EFFECT OF VARIOUS PRE-CRASH BRAKING STRATEGIES ON SIMULATED HUMAN KINEMATIC RESPONSE WITH VARYING LEVELS OF DRIVER ATTENTION

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Paper Number 11-0306

ABSTRACT

In this study, human kinematic response resulting from various pre-crash braking scenarios is quantified. The underlying question is what effect pre-crash braking systems have on the driver or the front seat passenger.

The vehicle deceleration pulses resulting from various pre-crash braking strategies are implemented on a vehicle interior model in a multi-body software code. The two most important strategies are based on 1) a brake assist system with modulated braking (BAS+) and 2) an autonomous braking system (AUT). In addition, simplified braking scenarios at various deceleration levels (3, 6 and 9.5 m/s^2) are simulated. The driver is represented by a numerical human model incorporating, besides all passive stiffness and damping properties, algorithms that simulate active stabilising behaviour in case of an induced acceleration on the body. The lumbar and thoracic spine are stabilised by torque actuators, while the cervical spine is stabilised by Hill-type muscle segments. The level of control, bracing and reaction time delays can be varied. This allows for the simulation of various attention schemes. A parameter study is performed, in which sensitivity of the kinematic response to vehicle braking strategies and to various human reaction types are discussed and compared to findings in literature.

This study provides insight in human kinematic motion in the vehicle under various braking scenarios and human attention levels. The methods currently lack specific validation for frontal pre-crash braking, due to the lack of available volunteer testing data. Also, due to the complexity of human behaviour and the current state-of-the-art regarding its characterisation or modelling, the models are empirical of nature, however provide practical guidance to the range of possible pre-crash kinematics as a result of varying human behavioural strategies. Conclusions from this research are that driver attention plays an important role in determining the effectiveness of pre-crash braking systems in preventing severe occupant motions and in positioning the occupant in an optimum position at time of impact.

INTRODUCTION

Pre-crash braking systems

Pre-crash braking systems are being employed in vehicles on the market currently, and the performance of them is being improved with the availability of more accurate sensor technology and risk estimation algorithms. Even though these pre-crash braking systems have limited penetration in the vehicle fleet, even in developed countries, numerous studies showing their efficacy have been performed. For example, Kuehn *et al.* (2009) showed that a collision mitigating braking system (of level 2) can avoid 12.1% of all crashes if all cars would be equipped with such a system. A level 2 system is defined as a system that, based on forward environmental detection and estimation of case vehicle speed, provides a warning at Time To Collision (TTC) of 2.6 s, performs automatic partial braking of 0.4-0.6 G if the driver has not braked at TTC 1.6 s or a applies modulated braking to avoid the crash if the driver has applied the brakes at TTC 1.6 s. Schittenhelm *et al.* (2009) assessed the effectiveness of various stages of pre-crash braking systems based on comparing registered crashes with numbers of sold cars with or without such a system. The presence of Brake Assist systems resulted in 8% less rear-end collisions to occur and 13% less serious impacts against pedestrians. More advanced systems, with warning, modulated braking when the driver reacts and partial autonomous braking, similar to as defined by Kuehn as level 2, showed to be able to avoid a collision with a vehicle in front in 20% of all cases and to reduce the severity in 25% of all cases. As such, pre-crash braking systems are entering the market that act differently when the driver does or does not apply the brakes, i.e. detects the oncoming crash. In this light, Ore *et al.* (1992) indicated that roughly half of all vehicle occupants apply the brakes prior to a frontal collision.

Woldrich *et al.* (2010) presented a pre-crash braking system that attempts to position the occupant in an optimum position at the time an apparently inevitable crash occurs. Moreover, the system attempts to provide the occupant with as much as safely possible rearward velocity, in order to mitigate the consequences of a possible oncoming crash. This safety system functions

during the pre-crash braking phase by means of seat belt actuation and as such highly depends on accurate prediction of occupant kinematics in the pre-crash braking phase.

