

Evaluation of Frontal Impact Restraint System in Integrated Safety Scenario Using Human Body Model with PID Controlled Active Muscles

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Abstract Automated Emergency Braking (AEB) systems have been on the market for the past decade, and are growing rapidly in use. Real-world accident analyses indicate the value of studying countermeasures in pre-impact braking phases. An AEB system can modify occupant pre-crash positioning, which has not yet been evaluated in rating and legal tests. The objective of this study was to evaluate frontal restraint systems in integrated safety scenarios: activation of diverse pre-impact AEB systems, followed by a frontal impact. A validated Active Human Body Model (AHBM), capable of mimicking human posture maintenance and reflexive responses under pre-impact braking, was used. The human body model includes active muscles controlled by a proportional integral derivative (PID) control system. Using the validated AHBM, frontal restraint system performances in combined pre-crash braking, and crash scenarios, were evaluated, including comparisons with conventional methods, i.e. the HBM without active muscles, and the THOR dummy. Effects of varying pre-pretensioner (PPT) forces were also evaluated. It was found that different braking durations and levels led to corresponding occupant positions. The PPT applied pre-crash maintained the occupant's initial position to some extent, resulting in a decrease in thoracic injuries. Using an AHBM appears to be a rational approach for the evaluation of integrated safety performance.

Keywords Active muscle, automated emergency braking, human body model, integrated safety, pre-pretensioner.

I. INTRODUCTION

Approximately a half-century of vehicle safety system history shows a variety of measures taken for the enhancement of automotive safety systems: starting from the three-point seat-belt system in the late 1950s; then, the anti-lock braking system (ABS) introduced in the 1970s; the launch of the airbag system in the 1980s; and the appearance of the electronic stability control (ESC) system in the 1990s.

With the advent of the 21st century, automated emergency braking (AEB) systems have been on the car market and are showing rapid penetration into the developed markets, such as in North America, Europe and Japan. The results of a study by Fildes et al. [1], concerning the effectiveness of low-speed AEB systems in real-world rear-end crashes, have shown a 38% overall reduction in striking rear-end crashes for vehicles with AEB systems. Kusano et al. [2] have also shown a reduction of crash severity (ΔV) in rear-end crashes by an average of 12–50%, and an avoidance of 0–14% of collisions according to braking pulse parameters (e.g. ramp-up time, brake level, and time to collision).

Intervention by an AEB system prior to a frontal impact can alter occupant position from the designed position. This has not been evaluated in crash tests where the crash dummy is placed in a specifically designed position. This positional alteration leads to a higher incidence of driver chest injury, at equivalent impact conditions [3]. In conjunction with the AEB system, the motorised pre-pretensioner (hereafter PPT) system was introduced in order to reduce the degree of an occupant leaving the designed-position [4]. To comprehensively assess the benefit of introducing so-called integrated safety systems (e.g. motorised PPT systems in conjunction with AEB) in order to reduce the degree that occupants are forced out of the designed position and subsequent in-crash injuries, an appropriate occupant model must be used. The occupant model should represent occupant responses, not only for in-crash loading, but preceding pre-crash loading. Since anthropomorphic test devices (ATDs), such as the Hybrid-III AM50, are designed to predict injury risk in relatively high-energy impacts (i.e.

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automotive collisions), they have limitations in the evaluation of low loading conditions (e.g. AEB loading) [5]. On the other hand, in contrast with ATDs, post-mortem human subjects (PMHS) exaggerate upper body forward excursion in low loading conditions compared to braced volunteers [5].

In order to simulate human posture maintenance and human-like reflexive responses during pre-impact brake loading, an FE HBM with proportional integral derivative (PID) controlled Hill-type active muscle system model was developed by Östh et al. [6]. The neuromuscular feedback control was implemented for the Total Human Model for Safety (THUMS) AM50 version 3.0 (Toyota Motor Corporation 2008), with some enhancements to the model [7]. The developed model – the so-called SAFER AHBM – with an active muscle system, was able to capture the kinematic responses during AEB events, and muscle activation magnitude was similar to that of the volunteers [8]. The SAFER AHBM uses a 1D Hill-type model, as muscle representation, with muscles controlled by PID feedback, via stabilising muscle activation generated in response to external perturbation.

For the purpose of full-scale application of the AHBM to industrial usage, i.e. to make it usable for new product development in an industrial context, the model also needed to be numerically stable from pre-crash to in-crash, and to have a reasonable calculation time and scalability with parallel computing. These challenges have been met appropriately and as a result the AHBM has been industrialised [9]. In this study, this industrialised AHBM was used as the active human body model. This introduction of HBM with a PID-controlled muscle system has enabled the virtual development of integrated safety restraint systems throughout the pre-crash and in-crash scenarios (hereafter integrated safety scenario).

The objective of this study was to evaluate a frontal restraint system's performance – including the seat belt's PPT effect – in integrated safety scenarios using a validated AHBM. The evaluation using a passive HBM (hereafter PHBM), and THOR dummy, as the ATD was also carried out for comparison with the AHBM.

II. METHODS

In this study, all numerical simulations were performed using the FE program LS-DYNA version 971 (LSTC Inc., Livermore, CA). Pre- and post-processing were done with Is-prepost (LSTC Inc.) and Matlab (The Mathworks Inc., Natick, MA).

Firstly, a description of the car interior model used throughout the study is presented. Secondly, a description of the FE driver-side occupant models used in this study are also presented. Thirdly, the study cases are described, comprising three sections: *Study I*: An occupant kinematics comparison with human volunteer tests under AEB conditions; *Study II*: the pre-crash parameter study; and *Study III*: The integrated safety evaluations.

