COMPARISON OF THE INJURY RISK PREDICTION OF THE THOR-RECLINED DUMMY AND THE THUMS HBM

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ABSTRACT

Autonomous vehicles are expected to allow car occupants to position themselves in more relaxed positions inside the vehicle. These new seating positions constitute a new challenge for crash safety analysis. Therefore, new crash test protocols, adapted to this new paradigm, may be required in the future.
In the literature, most of the virtual reclined posture analysis has been performed using Human Body Models (HBMs) which are increasingly used to assess vehicle safety and injury risk, as currently regulated ATDs (Anthropomorphic Test Devices) are neither designed nor validated for reclined seating configurations. Nevertheless, these HBM simulation studies need to be correlated against repeatable physical tests that allow future cars to be rated according to regulation and consumer testing protocols. New options for crash dummies such as the THOR-Reclined kit from CELLBOND; which allows adapting the THOR ATD for these new reclined seating postures, are being developed and may enable the performance of physical tests in reclined occupant positions. However, the question of whether its performance is comparable to that of an HBM remains unanswered.
A series of simulations were then conducted comparing the behavior of the THOR-Reclined simulation model and the THUMS v4.1 by means of kinematics and injury risk prediction. Also, a series of tests using the THOR-Reclined in IDIADA's deceleration facility have been planned and the results will be shared in future publications. Injury risk prediction was then compared between the HBM and the ATD.
The ATD and the HBM FE models were compared by means of kinematics, restraint system outputs, injury criteria, and injury risk prediction. The result of this comparison will be discussed in this paper.
Some differences were observed between the models. THUMS allowed to study injury risk criteria based on the strain of the rib cage, while the ATD is mainly designed for measuring displacements and accelerations.
The primary limitation of this work is the lack of thorough validation data of the active HBM and the ATD model in the studied position. However, this work provides further insight into the comparability of their performance and the differences found between the studied models.
Differences have been found between the two models, mainly due to their physical dissimilarities. Nevertheless, some comparisons can be made between them from a kinematic and injury criteria perspective and will be shared in this paper.

INTRODUCTION

Future traffic scenarios, in 5 to 10 years, will most likely include mixed traffic (vehicle fleets composed of both traditional vehicles and vehicles with a high level of automation) or automated ones. Based on this information, Level 3 [1] vehicles are considered in the study. Level 3 vehicles represent "conditional automation" meaning that a human driver will respond appropriately to a request if needed. This level of automation allows the driver to be seated in a more relaxed, reclined position.

A current state-of-the-art frontal restraint system, i.e., a 3-point seat belt with B-pillar mounted belt guide, driver airbag in the steering wheel and knee bolster in the instrument panel, has limited protection functionality in the new proposed seating positions in which seatback angles are further from nowadays regulations' seatback angles. In particular, a reclined occupant posture may increase the risk of submarining [2] [3] [4], which is where the lap belt translates over the anterior superior iliac spines (ASIS) to load the abdomen directly and can result in injuries to the lumbar spine and hollow organs of the lower digestive system.

To assess in detail the injuries that may result from a crash, the injury output from a HBM or ATD is needed. The injury criteria output from an ATD is generally limited to specific scenarios (one specific ATD for one specific type of crash test) and limited by the mechanical elements that form the surrogate. Therefore, some of the measurements that can be done with these ATDs, like the chest displacement, are limited to four specific measurement points in the case of the THOR dummy (Upper Left, Upper Right, Lower Left, and Lower Right) or even one single measurement point for the Hybrid III dummy. HBMs, however, can measure the strain of every point of the rib cage, measuring all the circumference of the body and estimating the probability of 2, 3, 4, or 5 rib fractures depending on the age of the occupant. They can even be used to calculate if the structural integrity of the rib cage is in danger. Nevertheless, it is important to understand the variation of the outputs of both technologies, HBMs and ATD.

In the literature, most of the virtual reclined posture analysis has been performed using Human Body Models (HBMs) which are increasingly used to assess vehicle safety and injury risk, as currently regulated ATDs (Anthropomorphic Test Devices) are neither designed nor validated for reclined seating configurations. Nevertheless, these HBM simulation studies need to be correlated against repeatable physical tests that allow future cars to be rated according to regulation and consumer testing protocols. New options for crash dummies such as the THOR-Reclined kit from CELLBOND; which allows adapting the THOR ATD for these new reclined seating postures, are being developed and may enable the performance of physical tests in reclined occupant positions. However, the question of whether its performance is comparable to that of an HBM remains unanswered. The purpose of this study was to compare the outputs of a HBM and an ATD.

METHODOLOGY

Two simulation models (one HBM -THUMS- and one ATD -THOR-Reclined-) were compared by means of kinematics, injury criteria, and injury risk prediction in a generic frontal simulation environment [5] with a semi-rigid seat[1] during a frontal crash simulation. This semi-rigid seat was proposed by Uriot et al. [6] in 2015 and consists of two plates attached to a set of strings that can be changed to adjust the stiffness of the seat. The plate in the front recreates the anti-submarining foam of a standard seat and the second plate recreates the seat pan. This allows recreating a foam seat in a repeatable way while being a simple seat to model for simulation. The simulation model used in this study is shown in Figure 1.

