A global and a detailed mathematical model for head-neck dynamics

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A Global and a Detailed Mathematical Model for Head-Neck Dynamics

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ABSTRACT

Two mathematical head-neck models have been developed using MADYMO: a global model and a detailed one. The global model comprises rigid head and vertebrae connected through nonlinear viscoelastic intervertebral joints representing the lumped behaviour of disc, ligaments, facet joints and muscles. The model response to frontal impacts agreed reasonably with volunteer responses. The detailed model comprises rigid head and vertebrae connected through linear viscoelastic discs, nonlinear viscoelastic ligaments, frictionless facet joints and contractile muscles. The model response to lateral impacts agreed excellently with volunteer responses, whereas the response to frontal impacts showed that the model was too flexible. The global model is especially suited for use in complex simulations as occupant behaviour in car crashes, whereas the detailed model is particularly suited for neck injury assessment.

Introduction

The human neck is vulnerable. In car accidents, inertia forces of the head can load or deform the tissues of the neck beyond tolerable limits, resulting in injury. Neck injuries appear to occur frequently in automotive accidents [1–4]. Most neck injuries are minor injuries (whiplash), but they may lead to long-lasting and irritating complaints such as a painful stiff neck, headache, cognitive function loss and numbness of the upper limbs [5]. Severe neck injuries are often disabling or fatal. Knowledge of the (injury) mechanisms through which loads cause injuries to the neck is incomplete, especially for the minor injuries, for which usually no clearly identifiable damage in the neck can be found. Furthermore, reliable neck injury criteria hardly exist to date. Experimental and numerical studies may aid to both qualify and quantify the various injury mechanisms and the corresponding injury criteria.

The objective of our study is to develop a detailed three-dimensional mathematical model describing the dynamic behaviour of the human head and neck in automotive accidents without head contact. With a sufficiently detailed model, the loads applied to the head and neck can be converted into loads and deformations of the tissues within the neck, which can be compared with failure limits of the tissues to check whether injury took place. The strategy is to proceed from a relatively simple model (*the global model*) for gaining insight into head-neck dynamics towards a complex model (*the detailed model*) providing the loads and deformations of the tissues of the neck. This paper deals with both models, which are briefly outlined by summarizing their main characteristics. Focus is on the response of both models in comparison with human volunteer responses to frontal and lateral sled acceleration impacts. Full details on the models may be found in Ref. [6]. The models have been implemented in the integrated multibody/finite-element package MADYMO, version 5.1.1, of the TNO Crash Safety Research Centre [7].

The Global Model

The global model comprises rigid bodies for the head (C0), the cervical vertebrae (C1-C7) and the first thoracic vertebra (T1), to which the inertia characteristics of head and neck are lumped, Fig. 1. The initial configuration includes the cervical lordosis reconstructed from lateral x-rays of male volunteers [8]. The rigid bodies are connected through three-dimensional nonlinear viscoelastic intervertebral joints representing the lumped mechanical behaviour of the intervertebral disc, ligaments, facet joints and muscles. Joint characteristics were derived from the experimental data on the mechanics of lower cervical motion segments and upper cervical spine specimens [9–16], Fig. 2. The lower cervical joints had initially the same (*in vitro*) characteristics, but their (excessive)

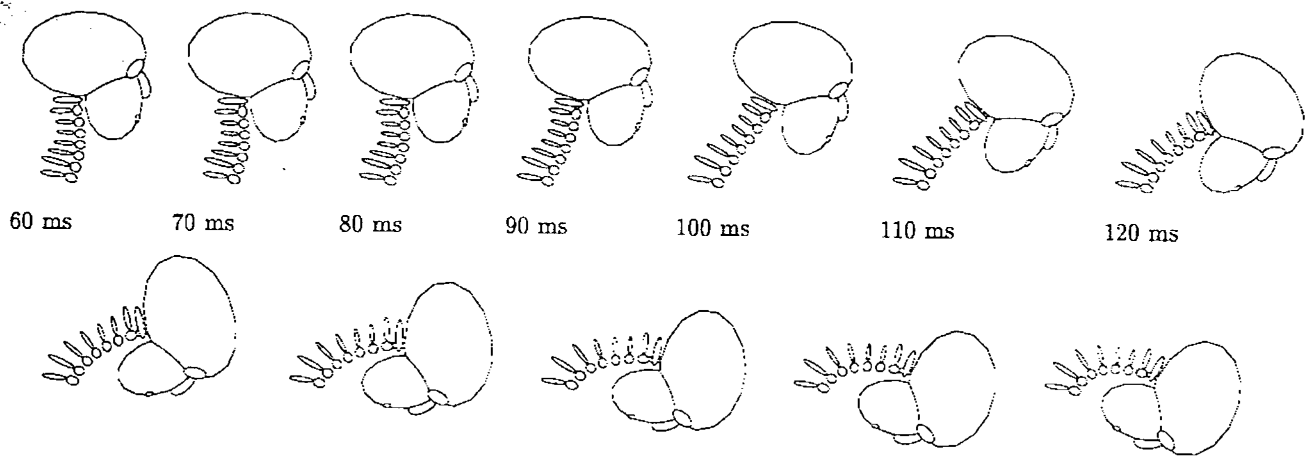


Figure 3: Motion of the global model for the frontal impact at 10 ms intervals.

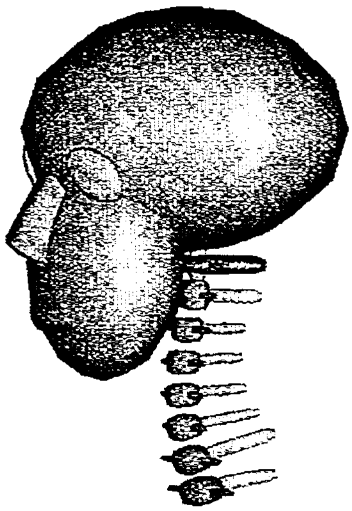


Figure 1: Frontal oblique view of the global head-neck model. Shown are the ellipsoids representing the vertebrae C1–C7, T1 (vertebral body with arch and spinous process) and the skull (with nose, mouth and eyes for visual purposes only). The intervertebral joints are not shown.

