# Validation of a Simplified Human Body Model for Predicting Occupant Kinematics during Autonomous Braking

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## ABSTRACT

*Finite element human body models (FEHBMs) without active musculature fail to accurately predict volunteer kinematics in various pre-crash scenarios that created the need for active FEHBMs. The objective of this study was to develop and validate a simplified human body model with active musculature for predicting occupant response during autonomous braking with and without active restraint systems. The Global Human Body Models Consortium (GHBMC) average male simplified occupant model (M50-OS) was updated with active musculature (M50-OS+Active) as part of this study. The two sets of simulations were carried out, one set with pre-tensioner and one set without pre-tensioner. The baseline M50-OS model was updated with one-dimensional beam elements representing all the major skeletal muscles. These beam elements were assigned Hill-Type muscle material. The updated model M50-OS+Active employs 32 proportional–integral–derivative (PID) controllers for maintaining the initial posture of the model, which is the assumed behavior of the volunteers. The output of PID controllers along with initial activation is used for calculating muscle activation levels for each muscle using the firing rate of motor neurons sigmoid function. Volunteer (n = 11) data from autonomous braking tests by Olafsdottir et. al. was used in this study. The data consists of occupant kinematics with subjects in passenger seat position. The autonomous braking event was simulated by applying the braking acceleration pulse of 1.1g in the x-direction in LS-DYNA (R. 9.1, LSTC, Livermore, CA). The braking pulse and seatbelt pre-tensioner loading pattern used in the simulations were taken from this volunteer study. Preliminary results from volunteer simulations have shown a strong dependence of reaction loads and kinematics on muscle activation. The results from the simulation without muscle activation were falling outside the experimental corridors. The head CG displacements in the x-direction are similar to the displacements observed in volunteer tests, demonstrating the ability of the selected approach in capturing the active muscle response. An active muscle version of the GHBMC simplified occupant model was developed and validated for predicting occupant kinematics in autonomous braking.*

## INTRODUCTION

In the year 2016, on average 102 fatalities per day were reported due to vehicle crashes (NHTSA, 2017). Motor vehicle accidents are a serious health and economical issue that has been motivating researchers and vehicles manufacturers around the world to design new safety systems. Great efforts are being carried out in developing integrated safety systems that work well before and during the crash to avoid a crash or when a crash is imminent, minimize injury to all the occupants. Autonomous baking is one of the safety systems that helps in reducing accident severity.

As the new cars will adopt this technology in the coming years, the frequency of autonomous braking will increase. It is necessary to study occupant behavior in autonomous braking events to understand occupant-vehicle interaction for optimum design of active safety systems. A reduction of 73% in Maximum Abbreviated Injury Scale 2+ injuries was estimated for vehicle collision between a passenger car and a heavy goods vehicle, both equipped with the autonomous braking system (Osth, et al. 2013). Autonomous braking coupled with motorized reversible pre-tensioning has the ability to reduce occupant injuries by reducing belt slack and reposition occupant in optimal pre-crash position (Cutcliffe, et al. 2015).

All the pre-crash braking events are of low severity (0.2-1.5g), longer duration (0.1-2 or more seconds), and low loads. Due to the longer duration of pre-crash events, there is enough time for muscle activity that alters the occupant posture from a nominal position. This deviation from nominal posture may increase injuries to the occupant as the restraint systems are designed assuming occupant is in nominal position. The reduction in forward kinematics with different level of muscle activity was reported by various studies (Bastien, et al. 2012, Beeman, et al. 2011, Choi, et al. 2005) that supports the fact that muscle activation has a greater influence on occupant kinematics.

It is necessary to evaluate the performance of these restraint systems with various pre-crash maneuvers and develop an optimal design. The finite element human body model (HBMs) is a useful tool for assessing the performance of the safety system and develop an optimal safety system. Various researchers (Iwamoto, et al. 2015, Iwamoto, et al. 2012, Kato, et al. 2017, Osth, et al. 2015, Osth, et al. 2012) have demonstrated the use of active HBMs in various scenarios. To use any model it is necessary to validate it in all the pre-crash and crash phase scenarios before using it in the design process. After successful validation of the model, it can be used for predicting occupant kinematics or assessing the performance of safety systems.

