SPINAL CORD DEFORMATION DURING INJURY OF THE CERVICAL SPINE IN HEAD-FIRST IMPACT

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ABSTRACT

There is a need to better understand the biomechanics of spinal cord injuries especially at the level of the spinal cord itself. We report results from a series of head-first impacts using human cadaveric cervical spines (N = 6) and a custom drop tower. To improve the biofidelity of the column response during impact muscle forces were simulated using a constant follower load of 150N. A radio-opaque, biofidelic surrogate cord was imaged using cineradiography at 1000fps to quantify the transverse spinal cord deformation during injury. A variety of injury modes commonly associated with axial head first impact were induced. The average maximum cord compression was 49%, occurring at 12.3ms after impact. The follower load appeared to eliminate "snap-through" buckling in these tests.
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HEAD FIRST IMPACT can occur during home and occupational falls, automotive rollovers, and during various sports and other activities. These incidents are often associated with both spinal column and spinal cord injuries. To better understand these injuries and to develop devices to prevent them, many investigators have studied the response of the cervical spinal column under axial impact. In general these studies have not related the spinal column injuries sustained to spinal cord injury.

A limited number of groups have studied spinal injuries at the level of the spinal cord. These groups have applied axial impacts and studied spinal canal occlusion at discrete points (Carter, 2000), or pressure distribution at discrete points using a surrogate spinal cord (Pintar 1995). Spinal cord deformation has also been studied directly under flexion-extension trauma (Bilston 1997). There is limited data available on spinal cord deformation during spinal cord injury and none of these groups have directly reported cervical spinal cord deformation during axial impact to the head.

Spinal cord injury and treatment has been extensively studied using animal models (Kwon, 2002). An understanding of the conditions of human spinal cord injury in terms of strain rates and deformations would provide more realistic parameters for simulating injury in the animal models.

A dynamic head to ground impact model that allows for the visualization of the cord, within the bony spinal canal, during injury would allow for the characterization of cord deformation resulting from bony injury. In considering experimental creation of these injuries in a human cadaveric model, there is some concern as to the biofidelity of the anatomic specimen. Nightingale (1996) reported a higher order serpentine buckling pattern of the human cadaveric cervical spine during impact and complex contiguous injury patterns. It has been shown that the ligamentous cervical spine buckles under 10N of load (Panjabi, 1998) and this may be why the higher order buckling modes were exhibited. The use of a follower load allows the cervical spine to support physiologic loads without buckling (Patwardhan, 2000). Follower preloads have been used in many quasi-static biomechanical tests and in some horizontal impact tests (Ivancic, 2005) but never during head first impact. We hypothesize that follower preload will prevent higher order buckling behaviour in head-first impact experiments.

The objectives of our work are therefore to quantify cord deformation during injury sustained during head to ground impact and to measure the kinematic response of the cervical spine in the presence of a follower preload.

METHODS

Six human cadaveric cervical spines (occiput to T2) were prepared for impact testing. Each specimen was dissected free of soft tissues while preserving the osteo-ligamentous structures. The

transverse processes of C3-C7 were removed to allow anterior access to the lateral masses for preload guides. Care was taken to ensure that the guide screws did not disrupt the facet joints. Four photoreflective marker pins were inserted into the vertebral bodies and lateral masses of at each level between C2-C7 for kinematic analysis. The T1 and T2 vertebrae were mounted in dental stone such that the C4-C5 disc was horizontal while the neutral posture lordosis of the spine was maintained. A surrogate head with inertia, mass and head shape properties that match those of the human head was mounted to the occiput. Use of cadaveric specimens and our procedures were reviewed and approved by our institutional Clinical Research Ethics Board.

Axial impact was applied using a four shaft, self supporting drop tower with a light-weight carriage. The carriage was mounted on linear bearings. The carriage weight approximated the upper torso weight of a cadaver specimen (approx 15kg). The cervical spine specimens were attached to the carriage in an inverted orientation such that when the carriage was released head-first impact occurred against an impact platen. The experiment thus mimics head first impact that could occur when diving into a shallow pool head first. Prior to head impact, the specimens had a neutral posture with physiologic lordosis due to the application of a follower load compression to the spine that was applied to simulate pre-impact muscle forces. A 6-axis load cell (MC3A, AMTI) was mounted to the underside of the carriage at the caudal end of the cervical spine specimen to measure the reaction forces and moments at the neck during impact. A single axis load cell (LC402-5K, Omega Engineering) was mounted under the impact platform to measure the impact forces at the head. Two high speed digital cameras (Phantom v9.0, Vision Research Inc., Wayne, NJ, USA) recorded the impact at 1000 fps for kinematic analysis purposes. A drop height of 60cm was used resulting in an impact speed of 3m/s. The axial compression of the spine specimen was constrained to 2cm by physical stops.

The follower load system used in this experiment was a modified version of that developed by Miura (2002). The follower load was applied laterally by cables passing through guides mounted to the anterior aspect of the lateral masses. This constrained the preload to pass through the approximate flexion/extension centre of rotation of each vertebra (Dvorak, 1991). It was anchored at the caudal end of the specimen by eyebolts embedded in the dental stone. A 75N load was applied to each side (for 150N total preload) using compression springs at the head. This level of preload represents the cervical spine compression with some level of neck muscle activation (Hattori, 1981; White, 1990; and Moroney, 1988) as may occur in people during head first motion prior to head impact.