Human kinematic behaviour characterisation

As discussed in the previous section, the characterisation and quantification of human kinematic behaviour in the phase prior to the crash is of importance for optimal restraint performance and as such for mitigating fatalities and injuries in the case a crash occurs. Based on volunteer tests, Begeman *et al.* (1980) identified muscle activation reaction times of more than 200 ms when exposed to frontal impact acceleration. Choi *et al.* (2005) performed volunteer tests to assess the change in driver posture as a result of bracing for an impact that was detected by the driver. In addition, muscle activation levels were computed from EMG measurements as well as forces applied on the vehicle structures. Occupant motion as a result of acceleration was not quantified. Ejima *et al.* (2009) performed a series of tests with volunteer seated on rigid seats, restrained by a three-point belt system and subjected to a 600 ms 0.8 G constant deceleration. For a tensed volunteer, kinematic figures indicate that head forward displacement was in the order of 100 mm at 200 ms after impact, while T1 forward displacement was in the order of 25 mm and hip forward displacement around 10 mm. For a relaxed occupant restrained by a lap belt only, the head displacement was in the order of 600 mm at 600 ms after impact with T1 displacement around 400 mm. In an earlier study with tensed volunteers on simple seats and an approximate 200 ms duration 1.0 G pulse Ejima *et al.* (2007) identified that the sternocleidomastoid muscles in the neck were activated around 100-200 ms after impact at the time when the torso was moving forward more than the head, i.e. the head moved rearward with respect to T1. In a later phase, when the head/neck goes into flexion the paravertebral muscles (i.e. longus colli and longus capitis) were activated. In addition, the latissimus dorsi muslces in the torso were activated. Behr *et al.* (2010) focussed on lower extremity kinematic and muscle activation behaviour during emergency braking and established reaction times for first movement of the foot after the emergency situation was visually detected of 0.285 s (0.042 SD). Muscle activation levels were up to 57% of the maximum possible activation level for muscles in the lower extremity.

Numerical human modelling

From Crandall (2008) it can be stated that due to the breadth of variations in which collision-induced injuries occur, in order to achieve goals set in further injury and fatality reduction, numerical simulation methods allow for vehicle (restraint) design for optimising towards real-world protection, as opposed to protection in a specific scenarios. In doing so, a concise review was presented on the state the art in numerical human modelling for injury reduction. Bose *et al.* (2008) used a numerical human model (de Lange *et al.*, 2005 & Cappon *et al.*, 1999) to study the effect of pre-impact posture, as well as levels of muscle bracing in the lower extremities and body mass and stature, on the injury risk in the event a crash was unavoidable. Pre-impact posture was shown to be the parameter affecting the injury risk the most. In an optimisation routine it was found that with a seat belt system with adaptive force limiting settings and variable pretensioner firing time, a reduction of injury risk of up to 35% could be achieved. While this study showed the necessity for the prediction of occupant kinematics, the human model used could not predict this in the pre-crash phase.

In order to develop human models that predict occupant kinematics during emergency braking manoeuvres, the active muscle response behaviour of occupants needs to be simulated. While numerous human models have been developed that simulate muscle behaviour at various levels of detail, limited models are able to predict human reactive response to an external stimuli, such as vehicle braking. Most models merely prescribe muscle activation dynamics based on electromyography (EMG) measurements in similar test environments. The first known approach to predicting human reactive response was developed by Cappon *et al.* (2007). A passive human model, validated for the crash scenarios (de Lange *et al.*, 2005) was extended with torque actuators acting on each spinal vertebrae, being controlled by a set of PID-controller, thus stabilising the spine resulting in human-like kinematics. Obviously, body internal loads as well as the stabilising algorithm were not human-like. In order to overcome this deficit, Fraga *et al.* (2009) applied similar PID controllers, however acting on Hill-type line element muscles present in a multi-body neck model. This controller approach was given a higher degree of biofidelity by developing a control algorithm that allowed for a definition of muscle recruitment strategies, provision of a level of co-contraction and uncoupled control in three main degrees of freedom of the neck, i.e. head roll, pitch and yaw motion (Nemirovsky *et al.*, 2010). Similar approaches are taken currently by Östh *et al.* (2011) and Prüggler *et al.* (2010).

Objectives

The objective of this study is to predict human kinematic response resulting from various pre-crash braking scenarios, based on simulations with numerical human models that are developed to be suitable for such simulations and to study sensitivity to driver attention schemes.

METHODS

Simulation setup

For this study a human model was developed in MADYMO that was a combination of two models:

- The human model with stabilising spine (Cappon *et al.* 2007) was used for actuation of lumbar and thoracic spine
- The human neck model with Hill-type line-element muscle control (Nemirovsky *et al.*, 2010) was coupled to the above human model

This combined model is shown in figure 1.