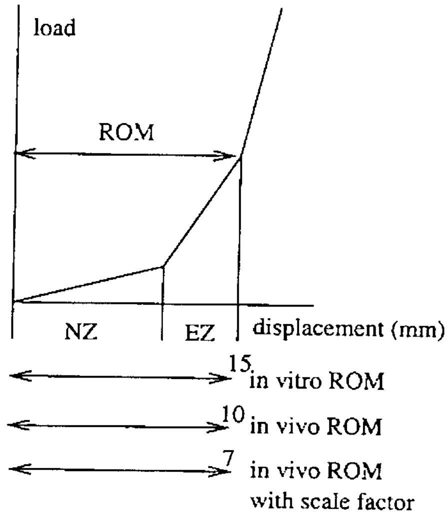


Figure 2: Outline of the nonlinear load-displacement curves for the intervertebral joints of the global model. The neutral zone (NZ) and elastic zone (EZ), together forming the range of motion (ROM) of the joint, were derived from the *in vitro* studies of Panjabi and co-workers [10-15]. The part of the curve beyond the ROM were derived from Shea *et al.* [16] and Chang *et al.* [9]. The displacement axis was adjusted to obtain the *in vivo* ROMs in rotation reported by White and Panjabi [17]. A scale factor was introduced to further adjust the curves to represent increased joint stiffness due to muscle tensioning as found by Wilke *et al.* [18].

ranges of motion in rotation were modified to agree with the *in vivo* ranges of rotation reported by White and Panjabi [17]. Both upper cervical joints (C0-C1 and C1-C2) have different characteristics reflecting the unique mechanical behaviour of these joints. Because the joint characteristics resulted in a too flexible model, a scaling factor was introduced for modifying the joint stiffnesses in order to account for the increase in joint stiffness due to muscle tensioning [18]. This scaling factor was used to calibrate the model response to frontal impacts.

The model response to frontal impacts was verified using the head-neck responses of human volunteers subjected to sled acceleration impacts performed at the Naval Biodynamics Laboratory (NBDL) [19]. The most severe frontal impacts, which were recently re-analyzed by Thunnissen *et al.* [20], are used here. The average horizontal forward acceleration of vertebrae T1 was used as input to the model to simulate the impact. It was not

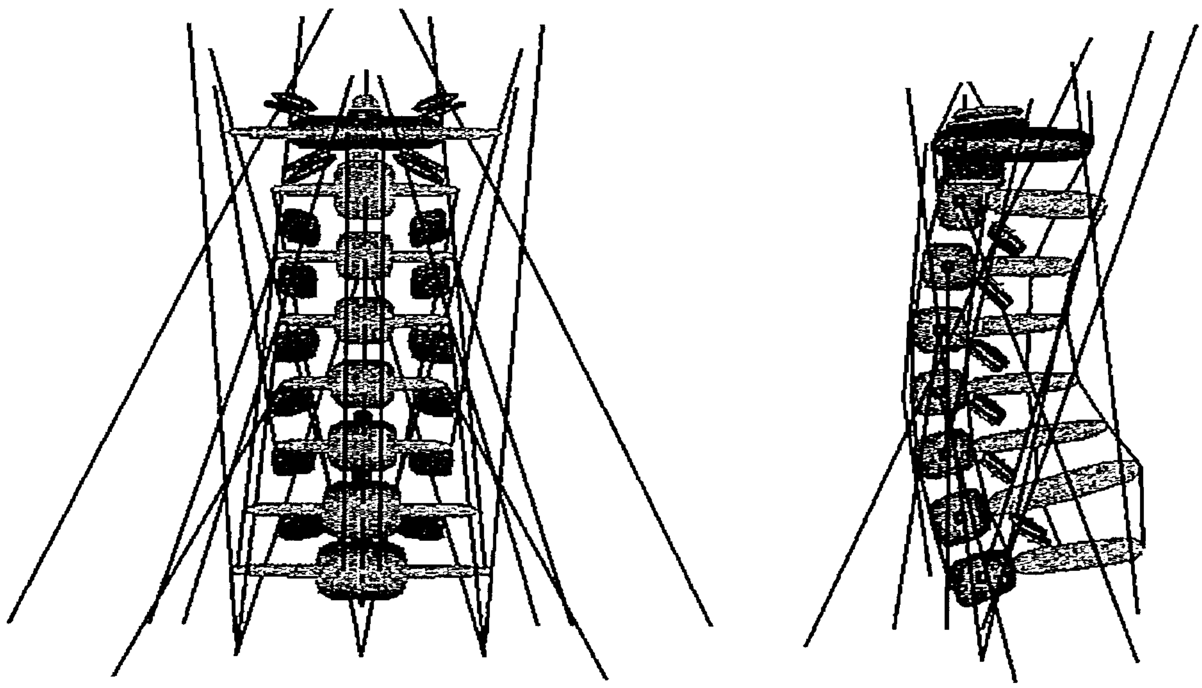


Figure 4: Frontal and lateral view of the detailed model. Ellipsoids represent the vertebrae and the articular facets, while line elements depict the ligaments and muscles. The skull and the intervertebral discs are not shown.

necessary to prescribe the forward rotation of T1 as it was accounted for in the response corridors used to compare the model responses with [20]. These corridors were defined at the average volunteer response plus or minus one standard deviation. Used corridors are the linear and angular acceleration of the head centre of gravity relative to the laboratory coordinate system; the trajectory of the occipital condyles and the centre of gravity of the head relative to the T1-vertebral body; the rotation of head and neck versus time; the neck rotation versus head rotation; and the neck length. Neck rotation was defined as the rotation of the straight line connecting T1 to the occipital condyles; neck length as the length of this line.

Fig. 3 illustrates the motion of the model and Fig. 7 shows the model responses and volunteer corridors. Overall, the model response is satisfactory, except for head rotation. The trajectories of the occipital condyles and the centre of gravity of the head accurately follow the corridors, but the downward displacements become too large for the centre of gravity. The neck length response is fairly accurate and shows that the head oscillates slightly too much in axial direction. The linear accelerations of the head centre of gravity agree qualitatively with the corridors, although the peaks outweigh the corridors. The linear accelerations of the model lag slightly behind the corridors. The head angular acceleration corresponds reasonably with the corridors and the maximum is well reflected. The angular acceleration increases somewhat too early as does the head rotation. Neck rotation falls well inside the corridors, while head rotation agrees poorly with the volunteer responses. The head lag response reflects that head rotation lags well behind neck rotation, although it deviates a little due to the early rise of head rotation. Eventually, head rotation exceeds the neck rotation for the model (overtipping) in contrast to the volunteers, of whom head and neck rotation increase uniformly (locking).

