A Numerical Investigation of Human Biomechanical Response under Vertical Loading Using Dummy and Human Finite Element Models

Costin D. Untaroiu, Jacob Putnam
Virginia Tech, Blacksburg, VA, USA

Abstract

The safety of aerospace transport for both fixed and rotary wing aircraft is evaluated primarily through testing of anthropometric test devices (ATDs), commonly known as crash test dummies. While the majority of the ATDs were certified under automotive horizontal impact conditions, their biofidelity under vertical loading is less known. The objective of this study was to compare the THOR-K dummy model response to the corresponding response of the THUMS human FE model in the same impact conditions. A series of vertical drop tests were performed on a THOR-K crash dummy. Impact conditions were replicated in the FE simulation based on pre-impact velocities and crash pulse decelerations measured during testing. FE simulations were run with both dummy and human FE models using LS-DYNA® software. Comparisons between injury prediction of dummy and human models were also performed. While a good correlation was observed in terms of neck load between both FE models, the THUMS spine showed a higher bending flexibility within the sagittal plane. In addition, differences were observed in pelvis region where a significant bouncing was observed in THOR model, but not in the THUMS model. The comparison of THOR FE model with THUMS human model may help to improve the THOR design and define better injury criteria for vertical loading.

Introduction

During a crash aircraft occupants can be exposed to a variety of complex loading conditions the primary component of which is typically a high rate vertical acceleration. Commonly, these extreme loading scenarios induce a variety of injuries during the initial impact acceleration phase. Recently, several studies reported that serious and fatal injuries could be potentially avoided if the protective systems within the aircraft are adequately designed. It has been estimated that 85 percent of all aircraft crashes were classified as survivable [1]. A better understanding of human response and injury mechanisms under vertical loading is necessary to improve the design of novel airplane protective systems (e.g. seats, restrain systems).

The safety of aerospace transport for both fixed and rotary wing aircraft is evaluated primarily through testing of anthropometric test devices (ATDs), commonly known as crash test dummies. Historically, the Hybrid II, Aerospace Hybrid III, and FAA Hybrid III dummies have been the most commonly used ATDs in aircraft crashworthiness testing. Recently, there has been an increased interest for testing the Test Device for Human Occupant Restraint (THOR), the most advanced automotive dummy, in aerospace loading conditions. ATD testing provides an effective method for vehicular safety evaluation [2, 3]. However, the high cost and limited availability of THOR ATD makes performing impact tests in the multitude of possible aerospace impact configurations difficult. Numerical simulations of impact provide an important compliment to ATD testing by evaluating performance in a limitless number of scenarios.

Vehicle structural response to a crash impact can be accurately modeled in currently available finite element codes such as LS-DYNA[4]. Yet to confidently model the safety of aircraft, a well validated test dummy model which exhibits biofidelic response in relevant impact scenarios is necessary. Further, simulation of dummy model response with a human model may help in the continuous improvement of dummy design and in the defining of better dummy injury criteria. In the current study a drop test is simulated using a FE model of THOR-NT based on the data recorded during a series of dummy tests performed at NASA LaRC [5, 6]. Then, the same drop test is simulated using THUMS (Total Human Model for Safety) FE model. Finally, the dummy and human responses are compared and the differences are discussed.

Methods

In the vertical ATD impact tests, the THOR-NT dummy was arranged in an upright position prior to each test. Lap and chest restraints were used to hold the dummy to the seat; minimum tension was applied. Dummy and seat were dropped from a certain height onto a honeycomb block setup which generated a specific deceleration pulse upon impact[7]. The model was updated and calibrated for vertical loading in our previous study[6]. A seat model was developed to the specifications of the seat used in the physical tests. The THOR-NT dummy FE model was originally positioned within the seat and the safety straps used in testing were modeled as seat belts in LS-PrePost®. The final pre-test posture of the dummy model was verified based on photogrammetric imagery of the dummy recorded prior impact (Fig. 1).