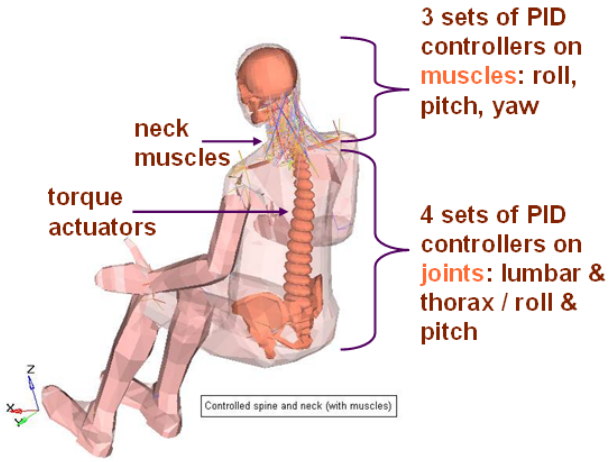

Figure 1. The MADYMO active human model with stabilising spine and neck model with line-element muscle control.

This model was positioned in a simplified vehicle interior model, in order to focus on occupant behaviour as opposed to vehicle model parameters. The human model was positioned on a rigid seat with flat surfaces at angles similar to an automotive seat. A rigid foot well surface was introduced, as well as a steering column with steering wheel. A three-point belt system with standard belt stiffness and retractor properties was fitted around the occupant. The occupant's hands were constrained to the steering wheel with a maximum force of 400 N per hand, simulating grip as based on Bao (2000). The model setup is shown in figure 2.

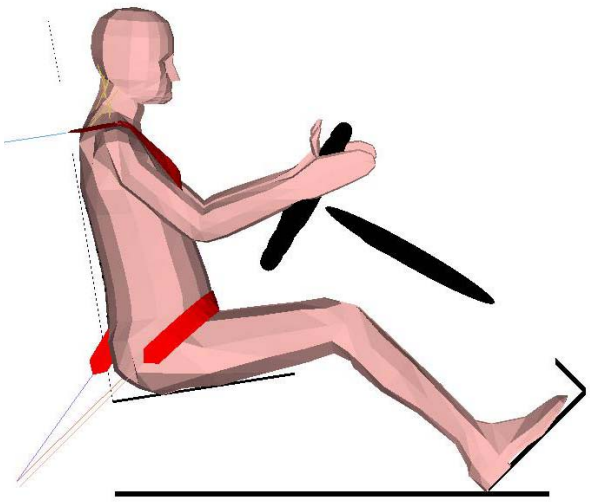

Figure 2. The MADYMO active human model in a simplified vehicle environment.

A uni-axial linear acceleration, without vehicle pitch motion, was implemented on the occupant environment in order to simulate vehicle braking. First, a set of three idealised vehicle braking pulses assuming constant deceleration from 50 km/h to 0 km/h were implemented, as shown in figure 3.

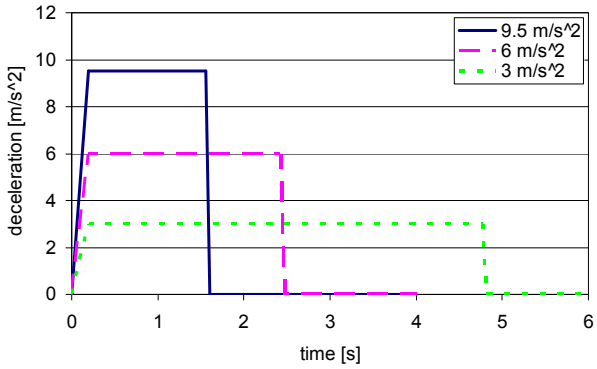

Figure 3. Idealised vehicle braking pulses at various deceleration levels, decelerating the vehicle from 50 km/h to 0 km/h.

Secondly, two pulses approximating possible responses from two types of pre-crash braking systems were implemented, as figure 4 shows:

- BAS+ is the deceleration profile arising from a Brake Assist system in which the driver applied the brake while the system applies the amount of modulated braking necessary to prevent a collision with an object in front.
- AUT is the deceleration profile from an autonomous braking system that first applies partial braking at 4 m/s^2, then full braking to assure collision avoidance.