Therefore, the objective of this study is to develop and validate the simplified human body model with active musculature for predicting occupant response during autonomous braking. The Global Human Body Model Consortium (GHBMC) average male simplified occupant model (M50-OS) is updated with active musculature (M50-OS+Active). To validate the active model; data from a volunteer study with a volunteer in the front passenger seat of a passenger car was used (Olafsdottir, et al. 2013). This study has two different restraint conditions, one with a reversible pre-tensioner seatbelt and other with a standard retractor. In the current study, a total of 4 simulations were carried out, two simulations were performed using M50-OS+Active model and other two with M50-OS model.

## METHODS

### Active muscle version of the simplified human body model

The GHBMC M50-OS model was updated with one-dimensional (1-D) beam elements representing all major skeletal muscle throughout the body (M50-OS+Active). The physiological cross-sectional area for each muscle was taken from various literature sources (Bulcke, et al. 1979, Cutts, et al. 1991, Delp, et al. 2001, Friederich, et al. 1990, Fukunaga, et al. 1992, Holzbaur, et al. 2007, Langenderfer, et al. 2004, Murray, et al. 2000, Osth, et al. 2012, Pierrynowski 1982, Veeger, et al. 1991). All these 1-D beam elements were assigned Hill-type muscle material. This material model requires muscle activation level time history data. These muscle activation values for each muscle are calculated using a sigmoid function based on the firing rate of motor neurons given by Eq. 1 (Kato, et al. 2017). The input to this sigmoid function is the output of the PID controller ($u_i$)

and contribution values of each muscle for various body motions ($R_{ij}$). $i, j$ represents each muscle and the joint motion respectively. In Eq. 1, $A_i^0$ is initial activation, $C_i$ is the muscle activation coefficient.

$$AL_i(t) = A_i^0 + C_i . s_i$$

$$s_i = \frac{1}{1 + e^{(-9.19 \times w_i + 4.60)}} \tag{1}$$

$$w_i = \sum_j R_{ij} . u_i(t)$$

The M50-OS+Active model employs 32 PID controllers to control 32 joint motions in the body. The PID controlled muscle activation calculation is based on the assumption that the volunteer tries to maintain their initial posture (Han, et al. 2016, Iwamoto, et al. 2015, Osth, et al. 2012, Osth, et al. 2014). The PID controller is a closed-loop feedback control mechanism that uses joint angles as a process variable and compares it with desired set points to calculate error values ($e_j$). This error term is used in Eq. 2 to calculate the control signal ($u_j$). The PID controller output is affected by three controller gains, proportional gain ($k_P$), integral gain ($k_I$), and differential gain ($k_D$). The ad-hoc simulations were carried out to tune these controller gains.

$$u_j(t) = k_{Pj} . e_j(t) + k_{Ij} . \int_0^t e_j(\tau) d\tau + k_{Dj} . \frac{de_j(t)}{dt} \tag{2}$$

The sigmoid function and the PID controllers are implemented in LS-Dyna using *DEFINE CURVE FUNCTION and *DEFINE FUNCTION keywords. The PIDCTL keyword is used for the PID controller. The Hill-type muscle material is modeled using *MAT MUSCLE.

**Simulation setup**

To validate M50-OS+Active model in autonomous braking, data from volunteer experiment in the autonomous braking event (Olafsdottir, et al. 2013) was used. These experiments were carried out in two different restraint conditions, one with a pre-tensioner and other with a standard retractor with a volunteer in the front passenger seat. The braking acceleration pulse (Figure 1) of 1.2g was designed in such a way that the vehicle speed is reduced from 70 kph to 50 kph. The volunteers were unaware of the start of the braking pulse. The simulation setup used in this study is shown in Figure 2. The model was settled into a deformable car passenger seat model. A total of 4 simulations were carried out, two in each restraint condition (with or without pre-tensioner) with all muscles active in one set while no muscles in another set. This strategy was adopted to compare the results of active muscle model with the model without muscle activation.

