The validation of a full body finite element model in lateral full body sled and drop tests

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ABSTRACT

This study presents three validation cases of the Global Human Body Models Consortium (GHBMC) mid-sized male model (M50) – full body sled tests at two speeds, and a lateral drop test. The lateral sled tests involved propelling a seated cadaver into a rigid wall at 6.7 m/s and 8.9 m/s. Drop tests were performed from 1 m above a rigid surface. Model results were compared to force vs. time curves, peak force, and number of fractures from the studies. For the 6.7 m/s impact, the peak thoracic, abdominal and pelvic loads were 8.7 kN, 3.1 kN and 14.9 kN for the model and 6.0±1.1 kN, 3.6±1.3 kN, and 5.1±2.5 kN for the tests (n = 3). Similarly, in the 8.9 m/s case they were 12.6 kN, 6.1 kN, and 21.9 kN for the model and 8.75±4.5 kN (n=6), 5.0±2.0 kN (n=8), and 15.0±4.5 kN (n=8) for the experiments. The peak thorax load in the drop test was 6.7 kN for the model and the range was 5.8 kN to 7.4 kN in the cadavers. When analyzing rib fractures, the model predicted Abbreviated Injury Scale (AIS) 3 thoracic injury in the 6.7 m/s sled test while the test subjects injuries ranged from AIS 0 to 4. In the 8.9 m/s sled test, the model predicted AIS 4 thoracic injury which matched the AIS scores in 7 of the 8 test subjects. In both sled test cases the model also predicted pelvic fractures while none were reported in the literature. The model predicted an ilio-ramus fracture in both cases in addition to ischio pubic ramus and sacrum fractures in the 8.9 m/s case. The model predicted 2 rib fractures in the drop test, which is within the reported range of 0 to 5 fractures for male subjects. These results provide confidence in the GHBMC model's performance in lateral impacts.

INTRODUCTION

More than 1.2 million people die worldwide in motor vehicle crashes (MVC's) and current trends indicate that MVC's will become the fifth leading cause of death worldwide by 2030 (2009). Side impacts are of particular concern; in the United States this crash mode represents just 19% of all crashes, but represents 32% of fatalities (Vander-Lugt 1999). This indicates that they have a higher mortality than other crash modes. In the field of injury biomechanics, computational models have been used for more than 50 years to study MVC's and related injuries (Yang 2006). Finite element analysis (FEA) has commonly been used as a computer modeling tool in studying blunt injury biomechanics. Validated human body FEA models can be used to determine the stresses and strains in conditions that are difficult to determine experimentally, thereby providing an advantage in safety system and injury mitigation research. Because of this, FEA is a tool that is used in studying various blunt injury mechanisms. With the advancement of computational capabilities, human body models have become increasingly detailed and complex allowing for more accurate predictions and a better understanding of injuries and their mitigation (Yang 2006).

Currently two human body models are commercially available to the injury biomechanics community; the Total HUman Model for Safety (THUMS) (Iwamoto, Kisanuki et al. 2002; 2010) and the HUman MOdel for Safety (HUMOS) (Robin 2001). The first version of THUMS was released in 2000 with 80,000 elements while the current version, released in 2010, has about 2 million elements. The original geometry for THUMS was from the ViewPoint Datalabs/Digimation (St. Rose, LA, USA) dataset, which employed digitization of physical bones and data from the Visible Human Project. The current model of THUMS includes geometry from supine CT images of an individual selected as a close match to the 50th percentile male. The HUMOS geometry was obtained from a frozen and sectioned male cadaver placed in the seated driving position. As with much cadaver work, the specimen was advanced in age and was also not a good match to the 50th percentile male, so scaling was necessary. While both of these models have provided advancements in human body modeling, they are both created from geometry in the supine position.

A state-of-the-art Full Human Body Finite Element Model, or Full Body Model (FBM), was developed as part of a global design effort, based on medical images of an individual (height – 174.9cm, weight – 78.6±0.77kg, and age – 25.7±0.25 years) chosen to represent the 50th percentile male (Gayzik 2011). Geometry for the model was created using a multi-modality imaging approach that included surface scanning, supine Magnetic Resonance Imaging (MRI), upright MRI, and supine Computed Tomography (CT). Upright MRI was used to obtain anatomical locations of organs with the subject sitting, making the FBM a more accurate representation of a seated occupant.

The aim of this study is to describe the validation of the FBM of the Global Human Body Models Consortium (GHBMC) 50th percentile male seated occupant model (M50) in lateral sled and drop tests. Computational models require validation for confidence in simulation results. In this study, validation involves matching a combination of force traces and injury prediction measures reported in the studies that are being simulated to the model outputs. This work also focuses only on gross measures of the model validation, meaning reaction loads measured during the tests. Regional validation of the M50 model has also been conducted and is ongoing. (Li 2010; Fice 2011; Yue 2011; Beillas 2012; DeWit 2012)

METHODS

The model being validated in this study is the 50th percentile male model (M50) from the Global Human Body Models Consortium (GHBMC). The computer aided design (CAD) geometry used for the model was derived from medical images of an individual selected to represent the 50th percentile male. The selection process and a detailed description of the imaging protocol can be found in the literature (Gayzik 2011; Gayzik 2012). The CAD geometries were meshed by partner universities on a region-by-region basis, including the Head, Neck, Thorax, Abdomen, Pelvis, and Lower Extremities. The current version of the model contains a total of 1.3 million nodes, 1.95 million elements, 847 parts, and weighs 75.5 kg. Rib fracture is predicted using a piecewise linear elastic material model with a failure strain, 0.02 for cortical bone, at which elements are deleted (Li 2010). Similarly, pelvic fracture is predicted by deleting elements past a prescribed failure strain, 0.03 for cortical bone, using a plastic kinematic material model.

