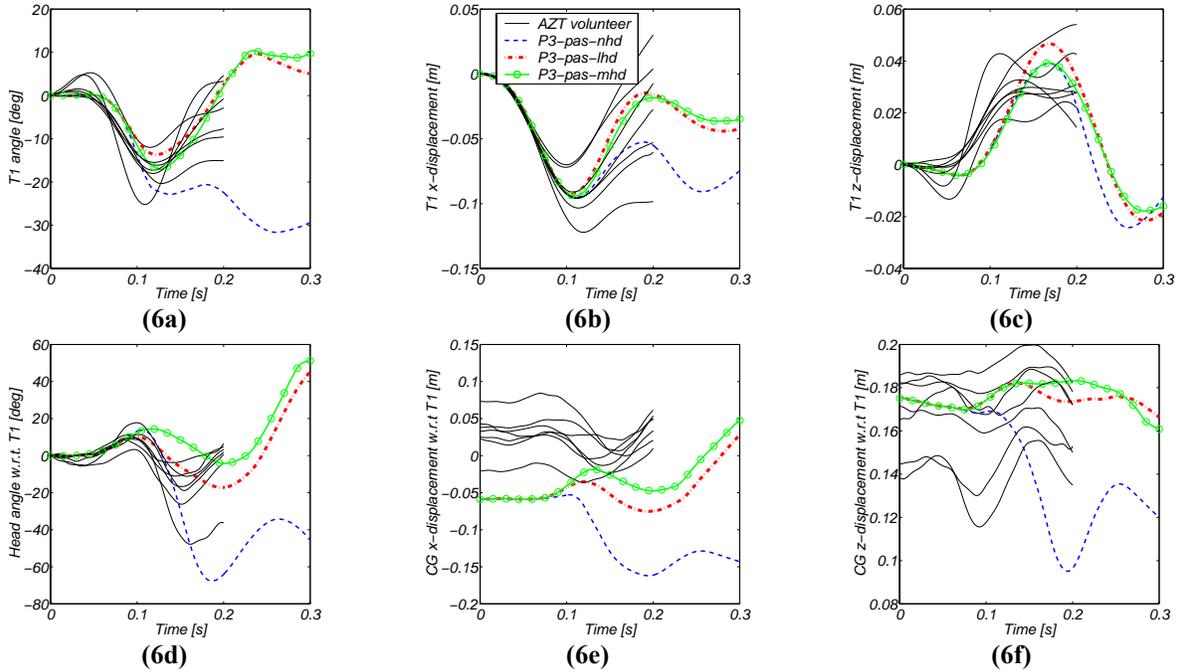


Figure 6. Variation in vertical head restraint position. Head and T1 kinematics versus time. Response to 5g rear end impact of the human body model with vertical head restraint position (mhd=normal, lhd=low, nhd=no head restraint), initial seating posture P3, and passive muscles (pas) compared to AZT volunteer response. (+x is forward, +y is to the left, +z is upward; thus, flexion is positive and extension negative).

DISCUSSION

The model presented is intended to predict local loads in the various structures in the neck. As a first indication of the quality of the model in this study global kinematics and accelerations of the head neck system of experiments and model are compared.

Simulation of PMHS experiments are needed for model validation for high severity impacts. For the PMHSs in the LAB tests, a satisfactory model response was obtained after adapting the muscle tensile properties towards published PMHS tissue properties. The assumption that the muscles of the PMHS were stiffer than for volunteers can be justified when the PMHSs of the LAB experiments were not preconditioned before testing, resulting in unrepeatable but stiffer response than live passive muscle response [12]. Although the PMHSs showed repeatable responses, nothing has been reported about preconditioning. The mechanical properties of the muscle varied significantly over the postmortem period [12]. Therefore, the preparation of the PMHS used in experiments should be documented, making it possible to adjust the model muscle tensile stiffness towards the condition of the PMHS.

For a further test of the model, experimental data on local mechanics are needed, e.g. vertebral motion during impact, extracted from X-ray images [27,28]. In addition substructure testing can provide additional confidence for instance testing without muscles to avoid the effect of tissue property changes.

