Lower Cervical Spine Hard Tissue Injury Prediction in Axial Compression

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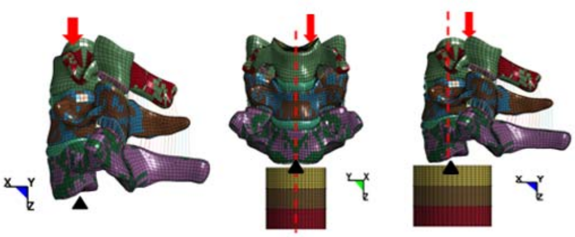
I. INTRODUCTION

Motor vehicle accidents (MVAs) are the leading cause of traumatic spinal cord injuries (SCI) [1]. Although rollovers account for 3% of MVAs [2], a high proportion of vehicle-related SCI cases are associated with rollovers [3] and they have the highest incidence rate of AIS 2+ cervical spine injuries [4]. Spine fractures are present in 64% of patients with SCI and burst fractures are reportedly the most common fracture type (48%) [5]. In a rollover crash scenario, the neck may be subjected to axial compression loading, which may lead to a fracture of the cervical spine [6]. Detailed human body models can help advance our understanding of the mechanics of these injuries and they can provide data that is not possible to collect experimentally. Critical requirements for these models are accurate material properties and tissue level failure criteria, and most often trabecular and cortical bones in computational models are assigned linear isotropic material properties [7-9]. The objective of this study was to investigate injuries of the lower cervical vertebrae under axial compression impact conditions using a detailed human male 50th percentile neck model (Global Human Body Models Consortium (GHBMC) M50-O v4.3) with updated hard tissue constitutive models, and to validate this model against available experimental data.

II. METHODS

A literature survey was performed to identify sets of material properties for trabecular and cortical bones in both young and aged donor populations [10-14]. For cortical bone, an asymmetric constitutive model was used, while a crushable foam constitutive model was used for trabecular bone [15]. In order to model fracture initiation and propagation, an element deletion approach based on failure strain determined from the literature was utilized [12]. Single element simulations in tension and compression were performed to verify the constitutive models and material properties. Subsequently, simulations replicating pure axial [16], posterior eccentricity [16] and low lateral eccentricity [17] compression experiments were performed on the C5-C6-C7 (C57) segment of the 50th percentile neck model (Figure 1). The kinematic and kinetic responses and the ultimate loads were compared with experimental results. For a suitable comparison of the laterally eccentric experiments with the simulation, the fracture initiation values, rather than the ultimate loads, were compared to those when fracture was predicted in the computational model by failure or erosion of an element [17].



Young Samples (40-49 years old)			Aged Samples (70-79 years old)		
Cortical Bone [10,18]	Tension	Compression	**Cortical Bone [10,18],**	Tension	Compression
Ultimate Stress (GPa)	0.141	0.175	Ultimate Stress (GPa)	0.132	0.178
Ultimate Strain	0.0196	0.0435	Ultimate Strain	0.0167	0.037
Trabecular Bone [12]			**Trabecular Bone [12,14]**		
Ultimate Stress (GPa)	0.007	-	Ultimate Stress (GPa)	0.002	-
Ultimate Strain before densification	-	0.457	Ultimate Strain before densification	-	0.6

Figure 1: (from left to right) Pure axial compression simulation [16], Low lateral eccentricity simulation [17], Posterior eccentricity simulation [16]; Summary table of mechanical properties of hard tissues in young samples [10,12,14,18]

III. INITIAL FINDINGS

Results with constitutive models from the young and aged populations in the C5-C6-C7 segment in pure axial compression compared relatively well (8% lower for young model; 2% higher in failure load and 28% in failure displacement for aged model) with experimental data (Figure 2). For the eccentricity cases, only the aged constitutive model was used to be consistent with the donor ages of the specimens used in the experiments. The failure loads were similar (within 5%) to the average experimental values for the lateral eccentricity case (Figure 3) but they were lower (63%) for the posterior eccentricity case (Figure 4). On the other hand, the

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simulated moments were similar (10%) to the average experimental values for the lateral eccentricity case (Figure 3) and were 48% lower for the posterior eccentricity case (Figure 4). The fracture locations in pure axial compression and lateral eccentricity agreed with experimental findings (vertebral body, articular process and lamina in pure axial compression; vertebral body and lamina in low lateral eccentricity) [16, 17] (Figures 2, 3). Fracture in the articular process was predicted for posterior eccentricity loading, and not in the posterior elements as reported in the experiments [16] (Figure 4).

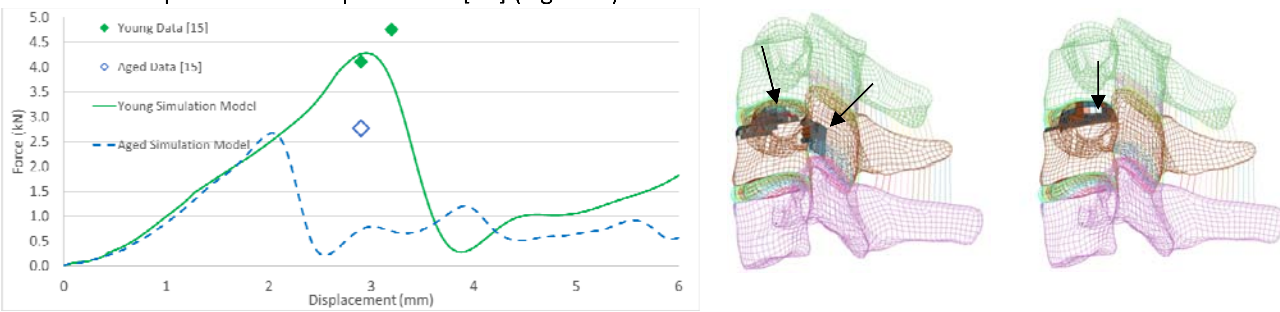

Figure 2: (from left to right) Force-displacement response of young and aged models; Fracture locations (in arrows) in axial compression in the young model; Fracture locations (in arrows) in axial compression in the aged model

