GHBMC-M50-specific Lumbar Spine Fracture Risk Prediction Considering Two Different Metrics

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I. INTRODUCTION

The use of human body models (HBMs) within biomechanics research for studying motor vehicle crash (MVC)-related kinematics and injuries has increased substantially over the years. Further, their inclusion in the vehicle assessment protocols for Euro NCAP [1] suggests that HBM simulation and virtual testing may play a larger role in the future. The lumbar spine does not yet have associated injury risk assessment values for any surrogate, including HBMs, in spite of the fact that injuries still occur [2-5], mostly in frontal MVCs [4], and may increase in frequency and severity in future vehicles with reclined occupants [6-8]. Ultimately, researchers need validated tools to interpret simulation results and predict injury risk with the latest HBMs, and there is a specific need for such tools for the lumbar spine. Additionally, due to the inherent differences between humans and HBMs, surrogate-specific injury risk prediction can be developed, which has been previously demonstrated by Forman *et al*. [9] for rib fractures. Thus, the goal of this study was to develop a series of HBM-specific injury risk functions (IRFs) and injury risk curves (IRCs) from simulations of the GHBMC-M50-O (v6.0.0).

II. METHODS

Three-vertebra segments (T12-L2 or L3-L5) of the GHBMC lumbar spine were simulated in injurious compression-flexion loading conditions with test-specific input pulses and axial compression levels to match the 40 post-mortem human subject (PMHS) experiments [10]. Each segment was first axially compressed, quasi-statically, to one of three force levels (2200 N, 3300 N, 4500 N), and then dynamically flexed at a rate of ~600 deg/s (Fig. 1). For the PMHS, an IRF was previously developed using a structural metric, L_{fx}, and survival analysis (Weibull distribution) with appropriate data censoring when middle vertebrae fractures did not occur or when exact timing could not be determined [11]. L_{fx} was a combination of "stress"-like terms that were calculated using the axial compression force, flexion moment, and the cross-sectional area of the middle vertebral body, and modified by an optimised tuning factor, α [11] (Fig. 2). For the GHBMC, structural and material response metrics were extracted from each simulation at the time when PMHS injury occurred in the corresponding experiment and was used to develop separate sets of IRFs via survival analysis (Weibull distribution). First, the structural metric was the same as in the PMHS IRF: L_{fx} calculated from GHBMC forces, moments and CSA. The optimal value of α was re-tuned to GHBMC data and selected as the value that best fit the data (lowest Akaike information criteria (AIC)). Then, five candidate material response metrics indicating local element stresses and strains in the middle vertebrae (e.g. 95th percentile: maximum and minimum principal strain in the shell and solid elements, and von Mises stress in the shell elements) were considered. The optimal of the five candidate metrics was determined by qualitative review of the IRCs as well quantitative review and comparison of the IRFs' Brier metric scores (BMS), as suggested by [12]. Age was included as a covariate for both IRFs, similar to the PMHS. Data censoring was consistent for PMHS and the corresponding GHBMC simulation (same injury/no injury designation for both).

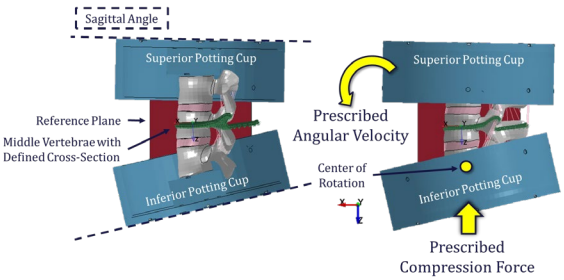

Fig. 1. Simulation setup that mimicked the PMHS experiments, with loads measured in the middle vertebrae cross-section.