Clearly, head rotation is unrealistically large. If chin-torso contact had been incorporated into the model, head rotation might have been limited, but would still be too large as chin-torso contact did not occur for the volunteers. The model responses can also be compared with the responses of human cadavers subjected to similar sled tests as presented by Wismans *et al.* [21]. The neck muscles of the cadavers were artificially stiffened to represent muscular tension needed to keep head and neck upright. Main differences in response were that the cadavers had a smaller head lag, had a maximum head rotation that was about 20 deg larger, and showed overtipping of head rotation. Thus, the cadavers show a similar difference in response with the volunteers as the model, indicating that muscle tensioning limits head rotation and prevents overtipping for the volunteers.

The Detailed Model

After the global model, an anatomically more detailed model was created. Initially, detailed models of the lower cervical motion segments and the upper cervical spine were developed and (satisfactorily) validated against ex-

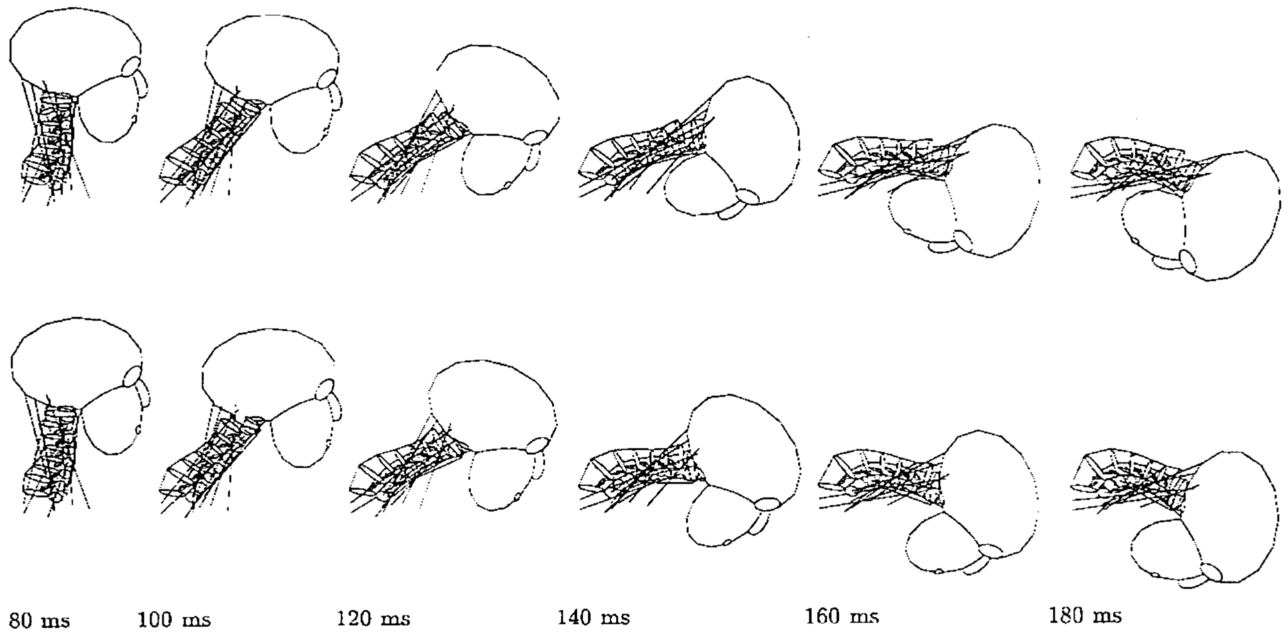


Figure 5: Motion of the detailed model with passive (top) and active (bottom) muscle behaviour between 80 and 180 ms for the frontal impact.

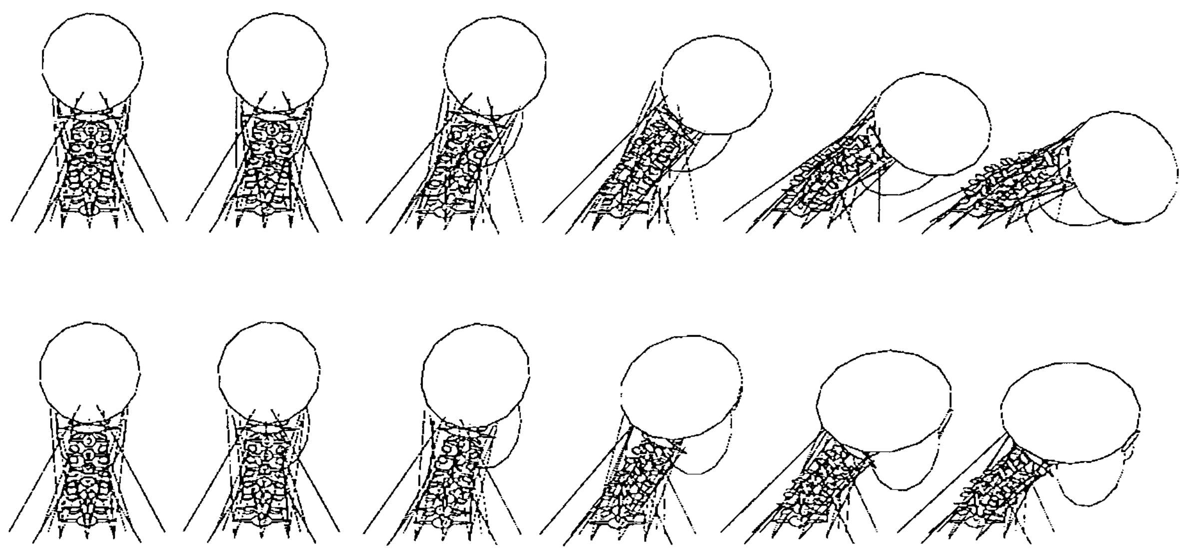
perimental data [6]. These segment models were then joined to form the detailed head-neck model. This detailed model comprises rigid head and vertebrae (C1-C7, T1) connected through linear viscoelastic intervertebral discs, nonlinear viscoelastic ligaments, frictionless facet joints and contractile muscles, Fig. 4.

The intervertebral discs are modelled as three-dimensional linear viscoelastic elements using the stiffness of disc segments reported by Moroney *et al.* [22]; in tension, the data of Pintar *et al.* [23] were used. Nonlinear viscoelastic line elements were used for the cervical ligaments, resisting load in tension only. The load-displacement characteristics were based on Refs. [24, 25]. Frictionless contact interactions between almost rigid bodies were used for the facet joints, the geometric characteristics of which were derived from Panjabi *et al.* [26].

