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DEVELOPMENT AND PRELIMINARY VALIDATION OF CHESTBAND DATA FROM A FULL BODY FINITE ELEMENT MODEL

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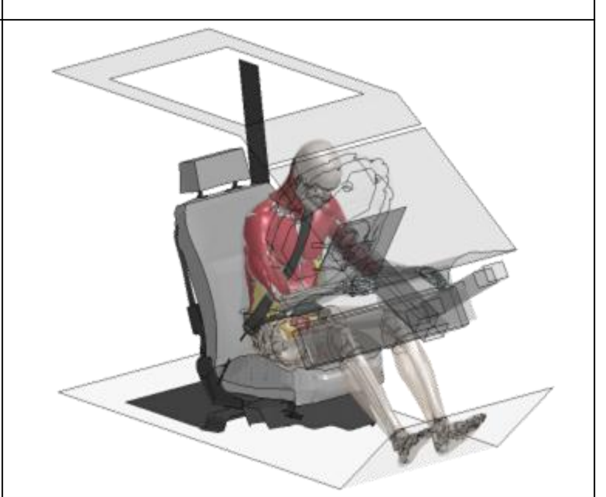
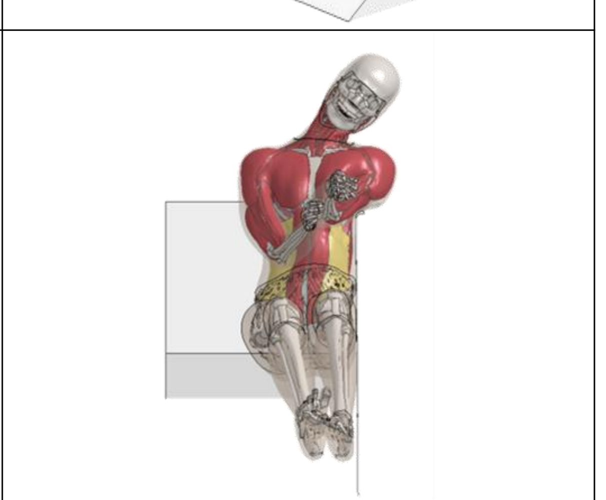
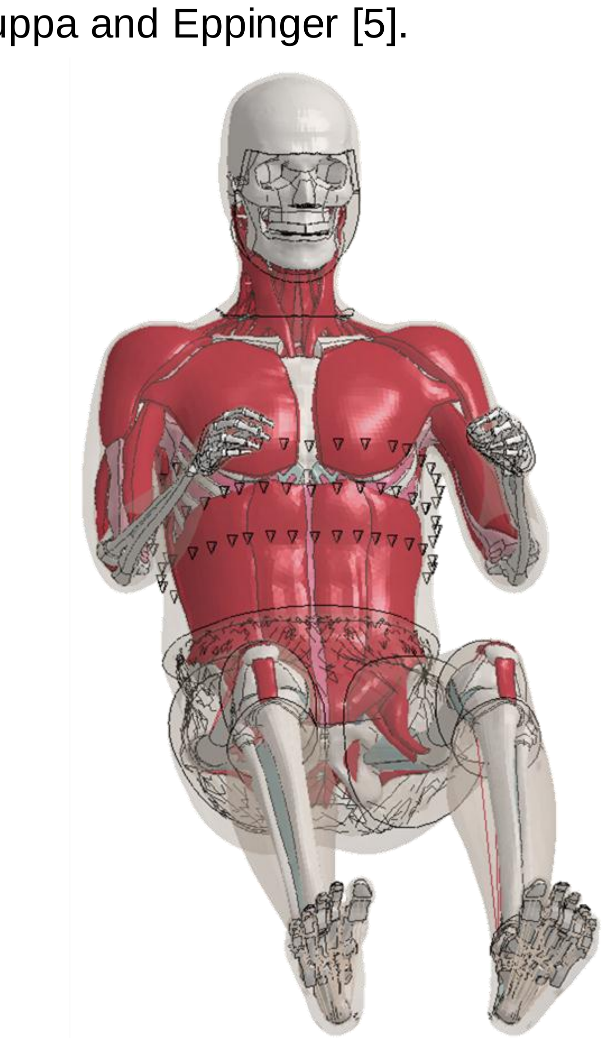
Introduction