Two studies from the injury biomechanics literature were simulated in the lateral full body validation of the model: lateral sled tests by Pintar, et. al. (Pintar 1997) and lateral drop tests by Stalnaker, et. al. (Stalnaker 1979). The lateral sled tests were conducted at two velocities, 6.7 m/s and 8.9 m/s and employed a Heidelberg-type side impact sled to create a left sided impact (Kallieris 1981). Only tests that were conducted without padding were modeled to remove the confounding factor of modeling the foam. Force vs. time data for these experiments was obtained from the National Highway Traffic Safety Administration's (NHTSA) online biomechanics database. The raw literature data was filtered to SAE CFC 180, truncated so time zero occurred at the onset of thorax loading, and mass-scaled to 76 kg using the method described by Eppinger (Eppinger 1976). The lateral drop test that was used for validation occurred from 1.0 m into an instrumented rigid wall. Force vs. time traces were obtained from the literature (Stalnaker 1979) and a force vs. time corridor was obtained from ISO TR 9790 (1999). The literature data was filtered to SAE CFC 180. In both setups the walls were modeled as perfectly rigid.

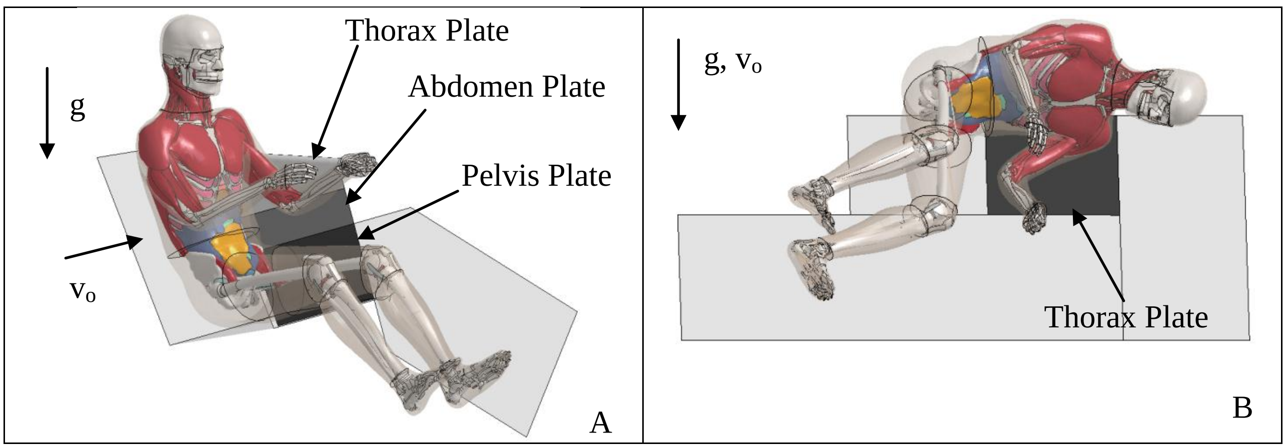

Figure 1: Setup for A) sled tests and B) drop test with gravity (g) and initial velocity (v_o) indicated.

Model results were compared to force vs. time curves in the sled tests and the force vs. time data from the drop test. Average experimental data was compared to model outputs using a method described by Sprague and Geers (Sprague 2004), which calculates a comprehensive error term attributed to the magnitude and phase. The sled tests utilized thorax, abdomen and pelvis force plates, the locations of which can be seen in Error! Reference source not found.A. One thoracic force plate was used in the drop test, the location of which was determined by the study description as seen in Error! Reference source not found.B. Additionally, several common injury prediction values were calculated from model results and compared to those from the literature. The model values of maximum chest deflection in two locations (the axilla and rib 8 at the lateral-most portion) and percent compression in both locations were compared to the sled test literature. Both rib deflection (at the axilla) and peak pelvic acceleration were calculated for the drop tests and compared to values given in the literature. The number of fractures and concurrent Abbreviated Injury Scale (AIS) score predicted by the model were also compared to those seen in the experiments.

RESULTS

Model Kinematics

Model kinematics looked reasonable in all three simulations. As would be expected, there is more head excursion and arm flail in the 8.9 m/s sled test than the 6.7 m/s sled test. Comparing the kinematics shows that the drop test is the least severe of the three cases with less head excursion and arm flail than the other two cases. Kinematics can be seen in the frames shown in Figure 2 for the three model runs. Note that the drop test is shown with the body vertically oriented and with gravity applied in the horizontal direction displayed by the arrow.

