Development of a Multi-Body Human Model that Predicts Active and Passive Human Behaviour

Riske Meijer[1], Edwin van Hassel[2], Jeroen Broos[1], Hala Elrofai[1], Lex van Rooij[1], Paul van Hooijdonk[2]

Abstract Active safety systems that start to act moments before the crash might be capable of anticipating the occupant's position, either by correcting it, or by taking the out-of-position into account. To develop such active safety systems, computer simulations of the occupant's pre-crash behaviour are very valuable in determining system performance. The objective of this study was to develop a run-time efficient computer human model that can simulate active as well as passive human behaviour such that it can be used to simulate the pre-crash and in-crash phase in one simulation run. The so-called active human model is a multi-body model, and is based on earlier developed human models in MADYMO and techniques for controlled active behaviour. The model's responses in a 1 g car braking, 15 g frontal, 7 g lateral and 3.6 g rear impact were compared to that of the volunteers. It was concluded that the active human model with controlled active behaviour with co-contraction of the neck muscles better predicts the behaviour of the volunteers than without co-contraction or completely passive behaviour. With the best fitting co-contraction levels and reaction times the maximum deviation from the average peak responses of the volunteers was at most 20%.

Keywords Human Model, active behaviour, pre-crash loading, multi-body dynamics.

I. INTRODUCTION

If an occupant is out-of-position (due to e.g. onset of rollover, vehicle dynamics or a secondary human task) just before a car crash, the outcome of the injury may be a lot worse than in a normal driving posture for which the restraint systems were designed [1]. Active safety systems that start to act moments before the crash, might be capable of anticipating the occupant's position, either by correcting it, or by taking the out-of-position into account. At the moment these active safety systems are being evaluated by means of volunteer tests in a car [2]. However, volunteer tests for optimization of the timing of an active restraint system has limitations in being very time consuming, less repeatable and possibly ethically difficult. Also, attempts were made to adapt the performance of a crash test dummy for pre-crash braking by adding foam between the belt and the ATD [2] and by adding active components to a numerical model of an ATD [3]. However, crash test dummies were developed for severe impacts, and there are quite a few challenges to adapt or develop a hardware dummy for low severity impacts, such as pre-crash car movements. Development of a computer human model that can predict pre-crash kinematics may be done much faster than the development of a hardware dummy. Such a computer human model could also be used in virtual testing of active restraint systems in the future, like the pedestrian human models in the Euro NCAP Pedestrian Safety Assessment for active bonnets [4]. As such, to develop active safety systems, computer simulations of the pre-crash behaviour of the occupant are very valuable in determining system performance.

In order to evaluate the effect of an active safety system during the crash, it would be most effective if a human model for pre-crash kinematics could also predict the kinematics as well as the risk of sustaining injuries during the crash. Risk of injuries are defined by injury criteria and accompanying limits. These limits were defined for the hardware dummies, and therefore might be different for human models. Nevertheless, measures resulting from a simulation with a human model, kinematics as well as existing injury criteria, can be used to improve an integrated safety system as well. Furthermore, several studies [5]-[7] showed that the muscle activation significantly affects the kinematics in low severity impacts or pre-crash car movements. As such, for real-world safety purposes, the various states of awareness and reactions of drivers and occupants on

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the impending crash should be taken into account. To be able to simulate the various states of awareness and reactions, the active human behaviour should be modelled in addition to the passive behaviour [8]. Besides models discussed in this study, various other detailed human models have added components simulating active muscle features [9]-[11]. The objective of this study was to develop a run-time efficient computer human model that can simulate active as well as passive human behaviour such that it can be used to simulate the pre-crash and in-crash phase in one simulation run.

II. METHODS

Creation of the Active Human Model

The active human model has been developed in the multi-body and finite element software package MADYMO [12]. For run-time efficiency only multi-body techniques were used. Since, in the past various human models were developed in MADYMO, and also various techniques to make certain human body parts active and even controlled active, these were used to create a basis for the active human model. Besides taking the (controlled) active components from the available human models, other human models or components thereof were used which improve the active human model for real-world loading conditions. The available models and techniques that were used were:

- Facet occupant model [13]-[15]
- Ellipsoid pedestrian model [15]-[17]
- Controllers and actuators to stabilise the spine of the facet occupant model [15][18]
- Detailed neck model with muscles [15][19]
- Controller to stabilise the detailed neck model via the muscles [7]
- Detailed arm models with muscles of a preliminary version of the active human model [5]
- Detailed leg models with muscles [15][20]
- Shoe model

Since occupant and pedestrian load cases have a lot in common, and both kind of models were available, the most biofidelic model components were used for the creation of the active human model. In doing so, the first step was to create a passive human model that is biofidelic for occupant as well as pedestrian impacts. The head, arms and thorax of the latest version of the facet occupant model [15] were combined with the detailed leg models with muscles [20]. The main difference between the latest version of the facet occupant model and that of [13][14] is that it has a refined skin for the neck, thorax and abdomen, which is important for the accuracy of the contact with restraint systems like belts and airbags in a crash simulation. First, the muscles in the legs were removed. Next, the knee characteristics for lateral loading as well as the bending and fracture joints of the ellipsoid pedestrian model were copied to the legs of the new human model. The bending and fracture joints were positioned on the bones in the legs in order to create a realistic bending and/or fracture behaviour of the whole leg. Next, the mesh of the legs was refined for a better contact with the front area of finite element car models for pedestrian impact simulations and with the interior for occupant impact simulations. By means of simulating the head impact tests of Melvin *et al.* [21][22], the contact characteristic of the head was updated. Finally, left and right shoe models were fit to the left and right feet of the new human model, respectively, and contacts between the feet and the shoes were defined. The pelvis, femur and tibia were added to the contact definitions of the lower extremities in order to simulate a location dependent penetration at the lower extremities in contact with the environment. The models that were used to create the new human model are shown in Fig. 1 to Fig. 5. The resulting human model is called the facet pedestrian model [15] and is shown in Fig. 6.

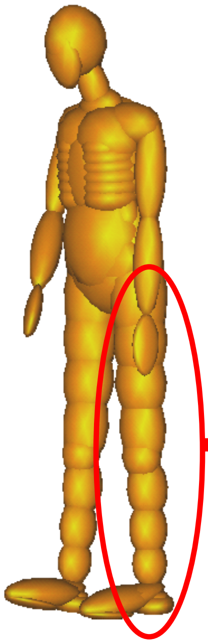

Fig. 1. Ellipsoid pedestrian model.

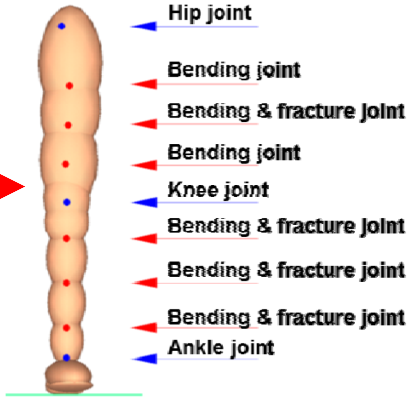

Fig. 2. Bending and fracture joints in ellipsoid pedestrian model's legs.

