

Validation of a Human Body Model for Frontal Crash and its Use for Chest Injury Prediction

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Autoliv Research

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ABSTRACT

Whole-body kinematics of the finite element human body model THUMS was evaluated by means of sled tests. A model of a crash test dummy (Hybrid-III 50%-ile) was used to validate the test environment by matching the model predictions to the experimentally measured dummy sled test responses. Once the environment was validated, the THUMS model was placed in the sled model and the post mortem human subject (PMHS) sled tests were replicated.

Two test configurations were used for the evaluation. One configuration was high impact velocity sled tests with an advanced restraint system. The other configuration was low impact velocity sled tests with a basic restraint system. The test velocities were 48 km/h and 29 km/h respectively.

The evaluation was carried out by an objective rating method that compared predictions from the model to results from the mechanical tests. The method assessed the peak level, peak timing and curve shapes of the predictions relative to the test results.

The mathematical crash dummy model predictions matched the mechanical counterpart responses well, confirming that the test environment was replicated. The match between the THUMS model predictions and PMHS responses was not as good.

Generally the chest deflection predicted by THUMS was greater than the chest deflection predicted by the PMHS. There was also a discrepancy between model prediction of chest injury and PMHS injury.

INTRODUCTION

In automobile crashes thoracic injuries are very common. Thoracic injuries rank second only to head injuries in three categories; 1) the thorax is the second area most often injured [1, 2], 2) chest injuries results in the second highest number of fatalities and serious injuries [3], 3) chest injuries have the second highest overall societal harm [4]. It has been shown that thorax

injuries account for approximately 13% of all AIS 1-2 injuries and 29% of all AIS 3-6 injuries [2]. Rib fractures and flail chest are the most frequent types of thoracic injuries for both drivers and passengers, followed by pulmonary, liver and arterial injuries. The prevention of these injuries remains a major concern.

There is also a need to study human thoracic response to dynamic loading under various impact conditions. Such conditions involve loading through restraint systems, e.g., blunt loading from airbags and/or steering system, seatbelt alone or in combination with airbags. Thoracic response under various loading conditions has been evaluated using post mortem human subjects (PMHS). These studies have been time consuming and the vast majority of them have been conducted with PMHS's representing the older population. Thoracic response evaluations can be done more efficiently with finite element (FE) models of the human body. The mathematical human body models can improve our understanding of injury mechanisms and predict injury risks. The models can enable evaluation of physical variables mechanically related to injury, e.g. strain [5]. Also, the models can be used to evaluate complex chest deformation patterns that can not be assessed using mechanical anthropometric test devices (ATD:s) with a single-point slider as in used in the Hybrid-III dummy. It was found in testing with post mortem human subjects (PMHS) loaded by a single diagonal belt that max chest deflection was observed at points other than mid sternum [6].

There is a considerable number of mathematical human body models developed. Such models are the THUMS model (Total Human Model for Safety), the MADYMO human body model, the H-model etc. These models are developed to predict gross motions and stress or strain distributions of the whole human body at impacts. The models reproduce anatomical geometry and biomechanical properties of the human body such as the stiffness of the bone and the flexibility of the skin. Each joint is modeled anatomically with the major ligaments and tendons.

The modeling approach used for the chest is similar in most human body models. In the human body model THUMS the ribs are modeled using a combination of solid and shell elements. The trabecular (spongy) bone is modeled using one solid element through the thickness of the rib. The cortical bone is modeled with shell elements. The material model used for the trabecular bone is an elasto-plastic material with a stress vs strain curve defined. The material model used for the cortical bone is an elasto-visco-plastic material model with a stress vs. strain curve defined. No strain-rate dependency is defined. However, there is a failure criterion assigned to the cortical bone. The column vertebrae are modeled with rigid bodies connected by spring elements. The intervertebral disks are modeled with an elastic material model. The ligaments are represented by spring elements. The sternum is modeled with a similar material model as the ribs and the cartilage is modeled using an elasto-plastic material model with a failure criterion defined. The spino-costal joint is defined with spring element and the sterno-costal joint is defined with solid elements.

Human body models have been validated by means of sled, impactor and belt compression tests over range of impact energies [7]. Various studies for the validation of different parts of the THUMS model have also been reported [8, 9, 10]. The cervical spine model was validated for both frontal and rear loading conditions. It was found that the model predicted mechanical test results in some load cases and in some other load cases the model predictions were not correlated to the mechanical test results [11]. However, it is not likely that a component level test alone can evaluate the biofidelity of a model on whole-body level. The reason is that there are a large number of structures in the body, each structure has multiple degrees of freedom and the loading pattern is often poorly defined. Therefore there is a need to evaluate the biofidelity of whole-body kinematics predictions in a restraint and collision environment.

Few tools in mathematical crash modeling have been developed to objectively evaluate the ability of a mathematical model to predict experimental test results. Such a tool can enable an objective rating of a mathematical model. The quality values resulting from the rating procedure provides a judgment of the performance of the dummy or human body model in its environment. The quality values states how "good" the simulation model is. One such tool uses three criteria to evaluate the quality of the signals: the peak criterion, the peak timing criterion and a signal trend criterion.