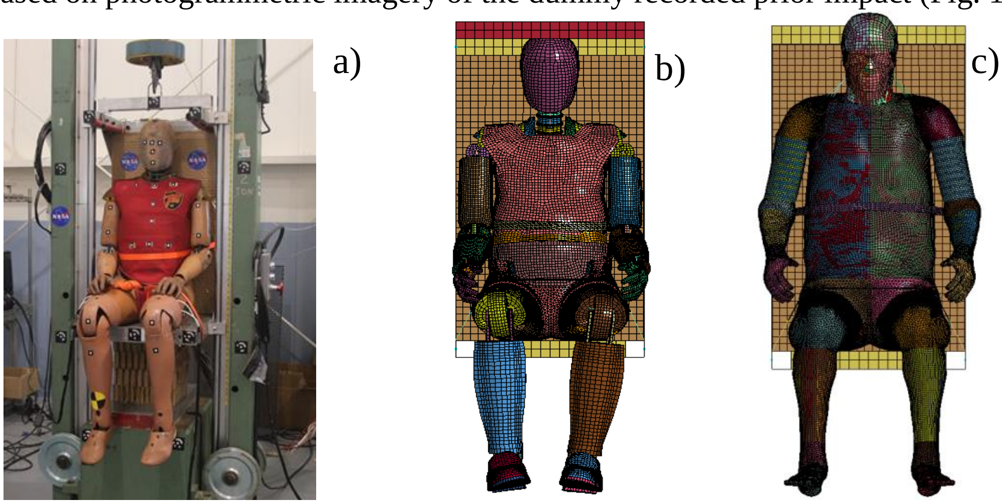

Figure 1. Pre-test conditions of THOR ATD a), THOR-NT FE model (b), and THUMS FE model (c).

Initial velocity in was set to all dummy and seat model parts based on the test pre-impact velocity. The acceleration time history of the seat, recorded in testing and filtered in accordance with guidelines set forth by SAE J211[8], was assigned to the seat model (Fig. 2). In addition, gravitational acceleration (9.81 m/s^2) was applied to all parts in the FE simulation. LS-DYNA FE software (LSTC, Livermore, CA, USA) was used to run all simulations. Simulations were performed on a desktop PC with an Intel® Core™ i7-2600 CPU @ 3.4 GHz processor. Simulation time step of the THOR-NT was 0.63 µs and an average computation time for a 150 ms impact condition was approximately 30 hours.

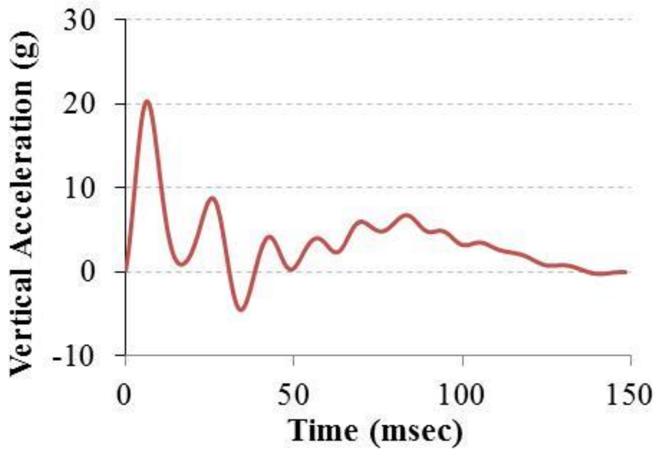

Figure 2. Impact deceleration pulse used in simulation.

Dummy head and spine kinematics were calculated were calculated at the locations of head CG, T1, and T12 accelerometers (Fig. 3a). Three nodes were defined at corresponding locations in THUMS FE model (Fig. 3b). The trajectories of the head CG and T1 were calculated based on displacement data recorded in both models relative to the chair.

