Lateral Regional Impact Validation of a Full Body Finite Element Model for Crash Injury Prediction

N.A. Vavalle†‡, D.P. Moreno†‡, A.R. Hayes†‡, J.D. Sitzel†‡, and F.S. Gayzik*†‡

† Wake Forest University School of Medicine, Medical Center Blvd, Winston-Salem, NC 27157
‡ Virginia Tech – Wake Forest University Center for Injury Biomechanics, 573 N. Patterson Avenue Suite 120, Winston-Salem, NC 27101

Abstract

With increasing computing power, finite element models (FEMs) of the human body are becoming more anatomically detailed, and have the potential to serve as powerful tools for injury mitigation research. These models must be validated in specific loading modes to quantify their accuracy. The purpose of this study is to examine lateral hub impact validations of a human body finite element model and to examine the robustness of the model a sensitivity study.

A full body FEM of the 50th percentile male (M50) was developed from the medical images of a living subject. This study focuses on the validation of M50 in a shoulder impact and a pelvic impact. The shoulder impact applied a cylindrical impactor, weighing 23.4 kg, at a velocity of 4.5 m/s while the pelvic impact used a square-faced block impactor, weighing 16 kg, at a velocity of 10.0 m/s. Transient force traces from model outputs were compared to literature data using Correlation and Analysis (CORA) software. Pelvic fractures were predicted in the model via element deletion at a strain of 0.03 and compared to experimental outcomes. A sensitivity study using the pelvic impact was also performed to determine model robustness.

The model showed a good match in the pelvic impact with CORA values of 0.97 (Corridor), 0.98 (Shape), 0.48 (Size), and 1.0 (Phase). CORA values in the shoulder impact were: 0.51 (Corridor), 0.85 (Shape), 0.62 (Size), and 0.09 (Phase). The model output had peak pelvis and shoulder forces of 13.0 kN and 4.6 kN compared to 10.7±2.8 kN and 2.4±0.8 kN in the experiments, respectively. In the pelvic impact the model predicted an ilio-pubic ramus fracture whereas experiments showed a range from no fractures to four fractures, with rami fractures also observed. Results from the sensitivity study indicate that perturbations to the impactor positioning changed peak force by no more than 3% and deflection by no more than 6%. The results of the shoulder impact show that while the energy absorbed values matched well, the peak force in the model was larger than the experimental average. In the pelvis case, both the energy absorbed and the peak force were larger in the model than in the experiments. The sensitivity study determined good model robustness since small input perturbations did not change the outputs greatly. This study represents one component of a range of validation efforts currently focused on M50. This model will become an important tool used by engineers in studying and advancing occupant safety.

Keywords: FEM, Human Body Model, Injury, Validation, Sensitivity Study.

1. Introduction

Every year more than 1.2 million people die worldwide in motor vehicle crashes (MVC’s) and it is estimated that MVC’s will become the fifth leading cause of death by 2030 (WHO 2009). Lateral impacts are of particular concern; in the United States this crash mode represents just 19% of all crashes, but accounts for 32% of fatalities (Vander-Lugt 1999). Occupants on the side of the impact in a lateral collision (near-side occupants) are at an especially great risk when there is intrusion of the impacting vehicle (Yoganandan 2007). Three of the four most commonly injured body regions in side impact are the chest, lower extremity, and upper extremity (Gabler 2005; Yoganandan 2007). For these reasons, it is important that any predictive model for use in studying MVC injury be validated not only in frontal impact situations, but also lateral impacts, and especially in the body regions at highest risk for injury.

Modeling is a common tool used in studying blunt injury biomechanics (Yang et al. 2006). Particularly, finite element modeling (FEM) is

*Corresponding author. Email: sgayzik@wakehealth.edu
applicable since it can approximate both the complex geometry and nonlinear material properties of the human body. However, before a model can be used in this matter it must first be validated. Validation of human body models often involves comparison to experimental results from post mortem human surrogate (PMHS) testing. Corridors that give an expected range of outcomes can be determined from experimental results and in turn can be used to assess if a model fits within expectations. Mathematical methods that provide firm quantitative comparisons of model outcomes to the average experimental results can be used as well. However, it is not only important to develop well-validated human body FEMs, but also robust models. When models are robust, they can not only give solutions for a variety of impact scenarios, but also provide only small differences in solutions when inputs are slightly perturbed. This study presents the validation of the Global Human Body Models Consortium (GHBMC) 50th percentile male model (M50) in two regional lateral loading scenarios. The loading includes a pelvis impact and a shoulder impact. These regions are commonly injured in side impacts (Gabler 2005; Yoganandan 2007) and represent a spectrum of locations from the upper extremity through lower extremity. Both qualitative and quantitative comparisons are made between the model and the experimental data being validated against using corridor comparisons and Correlation and Analysis (CORA). Additionally a small sensitivity study was completed using the pelvic impact. The cases presented here are a subset of a larger series of simulations used in validating the M50 model.

