Finite Element Analysis of Airbag Interaction using an Out-of-Position Seated Human Body Model

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I. INTRODUCTION

Airbags are a mature safety countermeasure ubiquitous in the modern automotive fleet [1]. However, in a vehicle crash, posture and position relative to the airbag strongly influence injury outcome for occupants. For the driver, ideal posture is at least 12 inches (304.8 mm) from the steering wheel, while faced directly forward, and the seatbelt placed over the ASIS and diagonally across the thoracoabdominal region, overlapping the mid-portion clavicle [2]. This restraint system, along with proper positioning within the seat, is crucial for the occupants, and deviations can be fatal. However, seating preference, or being out-of-position due to automated or driver-initiated pre-crash manoeuvring can result in positioning the occupant closer to the steering wheel. Anthropometric differences in humans also result in changes in kinematics, which can influence the body and airbag contact interface [3]. We analysed outcomes of a diffuse airbag-only loading case on two HBMs, the M50-O and the age targeted M50-O+aged, using an FE airbag representing the experimental tests performed by LeBarbé *et al.* 2005 [4] Fig. 1. Analysis of diffuse airbag loading cases may provide insight into how the airbag-human contact interface imparts force onto the chest during a symmetrical out-of-position impact, and how distribution of the airbag may change rib fracture outcomes. In addition, relatively few studies focused on airbag-HBM interactions have been published in the literature.

II. METHODS

The Global Human Body Model Consortium (GHBMC) 50th percentile male detailed occupant model (M50-O) and an age-adjusted version of the same (M50-O+70YO) were run in 12 Out of Position Seated (OOPS) simulations per experimental PMHS testing by [4]. Simulations were run using LS-DYNA version R10.2.0 at three distances of chest standoff: 13 mm, 78 mm, and 128 mm. Simulations were run with an industry-sourced FE airbag. To match the methods of the PMHS testing, the legs of the HBMs were removed at the proximal femur and the pelvis was added to a ballast mass via a freely rotating joint. Corridors were developed using the min-max boundaries of the force versus time PMHS data in the 13 mm and 78 mm tests (n=2 each); in the 128 mm configuration there was only one PMHS, therefore no corridors were developed. Airbag parameters were adjusted for pressure versus time in order to match the force versus time corridors from the 78 mm experimental testing to better assess rib fracture. In addition to the two baseline models, yield stress adjustments to the cortical bone of the ribs were implemented, based on data taken from [5], to assess the effect of bone quality differences between the data used to develop the rib models in the GHBMC M50-O and the bone quality of the PMHS, which could only be inferred [5]. Yield stress of the cortical rib was adjusted to 20% of its initial value (yield stress value: 0.088), as this is where the asymptote for yield stress was present in [5]. As a secondary aspect of these tests, we sought to elucidate differences in rib fractures observed in the PMHS testing versus the HBMs. Rib fracture analysis was performed using Metriks (Elemance, Winston-Salem, NC) using a 50% probability or greater as the threshold to count the rib as fractured. Probability of fracture was based on the risk curves of Larsson *et al.* [6]. All simulations were run on a Linux red-hat HPC cluster hosted at Wake Forest School of Medicine using 44 cores.

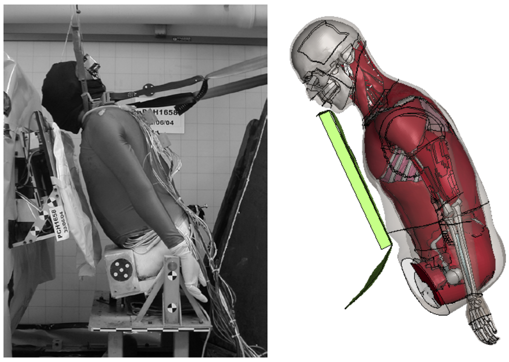

Fig. 1. A comparison of the PMHS in experimental testing from [4] to the M50-O+aged HBM from the simulations.

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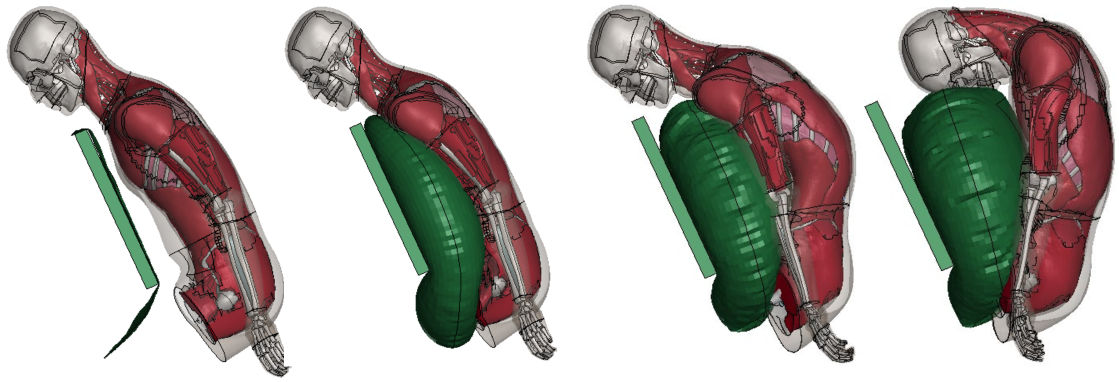

Fig. 2. Stepwise lateral images of the M50-O baseline model impact, in 25 ms increments starting at t = 0. The gross body mechanics had good agreement to the PMHS experimental videos.