Fourteen mid-sagittal symmetrical pairs of MADYMO's contractile Hill-type elements were used to represent the stronger and more superficially located neck muscles. A simplified geometric representation was chosen in which each muscle force is directed along the straight line connecting origin and insertion (line of action). This representation seemed justifiable to gain a first impression of the relevance of muscles in impacts. An important limitation is that the muscles cannot curve around the vertebrae, which may lead to inaccurate lines of action for large intervertebral rotations, but this could not be prevented with the muscle elements of MADYMO, version 5.1.1. The chosen insertions of the muscle elements represent an average position, since most cervical muscles insert on several vertebrae. These positions were based on Refs. [27–32].

The detailed model was validated for frontal and lateral impacts using human volunteer responses to sled acceleration impacts, for which the most severe frontal [20] and lateral impacts [33, 34] were used. Both active and passive muscle behaviour were simulated to study the effect of muscle contraction on the head-neck response. To have a maximal effect on the head-neck response, all muscles start to contract at 50 ms: the time at which the sled begins to accelerate after the volunteer initiated the impact (at 0 ms). An earlier contraction time would lead to an unrealistic large extension of the head relative to the neck, since muscle forces were not balanced to keep head and neck upright (as a volunteer would do) when the sled is still at rest. The active and passive response are the response of the model with and without muscle activation, respectively.

Response to Frontal Impact The detailed model was subjected to the same frontal impact as the global model and the same response corridors are used for validation. Fig. 5 depicts the model kinematics and shows that the muscle lines of action appear to become unrealistic after about 140 ms for the active and passive response. Fig. 8 compares the active model response with the response corridors. The linear accelerations of the head centre of gravity compare well with the corridors, although the minimum about 110 ms for the resultant acceleration is too strongly present in the model. Initially, the angular acceleration of the head agrees favourably, but, after 120 ms, it is minimal for the model, while it is still sig-



80 ms 100 ms 120 ms 140 ms 160 ms 180 ms

Figure 6: Motion of the detailed model with passive (top) and active (bottom) muscle behaviour between 80 and 180 ms for the lateral impact.

nificant for the volunteers. The neck length response is accurate up to 170 ms. The trajectories of the occipital condyles and centre of gravity of the head nicely follow the response corridors but, eventually, the downward displacements (z) become too large, showing that the model is more flexible than the human volunteers. This is also reflected by the head and neck rotation, which fall within the corridors up to 150 ms and 100 ms, respectively, before they become much too large. The maximum head rotation would have been limited if chin-torso contact had been included in the model (Fig. 5); chin-torso contact did not occur for the volunteers. The head lag is too strongly present in the model, because the head flexion starts somewhat late while neck rotation begins early in comparison with the volunteer responses. The head lag response (after the head lag and up to 80 deg head rotation) also shows that head and neck move more or less as one unit; that is, rotation of the head relative to the neck is almost absent. This locking phenomenon was also seen for the volunteers and causes the neck rotation to be larger than the head rotation during the entire impact. Locking is, thus, adequately reflected by the model.

Fig. 9 compares the active response with the passive response. Muscle contraction, clearly, influences the response. Head angular acceleration is strongly affected: the acceleration is minimal after 120 ms for the active response, while it strongly oscillates for the passive response. The linear accelerations as well as the trajectories of the occipital condyles and centre of gravity of the head change little due to muscle contraction; this is also reflected by the model configurations (Fig. 5). The head rotation and, consequently, the head lag show a stronger change in response. For the passive response: the head lag decreases somewhat, the head rotation increases, the rotation of head and neck are no longer locked, and the head rotation eventually exceeds the neck rotation. This indicates that the muscles are responsible for both the head lag and locking of head and neck rotation, which confirms the observations made by Wismans *et al.* [21], who found similar differences between the experimental head-neck responses of human volunteers and human cadavers subjected to sled acceleration impacts. The neck length response shows that the muscles compress the neck to some extent. Neck rotation is larger for the active model because the lines of action of the muscles become unrealistic for large rotations such that the head is pulled towards the torso rather than pulled back and upwards.

Response to Lateral Impacts The lateral impact was simulated with the model using active and passive muscle behaviour. For the model with muscle contraction, only the muscles on the left side were activated to oppose the motion of the head and neck to the right. The model kinematics are depicted in Fig. 6, showing that the lines of action of the muscle appear realistic for the active model, due to the moderate lateral bending of the neck, while they tend to become unrealistic for the passive model.

Fig. 10 shows the active response and the response corridors. All responses compare well with the corridors. The trajectories of the occipital condyles and centre of gravity follow the corridors adequately and exceed the corridors only slightly. The neck length, which differs somewhat initially, increases slightly for the model during the impact, while it decreases for the volunteers. The linear acceleration of the head centre of gravity agree well with the corridors, although the x- and y-acceleration increase a little too early, and the z-acceleration is smaller compared with the volunteers. The angular accelerations compare favourably with the volunteer responses, but, after 130 ms, the angular acceleration of the model hardly changes in contrast to the accelerations of the volunteers. The z-angular acceleration increases too soon, which is also reflected by the axial rotation of the head, causing it to lie outside the corridor. Even though the trends are similar, axial rotation becomes too large for the model. The lateral head rotation closely resembles the corridor. Both upper cervical joints bend slightly to the left (less than 5 deg) compensating, in part, the large rotations of the lower joints to the right, which causes the cg-trajectory to fit the corridors more closely than the oc-trajectory does.

Fig. 11 compares the active and passive responses. Muscle contraction affects the head-neck motion strongly, which is also reflected by Fig. 6, indicating that muscles are strong enough to significantly alter the head-neck response to impacts. Especially, the trajectories and the lateral head rotation are strongly improved due to muscle tensioning, but axial rotation becomes too large. Comparing Figs. 10 and 11, it follows that muscle tensioning causes most model responses to change sooner than the volunteer responses. Also, it appears that the model would have followed most corridors even better if the muscles would have been tensed less than maximally or would have started to contract after 50 ms, noting that most passive responses appear to lie at the other side of the corridors in comparison with the active responses. If the upper torso had been included, the head would have contacted the right shoulder for the passive response limiting the lateral head rotation. Muscle tensioning leads to smaller lateral rotations for all joints, but the extension of the upper cervical joints were found to increase.