Numerical Model I: Generic Car Interior Model

As pre-crash computational time using explicit FE is significantly higher compared with the in-crash calculation alone, a generic car interior model was created to reduce computational time for studies, including pre-crash (Fig. 1). The model includes frontal restraint systems, such as a seatbelt retractor with a pyrotechnic pre-tensioner, load limiter, and PPT. A driver airbag is folded and packaged in a bag container at the centre of the steering wheel. Seat cushion, knee bolsters, and foot rest are modelled as rigid shell elements connected with 1D spring elements with corresponding force-stroke properties.

The car interior model was validated by means of a 56 km/h full frontal collision with a THOR dummy in a mechanical sled test. The THOR dummy was seated in a D-segment sedan-type body (Fig. 2). A generic pulse of a D-segment sedan-type vehicle for US NCAP 56 km/h frontal collision was applied (Fig. 3). No column collapse function was applied in either test or simulation.

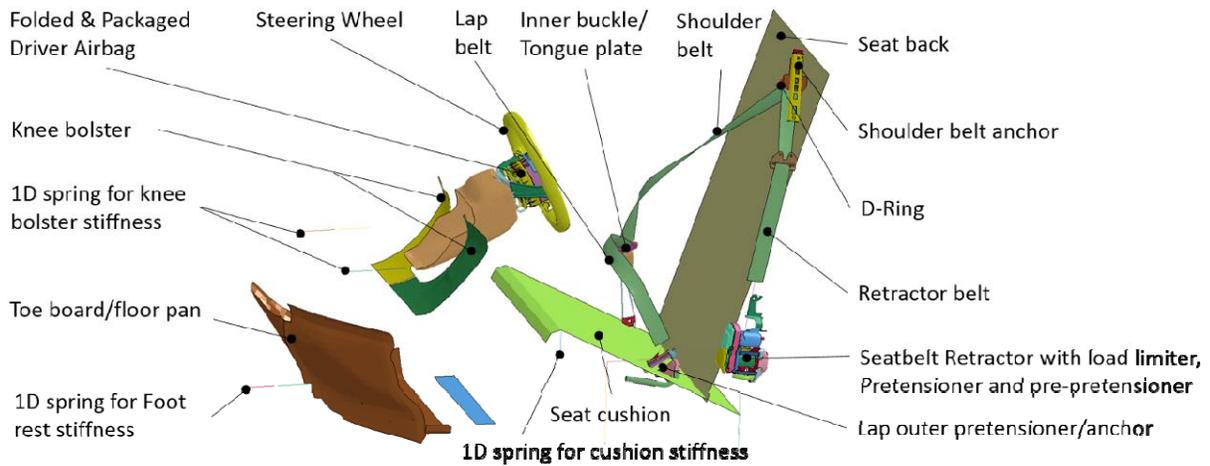


Fig. 1. Generic car interior FE model with frontal restraint system.

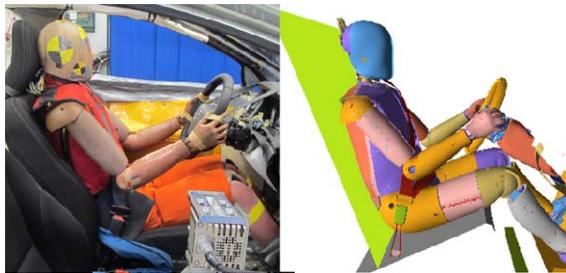


Fig. 2. Sled set-up with THOR dummy (test and simulation).

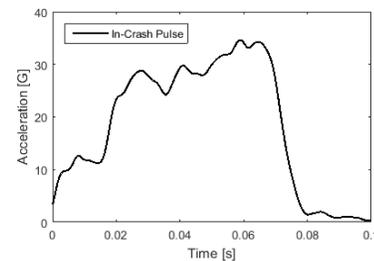


Fig. 3. Crash pulse used in sled test.

In this study, the objective rating method (ORM) [10] was used for the numerical model predictability assessment, with regard to the mechanical test. The ORM quantifies the correlation level between two output signals by scalar values, such as peak value of the signal and its peak times, and shape of the curve. A complete ORM value, which assesses the overall model validity, can be calculated from each group of ORM values. To calculate complete ORM values, the weight factor was set to 1.0 for all ORM values in this study. ORM values of 65% or higher were defined as good correlation between model predictions and mechanical tests according to Hovenga *et al.* [11].

Numerical Model II: FE Driver-side Occupant Models

In this study, three different types of FE occupant models were used as follows:

- The AHBM (Active Human Body Model with PID controlled muscles) – a validated AHBM [9], as described in the previous section.
- The PHBM (Passive Human Body Model without muscle activation) – for comparison. The PID-controlled muscle functions were deactivated in this model.
- The THOR dummy model – Humanetics THOR 50th Metric dummy ver.0.6 (Humanetics Innovative Solutions, Inc., Plymouth, MI) as the THOR ATD model for comparison with the AHBM.

Study I: Comparison with Human Volunteer Tests

As described in the previous section, a series of human volunteer tests were carried out by Östh *et al.* [8] for the AHBM validation prior to this study. In *Study I*, the responses from the numerical occupant models of AHBM, PHBM, and THOR were compared with human volunteer test responses, which consisted of the occupant's longitudinal head displacement, chest displacement, and lap-belt force. The applied AEB pulse in the numerical simulation was the same used in a series of human volunteer tests in [8].

Study II: Pre-Crash Parameter Study

Prior to the full-scale integrated safety study, a pre-crash phase study was carried out. This study focused on

the AEB scenarios in a longitudinal direction. The pitching effect during AEB activation was not taken into account in this study since the effect of pitching was considered to have only a minor influence on the measurements from the series of human volunteer tests [12]. For the purpose of the parameter study, the simplified AEB pulse was used, which consisted of: (1) a mechanical transfer phase; (2) a ramp-up phase; and (3) a peak hold phase (Fig. 4). After the AEB trigger, there was an initial nominal delay of 0.2 s for the hydraulic pump pressure in the brake booster unit to rise, allowing for brake acceleration as in the test measurement data [8]. This time period, from the AEB system’s trigger timing to the acceleration rising apparition timing, is denoted as the mechanical transfer phase. In this study, the mechanical transfer phase is set at 0.2 s for all pre-crash scenarios. Ramp-up phase was defined as time duration from the beginning of the initial rise until reaching the peak level. Peak hold phase was defined as time duration of the peak brake level until the pre-crash phase ends.

