A Human Body Model With Active Muscles for Simulation of Pretensioned Restraints in Autonomous Braking Interventions

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Objective: The aim of this work is to study driver and passenger kinematics in autonomous braking scenarios, with and without pretensioned seat belts, using a whole-body finite element (FE) human body model (HBM) with active muscles.

Methods: Upper extremity musculature for elbow and shoulder flexion–extension feedback control was added to an HBM that was previously complemented with feedback controlled muscles for the trunk and neck. Controller gains were found using a radial basis function metamodel sampled by making 144 simulations of an 8 ms$^{-2}$ volunteer sled test. The HBM kinematics, interaction forces, and muscle activations were validated using a second volunteer data set for the passenger and driver positions, with and without 170 N seat belt pretension, in 11 ms$^{-2}$ autonomous braking deceleration. The HBM was then used for a parameter study in which seat belt pretension force and timing were varied from 170 to 570 N and from 0.25 s before to 0.15 s after deceleration onset, in an 11 ms$^{-2}$ autonomous braking scenario.

Results: The model validation showed that the forward displacements and interaction forces of the HBM correlated with those of corresponding volunteer tests. Muscle activations and head rotation angles were overestimated in the HBM when compared with volunteer data. With a standard seat belt in 11 ms$^{-2}$ autonomous braking interventions, the HBM exhibited peak forward head displacements of 153 and 232 mm for the driver and passenger positions. When 570 N seat belt pretension was applied 0.15 s before deceleration onset, a reduction of peak head displacements to 60 and 75 mm was predicted.

Conclusions: Driver and passenger responses to autonomous braking with standard and pretensioned restraints were successfully modeled in a whole-body FE HBM with feedback controlled active muscles. Variations of belt pretension force level and timing revealed that belt pretension 0.15 s before deceleration onset had the largest effect in reducing forward head and torso movement caused by the autonomous brake intervention. The displacement of the head relative to the torso for the HBM is quite constant for all variations in timing and belt force; it is the reduced torso displacements that lead to reduced forward head displacements.

Keywords: human body model, active muscle, feedback control, finite element, occupant kinematics, seat belt pretension

Introduction

Integrated safety systems show a promising potential to reduce the traffic trauma problem. One such system, already implemented by several car manufacturers (e.g., Coelingh et al. 2007; Schittenhelm 2009), is autonomous braking prior to an accident. An injury reduction potential of up to 73% for Maximum Abbreviated Injury Scale 2+ injuries in car-to-heavy goods vehicle collisions was estimated provided that both vehicles were equipped with autonomous braking systems (Strandroth et al. 2012). With the emergence of integrated safety systems, studies that evaluate car occupant kinematics and muscle responses in precrash conditions have become more frequent. Several studies have investigated volunteer responses to precrash braking deceleration pulses (Ejima et al. 2007, 2008, 2009; van Rooij et al. 2013) and found that the level of muscle contraction is a significant factor that determines forward displacement and belt interaction forces. For occupants in the front passenger seat, a tendency toward larger forward displacements than for drivers was found by Carlsson and Davidsson (2011). Integrated restraint systems with reversible pretension, reducing belt slack, and having the potential to reposition the occupant prior to the impact have also been investigated (Develet et al. 2013; Good et al. 2008; Ito et al. 2013; Mages et al. 2011; Ölsth et al. 2013; Schöneburg et al. 2011). It has been indicated that an occupant position close to the steering wheel prior to a collision increases injury risk (Bose et al. 2010). The displacement...
of the occupant in the forward direction caused by braking can be reduced by reversible pretensioning (Ito et al. 2013; Mages et al. 2011; Ölafsdóttir et al. 2013; Östh et al. 2013; Schöneburg et al. 2011). In parallel, research on numerical occupant human body models (HBMs) has been conducted, with the aim to incorporate human muscle activation schemes for the modeling of occupant precrash responses. Early implementations of active musculature in HBMs were made for the cervical spine. Muscle activity was applied, in the form of a ramp function, to all cervical muscles to model volunteer neck kinematics in frontal, lateral, and rear-end impacts (Brolin et al. 2005; de Jager 1996; van der Horst 2002). This simultaneous activation of all muscles can be hypothesized to represent a startle response to the impact deceleration. However, to model precrash events with a duration of 0.1–2 or more seconds, balancing muscle activation levels must be applied. For this purpose, neuromuscular feedback control to regulate muscle activation levels has been implemented in HBMs. Meijer et al. (2008) noticed that, in far-side impacts, there is sufficient time for the passenger to react. They integrated feedback-controlled torque actuators for the spine, previously developed by Cappon et al. (2007), with active line muscle elements for the arms and lower extremities. In simulations of far-side impacts situations, Meijer et al. (2008) found that their active multibody HBM was not only successful in capturing volunteer reactions in a driver simulator but its response was also significantly different from that of an anthropometric test device. The same multibody HBM was later complemented with a detailed neck (Fraga et al. 2009; Nemirovsky et al. 2010); evaluated in autonomous braking and frontal, lateral, and rear-end impacts (Meijer et al. 2012); and enhanced with a more detailed neck and elbow model (Meijer et al. 2013). Neuromuscular feedback control has also been implemented in a finite element (FE) HBM. Östh, Brolin, and Happee (2012) used the elbow flexion angle to control elbow line muscle flexor and extensor activations; they then simulated volunteer responses to 15 N impacts to the hand. Using the angle of the head, neck, and lumbar spine, feedback control of line muscle elements was utilized to model passenger kinematic responses to autonomous braking interventions (Östh, Brolin, Karlsson, et al. 2012). There are a number of other implementations of active musculature for FE HBM controlled through an open-loop approach; for example, Iwamoto et al. (2012) and Behr et al. (2006). Feedback control to regulate muscle activation levels in autonomous braking events has been shown to be a feasible approach (Meijer et al. 2012; Östh, Brolin, Karlsson, et al. 2012). However, to the best of our knowledge, feedback control has not been used to study the combination of autonomous braking and reversible pretensioned restraints.