2. Materials and Methods

2.1. Model Development

A state-of-the-art full human body finite element model, or full body model (FBM), was developed as part of a global effort, based on medical images of an individual (height – 175cm, weight – 78.6 ±0.77kg, and age – 26 years) chosen to represent the 50th percentile male. The development of the model has been documented in the literature (Gayzik et al. 2011; Gayzik et al. 2012; Vavalle et al. 2013); only a brief discussion of this will be included here. The model was developed within a consortium of automotive manufactures known as the Global Human Body Models Consortium (GHBMC). Geometry for the model was created using a comprehensive multi-modality imaging approach that included surface scanning, supine MRI, upright MRI, and supine CT of the recruited subject (Gayzik et al. 2011). 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. A region specific validation approach was taken to model development (Li et al. 2010a; Li et al. 2010b; Fice et al. 2011; Beillas and Berthet 2012; DeWit and Cronin 2012; Shin et al. 2012; Cronin 2013; Untaroiu et al. 2013) wherein university consortium members developed regional models that were then integrated into one full body model. The version of the full body model used in this study (FBM v. 4.0) contains 1.3 million nodes, 2.2 million elements, 965 parts, and represents a weight of 76.9 kg. The model is designed to predict a range of injuries, including bone fracture. Cortical pelvis fracture is predicted using a plastic kinematic material model with a failure strain of 0.03, past which elements are deleted.

2.2. Regional Lateral Validation

Two regional lateral loading regimes were simulated in this validation study, a pelvic impact (Bouquet 1998) and a shoulder impact (Koh et al. 2005). The pelvic impact involved a square-faced impactor weighing 16 kg impacting with 800 J of energy (n=4 subjects). This required propelling the impactor at 10 m/s. The shoulder impact used a 23 kg cylindrical hub impactor (n=6 subjects). This impact occurred at a nominal velocity of 4.5 m/s. The cadavers subjected to pelvic and shoulder impacts were in a seated position. In the pelvic impact, the seat was coated in Teflon to reduce the effects of friction between the cadaver and the seat. This was modeled as a frictionless contact during these simulations. The pelvis impactor contacted the trochanter and iliac crest, according to the literature. The shoulder impact occurred through the head of the humerus. In both cases, experiments utilizing unpadded impactors were chosen to reduce the modeling complexity, ensure accurate boundary conditions, and maximize efforts on the FBM development rather than matching the foam padding used the experiment. The simulation environments mimicking these experiments can be seen in Figure 1.

Figure 1: Simulation setup for A) pelvic impact and B) shoulder impact, with gravity (g) and initial impactor velocity (v_0).

The model remained in the seated driver occupant position during all simulations. The impactors were modeled as rigid bodies with the appropriate mass assigned to the part. Initial velocities were applied to the impactors and in the pelvis impact a gravitational load was applied to the full body model. Gravity was excluded from the shoulder
impact since it would have little effect on simulation outcome and only add to computation time. Reaction force in the impactor was output from the model for comparison to response corridors determined in the literature. In the pelvis impact, pelvic fractures predicted by the model were determined through visual inspection and compared to those found experimentally. Since only clavicle and acromion fractures were seen in the shoulder impact and the model is not programmed to predict clavicle or scapula fracture, this part of the study did not focus on fracture. However, the model is designed to predict rib fractures so simulation outputs were used to confirm that no rib fractures were predicted in this case. Additionally, Abbreviated Injury Scale (AIS) scoring was used to compare injury severities (AIS Revision 2008).

In addition to qualitative analyses, we have made quantitative comparisons of the model to experimental data using correlation and analysis (CORA). This method is commonly used in injury biomechanics (Gehre et al. 2009). The strength of this method is that it can give independent scores for how well the model response fits the corridor, shape, size, and phase of the experimental data. For all four metrics of CORA, the best score is a “1” and the worst score is a “0.” A more detailed explanation of the method can be found in the literature (Gehre et al. 2009). Transient force curves from both impacts were used as the input for comparison via CORA. Cross-plots which are not monotonically increasing (such as force vs. deflection) cannot be assessed using CORA and must be broken down into the individual components.

2.3. Location Sensitivity

In order to test the robustness of the model, a sensitivity study was performed. If a model is robust, then small perturbations in the input boundary conditions will amount to small changes in the output of the model. To examine this in the GHBMC M50, the pelvic impact was used. The location of the impactor was moved a distance of 12.5 mm in the anterior, inferior, posterior, and superior directions (Figure 2). The distance was calculated by taking 25% of the diameter of the femur at the trochanter, since the impactor location was based on this landmark. The peak forces and deflections from the four variations were compared to the nominal location case by looking at percent change.

Figure 2: Location deviations examined in the sensitivity study.

3. Results

3.1. Regional Lateral Validation

The model simulated both the pelvis impact (40 ms) and the shoulder impact (60 ms) to completion. The simulations can be seen at peak force in Figure 3.

The force vs. displacement results from the pelvis impact can be seen compared to experimental corridors in Figure 4. The model output remains within the experimental corridors until approximately 40 mm of deflection at which point the model force output peaks slightly above the corridors. The peak force occurs with a higher deflection in the model than in the experimental corridors. The peak force found in the model was 13.0 kN while the peak found experimentally was 10.7±2.75 kN. In this impact the model predicted an ilio-pubic ramus fracture. PHMS results from this impact indicated a range of no fracture to four fractures, with rami fractures observed as well. The model predicted fracture amounted to an AIS 2 injury while the experimental subjects sustained Maximum AIS injuries ranging from 0 – 3.