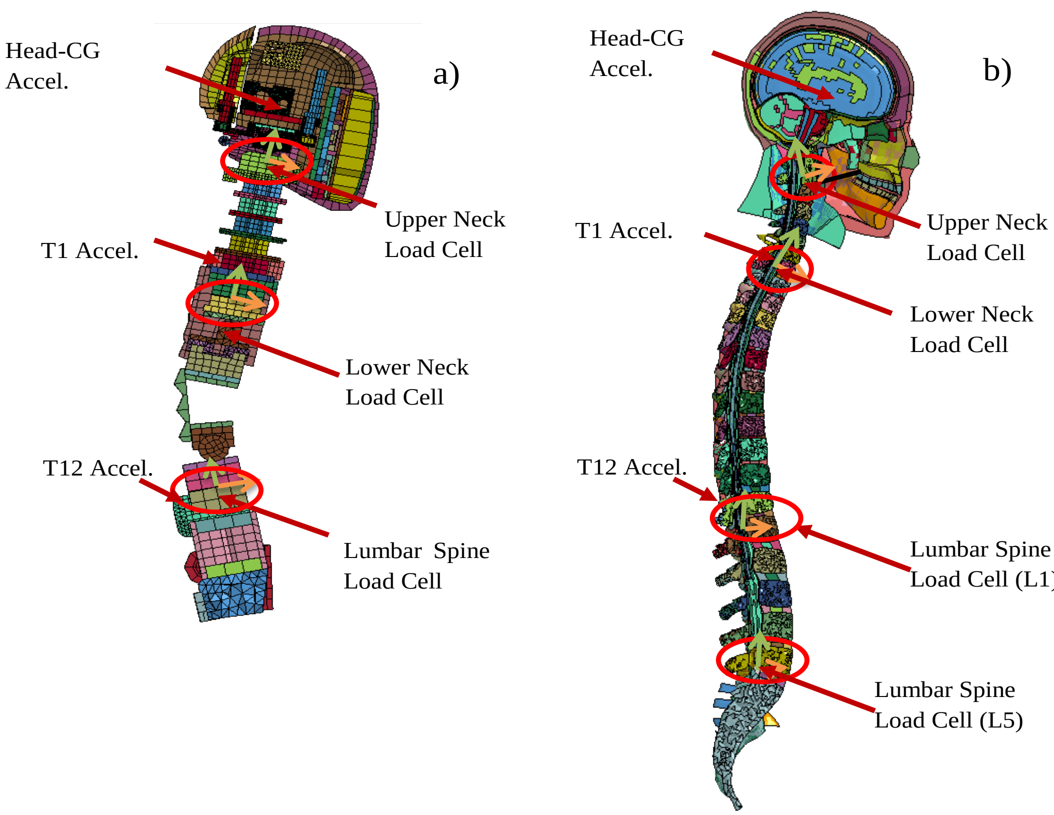

Figure 3. The cross-sections through the head-neck-spine complex a) THOR-NT dummy FE model b) THUMS human FE model

Loading data in the FE model of THOR-NT dummy during drop simulation was calculated at its upper neck, lower neck, and lumbar spine load cells (Fig. 3a). Similar load cells were defined in THUMS FE model using cross-section planes at locations of C1, T1 and T12/L1 with their respective local coordinate systems aligned with the loading orientation up the spine (Fig. 3b). To calculate the compressive force between the lumbar spine and pelvis, an additional load cell was defined at L5 in THUMS FE model (Fig. 3b).

Results and Discussion

Overall, the THOR-NT FE dummy model (Fig. 4a) shows a stiffer kinematic response than the THUMS FE model during the drop simulation (Fig. 4b). The THUMS FE model exhibits a relatively large flexibility of its vertebral column in the sagittal plane and significant pelvis flesh compression during vertical loading. The THOR FE model shows a less spinal flexibility and less deformation of the pelvis flesh.