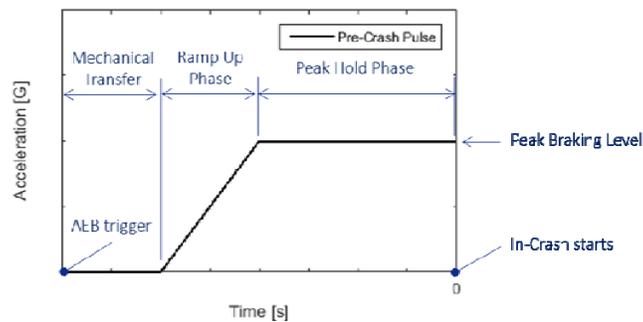


Fig. 4. Parametrisation of simplified pre-crash AEB pulse.

Table I shows the test matrix used for the pre-crash study by varying AEB pulse parameters. Case 1-1 and 1-2 are used to compare the influence of peak hold duration to occupant responses. The effects of seat-belt PPT force were evaluated as well, by varying PPT force from 0 N (i.e. without PPT) to a maximum of 600 N, with an interval of 150 N. PPT activation was set for the same timing as the AEB trigger timing. Each occupant model response, such as longitudinal displacement of the head, thorax and pelvis, was evaluated.

TABLE I
TEST MATRIX FOR PRE-CRASH STUDY

Pulse case	Ramp-up Phase (sec)	Peak Hold Phase (sec)	Peak Braking Level (g)	Speed Reduction ΔV (km/h)	PPT force (N)
Case 1-1	0.2	1.0	1.0	39	0, 150, 300, 450, 600
Case 1-2	0.2	0.4	1.0	18	0, 150, 300, 450, 600

Study III: Integrated Safety Evaluation

Following the pre-crash AEB scenario studies described in the previous section, the full scale, integrated safety scenario study has been evaluated. Figure 5 shows the designed integrated safety scenario pulse starting from a pre-crash AEB pulse, followed by a 56 km/h, full frontal rigid barrier collision pulse. Supposed initial cruising car speed is 95 km/h. After the AEB trigger, the brake pulse begins to rise with a 0.2 s ramp-up time until reaching 1 g, while the peak hold phase continues for 1 s, and then an in-crash pulse of full frontal collision starts (Table II).

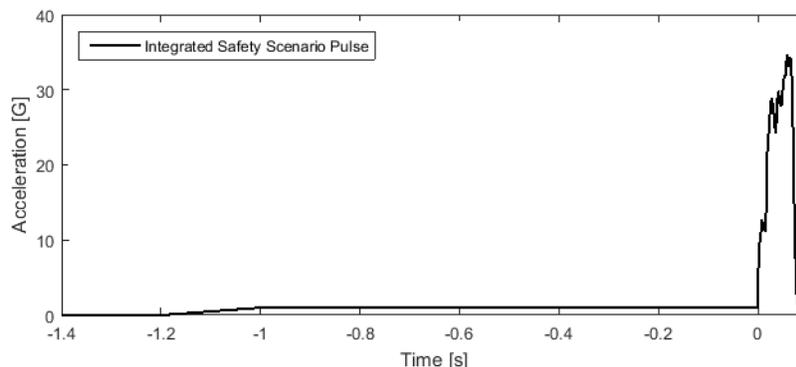


Fig. 5. Integrated safety scenario pulse, pre-crash AEB pulse, followed by 56 km/h full frontal collision pulse.

TABLE II
INTEGRATED SAFETY SCENARIO PULSE PARAMETER

Initial Cruising Speed (km/h)	Ramp-up Phase (sec)	Peak Hold Phase (sec)	Peak Braking Level (g)	In-Crash Speed (km/h)	PPT Force during pre-crash (N)
95	0.2	1.0	1	56	0, 150, 300, 450, 600

For the purpose of assessing the effects of seat-belt PPT forces in integrated safety scenarios, seat-belt PPT forces were varied from 0 N (i.e. without PPT) to a maximum of 600 N, with an interval of 150 N for each case.

For the injury criterion evaluation, HIC15 and BrIC [13] were evaluated for head injuries. In case of AHBM and PHBM, chest deflection was evaluated by measuring the distance between the 8th thoracic vertebra and the centre of the sternum, as a representative value. For THOR, peak resultant deflection R_{max} value, calculated by the four IR-TRACC peak resultant deflection [14], was evaluated.

III. RESULTS

Validation Results: Car Interior Model

Figure 6 shows physical and numerical sled test dummy responses with a 56 km, full frontal collision pulse, as described in Fig. 3, for validation of the car interior model.

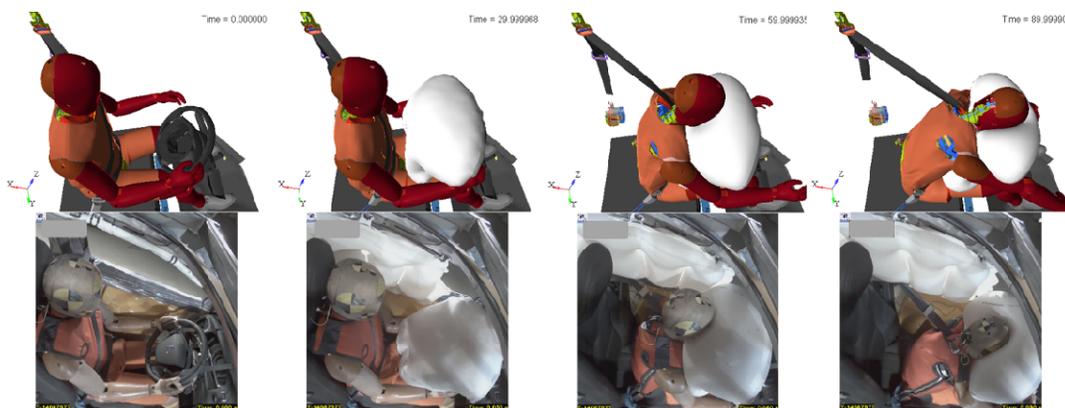


Fig. 6. The THOR dummy response in physical and numerical tests, from left to right: 0 ms; 30 ms; 60 ms; 90 ms.

