EVALUATION OF AN ACTIVE MULTI-BODY HUMAN MODEL FOR BRAKING AND FRONTAL CRASH EVENTS

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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. For the development and evaluation of such active safety systems, recently a run-time efficient multi-directional computer human model that can simulate active as well as passive human behaviour has been developed. The objective of this study was to evaluate this so-called active human model for simulations of braking followed by a frontal crash.
Simulations of published PMHS blunt frontal impact tests on the head, thorax, and abdomen showed that the model is capable of predicting the PMHS peak responses within 20% deviation from the PMHS response corridors. Here, the active behaviour was switched off. Simulations of published 1 g to 15 g full-body frontal impact tests with belted volunteers showed that the model is capable of predicting the volunteer peak responses within 20% deviation from the volunteer response corridors. Here, values for the unknown parameters reaction time and level of bracing in the neck (co-contraction of neck muscles) were assumed.
Also, simulations with the active human model in a car interior to which high severity impacts were applied (pulses from Euro NCAP car-barrier frontal impact tests), with and without preceding braking as well as with active behaviour switched on and switched off were performed. The results of these simulations showed that the model is robust and sensitive to effects of braking and active behaviour, and the effects of braking on the injury values are dominant over the effects of the active behaviour itself. However, the active behaviour is indispensable for correct simulation of the human pre-crash kinematics.
From this study it was concluded that the current active human model is capable of simulating realistic human full-body kinematics as well as realistic injury values for the head and the thorax in one single simulation of braking followed by a frontal crash. As such, the current active human model can be used for evaluating the effectiveness of active safety systems in frontal impacts.

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. As examples, a system accelerating the occupant rearward prior to a frontal collision in combination with a reversible belt pretensioner may result in an optimal position of an occupant at the time the crash occurs [2], and a reactive reversible belt pre-pretensioner that was designed to provide benefit in frontal, rearward and lateral crashes [3].
Thereby, several studies showed that the muscle activation significantly affects the kinematics in low severity impacts or pre-crash car movements. From volunteer sled tests at 2.5 G frontal impact a 47±12% decrease in head forward excursion due to bracing was observed, while it was 36±12% at 5.0 G [4]. In lap-belt only frontal sled tests up to 1.0 G it was shown that being tensed reduces head and neck flexion to nearly zero, while flexion of the lower spine was substantially reduced as well [5]. Based on vehicle driving tests in normal traffic, it was observed that head excursion and head flexion were much smaller when drivers applied the brakes themselves versus surprise autonomous braking [6]. As such, the various states of awareness and reactions of drivers and occupants on the impending crash should be taken into account in the development of active safety systems.
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. For this aim, recently, a run-time efficient multi-directional computer human model that can simulate active as well as passive human behaviour has been developed [7]. The objective of this study was to evaluate this so-called active human model for simulations of braking followed by a frontal crash.

METHODS

The Active Human Model

The active human model evaluated in this study is version 1.1 [8] available in the multi-body and finite element software package MADYMO version 7.4.1 [9]. The development of this model, which is based on earlier developed MADYMO whole body human models and human segment models, is described in [7]. A detailed description of this model can be found in [8]. The active human model in a frontal crash simulation is shown in Figure1. Below follows a short description of this model.

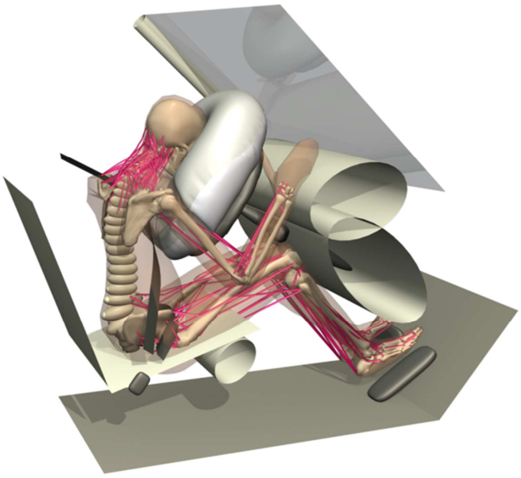

Figure1. MADYMO Active Human Model.

For run-time efficiency and for focussing on kinematics and global injury criteria only multi-body techniques are used in this model. The model consists of 186 bodies (178 rigid bodies and 8 flexible bodies). Inertial properties of the human body segments are represented fully by the inertial properties of the rigid and flexible bodies. The body segments are connected by kinematic joints representing the articular joints in a human body. For modelling the interaction with the environment and for visual representation the outer surface of the whole human body as well as for many bones is represented by meshes of shell-type massless contact elements (further referred to as facets). These facets are connected to the rigid and/or flexible bodies. The ribcage and the hand bones are not represented by facets, and the spinal vertebrae are represented by ellipsoids. The contact force and penetration of the model is calculated from stress-strain functions defined for the facets of the outer surface as well as for the underlying bones in the arms, legs and thorax. Since the bones were defined much stiffer than the outer surface, the penetration is dependent on the location of the contact at these body parts. For the rest of the body parts a combined flesh and bone stress-strain function was defined. The thorax includes flexible bodies for modelling the high deflection that can be seen at the ribcage and abdomen in frontal and side impacts.
For modelling active behaviour Hill-type muscle elements [10] are included in the neck, arms and legs. Because of the complexity of the musculature of the spine, the active behaviour of the spine is modelled by actuators on the vertebral joints. The muscle elements have a realistic curvature that is maintained during movements. Sensors in the model measure the position of the head, elbows, hips and spine. Based on the output of these sensors, control systems determine the activation levels for the various active elements in the model (muscles and actuators) to stabilise to a defined position. The control systems in the active human model are all based on the scheme illustrated in Figure2. Below, this basic control scheme is explained, followed by the implementation and some modifications per body part (neck, elbows, hips, and spine).