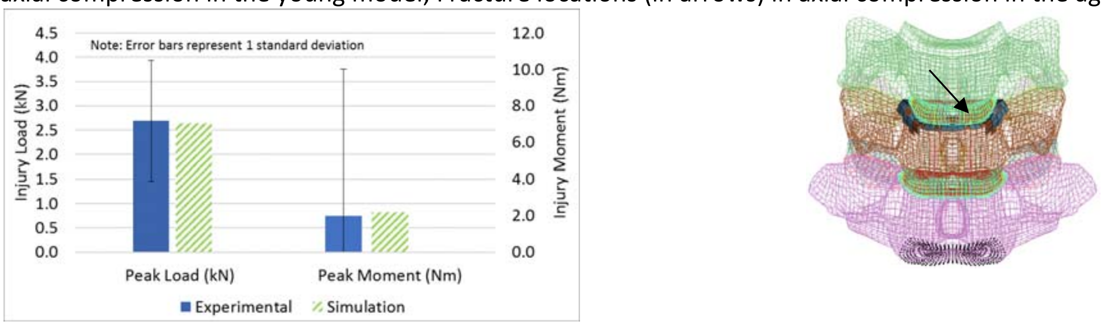

Figure 3: (from left to right) Injury load and moment results for the low lateral eccentricity case (experimental n=6); Fracture location (in arrow) in the aged model in low lateral eccentricity case

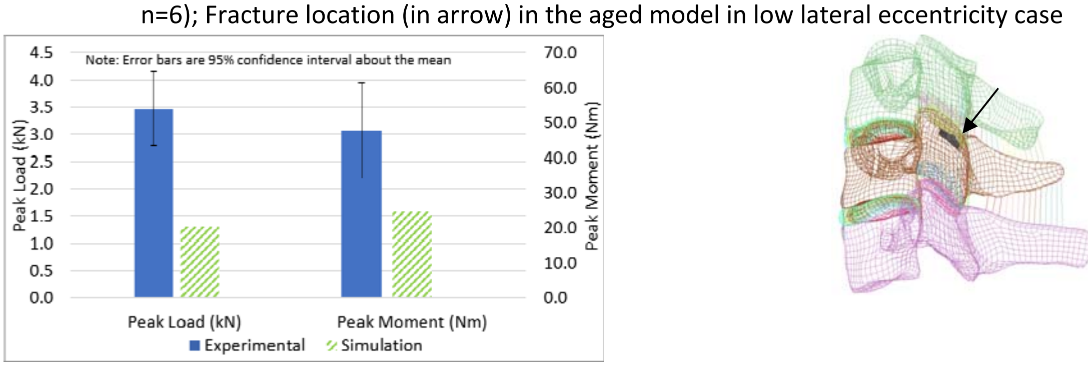

Figure 4: (from left to right) Peak load and moment results for the posterior eccentricity case (experimental n=8); Fracture location (in arrow) in the aged model in compression-extension loading case.

IV. DISCUSSION

The disparity in values and fracture locations in the posterior eccentricity case may be due to differences in the test set up and therefore, further investigation into modeling the experimental boundary conditions is underway. The computational model is also sensitive to the orientation of the loading vector in the spine and lack of this information for the experimental tests may affect the results. Furthermore, the variation in bone mineral density (BMD) between individuals may affect the results as the young and aged constitutive models were based on a collection of properties from various studies [10,12,14,18] that are different from the donors of the specimens used the compression experiments. In conclusion, this study provides an initial investigation into the constitutive models for both cortical and trabecular bone in a detailed neck model under axial compression loading with eccentricity, where future work will apply these findings to full neck impact studies.

V. REFERENCES

[1] O'Connor et al., Acc Annal and Prev, 2002b
[2] Foster, UVA catalog, 2013
[3] O'Connor et al., Acc Annal and Prev, 2002a
[4] Yoganandan et al., SAE, 1989
[5] Pickett et al., Spine, 2006
[6] Argenson et al., Eur J of Orthop Surg & Trauma, 1997
[7] Kumaresan et al., MedEng&Phys, 1999
[8] Yoganandan et al., MedEng&Phys, 1996
[9] Teo et al., MedEng&Phys, 2001
[10] Hansen et al., J of Biomech Eng, 2008
[11] Reilly DT et al, J of Biomech, 1975
[12] Liu et al,Mech of Bio Systems and Mtls,2012
[13]Mosekilde et al., Bone, 1987
[14] Hayes et al., J Biomed Mat Res Sym, 1976
[15] Khor et al., IBRC conf proc, OSU, 2016
[16] Carter et al., Stapp Car Crash Journal, 2002
[17] Van Toen et al., JBiomechanics, 2014
[18] Burstein et al., JBone&Joint Surg, 1976