Discussion

Two mathematical head-neck models have been developed using MADYMO: a global and a detailed one. The global model is a relatively simple model with few anatomical details. It comprises a rigid head and rigid vertebrae, connected through nonlinear viscoelastic intervertebral joints describing the mechanical behaviour of the intervertebral disc, ligaments and facet joints. Joint characteristics were derived from the behaviour of motion segments of the lower and upper cervical spine. Because these *in vitro* characteristics result in a too flexible model, the characteristics can be scaled to incorporate the stiffening effect of muscle tensioning on the neck, and to allow for calibration of the model response to impacts. The model was calibrated to match the response of human volunteers to frontal impacts and a reasonable agreement could be obtained. The linear and angular accelerations of the head and the neck rotation agreed satisfactorily, but head rotation was too large.

Since the oc-trajectory and, consequently, the neck rotation lie well within the corridors, it appears that the *overall* neck response is described accurately. It was found that the maximum rotations of the upper four joints appeared accurate compared with the ranges of motion. The maximum rotations of the lowest four joints, in contrast, were much larger than the ranges of motion, indicating that the neck is curved too strongly, and that the large head rotation is primarily caused by too much rotation of the lower joints. Another cause is the absence of active muscle behaviour needed to prevent overtipping of the head relative to the neck. Thus, the *segmental* neck response needs to be improved, which may be done by modifying the stiffnesses of the lowest joints. However, *in vitro* studies did not find significant variations in intervertebral joint stiffness with vertebral level. Possibly, muscles can more effectively stiffen the lower joint compared with the upper joints, but experimental evidence is lacking.

The global model has several advantages making it particularly useful in car safety design studies. First, it is a numerically efficient model: a simulation of an impact lasting 200 ms typically takes about 40 CPU-seconds on an SG Indigo 2 R4400 workstation using MADYMO with a fifth order Runge-Kutta-Merson method with variable time step. Second, the model can easily be modified to represent a taller/shorter or a stronger/weaker neck, since few geometric and mechanical parameters characterize the model. Third, a reasonably accurate model response was found for frontal impacts, while the model response to other impacts may readily be calibrated with experimental responses by adjusting the scale factors. Thus, the model might be useful in a model of a car occupant to study the effect of head rests, airbags and the car interior on the occupant's head-neck response in simulated crashes. Further model validation, however, is needed for contact situations as well as other impact directions and severities.

The detailed head-neck model comprises rigid head and vertebrae, linear viscoelastic discs, frictionless facet joints, nonlinear viscoelastic ligaments and contractile muscles. Human volunteer responses were used to validate the model. In the lateral impact, the model agreed excellently with the volunteers for the linear and angular accelerations of the head, the trajectories of the occipital condyles and centre of gravity of the head, and the lateral head rotation. Only the axial head rotation was too large. In the frontal impact, the linear and angular head accelerations agreed reasonably with the volunteer responses, but head and neck rotation were too large. Rotation of the head relative to the neck, however, was accurately predicted due to muscle tensioning. The trajectories also reflected that the model was too flexible. This was mainly attributed to the incapability of the muscle elements to curve around the vertebrae: their straight lines of action became unrealistic for large neck rotations, such that the muscles failed to effectively constrain the head-neck motion and stiffen the joints.

Thus, the muscle representation seems to be the major limitation of the model, especially when joint rotations become large. The lateral impact illustrated that the muscles do influence (and improve) the model response significantly and that the muscle elements function adequately for moderate joint rotations. As the lateral response is good, it is expected that the response to frontal impacts can be improved with better geometric modelling of the muscles. It is also expected that the present model is suitable for frontal impacts in which the neck rotation is limited, because a less severe impact is simulated, or because the head contacts an airbag or a head rest.

The detailed model may also be used in car safety studies, although a simulation takes about ten times the CPU-time needed for the global model. Thus, the global model is preferable for these applications, unless more detailed information about the head-neck response is needed.

The detailed model is, in principle, useful for studying injury mechanisms because deformations and loads of the individual soft tissues can be assessed. The calculated loads and deformations can be used to formulate failure tolerances by reproducing experiments done with either motion segments or complete cervical spines and correlating the experimentally obtained injuries with the predicted tissue loads and deformations. Once appropriate tolerance levels for the tissue are known, injury mechanisms can be incorporated in the model such that the model will, eventually, be able to predict the probability of the occurrence of injuries in reconstructed accidents.

Summary and Conclusions

- A global and a detailed head-neck model have been developed. In the global model, rigid head and vertebrae are connected by intervertebral joints only. In the detailed model, head and vertebrae are connected by intervertebral discs, ligaments, facet joints and muscles.
- The global model was validated for frontal impacts and a reasonably accurate response was found.
- The global model is a computationally efficient model, which can easily be modified and calibrated, making it especially suited for use in complex simulations as occupant behaviour in car crashes
- The detailed model was validated for frontal and lateral impacts. The lateral response agrees well with volunteer responses. The model is too flexible in frontal impacts.
- Active muscle behaviour appears essential to accurately describe the human head-neck response to impacts.
- The detailed model is particularly suited to study neck injury mechanisms and criteria, because it reveals the loads and deformations of the individual soft tissues.