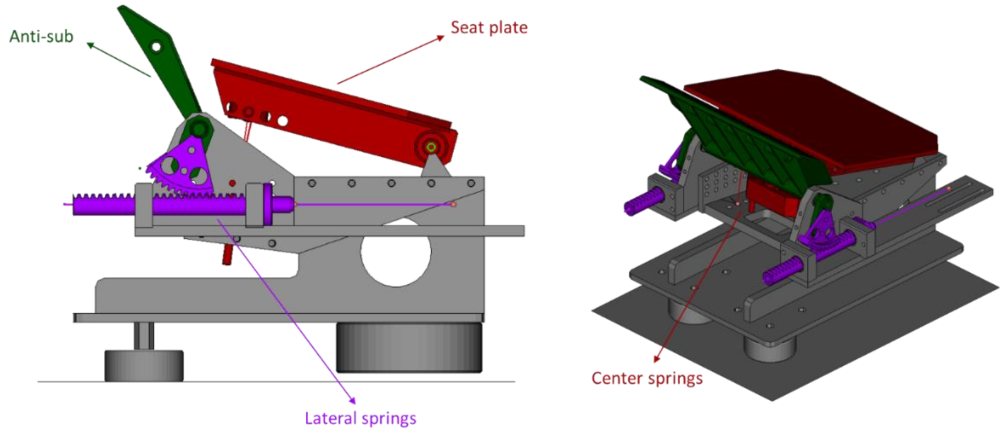
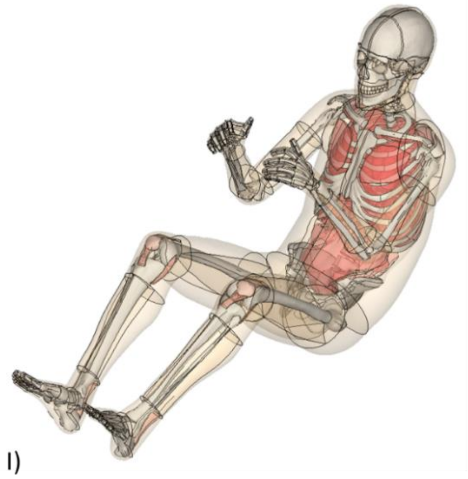


Figure 1: LAB CEESAR Semi-Rigid seat simulation model in the frontal configuration

Two impact simulations were performed using LS-DYNA MPP R9.3.1 (ANSYS/LST, Livermore, CA, USA) as solver and the Total HUman Model for Safety (THUMS, version 4.1) AM50 Occupant model (Toyota Motor Corporation, Japan) and the ATD-TH50R-D00.17_R00.06 model (ATD-MODELS GmbH, Weißwasser, Germany) as surrogates for the study. These two models are presented in Figure 2.

[1] The semi-rigid seat CAE model was provided by Autoliv [17] from the EU project OSCCAR [20]. This model replicates the lab CEESAR [18] semi-rigid seat which has been used in previous studies regarding frontal collisions [6] [19] [3].



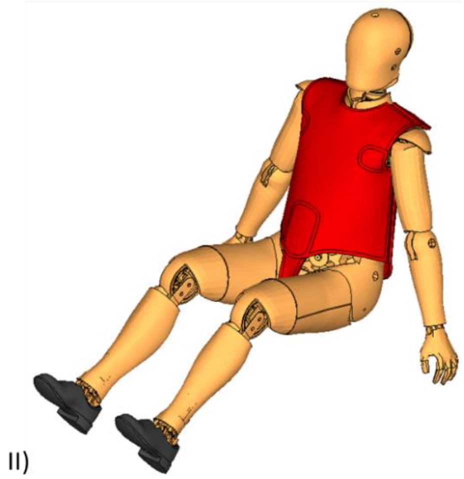


Figure 2: THUMS (Left, I) and THOR-Reclined (Right, II) simulation models in their original postures

Simulation Environment

The simulation environment used for this study is an adaptation of the generic frontal simulation environment used in the SAFE-UP project [5] (Figure 3). This environment consisted of a generic floor geometry and foot support, a semi-rigid seat, a generic seatback, a generic knee bolster, a State-Of-The-Art (SOTA) belt system installed in the seat, a simplified retractor with pre-tensioning and load limiting capabilities, a buckle with a crash locking tongue, an end bracket with pre-tensioner, a simplified belt webbing (defined using *MAT_SEATBELT card from LS-DYNA), a generic steering column (SC) with a production steering wheel, and a generic driver airbag (DAB). This generic environment model was validated by Autoliv.

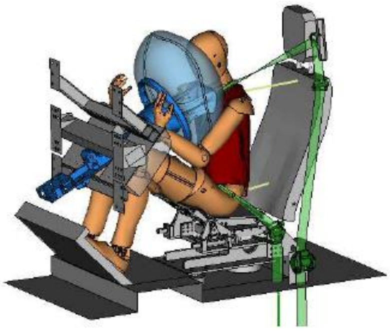


Figure 3: SAFE-UP Generic environment simulation model [5]

Some modifications were made to adapt the proposed environment to the IDIADA's sled testing facilities and to be able to reproduce it physically. The semi-rigid seat was used in the frontal configuration proposed by Richardson in 2020 [3] using 128N/mm seat pan lateral springs, a 379N/mm seat pan center spring, and 132 N/mm anti-submarining springs.

The seatback was simplified for easier construction as a physical part. The adapted model consisted of a rigid steel plate with foam on top to protect the dummy on the rebound phase for future sled tests. This foam was modeled as Ethafoam 220 and had no specific function during the crash phase. The dynamic model of this foam was provided by Autoliv for the SAFE-UP project. The seatback was positioned at a 45-degree angle according to the SAE standard [7] [8].

Regarding the footrest, an expanded polypropylene (EPP) foam with a density of 30g/l was used to reduce tibia loads and stabilize the contacts between the surrogates and the floor. The material characterization was done internally in IDIADA.

The geometry of the knee bolster was maintained from the original model. However, the foam material was changed to an EPP of 60g/l to recreate a stiffer dashboard that can apply higher loads to the occupant's femurs.

The belt system elements were adapted to the new seatback configuration. The D-ring was positioned close to the seatback, simulating a belt-in-seat mount. The retractor was positioned right below the D-ring to emulate a physical testing routing. The firing parameters of the belt remained the same as in the original model. The 3-point belt system consisted of a shoulder belt retractor with two load limiters (3.5 kN and 10 kN) and 2 kN pre-tensioners, a 2 kN lap belt pre-tensioners, and a crash locking tongue.

The steering column and the belt system remained unchanged from the original model. The collapsible column had a force level of 4.5kN with 100mm of maximum stroke.
The updated environment model is presented in Figure 4.