The aim of the present work is to study driver and passenger kinematics in unexpected autonomous braking scenarios with and without reversible pretensioned seat belts, using a whole-body FE HBM with active muscles. Furthermore, the presently available studies of reversible pretensioned restraints typically use one fixed level of pretension force—for example 110 N (Mages et al. 2011) or 120 N (Ito et al. 2013)—and a fixed activation timing (Ölafsdóttir et al. 2013; Östh et al. 2013). Therefore, the HBM was utilized to study variations in the timing and force of a reversible pretensioned 3-point seat belt for both drivers and front seat passengers in cars, in order to find suitable parameters for the activation of reversible pretensioned restraints.

**Method**

A whole-body FE HBM, the Total HUman Model for Safety AM50 version 3.0 (Toyota Motor Corporation 2008) was used in this study, hereafter denoted THUMS. The anthropometry and seating position of the THUMS is based on the 50th percentile male reported by Robbins et al. (1983); it consists of approximately 68,100 solid elements, 75,700 shell elements, and 3,400 one-dimensional elements. The model contains rigid bodies (e.g., the vertebrae) and deformable bodies (e.g., the intervertebral discs, ribs, skin, and internal organs). However, the original model does not contain any active muscles. Some enhancements were made to the THUMS as reported in Östh, Brolin, Carlsson, et al. (2012): constraints that fix nodes in the skin to the vertebrae were removed; the elastic moduli and the linear stiffness of the intervertebral discs, ligaments of the spine, and the skin were reduced. In addition, upper extremity joint ligaments and contacts were replaced with rigid-body joint definitions (Östh, Brolin, and Happee 2012). The FE solver LS-DYNA version 971, release 6.1.0 (LSTC Inc., Livermore, CA) was used. To implement active muscle control, the user function subroutine was used. Pre- and postprocessing were done with LS-PREPOST (LSTC Inc.) and MatLab (The Mathworks Inc., Natick, MA). Controller gains were tuned using a metamodel generated in LS-OPT (LSTC Inc.). A right-hand coordinate system with positive X in the forward direction and positive Z upwards was used.