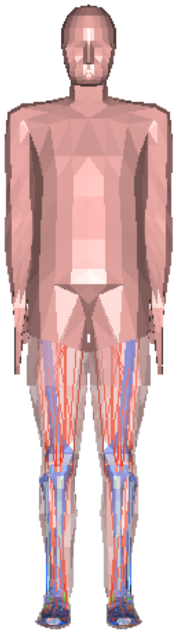

Fig. 3. Detailed leg models in facet occupant model.

Fig. 4. Facet occupant model.

Fig. 5. Shoe model.

Fig. 6. The facet pedestrian model.

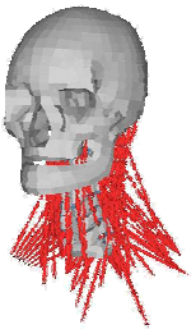

Fig. 7. Detailed neck model.

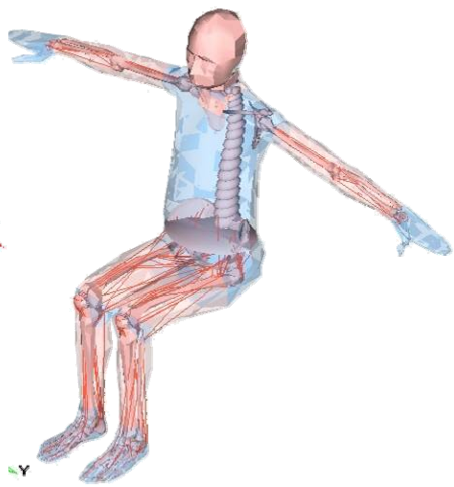

Fig. 8. Detailed arm models in a preliminary version of the active human model.

The second step was to create the basis of the active human model by adding the available detailed models with muscles to the facet pedestrian model that were necessary to simulate the active behaviour in a pre-crash phase in a biofidelic way. The neck of the facet pedestrian model was replaced by the detailed neck model [19], and the arms were replaced by detailed arm models of a preliminary version of the active human model [5]. The muscles of the detailed leg models [20] were added to the legs of the facet pedestrian model. The models that were used to create the basis of the active human model are shown in Fig. 3, Fig. 6, and Fig. 8.

Next, the contact characteristics of the thorax were updated in order to improve the active human model's interaction with belts. Since the outer mesh (skin) would wrinkle at the buttocks and knee locations when it is positioned from standing to seating position, separate meshes for the model in a standing position and in a seating position were generated for these body regions. A wrinkled mesh will cause unrealistic force-penetration calculations in contact interactions with the environment. From the model with the standing mesh other standing positions, e.g. walking, can be defined and from the model with the seating mesh other seating positions, without the mesh getting wrinkled. Apart from the mesh at these body regions, the standing and the seating active human model are identical. The resulting active human model in a standing position, seating position, and transparent are shown in Fig. 9 to Fig. 11.

Fig. 9. The active human model in pedestrian walking position.

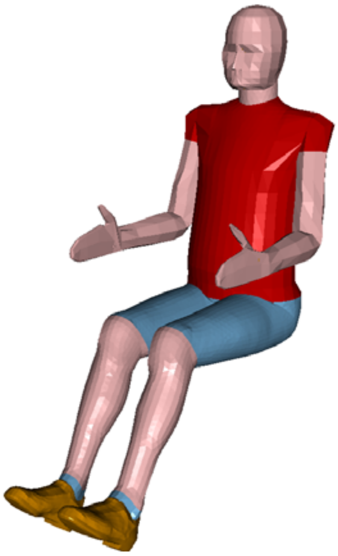

Fig. 10. The active human model in seating position.

Fig. 11. The active human model transparent showing the bones and muscles inside.

The third step in the creation of the active human model was the addition of sensors, controllers and joint torque actuators for simulating active behaviour in a pre-crash phase. For this study, the active behaviour in the pre-crash phase was assumed to be posture maintenance only, i.e. the model will try to maintain its initial position under the influences of external disturbances. However, the model was constructed in such a way that it is possible to simulate some voluntary movements or human cognitive reactions (such as bracing) as well, based on defined control parameters and muscle activation patterns. The sensors, controllers and joint torque actuators for stabilising the spine [18] were added between Sacrum and T1, and adapted as described below. The sensors and controllers for stabilising the neck by means of the neck muscles [7] were originally developed in the software program Matlab and worked via a Matlab-MADYMO coupling. This Matlab script was translated to the MADYMO code, the control scheme was adapted as described below, and added to the new model. Sensors and controllers were added to the muscles of the arms for stabilising the elbows. Also, sensors and controllers were added to the muscles of the legs for stabilising the hips. The stabilising of the elbows and the hips was modelled in a non-biofidelic but pragmatic way to just make the model move to its initial position, also without belts. The stabilising of the neck, hip, elbow and spine is described in more detail below.

All controllers in the active human model are in principle based on the scheme shown in Fig. 12. This controller scheme is present for each degree of freedom that is controlled. By changing parameters in the 'user input' block various situations can be simulated, e.g. different states of awareness and reactions. The basic controller scheme starts with the sensors. For each degree of freedom that is controlled, a sensor is defined to measure the current rotation. The target signal is by default 0, which means that the initial position of the active human model is the target position. The target signal could also be another rotation function, e.g. to simulate a voluntary movement or a reaction. The rotation error is then calculated as the difference between the sensor and target signals. The next step in the controller scheme is the reaction time. Here, the reaction time represents the total time from sensing, transfer of the signal to the brain, and the processing in the brain. For a stable response of the active human model, the reaction time was implemented in the following way:

- Rotation error due to pure stabilising behaviour causes a direct control action.
- Rotation error due to a new event only causes control action with a delay of the reaction time.

A new event is defined as a pulse from outside causing a rotation error that is larger than the maximum error that occurred in the simulation up to the current time step. New events are automatically detected by the active human model. If the rotation error remains below the maximum, the signal is transferred directly, but if the error is above the maximum, it is limited to the maximum during the reaction time before it increases further, see Fig. 13.

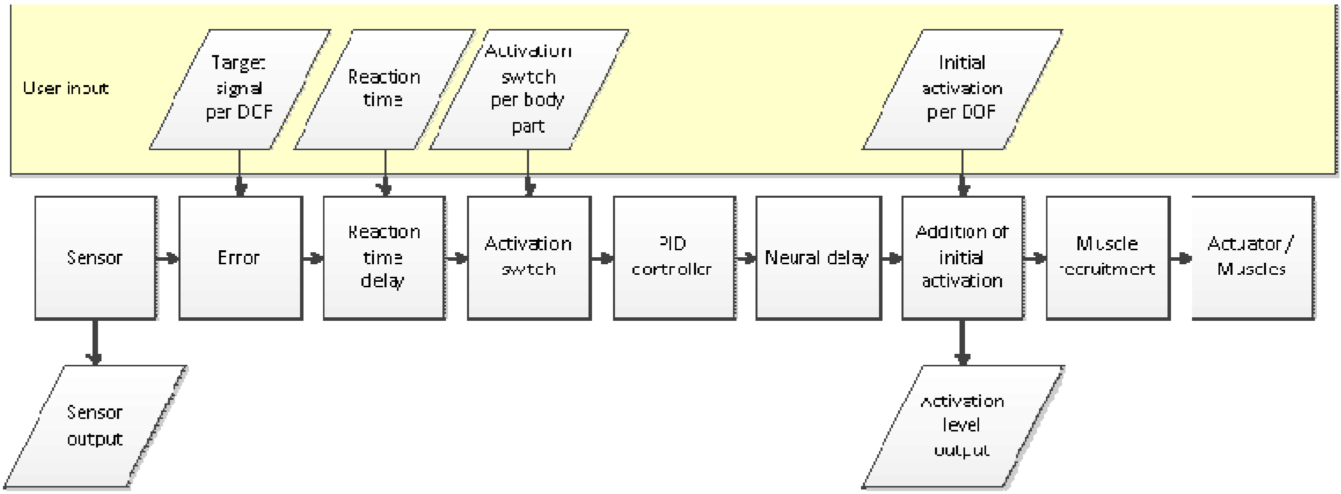

Fig. 12. Basic controller scheme of the neck, spine, hips and elbows of the active human model. The parameters in the 'user input' block can be changed to simulate various situations.