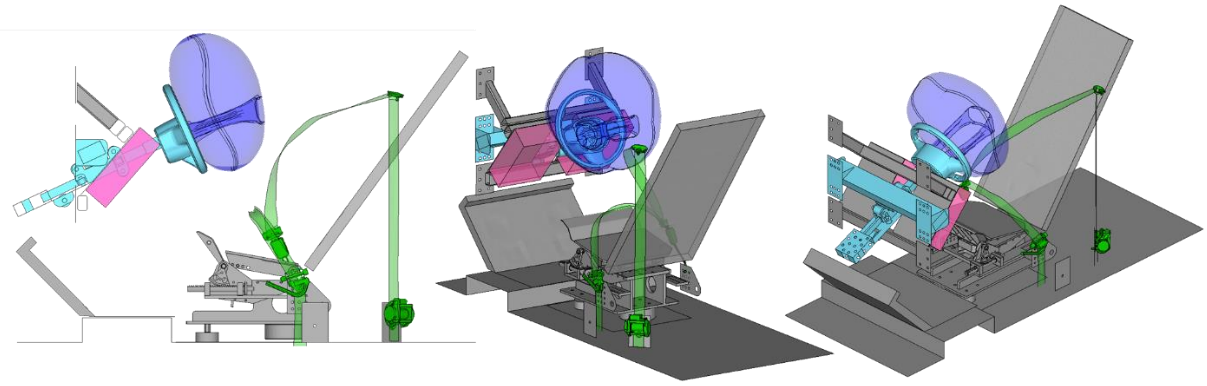


Figure 4: Adapted generic frontal model used in the study. In grey: the semi-rigid seat, seatback, foot support, and structures. In light blue: collapsible steering column and steering wheel. In dark blue: driver airbag. In pink: knee bolster. In green: seatbelt system.

Occupant Positioning and Belt Routing

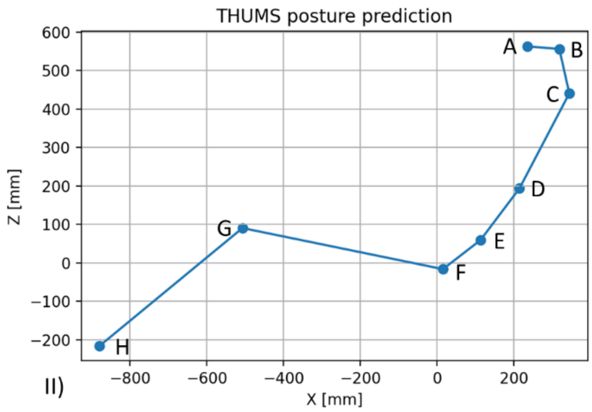
The study used the THUMS model as the main finite element HBM. This model represents a 50th-percentile male with a stature of 178.6 cm and a weight of 78.5 kg and was used as the baseline model. THUMS was positioned via a (pre)simulation and the THOR-Reclined ATD was positioned based on the achieved posture of the FE HBM. The two models were positioned in a reduced environment with just the semi-rigid seat plates (which were considered rigid), the seatback, and the foot support.

First, the final posture was estimated based on the UMTRI 2018 study [9]. The anatomical landmarks (ankle joint, knee joint, acetabulum joint, L5/S1 joint, T12/L1 joint, C7/T1, Head/C1 joint, head and center of the eye) were calculated using the complete regression model including the anthropometric predictors using Python v3.10.6 (Python Software Foundation, Beaverton, USA). The values used for the posture estimation and the obtained landmarks are shown in Figure 5.

I)

Variable	Value
Stature (mm)	1786
Weight (kg)	78,5
Erect sitting heihgt (mm)	893
Seatback angle (deg)	45
Presence of headrest (-)	0

II)



Letter	Anatomical landmark
A	Center of eye
B	Head/C1
C	C7/T1
D	T12/L1
E	L5/S1
F	Acetabulum joint
G	Knee
H	Ankle

Figure 5: I) Variable values used for the posture estimation, and II) coordinates of the anatomical landmarks obtained if the seat H-point was placed in the (0,0)

To position the occupant model, a (pre)simulation was run using the marionette method [10]. The previously mentioned anatomical landmarks were used as targets to achieve the desired pre-impact posture. The knee and ankle landmarks were modified so the leg's posture fitted the simulation model used for this study. Regarding the arms, they were placed in line with the torso and the hands were placed in contact with the seat to obtain an achievable posture for the THOR-Reclined dummy model. The THUMS nodes used for this (pre)simulation are shown in the following Table.

Table 1: Node IDs used for each anatomical landmark

Anatomical Landmarks	NODE ID	Anatomical Landmarks	NODE ID
Left Ankle	82002574	C7/T1 joint	89000708
Right Ankle	81002574	Left Shoulder	86007088
Left Knee	82074216	Right Shoulder	85007088
Right Knee	81074216	Left Elbow	86006275
Left Acetabulum	83001318	Right Elbow	85006275
Right Acetabulum	83501318	Left Wrist	86003794
L5/S1 joint	89066294	Right Wrist	85003794
T12/L1 joint	89065464	Head/C1 joint	87000782

The positioning was performed based on the work presented by Alexandros Leledakis et al. [11]. A two-step (pre)simulation was used to position the model. The first stage had a duration of 450 ms. During the first phase, one-dimensional elements were used, applying a force from 0 to 500 N to position the model. Simultaneously, the geometrical constraints of the generic environment that had contact with the HBM were moved to their original position. These surfaces (the anti-submarining plate and seat plate of the semi-rigid seat, the footrest, and the seatback) were originally moved 150 mm away from their original position in X and Z directions. The second stage had a duration of 300 ms. In this phase, the one-dimensional elements force was set to 0 and the model was allowed to reach equilibrium. Gravity was activated throughout the complete simulation and a global damping of 0.15 was used. This process is illustrated in Figure 6.