Table A-I (Appendix) shows corresponding ORM values for each of the main output signals of acceleration, displacement and force. The complete ORM value, which assesses the overall validity of a generic car interior model with a frontal restraint system, showed 83%. Accordingly, this car interior model was considered valid, and has been used in subsequent studies with integrated safety scenarios.

Study I Results: Comparison with Human Volunteer Tests

Figure 7 shows the responses from human volunteer tests [8] (mean and ± 1 standard deviation (S.D.)) compared with numerical occupant model (AHBM, PHBM and THOR) responses, which consisted of occupant longitudinal head displacement, chest displacement, and lap belt force. The simulated forward displacement of the head and chest using the AHBM are within the volunteer corridor. But for the PHBM, forward displacement was too large, and beyond the volunteer corridor for both the head and chest. For the THOR, head and chest displacement were within the volunteer corridor, even if the chest displacements were close enough to the lower boundary of the volunteer corridor. Table A-II (Appendix) shows corresponding ORM values for each model's output signals of head displacement, shoulder displacement, and lap-belt force. The complete ORM value, which assesses the overall validity of each occupant model, showed 74% for the AHBM, 61% for the THOR, and 46% for the PHBM.

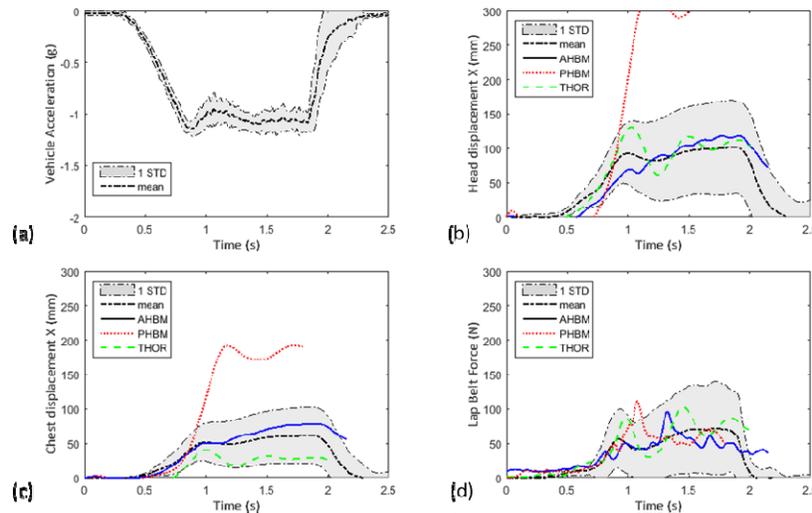


Fig. 7. Comparison of human volunteer tests and numerical simulation responses: (a) vehicle pulse; (b) head longitudinal displacement; (c) chest longitudinal displacement; and (d) lap-belt force.

Study II Results: Pre-Crash Parameter Study

According to the pre-crash study test matrix described in the previous section (Table I), driver occupant longitudinal excursion at the centre of gravity (CoG) of the head, T4, T12, and sacrum were measured for the AHBM (Fig. 8), PHBM (Fig. 9), and THOR (Fig. 10). The effects of varying the PPT force are also shown in these figures. The difference between Case 1-1 and case C-2 shows the effects of different brake peak hold times, i.e. 1.0 s and 0.4 s. For the AHBM, the effects of different brake peak hold times were observed, especially for the head longitudinal excursion in each PPT force (Fig. 8), e.g. head excursion 58 mm in Case 1-1, 44 mm in Case 1-2 with 150 N PPT applied. On the other hand, for the PHBM and THOR, little difference in the brake peak hold time effect was observed: PHBM, head excursion 151 mm in Case 1-1, 152 mm in Case 1-2 with 150 N PPT applied (Fig. 9); THOR, head excursion 68 mm in Case 1-1, 67 mm in Case 1-2 with 150 N PPT applied (Fig. 10). For the AHBM, the amount of occupant longitudinal excursion decreases as a function of the applied PPT force, i.e. a greater applied PPT force results in an occupant position closer to the initial occupant position. On the other hand, for the PHBM and the THOR, although applied PPT forces reduce occupant longitudinal excursion, it remains far from the initial position, as head excursion remains about 45 mm in PHBM and 40 mm for the THOR with 300 N PPT force or more.

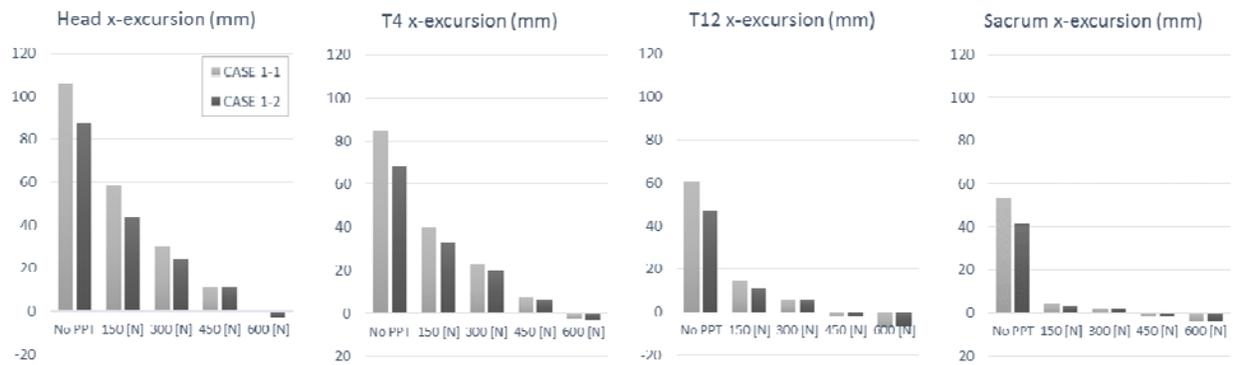


Fig. 8. AHBM occupant longitudinal excursion at Head CoG, T4, T12 and Sacrum with Case 1-1 and Case 1-2.