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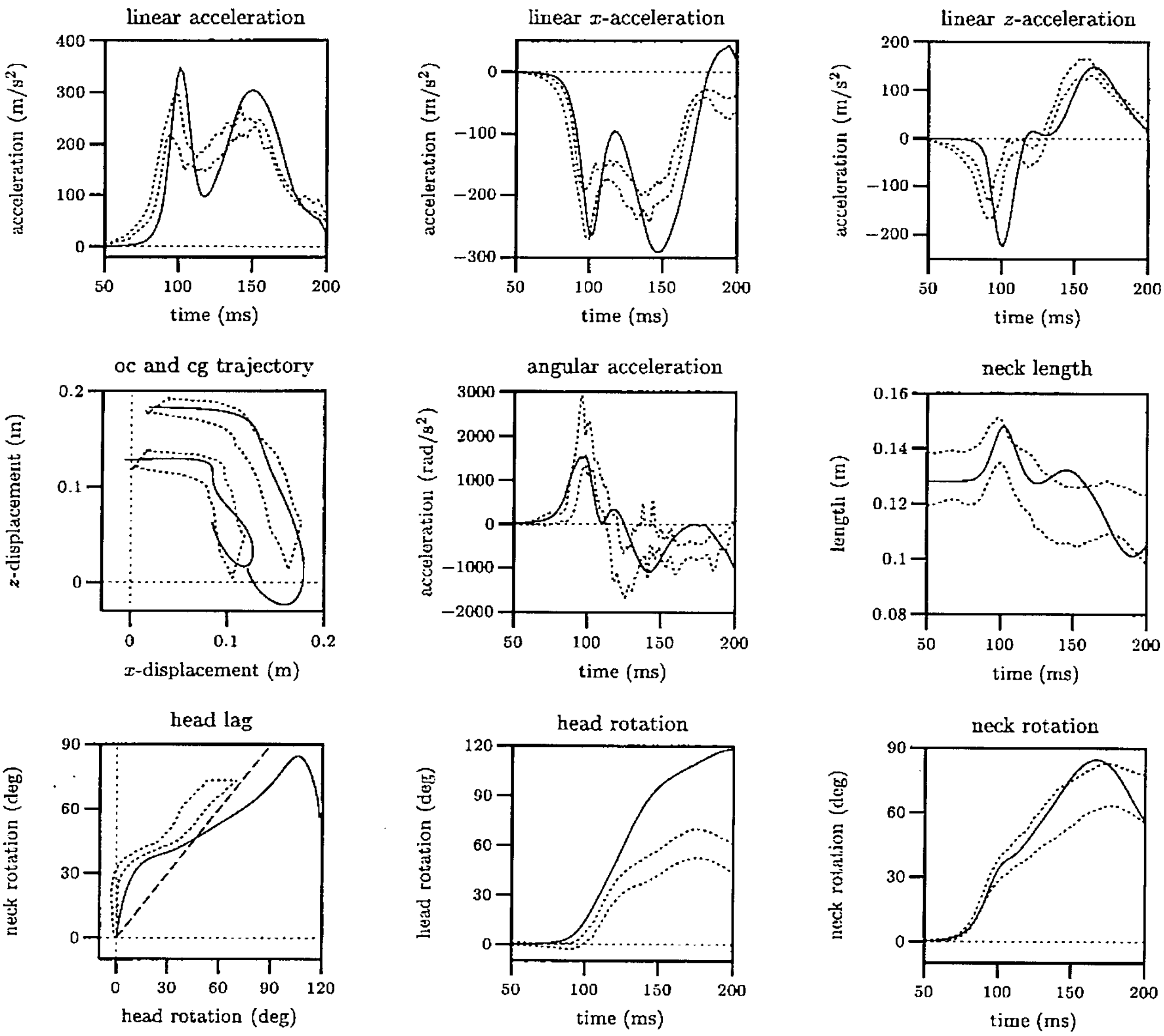

Figure 7: Response of the global model (solid line) in comparison with the human volunteer response corridors (dotted lines) for the frontal impact. $+x$ is forwards, $+z$ is upwards.

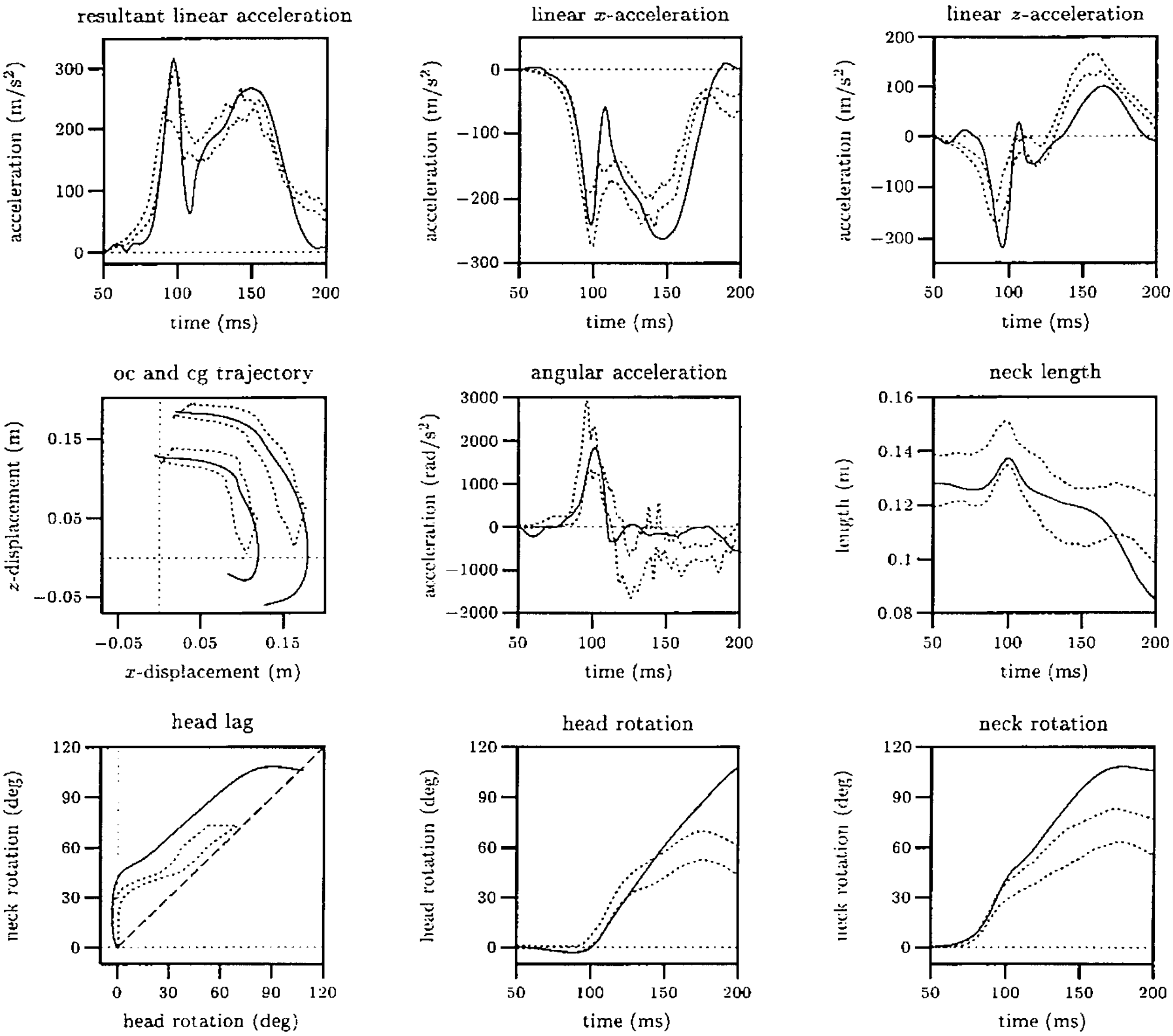

Figure 8: Response of the detailed model with active muscle behaviour (solid line) compared with human volunteer response corridors (dotted lines) for the frontal impact. $+x$ is forwards. $+z$ is upwards.