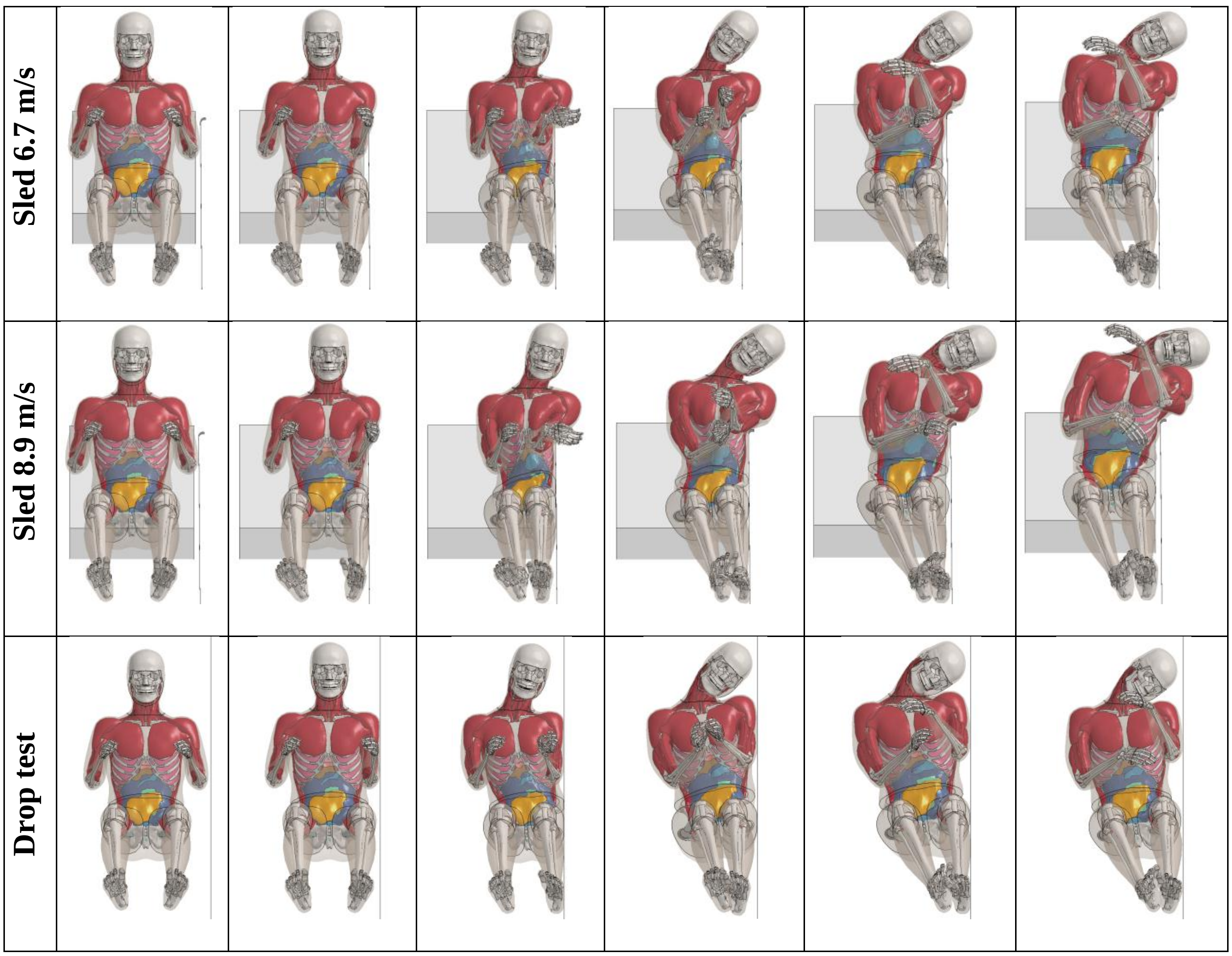

Figure 2: Frames showing model kinematics for the three validation tests. Note gravity is applied in the rightward direction for the drop test.

Force and Deflection Comparisons

The model outputs for force vs. time in the 6.7 m/s sled test can be seen alongside the experimental results in Figure 3. Model prediction for peak thoracic, abdominal, and pelvic forces were 8.7 kN, 3.1 kN, and 14.9 kN compared to 6.0±1.1 kN, 3.6±1.3 kN, and 5.1±2.5 kN, respectively. The peak chest compressions in the model were 8.8 cm and 6.1 cm in the upper and middle chestbands while experimental results were 12.5±2.0 cm and 9.3±0.75 cm. In terms of percent of chest width, the peak compressions predicted by the model were 27.4% and 20.0% in the upper and middle chestbands with matching experimental values of 36.1±4.7% and 29.4±0.85%.