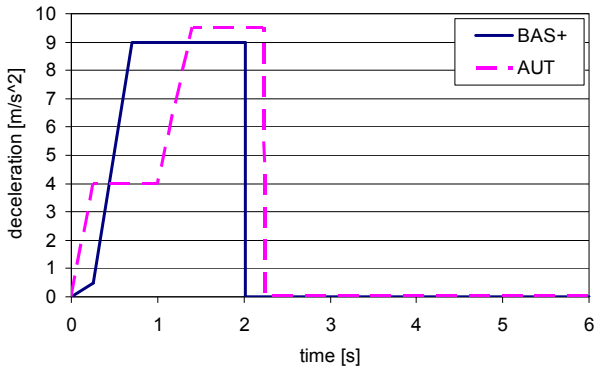

Figure 4. Pre-crash braking system pulses for Brake Assist with modulated braking (BAS+) and an autonomous braking system (AUT), both decelerating the vehicle from 50 km/h to 0 km/h.

In the human model various activation strategies or attention levels are simulated by varying settings of the controller. In table 1, three strategies are defined:

- Validated: represents the controller settings for which the human model was validated, as presented in Cappon *et al.* (2007) and Nemirovsky *et al.*, (2010). The PID controller settings for the various body parts determined through optimisation towards an experimental dataset. For the neck, frequency response perturbation tests (Keshner *et al.*, 2003) and 3.6 G rear impact tests performed by JARI (Ono *et al.*, 1999) served as validation dataset. For lumbar and thoracic spine it was based on Muggenthaler (2005)
- Attentive: originally, the controller settings for the validated strategy were believed to simulate an attentive driver due to the relatively high G level anticipated by the volunteers in the JARI laboratory tests. However, it was found that a constant level of co-contraction as high as 80% resulted in the neck locking up in a different position than the reference position due to the braking input. In order to overcome this, a co-contraction algorithm needs to be implemented that is variable for a change in head/neck pitch orientation. As such, the attentive scheme incorporated reduced co-contraction at 40% and tenfold increased PID settings in the neck. In addition, response time delay was reduced to 0 ms, since the driver is fully aware of the oncoming impact.
- The inattentive scheme presumes a person is not paying attention to the road or is even asleep. As such, the PID settings are reduced with respect to the attentive strategy. In addition, a response time of 500 ms is introduced, as well as a 10% level of co-contraction, barely able to hold the neck upright.

Table 1.
Muscle activation strategies employed with varying control, delay and co-contraction (CC) settings

strategy	control	delay	CC
validated	*neck:* P: 0.4 I: 1.25 D: 0.07 *thorax:* P: 12 I: 1.1 D: 3.3 *lumbar:* P: 18 I: 1.5 D: 3.2	excitation: 30 ms activation: 10 ms response: 100 ms	80%
attentive	*neck:* P: **4.0** I: **12.5** D: **0.7** *thorax:* P: 12 I: 1.1 D: 3.3 *lumbar:* P: 18 I: 1.5 D: 3.2	excitation: 30 ms activation: 10 ms response: **0 ms**	**40%**
inattentive	*neck:* P: 0.4 I: 1.25 D: 0.07 *thorax:* P: **0.5** I: **0.33** D: **0.33** *lumbar:* P: **5** I: **1.5E15** D: **0.32**	excitation: 30 ms activation: 10 ms response: **500 ms**	**10%**

The muscle recruitment strategy employed for this model was as commonly found in literature (Dul, 1984):

Minimise $$\sum_j \left(\frac{F_j}{F_{MAX\,j}}\right)^p \qquad (1)$$

Dul (1984) also proposed a value for p=3 to represent a minimum fatigue criterion. As such, this was adopted for this model. In order to minimise this sum, all muscles will contribute while the muscles that have the largest contribution in terms of moment in the desired direction will contribute more. The contribution of each muscle to any of the three desired head rotations (roll, pitch, yaw) is shown in table 2. Also, a division is

made for every muscle whether it contributes to head flexion or extension in the pitch direction. As such, in this model the longus colli is the strongest flexor, while the semispinalis cervicis is the strongest extensor. However, all other muscles are recruited as well only to a lesser degree (or power *p*).

Table 2.
Percentage contribution of each neck muscle to desired head rotation in roll, pitch and yaw direction for the MADYMO active human model.