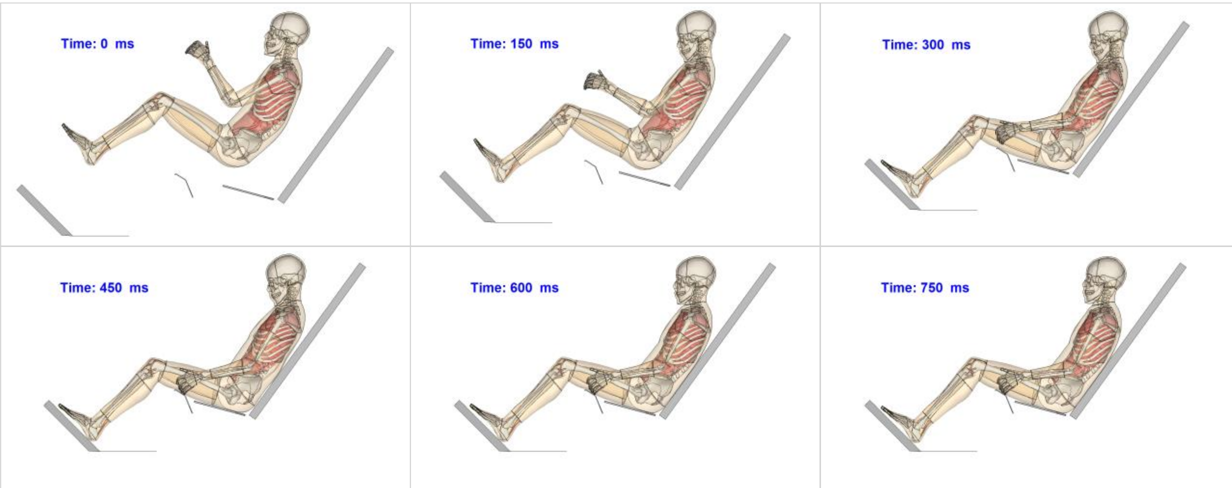


Figure 6: HBM positioning (pre)simulation. The environment elements (in grey) start 150 mm away from their original position and are repositioned during the first stage of the simulation. The last 300 ms are used to reach equilibrium in the model.

The position of the nodes of the HBM, the footrest foam, and the seatback foam were retained for the impact simulation. Foam and internal HBM stresses were not retained.

The same procedure was applied to the THOR-Reclined dummy model. In this case, the reference was the achieved posture of the HBM, so the dummy was positioned as close as possible to the THUMS. Due to the differences between both models, priority was given to the similar positioning of the internal structure of the dummy and the HBM skeleton, starting from the iliac spines. The dummy was (pre)simulated using the same simulation process of the HBM as can be seen in Figure 7.

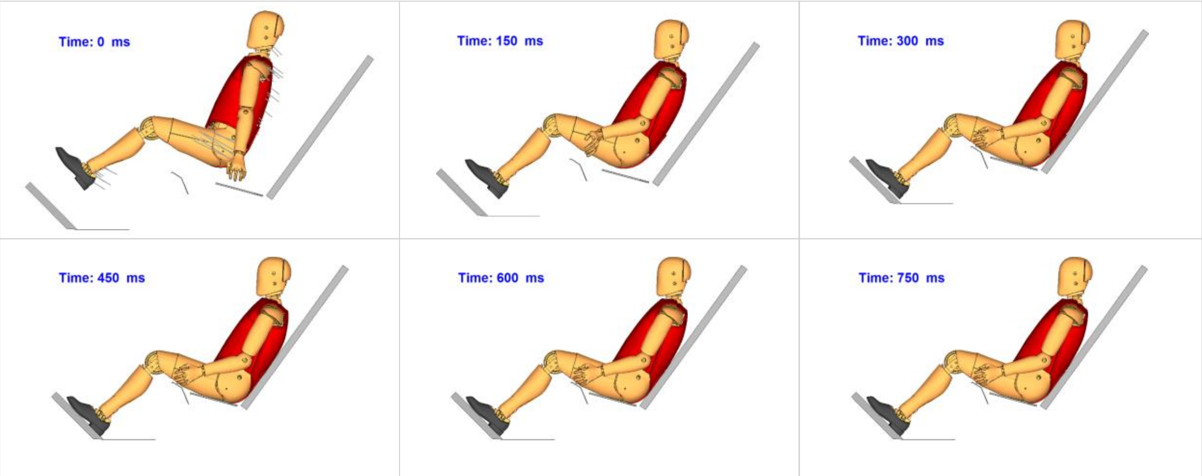


Figure 7: ATD positioning (pre)simulation. The environment elements (in grey) start 150 mm away from their original position and are repositioned during the first stage of the simulation. The last 300 ms are used to reach equilibrium in the model.

Following this method, a comparable posture of the ATD was obtained. A comparison between the posture achieved with each model is shown in Figure 8.