The objective of the study is to assess the whole-body kinematics of the total human model for safety (THUMS) finite element model relative to post mortem human subject (PMHS) response. An extended aim is to evaluate the ability of THUMS to predict the risk for an occupant to sustain chest injuries.

METHOD

The 50%-ile Hybrid-III finite element model was used to validate the test environment (restraint system and seat) by matching model predictions to experimentally measured Hybrid-III sled test responses. Once the environment was validated, the THUMS model was placed in the sled model and the PMHS sled tests were replicated. The predictions from the models were compared to the responses from the mechanical tests and objective rating scores were computed.

SLED CONFIGURATIONS – Two test configurations were used for the evaluation. One configuration was driver side high impact velocity sled tests with advanced restraint systems. Five tests with PMHS and three tests with 50%-ile Hybrid III dummy were carried out. Data from these tests were presented by Shaw et al. (2000) [12]. The other configuration was passenger side low impact velocity sled tests with basic restraint systems. Three tests with PMHS and three tests with 50%-ile Hybrid III were carried out. Data from these tests were presented by Forman et al. (2006) [13]. The advanced restraint system consisted of an airbag in combination with a pretensioned and force-limited belt system. The basic restraint system had no airbag and used a belt system without pretensioning and load limiting.

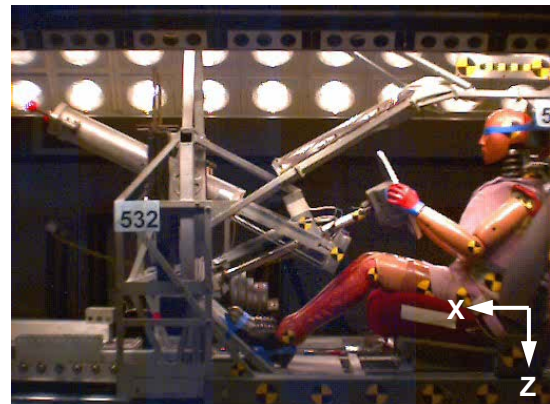


Figure 1: Driver side mechanical and mathematical Hybrid-III 50%-ile dummy.

The driver side evaluation was carried out using a buck representing the interior of a 1993 model-year mid-sized sedan (Figure 1 and Figure 2). For all driver-side tests there was a pretensioner at the buckle and the seatbelt was force limited to nominally 3.5 kN. Prior to test the belt was preloaded (tightened) by hand.

to achieve approximately 9 N of tension in the shoulder belt (between the shoulder and the D-ring). The driver side used a non depowered driver's side airbag. A windshield and a generic energy absorbing knee bolster were also used on the driver side.

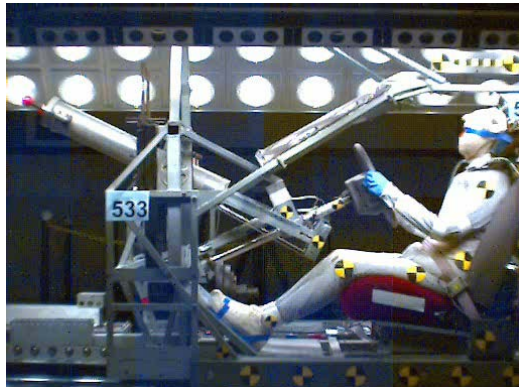


Figure 2: Driver Side PMHS and THUMS.

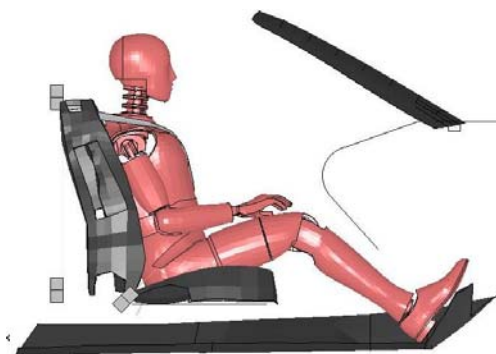


Figure 3: Passenger side mechanical and mathematical Hybrid-III 50%-ile dummy.

All passenger-side tests were conducted with the occupant placed in the right front passenger position of a

1997 mid-sized sedan buck (Figure 3 and Figure 4) [14]. A standard (not pretensioned nor force limited) belt was used. The passenger side test included an OEM instrument panel. All tests (on both driver and passenger side) used seats whose frames and anti-submarining pans were reinforced to allow multiple uses with minimal seat and seat-pan deformations.

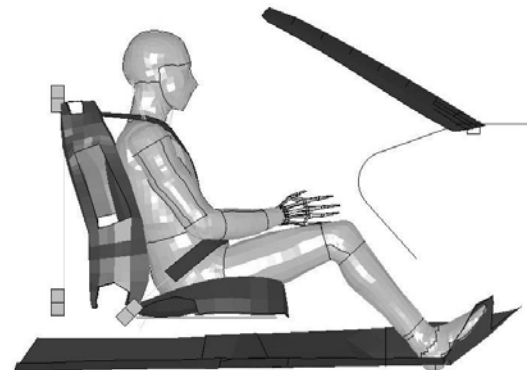
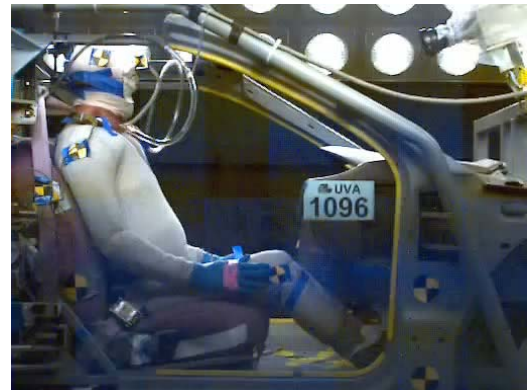


Figure 4: Passenger side PMHS and THUMS.