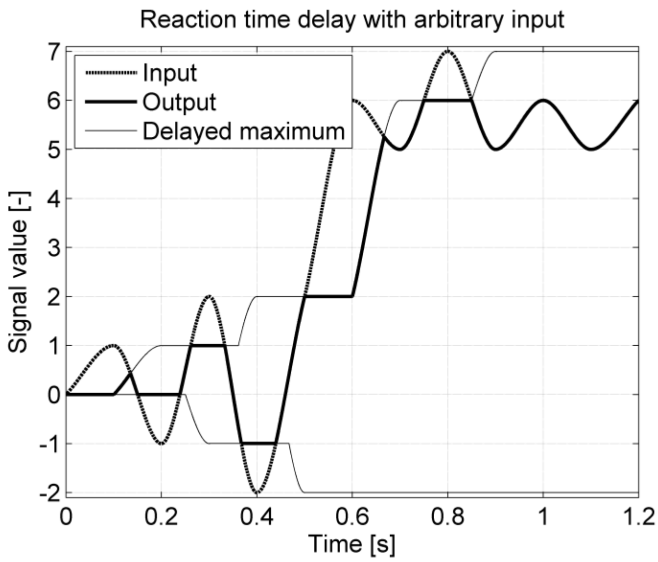

Fig. 13. Example of the effect of the reaction times on the signal output of the controllers. The output signal is limited by the delayed (reaction time) maximum of the input.

The activation of the active human model can be changed by switching the active behaviour per body part (neck, spine, elbow, hip) on or off. This is done by multiplying the control signal with the activation parameter, where '0' results in no active behaviour, and '1' results in active behaviour (posture maintenance).

All controllers are PID controllers, with a transfer function that is defined as:

$$Hcm = Km\,(1 + \frac{1}{\tau_{im} s} + \tau_{dm} s) \tag{1}$$

where *Hcm* is the transfer function of the controller, *Km* is the proportional action (P), τ_{im} is the integration time (I) and τ_{dm} is the differentiation time (D). The PID controller aims to reduce the error by calculating an activation level. The P-action changes the controller action based on the present error. The I-action makes sure the controller will reduce the error to zero by integrating the past errors. To damp out future errors (oscillations) the D-action will reduce the error based on the current rate of change. Without the I-action the controller causes the activation level to be in equilibrium with the external load, so a constant motion error might remain.

After the PID-controller, there is a neural delay defined. The neural delay represents the time it takes for the signal transfer from the brain to the muscle and the time it takes for the muscle to convert the signal into a force. The neural delay is defined differently for each body part depending on the distance to the brain, so being the shortest for the neck and longest for the legs. The behaviour of this neural delay is frequency dependent and defined as:

$$d(output)/dt = (input - output)/\tau_{neural\_delay} \quad (2)$$

So, signals with lower frequencies are transferred better than signals with higher frequencies, and the delay decreases with increasing frequencies. Fig. 14 shows the effect of the neural delay on the output signal for a step function, and Fig. 15 for a swept sine wave with increasing frequency as input.

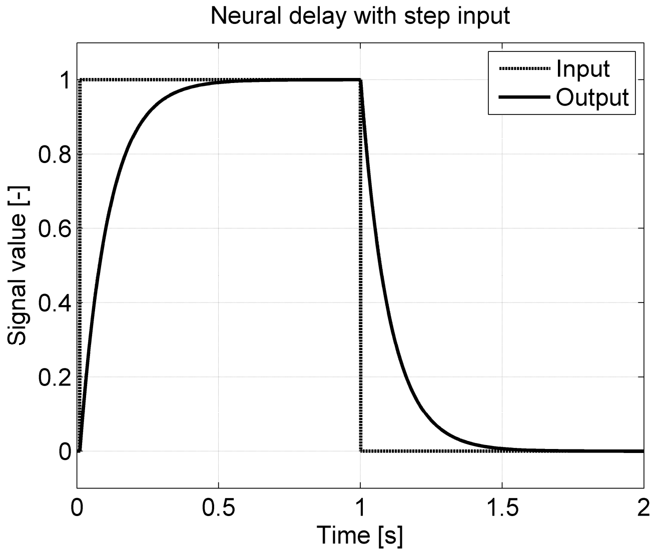

Fig. 14. Frequency dependent behaviour of the neural delay for a step function.

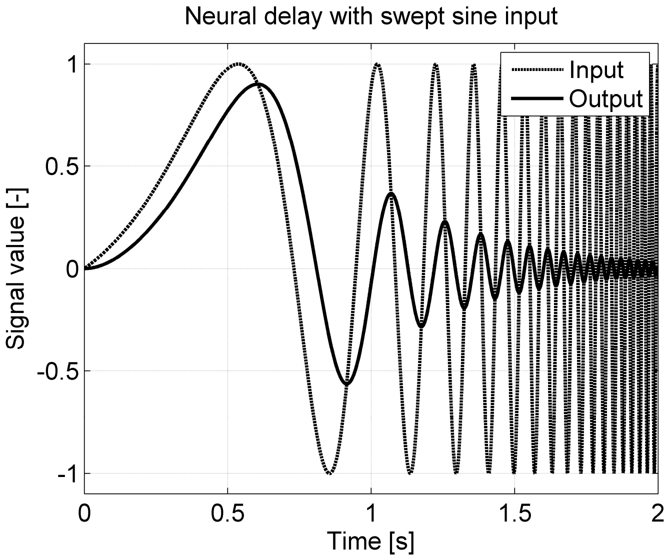

Fig. 15. Frequency dependent behaviour of the neural delay for a swept sine wave.

For both a passive occupant model and the active human model, a settling simulation needs to be performed in order to have the model in equilibrium with the seat and the gravity load. From the settling simulation the joint position outputs at the end of the simulation are defined and used as input for the pre-crash/crash simulation. Also, for the active human model initialisation of the controllers is required. Similar to the joint positions, from the same settling simulation the activation level outputs are defined and used as input for the pre-crash/crash simulation.

Finally, the signal from the controller, being one signal per degree of freedom, is converted to a signal for each actuator. This is done by means of a recruitment table in which there is for each degree of freedom and for each actuator a constant factor defined, to obtain a weighted combination of the three degrees of freedom. The converted signals are then used as input for the actuators, which can either be muscles or multi-body actuators.