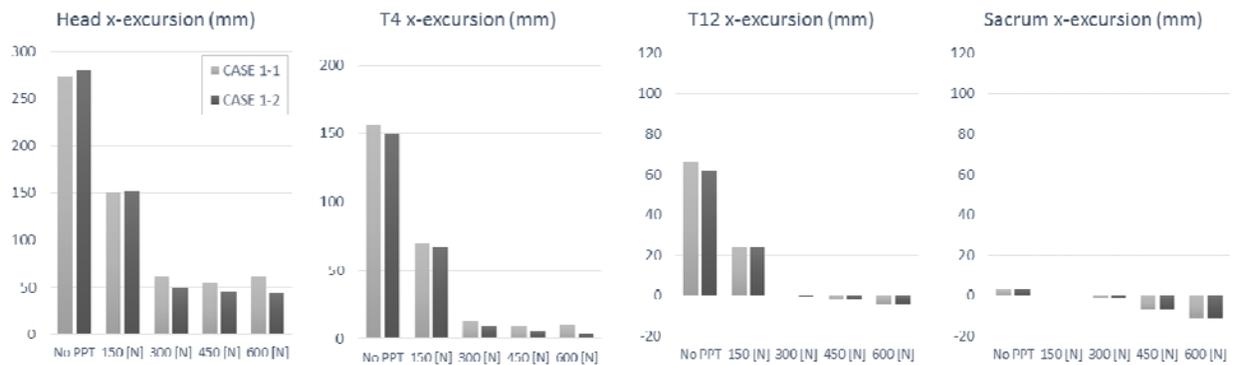


Fig. 9. PHBM occupant longitudinal excursion at Head CoG, T4, T12 and Sacrum with Case 1-1 and Case 1-2.

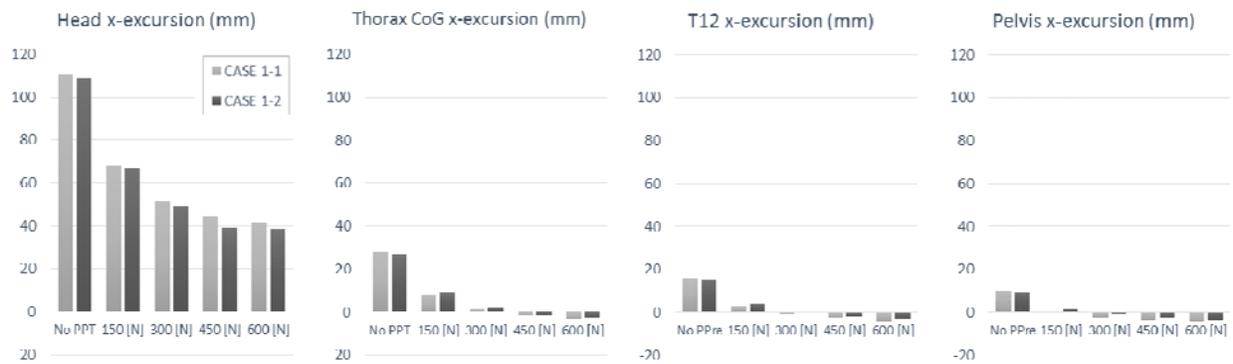


Fig. 10. THOR occupant longitudinal excursion at Head CoG, T4, T12 and Pelvis with Case 1-1 and Case 1-2.

Figure 11 shows each occupant (AHBM, PHBM and THOR) position in the sagittal plane with 150 N of PPT force with the Case 1-1 pre-crash scenario. PHBM's longitudinal excursion at the head CoG is about 160% larger than for the AHBM, 75% larger at T4, and 70% larger at T12. For the THOR, although head excursion is relatively close to the AHBM, chest excursions are about 80% shorter compared to the AHBM.

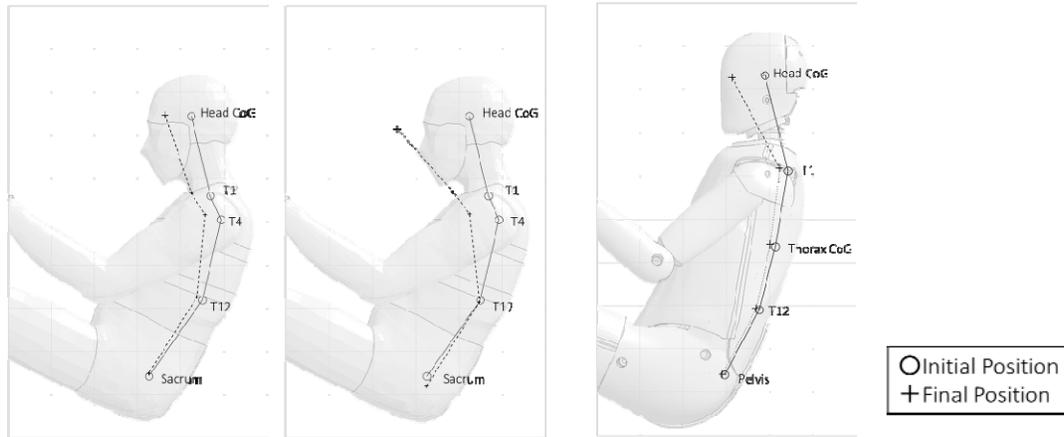


Fig. 11. Occupant position before/after AEB with PPT 150 N; AHBM (left), PHBM (centre) and THOR (right).

