Effect of Postural Changes in Human Body Modeling of Ejection Seats.

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I. INTRODUCTION

Ejection seats are an important safety feature of military aircraft and understanding how natural variation in human posture and tissue properties influences ejection injury risk is essential. The I-PREDICT Human Body Model (HBM) was developed for military-relevant loading scenarios [1–2] within a probabilistic framework, enabling, population-level response predictions through propagation of uncertainty and variability. This study presents a probabilistic validation of the 50th Percentile Male (M50) I-PREDICT HBM's response to rapid deceleration from a vertical drop-tower, simulating aircraft ejection. The influence of uncertainties in initial spinal posture, seat belt characteristics, and HBM material properties on predicted kinematics and intervertebral disc (IVD) forces is quantified and compared against post-mortem human subject (PMHS) data [3–4].

II. METHODS

The M50 I-PREDICT HBM was positioned to replicate a vertical drop-tower test with a peak acceleration of 15 G, for which spinal kinematics and IVD forces were reported for three PMHS [3–4]. This loading condition was applied as a deceleration pulse on the seat-carriage in the upward direction with no initial velocity (Fig. 1).

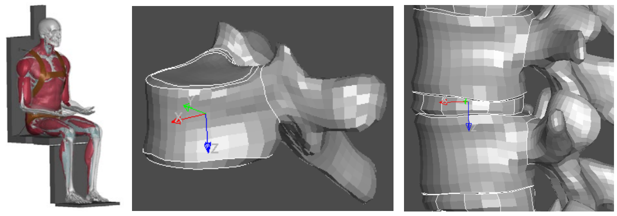

Fig. 1: I-PREDICT HBM model (left). Local coordinate systems in the vertebrae (center) and cross-section in the intervertebral disc (right) used to report spinal kinematics and cross-sectional forces.

Model kinematics were evaluated using linear accelerations and rotational velocities in local coordinate systems (LCS) defined at T1, T4, T12, L3 and S1 (Fig. 1). These outputs were compared directly with PMHS measurements. Similarly, IVD cross-sectional forces were extracted at lumbar IVD levels and compared to reported PMHS force data (Fig. 1).

To characterise system uncertainty, twelve material and structural properties were modeled as random variables using literature-derived distributions. Spinal cortical and trabecular bone were modeled using a piecewise linear plasticity material model with Young's modulus, yield stress, and failure strain treated as independent, normally distributed random variables [5–6]. Spinal ligaments are modeled as nonlinear elastic springs with two correlated stiffness scale factors (cervical and thoracolumbar, ρ = 0.8) [7-9]. Seat belt pretension force [3–4] and retractor pullout force were modeled as normally distributed variables. To allow for variability in the retractor response, a scale factor for the loading and unloading pullout curves was modeled as normally distributed an assumed standard deviation of 15% of the mean. Muscle tissue was modeled as a rate-dependent hyperelastic rubber with a normally distributed scaling parameter applied to its constitutive response [10–11]. Spinal curvature variability was incorporated using a uniform distribution of three initial configurations representing observed PMHS postures [3–4]. The first orientation (Fig. 2, Spine00) was the spinal position produced by gravity settling the HBM (Fig. 1 and 2). The two less lordotic orientations (Fig. 2, Spine01 and 02) were produced by applying a displacement to T9-L5 to the settled model.

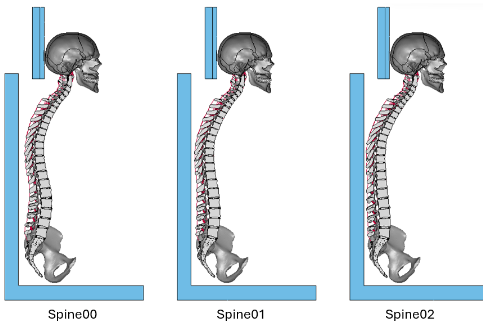

Fig 2: Model cross sections showing starting spinal orientations

A total of 120 simulations were performed in LSDYNA with properties sampled from the defined distributions using Latin Hypercube Sampling; 106 simulations completed successfully. Outputs were sampled at 10 kHz and filtered using a fourth-order Butterworth filter (300 Hz cutoff).