The neck controller acts in three degrees of freedom, being the three rotations of the head. For each degree of freedom the neck controller follows the basic scheme as explained above. Depending on the initial settings, the head rotations are either calculated relative to the reference space, to keep the head upright, or relative to T1, to keep the neck straight. As the vestibular system is in the head, usually a human will aim to keep its head upright. However, for large rotations of the body (more than 90 degrees with respect to inertial space), e.g. in a pedestrian impact, a strategy which keeps the neck straight is assumed to be more realistic. The muscle recruitment table for the neck is taken from the model of [7]. Here, the recruitment table is balanced, which means that an error in one degree of freedom results in a torque in only that degree of freedom. Besides the control on the three degrees of freedom of the head, co-contraction can be added to the neck muscles. Here, co-contraction is defined as the simultaneous tension of all muscles without giving any resultant torques. Co-contraction will always be present to some extent, and is possibly higher if a person is tensed. In the active human model the co-contraction level is defined by the initial input as a relative value (0-1) of the active part of the muscle force. Standard the co-contraction starts at the start of the simulation and is constant over the whole simulation, however the co-contraction can be made variable and delays can be included. So, standard the reaction time and the neural delays do not work on the co-contraction levels. The co-contraction level is included in the calculation of the muscle recruitment. In fact the co-contraction levels stiffen the neck.

The controllers on the left and right hip each act in three degrees of freedom, being the three rotations of the hip joint, flexion-extension, medial-lateral rotation, and abduction-adduction. For each degree of freedom the hip controller follows the basic scheme as explained above. The muscle recruitment table for the hip is set up such that for a specific degree of freedom the muscles that have most effect in that degree of freedom are activated the most.

The controllers on the left and right arm each act in only one degree of freedom per side, being the elbow flexion-extension. For flexion-extension the elbow controller follows the basic controller scheme as explained above. The muscle recruitment for the elbow is very simple, just dividing the muscles in a group of flexors and a group of extensors and activating all muscles in one group to the same extent.

For the spine, no muscles are included because of the complexity of the musculature of the thorax. Instead, multi-body actuators are used to directly apply a torque between two successive vertebrae. The spine controller acts in three degrees of freedom per vertebra for each of the 5 lumbar and 12 thoracic vertebrae, so 17 vertebrae in total. For each degree of freedom the spine controller follows the basic scheme as explained above, except for the muscle recruitment. For each vertebra, sensors are defined to measure the angle of the vertebra relative to the sacrum (pelvis). For the spine no target functions are defined. Hence, the rotation error for the spine is equal to the sensor output. If a vertebra rotates, the controller applies a torque to that vertebra as well as to the vertebrae below, such that the spine is in a stable position. This results in the following activation signals:

- T1 activation: T1 controller
- T2 activation: T2 controller + T1 activation = T2 controller + T1 controller
- T3 activation: T3 controller + T2 activation = T3 controller + T2 controller + controller T1
- Etc.

Finally, the initial activation levels are added to the activation signal (as in the basic scheme), and the signal is used in the actuators.

The active human model's response was evaluated using almost all the validation data that was used for the human models and components the model is comprised of. The validation data comprise of low to high severity blunt impact tests on various body parts as well as low to high severity full-body load tests. The validation data are from published post mortem and living human subject tests. In order to simulate the tests in a most biofidelic way, the post mortem human subject (PMHS) tests were simulated with the activation of all controlled body parts switched off, and the living human subject tests were simulated with the activation of all these body parts switched on. By doing so, the passive behaviour of the model is validated with the PMHS tests, and the combined active and passive behaviour with the living human subject tests.

Evaluation of the Active Behaviour

This study focuses on the evaluation of the active behaviour of the active human model. Therefore, some full-body volunteer impact tests were selected to show the active human model's response in frontal, lateral and rear direction, see TABLE I. Compared to the validation data that were available from the earlier developed human models one new data set was added for this study, i.e. the 1 g car braking test with 3-point belt (Test nr. 1 in TABLE I).

As the co-contraction levels and the reaction times are unknown parameters in these volunteer tests, these parameters were varied in the simulations. Reported motor reflex delays vary between 10 and 120 ms [23]-[27]. Since, the activation of the muscles or actuators in each body part (head, spine, elbows and hips) in the active human model starts at the onset of rotation of that body part delayed by the reaction time, and the muscles activations in the volunteer tests could be triggered earlier, e.g. visual, vestibular etc., reaction times (RT) of 0 ms, 50 ms and 100 ms were simulated. For the neck muscles co-contraction levels (CCR) of 0.0, 0.3, 0.5, 0.7 and

0.9 were simulated for each of the reaction times. The active human model was settled in the environment model of the test set-up for each co-contraction level separately. The co-contraction level was kept constant during the whole simulation. Next, belts as were used in the tests were modelled and fitted around the active human model for the simulations of Test nr. 1, 2 and 3 in TABLE I.

The output signals from the simulations were each filtered the same as the measurement signals from the tests. The best fitting co-contraction levels and reaction times for the simulation of each volunteer test were estimated based on that the active human model's response deviates least from the average peak of the volunteer responses with a maximum of 20% as well as on the time period its response is within the response corridors of the volunteers.

TABLE I
FULL-BODY LOAD TESTS

Test nr.	Test Type	Subject	Direction	Position	Pulse	Ref.
1	car braking 3-p belt	volunteer	frontal	sitting	1 g during 1.7 s	[2]
2	sled rigid seat 5-p belt	Volunteer	frontal	sitting	peak 15 g	[28]-[31]
3	sled rigid seat 5-p belt and side-wall	volunteer	lateral	sitting	peak 7 g	[30] [32]
4	sled rigid seat	volunteer	rear	sitting	peak 3.6 g	[33]

III. RESULTS

The horizontal displacements of the chest and neck of the active human model in the 1 g car braking test (Test nr. 1 in TABLE I) are compared to the volunteer responses in Fig. 16 and Fig. 17. These figures only show the simulation results at a co-contraction level of 0.5, since the chest and neck displacements are hardly affected by the co-contraction level. This is explained by that the co-contraction level mainly affects the displacement of the head with respect to T1 vertebra, and the neck displacement in this test was measured at a collar around the neck of the volunteer at vertical location similar to C7 vertebra. The co-contraction level of 0.5 was chosen based on engineering judgement and on the fact that the volunteers knew that a braking event would happen, but were not informed about the timing. Fig. 16 and Fig. 17 show that the reaction time hardly affects the chest and neck displacements either. The reason for this is that the reaction times are much shorter than the braking event, so the activation already takes place at the start of the braking event. Fig. 16 shows that the maximum chest displacement is less than 20% smaller than the average maximum chest displacement of the volunteers. Fig. 17 shows that the maximum neck displacement is approximately 20% smaller than the average maximum neck displacement of the volunteers. This can be explained by the fact that the volunteers were wearing a thick winter coat which was not accounted for in the simulation. This caused the active human model to be restrained by the belt a bit earlier than the volunteers, resulting in a smaller chest displacement and smaller neck displacement than the volunteers.