**Shoulder Muscle Implementation**

Previously, proportional–integral–derivative (PID) feedback controllers have been implemented to control 178 cervical, 110 lumbar, and 14 abdominal Hill-type muscle elements, added to the whole-body THUMS and used to model car passenger responses to autonomous braking (Östh, Brolin, Carlsson, et al. 2012). Elbow flexion impulse responses were modeled for the right elbow of the THUMS, with a muscle representation consisting of 12 Hill-type muscle elements (Östh, Brolin, and Happee 2012). In the present study, the elbow muscle model and controller were mirrored and incorporated in the left and right arms of the whole-body THUMS. Eleven Hill-type line muscle elements were added to each shoulder (Table A1 and Figure A1, see online supplement), with physiological cross-sectional areas from Holzbaur et al. (2005) and origin and insertion points according to Staudinger (2008). To capture the line of action of the deltoid and latissimus dorsi, the muscle elements were coupled in series with seat belt elements that were fed through slip rings. A high stiffness (10,000 N/engineering strain) was used for the seat belt elements to minimize the influence on muscle strain. The deltoide muscles were routed around a
The thoracic part of the latissimus dorsi was routed around the eighth rib and the lumbar and iliac parts on the ninth rib on the lateral side. In addition, the glenohumeral joint was allowed; adduction–abduction and humeral rotation were restricted with passive rotational stiffness of 6,000 Nm rad$^{-1}$.

The muscle elements were modeled according to

$$\sigma = [N_d(t) \cdot f_c(v) \cdot f_l(l) + f_{pe}(l)] \cdot \sigma_{\text{max}} + \sigma_d,$$  \hspace{1cm} (1)

where $N_d(t)$ is the muscle activation level, with a range of 0–1, $f_l$ and $f_c$ are the force–length and force–velocity relations of the Hill element, and $f_{pe}$ and $\sigma_d$ are the contributions of the nonlinear stiffness and damping of the passive muscle. Hill muscle material constants for all shoulder muscles were the same as for the elbow flexor muscles, reported in Östh, Brolin, and Happee (2012). Exceptions are the maximum shortening velocity, $V_{\text{max}}$, which was set to 10$^{\text{opt}}$ for the latissimus dorsi and deltoid elements based on the recommendation of Zajac (1989). Elbow muscles have a lower $V_{\text{max}}$ set to 5$^{\text{opt}}$ to compensate for the fact that, in the model, the muscle fiber length is about twice the physiological due to the omission of tendons for these muscles. Furthermore, for all upper extremity muscles the maximum isometric stress, $\sigma_{\text{max}}$, was set to 1 MPa (An et al. 1989). For the elbow muscles the physiological cross-sectional area reported in Östh, Brolin, and Happee (2012) was used without scaling of the area. For the muscles of the trunk and neck, $\sigma_{\text{max}} = 0.5$ MPa. Two different values were chosen to give the model maximum isometric strengths of similar magnitude as that of volunteers. For example, in elbow flexion and extension, the model strength is 86 and 48 Nm compared to volunteers, 78 (SD = 11) Nm and 50 (SD = 11) Nm (Buchanan et al. 1998). For cervical flexion and extension the model strength is 32 and 48 Nm compared to volunteers, 30 (SD = 5) Nm and 40 (SD = 8) Nm (Östh et al. 2013), measured relative to T1.

### Controller Implementation

The muscle activation levels, $N_d(t)$ in Eq. (1), for all muscles are determined by 7 PID feedback controllers, hypothesized to represent vestibular and proprioceptive feedback. For each of the angles shown in Figure 1, a control signal, $u(t)$, is computed according to

$$u(t) = k_p \cdot e(t) + k_i \cdot \int_0^t e(\tau)d\tau + k_d \cdot \frac{de(t)}{dt},$$  \hspace{1cm} (2)

where the error, $e(t)$, is the difference between the joint angle, $\gamma(t - T_{de})$, and the reference angle $r(t)$. In the present study, the reference is equal to the initial position of the model; hence, the controllers act to maintain the initial posture of the HBM. The transport delay, $T_{de}$, is used to account for the time needed for the neural signal to be conveyed to and from the central nervous system. The requested control torque, $u(t)$, is converted to a muscle activation request by scaling with the maximum isometric strength of the muscles. The scaled activation request is directed to the flexors or extensors depending on the sign of the signal; it is passed through a muscle activation dynamics model consisting of two coupled first-order filters, which gives muscle activation levels $N_d(t)$. The model utilizes $T_{de} = 34$ ms for the elbow and 30 ms for the shoulder as reported by de Vlugt et al. (2006). For the head and neck, a shorter delay of $T_{de} = 20$ ms is estimated due to the proximity to the spinal cord, matching the 18 ms delay for the cervico-ocular reflex in cats reported by Peterson et al. (1985). For the lumbar spine muscles, $T_{de} = 25$ ms is used, relatively close to the 30 ms for the lumbar spine muscles reported by van
Druten et al. (2014). The shoulder muscle activation dynamics constants are the same as for the elbow muscles reported in Östhus, Brolin, and Happee (2012). Furthermore, for all muscles an initial cocontraction level as a percentage of full muscular activation was estimated; in this study, the head, cervical, and lumbar muscles had a cocontraction of 3% and the upper extremity muscles 4%, which is based on the range found for volunteers during quiet driving (Östh et al. 2013).