In the driver and passenger side tests the impact velocities were 48 km/h and 29 km/h respectively. The acceleration pulses were trapezoidal in shape. For the driver side the pulse was chosen to represent the vehicle decelerations experienced in offset frontal barrier tests of a mid sized sedan. The stopping distances were 0.93 m in the 48 km/h test and 0.45 m in the 29 km/h test (Figure 5).

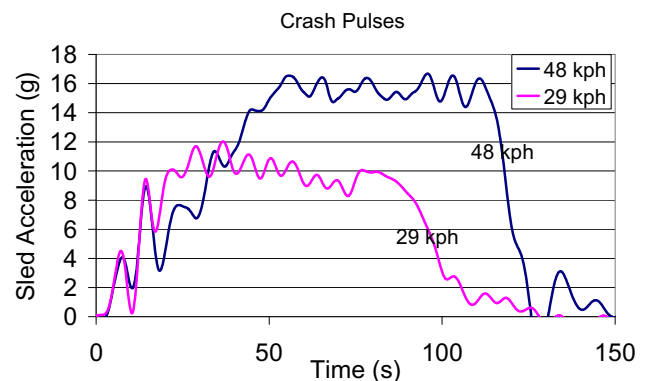


Figure 5: Sled Acceleration

Finite element models of the driver side and passenger side of the sled buck were developed. The test

configuration included the seat, the seat belt and the airbag (driver side only). For validation of the test configuration, the mechanical sled tests with the 50%-ile Hybrid-III dummy were replicated with the finite element model, where the FTSS 50%-ile Hybrid-III version 5.1S1 model was used. The predictions from the model were compared to the response from three mechanical sled tests. Once a satisfactory level of validation of the test configuration was obtained an objective rating score was computed and the Hybrid-III dummy model was replaced with THUMS (Version 2.21-040407).

All PMHS response data used in the evaluation of THUMS was obtained from a study by Forman et al. [15]. In that study the response data was calculated from the results from five driver side sled tests and three passenger side sled tests. The PMHS data from the sled tests was scaled to a 50%-ile male using a variety of methods. For sets of data that exhibited similar shapes and timing, response envelopes were defined as the region enclosed by the upper and lower bound of the data set. For these data sets, average response curves were defined as averages of the upper and lower bounds. For sets of data that exhibited dissimilar shapes or dissimilar timing, characteristic average responses were calculated by normalizing each data curve in time by a characteristic time (e.g. time of peak value), calculating an average curve from the set of normalized curves, and de-normalizing this to obtain a characteristic average curve [16]. These average responses were used for the evaluations of THUMS. For the Hybrid-III response data, the average signals based on three sled tests on the driver side and three tests on the passenger side were used.

In this study, the signals according to Table 1 were included the evaluation. The sign conventions follow SAE/J1733 for the occupant local accelerometers and forces. The occupant displacements were derived by film analysis in which the reference frame was moving with the sled buck. In addition to the signals in Table 1, each PMHS was also equipped with two chestbands. The upper band was wrapped around the chest at the level of the 4:th cervical rib and the lower chest band at the 7:th cervical rib. For the chest band evaluation, only one test from each PMHS load case was used.

OBJECTIVE RATING METHOD – The assessment of the predictability of mathematical models involves comparisons of a large number of time history signals to the corresponding mechanical test results. This kind of evaluation benefits from a systematic and objective methodology which can identify weak as well as strong prediction capability of a model, as well as deriving a single or a few quality values. In this study, the objective rating method (ORM) [17] was used to asses the quality of the model predictions relative to the mechanical tests. The method evaluates the correlation between two signals by scalar values, such as peaks and peak times, and curve shapes (Figure 6). A complete ORM value for each of the four load cases was also calculated based on the individual scalar and shape ORM values.

Table 1: Compared signals of the Hybrid-III and THUMS.

Type	Hybrid-III	THUMS
Accelerations	Head X-Acc	Head X-Acc
	Head Z-Acc	Head Z-Acc
	T1 X-Acc	T1 X-Acc
	T1 Z-Acc	T1 Z-Acc
	Chest X-Acc	T8 X-Acc
	Chest Z-Acc	T8 Z-Acc
	Pelvis X-Acc	L2 X-Acc
	Pelvis Z-Acc	L2 Z-Acc
Displacements		Pelvis X-Acc
		Pelvis Z-Acc
	Head X-Disp	Head X-Disp
	Head Z-Disp	Head Z-Disp
	T1 X-Disp (dr side only)	T1 X-Disp (dr side only)
	T1 Z-Disp (dr side only)	T1 Z-Disp (dr side only)
	Pelvis X-Disp	Pelvis X-Disp
	Pelvis Z-Disp	Pelvis Z-Disp
Forces		Knee X-Disp
		Knee Z-Disp
	Femur Z-Force Left	Belt Diagonal Force Upper
	Femur Z-Force Right	Belt Diagonal Force Lower
	Belt Diagonal Force Upper	Belt Lap Force Lower
	Belt Diagonal Force Lower	
	Belt Lap Force Lower	

