Comparison of thorax and upper-extremity responses between GHBMC M50-OS and M50-O in a high-speed, rear-facing, frontal impact

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I. INTRODUCTION

Following a transition to self-driving vehicles in the future, non-traditional seating configurations and reclined seatbacks may be implemented in vehicle cabins for occupant comfort. One such configuration is the “campfire” arrangement, where front-row occupants are rear-facing (RF) [1]. To understand RF occupant responses in high-speed frontal impacts, Kang *et al*. [2-3] performed sled tests on six post-mortem human subjects (PMHS) seated on 25-degree Honda Accord seats with conventional restraints. Rib fractures were found to be the most common injuries in these PMHS. In addition to skeletal and organ kinematics, seat interaction and bone deformation patterns may be obtained using finite element (FE) simulations with human body models (HBMs), which are otherwise difficult to obtain from PMHS tests. Global Human Body Models Consortium (GHBMC) 50^{th} male (M50) – detailed (O) model has a detailed representation of the human body, as well as injury prediction capability. The GHBMC M50 – simplified (OS) model is limited in its injury prediction capability [4], but is nearly five times computationally faster than the M50-O, making it time- and cost-effective for safety evaluation in different seating configurations. The goal of this study was to compare the thorax and upper extremity responses of the GHBMC M50-OS v2.3 and M50-O v6.0 in a high-speed RF frontal impact (delta V = 56km/h) using biomechanical data from PMHS.

II. METHODS

GHBMC models were positioned on a FE Honda Accord seat model (TS-Tech Americas, Inc.), using PMHS information. Sled boundary conditions from the PMHS tests were replicated in the simulations, and a high-severity frontal pulse (delta V = 56km/h) was applied to the sled floor with the RF seat. Simulations were run using LS-DYNA v11.0. PMHS biomechanical corridor for chest deflection in the anterior-posterior (AP) direction, obtained from a chestband in the tests, was generated using the method outlined in Kang *et al*. [5]. AP chest deflections in the GHBMC models were obtained from pre-defined upper-chestband node sets on the skin. Fig. 1. shows AP chest deflections, mass normalised to the M50 and subsequently normalized by half chest depth. Thorax responses of the GHBMC models were optimally aligned to the PMHS mean response via dummy phase shifts (DPS), and BioRank scores were evaluated over a time duration of 100 ms, using the PMHS corridor and the most recent NHTSA Biofidelity Ranking System (BRS). A BRS score under 2 indicates good biofidelity [6]. XZ-trajectories of left mid-humerus nodes in the GHBMC models were compared to PMHS-21 [3] XZ-trajectory obtained from a motion block sensor on the left mid-humerus.

III. INITIAL FINDINGS

The time history for normalised chest deflection in the M50-OS was consistent with the PMHS, and the peak coincided with the PMHS mean, indicating good biofidelity (BRS=1.36) (Fig. 1). A smaller peak normalised chest deflection and a large deviation from the PMHS corridor was observed in the M50-O in the ramping phase, which resulted in a poor biofidelity (BRS=2.61). Arm flailing was observed in the M50-OS as indicated by the XZ-trajectory of the left mid-humerus, similar to the PMHS on the left side (Fig. 2). The right arm of the PMHS was obstructed by metal components in the support structure, which may have influenced right arm kinematics. For simplicity, these components were not included in the simulations. By contrast, arm flailing was limited in the M50-O as compared to the PMHS and M50-OS (Fig. 2). Shoulder-belt slip-off was seen in the PMHS and GHBMC models, but the slip-off was limited in the M50-O due to limited arm flailing.

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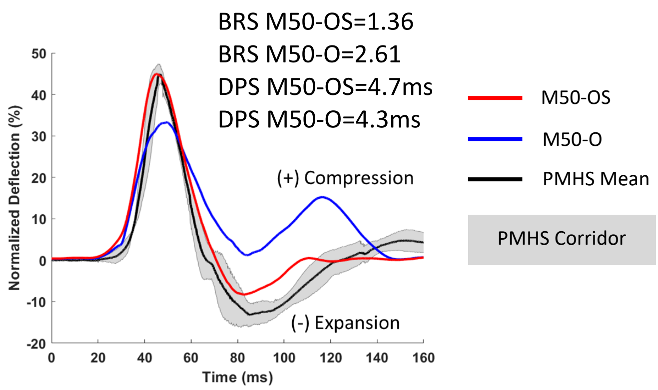

Fig. 1. Normalised chest deflection.