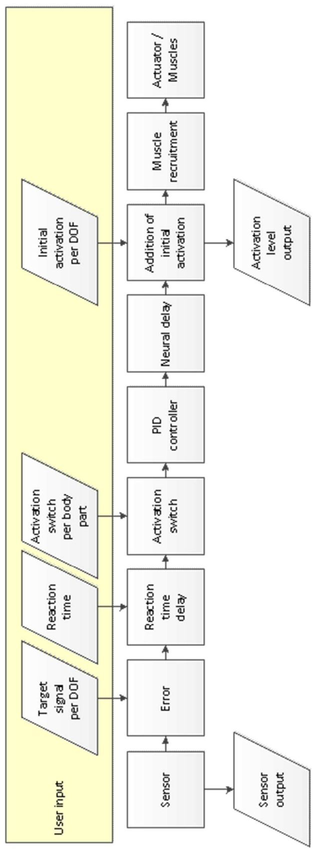

Figure2. Basic control scheme of the active human model.

For each degree of freedom that is controlled, a sensor is defined to measure the motion. Also, a target signal is defined which can be changed (to simulate e.g. voluntary or reflex movement), but which by default is 0 in order to stabilise towards the initial position. The control error is calculated as the difference between the sensor and target signals.
The next step is the reaction time. Here, the reaction time represents the time it takes for the human brain to start responding to any new event. This includes the time needed for sensing, transfer of the signal to the brain, and processing in the brain. The reaction time is implemented such that:

• Control errors related to pure stabilising behaviour, without any new events, cause a direct response;

• New events cause a response with a delay of the reaction time.

New events are automatically detected by the active human model. A new event is defined as any external load causing a control error that is larger than the maximum error occurred in the simulation up to the current time step. If the 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.
For each body part that is controlled the active behaviour can be switched 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.
Proportional-integral-derivative (PID) controllers aim to reduce the error by calculating a correcting load. 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 oscillations and reduce future errors the D-action makes the controller action larger, if the error is increasing, and smaller, if the error is decreasing.
After the PID-controller a neural delay is implemented. 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. This neural delay is defined as (Equation 1):

$$\frac{d(ouput)}{dt} = \frac{(input - output)}{\tau_{delay\_time}} \quad (1).$$

For the controllers on the neck muscles τ_{delay_time} is set to 40 ms, for the controllers on the actuators of the spine and on the muscles in the arms to 70 ms, and for the controllers on the muscles in the legs to 100 ms. The neural delay behaves frequency dependent, i.e. signals with lower frequencies are transferred better than signals with higher frequencies.
For initial equilibrium initialisation of the controllers is required. In order to achieve this, initial activation levels need to be defined from the output of a settling simulation in which the model is run with only gravity applied to find an equilibrium. The initial activation levels are added to the controller signals. The last step in the control scheme is different per body part, and explained below.
The neck controller acts in three degrees of freedom, being the three rotations of the head. Depending on the user 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. The muscle recruitment table for the neck was taken from the model of [11]. For the neck 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, also co-contraction of neck muscles can be defined. Co-contraction is 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 and/or bracing. In the active human model the co-contraction level can be defined in the initial input as a relative value (0-1) of the maximum possible muscle activation. The co-contraction level is included in the calculation of the muscle recruitment, and is balanced for any pitch angle. For the co-contraction the reaction time and neural delay can be switched off in the user input in order to avoid the co-contraction building up from zero during the first part of the simulation.
The controllers on the left and right elbow each act in only one degree of freedom per side, being the elbow flexion-extension. The muscle recruitment for the elbow divides the muscles in a group of flexors and a group of extensors and activates all muscles in one group to the same extent.
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. 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.
For the spine no target functions are defined. Hence, the rotation error for the spine is equal to the sensor output. 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 vertebra sensors are defined to measure the angle of the vertebra relative to the sacrum (pelvis). The activation signal for each vertebra is then applied to that vertebra as well as

to the vertebrae below, such that the spine is in a stable position.

Comparison to Human Subjects in Frontal Impact Tests

The active human model's response was tested using almost all the human subject test data that was used for the human and segment models the model is based on, and new tests were added. The test data comprise of blunt impact tests on various body parts as well as full-body impact tests. Frontal, lateral, rear and vertical human subject loading tests at several loading severities and conditions are included. The simulation set-ups of these tests, the human subject responses or response corridors, and the simulation results were put together in an internally developed automated test suite in which the active human model's responses are compared to that of the human subjects. The tests as well as the active human model's responses compared to the human subject responses (or response corridors) are described and shown in a MADYMO Quality Report [12]. The test data are from published post mortem human subject (PMHS) tests as well as from published volunteer tests. In order to simulate the tests in a most biofidelic way, the post mortem human subject tests were simulated with the activation of all controlled body parts switched off (passive model), and the volunteer tests were simulated with the activation of all these body parts switched on (active model). By doing so, the passive behaviour of the model is tested with the PMHS tests, and the combined active and passive behaviour with the volunteer tests.
This study focuses on the evaluation of the active human model's response in braking and frontal impact. Therefore, a number of impact tests that are essential for the evaluation of the response in braking and frontal impact were selected from the test suite for this study. The selected tests comprise of PMHS blunt frontal impact tests on the head, thorax and abdomen at several loading severities (See Table1) as well as volunteer full-body frontal impact tests at several loading severities (See Table2). The 1 g car braking test (See test no 1 in Table2) and the 3.8 g frontal impact sled test (See test no 2 in Table2) are new tests added to the test suite of the active human model.