The T1 motion of LAB and AZT P3 simulations showed acceptable agreement compared to the experiments, except for the ramping up of the LAB PMHS test and the sudden increase of the T1 angle (Figure 3d, Figure 5a, Figure 6a). Although the sudden increase is not seen in the experimental data used, it can be seen in other experimental data with volunteers [28] and cadavers [27], but no attention has been paid to this phenomenon. The parametric study of initial posture showed that the sudden increase in the T1 angle becomes smaller for the more forward placed model (P1), indicating that this phenomenon is influenced by initial position. Since the model shows small T1 z-displacement for the LAB PMHS tests, additional simulations were performed in which the following parameters were varied, friction of the seat back, belting of the PMHS, and gravity. However the head angle was hardly influenced while only neglecting the gravity showed acceptable increase of the T1 z-displacement. In Figure 5c it is seen that varying initial posture

influences the ramping up as well. Therefore it is assumed that the variance in initial position between PMHSs and between model and PMHS causes the differences in ramping up.

RAMSIS predicts one initial seating posture, which corresponds with posture P3 of the model, however in reality the seating postures of drivers and their passengers show a large variability [15]. The results from the posture variance simulations support the statement made in literature [14,15,16] that posture variation has major effects on T1 and head response. The model response showed larger variability due to posture variation than the experimental response, however the range of postures of the simulations was much larger compared to the differences in initial seating posture of the volunteers. The increasing displacement and rotation of T1 from P1 through P2 to P3 is caused by the fact that the model is sitting more bent forward, having more space between the back and the seat, resulting in larger displacements. The larger ramping up for P3 can be explained by the interaction of the back and the seat back occurring earlier, resulting in upward movement of the body. Head contact occurs first for model P3, being closest to the head restraint at the start. This results in the smallest forward head displacement for P3 in the first phase. Then the head rotates backwards, resulting in backward translating with respect to T1. Since this rotation starts late for model P1 the head never reaches extension, while the model P3 starts earlier with rotation, but the rotation is limited by the head restraint, ending in head flexion.

The effect of head restraint height is relatively small. However, removing the head restraint shows much larger head and T1 motions compared to the response with head restraint. Lowering the head restraint showed better correspondence for the head rotation, but still the backward rotation is too slow. The difference in head kinematics at varying head restraint position is caused by the variation of contact point resulting from the variance in vertical height. The lower the contact point with respect to the centre of gravity, the larger the head extension will be. It has to be remarked that in the test with head restraint, head translations and rotations are much smaller than in the test series without head restraint, this trend was also seen for the simulations. An additional study is recommended in which the force and line of action for the contact between head and head restraint will be evaluated, in order to reach better correlation for the model head kinematics when contacting a head restraint.

In accordance with other studies (see review by Szabo [29]) it is shown by mathematical modelling that seating in an upright position together with an head restraint adjusted in line with the top of the head reduces the head motion compared to a more forward seating position and a low head restraint.

Not only seating posture and head restraint position influence the head neck response. Also the anthropometric variability of the human will have effect on the response. Studying the influence of anthropometric variability requires scaling of the model [30]. For true model validation exact information on seating posture, position of seat and head restraint and the anthropometric data of the subjects should be known. Based on that information a decision should be made if one to one evaluation (one model with one experiment) is needed, or that the model can be compared to an envelope of experiments.

As described earlier, in reality the muscles in neck and spine are slightly activated to maintain the initial posture. Although not shown here, the model has also the potential to simulate muscle activation [13]. Muscles are also being implemented in the rest of the spine allowing simulation of postural and reflex induced muscular activation.

An important benefit of the model is the extended validation and the possibility to simulate different initial positions. Due to the integration of the detailed neck model, in principle the model can be used for studying injury mechanisms because deformation and loads of the individual soft tissues can be assessed.

CONCLUSIONS

- ◆ A full body human model with detailed neck has been presented. The model was used to simulate a PMHS study at 12g rear end impact on a rigid seat without head restraint and a volunteer study at 5g rear end impact on a standard seat with head restraint. Validation is provided for kinematics and accelerations of head and neck.
- ◆ The model shows good agreement with the PMHS responses when muscle tensile stiffness is increased towards published PMHS tissue properties.
- ◆ Initial seating posture strongly influences the model response. More leaning forward results in larger T1 and head motions.
- ◆ The effect of head restraint height is relatively small compared to the influence of posture variability (horizontal head restraint variance). A correct vertical position of the head restraint (top of head in one line with top of head restraint) can reduce head extension angle.

ACKNOWLEDGEMENTS

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