**Tuning of Controller Gains**

To determine a set of controller gains, $k_p$, $k_i$, and $k_d$ in Eq. (2), that represent the driver response to autonomous braking, a volunteer sled test with 8 ms$^{-2}$ rearward acceleration (Ejima et al. 2008) was simulated. The volunteers sat in a rigid seat with a seat back angle of 20°, held onto a 340-mm-diameter steering wheel, and were restrained by a 3-point seat belt. In the present study, the HBM was positioned in the rigid seat with a 3-point seat belt as shown in Figure 1, and the acceleration pulse recorded in the volunteer test was applied to the rigid seat. First, a 0.5 s phase in which the model settled due to the influence of gravity was simulated and then 0.4 s of the sled test.

A total of 21 gains, 3 per controller, had to be tuned. To reduce the number of variables, first symmetric gains for the left and right upper extremities were used. Second, the integral gains, which mainly affect the stationary response of the HBM, were kept constant at the base line values. To tune the remaining 10 proportional and derivative controller gains, a radial basis function metamodel of the HBM response was generated. A space-filling design was used and the metamodel was sampled by a single-stage iteration consisting of 144 sled test simulations. For the head, cervical, and lumbar controllers, the baseline gains were taken from the simulation of passenger responses to 6 ms$^{-2}$ autonomous braking interventions reported in Östhus, Brolin, Carlsson, et al. (2012). For the elbow and shoulder gains, the baseline was estimated.

The fit of the HBM relative to the volunteer data was calculated with the weighted integrated factor (WIF) method (Hovenga et al. 2005). This method compares 2 time-history curves and yields a WIF value that is normalized to the range 0–1 to allow for objective rating of the similarity of response curves. Four metamodel responses were calculated by comparing the HBM time-histories to the volunteer data: the seat belt force, steering column axial force, and head and T1 forward displacements from one volunteer who was instructed to be relaxed in the sled test. A combined metamodel response was calculated, using a root mean square (RMS) addition with equal weights, $w$ (Hovenga et al. 2005), of the force and displacement WIF values:

$$\text{Combined WIF value} = 1 - \sqrt{\frac{\sum w_i (WIF_i^2)}{w}}.$$ 

The metamodel was then used to predict a set of tuned controller gains that would provide the highest combined WIF value.

**Model Validation**

Controller gains for the validation simulations were taken from the resulting tuned gains predicted by the metamodel; see Table 1. In the passenger simulations the controllers in the upper extremities were deactivated, thereby representing a passenger not using his or her arms for additional restraint.

To validate the ability of the HBM to predict occupant responses in potential precrash scenarios, simulations of the autonomous braking test cases reported by Östhus et al. (2013) and Olafsdottir et al. (2013) were made. In these experiments, the volunteers were instrumented with surface electromyography (EMG) electrodes on the flexor and extensor muscles of the neck, lumbar area, shoulder, and elbow. Maximum voluntary contractions were performed in a driving posture for normalization of the EMG signals. The volunteers were driving and riding in a passenger car on rural roads and were subjected to 20 autonomous and driver braking events in the driver seat (Östhus et al. 2013) and to 9 autonomous braking events in the passenger seat (Olafsdottir et al. 2013). A standard seat belt and another with 170 N reversible pretension applied 0.2 s before the braking were used. Kinematic data were collected by film analysis at 50 Hz; interaction forces between the occupants and the vehicle were measured, as well as driver seat indentions. In the present study, data from the autonomous braking test cases for 11 male subjects who had an approximate 50th percentile male anthropometry with regard to their average height (mean = 178 cm, SD = 5 cm) and weight (mean = 78 kg, SD = 6 kg) were selected for validation of the HBM response.

An FE model of the test vehicle seat (previously validated for quasistatic indention in Östhus, Brolin, Carlsson, et al. 2012) was used; see Figure A2, online supplement. The seat cushion was in its lowest position, in the middle of the fore–aft travel range, and a seatback angle of 25° was used. The seat belt attachment points were taken from the vehicle geometry and the average steering wheel position reported in Östhus et al. (2013) was used. In all simulations the HBM was initially positioned over the seat so that no initial penetrations in the contact between the seat and HBM were present. Footrest plates were positioned directly under the feet of the HBM. Belt slack characteristics were simulated with a nonlinear elastic element in series with the shoulder belt. The belt slack element stiffness characteristics were 0, 10, and 47 N at 0, 13, and 32 mm payout respectively, to match measured data from Östhus et al. (2013). Above 32 mm payout, a stiffness of 1,000 Nmm$^{-1}$ was used.