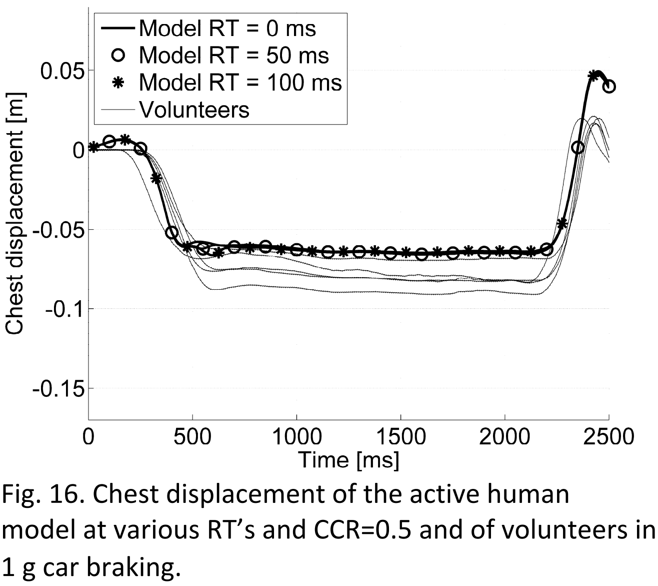

Fig. 16. Chest displacement of the active human model at various RT's and CCR=0.5 and of volunteers in 1 g car braking.

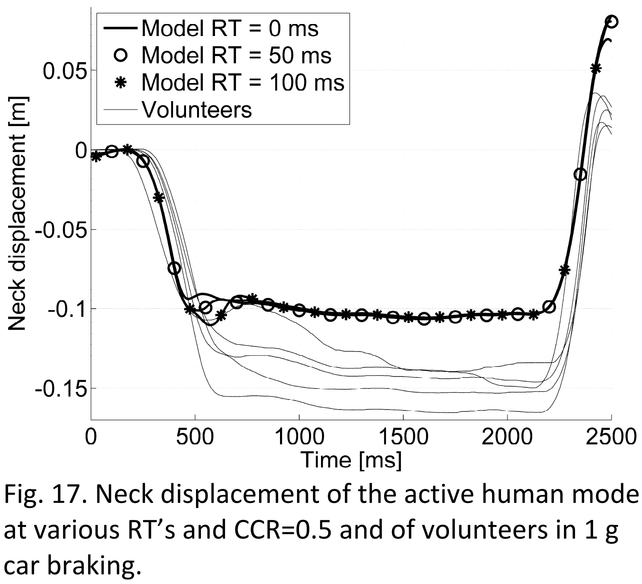

Fig. 17. Neck displacement of the active human model at various RT's and CCR=0.5 and of volunteers in 1 g car braking.

The head rotation and head centre of gravity (COG) horizontal displacement of the active human model in the 15 g frontal impact test (Test nr. 2 in TABLE I) are compared to the response corridors of the volunteers in Fig. 18 to Fig. 23. These figures show that a shorter reaction time as well as a higher co-contraction each decrease the head rotation and head forward displacement. The active human model best predicts the volunteer responses at RT=0-50 ms with CCR=0.5. At these parameter values the peak head rotation and the peak head forward displacement deviate less than 10% from the average peak of the volunteers, and the responses are for a large extent within the response corridors of the volunteers. Fig. 18 to Fig. 23 also show the active human model's response with active behaviour switched off, indicated as 'Passive'. These figures show that at RT=0 ms the active behaviour of the model decreases the peak head rotation by more than 21% and the peak head displacement by more than 6%. At RT=100 ms the response of the model with active behaviour switched off is almost similar as with active behaviour with CCR=0.

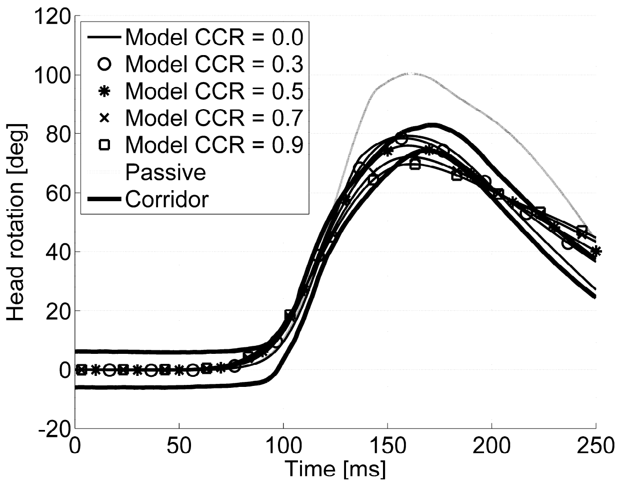

Fig. 18. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 15 g frontal impact.

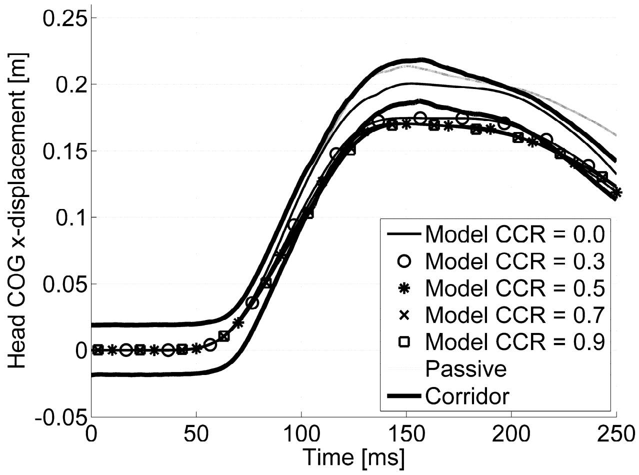

Fig. 19. Head x-displacement w.r.t. to the sled of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 15 g frontal impact.

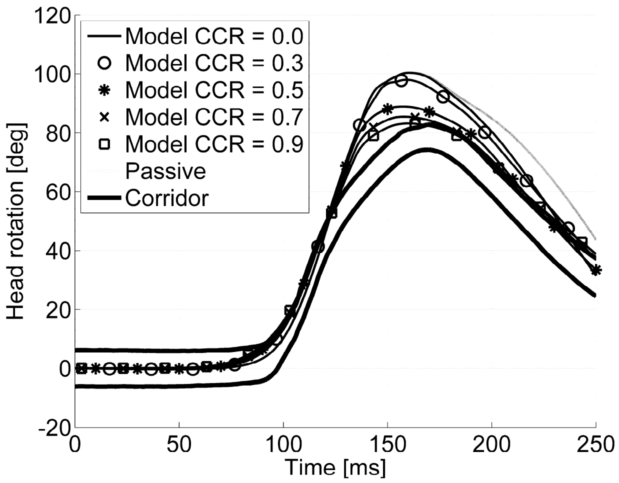

Fig. 20. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 15 g frontal impact.

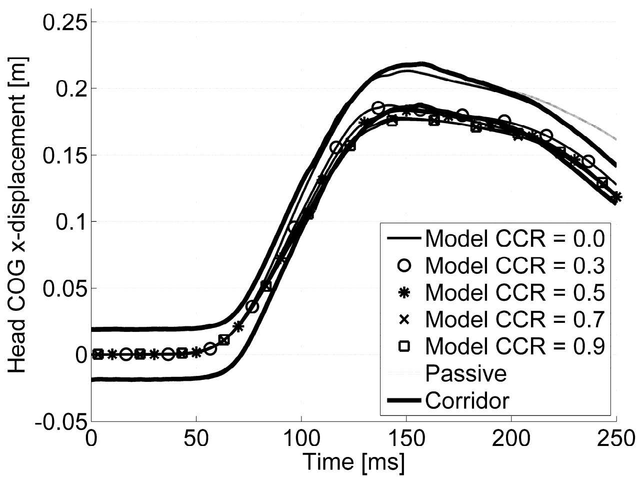

Fig. 21. Head x-displacement w.r.t. to the sled of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 15 g frontal impact.