The simulation with pre-tensioner was carried out but applying 170 N pretension as mentioned in the experimental study. The pre-tensioner curve was taken from the literature (Osth, et al. 2015). The pre-tensioner was activated at 110 ms as shown in Figure 1 by the blue line. The second set of simulations was performed by modeling a standard retractor with force-pull-out data taken from the literature (Osth, et al. 2015). The retractor locks if the vehicle acceleration if above 0.45 g or if the pull-out rate is higher than of 1.5 g.

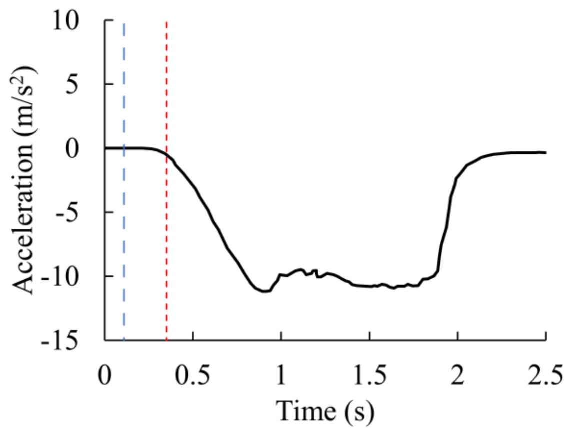


Figure 1: Autonomous braking pulse (Olafsdottir, et al. 2013), the blue line represents activation of pre-tensioner and the red line is for acceleration onset

The model kinematic and kinetic data from the simulation were compared with the results of the experimental study. The results of both the model with or without muscle activation are compared against the experimental data. This comparative data includes Head CG and T1 kinematics, and shoulder belt and footwell force. To quantitatively asses, the model performance was quantitatively assessed by performing CORA (CORrelation and Analysis, R 3.51) analysis. The results of simulations and CORA analysis are discussed in detail in later sections of this paper.

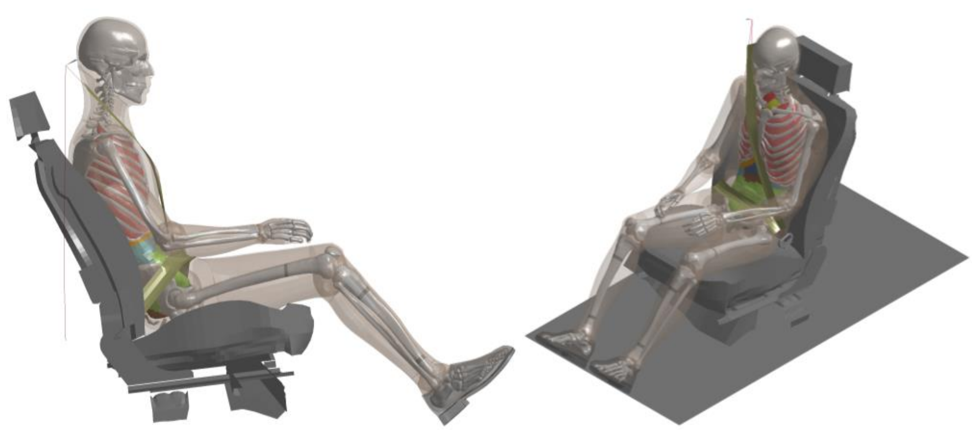

Figure 2: Left and Isometric view of the simulation setup

## RESULTS

### Kinematic and kinetic data

Head CG anterior-posterior (x-direction) displacement, vertical (z-direction) displacement, head CG rotation about y-direction time-history is shown in Figure 3. T1 anterior-posterior (x-direction) displacement, vertical (z-direction) displacement time-history is shown in Figure 4; while shoulder belt and footwell force time-history are given in Figure 5. In all these graphs, the solid black line represents M50-OS+Active model results with muscle activation whereas the dashed black line is for M50-OS model without active musculature. The experimental data are plotted with blue lines, solid line for mean and dotted lines for standard deviation.

