COMPARISON OF HYBRID III AND HUMAN BODY MODELS IN EVALUATING THORACIC RESPONSE FOR VARIOUS SEAT BELT AND AIRBAG LOADING CONDITIONS

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ABSTRACT

Thoracic responses between the Hybrid III model and the THUMS human body model were compared in three frontal impact severities for two belt (3-point and criss-cross), two belt load limiter (constant and degressive) and two airbag types (symmetric and non-symmetric). The thoracic responses were evaluated by measurements of chest deflections (mid-sternum and multi-point measured), chest excursions and, for the human body model, supplemented by maximum rib strains. For the 3-point belt, an overall correlation was found between Hybrid III and THUMS chest deflections and excursions as well as between THUMS multi-point chest deflections and rib strains. For the criss-cross belt, Hybrid III chest deflections increased and THUMS rib strains decreased. For the non-symmetric airbag, Hybrid III and THUMS chest deflections decreased.

Keywords: Finite element method, Hybrid III, Human body, Thorax, Restraint systems

ENHANCED CRASH SAFETY REQUIREMENTS drive the development of more advanced interior safety systems, which increase the demands on occupant surrogates. The biofidelity of anthropometric test devices (ATDs) is necessarily limited due to durability and reliability requirements, limitations which are also reflected in the mathematical models of the ATDs. The limitations of the single point chest deflection measurement, which is used in the Hybrid III dummy, was indicated by Cesari et al. (1994) for diagonal belt loading where it was found that the location of maximum chest deflection was observed at points other than the mid-sternum. Also Petitjean et al. (2002) reported that the mid-sternum deflection did not fully assess the effectiveness of systems with various ways of restraint by belts and airbags.

The force-deflection response of the chest increases if a diagonal belt is added to an existing 3-point belt, creating a criss-cross loading configuration (Kent et al., 2004, Kent et al., 2005, Forman et al., 2005). The total belt force can thus be increased without exceeding injurious chest deflection levels. Thoracic response of post mortem human subjects (PMHS) subjected to varying anterior chest loading in quasi-static table-top tests was evaluated by Kent et al. (2004). The response data was normalized to a 45-year-old 50th percentile male and the force response of the thorax was calculated at 20% of chest deflection for a 50th percentile male (46 mm). Compared to a single diagonal belt loading, the force response of the thorax increased by 45% for a distributed chest loading, by 40% for a double diagonal (criss-cross) belt loading and decreased by 50% for a hub loading. Kent et al. (2004) also recognized that not only the size of the loaded area, but also the anatomical structures engaged, affects the stiffness of the thorax. While the distributed conditions loaded the center of the rib cage, the double diagonal belt engaged the short and thick upper ribs and clavicles. Consequently, although the loading area by the double diagonal belt was only approx. 60% of the distributed loading area, nearly the same stiffness response of the thorax was obtained.

In addition to PMHS, Forman et al. (2005) included mechanical Hybrid III and THOR dummies for investigation of force-deflection thoracic responses from table-top loading conditions where a second diagonal belt was added to a single diagonal belt system. For double diagonal loading, with a force ratio of 0.5 between the two belts, the effective stiffness of the human thorax increased by 30% compared to a single diagonal belt loading. For the same loading condition, the effective stiffness of the THOR thorax increased by 80% and the Hybrid III thorax by 5%. The THOR thorax was thus found to be over-sensitive and the Hybrid III thorax much less sensitive than the human thorax for loading by double diagonal belts. Forman et al. hypothesized that one of the reasons for the different responses between the dummy and the human thorax can be due to the position of the shoulder/clavicle geometry relative to the upper rib cage. The sensitivity of the thorax to the additional belt will thus be dependent on to what degree the shoulder/clavicle complex acts as a loading path. Consequently, Forman et al. results indicate that both the Hybrid III and THOR dummies have biofidelic limitations for evaluating X-type belt systems.

Investigations of criss-cross type belt systems in frontal driver side sled conditions using the THOR mechanical dummy has resulted in decreased maximum chest deflections, increased deflections in the lower left point and a tendency to group the deflection values together thus resulting in smaller deflection gradients between the four measurement points (Bostrom et al., 2005, Rouhana et al., 2003, Forman et al., 2005). Bostrom et al. (2005) investigated the benefit of a 3+2 belt system where, in addition to the existing 3-point belt, a shoulder belt was mounted diagonally between the upper seat back and the lower seat frame points. The maximum chest deflection was reduced (by 16% in the upper and 22% in the lower point measurement points) while the minimum chest deflection (in the lower left measurement point) increased. Bostrom at al. estimated that the reduction in the maximum chest deflection would correspond to an AIS3+ risk reduction of about 30%. Rouhana et al. (2003) evaluated a criss-cross belt system (called X4) with the two shoulder belts attached to positions which would correspond to the upper seat back and with buckle type slip of the belt webbing through both lower lap belt points. Although the sled tests were carried out with higher belt load limiting forces than was intended, the maximum chest deflection for the X4 belt was reduced by 20% in the upper right point while deflection increased in the upper left point by 35%.

Rouhana et al. (2003) used the Hybrid III 50th percentile dummy and PMHS in driver side sled configurations to evaluate the biomechanics of 4-point belts in frontal impacts. The 4-point belts, a harness type (called V4) and the X4 belt, were compared to a baseline 3-point belt. For the Hybrid III tests, the chest deflections decreased by a factor of two by using the V4 belt and increased by 10% for the X4 belt. For the PMHS tests, the chest deflection was reduced from 38-58 mm to negative chest deflection values and the number of rib fractures were reduced from 16-32 to 0-3 by using the V4 belt. Rouhana et al. (2003) presented a hypothesis that V4 belt performed so well because the load from the belt to the human body was transferred mainly through the clavicles and pelvis.

Several mathematical models of humans have been developed to produce biofidelic predictions of human behavior in a crash and to improve the understanding of human impact response and injury mechanisms (Yang et al., 2006). Human models are a cost-effective alternative to PMHS tests without the need of ethical considerations. Human models have the potential to predict physical variables related to injury, e.g. strain, and to evaluate complex chest deformation patterns where mechanical dummies can be inadequate. One of the mathematical human body models developed is the Total Human Model for Safety (THUMS) finite element model (Iwamoto et al., 2000, Oshita et al., 2002, Iwamoto et al., 2002). The THUMS model represents an adult mid-sized male (30-40 years old) with respect to anatomical geometry and biomechanical properties such as bone stiffness and skin flexibility. The joints are modeled anatomically with the major ligaments and tendons. The chest is modeled using solid elements for the trabecular bone and shell elements for the cortical bone. For the trabecular bone an elasto-plastic material is used while an elasto-viscoplastic material is used for the cortical bone. No strain-rate dependency is defined but a failure criterion has been assigned to the cortical bone in the original THUMS model.