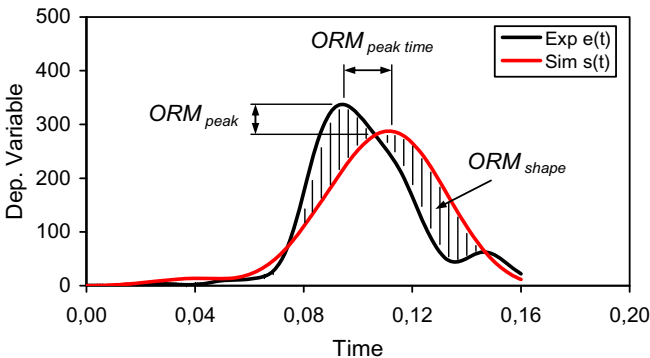


Figure 6: Objective Rating Method - ORM

In this study, the signals according to Table 1 were used for the ORM evaluation. The signals were grouped in accelerations, displacements and forces. In addition to the occupant response signals, also belt forces at three locations were included in the ORM evaluation; at the diagonal belt between the occupant shoulder and the D-ring, at the diagonal belt above the buckle and at the lap belt on the outer sill side. On the passenger side, the T1-displacements were not available for both the Hybrid-III and the PMHS and were therefore not included in the ORM evaluation. The ORM scalar and shape comparisons were limited to the time frame of 0-150 ms, except for the displacement comparisons for the Hybrid-III passenger head and pelvis together with the THUMS driver head, pelvis and knee, all which were available only up to 140 ms. The filter SAE180 was applied to all acceleration signals and SAE600 to all force and chest deflection signals. No filtering was applied to the displacement signals. Also, no signals in the lateral y-direction were included in the evaluation due to the pure

zero degree sled trajectory and the dominant x- and z-signals of the occupant trajectory.

The scalar ORM values were determined based on the *Factor Method* according to Equation 1. The resulting comparison between the experiment (*exp*) and the simulation (*sim*) range between 0 and 100%, where 100% corresponds to a perfect match.

$$ORM_{scalar} = \frac{\max(0, exp \cdot sim)}{\max(exp^2, sim^2)} \quad (1)$$

The correlation of the curve shapes between the experiment ($e(t)$) and the simulation ($s(t)$) was determined based on equation 2 which is called the *Weighted Integrated Factor Method*. In this method, a scalar value is calculated according to Equation 1 for each point of the curve. Also, a weight factor, based on the maximum value of the experimental and simulation signal at each point, is applied. The sum of all points was then determined using the *Root Mean Square Addition Method*.

$$ORM_{shape} = 1 - \sqrt{\frac{\int \left(\max(|e(t)|, |s(t)|) \cdot \left(1 - \frac{\max(0, e(t) \cdot s(t))}{\max(e(t)^2, s(t)^2)} \right)^2 dt \right)}{\int \max(|e(t)|, |s(t)|) dt}} \quad (2)$$

The ORM value for the complete system was calculated according to equation 3, where the root mean square addition method was applied to all scalar values of the system. W was a weight factor for each scalar value. In this study the weight $W=1.0$ was applied to all scalar values.

$$ORM_{complete} = 1 - \sqrt{\frac{\sum W_{scalar \text{ or } shape} \cdot (1 - ORM_{scalar \text{ or } shape})^2}{\sum W_{scalar \text{ or } shape}}} \quad (3)$$

The end time was defined to be the time when all peak values during the loading phase of the sled tests were reached.

RESULTS

ORM – DRIVER HIGH SPEED – The evaluations of the driver side high speed load cases with the Hybrid-III and THUMS models were carried out using the objective rating method. The signal curves of the two load cases can be observed in Appendix 1 and Appendix 3.

For the Hybrid-III driver side, the ORM values for the peaks and peak occurrence times can be found in Figure 7. The ORM x-acceleration peak values ranged from 83 to 100% with a median value of 97%, while the z-acceleration peak values ranged from 23 to 52% with a median value of 39%. When considering the displacements, higher ORM values were also observed in the x-direction (range 70-94%, median 84%)

compared to the z-direction (range 38-82%, median 47%). The chest deflection ORM peak value was 95% and ORM peak time value was 76%. When considering only the acceleration and displacement signals in both directions, the ORM peak value ranged from 23 to 100% with a median of 82% while the ORM peak time value ranged from 55 to 100% with a median of 85%.

The ORM values for the curve shapes were generally lower compared to the scalar values (Figure 8). As for the scalar values, the shape value in the x-direction was higher compared to the z-direction for both acceleration and displacement signals. Good agreement was observed for the chest deflection (77%). Low shape values were observed for the femur forces.