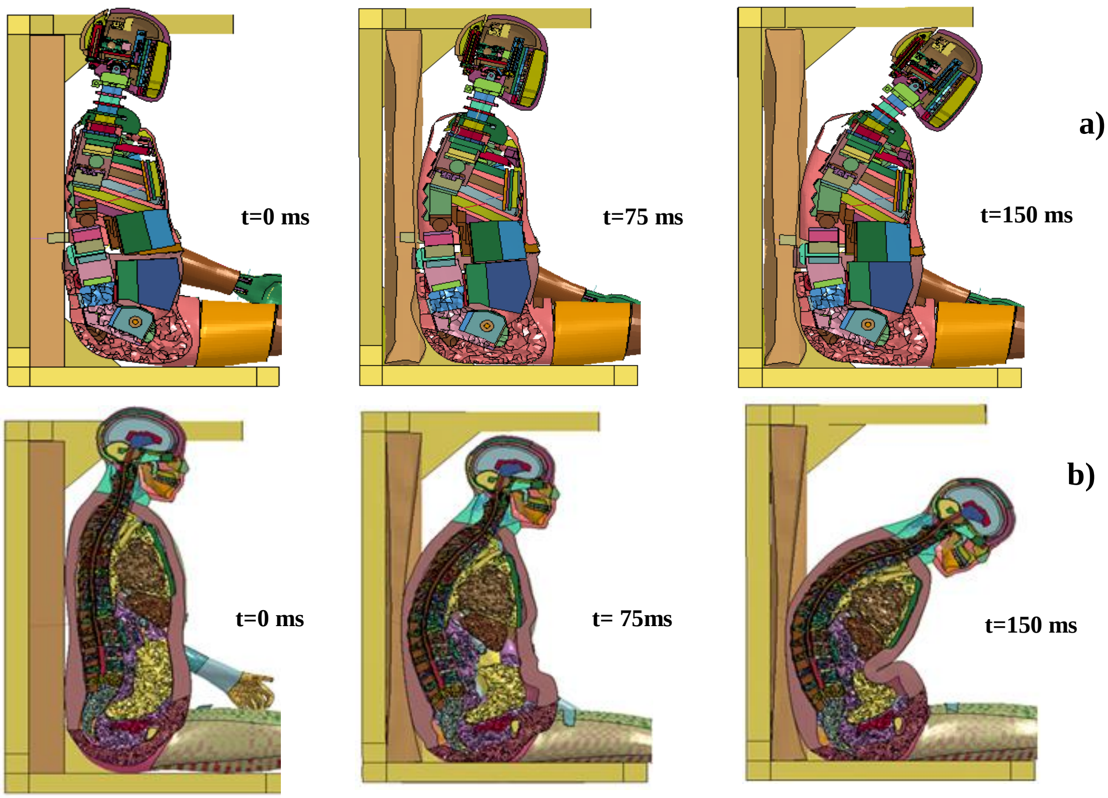

Figure 4. Comparison of the a) THOR FE model and b) THUMS FE model kinematics during drop simulation (sagittal cross-sections).

The two models begin with a similar trajectory (Fig. 5). However, as the simulation progresses the head CG in the THUMS model demonstrates a larger vertical displacement. This difference in vertical displacement is primarily developed early in simulation. Overall, both FE models predict a similar horizontal displacement (head excursion).

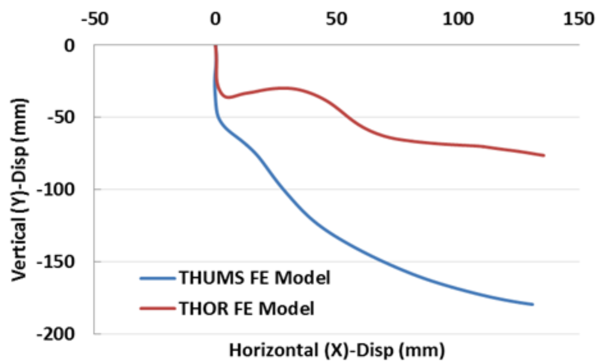

Figure 5. The trajectory of the Head CG in the THOR and THUMS FE Models.

