Subject-specific Lumbar Spine Finite Element Model Creation and Validation using Dynamic Compression

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I. INTRODUCTION

The advent of more advanced autonomous driving systems enables new seating postures by removing some of the constraints required to traditionally operate a vehicle [1]. Human Body Models (HBMs) are increasingly being used to assess the safety of these new systems and of alternative seating positions, due to their omnidirectionality and lack of a targeted hardware dummy [2-3]. Studies have shown higher lumbar spine compressive loading in reclined positions compared to traditional upright seating positions [4] and lumbar spine compressive fractures not only in reclined positions but also in upright seating positions [5]. The THUMS v5 HBM has recently been validated against one test of the non-injurious dynamic compression data from Stemper *et al.* [6-7]. Even though the use of individualized HBMs has been shown to be beneficial [8], until now few papers have been published investigating subject-specific models in dynamic compression of the lumbar spine at levels seen in automotive safety applications. By creating subject-specific lumbar spine models as documented in Stemper *et al.* [6], an initial impression of the importance of differences in anthropometry on kinematics and injury outcomes upon this data are investigated.

II. METHODS

In the experiments by Stemper *et al.* [6], T12-L5 lumbar spine specimens were mounted in a drop tower rig and accelerated onto a pulse shape foam, creating dynamic compression at both non-injurious and injurious levels. The simulation setup models one of the injurious tests with absent bony injury by applying the measured accelerations at the caudal ends of the whole lumbar spine specimens as measured in the experiments as model inputs. For the reconstruction of subject-specific Finite Element (FE) models of the full lumbar spine units, vertebral body geometry was extracted from pre-test CT scans. Discs and ligamentous structures as well as material properties were initially used from the THUMS v4.1 HBM. For measuring spinal posture, vertebral coordinate systems as proposed in [7] were implemented. A visualization of the setup can be seen in Fig. 1 – the subject-specific model with a natural lordotic curve and the THUMS v4.1 in the more kyphotic standard position. Both models were globally translated (sagittal axis) and rotated (sagittal and transversal axis), as described in [6], to perform a sensitivity analysis.

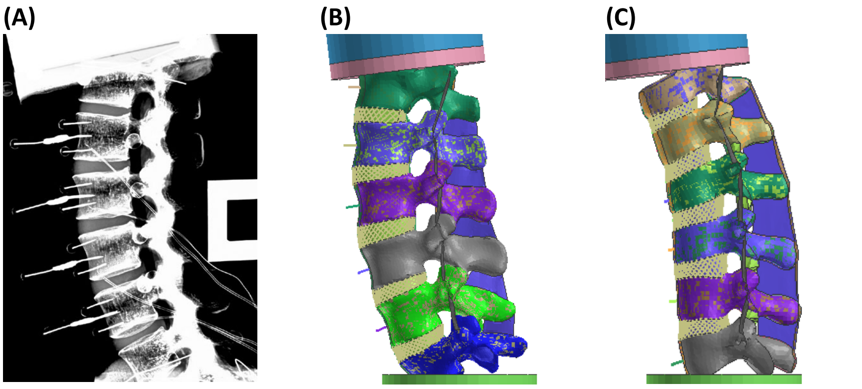

Fig. 1. Comparison of the experimental setup **(A)** to the simulation setups: subject-specific setup **(B)** and THUMS v4.1 setup **(C)**.

III. INITIAL FINDINGS

Fig. 1 **(B, C)** shows the best-matching global model orientation found via the sensitivity analysis. Angular displacements and acceleration curves for the corresponding simulation and experiment until time of peak load are presented in Fig. 2. Experimental results are represented by dashed lines, while solid lines represent simulation results with THUMS v4.1 and the subject-specific model, respectively. A similarity in the subject-specific kinematics to the experimental values was found: the simulation and experimental results for the relative angles of T12-L1 to L3-L4 all experienced flexion, whereas the L4-L5 response for both simulation and experiment showed extension. The simulation with the non-subject-specific THUMS v4.1 deviated from the experiment: the angular displacement of L2-L3 experienced flexion and L4-L5 experienced

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extension. Both accelerations at the cranial potting match with the accelerations measured in the experiment.