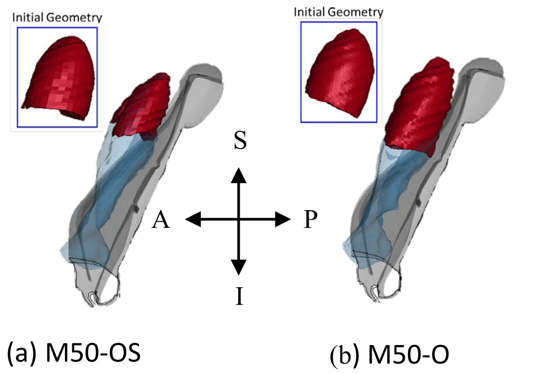

Fig. 3. Lung deformation at peak chest compression.

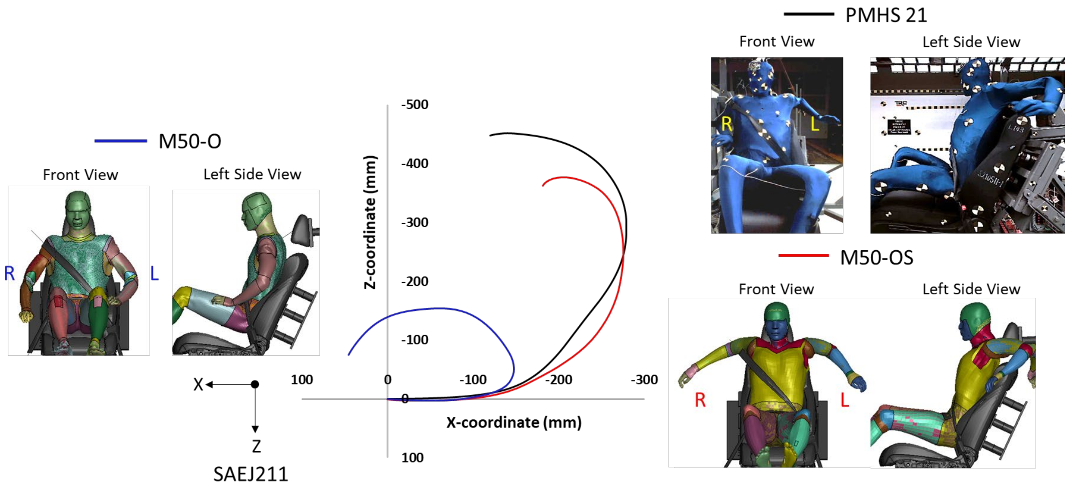

Fig. 2. XZ-trajectory of left mid-humerus. Occupant images correspond to t=118ms. L: Left and R: Right.

IV. DISCUSSION

The lungs in the M50-OS are modelled as low-density foam and underwent unopposed compression in both AP direction (from seatback interaction) and superior-inferior (SI) direction (from the mobility of abdominal organs) during the impact (Fig. 3(a)). The M50-O lungs underwent compression in the AP direction during the impact (Fig. 3(b)). However, the peak magnitude was lower than the M50-OS due to a stiff rubber material for the membrane tissue and the presence of fluid in the M50-O lungs. Expansion in the M50-O lungs in the SI direction was from fluid incompressibility and lack of abdominal organ mobility. PMHS lungs are expected to lack fluid pressure, and abdominal organs have been shown to move into the thoracic cavity during the impact [2-3]. Jansová *et al.* [7] have shown a human thorax FE model with M50-OS properties to correlate better with PMHS in frontal and oblique hub impacts, as compared to utilising the M50-O properties. AP chest expansion in the ramping phase in M50-OS (Fig. 1) was a consequence of upward deformation of the thorax in the ramping phase from abdominal organ mobility, similar to PMHS, as indicated by a relative Z-displacement of 28.3 mm between mid-sternum and T8. However, AP chest expansion was not seen in the M50-O due to a lack of abdominal organ mobility. As the seatbelt was in a fixed D-ring condition, it slipped off the occupant's shoulder from a rearward translation of the whole body during the impact. The shoulder-belt slip was enhanced in the PMHS and M50-OS from higher glenohumeral extension motion compared to the M50-O (Fig. 2). The lack of shoulder belt slip and lack of upward thorax deformation in the M50-O, unlike the PMHS and M50-OS, may have resulted in shoulder belt penetration into the chest in the rebound phase, generating a second AP chest compression peak at t=118ms (Fig.1). Scapula and clavicle fractures and sternoclavicular joint damage have been seen in the PMHS, likely due to excessive arm flailing motion [2-3]. Arm flailing in the M50-OS (Fig. 2) may be aberrant from simplifying the shoulder joint as a kinematic joint with associated joint stiffness, and lack of joint failure. The lack of arm flailing in the M50-O (Fig. 2) may be from stiffer shoulder joints compared to PMHS, and failure limits of the M50-O may have to be improved for high-severity rear impacts.

V. REFERENCES

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