Both THOR and THUMS FE models predict no risk of head injury in the simulation performed (Tab. 1). Both HIC criteria calculated the THOR FE data are below 1% of the maximum threshold for injury. In addition, the first principal strain of the THUMS cerebrum model was lower than .15 during the simulation which indicates a very low probability of injury.

Table 1: Head injury metrics calculated in THOR and THUMS FE models

Kinematic Injury Metrics	
Dummy Criteria	**THOR Value**
% Threshold HIC15	0.56%
% Threshold HIC36	0.37%
Human Criteria	**THUMS Value**
CSDM 15%	0
Max FPS	.0628514

The trajectory of T1 is seen to oscillate in the THOR model while continually decreasing in THUMS (Fig. 6). Similar to the head CG trajectory a significant increase in vertical displacement is observed in the THUMS model. The maximum difference in T1 vertical displacement (85 mm) is close to the corresponding distance of at the head CG (103 mm). The difference in total T1 horizontal displacement is approximately 18 mm compared to only 4 mm in the head CG.

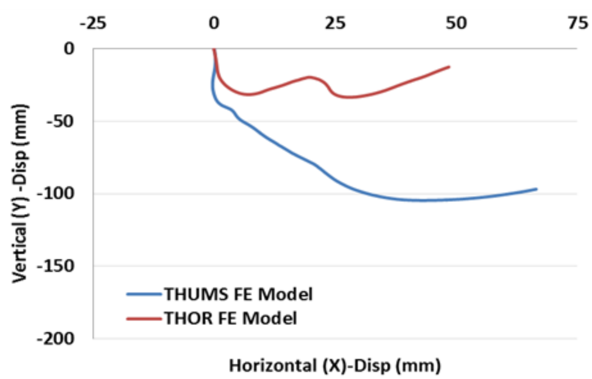

Figure 6. . The trajectory of T1 in the THOR and THUMS FE Models.

Overall, the time histories of vertical load in the neck show highly correlated values between the THOR FE load cell and vertebrae of the THUMS model (Fig. 7a). The C1 vertebra demonstrates the greatest correlation, with both initial and second peaks closely matching (Fig.

7a). The initial peak is also synonymous in the T1 and lower neck load cell (Fig. 7b), though the second peak is significantly higher in the THOR model. In both instrumented measurements the loads calculated in the THOR model demonstrate more dramatic unloading and secondary loading rates.

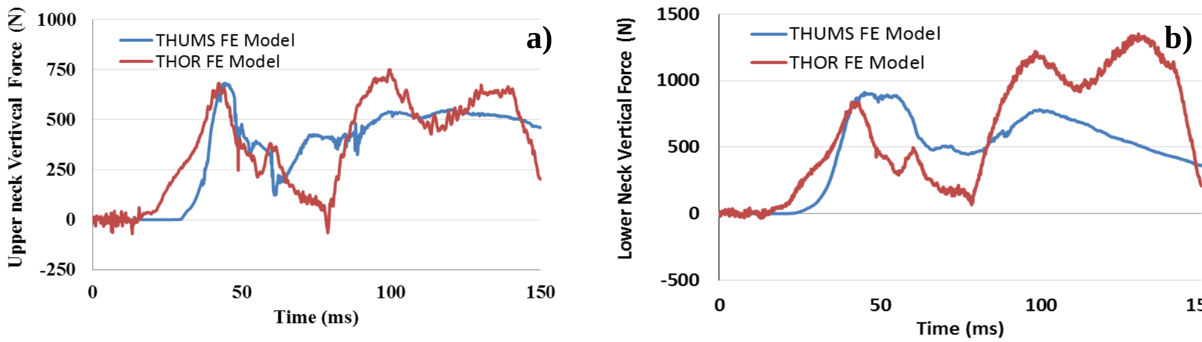

Figure 7. The time histories of vertical force at a) Upper neck b) Lower Neck in the THOR and THUMS FE models.