Various studies for validation and development of the THUMS model have been carried out by improving the model predictions based on results from PMHS tests. THUMS thorax was validated for frontal pendulum impact by Oshita et al. (2002) and for lateral pendulum impact by Furusu et al. (2001). Oshita et al. (2002) also carried out validation of THUMS abdomen for frontal impact by a half of a steering wheel. For all three load cases, the force-deflection responses predicted by the model were within the PMHS test corridors. The THUMS cervical spine model was validated in flexion-extension motion for frontal and rear end loading conditions by Chawla et al. (2005). It was found that the model predictions of the majority of the responses matched well against those measured experimentally for the PMHS. An elderly thoracic model based on the THUMS mid-sized adult male model was validated by Tamura et al. (2005) in four table-top loading conditions. By adjusting the material properties of the anterior, lateral and superior part of the cortical ribs to account for aging, the predicted force-deflection characteristics by the model matched the PMHS responses for distributed, hub, single and double diagonal belt loading conditions.

Pipkorn et al. (2008) evaluated whole-body kinematics of the THUMS model by means of two frontal sled test configurations; driver side high impact velocity with advanced restraint systems (airbag with a pretensioned and load limited belt system) and passenger side low impact velocity with basic restraint systems (belt system only without pretensioning and load limiting). First, a model of the Hybrid III 50th percentile dummy was used to validate the two sled test environments by comparing the Hybrid III model predictions to the experimentally measured dummy test responses. The Hybrid III model predictions matched the responses of the mechanical counterpart well, confirming that the sled test environments on driver and passenger sides were well replicated. The THUMS model was thereafter assembled with the validated sled models and the PMHS sled tests were replicated. The evaluation of the Hybrid III and THUMS occupant models was carried out by an objective rating method (ORM, Hovenga et al, 2005) which compared predictions from the model to results from the mechanical tests. The average responses based on five PMHS tests on the driver side and three PMHS tests on the passenger side (Forman et al., 2006) were used for the evaluations of THUMS. Overall the Hybrid III dummy model predictions were closer to the mechanical test results (ORM-value 61-64% for the two test configurations) than the THUMS predictions were to the PMHS test results (ORM-value 52-55%). For upper body x-displacement and belt force responses only, an ORM-value of 65-70% was obtained for THUMS in the two test configurations. The main responses in the frontal impact load case were thus considered well predicted by THUMS with respect to peak value, time of peak and curve shape (Pipkorn et al., 2008), indicating that the THUMS model can be used for comparative studies and trend analyses.

Although no average values of the PMHS chest compressions was constructed to take the variation into account, a comparison of the chest responses between THUMS model and two PMHS using chest band profiles was carried out (Pipkorn et al., 2008). The comparison indicated that the compressions predicted by THUMS were in these two cases greater than the corresponding PMHS chest compressions.

Previous work has indicated biofidelic limitations of current occupant surrogates for thoracic loading due to anthropometric design and response measurement feasibilities. Meanwhile, enhanced crash safety requirements drive the development of more advanced safety systems which increase the demands on the occupant surrogates. In this study, the objective was to improve the understanding of human impact response and injury mechanisms for various loading conditions on the thorax by comparison of the responses between a Hybrid III model and a previously validated human body model (THUMS). The loading conditions on the thorax were varied for three frontal impact severities using two belt types (3-point and criss-cross), two belt load limiter types (constant and degressive) and two airbag types (symmetric and non-symmetric). The thoracic responses were evaluated by chest mid-sternum and multi-point measured deflections and chest excursions. For the human body model, the chest deflections were supplemented by measurements of maximum rib strains.

METHOD

The comparison between the Hybrid III and the human body (THUMS) mathematical models predictions was carried out for belted driver occupants in a generic passenger vehicle interior. All mathematical simulations were performed using the finite element program LS-DYNA version 971 (Hallquist, 1998) and the data analyses using the software Hypergraph version 10 (Altair, 2009).

VEHICLE AND OCCUPANT MODELS: The evaluations were carried out using a generic vehicle interior model representing the driver side of a typical mid-sized sedan (Fig. 1). The interior model included seat, 3-point belt system with retractor pretensioner and load limiter, frontal airbag, steering wheel, compressible steering column with a constant force limit of 3.0 kN, knee-bolster and footwell. The depowered type airbags were inflated using both stages of a dual-stage inflator.

The mathematical occupant models included the FTSS finite element model of the Hybrid III 50th percentile male version 5.0 (FTSS, 2005) and the THUMS finite element model of a mid-sized adult male version 2.21-040407. The validation of the Hybrid III model is reported in two documents

(FTSS, 2005) where the model prediction of the responses matched well against those measured experimentally using the mechanical Hybrid III dummy. For this study the material model of the THUMS rib cortical bone was simplified by using an elasto-plastic material model with a Young's modulus of 10.2 GPa, yield stress of 65 MPa, and a tangent modulus of 2.3 GPa. Also, the strain-rate dependency and the fracture failure criterion were excluded. The occupant models were positioned according to the Euro NCAP (2009) positioning protocol.

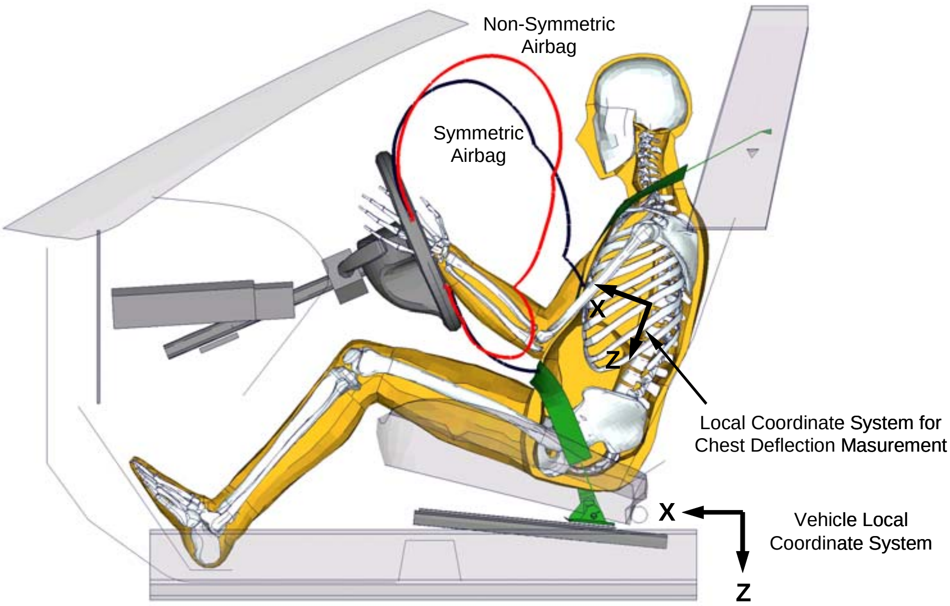


Fig. 1 – Generic interior vehicle model with the THUMS human body model.

PARAMETERS: The two occupant models were compared for three frontal impact severities based on the full frontal rigid barrier 40 km/h (FF40), the full frontal rigid barrier 56 km/h (FF56) and the offset deformable barrier 64 km/h (ODB64) load cases, all representative of a mid-sized sedan (Fig. 2). Since the main purpose was to vary the occupant kinetic energy to be managed by the safety systems, the crash pulses were applied in the longitudinal x-direction of the vehicle model only. Thus, the vehicle rotations (pitch and yaw) which are present to different degrees for the considered load cases were not taken into account.