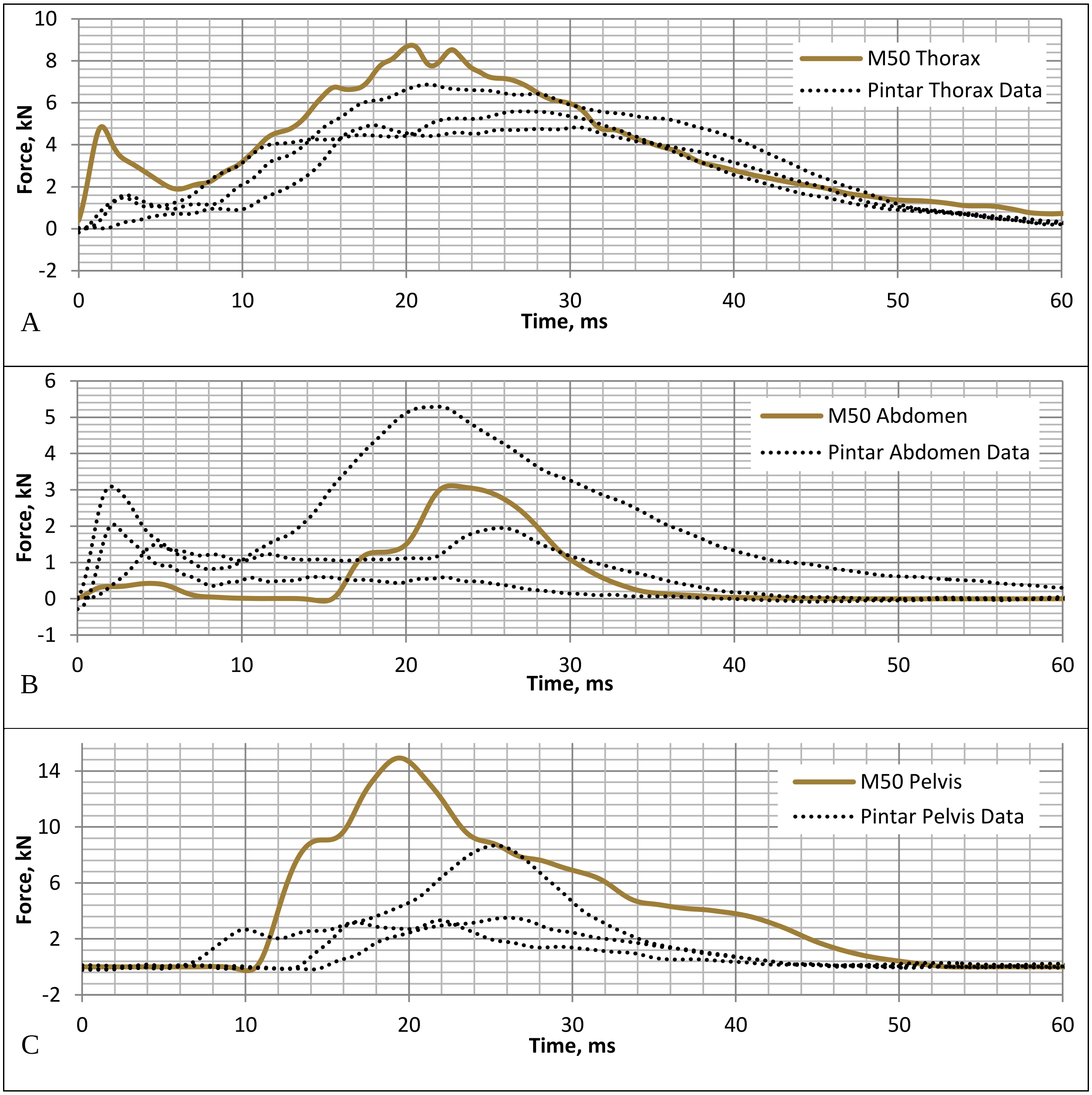

Figure 3: Force vs. time model outputs compared to data for the 6.7 m/s case in the A) thorax, B) abdomen, and C) pelvis.