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III. INITIAL FINDINGS

While the relative contribution of the force component vs. moment component (as indicated by the α value) in L_{fx} was optimised for the PMHS data (α = 0.11) [11], a different α value was optimal for the force component vs. moment component contributions when using the same optimisation methods for the GHBMC data (α = 0.40). These α values resulted in the best fit (by AIC) of each IRF to the respective injury data (PMHS and GHBMC). The optimised GHBMC IRC was positioned to the left relative to the PMHS risk curve with little overlap of confidence intervals (Fig. 2). When implementing material metrics extracted at the time of PMHS injury, von Mises stress in the shell elements representing cortical bone displayed the best qualitative illustration (i.e. visually appeared the most like a typical "S" IRF) and yielded the lowest BMS (Fig. 2).

$$L_{fx} = (1-\alpha)\frac{F}{CSA} + \alpha\frac{M}{CSA^{3/2}}$$

$$P(Injury|L_{fx}, Age) = 1 - e^{-\left[\frac{L_{fx}}{e^{1.41210-0.01088*Age}}\right]^{\frac{1}{0.209}}} \qquad P(Injury|VMstress, Age) = 1 - e^{-\left[\frac{VMstress}{e^{-0.72517-0.02448*Age}}\right]^{\frac{1}{0.493}}}$$

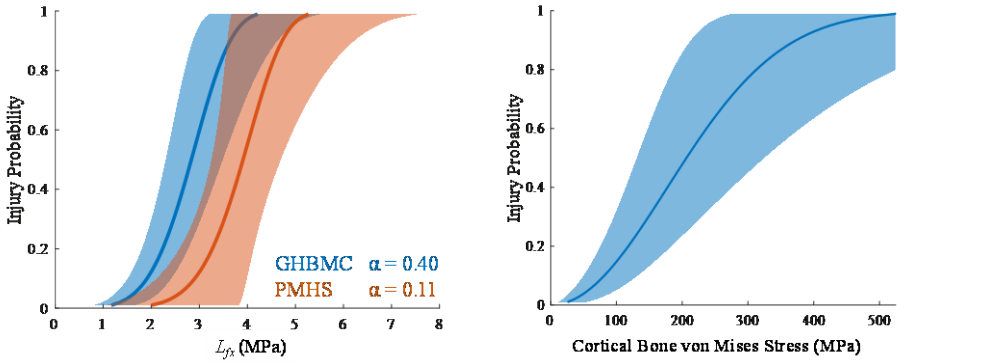

Fig. 2. IRFs and IRCs for the L_{fx} injury metric for the PMHS (orange) and GHBMC (blue), including the optimised alpha values (left). IRF for the PMHS not included (GHBMC only). IRF and IRC for the cortical bone von Mises stress metric for the GHBMC (right). Each plotted using the mean age (covariate). Mean risk curve with 95% confidence intervals.

IV. DISCUSSION

The exercise of comparing the GHBMC and PMHS demonstrated the value in creating HBM-specific IRFs and IRCs, similar to findings from Forman *et al*. [9]. In this study, and in [9], HBMs display elevated injury risk compared to PMHS for the same amount of mechanical input stimulus, therefore creating HBM-specific injury risk tools is necessary to avoid over-prediction of injury risk. If the original PMHS IRF was used to predict injury risk with the GHBMC, then GHBMC simulations would always under-predict the actual injury risk. In this study, the discrepancy between PMHS and GHBMC optimised α values and shifting of one IRC relative to another may be attributed to the GHBMC generally predicting lower moments than those measured in the PMHS in these loading conditions [13], and the GHBMC representing larger CSA than most of the PMHS CSA [13], although more analysis is needed. Further, the material-based IRFs and IRCs allowed for localised measurement of stress that is less likely to be tied to specific loading conditions used in the experiments/simulations than the structural metric (L_{fx}). Additionally, confidence intervals would likely tighten if either IRC included different levels of age (covariate). However, due to differences in biomechanical response and stiffness between PMHS and GHBMC [13], there may be cause to extract metrics from simulations at times different from PMHS failure time. Specifically, extracting peak metric values or matching an invariant metric (e.g. energy) between PMHS and GHBMC may be more suitable for injury risk prediction and will be explored in future work. Further, PMHS donor sex was not found to significantly affect injury risk [10], so all 40 PMHS (half male, half female) were utilized for the M50 simulations to harness statistical power of a larger sample size. Future efforts may include elucidating the potential need for sex-specific lumbar spine injury risk prediction and simulating female HBMs, and other male HBMs, in the same conditions to develop analogous surrogate-specific injury risk models.

V. REFERENCES

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