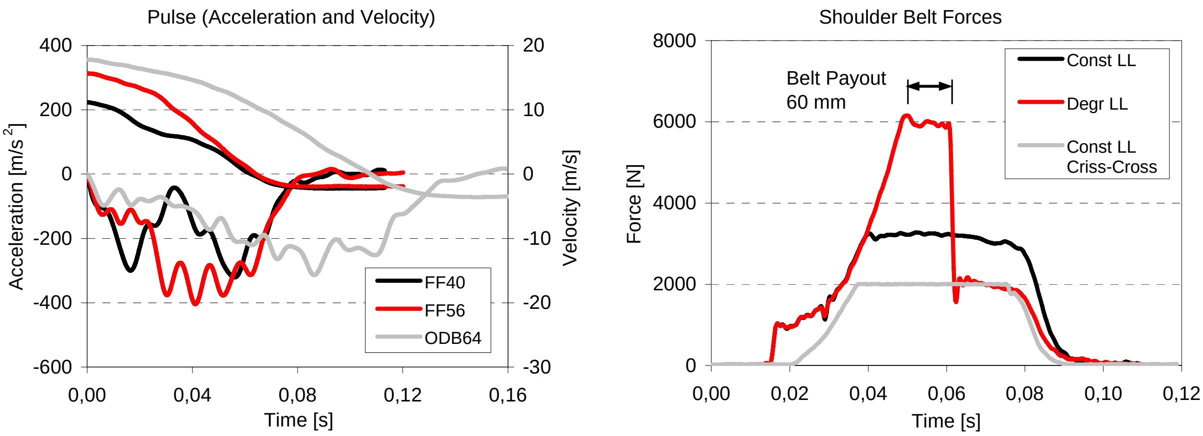


Fig. 2 – Crash pulses (left) and belt load limiters (right).

The loading condition on the thorax was varied using two belt types, two load limiting levels for the 3-point belt and two airbag types. For the pretensioned 3-point seat belt, the load limiting force, measured at the occupant's shoulder belt, was varied between constant (3.3 kN) and degressive characteristics (6.0 kN decreasing to 2.2 kN after a belt payout of 60 mm), see Fig. 2. A non-pretensioned 2 kN load-limited 2-point shoulder belt was added diagonally across the existing 3-point seat belt resulting in a criss-cross configuration (Fig. 3). The attachment of the 2-point belt to the seat back was not allowed to deform in order to reduce the variability that might have been caused by the seat back design. Since the use of the 2-point diagonal belt is intended to be optional, the load limiting force of the 3-point belt was not reduced due to the addition of an extra belt. In the case when the 2-point belt is not used, the existing 3-point belt alone must be able to fully restrain the occupant.

A non-symmetrically shaped frontal airbag with increased volume in the upper part and reduced volume in the lower part was compared to the standard symmetrical airbag (Fig. 1). The belt pretensioners were triggered at 12 ms. The airbags were triggered at 12 ms, except for the ODB64 load case. Due to the long duration of the ODB64 pulse, an airbag trigger time of 23 ms was used to achieve an acceptable occupant to airbag interaction. An overview of the parameters is presented in Table 1.

Table 1. Occupant model, pulse, belt and airbag type configurations.

Parameter	**Parameter variation**		
Occupant model	Hybrid III 50th perc.		THUMS AM 50th perc.
Pulse	Full frontal rigid barrier 40 km/h, ΔV=48 km/h (FF40)	Full frontal rigid barrier 56 km/h, ΔV=63 km/h (FF56)	Offset deformable barrier 64 km/h, ΔV=77 km/h (ODB64)
Driver airbag	Symmetric		Non-symmetric
Belt load limiter	Constant 3.3 kN		Degressive 6.0 kN - 2.2 kN
Additional belt	None		Criss-cross (load limiting 2.0 kN)

MEASUREMENTS: The chest excursions were measured at the chest accelerometer position for the Hybrid III model and at the anterior side of the 8th thoracic vertebral position for the THUMS model (Fig. 4). The chest excursions are reported in the local x-direction of the vehicle model (Fig. 1).

The risk to sustain thoracic injuries and rib fractures was assessed by chest deflections at several points and, for the human model, supplemented by measurements of maximum rib strains. The chest deflections were measured at mid-sternum and at four additional points. The additional multi-point deflection measurements were carried out at the chest positions corresponding to the Hybrid III Thorax Multi-Point and high Rate measurement device (THMPR) points (Fig. 3 and Fig. 4). For the Hybrid III model, the mid-sternum deflections were measured using the standard single-point slider device. The multi-point deflections (at four points) were measured relative to the origin of a local coordinate system, which was rigidly connected to the spine complex with respect to both translations and rotations. The multi-point deflection values are reported in the x-direction of the local coordinate systems, which was positioned parallel to the Hybrid III ribs. For the THUMS model, the chest deflection measurement system was defined by an overlay of the two occupant geometries. By assuming the same direction of the local coordinate systems x-axis between the models, the positions of the anterior deflection measurement points and the origin of the local coordinate system were defined along the x-axis line. In the same manner as for the Hybrid III model, the THUMS local coordinate system was rigidly connected to the 10th thoracic vertebrae. For THUMS, both mid-sternum and multi-point deflections are reported in the x-direction of the local coordinate system.

Strains were measured in the cortical rib bones of the rib cage for THUMS. The cortical bone was modeled using fully integrated shell elements with four integration points in the element plane and three Gauss integration points through the element thickness. The first (maximum) principal strain was measured for each rib based on the mid-surface value of the four in-plane integration points. Since the strains were compared to the deflections close to the sternum, only ribs in the vicinity of the chest deflection points were selected. Thus, ribs 9-12 were excluded due to their large geometrical distance from the chest deflection points. Rib 1 was excluded due to model discrepancies, since unrealistically high values were measured at the sternoclavicular joint. The remaining ribs 2-8 were used in the evaluations.

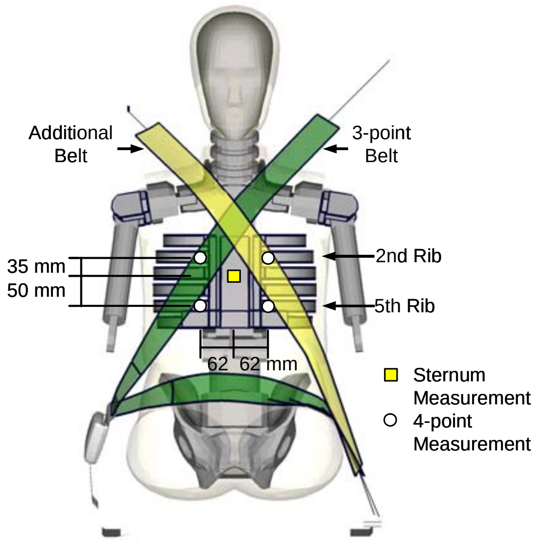


Fig. 3 – Belt geometry and chest deflection points for Hybrid III (left) and THUMS (right).