Model performance was quantified using root-mean-square-error (RMSE) between simulated mean responses and experimental mean responses generated via ARCGEN [12],

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with signals interpolated to 1000 points over 0 -200 ms. Global sensitivity was assessed using partial rank Spearman correlation coefficients (PRCC) between input variables and key response metrics: peak Z-direction IVD force (F_{zp}), peak resultant acceleration (A_{rp}), and peak Y-direction rotational velocity (V_{yp}) [13-14]. Significant PRCCs ($p<0.001$) are reported.

III. INITIAL FINDINGS

Good agreement between simulated and experimental kinematics was observed. RMSE values were below 4.7 G for LCS X- and Z-axis linear acceleration and below 377 rad/s for LCS Y-axis rotational velocity across instrumented vertebral levels (Fig. 3).

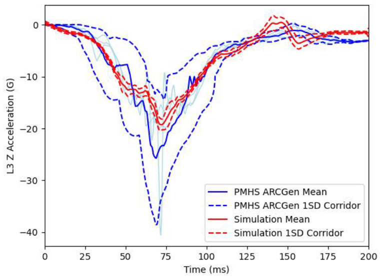

Fig. 3: L3 Linear Acceleration in Z; DeWitt PMHS data (light blue) from [3], ARCGen experimental mean response and corridor (dark blue) are shown along with the simulation mean and corridor (red).

Spinal posture was identified as a dominant driver of kinematic variability. Moderate PRCCs were observed between peak resultant acceleration (Arp) and spinal orientation for T1 (PRCC = 0.64), T4 (PRCC = 0.84), and T12 (PRCC = -0.46). These observations indicate that less lordotic lumbar initial postures (e.g., Spine01 and 02) had a higher T1 and T4 A_{rp} and lower T12 A_{rp}, when controlling for other model inputs. Seat belt pretension exhibited a weak negative correlation with Arp at T12 level (PRCC = -0.43).

Similarly, peak (most positive) Y-direction rotational velocity (Vyp) was affected by initial spinal orientation; with positive PRCC at the upper thoracic levels (T1 PRCC = 0.77; T4 PRCC = 0.76; and T12 PRCC = 0.91) and negative PRCC at L3 (PRCC = -0.93) and S1 (PRCC= -0.64). These results indicate less lordotic lumbar initial postures tended to have higher T1, T4, and T12 V_{yp} and lower L3 and S1 V_{yp}, when controlling for the effects of other model inputs. In addition, PRCCs were observed between V_{yp} and seat belt pretension, muscle stiffness parameters, ligament stiffness parameters, and spinal bone properties though these were generally weaker and depended on spinal level.

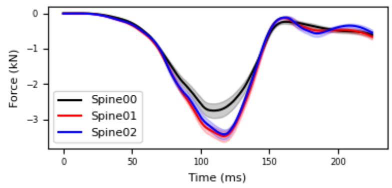

Fig. 4: HBM L1-L2 IVD Z-force; average (solid lines) and +1 standard deviation (shaded regions) are color coded by spinal levels.

The highest magnitude lumbar IVD force in the Z-direction (Fzp) was underpredicted (HBM: ~4-5% body weight; PMHS: ~7-12% body weight) [3-4]. PRCCs were observed between Fzp and initial spinal postures across lumbar and lumbar/sacral IVD levels ($0.75 \leq |PRCC| \leq 0.93$), with the trend direction varying by region. For the L1-L2, L2-L3, and L3-L4 IVDs, less lordotic lumbar initial postures were associated with higher magnitude F_{zp} (Fig. 4). For the L4-L5 and L5-S1 IVDs, more curved starting orientations were associated with lower $F_{zp.}$ Negative correlations between F_{zp} and trabecular bone properties (Young's modulus and yield stress) were also identified.

IV. DISCUSSION

The I-PREDICT HBM demonstrated good agreement with experimental kinematics under ejection-relevant loading [3–4], supporting its use in probabilistic injury assessment. While IVD forces were underpredicted, experimental measures relied on single pressure sensors and may not fully represent true cross-sectional loading [3]. Further, HBM IVD material properties were not varied in this study and may contribute to the observed discrepancies. The PRCCs indicate that initial spinal posture exerts a greater influence on response variability than HBM material property uncertainty for the metrics considered. This finding suggests that occupant posture should be considered in seat design and injury mitigation strategies. For alternative injury metrics, such as spinal strain-based criteria, material properties may play a more significant role.

V. REFERENCES

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