Table1.
PMHS blunt frontal impact tests

No	Body part	Impactor shape	Impactor mass [kg]	Impact velocity [m/s]	Ref.
1	head	rigid cylinder	23.4	2.0, 5.5	[13]
2	thorax	rigid disk	23.4	3.4, 5.8	[14]
3	thorax	rigid disk	23.4	4.9, 6.9, 9.9	[15], [16], [17]
4	thorax	rigid disk	10.4, 22.2	7.0, 9.9	[18], [19]
5	abdomen	rigid bar	31.4	6.9	[20]

Table2.
Volunteer full-body frontal impact tests

No	Test set-up	Subject position	Peak deceleration [g]	Ref.
1	car braking event	Seating occupant	1 during 1.7 s	[2]
2	sled rigid seat with 3-p belt	seating occupant	3.8	[21], [22]
3	sled rigid seat with 5-p belt	sitting straight-up	15	[23], [24], [25], [26]

Evaluation for Various Frontal Crash Scenarios

In order to evaluate the robustness as well as the sensitivity of the active human model for pre-crash braking followed by a crash, several frontal crash scenarios were simulated. The car interior consists of a seat with conventional 3-point belt (with retractor and load limiter) and airbag. The applied crash pulses were obtained from Euro NCAP 64 km/h 40% offset car-barrier frontal impact tests of three different car types, i.e. a small car with short bonnet, a large car with long bonnet, and a SUV. For confidentiality reasons the car types are not mentioned, and the crash pulses are not shown here.
For evaluating the effect of the active behaviour on the kinematics and the injury values of the model, simulations with the active human model with active behaviour switched on (active model) as well as switched off (passive model) for all body parts were performed. For evaluating the effect of braking on the kinematics and injury values all three simulations with the active model were repeated with a braking phase of 8.0 m/s^2 during 1 s preceding the same crash pulse.
A quantitative evaluation of the risk of injuries was not made, since the injury limits have not yet been validated for the active human model. Thereby, the car interior was not from an existing car, and was the same for all three pulses. However, the seat and

belt parameters were in the range of existing car seats and belts.

RESULTS

Comparison to Human Subjects in Frontal Impact Tests

The force-time responses of the head of the passive model in the blunt frontal head impact tests (See test no 1 in Table1) are compared to the PMHS response corridors in Figure3 and Figure4. Figure3 shows that the peak force as well as the timing of it in the 2 m/s impact are within the peak response corridors. Figure4 shows that the peak force of the head in the 5.5 m/s impact is 4% above the upper corridor, and the peak timing is within the peak response corridors. These simulation results indicate that the passive model's head is capable of predicting realistic forces-time responses of the human head in frontal impacts.

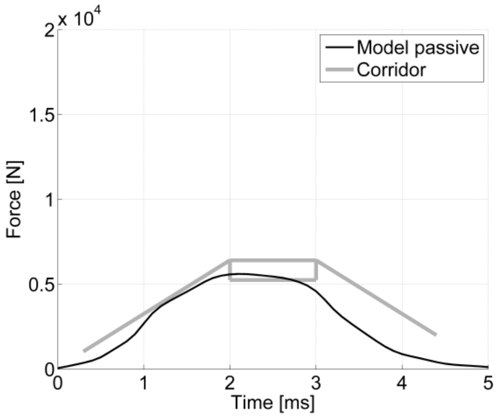

Figure3. PMHS response corridors and active human model (passive) response in blunt frontal head impact with 23.4 kg at 2 m/s.

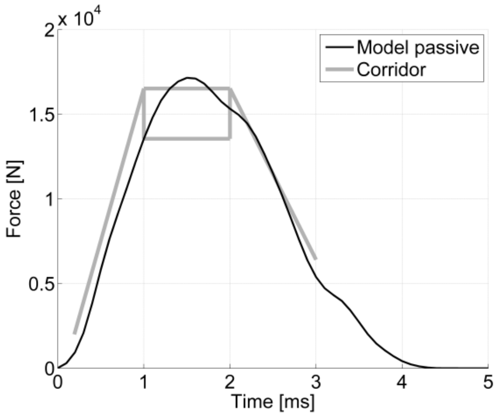

Figure4. PMHS response corridors and active human model (passive) response in blunt frontal head impact with 23.4 kg at 5.5 m/s.

The force-deflection responses of the thorax of the passive model in the blunt frontal thorax impact tests (See test no 2, 3 and 4 in Table1) are compared to the PMHS response corridors in Figure5 to Figure11. These figures show that in five of the seven thorax impact tests the peak force of the passive model is within the response corridors of the PMHS, and in the two most severe impacts (23.4 kg and 22.2 kg at 9.9 m/s) the peak forces are 4% below the peak of the lower corridor. These figures also show that in three impact tests the peak deflection is within the response corridors of the PMHS, in the three most severe impact tests (23.4 kg at 6.9 m/s and 9.9 m/s, and 22.2 kg at 9.9 m/s) the peak deflection is at most 20% above the peak of the upper corridor, and in the impact tests with the lowest mass (10.4 kg at 7 m/s) the peak deflection is 8% below the peak of the lower corridor. The force-deflection responses at the start of the impact are within the corridors for all tests except one (23.4 kg at 5.8 m/s). However, the corridor of this impact test is very narrow compared to that of similar impacts (23.4 kg at 4.6 m/s and 6.9 m/s). These simulation results indicate that the passive model's thorax is capable of predicting realistic force-deflection responses of the human thorax in frontal impacts.

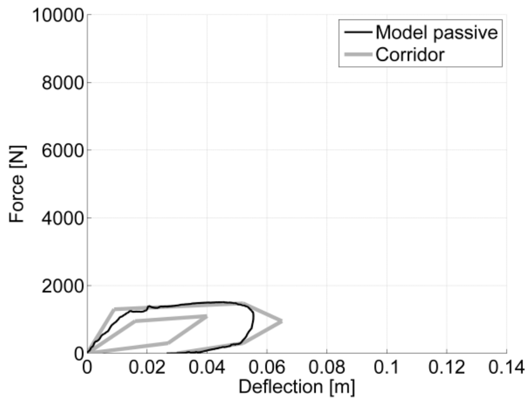

Figure5. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 23.4 kg at 3.4 m/s.