More than 1.2 million people die world-wide every year as a result of automobile accidents, and 25% of these deaths are attributed to thoracic injury [1, 2]. Given the statistics regarding injury and fatalities associated with vehicle crashes, finite element (FE) computer models are an emerging tool to examine the thoracic response of the human body in the simulated environment. Validation against experimental studies is essential to ensure biofidelity of the model. In this study, a new human body model, the Global Human Body Models Consortium (GHBMC) mid-sized male was used to examine chestband contour deformations in frontal and lateral impacts. Chestbands provide deformation contours in a given plane during impact. The purpose of this study is to present a methodology for extracting chestband data from a full body FE model, and to compare the model's results in frontal and lateral impacts to experimental results using chest bands.

Methods

Post Mortem Human Subjects (PMHS) data from two studies were used to compare chestband data from the Global Human Body Models Consortium (GHBMC) mid-sized male model. The GHBMC model was run using LS-DYNA (LSTC, Livermore, CA, R. 4.2.1) to simulate the frontal and lateral impacts from both studies [3, 4]. The model was pre-programmed with an upper, middle, and lower chestband each comprised of 32 nodes. The chestbands were placed around the circumference of the chest approximately at the level of the 4th, 6th, and 8th rib, and matched the description of chestband locations in the literature. One local coordinate system was defined per chestband using nodes on the chestband. The nodal data from the GHBMC model were exported to MATLAB (The Mathworks, R 10). Node locations on the chestbands at the maximum deflection state for the GHBMC model were plotted. Maximum deflection was determined using the methods of Kuppa and Eppinger [5].

Cases	
Frontal Impact Driver Position 13.3 m/s	
Frontal Impact Passenger 8.1 m/s	
Lateral Impact Into Rigid wall 8.9m/s	



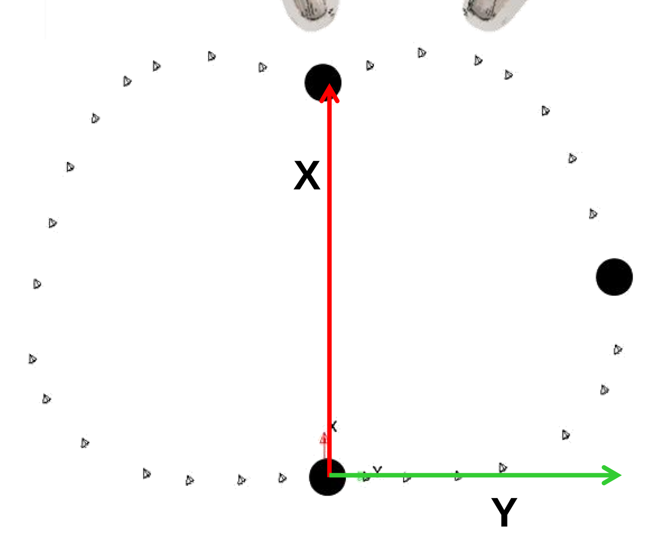


Figure 1. Simulation Cases (Left), GHBMC M50 Model with Chestbands (Right, Upper), and Coordinate System Definition (Right, Lower)

Results

The response at time equal to 0, 1/3 t_{max}, 2/3 t_{max}, and t_{max} of the simulation can be qualitatively examined for one of the frontal and lateral cases in Table 1 and Table2. The results of chestband contours is summarized in Figures 2 through 4, and Table 3.