Prior to all simulations, 0.5 s presimulations with only gravity loading were run to settle the HBM in the seat, remove seat belt slack with a constant retractor tension of 3 N, and position the hands of the HBM at the steering wheel in the 10 and 2 o’clock positions for the driver simulations or in the lap for the passenger simulations. For the driver simulations, the HBM was leaned 3.5° backward around its H-point. For the passenger simulations the model was leaned 8.5° backward around the H-point and translated 10 mm forward relative to the driver position.

Four validation simulations were conducted: in the (D) driver and (P) passenger seat, with a (SB) standard and a
(PT) pretensioned 3-point seat belt. The average deceleration pulse recorded in the volunteer tests, corresponding to a velocity change from 70 to 20 kmh$^{-1}$, was applied to the seat, steering wheel, and footrest plates. The deceleration started 0.31 s into the trial, peaked at 0.9 s, and decreased rapidly at 1.8 s; see Figure 2. For the PT simulations, seat belt pretension was modeled by a retractor force–time relation of 0, 130, 170, and 0 N at 0.11, 0.33, 0.7, 1.8, and 2.0 s; see Figure 2. The belt force is applied 0.25 s before deceleration onset, defined as when the vehicle deceleration first reaches 5% of the peak deceleration; see Figure 2.

**Seal Belt Pretension Parameter Study**

A parameter study in which the belt activation time was varied in steps of 0.1 s, from 0.25 s before the deceleration onset to 0.15 s after, was conducted with the validated HBM. For each belt activation time 3 levels of pretension force were applied: 170, 370, and 0 N at 0.11, 0.33, 0.7, 1.8, and 2.0 s; see Figure 2. The belt force is applied 0.25 s before deceleration onset, defined as when the vehicle deceleration first reaches 5% of the peak deceleration; see Figure 2.

**Model Validation**

For the validation simulations, WIF values comparing the HBM response with the average volunteer responses from Ölsföldlí et al. (2013) and Öst et al. (2013) were calculated for the period 0–1.8 s and are presented in Table 2. It can be seen that the model was best at predicting forward displacements and interaction forces, whereas the vertical displacements were less successfully predicted.

Shoulder belt and steering wheel forces for the HBM compared with the volunteer data are shown in Figures A4 and A5 (see online supplement). The shoulder belt force magnitude in the simulations varied with restraint conditions in a manner similar to that of the volunteers. For example, in the driver position fewer than 100 N was found for the standard belt; with the pretensioned belt, forces over 200 N were found for both the HBM and the volunteers. However, in the passenger simulation with a standard belt, the force onset was more abrupt than for the volunteers. For the driver simulation with the pretensioned belt, the HBM peak belt force was larger than the volunteer corridor. The steering column force for the standard belt simulation matches the volunteer average well. However, with a pretensioned seat belt, an initial positive (tensile) force was found, which indicates that the HBM is slightly resisting being pulled back by the seat belt; this was not seen in the volunteer average.

The kinematics for the head center of gravity is shown in Figure 3, together with sternum forward displacement for the driver simulation and T1 vertebra in the passenger simulations.
passenger simulations. The initial peaks in the forward displace-
ments of the head and torso were above the volunteer corridors for the simulations with a standard belt. The model then returned inside and stayed within the corridors until the rebound phase that took place after deceleration had stopped at approximately 1.8 s. After 1.8 s the HBM showed a larger rebound than the average volunteer response for all 4 conditions. With pretensioned belts the HBM response was close to the average volunteer response and within the corridor for the forward displacements until the rebound phase. The HBM head flexion rotations were too large in all of the simulations.

Muscle activation levels for the agonist muscles in the HBM, compared with volunteer data, in the standard belt simulations are shown in Figure A6 (see online supplement). Volunteer antagonist activations were in the order of a few percent, just as prescribed for the HBM. The exception was the shoulder extensor muscles, for which the volunteer drivers had 10% activation (Östh et al. 2013), which was larger than the HBM in the validation simulations. Although it can be seen that the HBM cervical extensors showed an initial oscillatory response and that steady-state activations were larger than for the volunteers, the onset timing was in phase. For the lumbar extensors the average volunteer response showed an initial peak at 0.5–0.8 s, which was absent from the HBM response. The HBM matched the volunteer amplitude of lumbar extensor activation during steady-state braking in the passenger simulation with a standard belt, but it seemed too low in the driver simulation.