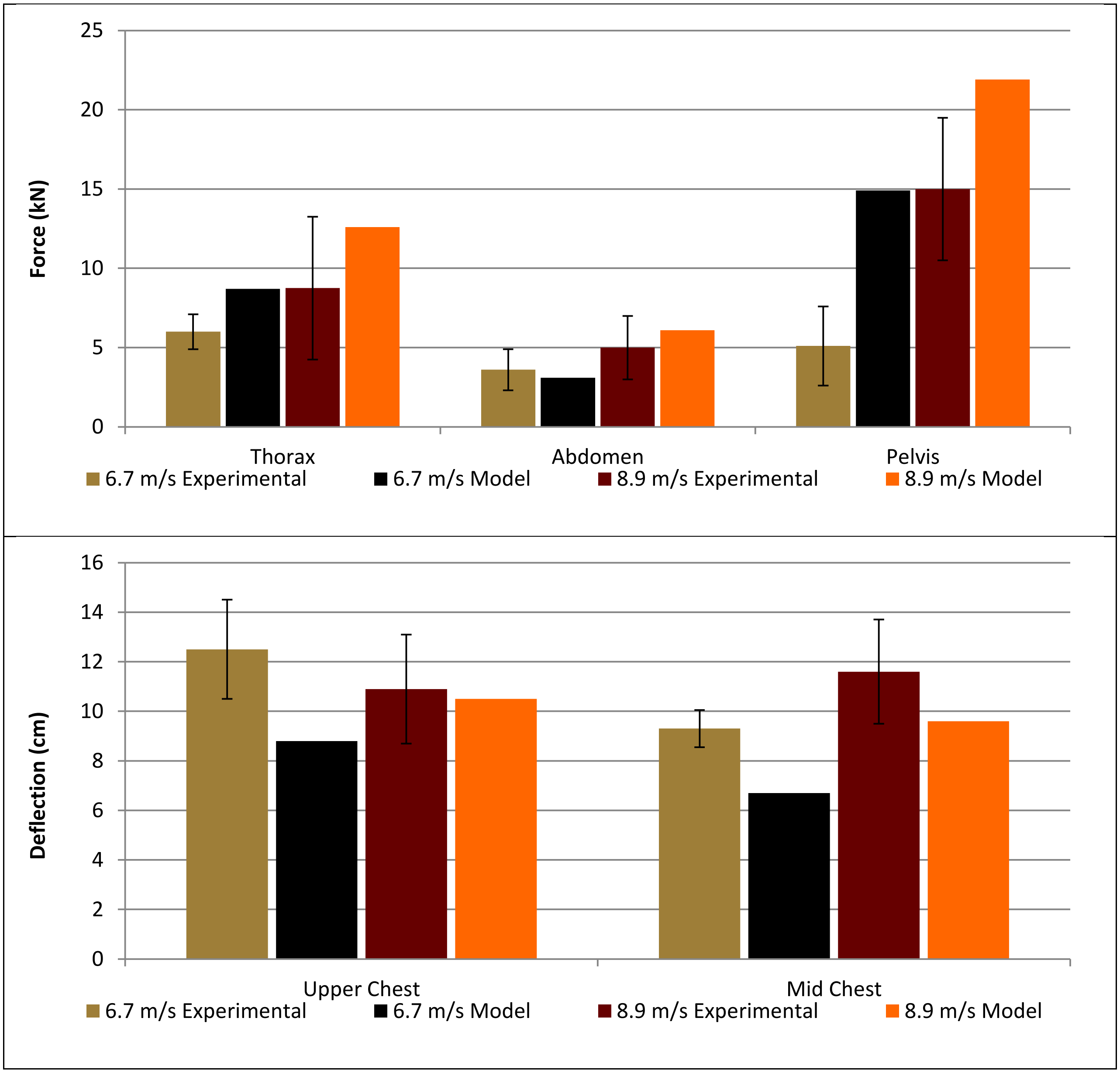

Figure 4: Peak values, for A) force and B) chest deflection, comparing experimental and model.

The model outputs of force vs. time in the 8.9 m/s sled test can be seen compared to the experimental results in Figure 5. Model predictions for peak thoracic, abdominal, and pelvic forces were 12.6 kN, 6.1 kN, and 21.9 kN compared to experimental values of 8.75±4.5 kN, 5.0±2.0 kN, and 15.0±4.5 kN, respectively. The peak chest compressions predicted by the model were 10.5 cm and 9.6 cm in the upper and middle chestbands while experimental results were 10.9±2.2 cm and 11.6±2.1 cm. Peak chest deflections as a percentage of chest width predicted by the model were 32.3% and 28.7% in the upper and middle chestbands compared to values of 34.8±7.0% and 36.8±6.7% in the experiments.

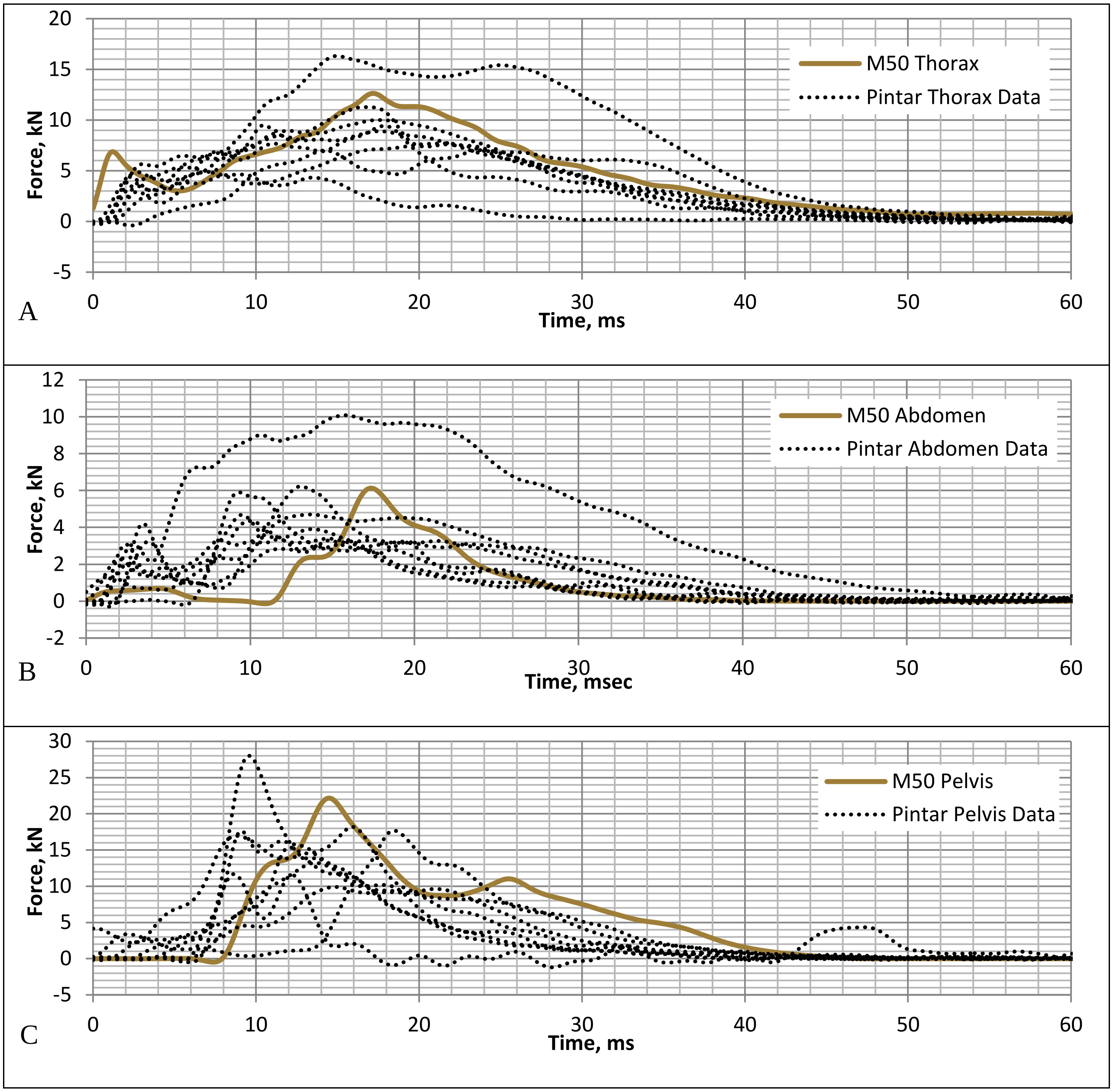

Figure 5: Force vs. time model outputs compared to data for the 8.9 m/s case in the A) thorax, B) abdomen, and C) pelvis.