Study III Results: Integrated Safety Evaluation

According to the integrated safety study scenario described in the previous section (Table II), driver occupant load responses were evaluated all through the pre-crash and subsequent in-crash phases. Figure 12 shows injury criteria for the AHBM in the integrated safety scenario, with different PPT force applications. As described in the previous section, by increasing PPT forces in the pre-crash phase, occupant positions at the end of the pre-crash phase differed for PPT force levels, and higher PPT forces caused occupants to remain closer to their initial position. In case of no applied PPT forces, the occupant’s forward-leaning position led to early contact with the driver airbag. This resulted in lower head acceleration because the head’s relative velocity to the car, at the time of the head-to-airbag contact, was lowered. However, as observed in Fig. 13 below, where no PPT was applied, head-to-airbag contact occurred during the airbag deployment phase, i.e. the risk for airbag close-set deployment increased relatively compared to the PPT applied cases.

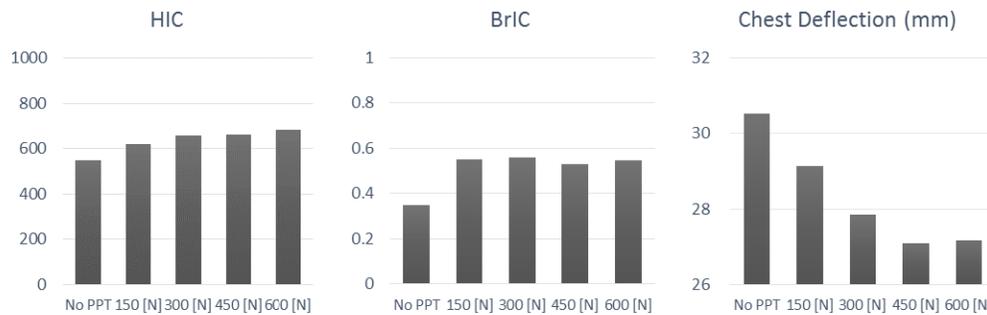


Fig. 12. Injury criteria for AHBM in the integrated safety scenario with different PPT forces in pre-crash.

Regarding maximum chest deflection values, higher PPT forces lead to lower chest deflection. As seen in Fig. 14, when no PPT force was applied, chest-steering wheel contact could be observed. Figure 15 shows plots of maximum chest displacement versus chest deflection with different PPT forces, which revealed a correlative relationship between these two variables ($p < 0.05$). Figure 16 shows plots for PPT force levels versus chest deflection, which also showed a correlative relationship ($p < 0.05$), hence PPT forces in pre-crash led to lower chest displacement during the in-crash phase. Accordingly, it led to lower chest deflection values. The higher PPT forces applied (>450 N) indicated no considerable differences neither for chest displacement nor deflection values, because as shown in Fig. 8, 450 N, and greater PPT forces applied in pre-crash, allowed the occupant to retain almost the same position as the initial one, which led to similar responses during the crash.

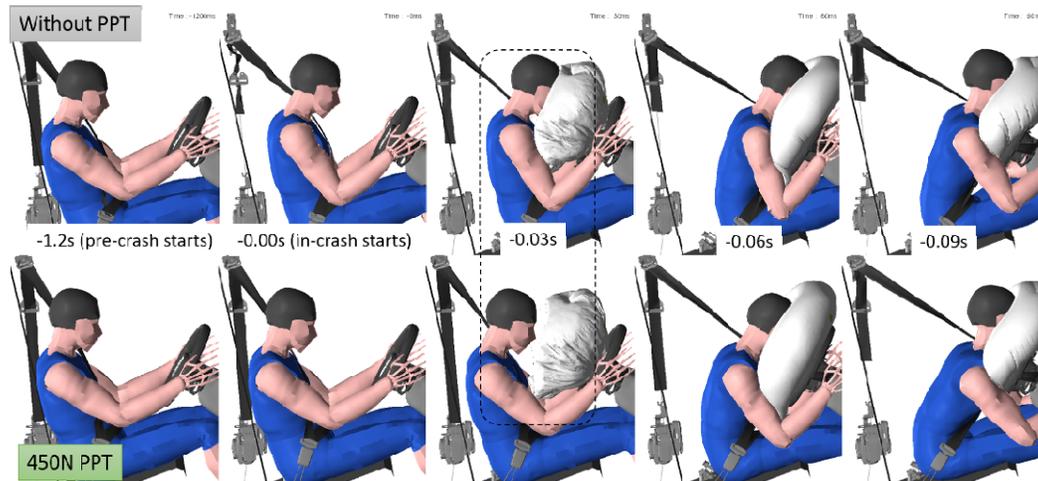


Fig. 13. AHBM occupant kinematics in an integrated safety scenario with no PPT force applied (*top*), and a 450 N PPT force applied (*bottom*) in pre-crash.

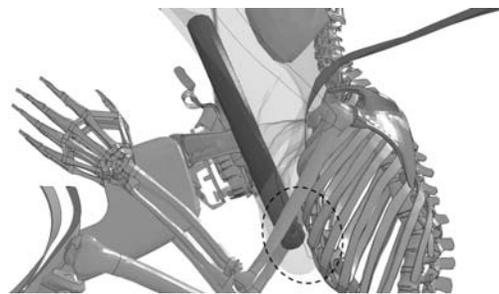


Fig. 14. AHBM at in-crash 0.08 s (max. chest displacement time) without PPT during pre-crash.

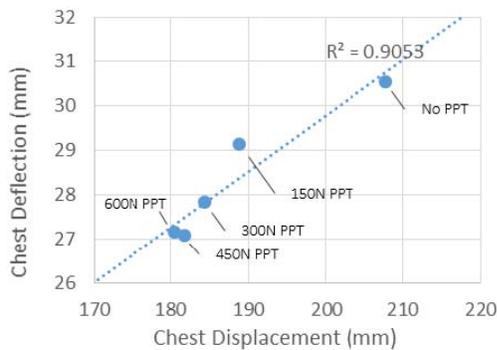


Fig. 15. Maximum chest displacement vs. chest deflection with different PPT forces.