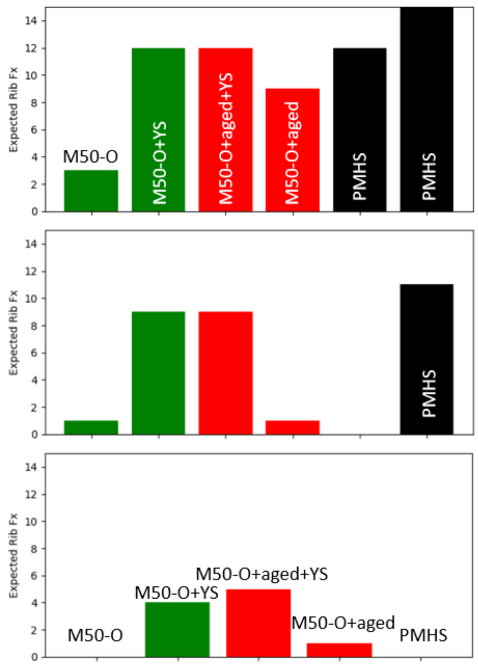

Fig. 3. Expected rib fracture and experimentally derived rib fracture for all three configurations. The order of models is repeated across each graph, except in the 128 mm (bottom), where there is only one *PMHS.*

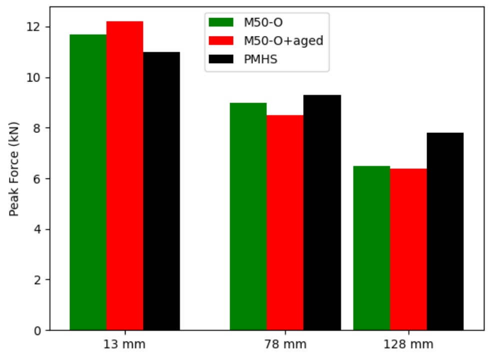

Fig. 4. Peak forces grouped by distance to airbag. The 78 mm cases were closest in peak force.

III. INITIAL FINDINGS

The gross kinematics of both the M50-O and M50-O+aged were similar to those of the PMHS in experimental testing by [4] (Fig. 2). The airbag peak force values are shown in Fig. 4. In the 13 mm and 128 mm cases, the peak force timing was similar between the HBMs and the PMHS (n=1 and n=2, respectively). The 13 mm and 128 mm simulations qualitatively matched the PMHS response, while the sample size was very limited. The baseline 50th percentile male model had less rib fracture compared to the aged model across almost all simulations, with both under-predicting fracture. The yield stress-adjusted models did not affect force response, but were closer to the average of the PMHS rib fracture in the 13 mm and 78 mm simulations, a cumulative average of all PMHS fractures across the three distances was 7.6, M50-O had an average of 1.3 fractures, the age-adjusted model had an average of 3.6 and the adjusted yield stress M50-O had an average of 8.3. Airbag proximity to the sternum was correlated to a greater number of rib fractures across all models.

IV. DISCUSSION

Diffuse loading cases are under-studied in HBMs and contemporary vehicle control systems may add to the prevalence of OOP occupants. The bone mineral density of the PMHS could not be directly related to mechanics from the data presented in [4], and this led to uncertainty in the bone quality of the ribs in the experimental testing. Rib fracture analysis of the GHBMC models is based on strain. Thus, as we decreased the yield stress of the ribs (which has been linked to bone quality), the loading reached the tangent modulus at lower stress, resulting in greater likelihood of failure [5]. The rib material properties for M50-O were taken from healthy rib which was experimentally characterised [7]. Therefore, to factor unknowns about bone quality in the PMHS, adjustment of the cortical bone yield stress was necessary. This simple adjustment to the yield stress of the cortical bone resulted in a marked change in the number of expected rib fractures across all three simulations in this condition (Fig. 3). To our knowledge, this is the first diffuse loading case reported on the GHBMC models and is indicative that they may perform well provided the airbag model is commensurate with the simulated test.

V. REFERENCES

[1] Segui-Gomez, M., *et al.*, *Am. J. Public Health*, 2000. [2] NHTSA, Safety 1N Numbers, 2015. [3] Hu, J., *et al.*, *Traffic Inj Prev*, 2019. [4] LeBarbé, M., *et al.*, *Stapp Car Crash J*, 2005. [5] Osterhoff, G., *et al.*, *Injury*, 2016. [6] Larsson, K., *et al.*, *Front Bioeng Biotechnol*, 2021. [7] Li, Z., *et al.*, *J Biomech*, 2010.