Muscle	Total Roll	Total Yaw	Total Pitch	Type of Pitch
Hyoids	38.9%	10.0%	51.1%	Flexor
Levator Scapulae	58.9%	7.8%	33.3%	Extensor
Longissimus Capitis	56.8%	13.8%	29.3%	Extensor
Longissimus Cervicis	74.3%	4.4%	21.3%	Extensor
Longus Capitis	38.3%	13.9%	47.7%	Flexor
Longus Colli	29.0%	4.8%	66.3%	Flexor
Multifidus Cervicis	8.9%	39.5%	51.6%	Extensor
Scalenus Anterior	67.3%	18.8%	13.9%	Flexor
Scalenus Medius	83.6%	12.2%	4.2%	Flexor
Scalenus Posterior	88.0%	8.0%	4.0%	Extensor
Semispinalis Capitis	29.2%	22.9%	47.9%	Extensor
Semispinalis Cervicis	3.0%	29.3%	67.8%	Extensor
Splenius Capitis	30.8%	16.3%	52.9%	Extensor
Splenius Cervicis	40.1%	18.7%	41.1%	Extensor
Sternocleidomastoideus	49.2%	28.4%	22.4%	Extensor
Trapezius	19.3%	43.8%	37.0%	Extensor

Simulations are performed with the five braking pulses and the three muscle recruitment strategies, resulting in in total 11 simulations.

RESULTS

Attentive driver with 9.5 m/s^2 braking

The occupant kinematics of the attentive driver in 9.5 m/s^2 braking serves as a base case. The kinematic images at various phases during the braking event are shown in figures 5 to 9. At 0.2 s head and torso have moved forward. Neck flexion starts to occur after that resulting in maximum neck flexion and head forward displacement at 0.82 s. This head position slowly returns to neutral, however once the deceleration is removed, the body rebounds into the seat back, resulting in neck extension around 2.35 s.

This kinematic behaviour is a result of deceleration imposed on the occupant, the passive properties of the human model and the muscle activation time history as determined by the controller. In figure 10 the muscle activation time histories are shown for all the muscles that result in head/neck flexion. In figure 11, the same for all extensors. The flexors are all activated by 0.4 (i.e. 40% of maximum activation as given by the Hill-muscle model) due to the 40% co-contraction setting. The extensors are activated to a smaller degree (around 15%) as dictated by the co-contraction algorithm contracting all muscles without head/neck motion to occur.

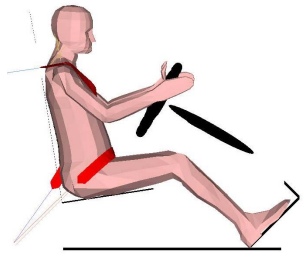

Figure 5. Attentive Active Human Model in 9.5 m/s^2 braking at t=0 s

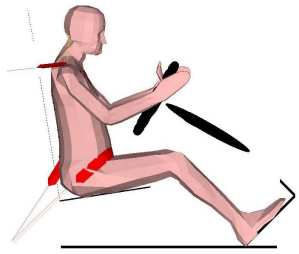

Figure 6. Attentive Active Human Model in 9.5 m/s^2 braking at t=0.2 s

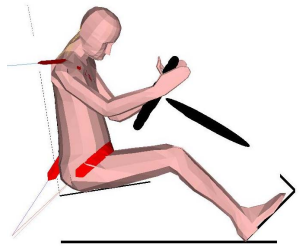

Figure 7. Attentive Active Human Model in 9.5 m/s^2 braking at t=0.4 s

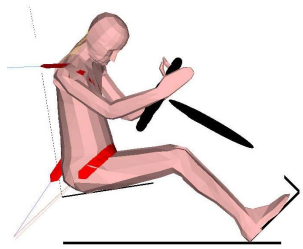

Figure 8. Attentive Active Human Model in 9.5 m/s^2 braking at t=0.8 s

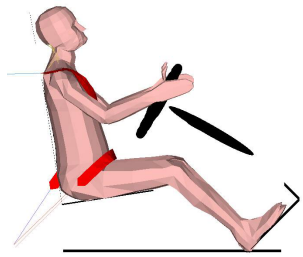

Figure 9. Attentive Active Human Model in 9.5 m/s^2 braking at t=2.35 s

Around 0.25 s after braking started the extensors start to activate more, attempting to overcome neck flexion observed in figure 7 and 8. The maximum activation level for the extensors is around 45% of the maximum. After 2.25 s the head is in rebound extension due to which the flexors start to activate. These figures also indicate that many muscles are activated however at different activation levels as given by the chosen muscle recruitment strategy. The flexor muscle with largest degree of activation is the longus colli, as dictated by table 2.