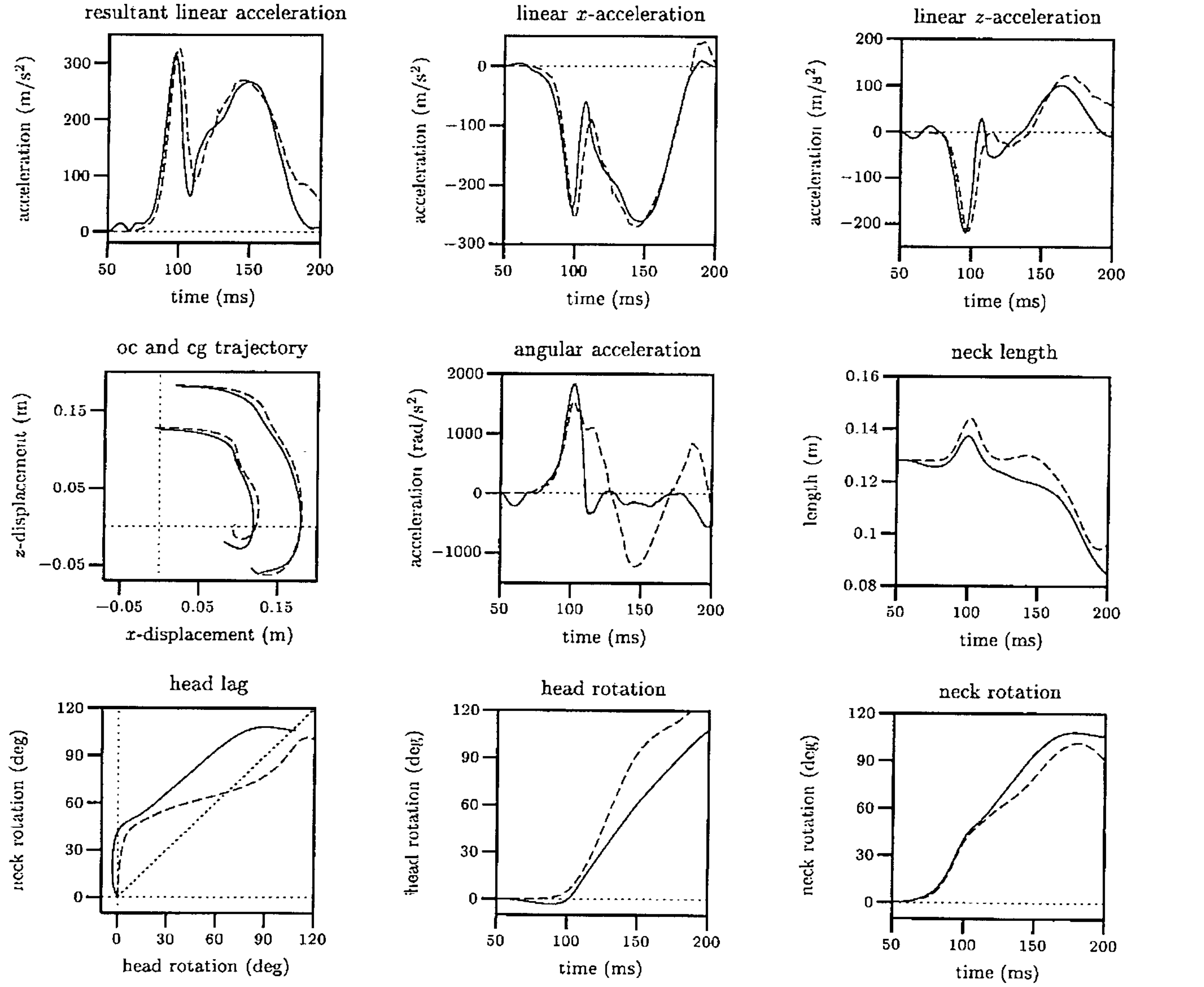

Figure 9: Response to frontal impact of the detailed model with active (solid line) and passive (dashed line) muscle behaviour. $+x$ is forwards, $+z$ is upwards.