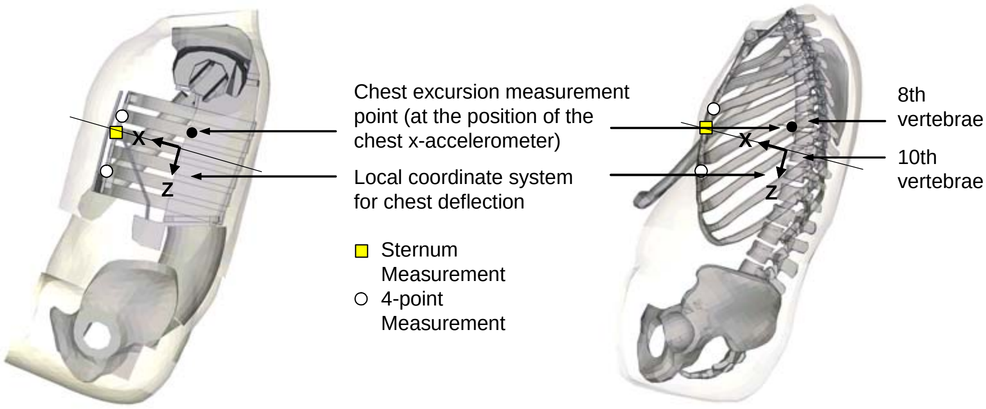


Fig. 4 – Chest deflection measurement points for Hybrid III (left) and THUMS (right).

RESULTS

The evaluation of chest deflection and excursion correlation values between the Hybrid III and THUMS models were based on all parameter combinations according to Table 1 except the criss-cross belt. For the remaining comparisons of the two occupant models, the airbag and belt load limiting parameters have been grouped in two main systems: the non-symmetric airbag together with constant load limiting and the symmetric airbag together with degressive load limiting.

The two occupant models were compared for three frontal impact severities based on the FF40, FF56 and ODB64 load cases. In general, similar chest deflections were obtained for the FF40 pulse as for the ODB64 pulse, while the chest excursions were smaller compared to the FF56 and ODB64 pulses. Due to the limited additional information compared to the FF56 and ODB64 pulses, the FF40 pulse was thus used only in the determination of the correlation values.

Detailed results including the rib number where maximum strain was measured for the THUMS model are presented in Appendix 1.

THUMS VERSUS HYBRID III: The chest deflection and excursion correlation values between the Hybrid III and THUMS models using the 3-point belt are shown in Fig. 5 for a variation of pulses (3), airbag types (2) and load limiter types (2). The THUMS model predicted higher chest deflections and excursions compared to the Hybrid III model. A correlation value of approx. 0.80 was obtained for chest deflections (independently of the measurement method) and 0.65 for chest excursions. Two outliers were identified where the chest excursions were considerably larger for the THUMS model compared to the Hybrid III model. For the Hybrid III model, larger deflections were obtained at mid-sternum compared to the point with the maximum value based on multi-point measurements (Fig. 6). For the THUMS model, larger differences between the maximum and minimum deflection values compared to Hybrid III model were obtained.

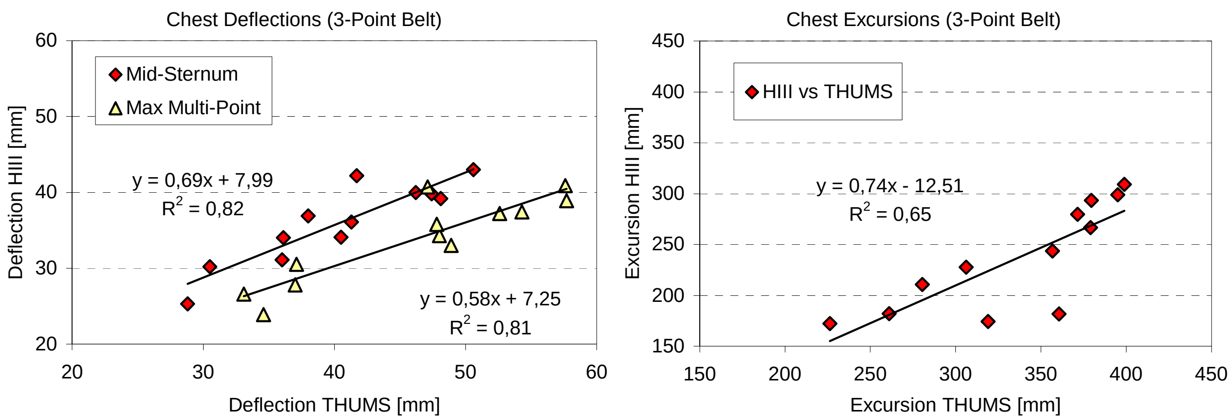


Fig. 5 – Correlation of chest deflections and excursions (Hybrid III vs. THUMS).

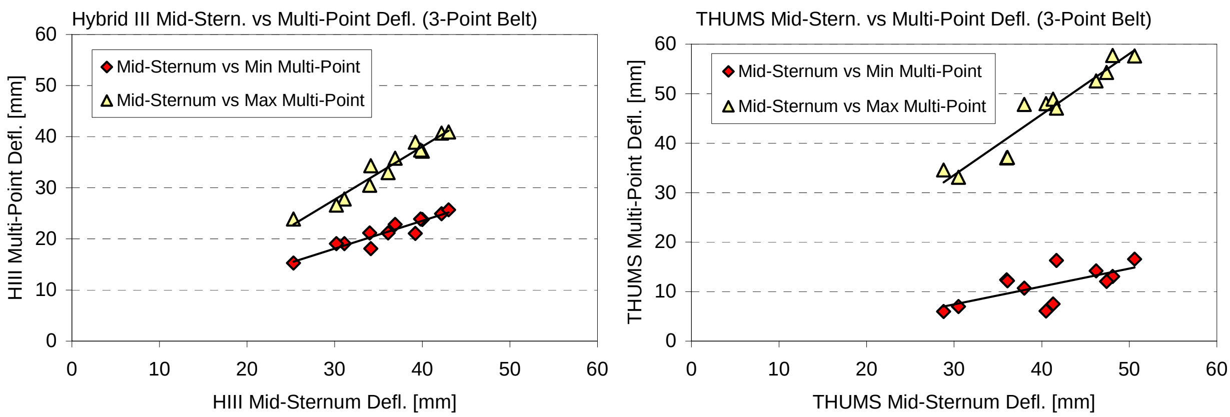


Fig. 6 – Correlation of mid-sternum to multi-point chest deflections for Hybrid III (left) and THUMS (right).

THUMS CHEST DEFLECTIONS VERSUS RIB STRAINS: The THUMS model predicted a higher correlation value of the chest deflections to rib strains using the maximum value of the multi-point measurement (0.49) compared to the mid-sternum measurement (0.34), Fig. 7. For all impact severities, lower chest deflections and rib strains were obtained using the non-symmetric airbag together with the constant belt load limiting compared to the symmetric airbag with degressive load limiting (Fig. 7).