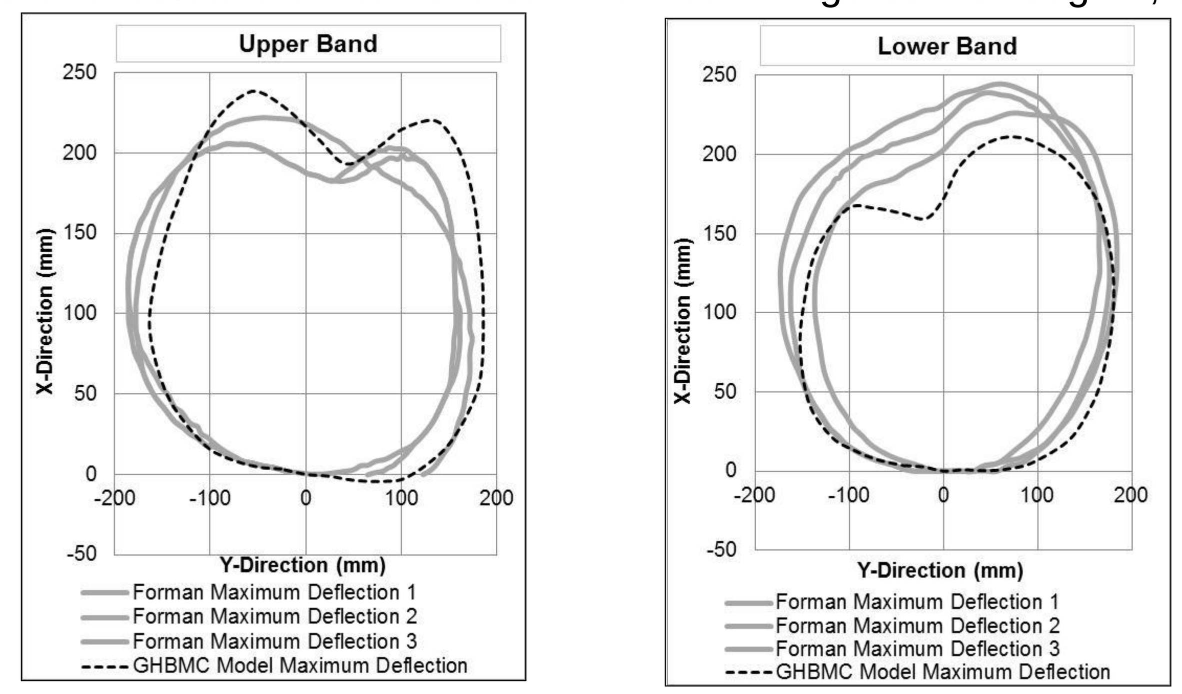


Figure 2. Maximum Chest Deflection of the Upper (left) and Lower (right) Band in the GHBMC model and Forman et al 29km/h frontal passenger impact.