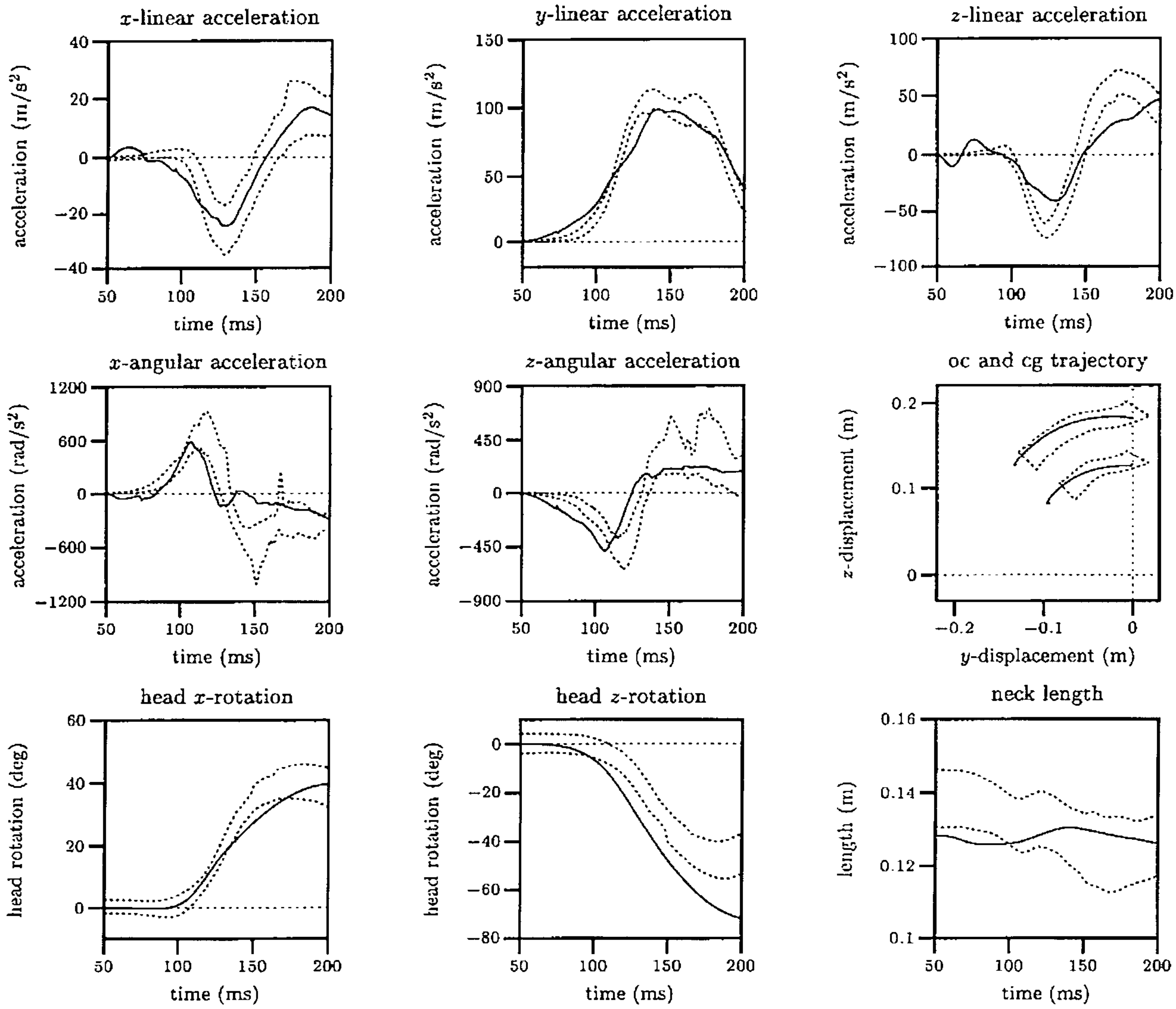

Figure 10: Response of the detailed model with active muscle behaviour (solid line) compared with human volunteer response corridors (dotted lines) for the lateral impact. $+x$ is forwards, $+y$ is to the left, $+z$ is upwards.

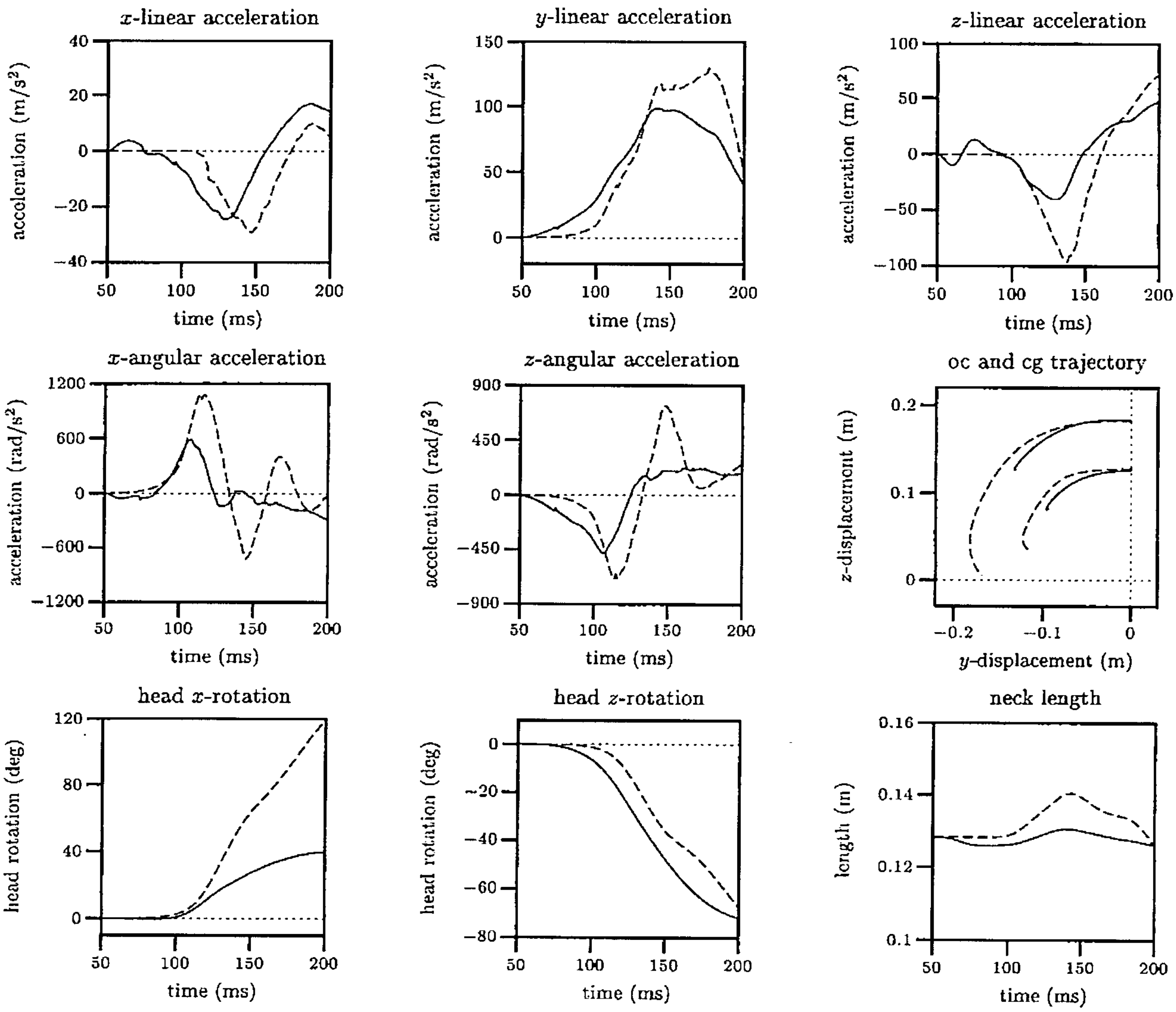

Figure 11: Response to lateral impact of the detailed model with passive (dashed line) and active (solid line) muscle behaviour. $+x$ is forwards, $+y$ is to the left, $+z$ is upwards.