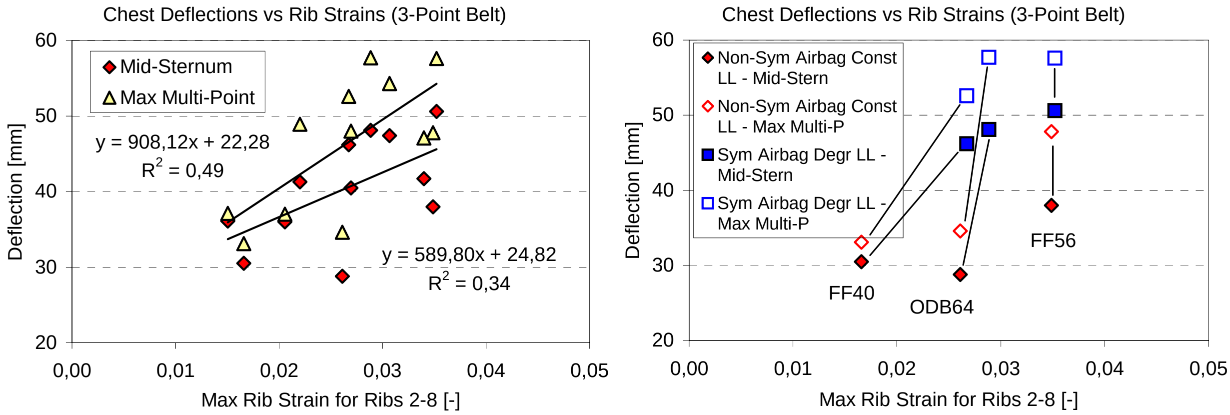


Fig. 7 - Chest deflections and rib strains for the 3-point belt (THUMS).

BELT LOADING: The Hybrid III and THUMS thoracic responses for loading by a criss-cross belt compared to the 3-point belt are shown in Fig. 8. Chest deflections increased for the Hybrid III model while increased multi-point measured deflections were predicted by the THUMS model using the criss-cross belt compared to the 3-point belt. For the THUMS model, reduced rib strains by 10-30% were obtained (Fig. 9 left). Reduced chest excursions were obtained for both occupant models (Fig. 8).

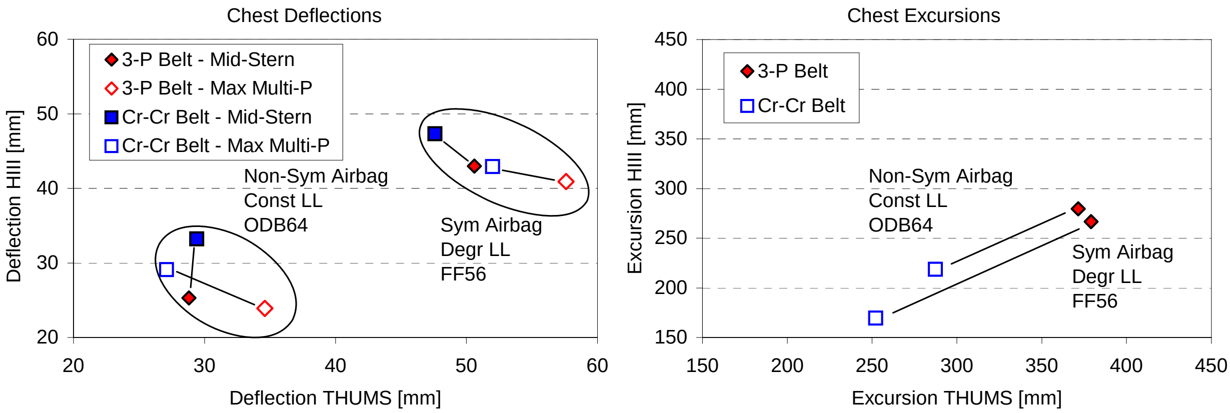


Fig. 8 – Chest deflections and chest excursions for belt loading (Hybrid III vs. THUMS).

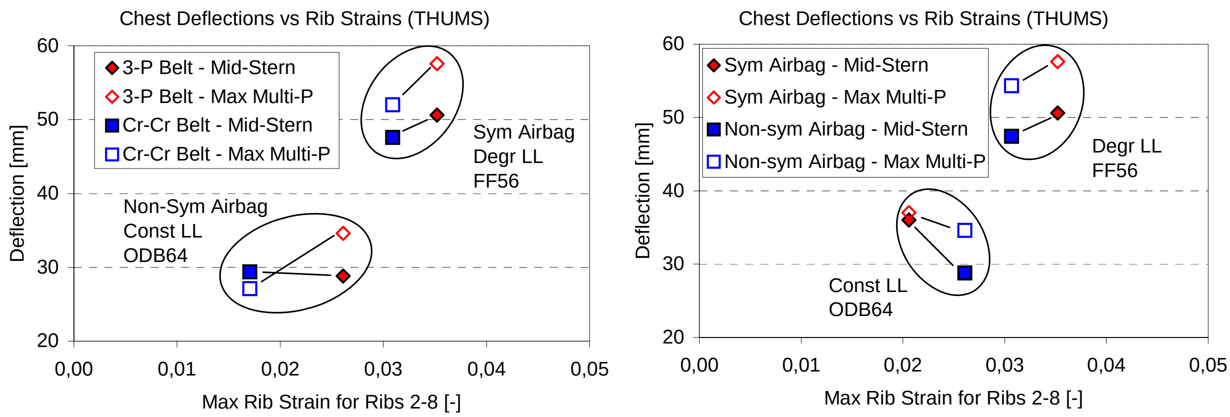


Fig. 9 – Chest deflections and rib strains for belt loading (left) and airbag loading (right) (THUMS).

AIRBAG LOADING: The Hybrid III and THUMS thoracic responses for loading by a non-symmetric frontal airbag compared the symmetric airbag are shown in Fig. 10. Chest deflections and chest excursions were reduced for both models using the non-symmetric compared to the symmetric airbag while inconsistent results for THUMS rib strains were observed (Fig. 9 right).

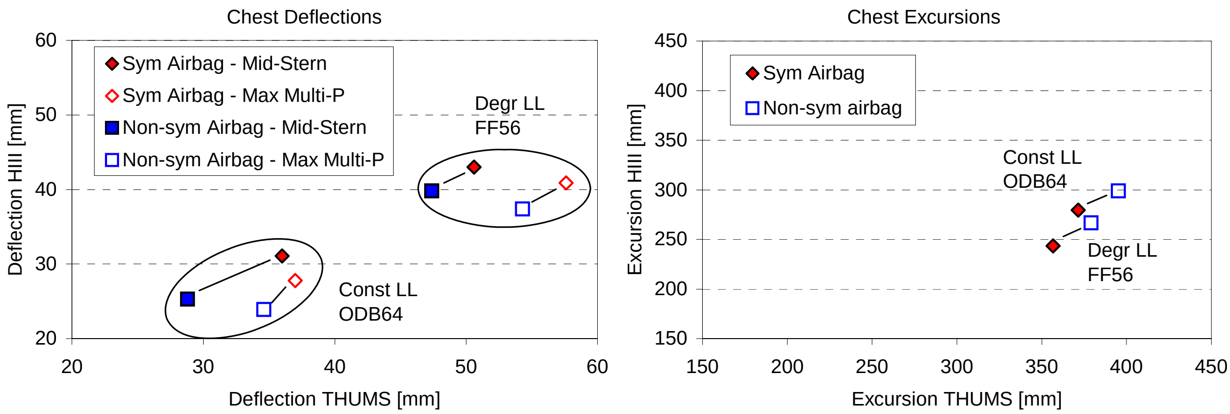


Fig. 10 – Chest deflections and chest excursions for airbag loading (Hybrid III vs. THUMS).