The model output of force vs. time can be seen in comparison with the average data from Stalnaker et. al. (Stalnaker 1979) and corridors from ISO TR 9790 in Figure 6. The peak thoracic load predicted by the model was 6.7 kN with a 40 mm peak rib deflection. Experimental results showed a range of 5.8 kN to 7.4 kN in peak force and 26 mm to 38 mm in peak deflection. The peak pelvic acceleration predicted by the model was 71 g which fit into the experimental range of 63 g to 77 g. These values are also summarized in Table 1.

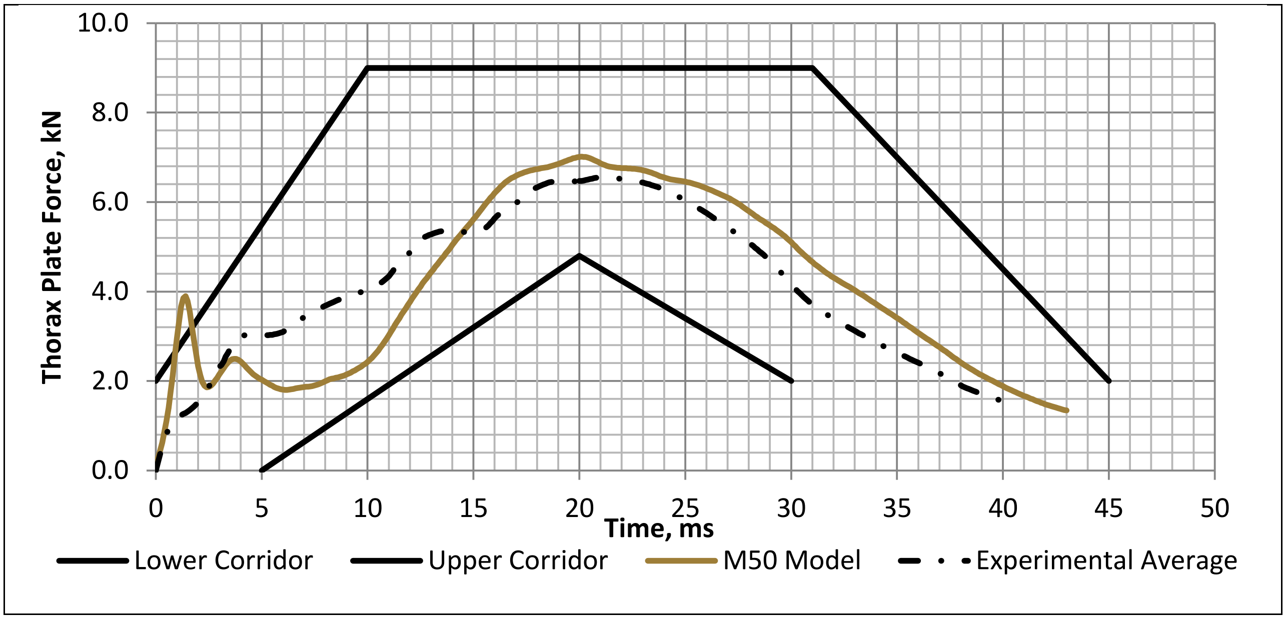

Figure 6: Model output for thorax force in the drop test compared to the corridor given in ISO TR 9790.

Table 1: Peak thoracic force and injury predictors for the drop test with ranges given from ISO TR 9790

	Drop Test - ISO 9790	**GHBMC FBM**
Load on Thorax (kN)	5.8-7.4	6.7
Peak rib deflection (mm)	26-38	40
Peak pelvis acceleration (g)	63-77	71
Rib Fx.	0-5	2

The comprehensive error values were calculated using the methods of Sprague and Geers (Sprague 2004), seen in Table 2. The 6.7 m/s sled test error values showed that the magnitude differences generally contributed more to the error than the phase differences. Values for the sled test at 8.9 m/s showed similar trends in that magnitude contributed more to the total error than phase. The negative M-value of the abdomen forces in both sled tests indicates that the model output was less than the average experimental data. Comprehensive error in the drop test was much less than that found in either sled test. Phase contributed more to the error of the drop test than magnitude.

Table 2: Coefficients calculated from the quantitative comparison of model output to experimental data. M-value represents magnitude contribution to error, P-value represents phase contribution, and C-value is comprehensive error

Validation Case	**Sled Test 6.7 m/s**			**Sled Test 8.9 m/s**			**Drop Test**
Force Location	Thorax	Abdomen	Pelvis	Thorax	Abdomen	Pelvis	Thorax
M-value	0.30	-0.15	2.12	0.24	-0.23	0.56	0.03
P-value	0.08	0.19	0.14	0.07	0.22	0.14	0.09
C-value	0.31	0.24	2.12	0.25	0.32	0.58	0.10

Fracture Prediction

The model predicted one fracture each in ribs 4-7 on the impacted side for the 6.7 m/s sled case, as is seen in **Error! Reference source not found.**, leading to AIS 3 thoracic injury. In the experimental data one subject sustained AIS 4 thoracic injury, one sustained AIS 3 thoracic injury, and the third sustained no thoracic injury (injury determined via fractures). The model also predicted a pubic ramus fracture which was not reported in the literature. The model predicted a total of 15 rib fractured which amounted to AIS 4 thoracic injury in the 8.9 m/s sled test. The literature reported that seven out of eight subjects sustained AIS 4 thoracic injury with the eighth subject receiving AIS 2 thoracic injury. The model also predicted three pubic ramus fractures and a sacral fracture that were not reported in the literature. In the drop test model predictions included two rib fractures, one each in ribs 4 and 5 on the struck side while the literature reported five rib fractures for one male subject and zero rib fractures for the other male subject.