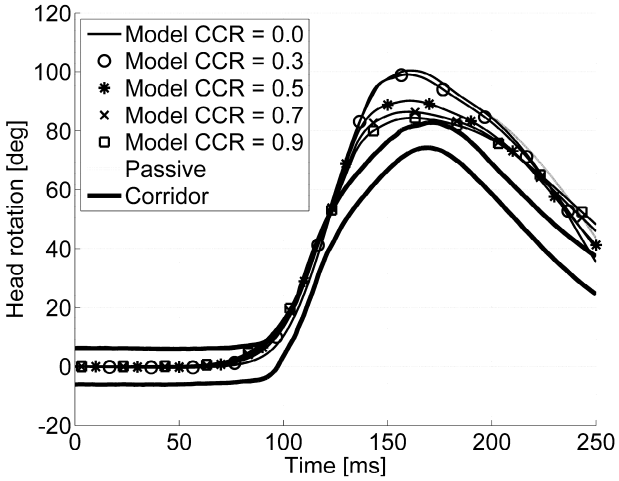

Fig. 22. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 15 g frontal impact.

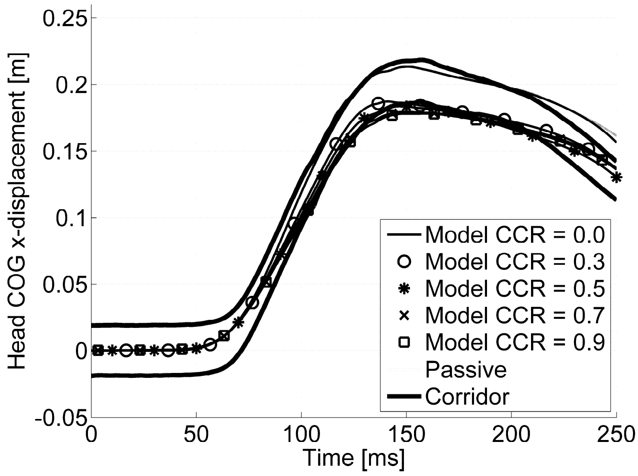

Fig. 23. Head x-displacement w.r.t. to the sled of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 15 g frontal impact.

The head rotation and head COG lateral displacement of the active human model in the 7 g lateral impact test (Test nr. 3 in TABLE I) are compared to the response corridors of the volunteers in Fig. 24 to Fig. 29. These figures show, that a shorter reaction time as well as a higher co-contraction each decrease the head rotation and head lateral displacement. This finding is in line with the active human model's response in the 15 g frontal impact shown in Fig. 18 to Fig. 23. The active human model best predicts the volunteer responses at RT=0-50 ms with CCR=0.9. At these parameter values the peak head rotation and the peak head lateral displacement deviate less than 20% from the average peaks of the volunteers, and the responses are for a large extent within the response corridors of the volunteers. However, the head lateral displacement starts earlier than that of the volunteers for all parameter variations. Comparing the movies of the volunteers with the active human model in the simulations, it was seen that the shoulder of the active human model is more compliant than that of the volunteers causing an earlier and larger translation of the head. The difference in shoulder compliance is explained by that the volunteers in this test were trained soldiers and the active human model represents an average male person.

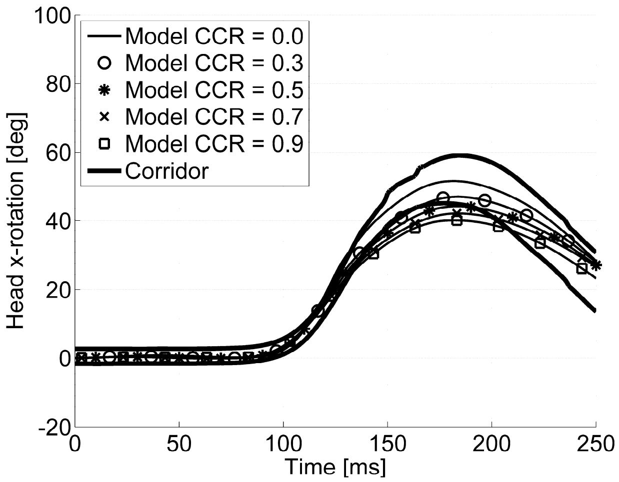

Fig. 24. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 7 g lateral impact.

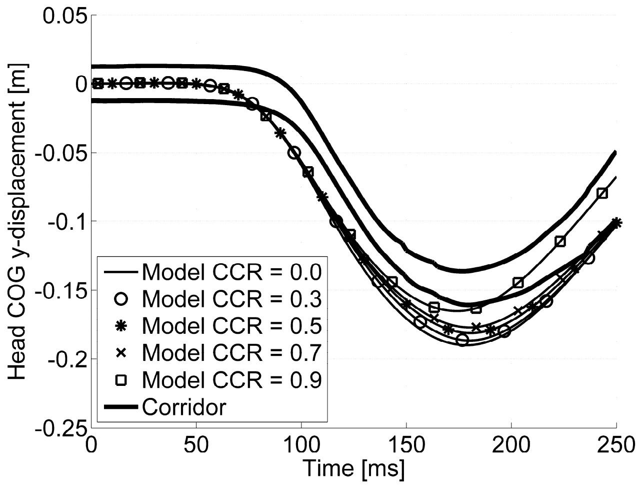

Fig. 25. Head y-displacement w.r.t. to the sled of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 7 g lateral impact.

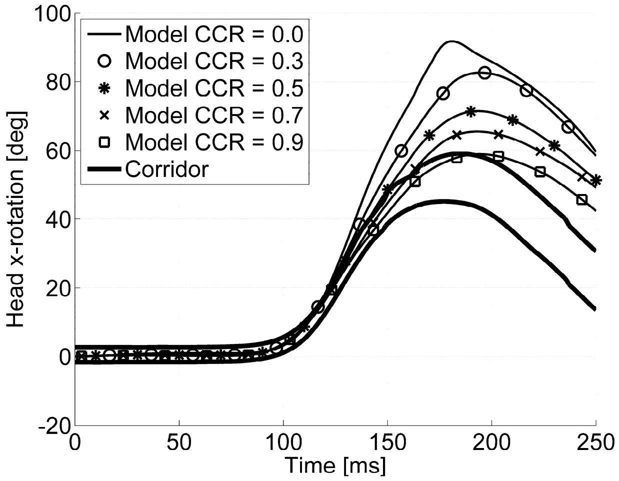

Fig. 26. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 7 g lateral impact.

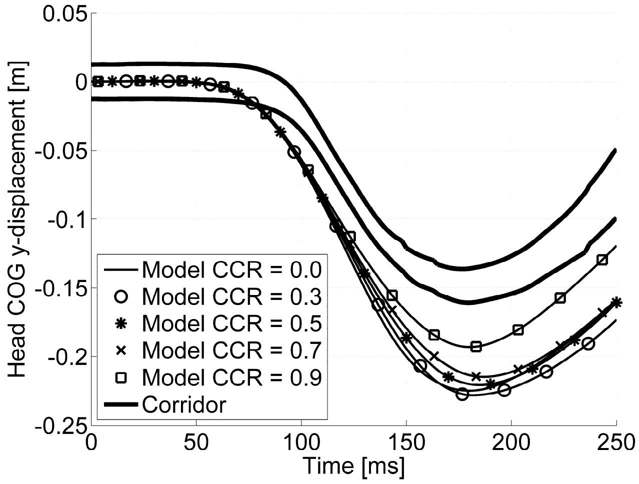

Fig. 27. Head y-displacement w.r.t. to the sled of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 7 g lateral impact.