DISCUSSION

The thoracic responses of a Hybrid III finite element model and the THUMS human body finite element model were compared in three frontal impacts for two belt types, two belt load limiter levels and two airbag types. The comparison was carried out for mid-sternum and multi-point measured chest deflections and chest excursions. For the human body model, the chest deflections were supplemented by measurements of maximum rib strains.

For the 3-point belt, an overall correlation was found between Hybrid III and THUMS mid-sternum deflections, multi-point deflections and chest excursions. The larger deflections at mid-sternum compared to the point with the maximum value based on multi-point measurements for the Hybrid III model has been observed previously (Kent et al., 2003). A larger sensitivity to localized loading for the THUMS thorax compared to the Hybrid III thorax was indicated by the larger difference between the maximum and minimum multi-point measured deflections (Fig. 6).

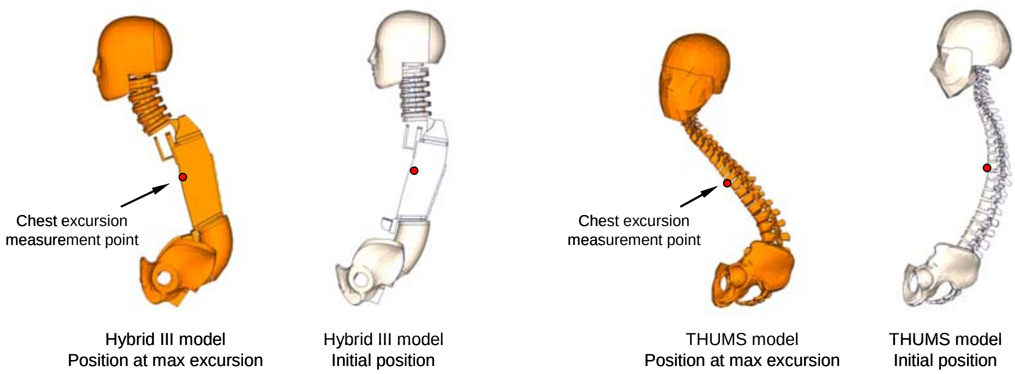


Fig. 11 – Typical spinal deformations for Hybrid III and THUMS models.

The THUMS model predicted approximately 40% higher chest excursions compared to the Hybrid III model. The excursion difference between the models can partly depend on the degree of spinal deformations (Fig. 11). The change in spinal curvature for the THUMS model during excursion can not occur for the Hybrid III model due to the rigid design of the spine. Also, two outlier points were included in the calculation of the excursion trend lines. For the long duration ODB64 pulse together with degressive load limiting, the belt force switched from the 6 kN to the 2 kN level for the THUMS model, which did not occur for the Hybrid III model. Thus, more energy needed to be managed by the restraint system for the THUMS model compared to the Hybrid III model.

Higher correlation of THUMS chest deflection to maximum rib strains was obtained for the multi-point deflection (R^2=0.49) compared to the mid-sternum deflection measurement (R^2=0.34), Fig. 7. This is consistent with the findings of Kuppa et al. (1998) who concluded that “the maximum central chest deflection alone was not as good a predictor of injury as the maximum deflection from one of five locations on the chest”. Kuppa et al. (1998) also observed that maximum chest deflection occurred at the central position in only 25% of the tests.

According to Kemper et al. (2005) the failure strain of cortical bone varies between 2-7%. For the majority of the 3-point belt load cases, rib strain values above the limit of 2% were predicted. The risk of fracture for these cases can thus potentially be reduced by a reduction in rib strains.

Dispersed properties for cortical bone material have been reported in the literature, ranging from 64-135 MPa for the yield stress (Kemper et al., 2005, Tamura et al., 2005, Ito et al., 2009) and 0.7-1.45% for the yield strain ((Tamura et al., 2005, Ito et al., 2009). The material properties for the rib cortical bone used in this study represent the lower range of the variation and further studies of the influence of the cortical bone material properties on the rib cage stiffness are needed.

The nonlinear stress-strain curve for the cortical bone has a convex shape without a true yield point. In THUMS, the nonlinear stress-strain curve was approximated by a bilinear relationship where the intermediate point has been denoted as the yield point, although it does not represent a true yield point. The choice of yield point in the THUMS model influences the unloading path but this is of less importance for frontal impacts. In Fig. 12, three strain levels between 0.6% (THUMS yield point) and 2% (fracture limit according to Kemper et al., 2005) are used to indicate the strain distribution for ribs 2-8 using the 3-point belt. For the lower deflection cases, strains larger than 0.6% were predicted for a number of ribs and strains larger than 2% for up to two ribs.

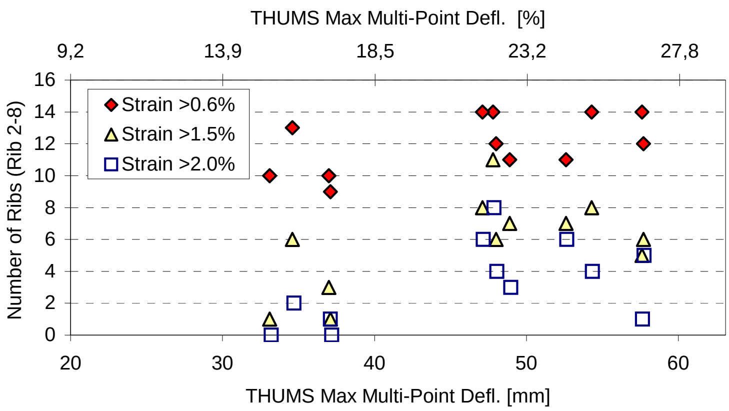


Fig. 12 – Strain distribution for the 3-point belt (THUMS).

For all impact severities using the 3-point belt system, the lowest THUMS sternum deflections and rib strains (all impact severities), but also the highest chest excursions, were obtained using the non-symmetric airbag with constant load limiting (Fig. 7 and Fig. 8). Based on the derived correlations for chest deflections and excursions between the occupant models, the same trends can be expected for the Hybrid III model.

For the criss-cross belt system, reduced chest excursions by up to 35% were obtained for both occupant models. As opposed to the increased sternum deflections for the Hybrid III model, reduced multi-point measured deflections and rib strains were obtained for the THUMS model. For the combination with the non-symmetric airbag and constant load limiting, strain levels below the fracture limit of 2% (Kemper et al., 2005) were obtained. In other words, reduced thoracic loading and excursions were obtained for the human model although the total belt force was increased by 30-90% (constant load limiting: 3.3 increased to 5.3 kN, degressive load limiting: 6.0 increased to 8.0 kN and 2.2 increased to 4.2 kN).

In previous studies of 4-point belts, reduced maximum chest deflections and a tendency of grouping of the deflection values from multi-point measurements have been reported (Rouhana et al., 2003, Bostrom et al., 2005). In an attempt to quantify the load distribution effect from 4-point belts, the difference of maximum and minimum chest deflections from multi-point measurements was calculated for previous test results and the results in this study (Fig. 13). Reductions in the deflection difference were obtained for the mechanical dummy THOR as well as the Hybrid III and THUMS models. Although improved load distribution is indicated by the results it is unclear how this is coupled to the risk of thoracic injury.