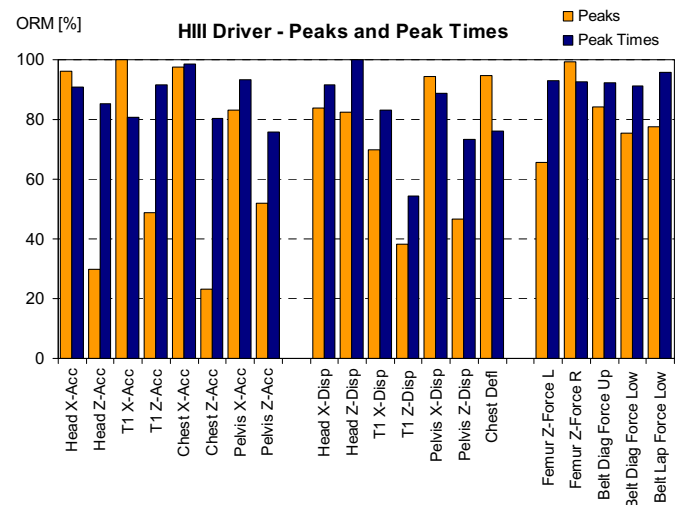


Figure 7: ORM values for the peaks and peak times of the Hybrid-III driver load case.

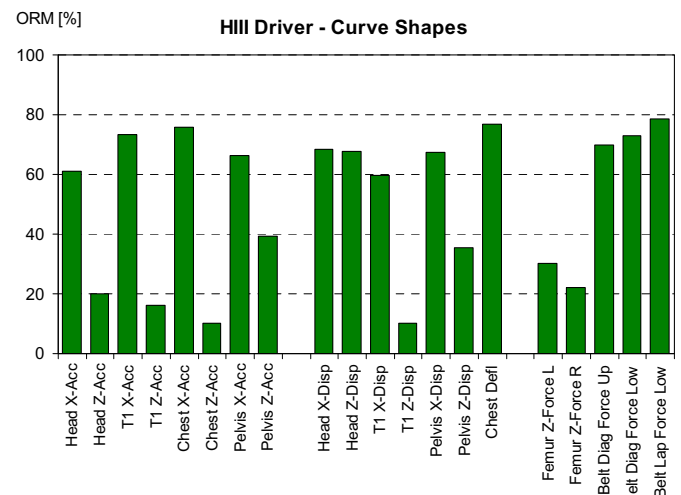


Figure 8: ORM values for the curve shapes of the Hybrid-III driver load case.

The ORM peak and peak time values for the THUMS driver side can be observed in Figure 9. The ORM x-acceleration peak values ranged from 17 to 95% with a median value of 58%, while the z-acceleration peak values ranged from 38 to 90% with a median value of 62%. For the displacements, slightly lower ORM values

were obtained in the x-direction (range 37-76%, median 61%) compared to the z-direction (range 26-60%, median 50%). For the acceleration and displacement signals in both directions, the ORM peak value ranged from 17 to 95% with a median of 58% while the ORM peak time value ranged from 57 to 100% with a median of 86%.

Figure 10 shows the ORM curve shape values for the THUMS driver side comparison. For the head x-acceleration and x-displacement and the T8 x-acceleration and pelvis z-displacement the ORM values were 59-64%. However, the majority of the curve shape values were below 50%. Good correlation was observed for the belt force shape values (79-84%).

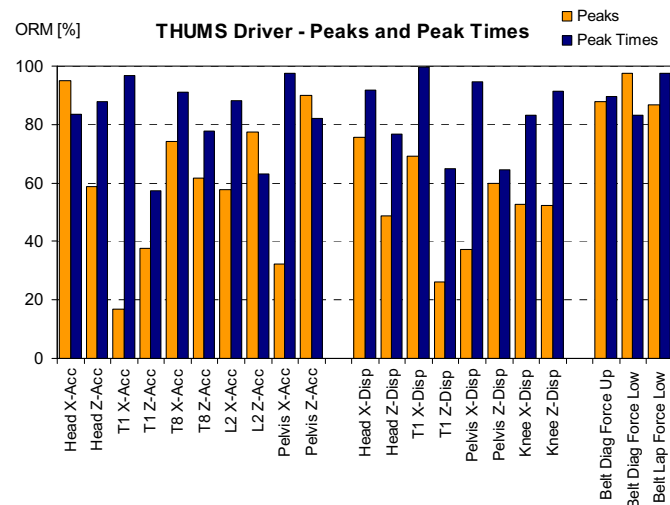


Figure 9: ORM values for the peaks and peak times of the THUMS driver load case.

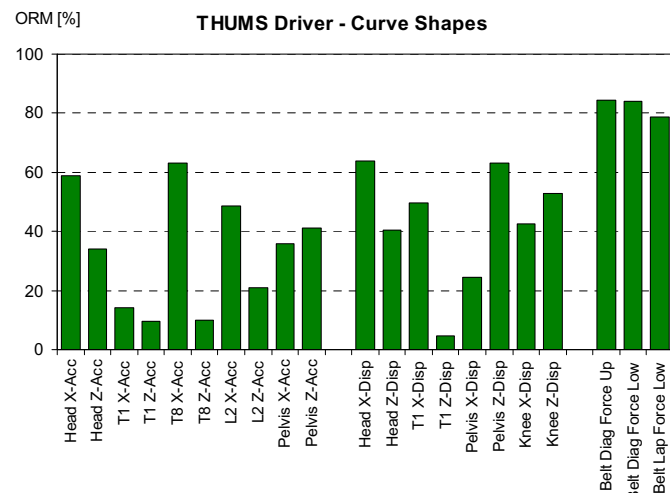


Figure 10: ORM values for the curve shapes of the THUMS driver load case.

