Lumbar Spinal Column Injuries and Loads from Inferior-to-Superior Impact: Applications to Military and Automotive Environments

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I. INTRODUCTION

Impact loading to the lumbar spinal column of a seated occupant via the seat structure and the pelvis, along the inferior-to-superior direction, is a mechanism of injury in military and civilian automobile and fall environments. From an automotive perspective, these injuries have been reported in European and US studies for restrained occupants in frontal impacts [1-4]. From a military perspective, they have been associated with underbody blast loadings, such as those from improvised explosive devices (IEDs) [5]. Injury levels have spanned across the entire lumbar column. While some studies have replicated these injuries, previous experimental models have not accurately mimicked the in vivo situations [6]. The objectives of the present study are to characterise the injuries to the human lumbar spinal column using a realistic experimental model and to apply the inferior-to-superior loading to the distal end of the preparation.

II. METHODS

Post Mortem Human Subjects (PMHS) were procured, x-rays obtained and bone mineral densities (BMD) of lumbar vertebrae computed from CT images. The thoracolumbar spinal columns were then isolated [7]. The specimens were fixed at the distal and proximal ends such that L5-S1 and T12-L1 intervertebral joints were unconstrained. Six-axis load cells were attached to the distal and proximal ends. The specimens were positioned in the neutral posture, i.e. T12 vertebra aligned at approximately 5° from the horizontal plane, while the sagittal angulation of the spine was approximately 12°. The prepared and aligned columns were positioned on a custom vertical accelerator (VertAc) to apply inferior-to-superior loading [8]. The rates of loading ranged from 9 to 31 km/h. The added weight of approximately 110 N, attached to the upper fixture, simulated the effective mass of the torso. Vertebrae superior to approximately the mid body level for T12 was in the PMMA fixation and hence, any motions above this level, encountered in vivo, was not included in the experimental model. From this point of view, this model is limited to the lumbar column without the full influence of the thoracic spine.

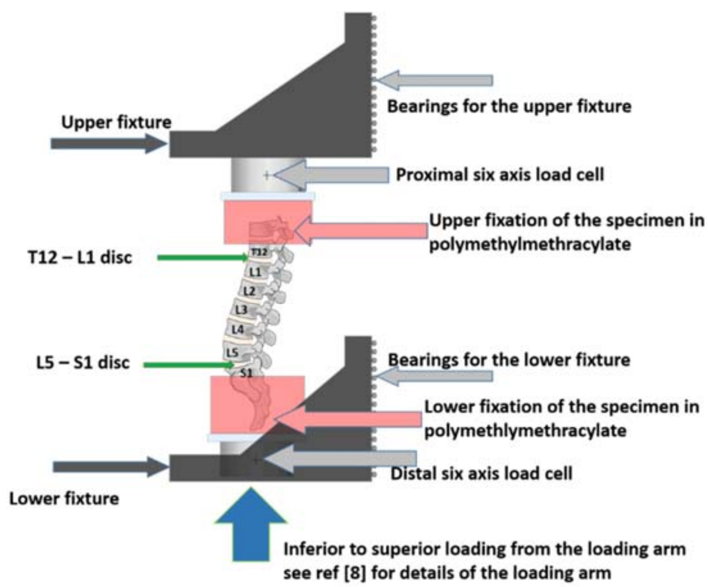

Fig. 1. Schematic of the lumbar spinal column in the VertAc.

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Pre-test x-rays were obtained while the specimen was positioned on the VertAc. Post-test x-rays were obtained while the spine was still on the VertAc; CT scans were taken and detailed dissections carried out. Posttest CT scans included the PMMA material. Injuries were identified using these modalities. Data were processed such that the loads from the two load cells were mass compensated and transformed to the proximal- and distal-most disc joint centres (T12-L1 and L5-S1). The peak resultant forces in the loading phase are reported at these two intervertebral disc levels.

III. INITIAL FINDINGS

The mean age, stature, total body mass, were 63 years, 1.8 m, and 85 kg, respectively. The mean proximal (T12-L1 disc level) and distal (L5-S1 disc level) forces were 6.5 kN and 8.5 kN, and the coefficient of variations were 0.30 and 0.39, respectively. All specimens sustained injuries at isolated vertebrae or spanning multi-vertebral levels: L1 only (n=6); L3 only (n=1); L1 and L2 (n=2); L1-L2 and L4-L5 (n=1); T12 and S1 (n=1); L1, L3 and L4 (n=1); L1-L4, S1 and sacrum (n=1); and L1 and L3 (n=1). Disc injuries also occurred in some cases. Fractures identified from the CT images were confirmed during dissection. All injuries were attributed to the compression-related mechanisms.

IV. DISCUSSION

The objective of the study was to replicate injuries observed in military field environments using a realistic model of the PMHS lumbar spinal column. Consequently, it was important to design an experimental setup that mimicked the real-world situation, i.e. compression-related injuries at different levels of the spine [5]. The posture of the occupant, realistic boundary conditions at the ends, preload on the lumbar spinal column, and load application from the inferior-to-superior direction are among the important factors that affect the output of any experiment [9-10]. The load transfer within the spinal column, i.e. the lower absorbed force at the T12-L1 disc level compared to the greater applied force at the L5-S1 disc level, depends on many factors. As the human spine is axially loaded, due to the body's biped nature, and carries the mass of the upper torso mass in vivo, it is necessary to simulate this feature. This was achieved by adding a weight to the proximal end. Likewise, the load vector in the real world is from seat to the pelvis to the vertebral column. This inferior end condition was simulated by using the VertAc, which accommodated the positioning of the spine to accept the cranially directed impact loading, therefore any motion of the spine was from the inferior-to-superior direction after impact [8]. Thus both the cranial and caudal end conditions were met. The loading is dynamic in the real world, thus rendering previous slow rate PMHS experimental models less realistic. The posture of the occupant was also based on field conditions applicable to this environment. The fact that the present setup was based on these considerations means that it is can effectively replicate injuries seen in clinical and field environments [5]. These are among the unique and novel features of the experimental design and outcomes, which can be matched with field outcomes in terms of injuries and biomechanical outcomes in terms of proximal and distal forces.

V. ACKNOWLEDGMENTS

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