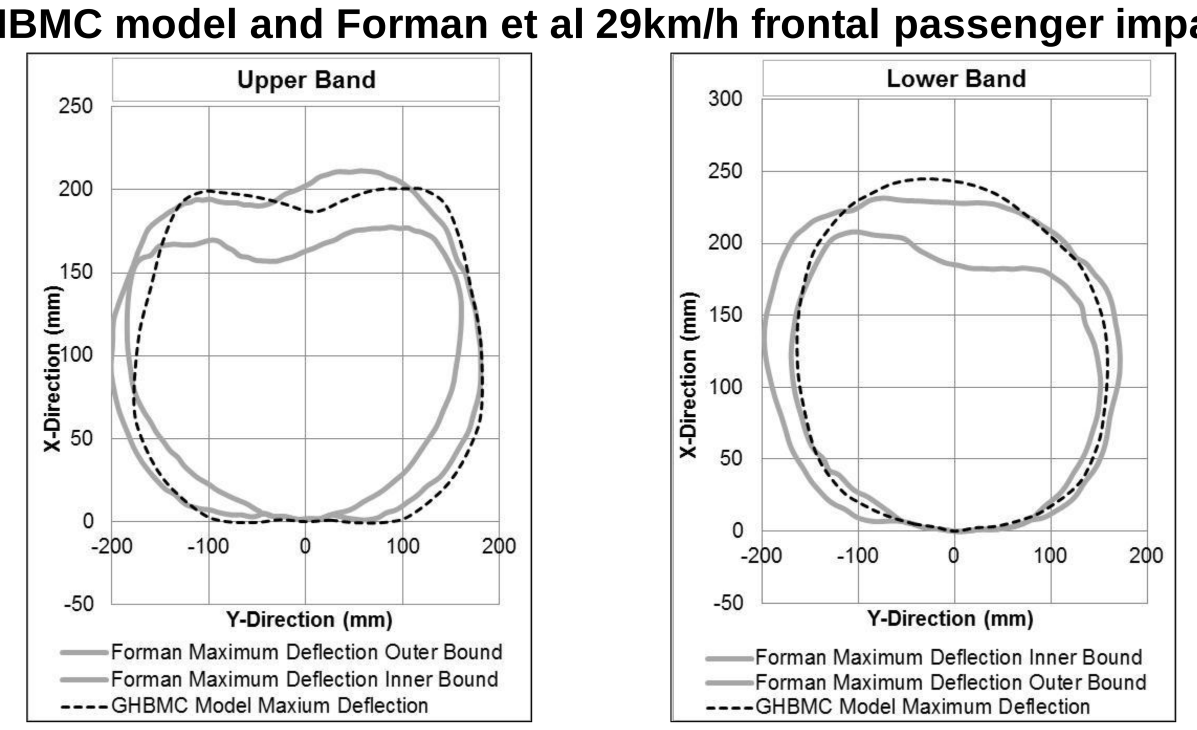


Figure 3. Maximum Chest Deflection of the Upper (left) and Lower (right) Band in the GHBMC model and Forman et al 29km/h frontal driver impact.