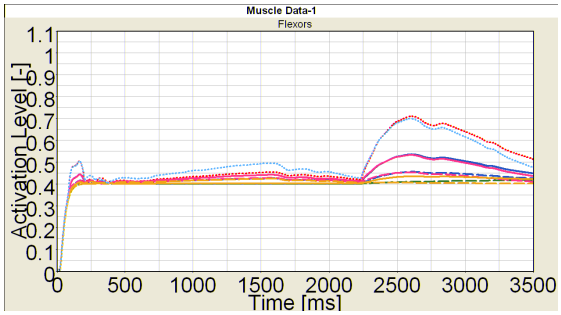

Figure 10. Flexor muscle activation signals for attentive Active Human Model in 9.5 m/s^2 braking.

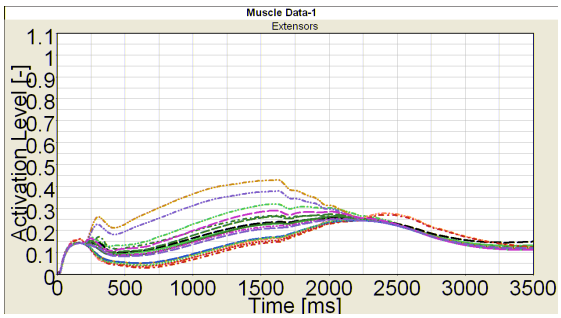

Figure 11. Extensor muscle activation signals for attentive Active Human Model in 9.5 m/s^2 braking.

Inattentive driver with 9.5 m/s^2 braking

The occupant kinematics of the inattentive driver in 9.5 m/s^2 braking is presented in comparison. The kinematic images at various phases during the braking event are shown in figures 12 to 15. At 0.2 s head and torso have moved forward slightly more than in the attentive scenario. Neck flexion starts to occur after that resulting in maximum neck flexion and head forward displacement at 0.78 s. The flexion is larger than in the attentive case, even resulting in the chin to contact the chest. This head position persists until the deceleration is removed and the body rebounds into the seat back, resulting in neck extension around 2.35 s.

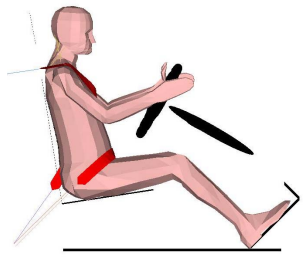

Figure 12. Inattentive Active Human Model in 9.5 m/s^2 braking at t=0 s

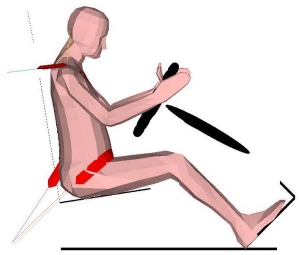

Figure 13. Inattentive Active Human Model in 9.5 m/s^2 braking at t=0.2 s

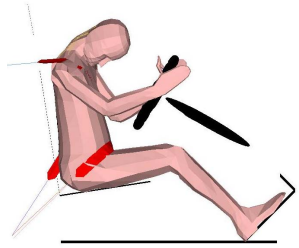

Figure 13. Inattentive Active Human Model in 9.5 m/s^2 braking at t=0.4 s

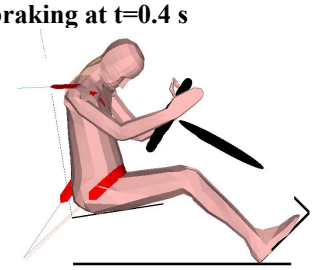

Figure 14. Inattentive Active Human Model in 9.5 m/s^2 braking at t=0.8 s

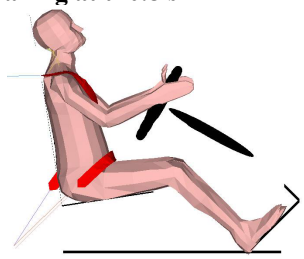

Figure 15. Inattentive Active Human Model in 9.5 m/s^2 braking at t=2.35 s

This kinematic behaviour is a result of deceleration imposed on the occupant, the passive properties of the human model and the muscle activation time history as determined by the controller. In figure 16 the muscle activation time histories are shown for all the muscles that result in head/neck flexion. In figure 17, the same for all extensors. The flexors are all activated by 0.1 (i.e. 10% of maximum activation as given by the Hill-muscle model) due to the 10% co-contraction setting. The extensors are activated to a smaller degree (around 3%) as dictated by the co-contraction algorithm contracting all muscles without head/neck motion to occur.