Table 3 : Rib fracture comparison

Sled 6.7 m/s	**Experimental**	**Model**
	0-15	4
Sled 8.9 m/s	**Experimental**	**Model**
	< 3 to > 10	15
Drop Test	**Experimental**	**Model**
	0-5	2

DISCUSSION

The model outputs for force vs. time compared reasonably to the experimental data for all three tests. In the sled tests the model typically predicted forces that were slightly larger than averages from the experiments with one notable exception of the pelvis force in the 6.7 m/s sled test. In this force plate the model predicted forces approximately twice as much as the largest force reported in the literature. Similarly, the pelvic force was larger than the average literature peak by more than one standard deviation, but was not nearly as divergent as in the 6.7 m/s case. This could be due to a difference in the effective mass contacting the particular force plates since the model was not gravitationally settled into the seat as a cadaveric subject would be. In the drop test, the model force vs. time output fit into the corridor well with a small exception early in the simulation. There is a spike in force output from the model, at 1.5 ms, that is approximately 0.75 kN above the corridor. While similar spikes were seen in the data presented in the literature (1999), the spikes were smaller in magnitude and were diluted by averaging the data. The spikes are due to the arm contacting the force plate. From literature descriptions of the experimental setup the arms of the cadavers were close to the body whereas the model arm is away from the body to model a driving position. This would cause the large initial spike in model output force followed by the dip before beginning to rise again, which was not seen as drastically in the experimental data. No large scale repositioning of the GHBMC model was undertaken in this study.

The model tended to under-predict chest compression in the sled tests. Both of the chest compressions in the 6.7 m/s sled test were more than one standard deviation less than the average reported in the literature. In the 8.9 m/s case both chest compressions were below the average, but within one standard deviation. On the other hand, the model predicted rib deflection that was slightly larger than the range given in the drop test literature. The caveat with this range is that it was calculated as a range around one test subject. A ±20% range was created around the deflection of the test subject which did not sustain rib fractures. The range was chosen in ISO TR 9790 without any further explanation. The under-prediction of compression in the sled tests combined with the forces seen indicates that the thorax of the current model is slightly stiff. However, we acknowledge the well-known limitations with using cadaver data and believe the model requires further validation.

While using cadaveric data for validating a human body model is the current standard in the field, there are limitations that coincide with using this data. Cadavers have both morphologic and material characteristics that make them unlike the living humans which any FBM ultimately aims to represent. This is the reason that the GHBMC M50 model was created from the scans of a living subject. It has been well established that many cadavers are older and frailer than an average living human. This is especially true in the ribcage where it has been shown that not only do the material properties change, but so do the cortical thickness and rib angle (Kent 2005) and shape (Gayzik 2008). It has also been shown that elderly people are more susceptible to thoracic injury (Stitzel 2010). In the sled tests, only one subject was close to the age of the model (27 years vs. 26 years, respectively) while the majority of subjects were over 60 years old. The two male subjects comprising the data in the drop test used for comparison were 52 years and 42 years. The morphologic and material differences of the elder subjects are a likely source of differences between the model output and experimental data.

It should be noted that the model prediction of fracture provides some uncertainty to this study. The material model used for the ribs was a piecewise linear plastic model a failure strain defining element deletion (Li 2010). This method for predicting bone fracture removes elements that surpass a given threshold. While this enables post fracture kinematics to be modeled, it is very difficult to precisely predict the timing of fracture, the location of fracture and the post-fracture kinematics. An alternative method of defining fracture would be to allow elements to remain in the model for the entire simulation and correlate the likelihood of fracture to a given the model output (i.e. strain). However, by not deleting elements during the simulation the kinematics are also altered. Loads are more distributed between bones in the vicinity of impact in this case and therefore other potential fractures may not be predicted. While neither model of predicting fracture is ideal, the developers chose to delete elements in an attempt to better capture the total number of rib fractures; however rib fracture data should be analyzed with caution. It is well established that there are many factors that may predispose an individual to sustaining rib fractures including age, bone mineral density, cortical shell thickness, and disease. The number of fractures is more commonly used in assigning injury scores which was a further consideration in choosing this material model.

Future work for this study will include modeling the walls as deformable bodies (composed of steel) rather than perfectly rigid. This is a possible solution to the contact forces being generally higher in both sled tests. Work has already begun on simulations to settle the model before simulating impacts. This may also aid in alleviating some of the larger contact forces between the body and the wall by distributing the model's mass as it would have been during the experiment. Additionally, our group has begun work on chest band validation for the sled tests as well as other validation cases. With the completion of gross external validation, research on soft tissue injuries using various experimentally determined predictors can begin. This work is a component of a range of model and validation efforts currently focused on the M50 model.