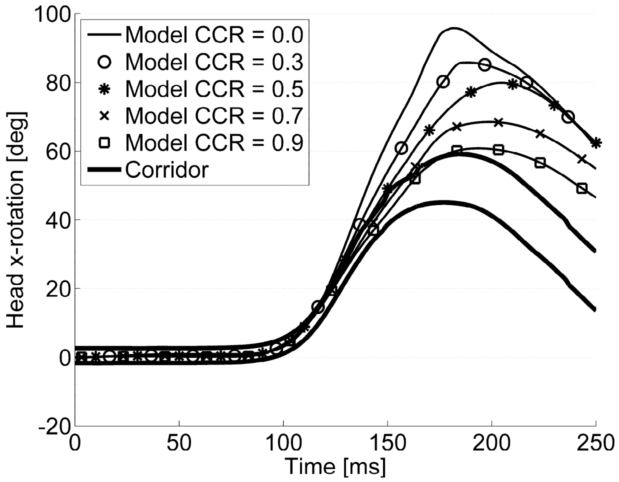

Fig. 28. Head rotation w.r.t. to the sled of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 7 g lateral impact.

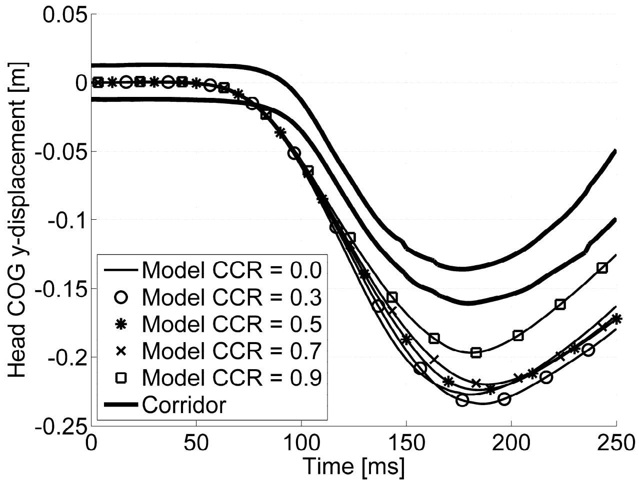

Fig. 29. Head y-displacement w.r.t. the sled of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 7 g lateral impact.

The head rotation with respect to T1 and the neck horizontal displacement with respect to T1 of the active human model in the 3.6 g rear impact test (Test nr. 4 in TABLE I) are compared to the response corridors of the volunteers in Fig. 30 to Fig. 35. In line with the active human model's response in the 15 g frontal impact and the 7 g lateral impact, these figures show that a shorter reaction time as well as a higher co-contraction decrease the head rotation and head rearward displacement. The active human model best predicts the volunteer responses at RT=100 ms with CCR=0.9. At these parameter values the peak head rearward rotation is approximately equal to the average peak of the volunteers, and the peak head rearward displacement deviates approximately 20% from the average peak of the volunteers. The active human model's responses are almost completely within the response corridors of the volunteers, however the timing of the peak responses is too early.

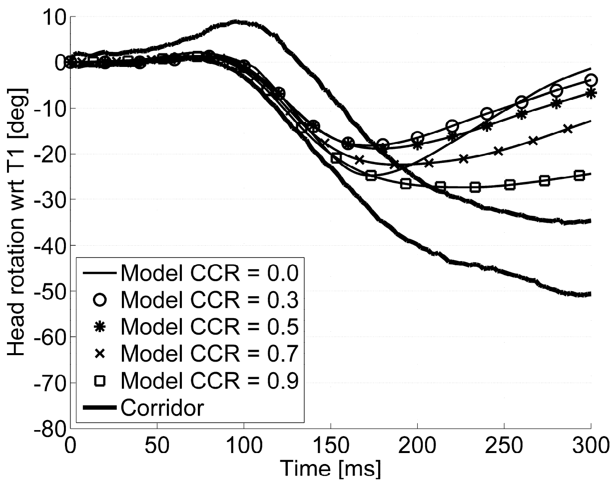

Fig. 30. Head rotation w.r.t. T1 of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 3.6 g rear impact.

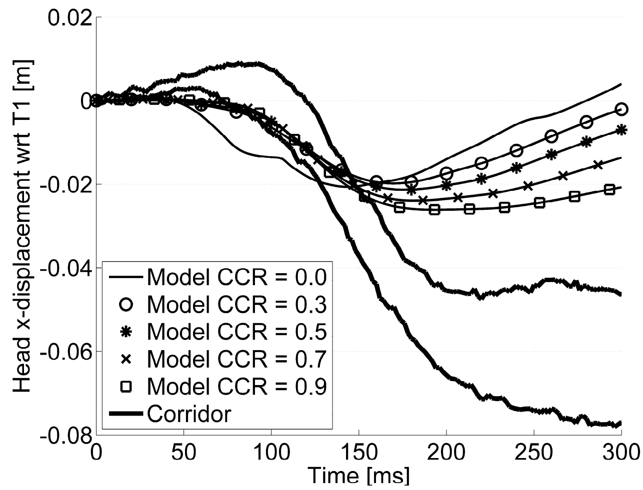

Fig. 31. Head x-displacement w.r.t. T1 of active human model at various CCR levels and RT=0 ms and corridors of volunteers in 3.6 g rear impact.

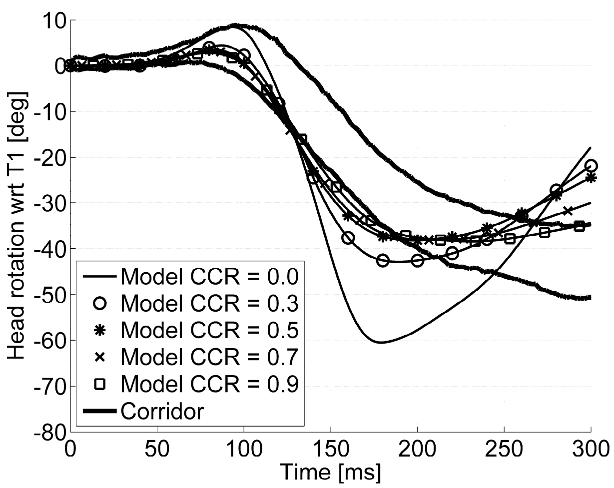

Fig. 32. Head rotation w.r.t. T1 of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 3.6 g rear impact.

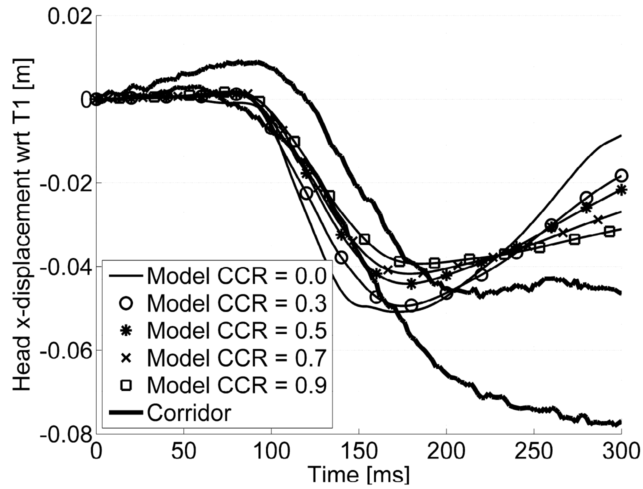

Fig. 33. Head x-displacement w.r.t. T1 of active human model at various CCR levels and RT=50 ms and corridors of volunteers in 3.6 g rear impact.