No risk of injury is predicted at the neck in either the THOR or THUMS FE models. Both dummy neck injury criteria predict very low risk of moderate injury with values. In the THUMS FE model, the peak stress in the T1/C1vertebrae were also well below failure (Tab. 2).

Table 2: Neck injury metrics calculated in THOR and THUMS FE models

Kinetic Injury Metrics	
Dummy Criteria	**THOR Value**
Nij (Upper Neck)	.2111 (No Predicted Injury)
BC (Lower Neck	.3746 (No Predicted Injury)
Human Criteria	**THUMS Value**
C1 % failure stress	43.02%
T1 %failure	36.94%

The trajectory of T12 in the THOR and THUMS models differ fairly radically (Fig. 8). As the THOR FE model bounces in the seat, the vertical displacement of T12 returns to zero.

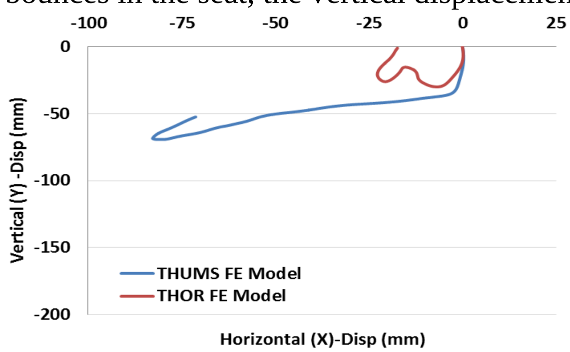

Figure 8. . The trajectory of T12 in the THOR and THUMS FE Models.

The THUMS demonstrates a much more prolonged vertical drop. In addition very little horizontal T12 displacement is observed in the THOR dummy model, while the THUMS T12 vertebra is demonstrates significant horizontal movement. The vertical load in the L5 vertebra of the THUMS shows a closer load history to the lumbar load cell of the THOR FE model, than the L1 vertebra (Fig. 9). The THOR model exhibits a dramatic initial load spike not seen in the THUMS model. The initial peak load is significantly larger in the THOR model, though secondary loading matches closely to L5. Initial load rise time is longer in the THUMS model and increases from L5 to L1.

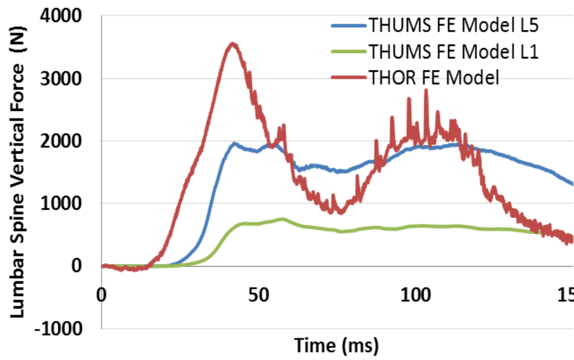

Figure 9. The time histories of vertical force at Lumbar spine

No risk of injury is predicted in the lumbar spine in both the THOR and THUMS models (Tab. 3). The lower lumbar peak force in the THOR model is almost 2,000 N below the injury threshold. Stress in both L1 and L5 are the less than defined yield stress, thus low probability of spinal damage due to column fracture is predicted.

Table 3: Lumbar spine injury metrics calculated in THOR and THUMS FE models

Kinetic Injury Metrics	
Dummy Criteria	**THOR Value**
LL	3547.01 (No Predicted Injury)
Human Criteria	**THUMS Value**
max. stress vs. L1 failure stress	59.76%
max. stress vs. L5 failure stress	43.02%

The distribution of peak seat force into the vertebral column of the THUMS and THOR model shows a similar decrease from the lumbar region into the neck though a significant difference in spinal loading (Fig. 10).