**CORA analysis**

In the CORA analysis, seven time-history data was used. The total CORA rating for all the test cases is given in Table 1. The center column represents values for M50-OS model without muscle activation and the right column is for M50-OS+active model with muscle activation.

Table 1: Total CORA rating for all simulations

| Simulation | M50-OS | M50-OS+Active |
|---|---|---|
| With Pretensioner | 0.659 | 0.747 |
| Standard Retractor | 0.728 | 0.729 |

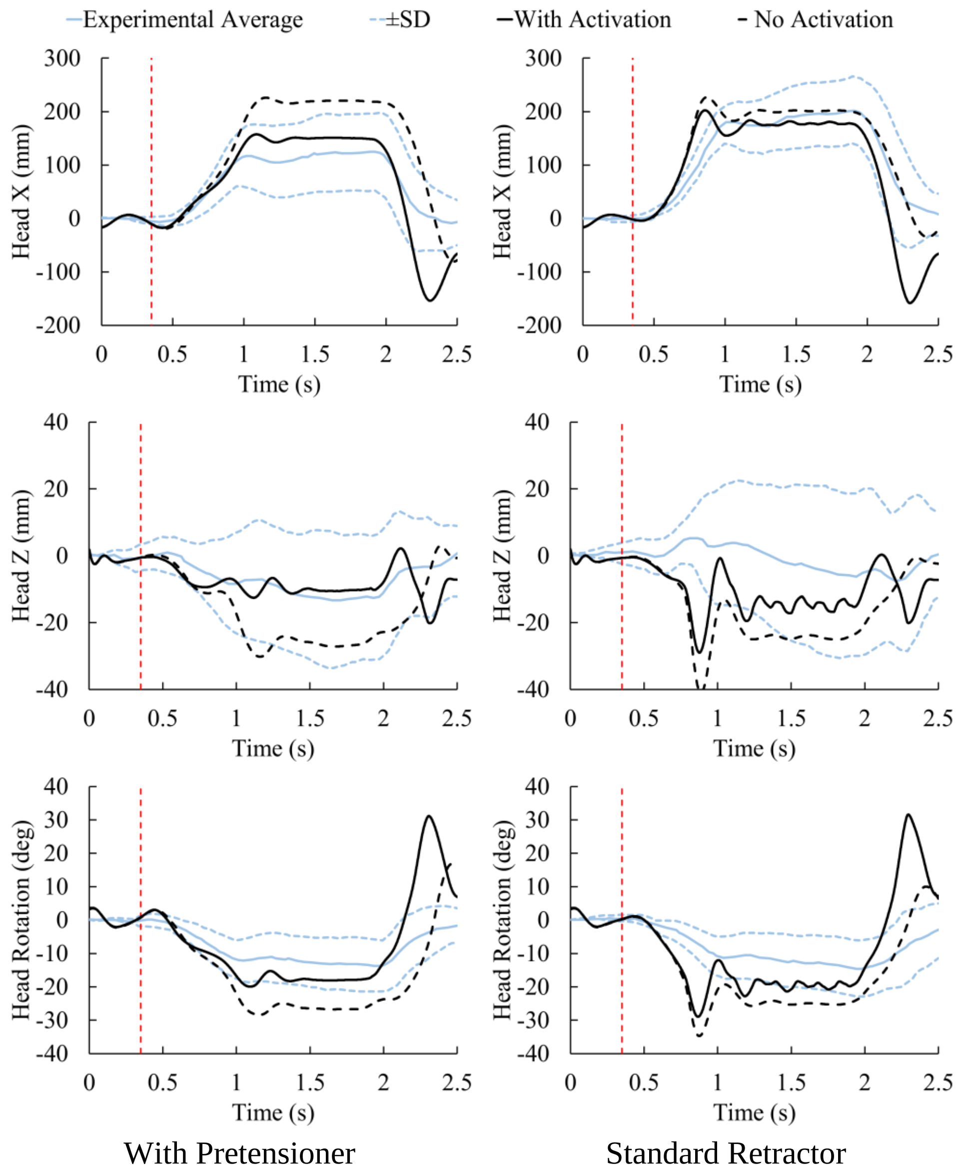


Figure 3: Head CG Kinematics, for simulations with pre-tensioner (left column) and with standard retractor (right column), Vertical dotted red line indicates deceleration onset.