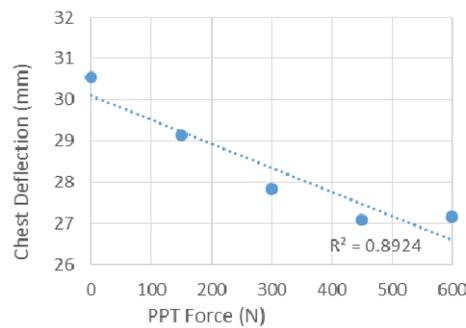


Fig. 16. PPT forces applied pre-crash vs. maximum chest deflection.

For PHBM, a calculation with 150 N PPT was done, where HIC, BrIC, and chest deflection were 640, 1.12 and 35.4 mm, respectively. By comparing the results of PHBM and AHBM under the same conditions of a 150 N PPT force applied, PHBM results were approximately 105% higher for the BrIC estimation, and 20% higher for the estimation of chest deflection. As shown in Fig. 11, PHBM exaggerated occupant forward excursion in pre-crash, for example, head excursion was 160% greater than AHBM, 75% greater at T4, and 70% greater at T12 with 150 N PPT applied – which resulted in a higher injury criterion estimate compared to AHBM results.

Finally, the evaluation using the THOR dummy model was conducted. In the case of THOR, as seen in the previous pre-crash results, the occupants' forward displacements were almost the same at both 1.0 s and 0.4 s of braking peak hold time (Fig. 10). Due to the limitation of the calculation cpu time, 0.4 s braking peak holding time applied, as well for integrated safety scenario evaluations for the THOR. HIC, BrIC, and each IR-TRACC displacement value were obtained (Fig. 17). As for HIC and BrIC values, a relatively similar trend was observed with different PPT forces compared to AHBM results – for example HIC values differed 13% between no PPT and

150 N PPT forces for the AHBM, while there was an 11% difference for the THOR dummy. Regarding the IR-TRACC displacement value, changing the PPT force from 0 N (i.e. without PPT) to 150 N, thorax upper right IR-TRACC displacement decreased from 51 mm to 45 mm. However, further increases of the PPT force caused no further decreases of IR-TRACC displacement, e.g. thorax upper right values showed 45 mm (150 N PPT), 48 mm (300 N PPT), 44 mm (450 N PPT), and 46 mm (600 N PPT), while for the AHBM, chest deflection values decreased monotonically by increasing the PPT force from 0 N to 450 N (Fig. 12).

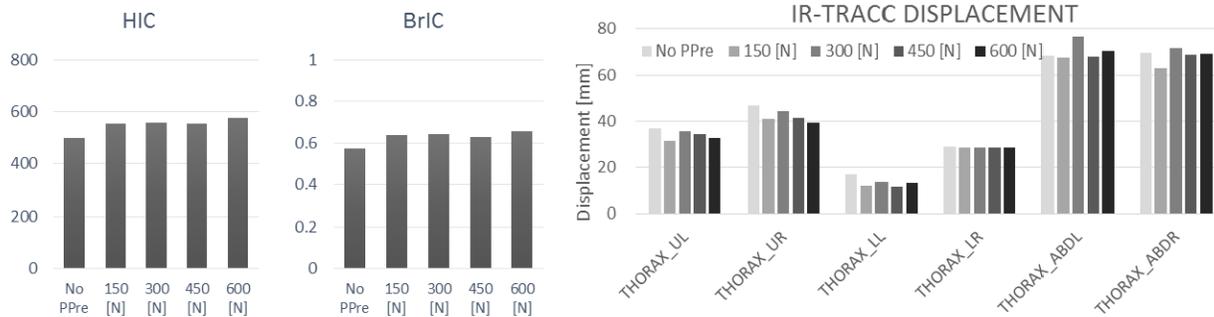


Fig. 17. Injury criteria and IR-TRACC displacement with different PPT forces applied.

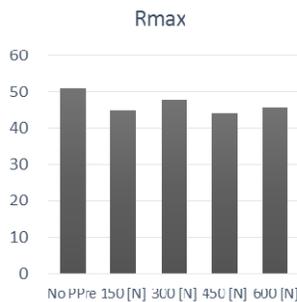


Fig. 18. R_{max} values with different PPT forces applied.

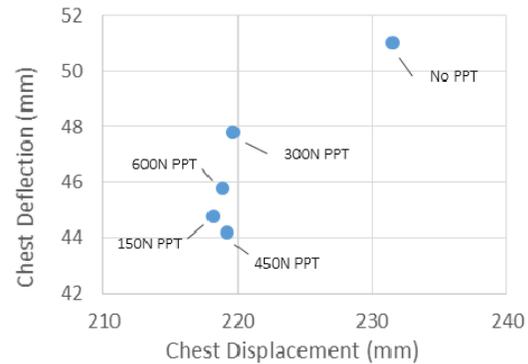


Fig. 19. Maximum chest displacement vs. R_{max} with different PPT forces applied.

Fig. 18 shows the peak resultant deflection R_{max} values calculated by the four IR-TRACC peak resultant deflections and Fig. 19 shows plots of maximum chest displacement versus R_{max} with different PPT forces applied. The effect of the absence or presence of PPT can be observed in Fig. 20, but no considerable effect of varying PPT forces on chest displacement could be observed for the THOR.