ORM – PASSENGER LOW SPEED – The evaluations of the passenger side low speed load cases with the Hybrid-III and THUMS models were also carried out using the objective rating method. The signal curves of the two load cases can be observed in Appendix 2 and Appendix 4.

The passenger side Hybrid-III ORM peak and peak time values can be observed in Figure 11. High ORM peak values (x-direction range 86-99% with median 91%, z-direction range 55-97% with median 90%) were obtained for all acceleration signals except for the T1 z-acceleration (55%). For the displacements signals, higher ORM peak values were observed in the x-direction (range 50-94%, median 72%) compared to the z-direction (range 29-82%, median 55%). The lowest peak values were obtained for the pelvis x- and z-directions, 50 and 29% respectively. A low peak value (39%) was also obtained for the chest deflection. For the acceleration and displacement signals in both directions, the ORM peak value ranged from 29 to 99% with a median of 86% while the ORM peak time value ranged from 57 to 100% with a median of 87%.

The lowest ORM shape values were obtained for the T1, chest and pelvis z-accelerations (14, 25 and 41%) together with the pelvis x- and z-displacements (50 and 21%). Low shape values were also observed for the chest deflection, 35%, and femur forces (34-51%).

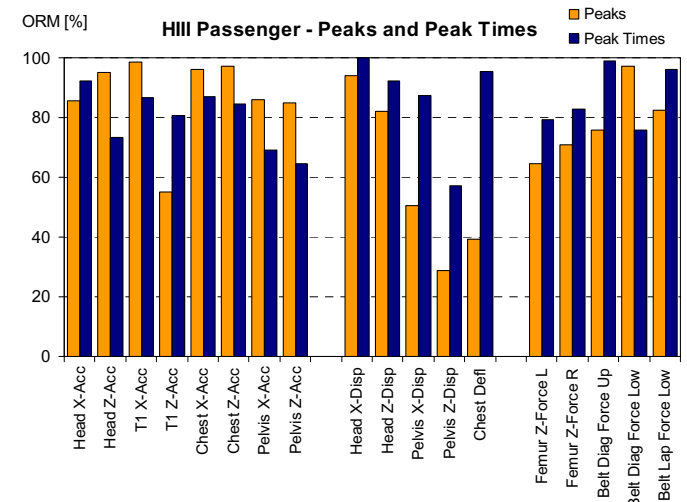


Figure 11: ORM values for the peaks and peak times of the Hybrid-III passenger load case.

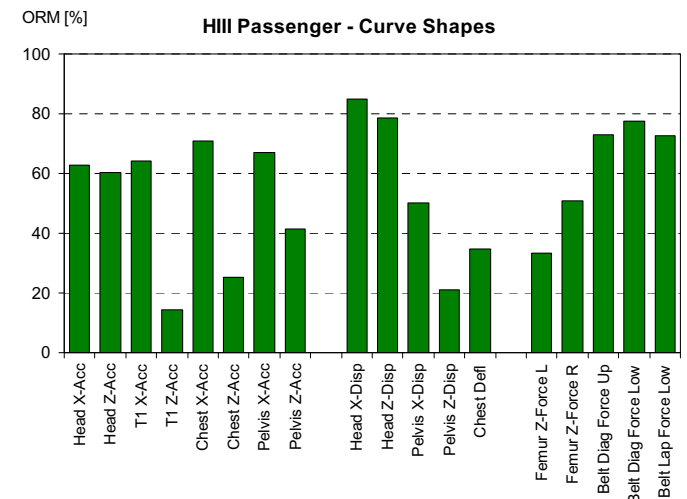


Figure 12: ORM values for the curve shapes of the Hybrid-III passenger load case.

The passenger side low speed THUMS peak and peak time ORM evaluation can be observed in Figure 13. Overall, the ORM x-acceleration peak values ranged from 16 to 86% (median value 54%), while the z-acceleration peak values ranged from 21 to 78% (median value 39%). For the displacements, lower ORM values were observed in the x-direction (range 20-91%, median 35%) compared to the z-direction (range 44-98%, median 69%). Considering only the acceleration and displacement signals in both directions, the ORM peak value ranged from 16 to 98% with a median of 47% while the ORM peak time value ranged from 41 to 100% with a median of 92%.

The ORM shape values for the THUMS passenger side comparison can be observed in Figure 11. The head x-displacement together with the belt forces showed ORM shape values of in the range of 57-92%. Several shape values were observed to be 20% or lower.

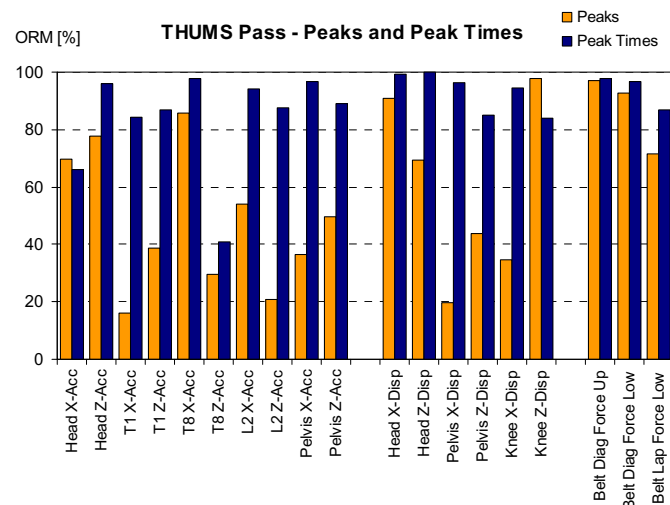


Figure 13: ORM values for the peaks and peak times of the THUMS passenger load case.