After 0.5 s, which was defined as the response time delay, the controller activates both flexors and extensors in an attempt to stabilise the neck. However, this approach is unsuccessful in counteracting the inertial load on the head as a result of braking. Only after braking has stopped and the head rebounds into extension do the flexors act to bring the head in a more neutral position.

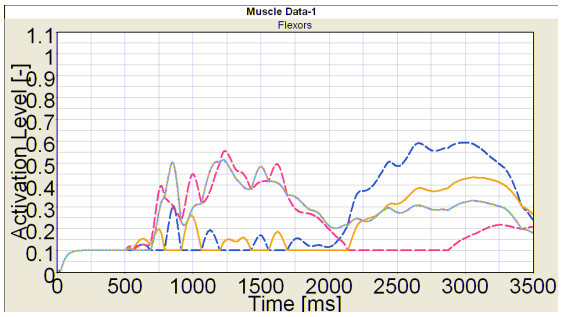

Figure 16. Flexor muscle activation signals for inattentive Active Human Model in 9.5 m/s^2 braking.

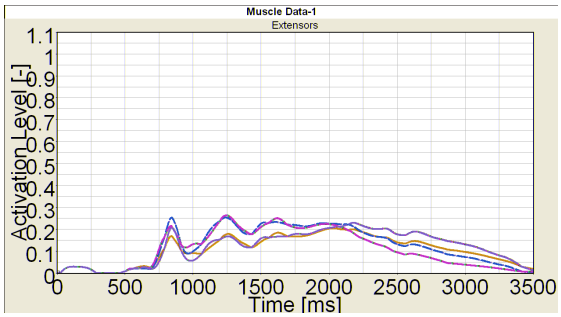

Figure 17. Extensor muscle activation signals for inattentive Active Human Model in 9.5 m/s^2 braking.

Parameter study

Results from the parameter sensitivity study are condensed in table 3 where maximum forward displacement of human model head centre of gravity, T1 and pelvis are shown together with the time at which this occurs. The simulation with the validated muscle recruitment strategy is shaded, as it was found not valid for braking simulations due to the lock up of the neck in forward flexed position. For all simulations pelvis forward displacement was negligible, most likely caused by the simulated environment with optimal seat belt and rigid seat and foot well.

Table 3.
Human model excursion in various braking scenarios with various muscle activation strategies

Braking accel.	Human activation	Head X[mm] *t [s]*	T1 X[mm] *t [s]*	Pelvis X[mm] *t [s]*
9.5 m/s^2	validated	181 *1.22*	87 *1.56*	12 *1.56*
9.5 m/s^2	attentive	157 *0.82*	91 *1.57*	10 *1.56*
6 m/s^2	attentive	129 *0.87*	68 *1.69*	1 *0.28*
3 m/s^2	attentive	93 *0.93*	49 *1.29*	0 *0.0*
BAS+	attentive	163 *1.37*	100 *2.02*	10 *2.01*
AUT	attentive	147 *1.50*	80 *1.49*	4 *1.51*
9.5 m/s^2	inattentive	180 *0.78*	90 *1.56*	11 *1.56*
6 m/s^2	inattentive	158 *0.45*	69 *1.71*	2 *0.28*
3 m/s^2	inattentive	146 *0.49*	53 *0.99*	0 *0.0*
BAS+	inattentive	291 *2.03*	207 *2.02*	6 *1.88*
AUT	inattentive	149 *0.49*	83 *1.48*	3 *1.49*

The 10 simulations with 2 muscle recruitment strategies (attentive, inattentive) and the 5 braking pulses are discussed based on figure 18 and 19 in which the results from table 3 are plotted. For the attentive scenario and simple (3, 6 or 9.5 m/s^2) braking the maximum head forward displacement occurs at around 0.8 to 0.9 s with varying levels of forward displacement: 157 mm for 9.5 m/s^2, 129 for 6 m/s^2 and 93 mm for 3 m/s^2. For the inattentive scenario and simple braking higher head forward displacements are seen for all braking severities. In addition, timing of maximum head displacement is lower for lower braking severity.