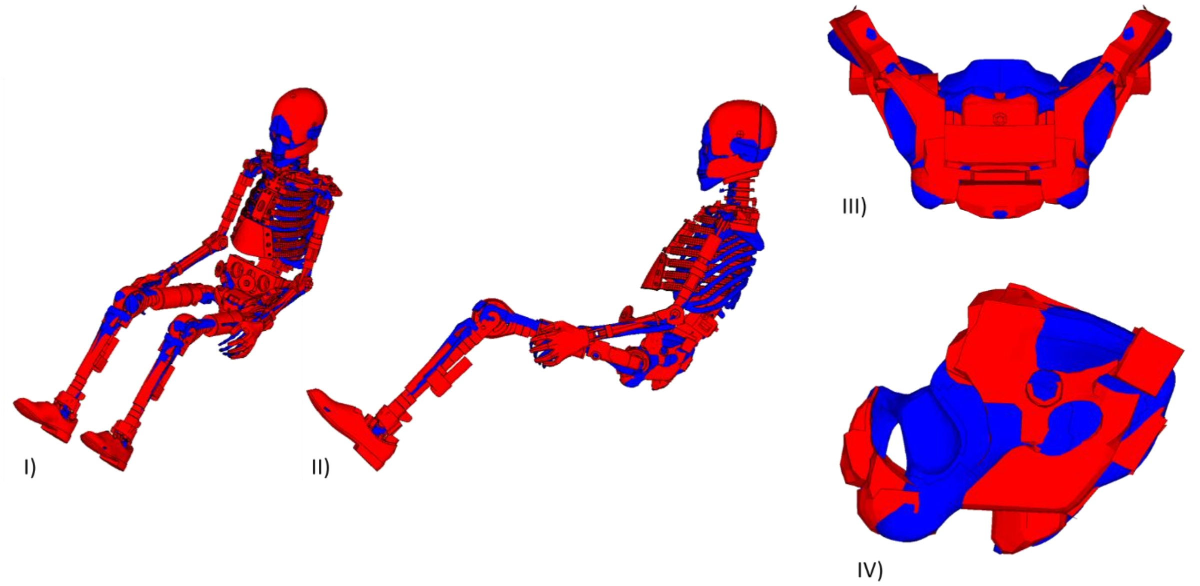


Figure 8: Comparison of the achieved postures of the HBM (blue) and the ATD (red). I) Isometric perspective. II) Lateral view. III) Frontal view of the pelvis of both models. IV) Lateral view of the pelvis of both models.

Regarding the belt routing, in both cases the shoulder belt was positioned following a straight line between the D-ring and the belt tongue, using the shortest path possible that allowed the belt to pass through the middle part of the collarbone of each model. The lap belt was positioned following a straight line between the buckle and the end bracket of the belt, placing the webbing in the lowest part of the abdomen possible for each model (Figure 9). No initial pretension was given to the belt.

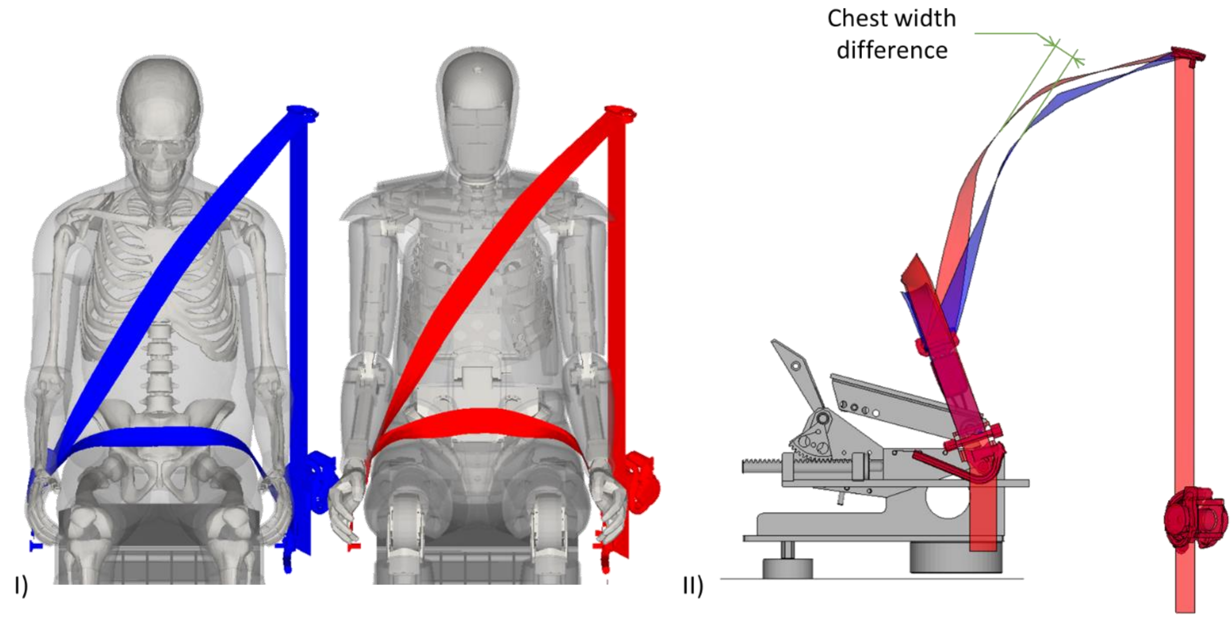


Figure 9: Belt fit comparison of both models. The THUMS belt is presented in blue and the THOR-Reclined belt is presented in red. I) Side-by-side comparison of belt fit over each surrogate. II) Lateral view of the belt and the semi-rigid seat. A gap between both belt models can be observed due to the difference in the chest width of both models.

Crash configuration

The full frontal 56km/h Car-to-Car (C2C) crash pulse from EU project OSCCAR [12] was used for this analysis. The characteristics of this pulse are presented in Figure 10. The pulse was chosen due to its high severity, as this would highlight the similarities and differences that may exist between the HBM and the ATD.

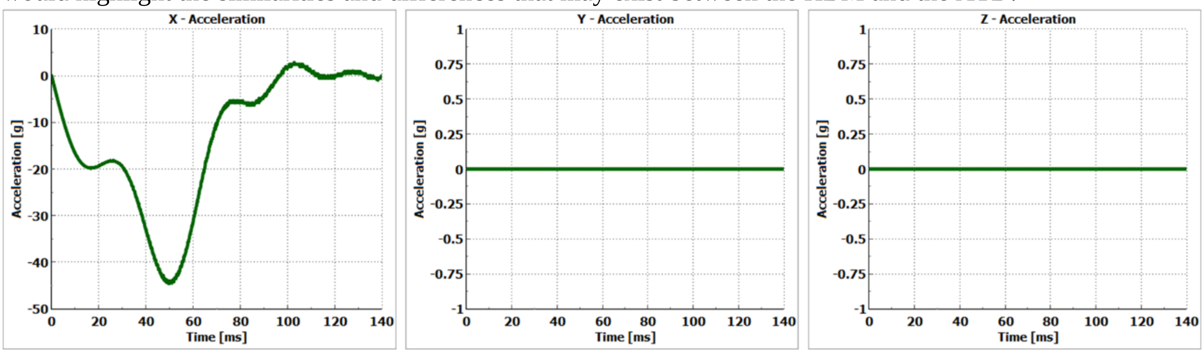


Figure 10: Full frontal 56 km/h C2C crash pulse from EU project OSCCAR

Analysis methods

Both simulation models were compared using surrogate kinematics, restraint system outputs, injury risk prediction, and visual inspection.