For the validation simulations with the pretensioned belt, the cervical extensors are within the volunteer corridors for both the driver and passenger simulations; in the driver simulation, the same is true for the shoulder extensors and elbow flexors as well. However, the lumbar extensors show only a small percentage of activation for both driver and passenger simulations, which is less than the average volunteer response.

**Seat Belt Pretension Parameter Study**

The HBM head and torso peak forward displacements resulting from the seat belt pretension parameter study are shown in Figure 4. Belt pretension at 0.15 s before deceleration onset had the largest effect in reducing forward head and torso movement caused by the autonomous brake intervention for the 170 and 370 N force levels. For the highest force level, the effect of timing was smaller. Furthermore, when the belt was activated at least 0.15 s before deceleration onset or 0.05 s for the 570 N force level, an initial retraction of the head was found; see Figure 4. It could be seen that the changes in torso peak forward displacements were of the same magnitude as for the head; that is, the reduced head displacements were due to the torso being constrained or pulled back by the seat belt.

Shoulder belt forces during steady-state braking (1.5–1.8 s) were similar for the same pretension forces and did not vary with belt activation timing. For the driver simulations the average forces during steady-state braking were 251 ± 39 N, 500 ± 36 N, and 690 ± 14 N and for the passenger simulations they were 344 ± 25 N, 476 ± 26 N, and 621 ± 32 N (±SD). In the driver simulations, early belt activation timings caused the HBM to resist the rearward pull of the seat belt initially. This led to a maximum steering column tensile force of 87 N when 570 N belt pretension was applied 0.25 s before deceleration onset.

**Discussion**

In this work, feedback-controlled muscles were added to the upper extremities, the trunk, and neck of an FE HBM. The
controllers were tuned by fitting the kinematic response of the model to that of a volunteer in an 8 ms$^{-2}$ volunteer sled test. The HBM was validated, with regard to driver and passenger kinematics, seat belt forces, and muscle activations in a second volunteer data set with 11 ms$^{-2}$ autonomous braking interventions both with and without a pretensioned seat belt. Finally, the HBM was employed for a parameter study in which pretension timing and force were varied. The HBM was found suitable to predict the forward displacements and interaction forces caused by autonomous brake interventions, in both the driver and front seat passenger positions.

**Tuning of Controller Gains**

The tuned gains as predicted by the metamodel compared to the baseline gains changed the most for the upper extremities. The baseline trunk and neck tuning was derived from a previous study of passenger responses to autonomous braking of similar magnitude (Östh, Brolin, Carlsson, et al. 2012), which is why it is not unexpected that these gains provided an acceptable match to the simulation setups in the present study. For the upper extremities, shoulder and elbow gains have been determined experimentally in other studies. Average proportional gains of 30 Nm rad$^{-1}$ for the shoulder and 20 Nm rad$^{-1}$ for the elbow were reported by de Vlugt et al. (2006), together with average derivative gains of 10 Nms rad$^{-1}$ for the shoulder and 5 Nms rad$^{-1}$ for the elbow. The tuned gains for the upper extremity controllers (see Table 1) were significantly larger than those experimentally found gains. However, this is probably due to the fact that in the experimental study (de Vlugt et al. 2006), the trunk was stationary and the hand was connected to a manipulator arm with an effective mass of 1.7 kg (de Vlugt et al. 2003). In the present study, the hand was stationary on the steering wheel, and the trunk, which has an effective mass that is on the order of 20 times larger, was moving. The influence of the upper extremities on the postural response of car drivers in braking has also been demonstrated with a multibody HBM (Meijer et al. 2013). However, the controller gains used were not described in their publication. Furthermore, in the present study, symmetric upper extremity gains were assumed and provided a reasonable response of the model. Due to the unsymmetrical loading of the 3-point seat belt, future studies might want to investigate possible benefits of differentiated upper extremity gains. The elbow proportional gain is on the limit of the allowed range, which indicates that it is possible that another set of controller gains that would fit the tuning simulations better exists for an extended controller gain domain.