CONCLUSIONS

This study presented validation of a state-of-the-art full human body finite element model in lateral sled and drop tests. Two main sources in the biomechanics literature were used in this validation study, Pintar et. al. and Stalnaker et. al. The results indicate that the model output is a reasonably good match to the literature data. This study provides confidence in the model's performance in lateral impacts involving the full body.

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REFERENCES

(1999). ISO TR 9790: Road Vehicles - Anthropomorphic side impact dummy - lateral impact response requirements to assess the biofidelity of the dummy.
(2009). Global status report on road safety. W. H. Organization. Geneva, Switzerland.

(2010). Documentation of Total Human Model for Safety (THUMS) AM50 Pedestrian/Occupant Model, Toyota Motor Corporation.

Beillas, P., Berthet, F. (2012). Performance of a 50th percentile abdominal model for impact: effect of size and mass. European Society of Biomechanics Conference, Lisbon, Portugal.

DeWit, J. A., Cronin, D.S. (2012). "Cervical spine segment finite element model for traumatic injury prediction." Journal of the Mechanical Behavior of Biomedical Materials **10**: 138-150.

Eppinger, R. H. (1976). Prediction of thoracic injuries using measurable experimental parameters. Proceedings of the Sixth International Technical Conference on the Enhanced Safety of Vehicles (ESV).

Fice, J. B., Cronin, D.S., Panzer, M.B. (2011). "Cervical spine model to predict capsular ligament response in rear impact." Annals of Biomedical Engineering **39**(8): 2152-2162.

Gayzik, F., Moreno, D., Danelson, K., McNally, C., Klinich, K., Stitzel, Joel (2012). "External Landmark, Body Surface, and Volume Data of a Mid-Sized Male in Seated and Standing Postures." Annals of Biomedical Engineering: 1-14.

Gayzik, F. S., Moreno, D.M., Geer, C.P., Wuertzer, S.D., Martin, R.S., Stitzel, J.D. (2011). "Development of a Full Body CAD Dataset for Computational Modeling: A Multi-modality Approach." Annals of Biomedical Engineering **39**(10): 2568-2583.

Gayzik, F. S., Yu, M., Danelson, K.A., Slice, D.E., Stitzel, J. D. (2008). "Quantification of age-related change of the human rib cage through geometric morphometrics." J. Biomech **41**(7): 1545-1554.

Iwamoto, M., Y. Kisanuki, et al. (2002). Development of a finite element model of the Total Human Model for Safety (THUMS) and application to injury reconstruction. International Research Council on the Biomechanics of Injury (IRCOBI). Munich, Germany.

Kallieris, D., Mattern, R, Schmidt, G, Eppinger, R. H. (1981). "Quantification of Side Impact Responses and Injuries." SAE: 38.

Kent, R., Lee, S.H., Darvish, K., Wang, S., Poster, C.S., Lange, A.W., Brede, C., Lange, D., Matsuoka, F. (2005). "Structural and material changes in the aging thorax and their role in crash protection for older occupants." Stapp Car Crash J **49**(1532-8546 (Print)): 231-249.

Li, Z., Kindig, M.W., Kerrigan, J.R., Untaroiu, C.D., Subit, D., Crandall, J.R., Kent, R.W. (2010). "Rib fractures under anterior-posterior dynamic loads: Experimental and finite-element study." Journal of Biomechanics **43**: 228.234.

Li, Z., Kindig, M.W., Subit, D., Kent, R.W. (2010). "Influence of mesh density, cortical thickness and material properties on human rib fracture prediction." Medicine Engineering and Physics **32**: 998-1008.

Pintar, F. A., Yoganandan, N., Hines, M.H., Maltese, M.R., McFadden, J., Saul, R., Eppinger, R., Khaewpong, N., Kleinberger, M. (1997). "Chestband analysis of human tolerance to side impact." Stapp Car Crash J **41**: 63-74.

Robin, S. (2001). Human Model for Safety – A joint effort towards the development of redefined human-like car-occupant models. 17th International Conference for the Enhanced Safety of Vehicles. Amsterdam.

Sprague, M. A., Geers, T.L. (2004). "A spectral-element method for modelling cavitation in transient fluid-structure interaction." Int J Numer Meth Engng(60): 2467-2499.

Stalnaker, R., Tarrière, C., Fayon, A., Walfisch, G., Balthazard, M., Masset, J., Got, C., Patel, A. (1979). "Modification of Part 572 Dummy for Lateral Impact According to Biomechanical Data." Stapp Car Crash J **23**: 843-872.

Stitzel, J. D., Kilgo, P.D., Weaver, A.A., Martin, R.S., Loftis, K.L., Meredith, J.W. (2010). "Age Thresholds for Increased Mortality of Predominant Crash Induced Thoracic Injuries." Annu Proc Assoc Adv Automot Med **54**: 41-50.

Vander-Lugt, D., Connolly, T, Bhalsod, D (1999). Vehicle Compatibility - Analysis of the Factors Influencing Side Impact Occupant Injur. Proceedings of the SAE International Congress and Exposition, Detroit, MI, Society of Automotive Engineers.

Yang, K. H., Hu, J., White, N. A., King, A. I., Chou, C. C., Prasad, P. (2006). "Development of numerical models for injury biomechanics research: a review of 50 years of publications in the Stapp Car Crash Conference." Stapp Car Crash J **50**: 429-490.

Yue, N., Shin, J., Untaroiu, C.D. (2011). "Development and validation of an occupant lower limb Finite element model." SAE Technical Paper 2011-01-1128.