Regarding kinematics, the head center of gravity (CG), T1, T4, T12, and pelvis Anterior Superior Iliac Spine (ASIS) kinematics were compared between both models.

Seat belt forces, webbing pay-outs at the retractor, belt tongue slip, webbing pay-ins at the end bracket pre-tensioner, DAB pressures and volume changes, steering column strokes and forces, and rotations of anti-submarining and seat pan plates were chosen as the main restraint system outputs to use for the comparison.

To evaluate the injury risk, outputs were defined for the HBM following the recommendations of the OSCCAR deliverable D3.3 [13]. Regarding the ATD, the standard outputs were used for the analysis, and injury risk criteria was evaluated based on [14]. For the head, HIC and BrIC injury risks were evaluated according to [14]. DAMAGE was as well evaluated based on [15]. Regarding the neck, cross sections were defined in the cervical vertebrae of the HBM (C1-C7) to analyze axial loading through cortical and spongy bones, left and right transverse processes, spinous process, and the ligaments connected to the respective vertebrae according to [13]. The maximum values

of forces from all the vertebrae were then compared with the maximum loads measured by the load cells of the upper and lower sections of the neck of the ATD. Regarding the thorax, rib fracture risk was assessed according to [13] [16] for the HBM using cortical bone maximum principal strain. This risk of rib fracture was then compared to the peak resultant chest deflection injury criterion for the ATD [14]. The anterior superior iliac spine (ASIS) peak force was also measured in the HBM by cross sections through cortical and spongy bone to assess iliac wing fracture risk. This load was then compared to the one measured in the THOR-Reclined ASIS load cells. The leg injury was assessed by femur force measurement. A comparison between the recorded values of the ATD's load cells and the HBM cross sections was made.
Submarining and overall behavior of both models in the generic frontal environment simulation model were assessed based on visual inspection.

RESULTS

The two FE (pre)simulations and the two impact simulations were performed successfully. All the simulations were checked regarding simulation quality. In all the simulations, the percentage of added mass was below the 5% limit and the hourglass energy remained below the 10% of the internal energy maximum of each simulation. The four simulations reached normal termination.

Surrogate kinematics

Head CG, T1, T4, T12 and left pelvis ASIS trajectories were compared between both models. Some differences and similarities were found between both models, which are presented in Figure 11.

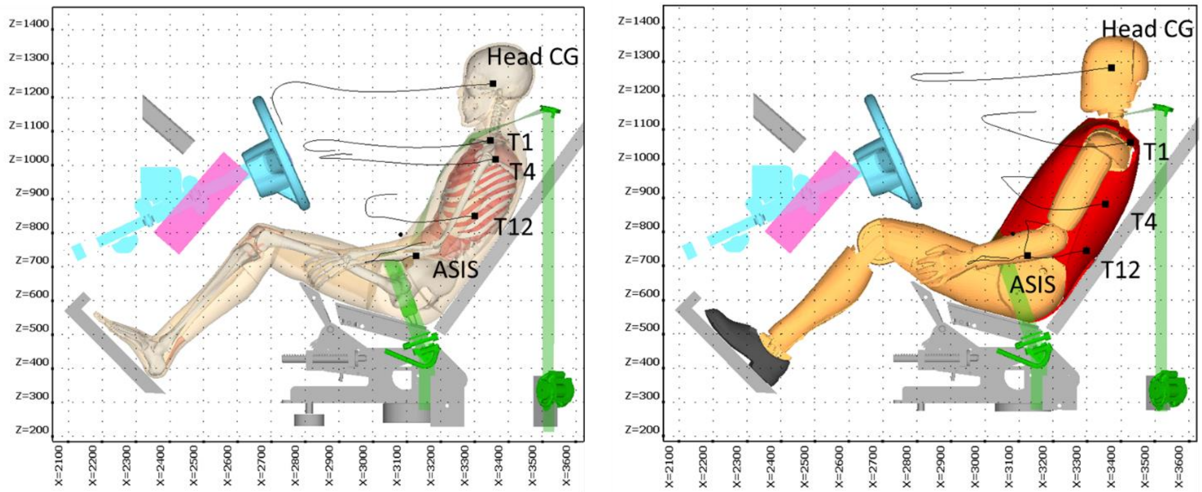


Figure 11: Kinematic comparison of THUMS (left) and THOR-Reclined (right) simulation models during the impact simulation.

Torso and head kinematics differed significantly between both models. THUMS had higher forward displacements, while the THOR-Reclined was stiffer and longitudinal displacements were smaller. Also, several differences were found regarding initial sensor positions between the ATD and the HBM. T4 and T12 dummy sensors had different positions compared to the initial position of the HBM T4 and T12. Nevertheless, the ASIS kinematics during the loading phase was very similar between both models, reaching a maximum of around 140 mm of forward displacement. Overall body movement during the first 80 ms is very similar (Figure 12) between both models as well. At 102 ms, the shoulder belt of the THUMS model wraps and slips from the shoulder of the surrogate. This does not happen to the THOR-Reclined simulation model and therefore the kinematics from that moment differ significantly.