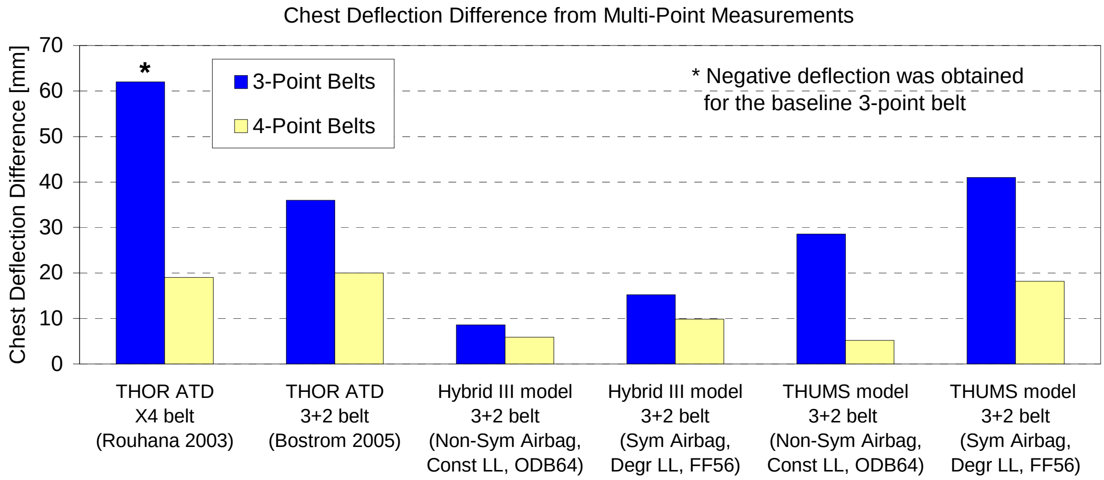


Fig. 13 – Chest deflection difference for 3-point and 4-point belts.

Although the 3-point belt and the additional 2-point belt were positioned symmetrically on the chest, the restraining of the occupant by the two belts was not entirely symmetric. In addition to the load limiting differences between the belts, the attachment point of the 2-point belt at the seat back was located below the 3-point belt D-ring position in the vertical direction. Consequently, at the occupant's shoulder, the angle of the 2-point belt in the lateral plane was closer to the horizontal compared to the 3-point belt. In contrast to the 3-point belt, the 2-point was not pretensioned since it was estimated that an extra belt will not likely include pretensioning. Also, the existing 3-point belt was able to transfer belt webbing between the diagonal and lap belt parts at the buckle. The effect of these parameters was not possible to investigate within this study but will be considered in future studies.

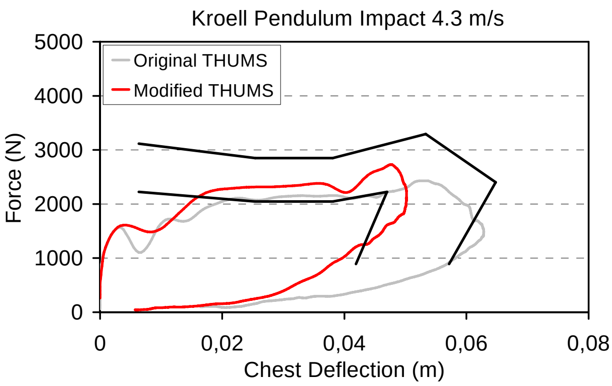
The shape of the non-symmetric airbag was designed to restrain the occupant mainly through the stiffer upper thorax (with short and thick ribs) and the head, while exerting less force on the softer lower thorax compared to the symmetric airbag. For the non-symmetric airbag, reduced chest deflections were obtained for both occupant models while blended results, depending on the degree of restraining by the belt, was observed for THUMS rib strains (Fig. 9). Due to the deformable spine of the THUMS model, an increased restraining force on the upper thorax did not necessarily unload the lower thorax to the same degree as for the Hybrid III model with its rigid spine design.

In the study by Pipkorn et al. (2008), an independent assessment of the THUMS model quality was carried out by comparing the model predictions to results from mechanical PMHS sled tests. In this validation, no tuning of the model to the frontal impact load cases was carried out due to the potential risk of reduced performance in other load cases. Anthropometric differences between THUMS and the PMHS chests were observed by comparison of chest band contours in the transverse plane. The upper part of the THUMS chest was smaller in the anterior-posterior direction than the

PMHS but larger in the lateral-medial direction. The lower part of the THUMS chest was similar in size in the anterior-posterior direction to the PMHS but larger in the lateral-medial direction. Due to the shape differences, a softer chest response for THUMS compared to the PMHS can be anticipated. Large variation in the PMHS test results can be expected due to difficulties in acquiring an adequate number of PMHS subjects in a specific size group and the inherent human variation in size, weight and body composition. It has also been shown that the mechanical characteristics of humans vary with age and gender. The variations of PMHS characteristics are often considered by using response scaling, where the responses are normalized to represent specific occupant sizes, such as small females, mid-sized males and large males. The two main response scaling techniques that have been proposed both assume geometrical similitude and constant modulus of elasticity between subjects. The main difference between the scaling methods is that the mass-scaling according to Eppinger et al. (1984) assumes no stiffness difference between the subjects while Mertz (1984) assumes the stiffness proportional to a characteristic length. However, none of the methods take into account variability of individual cadaver characteristics such as bone condition, cadaver shape, age and gender. Also, potential viscoelastic effects are not included.

Consequently, several simplified assumptions are made when scaling the test responses from a complex structure such as the human body. An alternative to response scaling of test data is anthropometric scaling of the model which is to be validated. The geometry of an existing human model can be scaled to match the geometry of the tested subject with respect to body size, shape and proportion. Total body weight as well as weights and mechanical properties of individual body segments can be modified in the model to match a specific test subject. Image measurements of PMHS using techniques such as magnetic resonance imaging (MRI) and computed tomography (CT) are available to complement the traditional anthropometric measurements of external landmarks. Detailed anthropometric information based on such techniques can be used to modify the geometry of the internal organs. Also, anthropometries of thin-tall or thick-short subjects, which are not covered by the standard mechanical dummy sizes, can be created.

In this study, several loading conditions have been mentioned where the responses are strongly dependent on the anthropometry of the anatomical structures activated. The influence of engaging the upper ribs and clavicles for various table-top loadings for PMHS was reported by Kent at al. (2004) and for mechanical dummies by Forman et al. (2005). Rouhana et al. (2003) hypothesized that the performance of the V4-belt was due to the load path through the clavicles and pelvis. Another case when anthropometric scaling can be important is for the angle of the shoulder belt, which restrains the upper body and is dependent on the sitting height. Also, the height of the torso affects the head excursions and the depth of the torso affects the interaction with the frontal interior structure and safety systems. The understanding of injury mechanisms and kinematics can thus be improved by anthropometrically improved models.



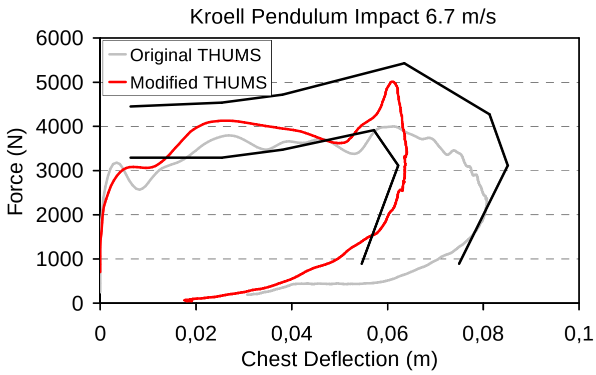


Fig. 14 – Comparison of original and modified THUMS in frontal pendulum impact.