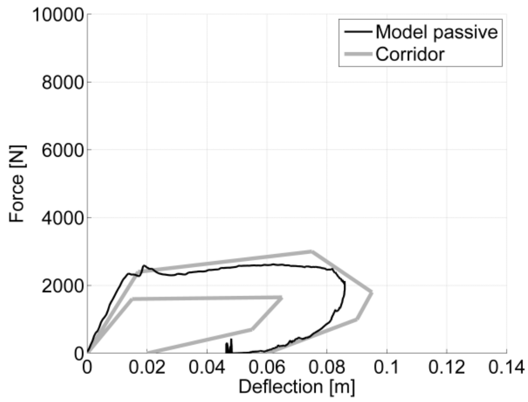

Figure6. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 23.4 kg at 5.8 m/s.

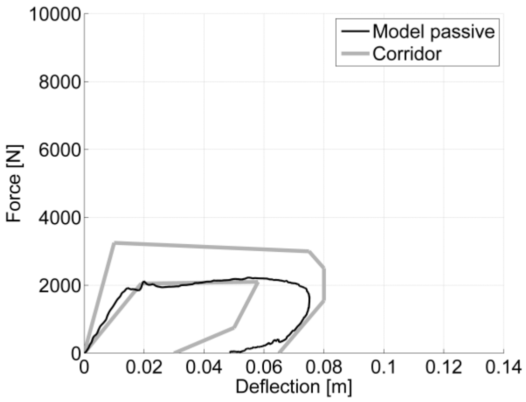

Figure7. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 23.4 kg at 4.9 m/s.

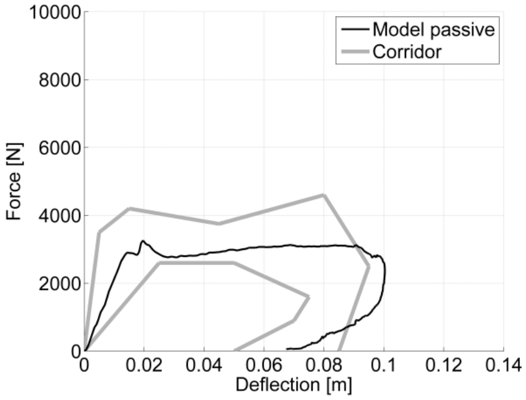

Figure8. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 23.4 kg at 6.9 m/s.

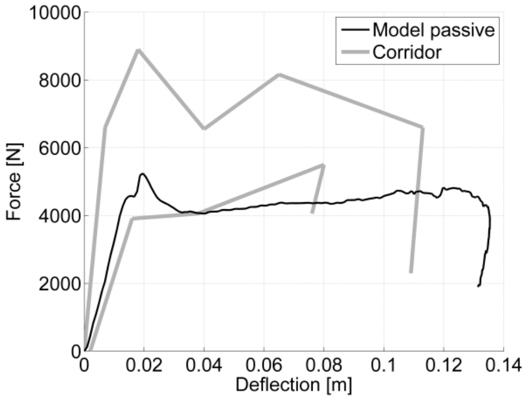

Figure9. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 23.4 kg at 9.9 m/s.

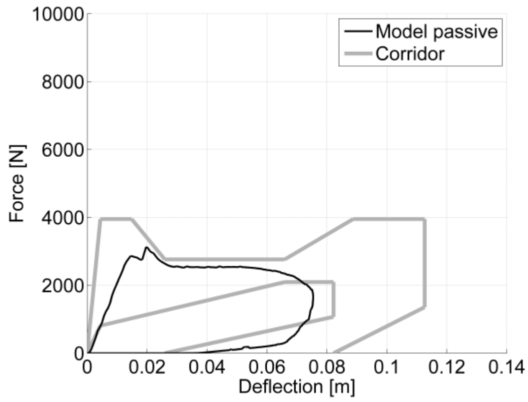

Figure10. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 10.4 kg at 7 m/s.

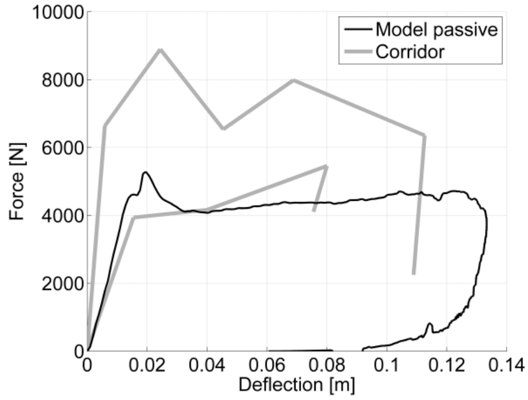

Figure11. PMHS response corridors and active human model (passive) response in blunt frontal thorax impact with 22.2 kg at 9.9 m/s.

The force-deflection response of the passive model in the blunt frontal abdomen impact test (See test no 5 in Table1) is compared to the PMHS response corridors in Figure12. This figure shows that at the start of the impact the force-deflection response is on top of the upper corridor of the PMHS responses. However, the peak force as well as the peak deflection are within the response corridors of the PMHS. This simulation result indicates that the passive model's abdomen is capable of predicting realistic force-deflection responses of the human abdomen in frontal impacts, although it is only evaluated for one impact severity.