The BAS+ system with an attentive occupant displays similar levels of head forward displacement as a 9.5 m/s^2 pulse however with 0.5 s delay. When referred back to figure 3 and 4, the BAS+ pulse is similar to the 9.5 m/s^2 pulse with a delayed start. As such, this explains the similarity.

The AUT system with inattentive occupant shows good performance since the head forward displacement is nearly identical to that of the 3 m/s^2 pulse, even though the deceleration level is higher up to 1.0 m/s^2.

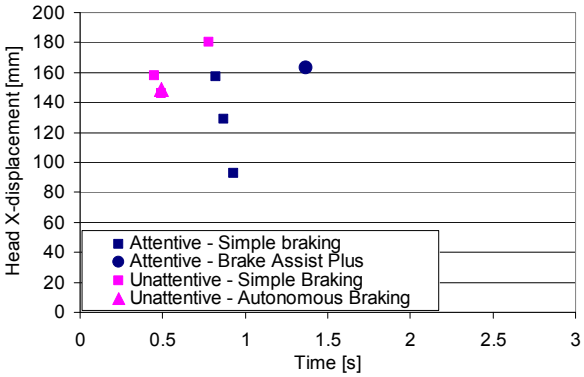

Figure 18. Head forward displacement as function of time for various combinations of braking scenarios and muscle activation strategies.

The T1 forward displacement is very similar between attentive and inattentive occupant for the simple braking cases. This indicates that the spinal stabilisation algorithm has limited influence in the simulated setup, possibly again due to the fairly optimal restraint with rigid seat. Again for BAS+ a delay is observed.

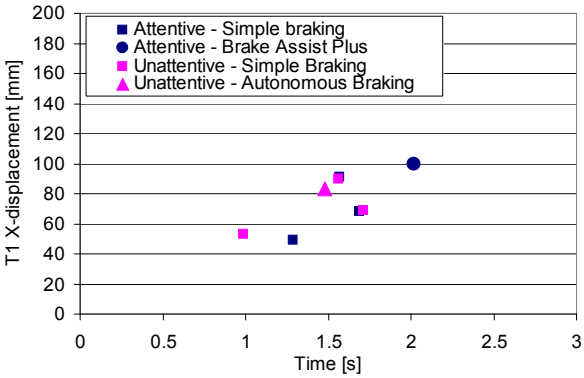

Figure 19. T1 forward displacement as function of time for various combinations of braking scenarios and muscle activation strategies.

DISCUSSION

The human model in this study was validated for specific dynamic loading conditions, such as frequency perturbations, rear impact and hub impactor tests. A first application in a braking scenario demonstrated that the validated controller settings were not valid for this application. The multi-body neck model with musculature was validated for front, rear and side impact. As such, the lack of validation in this case demonstrates that muscle control strategies of humans are more complex than currently implemented. The question that can not be answered based on the current study, but that would need to be answered is whether the chosen controller approach can be used to result in a model that can be validated for a number of scenarios. In other words, can a PID controller with delays and co-contraction setting be tuned to represent a number of scenarios while the parameters that define the controller are known instead of need to be tuned for every specific condition?

The fact that the validated setting did not create results that were anticipated is based on empirical findings as opposed to on the availability of a specific volunteer braking validation dataset. The braking tests performed by Ejima (2009&2007) are sufficiently similar to make a quantitative comparison with the results from this study. Head forward displacement in 8 m/s^2 deceleration was around 100 ms in Ejima's volunteer dataset, while it would be between 129 and 157 mm based on these results. T1 displacement was around 25 mm in Ejima's tests, while it would be between 68 and 91 mm in this study. As such, this model predicts around 50 mm larger T1 and head forward displacement. Since T1 is largely influenced by the seat belt, this may cause the better restraint of the torso and resulting lower T1 and head forward displacement. Additionally, Ejima discussed the activation of the sternocleidomastoid muscle early in the braking phase, which was not observed in the current study. The activation of longus colli, one of the paravertebral muscles, to overcome extension during rebound was observed in both Ejima's study as well as in the current.

CONCLUSIONS

The developed model showed applicable and sensitive to frontal pre-crash scenarios, however specific validation for frontal pre-crash braking based on kinematics and muscle activation patterns is required for assessing the controller parameters.

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