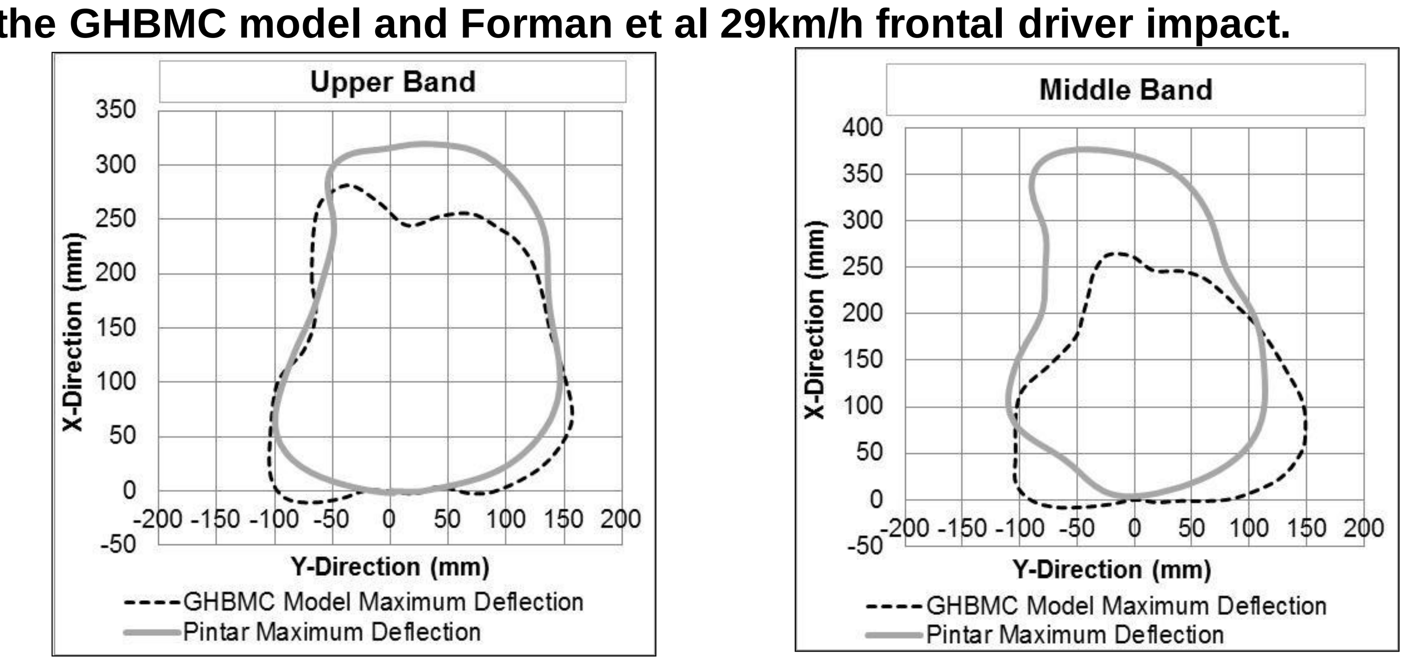


Figure 4. Maximum Chest Deflection of the Upper (left) and Middle (Right) Band in the GHBMC model and Pintar et al lateral impact.

Table 1. Frontal Chestband Simulated Case

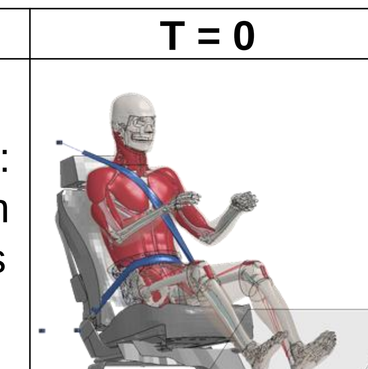
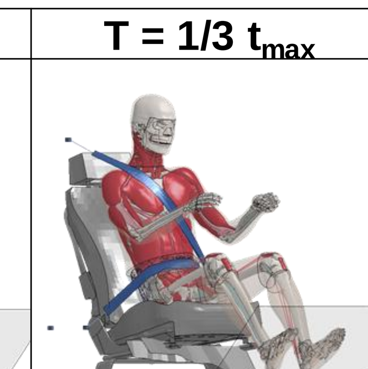
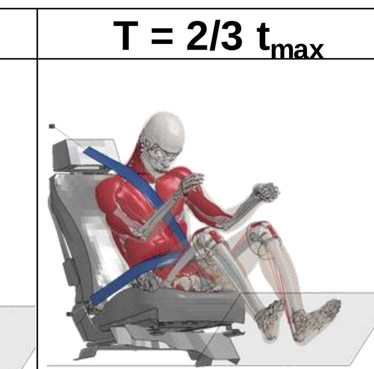
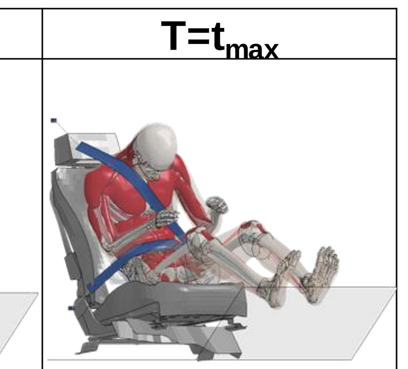
Cases	T = 0	T = 1/3 t_{max}	T = 2/3 t_{max}	T=t_{max}
Case 1: Forman 8.1 m/s [3]				