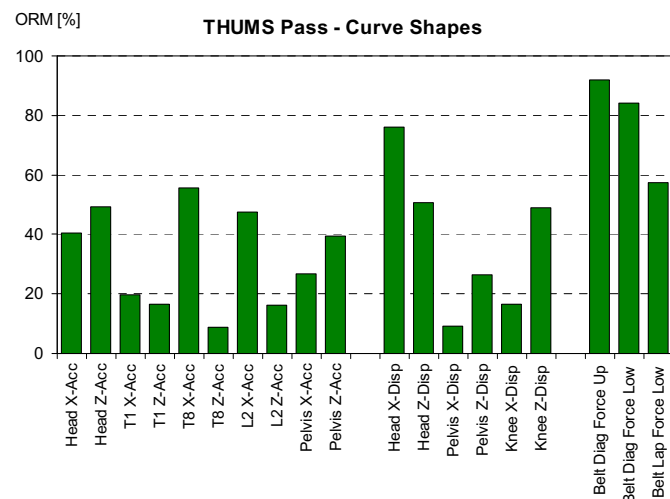


Figure 14: ORM values for the curve shapes of the THUMS passenger load case.

ORM – COMPLETE SYSTEM – The ORM evaluations for the four complete systems can be observed in Figure

15. The ORM complete values were based on all signals which were used in the previous detailed scalar and shape comparisons, see also Table 1. For the Hybrid-III driver 20 signals were used, while 18 signals were used for the Hybrid-III passenger. For the THUMS driver and passenger systems, a total of 21 signals were used for each configuration. No weighting between the values was applied, i.e. $W=1.0$ (Equation 3) was used for all ORM values. The ORM complete value for the two Hybrid-III model systems ranged between 61-64%, while the two THUMS model systems ranged between 52-55%. For the Hybrid-III model the highest value was obtained on the passenger side, while for the THUMS model the driver side showed the highest value.

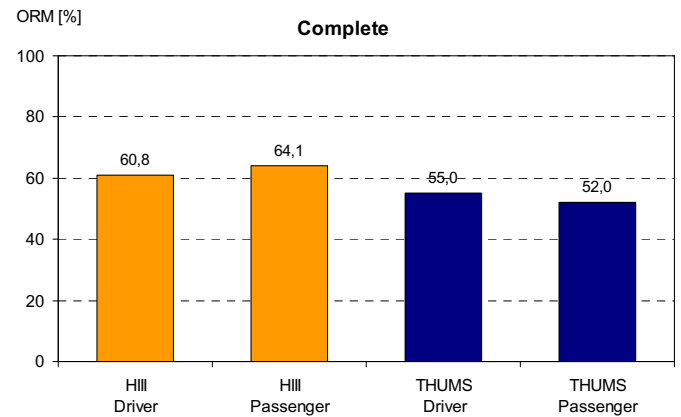


Figure 15: ORM values for the complete systems.

The correlation between model predictions and mechanical sled tests in various directions was investigated. the response signals were collected in four new groups for each load case: x-accelerations, z-accelerations, x-displacements and z-displacements. Especially in the x-direction (for both acceleration and displacement signals), the median of the ORM peak values for the THUMS models were lower compared to the Hybrid-III models (Figure 16). The value range in each group is also indicated by the vertical lines. It can be observed that the variation of the ORM peak values within each group was larger for THUMS model compared to the Hybrid-III model.

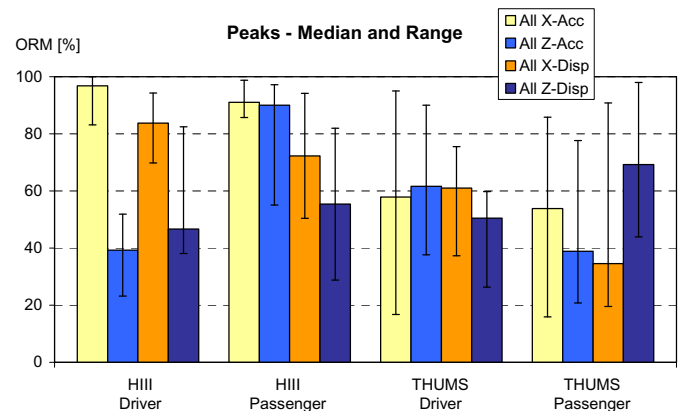


Figure 16: ORM peak values, median value and range.

The ORM curve shape values for the same four groups can be observed in Figure 17. For all load cases, the lowest median values for the curve shapes were established for the z-accelerations. For the two 50%-ile Hybrid-III load cases, the highest median values were reached for the x-accelerations and x-displacements. For the THUMS model, the passenger side was seen to have poor ORM median shape value for the x-displacements.