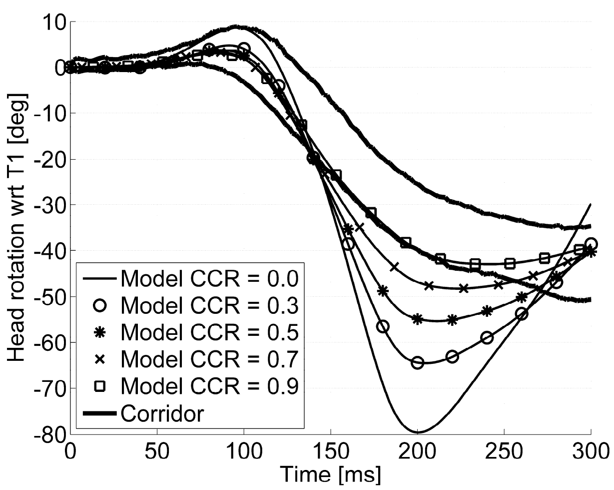

Fig. 34. Head rotation w.r.t. T1 of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 3.6 g rear impact.

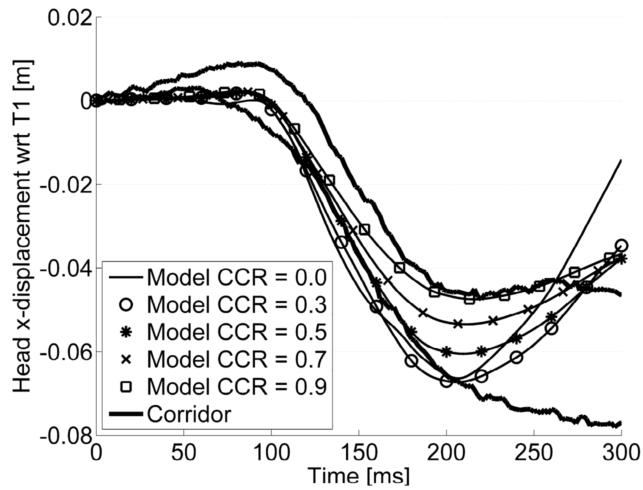

Fig. 35. Head x-displacement w.r.t. T1 of active human model at various CCR levels and RT=100 ms and corridors of volunteers in 3.6 g rear impact.

IV. DISCUSSION

The simulation results of the volunteer tests of TABLE I show that the newly developed active human model with controlled active behaviour with co-contraction of the neck muscles results in a better prediction of the kinematics of living human subjects than without co-contraction or completely passive behaviour. The simulation results also show that with the best fitting co-contraction levels and reaction times for each test the active human model predicts the average peaks of the volunteer responses with a maximum deviation of 20%.

A co-contraction level of 0.9 found for the 7 g lateral and 3.6 g rear impact seems to be rather high taking into account that human beings will probably only be able to activate their muscles at maximum extent for a very short period. Partly, this could be explained by the volunteers in the 7 g lateral impact were trained soldiers and the volunteers in the 3.6 g rear impact were young male persons. However, it could also be an indication that the maximum forces of the muscles or the passive resistance of the neck of the active human model is too low, or both.

A limitation of the neck of the active human model is that the muscles all have the same reaction time, and neural delay, which is not the case in reality. This is also the case for the muscles in the arms and in the legs and for the actuators in the spine. Only a difference in neural delay between the four controlled body parts is accounted for in the model. Further, limitations of the active human model are that for the hip and the elbows, the muscle recruitment is not balanced, as it is for the neck, and also no co-contraction is included. However, in the volunteer tests simulated in this study the active behaviour of the arms and legs did not or hardly affect the active human model's response, except for the legs in the 1 g car braking test. Muscle recruitment in general is based on presumed muscle strategies, which require detailed validation.

For improving the behaviour of the neck of the active human model, more information on the maximum force per muscle, reflex times and neural delays of each muscle and validation of the lumped passive resistance of the neck is needed. Co-contraction levels and reaction times vary per volunteer and per test set-up. Therefore, volunteer tests in which also EMG is measured are needed to be able to more accurately validate the response of the active human model. The EMG could give an indication of the co-contraction levels of the thick superficial muscles when the maximum voluntary contraction of each volunteer is measured, as well as an indication of the reaction times.

Further, bracing by means of the arms and legs is expected to be an important factor in the kinematics of the whole human body in pre-crash loading situations for the driver at least. It is possible to model bracing with the current active human model, however currently there is no validation data available for this. The influence of the co-contraction level and reaction time shown from the simulations of the 15 g frontal impact implicate to also add effects of inattention or drowsiness to the active human model. Also, for effects of inattention or drowsiness on the human kinematics in pre-crash and crash no validation data is currently available.

Despite the simplifications of the controlled active behaviour of the active human model mentioned above, the model is capable of predicting the kinematics of the volunteers in the 1 g car braking, the 15 g frontal, the 7 g lateral, and the 3.6 g rear impact tests. To evaluate if the active human model can predict the human responses in a pre-crash event followed by a crash, comparison of its responses to human subjects in low severity lateral and oblique impacts as well as in high severity frontal, lateral and rear impacts with a 3-point belt are still needed. Currently, the muscles and actuators keep working regardless of the magnitude of the external loading. However, at high severity impacts the loads on the joints by the external loading are probably much higher than that of the actuators, and will therefore hardly affect the kinematics and internal body loads. In this case PMHS tests can be used for validation of the active human model's kinematics and internal body loads in high severity impacts. However, this has to be studied yet.

V. CONCLUSION

In this study a new multi-body human model was developed that contains controlled active behaviour of the neck, spine, elbows and hips. The so-called active human model was based on earlier developed human models or components thereof and techniques for controlled active behaviour. With the controlled active behaviour switched on for all controlled body parts the model is capable of full posture maintenance in an automotive seating situation, also without belts. In order to evaluate the active behaviour of the new model tests with volunteers in a 1 g car braking with 3-point belt, 15 g frontal impact with 5-point belt, 7 g lateral impact with 5-

point belt, and 3.6 g rear impact were simulated and the responses were compared. From the results of these simulations it was concluded that the active human model with controlled active behaviour with co-contraction of the neck muscles results in a better prediction of the kinematics of living human subjects than without co-contraction or completely passive behaviour. Also, from the simulation results it was concluded that with the best fitting co-contraction levels for the neck muscles and reaction times for each test the active human model predicts the average peaks of the volunteer responses with a maximum deviation of 20%. To check if the active human model can predict the human responses in a pre-crash event followed by a crash, comparison of the model's responses to human subjects in low severity lateral and oblique impacts as well as in high severity frontal, lateral and rear impacts with a 3-point belt are still needed. Also, it is recommended to add the most common human activation patterns in a pre-crash phase to the active human model, such as anticipation (bracing), and effects of inattention or drowsiness. For this, research into human cognitive response and muscle activation patterns is needed as well as a more detailed modelling of the active behaviour of the arms and the legs.

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