IV. DISCUSSION

In this study, the full-scale application of an AHBM, throughout the pre-crash and in-crash, was carried out with changing pre-crash pulses and seat-belt PPT forces to assess final injury outcomes in an integrated safety scenario. These were compared with the results from the PHBM and THOR dummy. Since there are different approaches to assessing restraint system performance with respect to pre-crash influences, Bose et al. [15] simulated effects of pre-collision to the injury outcome by varying occupant posture, as well as occupant bracing levels. The results showed occupant posture to be the most significant parameter affecting the overall risk of injury in frontal collisions. Woitsch et al. [16] evaluated pre-crash effects on dummy injuries in frontal vehicle crash simulations by considering dummy position changes due to pre-impact vehicle motion. The results showed effects of occupant forward excursion to injury values. These methods indicated the effects of postural changes, due to pre-braking loading, to the resulting injury outcome. However, as indicated in reference [8] and [12], human muscle response activations are induced, not only by pre-braking loading, but also by applying seat-belt PPT forces. This indicates the value of using an HBM with PID feedback-controlled muscles – validated both with or without PPT applied conditions in prior studies – in evaluating integrated safety devices’ further adaptiveness,

as demonstrated in this study. In other words, introduction of the feedback muscle control framework to the integrated safety evaluation makes it possible to grasp the chain of events, e.g. seat-belt PPT forces or different AEB pulses applied in the pre-crash phase, and their effects on final injury outcome.

The results from pre-crash studies using a passive HBM (without active muscles) showed significantly larger occupant forward displacements than the THOR dummy and the AHBM. The previous study [5] has shown that forward excursions of PMHS in pre-braking were significantly greater than those of the ATD, which are in accordance with this study by understanding that the HBM without muscle activation can be considered analogous to the PMHS by the lack of active muscles.

The results of AHBM injury outcomes in integrated safety showed that higher occupant forward displacement resulted in a higher contact risk with the steering wheel, although this can be more and less mitigated by a collapsible steering column. For head acceleration values, higher occupant displacements in pre-crash, led to early contact with the airbag, which resulted in lower head acceleration because of a lower relative velocity to the vehicle at the moment of head-to-airbag contact. At the same time, however, greater occupant forward displacements in pre-crash increases risks posed by airbag close-set deployment, which is in accordance with the previous study using a pre-crash induced dummy method [16]. Even if the car interior model used in this study was not comprehensive or representative of the total car population, the result that seat-belt PPT force application in pre-crash can reduce steering wheel contact risk, and airbag close-set deployment risk, can be a robust argument for the full frontal collision. At the same time, further investigation is needed to optimize PPT force levels under various pre-crash scenarios.

Limitations of the study

The results are derived from one crash test pulse, a 56 km/h, full frontal crash scenario, with a D-segment sedan type car interior FE model, and no effects of occupant size or gender differences were taken into account. This study focused solely on the driver, not on passengers. Accordingly, the evaluation of integrated safety with an AHBM for passengers will be the next step, in which test conditions, relatively greater occupant forward excursions might be expected, due to the lack of an arm bracing effect with the steering wheel. A comparison between AHBM and THOR responses for these cases (without an arm bracing effect) also needs to be carried out. The pre-crash pulses used in the study focused only on longitudinal motions. Accordingly, further validation is probably necessary to confirm the applicability and extensibility of the PID control solution for various pre-crash loading scenarios, such as lane change maneuvers [17-18]. In addition, this study did not account for anticipatory postural control, such as driver braking actions. Östh *et al.* have shown that anticipatory responses can be modelled by the time-dependent change in the reference value of the feedback controllers [19], which can be applied to an integrated safety evaluation with anticipatory postural control. In this study, for the evaluation of chest injury outcome, the representative chest deflection values are used to confirm the trend of chest displacement and deflection relationship. But to assess the effect of the integrated safety devices to chest injuries in a direct way, a strain-based, probabilistic method to predict rib fracture risk, presented by Forman *et al.* [20], can be applied to future studies.

V. CONCLUSIONS

Using a validated-HBM with PID-controlled active muscles, frontal restraint system performances in integrated safety scenarios can be evaluated, and compared to conventional methods using the HBM without active muscles, and the THOR dummy. The effects of changing pre-crash pulse parameters, as well as varying seat-belt PPT forces, to the occupant's state after the pre-crash, were evaluated. Seat-belt PPT forces applied in the pre-crash phase leads to lower occupant forward excursion in AEB loading, and shows benefits, especially in reducing the contact risk with the steering wheel and airbag, close-set, deployment risk in the crash phase. In summary, the results indicate that the applying AHBM with PID controlled active muscles is a rational approach to the safety performance evaluation of integrated safety.

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VIII. APPENDIX

TABLE A-I
 COMPARED MAIN SIGNALS AND CORRESPONDING ORM VALUES

Type	Signals	ORM Peaks	ORM Peak Times	ORM Curve Shapes	ORM Complete
<i>Acceleration</i>	Head Acc.	0.94	0.93	0.75	0.83
	Pelvis Acc.	0.97	0.76	0.65	
<i>Displacement</i>	Head Disp.	0.95	0.98	0.94	
	Pelvis Disp.	0.89	0.91	0.88	
<i>Force</i>	Thoracic Spine FX	0.72	0.98	0.69	
	Thoracic Spine FZ	0.60	0.88	0.30	
	Shoulder-belt Upper	0.98	0.92	0.75	
	Lap-belt Outer	0.98	0.77	0.83	

TABLE A-II
 ORM VALUES OF AHBM/THOR/PHBM TO THE HUMAN VOLUNTEER TESTS

Model	Signals	ORM Peaks	ORM Peak Times	ORM Curve Shapes	ORM Complete
AHBM	Head Disp.	0.86	0.98	0.65	0.74
	Shoulder Disp.	0.79	0.99	0.66	
	Lap-belt Force	0.75	0.77	0.56	
THOR	Head Disp.	0.78	0.55	0.65	0.61
	Shoulder Disp.	0.66	0.55	0.36	
	Lap-belt Force	0.70	0.85	0.64	
PHBM	Head Disp.	0.31	0.64	0.26	0.46
	Shoulder Disp.	0.32	0.95	0.29	
	Lap-belt Force	0.65	0.63	0.51	