![Figure 3: Pelvic and shoulder impacts shown at 0, 1/3 t_c, 2/3 t_c, and t_c, where t_c is the termination time of the simulation.](image-url)
The model peak force output in the shoulder impact was 4.6 kN while the peak force from experimental results was 2.5±0.15 kN. Model output for force vs. time can be seen plotted with experimental corridors in Figure 5. Overall the match in the shoulder impact is not as strong as in the pelvis impact with model force output peaking earlier and higher than the experimental corridors.

The results of the CORA analysis can be seen in Table 1. Overall, the pelvis impact received better values than the shoulder impact with the exception of the size value. The shape values of both impacts showed strong matches to the data, indicating that the model is able to generate similar global curve characteristics. The model proved to have a relatively large phase difference in the shoulder impact which led to a lower corridor score than if the phases had been more in line with one another. This is evidenced by the early peak in response curve in Figure 5.

<table>
<thead>
<tr>
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<th>Pelvis</th>
<th>Shoulder</th>
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<tbody>
<tr>
<td>Corridor</td>
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<tr>
<td>Shape</td>
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<tr>
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<td>0.62</td>
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<tr>
<td>Phase</td>
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### 3.2. Location Sensitivity

The results of the sensitivity study with the pelvis impact can be seen summarized in the radar plot shown in Figure 6. Overall, the model showed good robustness with all peak values of both force and deflection within 6.1% of the nominal case. The greatest difference was seen in the deflection of the inferior case (6.1%) while the smallest difference was seen in the deflection of the superior case (-0.04%). In all four cases, the peak forces were larger than the nominal case, ranging from 0.2% (anterior) to 3.1% (superior). The peak deflection ranged from -1.6% to 6.1% change from the nominal case.

![Figure 6: Results of the sensitivity study showing peak force and deflection as a percent difference from the nominal case.](image)

### 4. Discussion

This study aimed to present the validation results of a full body finite element model in regional lateral loading. The loading conditions, which represent those of interest in side impact MVC's, included a pelvic impact and a shoulder impact. Model response in the pelvic impact proved to be a good match to experimental data. This is seen in both the comparison of force vs. deflection curve with the corridors and in the CORA scores in Table 1. The model was not as well correlated in the shoulder impact, but still received a strong CORA score for
shape. The phase score of this simulation indicates that the phase difference contributed significantly to a poor corridor score. While CORA does combine these scores into an overall biofidelity score, this shows the value of also considering the four metrics separately. Having four pieces of information rather than one allows the engineer to more easily identify the source of the differences. This is especially useful during the development of models; CORA results can be used to direct further model improvement. For instance, the low phase score indicates that the timing of the peak is off (too early). Updates to model will be focused on reducing this through material or contact adjustments.

A common limitation in human body finite element model validation studies is the use of cadaver data for validation. PMHS lack muscle tone, require reprofusion of fluids, may have relevant comorbidities, and often do not reflect a general population because they are often of an advanced age. The shortcoming of this data is that the simple mass scaling of cadaver data does not account for geometric or material differences between elderly cadavers and younger living human beings (Gayzik 2008). Therefore, scaling the data to the desired mass does not necessarily correspond to what it would be in a living person of that mass. However, PMHS testing is currently the gold standard for the injury biomechanics field and is the only widely available source of experimental data.

A further limitation of this study is that no preimpact gravitational settling was simulated. This would not be expected to change the outcome of the shoulder impact, but in the pelvic impact it could have potential ramifications. Since more flesh mass would be located between the impactor and the bony geometry of the pelvis in a settled case, the material properties of the flesh would dominate the resulting kinematics and kinetics more. Future work will include investigating the effects of settling on the pelvic impact.

It should be noted that the model prediction of fracture also provides some uncertainty to this study. The material model used for the pelvis was a plastic kinematic model a failure strain defining element deletion. 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 themselves. 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. While neither model of predicting fracture is ideal, the developers chose to delete elements in an attempt to better capture the total number of fractures; however, fracture data should be analyzed with caution.

5. Conclusion

This study presented the validation of the Global Human Body Models Consortium's mid-sized male model in regional lateral impacts. The impacts used in this study, pelvis and shoulder impacts, represent commonly injured areas of the body during side impact motor vehicle crashes. It is important to validate human body models in an omnidirectional manner to ensure accurate results in a wide array of impacts. These two simulations represent a subset of the full suite of scenarios used to validate the GHBMCM 50. It was found that the model output for peak force was larger than the experimental averages in both impact simulations. However, the model was within the established experimental corridors and one standard deviation of peak force for the pelvis impact. Additionally, the model showed good robustness in the sensitivity study results in which changes of no more than 6.1% were seen in the peak forces and deflections. This model will become an important asset to those studying injury biomechanics and this study is an important milestone in the development of the model.

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