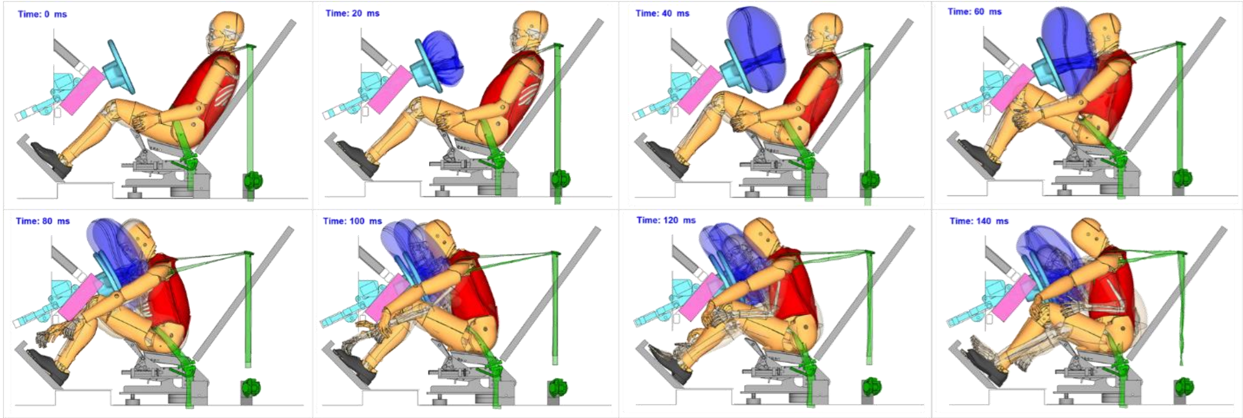


Figure 12: Visual comparison between both models during the impact simulation

Restraint system outputs

The restraint system had a comparable performance for both surrogates. Figure 13 shows a very similar behavior of the belt regarding both forces and payouts. The shoulder belt tension of the THUMS model decays a few milliseconds earlier than the THOR-Reclined due to the slippage of the belt. Also, higher forces are measured in section B6 for the THUMS model. The seat plate rotations are very similar for both models as well. The steering column and the DAB however show different behaviors in each simulation. The maximum collapsing distance of the SC for THUMS is 100mm, while the THOR-Reclined dummy reaches a maximum of 58 mm. The DAB registered higher pressure for the THUMS simulation.

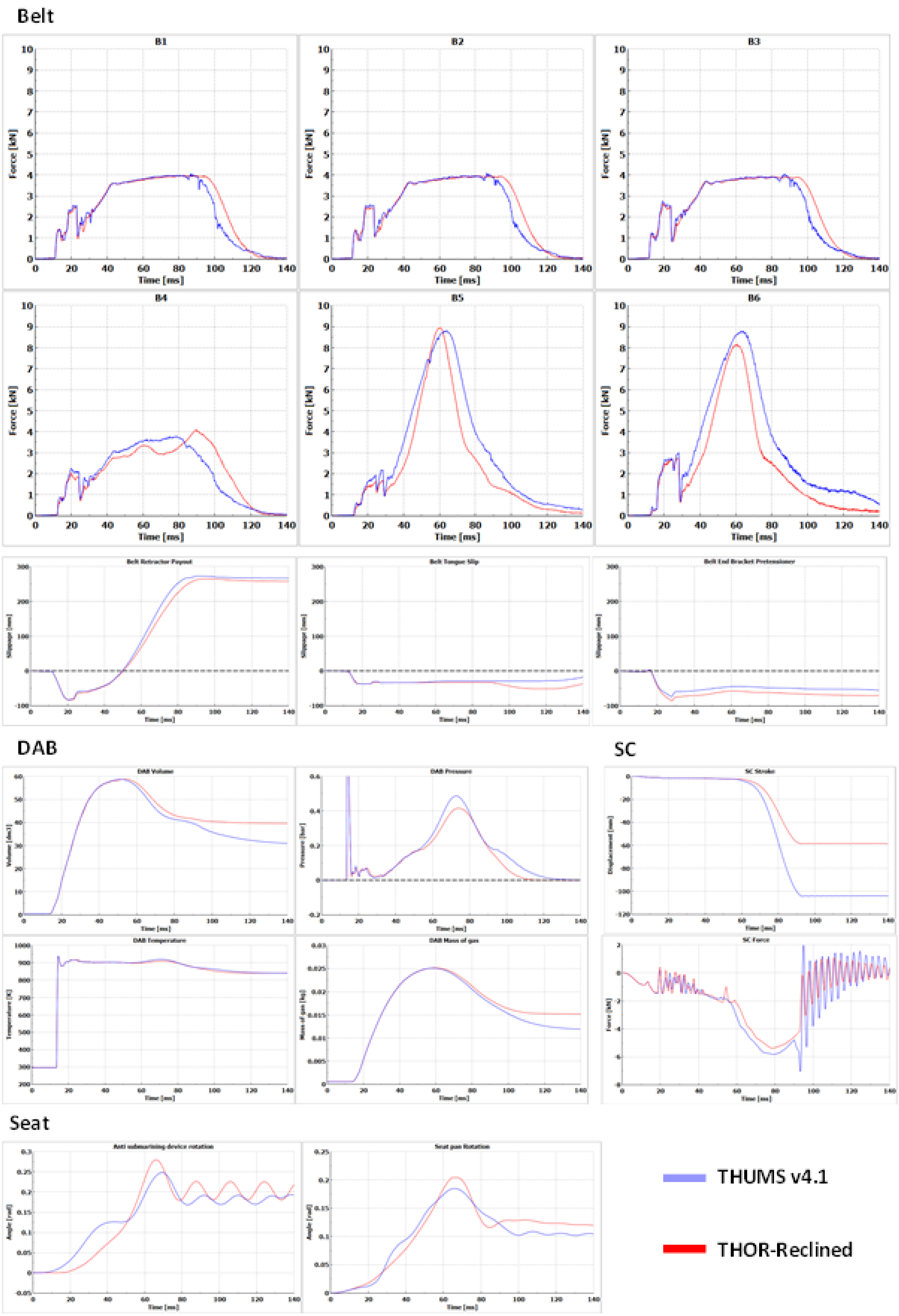


Figure 13: Restraint system comparison

Injury risk prediction

For the head, HIC 15, BrIC, and DAMAGE injury values were calculated based on each model head kinematics. The injury risk functions recommended by [14] for HIC and BrIC (based on cumulative strain damage measure) were as well applied. In general, values differed between both models. THUMS was always the model with higher injury values. The closest injury value between both models was the DAMAGE, with 0.543 for THUMS and 0.343 for the THOR-Reclined.