**Model Validation**

The controller gains generated by the tuning simulations led to a slight increase in the correlation between the HBM response and the volunteer response (the combined WIF increased from 0.5 to 0.58 for the tuned gains compared to the baseline). The WIF values for the HBM response in the validation simulations were better with an RMS sum WIF of 0.66–0.70 for the forward displacements and shoulder belt forces; see Table 2. Steering wheel forces also correlated fairly well (0.59), whereas muscle activations scored lower. There are multiple reasons for this. With the exception of the lumbar extensors, the model muscle activations were higher than those of the average volunteer. An explanation for this is that the volunteers on average (Östh et al. 2013) had a somewhat higher isometric strength than the HBM for the extensor muscles; for example, 48 Nm compared to the volunteer average of 58 Nm for the elbow extensors. For the lumbar and elbow extensors, initial activation peaks were seen in the volunteer responses; see Figure A6. This was not captured in the model because there was no relative movement to generate a controller error at this time in the simulation. Because the model was in phase or even slightly earlier with regard to the kinematic response (see Figure 3), this appears to be a muscle response in the volunteers triggered by something other than the actual displacement. It could be a slight startle or tactile reflex response.

The lowest correlation between the model and the volunteer responses was found for the vertical displacements of the sternum and T1. The volunteers moved approximately 5–25 mm upward with the sternum and T1: the least with standard belt in the driver position, the most with the standard belt in the passenger position (Olafsdóttir et al. 2013; Östh et al. 2013). The HBM, on the other hand, exhibited a downward movement for the sternum of a maximum 25 mm in the driver simulations. This was observed also in Östh, Brolin, Carlsson, et al. (2012); it is due to the rib cage pivoting slightly downwards in the HBM due to the lack of abdominal pressure to support it from below. For the passenger simulations, peak T1 motion was upward but only about 15 mm with a standard belt. With a pretensioned belt, the HBM T1 vertical displacement was 3.5 mm below the volunteer corridor. Moreover, in the validation simulation setup, the HBM was positioned in a posture resembling that of the corresponding volunteers. The spinal curvature affects the sitting height and it was noticed that the sitting height of the HBM is less than for the average volunteer (87.5 cm compared with 94.7 cm). This could also have an effect on the belt interaction with the thorax of the model.

As can be seen from the width of the volunteer response corridors, there was a considerable variation in the volunteer responses. Therefore, in addition to calculating the WIFs for the HBM response and the average volunteer curves, the WIF of each individual volunteer curve compared with the average for the period 0–1.8 s was calculated as well. For these calculations only the force and kinematic measurements in Table 2 were included, because individual EMG response data contain much higher frequency content than the average volunteer EMG curves; see Figure A6. The resulting RMS sums for each test case are given in Table A2 (see online supplement), for the best matching volunteer, median match volunteer, worst matching volunteer, and HBM. The HBM WIF RMS sums for the kinematics and interaction forces in all events appear to be of the same magnitude as those of the median match volunteer; that is, the median WIF RMS of each individual volunteer test compared with the average. Just as for the HBM, low scores for the volunteer individual WIFs were found for the Z displacements, indicating that the average curves for these measurements were somewhat unreliable. This is
probably because the amplitude of the movement was small; it might have opposite signs for individual volunteers.

To conclude, the model can usually distinguish the difference between the 2 conditions: pretensed and standard belt. The matching of its forward displacements and interaction forces is comparable to that of individual volunteers with respect to the average curves.

**Seat Belt Pretension Parameter Study**

The results of the belt force and activation timing parameter study revealed that there was an initial retraction of the head for pretension activation at least 0.15 s before deceleration onset, as well as for the highest force level, 0.05 s before deceleration onset; see Figure 4. However, for the sternum and T1 some initial retraction was found for pretension forces greater than 370 N, indicating that the head did not have time to follow the torso before brake deceleration onset. Larger retractions were found for the driver simulations than for the passenger ones, probably due to a more upright posture that gives more interaction with the seat belt. For analysis of the forward displacements caused by the braking deceleration, the initial peak displacements were chosen. Although the forward displacements during steady-state braking differ slightly from the peaks, there is a clear correlation between the 2 measures; see Figure 3.