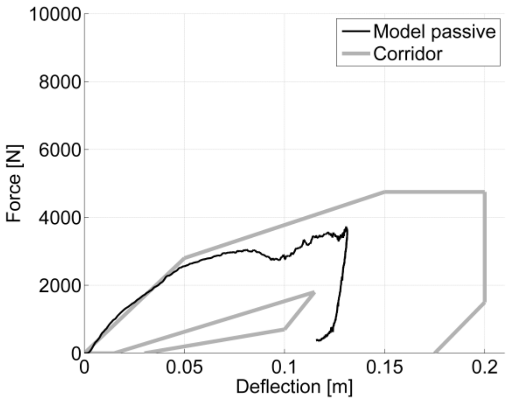

Figure12. PMHS response corridors and active human model (passive) response in blunt frontal abdomen impact 31.4 kg at 6.9 m/s.

The chest and neck horizontal displacements of the active human model in the 1 g car braking test (See test no 1 in Table2) are compared to the volunteer responses in Figure13 and Figure14. These figures were copied from [7] just to show the complete testing of the active model's responses in full-body frontal impact. The chest displacement was measured at the belt, and the neck displacement at a collar around the neck. Various reaction times (RT's) were simulated here. The co-contraction (CCR) was set to 0.5, which was an engineering judgement. However, the co-contraction of the neck muscles hardly affects the neck displacement in this simulation, since the displacement of the spine determines the displacement of the chest and neck most in this case. Figure13 shows that the peak chest displacement of the active model is equal to that of the volunteer with the smallest peak chest displacement. Figure14 shows that the peak neck displacement of the active model is approximately 20% smaller than that of the volunteer with the smallest peak neck displacement. The difference in neck displacement 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 model to be restrained by the belt a bit earlier than the volunteers, resulting in a smaller neck displacement than the volunteers. These simulation results indicate that the active model is capable of predicting realistic neck and chest displacements in car braking events.

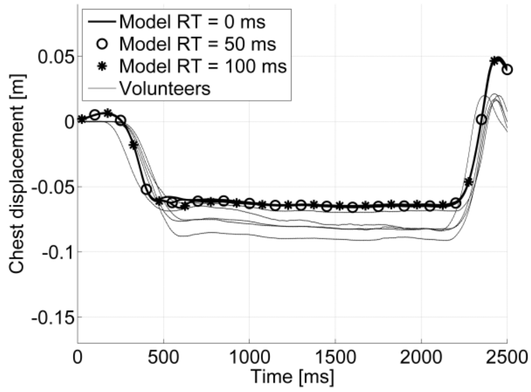

Figure13. Chest displacement of active human model at various RT's and CCR=0.5 and of volunteers in 1 g car braking.

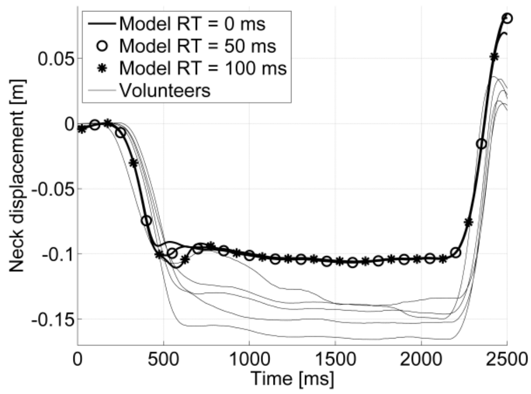

Figure14. Neck displacement of active human model at various RT's and CCR=0.5 and of volunteers in 1 g car braking.

The head, T1 and pelvis horizontal displacements of the active human model in the 3.8 g frontal impact tests (See test no 2 in Table2) are compared to the volunteer response corridors in Figure15 to Figure17. The head horizontal displacement was measured at the top of the head, that of T1 at a position close to T1. The iliac crest was used to measure the horizontal displacement of the pelvis. Various reaction times (RT's) were simulated here. The co-contraction (CCR) was set to 0.5. These figures show that the active model best predicts the volunteers kinematics at RT=25 ms. At this value the active model's head, T1, and pelvis horizontal displacement are completely within the volunteer response corridors. Thereby, Figure15 and Figure16 show that the spread of the volunteer head and T1 horizontal displacements as well as the timing of the peaks can partly be simulated by varying the reaction time. Although, Figure17 shows that the pelvis displacement shows a slightly different movement than that of the volunteers. This was caused by the pelvis sliding a few centimetres over the flesh of the upper legs, which is not modelled in the active human model. These simulation results indicate that the active model is capable of predicting realistic human full-body kinematics in 3.8 g full-body frontal impact.

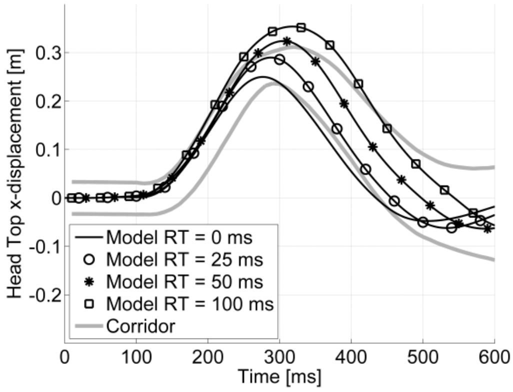

Figure15. Head top x-displacement w.r.t. sled of active human model at various RT's and CCR=0.5 and volunteer response corridors in 3.8 g frontal impact.

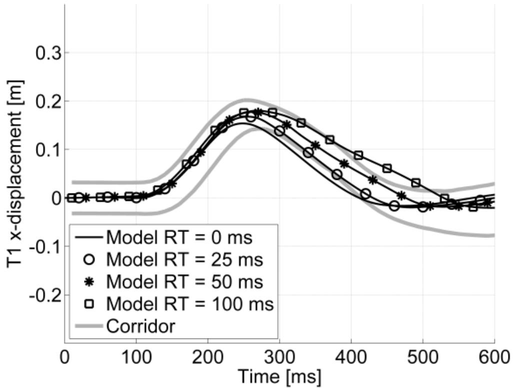

Figure16. T1 x-displacement w.r.t. sled of active human model at various RT's and CCR=0.5 and volunteer response corridors in 3.8 g frontal impact.