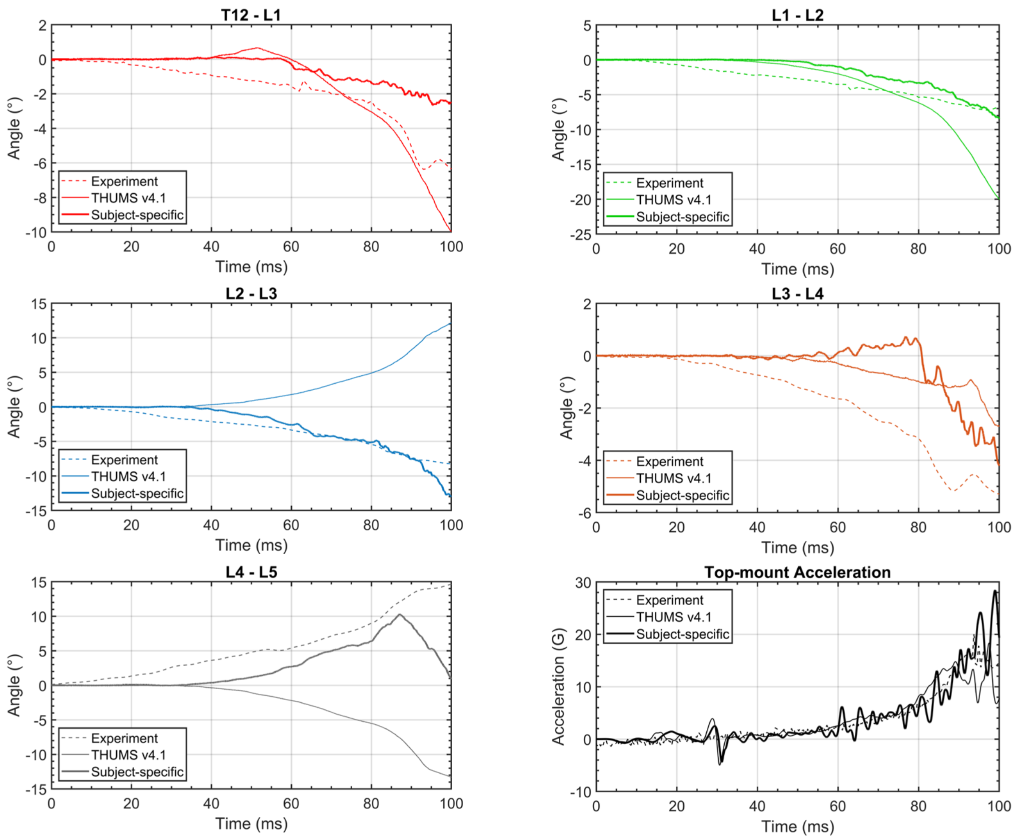

Fig. 2. Comparison of the angular displacements between experimental (dashed line), simulation with THUMS v4.1 (solid line) and with subject-specific models (thick solid line). The angular displacement for both models were zeroed by subtracting the respective initial angles.

IV. DISCUSSION

In this unique study, injurious dynamic datasets from whole lumbar spines were used for subject-specific FE model creation and the responses of the subject-specific lumbar spine and the THUMS v4.1 lumbar spine were compared against the dataset [6]. Whereas the subject-specific model responded consistently with the experiment, the THUMS v4.1 lumbar spine deviated from the experiment. These differences may be attributable to initial sagittal orientation of the spine. The initial sagittal intersegmental angles of the subject-specific models match the angles of the experimental specimen. However, only the L4-L5 angle of THUMS v4.1 matches well; all other angles are between 4° and 19° more positive than their experimental counterparts. These offsets have been removed in the comparison but might possibly explain this discrepancy.

The slope of the subject-specific curves is delayed and flatter than in the experiment. This might be due to subject-specific material properties that have not been adapted in the simulation model, but would influence the angular displacements. Another possible reason for the different slope might be preflexion in the experimental setup. As the focus of this investigation relied on initial positions, material property variation was not considered here. Another reason for deviations of the subject-specific curves from the experiment might be failure: particularly in the caudal region, high angular displacements were observed, which could provide an indication of failure. Integration of a failure criteria into the subject-specific models might overcome this issue. Furthermore, there are still uncertainties in the numerical setup configuration that require further investigation and clarification.

An investigation of sensitivity of fracture predictors due to differences in anthropometry can help to further understand the effects on dynamic spinal response. In future work, influential anthropometric parameters can then be taken into account before assessing global model performance using experimental corridors.

V. REFERENCES

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