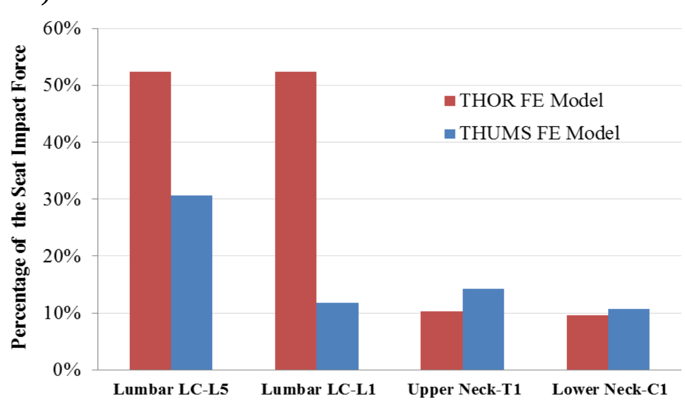

Figure 10. The Percentage of vertical impact force transmitted through the lumbar spine (L5 & L1), lower neck and upper neck. Comparison between the THOR and THUMS FE models.

The THOR lumbar load cell is shown to absorb over 50% of the vertical impact force in comparison to the L1 vertebra which is over 40% less at approximately 10%. The absorption rate of the THOR lumbar load cell is shown to be more representative of the L5 vertebra which absorbs approximately 30% of impact force. The THOR and THUMS neck models demonstrate similar absorption with approximately 10% of the distributed impact load in the upper and lower neck, though the absorption in in THUMS neck is slightly higher.

Discussion

In observing the kinematic responses of both human and dummy models a difference in spinal response is immediately apparent. The human model exhibits much greater flexibility in the spine, showing greater horizontal displacement in the lower back leading to increased vertical displacement in the head and neck. However, the THOR model shows good vertical load biofidelity in the neck. The peak initial load is very similar in both the lower and upper neck load cells in comparison to the corresponding locations in the human model. The drop in force observed in the THOR model is likely due to the bounce observed in kinematics and not a direct function of the neck response. This seat bouncing observed in THOR may be caused by the higher stiffness and lower damping of the THOR pelvis region than similar region of THUMS. While the deformation of pelvis has no a significant influence during the frontal crashes, its role in load transmission seem to be significant during vertical loading. Therefore, improving the biofidelity of THOR pelvis under compressive loading is suggested to be performed in the future.

As in the case of spinal kinematics, differences in lumbar load response between THUMS human model and THOR dummy model are observed as well. While the level of neck load cells (T1 and C1) correlated relatively well, higher peak loads are observed in THOR lower spine relative to THUMS (L1 and L5). These significant difference could be caused by the higher flexibility of THUMS spine which may cause that a significant part of the seat load to appear as a tangential force along the THUMS spine, not only as a compressive load as in the THOR spine. Therefore, in addition to pelvis improvement suggested previously, an increase of the flexibility of THOR spine is recommended as well.

HIC values of less than 1% of maximum in dummy head were shown to correlate well with the strain low values observed within the human model cerebrum. While this is a positive indication in correlation between the two models further studies are necessary to validate this correlation in more violent impact scenarios. The LL dummy injury criteria predicts no injury, as it is predicted by the low maximum stresses recorded within the THUMS L5 and L1 vertebrae. While some significant variations were observed in the normal force along the THUMS spine (e.g. L1, and L5), future studies are suggested to specifically evaluate differences in loading pathways up through the pelvis and thorax.

This study presents the vertical load response similarities and differences between the THUMS human and THOR dummy FE models. The results of this study are limited by a few factors which should be taken into account in future studies. Firstly, the THUMS model has been primarily validated in vertical impact conditions[9]. Future volunteer/PMHS studies are necessary to further validate its biofidelity in vertical impacts conditions. Secondly, the test simulated was a low injury risk impact, limiting the power of injury metric analysis. Lastly, material property differences have been speculated between the THOR FE model and dummy[6]. These differences may limit the ability to relate the results of this study to the physical dummy.

Acknowledgements

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