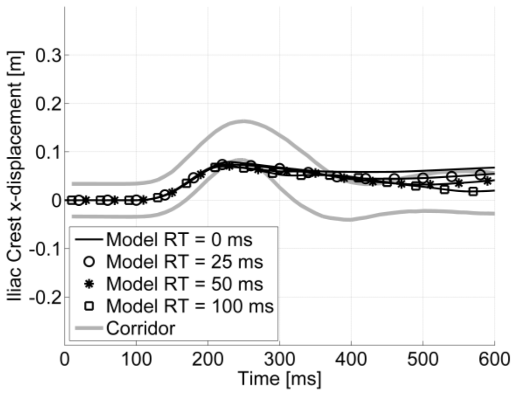

Figure17. Iliac crest x-displacement w.r.t. sled of active human model at various RT's and CCR=0.5 and volunteer response corridors in 3.8 g frontal impact.

The head rotation and head centre of gravity (COG) horizontal displacement of the active model in the 15 g frontal impact test (See test no 3 in Table2) are compared to the volunteer response corridors in Figure18 and Figure19. Various reaction times (RT's) were simulated here. The co-contraction (CCR) was set to 0.5. This test was also simulated in [7] for RT=0, 50 and 100 ms, and various levels of co-contraction. Figure18 and Figure19 show that the active model best predicts the volunteers kinematics at RT=25 ms. These figures show that at RT=25 ms the peak head rotation of the active model is approximately equal to the upper corridor, and the peak head rotation is 5% below the lower corridor. Both responses are almost within the volunteer response corridors, only the timing of the peak head rotation is 15 ms earlier than that of the volunteers. These simulation results indicate that the active model's head-neck complex is capable of predicting realistic human head-neck kinematics in 15 g full-body frontal impact.

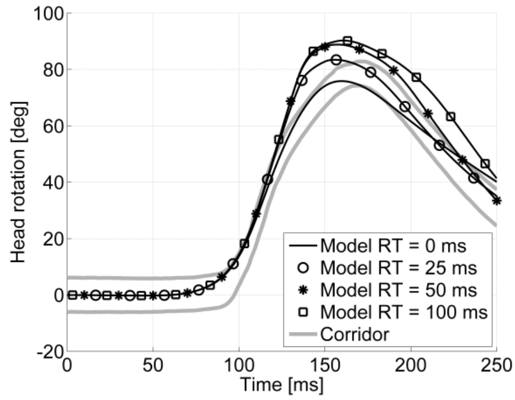

Figure18. Head rotation w.r.t. to sled of active human model at various RT's and CCR=0.5 ms and volunteer response corridors in 15 g frontal impact.

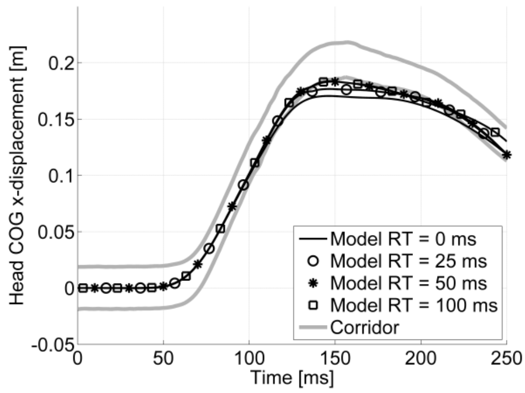

Figure19. Head COG x-displacement w.r.t. sled of active human model at various RT's and CCR=0.5 and volunteer response corridors in 15 g frontal impact.

Evaluation for Various Frontal Crash Scenarios

The position of the active model (red) and passive model (green) at onset of the frontal crash simulation for the small car is shown in Figure20, and at 60 ms (just before contact with airbag) in Figure21. The position of the active model at onset of the frontal crash simulation for the small car with preceding braking (red) and without preceding braking (blue) is shown in Figure22, and at 1000 ms (prior to start of crash pulse) in Figure23, and at 1060 ms (prior to contact with airbag) in Figure24. The simulations of the other cars showed similar differences between the position of the active model and passive model as well as between with and without preceding braking.

Figure21 shows that the head of the passive model just before contact with the airbag is slightly more upward rotated than that of the active model. Figure23 shows that braking causes the active model to move forward just before the impact. Also, Figure24 shows that due to braking before the crash the active model contacts the airbag earlier than without braking.

Figure20. Position of active (red) and passive (green) model at onset of the frontal impact simulations of the small car.

Figure21. Position of active (red) and passive (green) model at 60 ms of the frontal impact simulations of the small car.

Figure22. Position of the active model with (blue) and without (red) preceding braking at onset of the frontal impact simulations of the small car.

Figure23. Position of the active model with (blue) and without (red) preceding braking at 1000 ms of the frontal impact simulations of the small car.

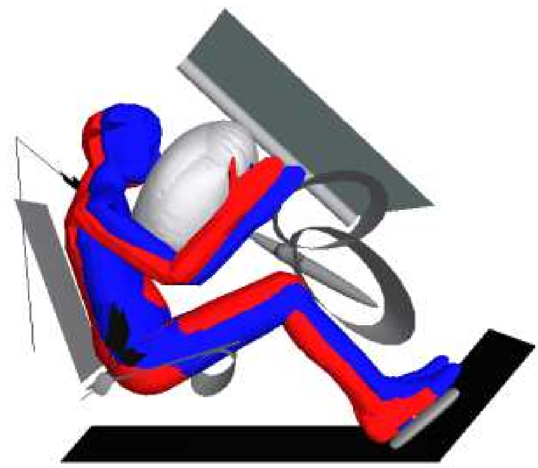

Figure24. Position of the active model with (blue) and without (red) preceding braking at 1060 ms of the frontal impact simulations of the small car.