Table 2. Lateral Chestband Simulated Case

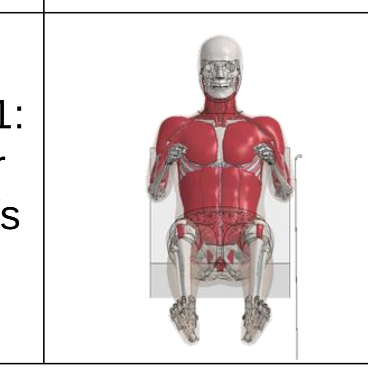
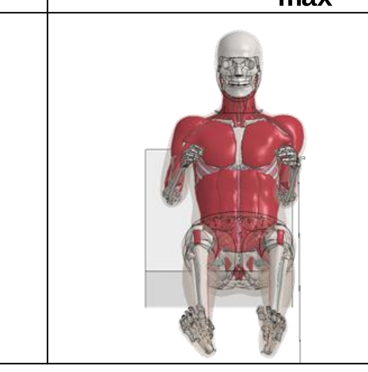
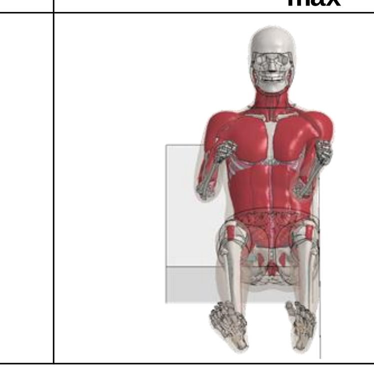
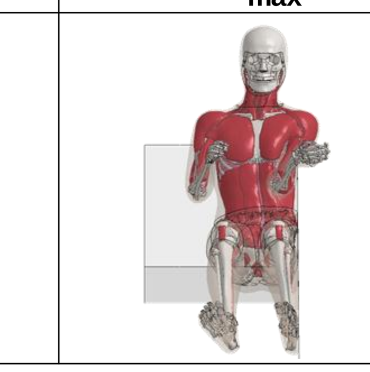
Cases	T = 0	T = 1/3 t_{max}	T = 2/3 t_{max}	T= t_{max}
Case 1: Pintar 8.9 m/s [4]				

Table 3. Percent Compression of Chestband in Model and Literature

Cases	Chestband Location	Model: % Compression	Literature: % Compression
Forman 8.1 m/s	Upper Chestband	5.2%	16±5.6 %
	Lower Chestband	16.1%	7±1.8 %
Forman 13.3 m/s	Upper Chestband	12.7%	23±5.6 %
	Lower Chestband	16.9%	12±6.9 %
Pintar 6.7 m/s	Upper Chestband	28.8%	36.1±4.7 %
	Middle Chestband	27.3%	29.4±0.9 %
Pintar 8.9 m/s	Upper Chestband	36.2%	36.0±6.0 %
	Middle Chestband	30.6%	36.8±6.7 %

Discussion

When comparing the GHBMC model chestband results to both the Forman frontal impact cases (8.1 meters/second(m/s) and 13.3 m/s) [3], there are strong similarities in shape that are clear functions of the belt path. For the frontal sled data at 8.1 m/s, the GHBMC model predicted peak deflection of the upper and lower chestband to be 5.2% and 16.1%, compared to the literature which reported a peak deflection 16±5.6% and 12±6.9%. For the frontal sled data at 13.3 m/s, the GHBMC model predicted the upper and lower chestband peak deflection to be 12.7% and 16.9%, again lower than the literature data. When examining the lateral sled case, closer agreement between the model and lateral case was observed. The somewhat larger discrepancies observed in frontal thoracic loading cases may indicate an overly stiff response in this region, however further investigation is required. For instance, whereas the lateral cases relied on a simplified rigid boundary, the frontal cases have many more variables including seat belt models, which play a role in the chest deflection. While this work was focused on the technique used to extract chestband data from the model, these initial results will also be used to further validate the GHBMC model.

Conclusion

The focus of this work was to determine a technique to extract chestband data. The results presented will be used to further validate the model. For example, this study indicates that the compliance of the thorax in frontal impact may need to be increased in subsequent iterations. Further investigation of simulated restraint systems will also be conducted. Ultimately, the GHBMC M50 model will be a new tool for engineers to improve safety systems and help mitigate the toll of blunt trauma.

Acknowledgments

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References

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