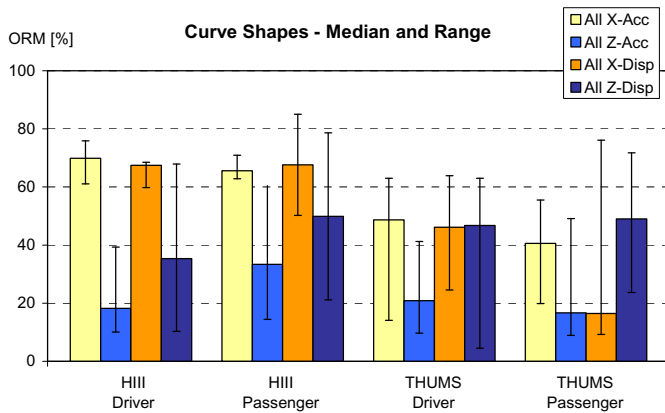


Figure 17: ORM curve shape values, median value and range.

CHEST ANALYSIS – In the chest band analysis it was observed that the upper part of the chest of THUMS was thinner but wider than the PMHS used for comparison (Figure 18). The lower part of the chest of THUMS was similar in thickness to the PMHS. However, the width was greater for THUMS (Figure 19).

In the high speed driver side tests at 80 ms the PMHS upper chest deflection was 19% and the chest upper deflection predicted by THUMS was 26% (Figure 18). At 120 ms the PMHS upper chest deflection was 19% while the deflection predicted by THUMS was 33%.

In the high speed driver side tests it was observed that the lower chest deflection at 80 ms was 13% for the PMHS and 25% for THUMS (Figure 19). At 130 ms the chest deflection of the PMHS was 6% and for THUMS the chest deflection was 34%.

In the low speed passenger side tests upper chest deflection at 80 ms was 9% for the PMHS and 12% for THUMS (Figure 20). At 160 ms upper chest deflection was 0% for the PMHS and 5% for THUMS.

In the low speed passenger side tests lower chest deflection was at 80 ms 15% for the PMHS and 14% for THUMS (Figure 21). At 160 ms lower chest deflection was 0% for the PMHS and 5% for THUMS.

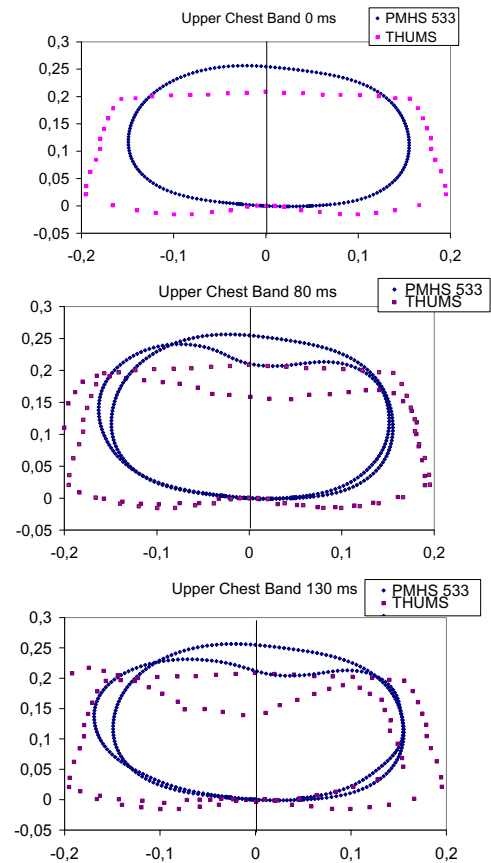


Figure 18: Upper Chest Band at 0, 80 and 130 ms in High Speed Driver Side Test (Units: m).

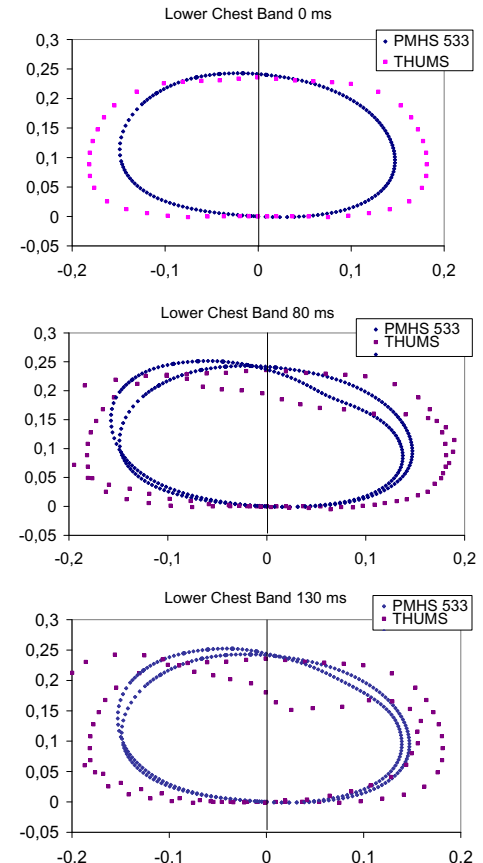


Figure 19: Lower Chest Band at 0, 80 and 130 ms in High Speed Driver Side Test (Units: m).