The resulting HIC_{36ms}, peak upper chest deflection, and peak T1 forward displacement from the simulations of the frontal crash scenarios are shown in Table3.

Table3.
Simulation results of various frontal crash scenarios with active and passive model

Simulation	HIC_{36ms} [-]	Lower chest deflection [m]	T1 x-displacement [m]
Passive model small car	930	0.024	0.361
Active model small car	883	0.026	0.352
Active model small car with braking	620	0.057	0.423
Passive model large car	725	0.023	0.357
Active model large car	733	0.027	0.349
Active model large car with braking	552	0.050	0.419
Passive model SUV	525	0.023	0.338
Active model SUV	477	0.023	0.325
Active model SUV with braking	328	0.050	0.388

Comparing the results of the active model to that of the passive model this table shows that the active behaviour decreases the HIC_{36ms} as well as the peak T1 forward displacement, and increases the peak lower chest deflection for all three crash pulses. This indicates that the active human model robustly predicts effects of active behaviour in frontal crash simulations. Comparing the results of the active model with braking to that without braking this table shows that braking decreases the HIC_{36ms}, and increases the peak T1 forward displacement and peal lower chest deflection for all three crash pulses. This indicates that the active human model robustly predicts effects of braking in frontal crash simulations.
Table3 shows that the active behaviour decreases the HIC_{36ms} by 9% at most, and the braking decreases it by 31% at most. The active behaviour increases the peak lower chest deflection by 14% at most, and the braking increases it by 121% at most. Further, the active behaviour decreases the peak T1 forward displacement with 4% at most, and the braking increases it by 20% at most. So, in these simulations a different position prior to the crash due to braking affects the HIC_{36ms} as well as the chest deflection at least three times more than the active behaviour itself does. Also, the simulation results show that a more forward position prior to the crash, due to the braking, results in approximately twice as high chest deflection. However, in these simulations a shorter distance to the airbag prior to the crash results in a smaller head impact. These results indicate that the active human model is sensitive to effects of braking and active behaviour, and the effects of braking on the injury values are dominant over the effects of active behaviour.

DISCUSSION

The simulation results of the head, thorax and abdomen blunt frontal impact tests showed that the active human model is capable of predicting the PMHS peak responses within 20% deviation from the PMHS response corridors. The maximum deviation from the corridors was found for the peak thorax deflection in the two most severe thorax impact tests. However, the impact severity of these blunt impact tests may be higher than observed in current automotive crashes, due to advances in crashworthiness and restraint systems. It must be noted that the blunt impact test data were all from PMHS. So, only the model’s passive thorax and abdomen response could be evaluated. Nevertheless, for the severity of these blunt impact tests the thorax and abdomen muscle activation is assumed to have a minor effect on the response. For this reason, active behaviour of the ribcage and abdomen are not included in the model. For pre-crash kinematics the active behaviour of the thorax is included in the spine.
The simulation results of the volunteer full-body frontal impact tests showed that the active human model is capable of predicting the volunteer peak responses within 20% deviation from the volunteer response corridors. The maximum deviation from the corridors was found for the peak neck forward displacement in the 1 g car braking test. The simulation results of the two other volunteer full-body frontal impact tests showed a better fit with the volunteer response corridors. These two volunteer tests were performed in a well-defined lab environment, and therefore could be simulated more accurately. For the simulation results of these two tests the best fit with the volunteer response corridors was obtained with the reaction time set to 25 ms, while the co-contraction level of the neck muscles was assumed to be 0.5 (50% of the maximum possible muscle activation).
The simulation results of the various frontal crash scenarios showed that the active human model is robust and sensitive to effects of braking and active behaviour. Also, these simulation results showed that the effects of braking on the injury values are dominant over the effects of the active behaviour itself. However, the active behaviour is indispensable for correct simulation of the pre-crash kinematics.

It must be noted that the active human model has not yet been validated for injury prediction. So, a quantitative evaluation of the effect of active safety systems on the risk of injuries is not possible yet. However, the current model is capable of predicting realistic human head, thorax and abdomen blunt impact responses at several loading severities and conditions, as well as realistic human kinematics in 1 g to 15 g full-body impact. As such, the current active human model can be used for evaluating the effectiveness of active safety systems in frontal impacts.

CONCLUSION

From this study it was concluded that the current active human model is capable of simulating realistic human full-body kinematics as well as realistic injury values for the head and the thorax in one single simulation of braking followed by a frontal crash. As such, this model can be used as a tool in the development process of an active safety system for evaluating its effectiveness in frontal impacts.

REFERENCES

[1] Bose, D., Crandall, J.R., Untaroiu, C.D., Maslen, E.H. 2010. “Influence of Pre-Collision Occupant Parameters on Injury Outcome in a Frontal Collision.” Accident Analysis and Prevention, 42, 1398–1407

[2] Woldrich, M. 2010. “Movement in the PRE-Crash Situation - A Simulation Research with a Reversible PRE-SAFE® Pulse System.” Abstract for the poster session (P24) Airbag 2010, 10th International Symposium and Exhibition on sophisticated Car Occupant Safety Systems

[3] Mages, M., Seyffert, M., Class, U. 2011. “Analysis of the Pre-Crash Benefit of Reversible Belt Pre-Pretensioning in Different Accident Scenarios.” Proceedings of the 22th ESV-Conference, Paper Number 11-0442, Washington DC-US