Regarding the neck, the THUMS model registered the maximum tension force of 2.02 kN in C3 at 85 ms. The THOR-Reclined registered a maximum tension force of 0.3 kN at 78 ms in the upper cell of the neck. Maximum tension was obtained in similar places of the neck at approximate times. The peak values obtained however differed by almost 2 kN.

Thorax compression was measured in the THOR-Reclined dummy in the upper-left, lower-left, upper-right, and lower-right IR-TRACCs, obtaining a peak resultant chest deflection of 49.13 mm in the upper-left section of the chest. This deflection translates into a 45.36% probability of AIS 3+ injury. In the case of the HBM, the maximum principal strain of the rib cortical bone was measured. Strain values were obtained at a frequency of 5000 Hz (every 0.2 ms) and elements with a strain rate higher than 0.04 ms^{-1} were eliminated from the analysis as suggested by [13]. Results are presented in Table 2. The risk of two or more rib fractures for a 65-year-old person resulted in a close value to the one predicted by the THOR-Reclined chest deflection.

Pelvis and femur forces were compared, showing similar results in the left iliac wing and the right femur (Table 2).

Table 2: Injury values obtained for each simulation model

		THUMS	THOR-Reclined
HEAD	HIC 15 [-]	1942,240	487,624
	BrIC (CSDM) [-]	1,008	0,524
	DAMAGE	0,543	0,343
	HIC P(AIS 2+)	76,36%	18,03%
	HIC P(AIS 3+)	56,40%	4,40%
	BrIC P(AIS 3+)	57,24%	0,00%
	BrIC P(AIS 4+)	44,86%	0,00%
THORAX	Peak resultant chest deflection [mm]		49,130
	Chest P(AIS 3+)		45,36%
	Risk for rib fractures 2+ (45 yr)	4,59%	
	Risk for rib fractures 2+ (65 yr)	48,28%	
PELVIS	Left iliac wing peak compression [kN]	3,810	3,524
	Right iliac wing peak compression [kN]	3,657	4,915
FEMURS	Left femur peak compression [kN]	1,083	0,212
	Right femur peak compression [kN]	0,450	0,204

LIMITATIONS

The following aspects were not included in the study and may have affected some of the results here obtained:

- Only one ATD was used for the analysis, so results do not represent all the dummy population.
- Only one FE HBM was used for the analysis. Further work needs to be done to get a general overview of the outputs generated by each HBM.
- One reclined posture was used for the study. More seating postures with different seat pan and seatback angles may be necessary to do a full comparison between the THUMS and the THOR-Reclined.

- No physical tests have been performed, in the scope of this work, to correlate the dummy FE model behavior. These will be done in the near future and the results will be published accordingly. In addition, further validation test results (found in the literature) will be used for the analysis.
- A simplified belt material card (*MAT_SEATBELT) was used for the FE simulations. A more detailed material definition like *MAT_FABRIC could help clarify the difference in shoulder kinematics, as the wrapping of the seatbelt could be avoided.
- No optimization of the restraint system was made for the studied posture. A SOTA restraint system was used with the only objective of comparing both simulation models in the same conditions.

CONCLUSIONS

A one-to-one comparison was made between the THOR-Reclined dummy and the THUMS HBM regarding kinematics, restraint system outputs, and injury criteria. Some differences have been found between both models, mainly due to their physical dissimilarities but also in the way both models are designed to output results. While the ATD is mainly designed for generating acceleration and displacement results, the HBM could not only offer acceleration and displacement outputs but stress and strain outputs as well.

Results suggest that the THOR-Reclined ATD has a stiffer behavior compared to the HBM. Low deformations were observed in the dummy compared to the HBM. One main difference was found in the shoulder behavior. The HBM generated a concave area around the shoulder that provoked the wrapping and slippage of the shoulder belt, changing the kinematics of the surrogate towards the end of the simulation. The THOR-Reclined however did not show that deformation and the shoulder belt stayed in position during the simulation, even though both models had comparable seat belt routings. Both models had similar pelvis kinematics and no submarining was observed in any simulation.

Regarding restraint systems, even though the seat behaved similarly, and the belt loads were comparable in both impact simulations, higher loads were transferred to the DAB and the SC in the HBM simulation, as higher pressures were obtained in the airbag and a larger stroke in the SC. This higher transfer of loads might have occurred due to the shoulder belt slippage, the more flexible behavior of the HBM, and the physical dissimilarities of both models.

THUMS consistently sustained higher injury values and injury risk predictions compared to the ones obtained by the THOR-Reclined. The biggest differences were found in the head. The HBM showed injury risk predictions of over 50% (regarding BrIC) while the ATD had a 0% of injury risk.

These results suggest that while nowadays cars are being rated with dummies, properly restraining a human body might be more challenging, as in the same environment the THUMS model has shown to have larger displacements and strains, worse coupling with the seat belt, and higher injury outputs compared to the THOR-Reclined.

This study offers the first direct comparison between a HBM and a dummy. Nevertheless, further work needs to be done to be able to fully compare both models regarding postures and restraint systems. Also, the addition of more models to the study could be beneficial to fully understand the similarities and differences between ATDs and HBMs to be able to properly improve and test restraint systems and their impact on real humans.

FUTURE WORK

Currently, the authors are undergoing a similar study using the MADYMO Active Human Model (AHM). This work will be compared with the results from this study to increase the number of studied models.

Physical tests will be performed shortly to correlate the dummy FE model behavior and the environment used for this study. This future work will as well be published accordingly.

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