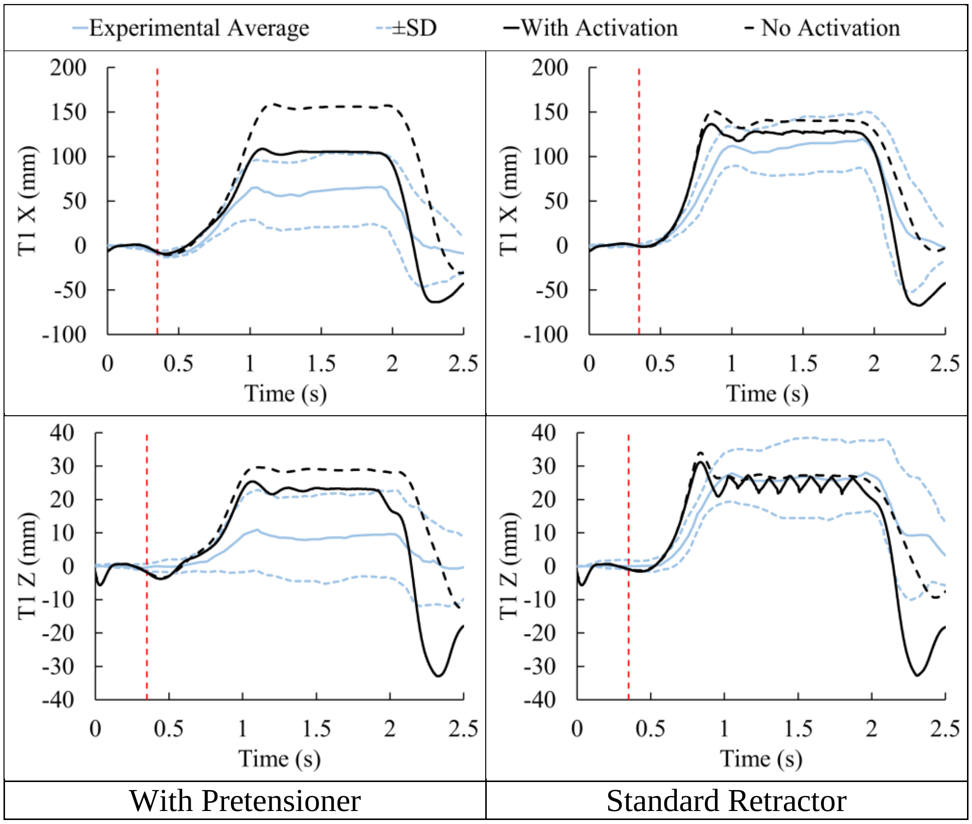


Figure 4: T1 Kinematics, for simulations with pre-tensioner (left column) and with standard retractor (right column), Vertical dotted red line indicates deceleration onset.

## DISCUSSION

Reduction in head CG kinematics as compared with the non-active model was observed in simulations with M50-OS+Active for both belt conditions, as shown in Figure 3. A similar reduction was observed in the T1 kinematics (Figure 4). Shoulder belt forces for both the models showed a similar trend in both simulations cases whereas footwell forces were lower for M50-OS model than M50-OS+Active, as shown in Figure 5. The M50-OS+active model showed overshooting in head CG and T1 kinematics, this might be the effect of aggressive control of trunk muscles. As acceleration magnitude started decreasing, the active model returned to initial posture faster than what was observed in the volunteer experiments.

In simulations with standard retractor with an active model, oscillations in response were observed. These oscillations are more prominent in T1 z-displacement (Figure 4), shoulder belt force (Figure 5), and footwell force (Figure 5). Tuning of trunk muscle PID controller gain will help in reducing these oscillations. The future work will focus on better ways of tuning PID controllers.

The CORA analysis shows 13.4% improvement with an active model in simulations with pre-tensioner as compared to M50-OS model. No significant improvement in CORA rating was observed in case of the standard retractor. For calculating CORA scores, an outer corridor for all the time-history data was calculated as ±2 SD.