The THUMS model as used in this study was modified by exclusion of the strain-rate dependency and fracture failure criterion for the rib cortical bone. Missing superior muscle elements were added to stabilize the kinematics of the scapula and the finite element meshes of ribs, sternum and chest soft tissue were refined. The modifications to the THUMS model were evaluated in frontal pendulum impacts (Kroell et al., 1971, Kroell et al., 1974), where higher chest stiffness was predicted by the modified THUMS compared to the original THUMS for both impact velocities (Fig. 14).

Kuppa et al. (1998) reported that the maximum resultant chest deflection was a better prediction of thoracic injury than the deflection in x-direction alone. However, in this study the chest deflection in the local x-direction was used because the mid-sternum deflection using the Hybrid III dummy standard single-point slider device is measuring mainly in the x-direction.

For the human body model, the risk to sustain thoracic injuries and rib fractures was evaluated partly by measurements of maximum rib strains. Rib fractures have not been explicitly modeled. The potential redistribution of forces due to the stiffness reduction for fractured ribs has not been considered.

In THUMS, the sternoclavicular joint is modeled using a few beam elements between the clavicle, rib 1 and sternum. The unrealistically high strain values which were measured for rib 1 are a result of the coarse modeling technique, which is estimated not to affect the stiffness of either rib 1 or the clavicle.

The results of the study have been based on a single adult male occupant size only. Future work can include investigations of the applicability of the belt and airbag systems to other occupant sizes. Moreover, implications for the neck and lumbar spine need to be investigated.

CONCLUSIONS

For the 3-point belt, an overall correlation was found between the Hybrid III and THUMS models chest deflections (approx. R^2=0.80) and chest excursions (R^2=0.65), although chest deflections and excursions predicted by the THUMS model were higher.

For the THUMS model, a higher correlation value of chest deflections to maximum rib strains (based on ribs 2-8) was obtained for the multi-point (R^2=0.49) compared to the mid-sternum deflection measurements (R^2=0.34).

For the criss-cross belt, reduced chest excursions by up to 35% were obtained for both occupant models. Increased mid-sternum deflections were obtained for the Hybrid III model and reduced rib strains for the THUMS model (10-30%).

For the non-symmetric airbag, reduced chest deflections were obtained for the Hybrid III and THUMS models.

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APPENDIX 1 – Result Summary

Num	Occupant Model	Pulse	Airbag Type	Belt Load Limiter [kN]	Belt Type	Chest X-Displ. [mm]	Chest Deflection [mm]					Max Strain Rib 2-8 [-]	Max Strain Rib Position
							Mid-Sternum	Upper Left	Upper Right	Lower Left	Lower Right		
c101	Hybrid III	ODB64	Non-sym	Const 3.3	3-Point	267	25,3	18,2	23,9	15,3	19,8	-	-
c102	Hybrid III	ODB64	Non-sym	Degr 6.0-2.2	3-Point	182	34,1	24,5	34,3	18,1	27,7	-	-
c103	Hybrid III	ODB64	Sym	Const 3.3	3-Point	244	31,1	23,7	27,8	19,1	23,5	-	-
c104	Hybrid III	ODB64	Sym	Degr 6.0-2.2	3-Point	174	39,2	29,0	38,9	21,1	31,9	-	-
c105	Hybrid III	FF40	Non-sym	Const 3.3	3-Point	228	30,2	21,5	26,6	19,1	24,8	-	-
c106	Hybrid III	FF40	Non-sym	Degr 6.0-2.2	3-Point	182	36,1	25,8	33,0	21,2	29,3	-	-
c107	Hybrid III	FF40	Sym	Const 3.3	3-Point	211	34,0	25,2	30,5	21,2	27,1	-	-
c108	Hybrid III	FF40	Sym	Degr 6.0-2.2	3-Point	172	40,0	29,8	37,2	23,8	32,0	-	-
c109	Hybrid III	FF56	Non-sym	Const 3.3	3-Point	309	36,9	25,6	35,8	22,8	33,6	-	-
c110	Hybrid III	FF56	Non-sym	Degr 6.0-2.2	3-Point	299	39,8	27,0	37,4	23,9	35,3	-	-
c111	Hybrid III	FF56	Sym	Const 3.3	3-Point	293	42,2	29,8	40,7	24,9	37,0	-	-
c112	Hybrid III	FF56	Sym	Degr 6.0-2.2	3-Point	280	43,0	30,0	40,9	25,7	37,8	-	-
c113	Hybrid III	ODB64	Non-sym	Const 3.3	Cr-cross	169	33,2	28,4	29,1	23,2	24,1	-	-
c124	Hybrid III	FF56	Sym	Degr 6.0-2.2	Cr-cross	219	47,3	36,2	42,9	33,0	40,5	-	-
c131	THUMS	ODB64	Non-sym	Const 3.3	3-Point	379	28,8	20,5	34,6	6,0	31,7	0,026	8R
c132	THUMS	ODB64	Non-sym	Degr 6.0-2.2	3-Point	361	40,5	21,0	36,8	6,1	48,0	0,027	8R
c133	THUMS	ODB64	Sym	Const 3.3	3-Point	357	36,0	23,8	33,2	12,4	37,0	0,021	8R
c134	THUMS	ODB64	Sym	Degr 6.0-2.2	3-Point	319	48,1	33,6	42,8	13,1	57,7	0,029	8R
c135	THUMS	FF40	Non-sym	Const 3.3	3-Point	306	30,5	17,8	28,9	7,0	33,1	0,017	8R
c136	THUMS	FF40	Non-sym	Degr 6.0-2.2	3-Point	261	41,3	23,0	39,6	7,5	48,9	0,022	8R
c137	THUMS	FF40	Sym	Const 3.3	3-Point	281	36,1	24,2	32,9	12,2	37,1	0,015	8R
c138	THUMS	FF40	Sym	Degr 6.0-2.2	3-Point	226	46,2	29,9	41,8	14,2	52,6	0,027	8R
c139	THUMS	FF56	Non-sym	Const 3.3	3-Point	399	38,0	38,8	47,8	10,7	44,6	0,035	4L
c140	THUMS	FF56	Non-sym	Degr 6.0-2.2	3-Point	395	47,4	31,3	45,7	12,1	54,3	0,031	4L
c141	THUMS	FF56	Sym	Const 3.3	3-Point	380	41,7	28,6	43,4	16,3	47,1	0,034	8R
c142	THUMS	FF56	Sym	Degr 6.0-2.2	3-Point	372	50,6	32,5	45,7	16,6	57,6	0,035	8R
c143	THUMS	ODB64	Non-sym	Const 3.3	Cr-cross	252	29,4	23,2	24,1	21,9	27,1	0,017	8R
c154	THUMS	FF56	Sym	Degr 6.0-2.2	Cr-cross	287	47,6	35,9	41,2	33,8	52,0	0,031	8R