An analysis of the head peak displacements relative to the sternum and T1 peak displacements (see Figure 4) shows that for all pretension forces and timings, the difference between the head and the torso was between 55 and 64 mm for the driver position and between 63 and 74 mm for the passenger position in all simulations. This indicates that the reduction of forward head displacement was a direct effect of the torso of the HBM being restrained better by the seat belt, because the head movement relative to the torso was quite constant under all conditions. This result differs from that presented by Woitsch and Sinz (2014). They found that when 200 N seat belt pretension was activated simultaneously with 8 m s\(^{-2}\) braking deceleration, the forward head displacement of a Hybrid III anthropometric test device FE model was reduced from 141 to 44 mm, and chest forward displacement decreased from 57 to 11 mm. This means that a nonconstant relation between the head displacement relative to the torso was found, indicating a rotation of the whole head–neck complex rather than a protraction of the head relative to the torso as found in the present study. A constant protraction of the head relative to the torso was also found for the volunteers in autonomous braking with and without pretensioned belts (Östh et al. 2013; Schöneburg et al. 2011). It appears the HBM with active muscles can capture this human response better than the Hybrid III, which may overestimate the ability of belt pretension to reduce forward head displacements during autonomous braking.

Larger pretension forces systematically led to reduced forward head displacements, with the lowest forward head peak displacements, 60 and 75 mm at 570 N, for drivers and passengers, respectively, both when activated 0.15 s before deceleration onset. Without belt pretension, the peak head displacements in the validation simulations were 153 and 232 mm for driver and passenger, respectively; that is, a 61 and 68% decrease in forward head displacement is predicted for 570 N applied 0.15 s before deceleration onset. Due to the protraction of the head relative to the torso, a complete reduction of the forward head displacement caused by the autonomous braking deceleration is not likely to be achieved with belt pretension, because the torso would be required to move at least 60 mm rearward of the initial position. Even 570 N belt pretension applied 0.25 s before deceleration onset is not enough to move the sternum rearward relative to the initial position; the peak forward displacement of the sternum is still positive at 1 mm (see Figure 4).

A limitation of the parameter study carried out here is that the response of the HBM was tuned and validated with respect to volunteer data collected in test configurations without belt pretension or with maximally 170 N pretension. This level of belt pretension was seen to invoke a startle response in a majority of female volunteers but not for males (Östh et al. 2013). The potential for a startle response in 50th percentile male occupants also, as represented by the HBM, cannot be ruled out, because belt pretension forces increase to 370 and 570 N. However, the present study is likely to be representative of the kinematics of 50th percentile male occupants subjected to autonomous braking with the 370 and 570 N belt pretensions for the first time, because no effect of the startle on the kinematics was found for the first trial (Östh et al. 2013). The effect of repeated exposure to belt pretension forces higher than 170 N for 50th percentile male occupants combined with autonomous braking remains to be investigated, because no such studies have been published to the best of our knowledge.

Furthermore, 2 characteristics of the test setup are important, because they affect the predicted peak displacements and initial retractions presented in Figure 4. First, the deceleration gradient from onset to peak is approximately 23 ms\(^{-2}\); see Figure 2. This rate is representative of a modern autonomous braking system that utilizes the hydraulic brake pump for actuation. However, it is slower than the maximum rate during driver braking interventions, such as those employed in volunteer testing of pretensioned seat belts by Schöneburg et al. (2011). They used the brake pedal and a pneumatic ram for actuation, achieving a deceleration rate of 63 ms\(^{-2}\). Second, the initial force gradients of the belt pretension in the present study were 591, 1,290, and 1,980 N s\(^{-1}\). Changes to deceleration and force gradients are likely to affect the resulting occupant kinematics.

Driver and passenger reactions to autonomous braking, in combination with pretensioned and standard 3-point seat belts, were simulated using an FE HBM with feedback-controlled active muscles. The HBM was tuned using kinematic data from one volunteer sled test and validated for another data set with 4 combinations of autonomous braking with volunteers: in the driver and passenger seats, with and without belt pretension. The HBM correlates well for forward displacements and interaction forces, whereas head rotation angle and muscle activations are overstated. Variations in seat belt pretension force level and timing reveal that belt pretension should be activated at least 0.15 s before deceleration onset to provide the greatest effect in reducing forward head
and torso movement caused by the autonomous brake intervention; however, the effect of timing was less important with the high force level of 570 N. With increasing pretension force, more reduction of forward displacements is found in general; this effect is greatest for the HBM in the passenger seat. Moreover, the displacement of the head relative to the torso of the HBM is relatively constant for all variations in timing and belt force. It is the reduced torso displacements that lead to reduced forward head displacements.

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**Supplemental Materials**

Supplemental data for this article can be accessed on the publisher’s website.

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