The restraint systems greatly affect the model response. A lot of assumptions were made while modeling the belt system. As per the experimental study (Olafsdottir, et al. 2013), retractor was locked after 320 ms of the acceleration onset for standard retractor, while no locking was mentioned for the pre-tensioner case. In modeling of pre-tensioner case simulation, no locking was modeled and a 170 N pretension force was applied similar to experiments, but a plateau was observed in the shoulder belt force response between 650 ms to 1110 ms, which is not evident in experiments. This suggests the need for exhaustive belt modeling data. Another limitation of the study is the physiological cross-sectional area used for various muscles in this model are taken from various literature sources, that may not represent 50$^{th}$ percentile male population. Apart from this, there is a need for a better method for tuning PID controller gains.

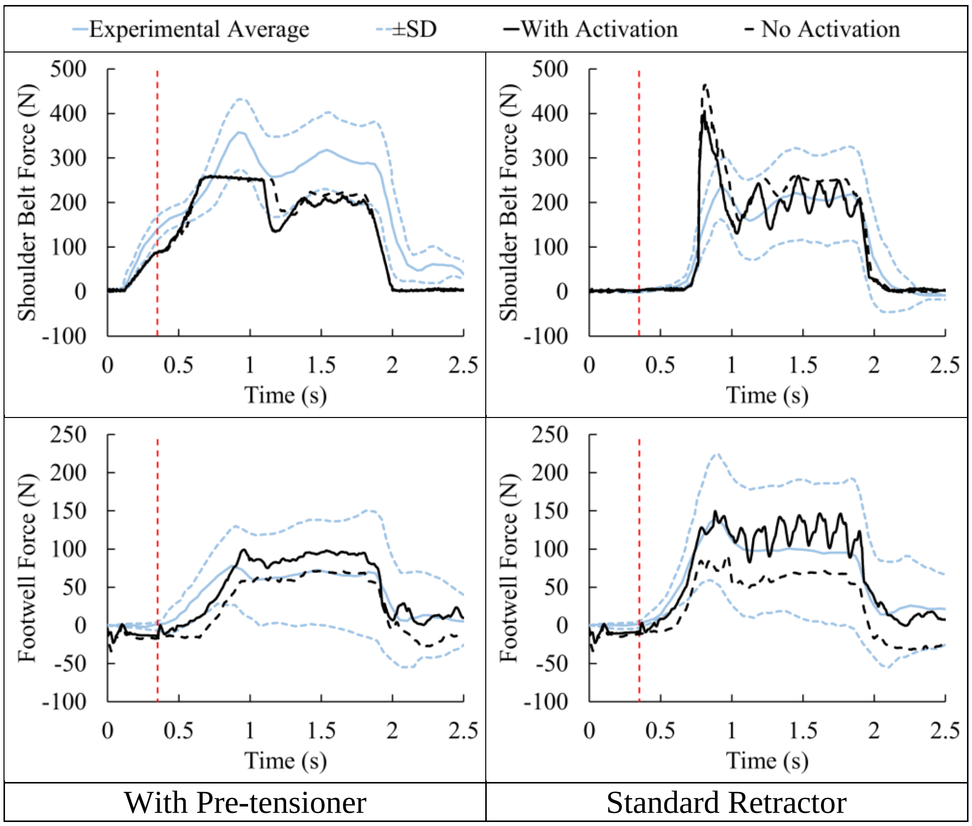


Figure 5: Shoulder belt and footwell forces, for simulations with pre-tensioner (left column) and with standard retractor (right column), Vertical dotted red line indicates deceleration onset.

## CONCLUSIONS

An active muscle model was developed and validated in an autonomous braking scenario. The M50-OS+Active model showed 13.4% improvement over M50-OS model in pretension case, whereas no improvement was observed in standard retractor case. The active model response emphasizes the fact that there is a strong dependence of reaction loads and kinematics on muscle activation. The active version of the simplified model has the potential for predicting occupant response in autonomous braking as well as its use in the design of better restraint systems.

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