[4] Beeman, S.M., Kemper, A.R., Madigan, M.L., Duma, S.M. 2011. “Effects of Bracing on Human Kinematics in Low-Speed Frontal Sled Tests.” Annals of Biomedical Engineering, Vol 39, No 12, 2998-3011

[5] Ejima, S., Ono, K., Holcombe, S., Kaneoka, K., Fukushima, M. 2007. “A Study on Occupant Kinematics Behaviour and Muscle Activities During Pre-Impact Braking Based on Volunteer Tests.” Proceedings of the International Conference on Biomechanics of Impact IRCOBI

[6] Carlsson, S., Davidsson, J. 2011. “Volunteer Occupant Kinematics During Driver Initiated and Autonomous Braking when Driving in Real Traffic Environments.”, Proceedings of the International Conference on Biomechanics of Impact IRCOBI, Krakow-Poland

[7] Meijer, R., Hassel, E. van, Broos, J., Elrofai, H., Rooij, L., Hooijdonk, P. 2012. “Development of a Multi-Body Human Model that Predicts Active and Passive Human Behaviour.”, Proceedings of the International Conference on Biomechanics of Impact IRCOBI, Dublin-Ireland

[8] MADYMO Human Body Models Manual, 2012. Version 7.4.1., TASS B.V., Rijswijk-The Netherlands

[9] MADYMO Theory Manual, 2012. Version 7.4.1, TASS B.V., Rijswijk-The Netherlands

[10] Winters, J.M. and Stark. L. 1988. “Estimated Mechanical Properties of Synergistic Muscles Involved in Movements of a Variety of Human Joints.” Journal of Biomechanics, 21, 1027-1041

[11] Nemirovsky, N., Rooij, L. van, 2010. “A New Methodology for Biofidelic Head-Neck Postural Control.” Proceedings of the International Conference on Biomechanics of Impact IRCOBI, Hannover-Germany, 71-84

[12] MADYMO Quality Report Release Update – Active Human 50th Facet Q Model. 2013. Version 1.1 (R7.4.2), Report Number QAHM50FC-130211, TASS B.V., Rijswijk-The Netherlands

[13] Don, B. van, Ratingen, M. van, Bermond, F., Masson, C., Vezin, P., et al. 2003. “Biofidelity Impact Response Requirements for an Advanced Mid-Sized Male Crash Test Dummy.” Proceedings of the 18th International Technical Conference on Enhanced Safety of Vehicles, Nagoya-Japan

[14] Bouquet, R., Ramet, M., Bermond, F., Cesari, D. 1994. “Thoracic and Pelvis Human Response to Impact.” Proceedings of the 14th International Technical Conference On Enhanced Safety of Vehicles, Munich-Germany, 100-109

[15] Kroell, C.K., Schneider, D.C., Nahum, M. 1971. “Impact Tolerance and Response of the Human Thorax.” SAE Paper 710851, Proceedings of the 15th Stapp Car Crash Conference, 163-213

[16] Kroell, C.K., Schneider, D.C., Nahum M. 1974. “Impact Tolerance and Response of the Human Thorax II.” SAE Paper 741187, Proceedings of the 18th Stapp Car Crash Conference, 215-253

[17] Kroell, C.K. 1976. “Thoracic Response to Blunt Frontal Loading.” SAE Paper 67, 49-77

[18] Nahum, A., Gadd, C., Schneider, D., Kroell, C. 1970. “Deflection of the Human Thorax Under Sternal Impact.” SAE Paper 700400, Proceedings of the International Automobile Safety Conference, 797-807

[19] Nahum, A., Schneider, D., Kroell, C. 1975. "Cadaver Skeletal Response to Blunt Thoracic Impact." SAE Paper 751150, Proceedings of the 19th Stapp Car Crash Conference, 259-293

[20] Cavanaugh, J.M., Nyquist, G.W., Goldberg, S.J., King, A.I. 1986. "Lower Abdominal Tolerance and Response." Society of Automotive Engineers, SAE Paper 861878, 41-63

[21] Balasubramanian, S., Seacrist, T., Hopely, T., Sterner, R., Maltese, R.M., Constans, E., Arbogast, B.K. 2009. "Head and Spinal Trajectories in Children and Adults Exposed to Low Speed Frontal Acceleration." Proceedings of the 21st Enhanced Safety of Vehicles Conference, Stuttgart-Germany

[22] Arbogast, K., Balasubramanian, S., Seacrist, T., Maltese, M.R, García-España, J.F., et.al. 2009. "Comparison of Kinematic Responses of the Head and Spine for Children and Adults in Low-Apeed Frontal Sled Tests." Proceedings of the 53rd Stapp Car Crash Conference, Savannah, Georgia-USA

[23] Ewing, C., Thomas, D., Lustick, L., Muzzy, W., Willem, G. et al. 1976. "The Effect of Duration, Rate of Onset, and Peak Sled Acceleration on the Dynamic Response of the Human Head and Neck." Proceedings of the 20th Stapp Car Crash Conference, Dearborn-Michigan, 3-41, Society of Automotive Engineers, SAE Paper 760800

[24] Wismans, J. and Spenny, C. 1984. "Head-Neck Response in Frontal Flexion." SAE Paper 841666, Proceedings of the 28th Stapp Car Crash Conference, Chicago-USA, 161-171

[25] Wismans, J., Oorschot, H. van, Woltring, H. 1986. "Omni-Directional Human Head-Neck Response." SAE Paper 861893, Proceedings of the 30th Stapp Car Crash Conference, San Diego CA-USA, 313-331

[26] Thunnissen, J., Wismans, J., Ewing, C., Thomas, D. 1995. "Human Volunteer Head-Neck Response in Frontal Flexion: A New Analysis." SAE Paper 952721, Proceedings of the 39th Stapp Car Crash Conference, Coronado CA-USA, 439-460