A biofidelic surrogate spinal cord (Reed, 2005) was placed in the spinal canal after removal of the natural cord. The surrogate cord was made from an elastomeric gel (QM Skin 30, Quantum Silicones). This material has similar material properties to the *in vivo* canine spinal cord in tension. The cord was oval shaped with diameters of 12mm transversely and 6mm sagittally (Kameyama, 1996). High speed cineradiography was used to observe the surrogate cord during injury. This setup was composed of a high powered industrial x-ray source (Philips MG-160 generator with Comet MXR-160 tube) and image intensifier (PT93XP43, Precise Optics) with an internally mounted high speed digital camera (Kodak Ektapro). Barium sulphate was added to the QM skin to increase the radiodensity of the surrogate cord. The deformation of the radioopaque spinal cord was measured at intervals of approximately 1.5mm along the length of the cord by digitizing each of the high-speed cineradiograph images with a precision digitizing tablet and image analysis software (ImageJ).

RESULTS

The injury mode of each specimen is reported in Table 1. The qualitative analysis of the high speed digital footage during impact indicates that no serpentine higher-order buckling pattern was seen in any of the six specimens. Each specimen had a similar response to axial impact. Upon impact each spine underwent traumatic extension as the carriage continued to load the spine for approximately12ms after the head was arrested. The carriage rebounded at this point due to contact with the physical stops.

Results of the analysis of the cineradiographic footage for all six specimens is summerized in Table 1. These results include the percent maximum compression, time of maximum compression, and the total time while the cord was compressed at least 10% and 75% of its maximum. In all cases there was no residual compression of the cord as the specimen rebounded.

Table 1: Spinal Cord Deformation - Maximum Transverse Compression Conditions

Specimen	Percent Compression	Time to Max.	Time spent at >10% of Max	Time spent at >75% of Max	Associated Injury Mode
H1091	50%	11ms	23ms	5ms	Hyperextension about C4/5
H1096	53%	15ms	14ms	6ms	Atlantoaxial dislocation
H1116	58%	15ms	31ms	9ms	Fracture disloaction at C3/4
H1177	36%	8ms	23ms	1ms	Hyperextension about C5/6
H1183	78%	14ms	29ms	21ms	Atlantoaxial dislocation
H1184	19%	11ms	21ms	2ms	Dislocation at C4/5
Average (STD)	**49% (18.3%)**	**12.3ms (2.8ms)**	**23.5ms (6.1ms)**	**7.3ms (7.3ms)**	

The cord compression profile for specimen H1116 illustrates a typical profile plot (Figure 1a). Due to the variation between specimens an "average" profile was not possible. The three dimensional surface graph shows the percent compression along the length of the cord labelled in Figure 1b with respect to time. The "negative" compression values arise from the cord diameter increasing due to longitudinal compression of the cord in the canal as the spinal column was compressed.

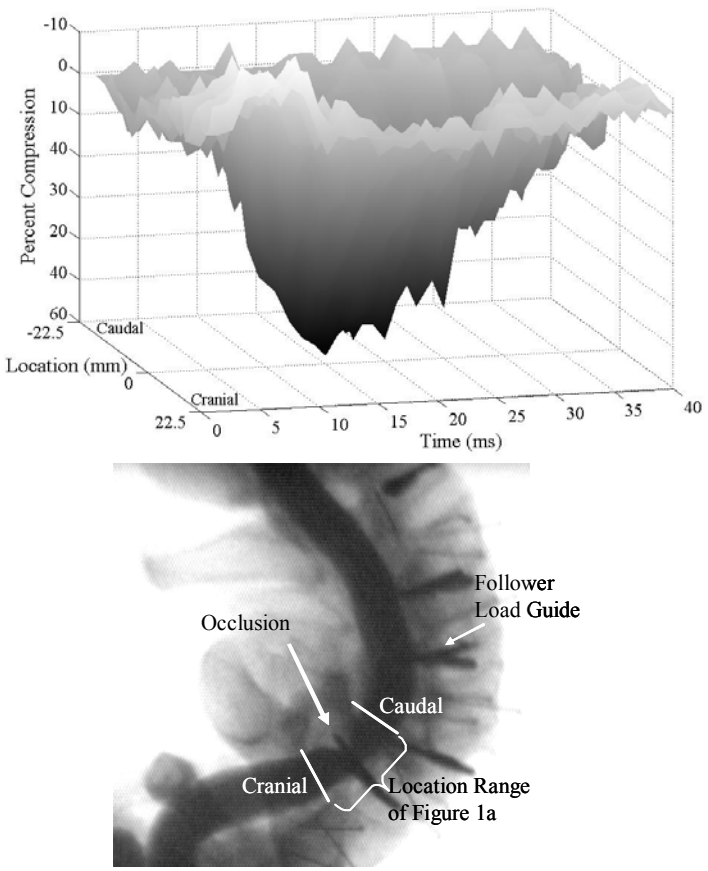

Figure 1: a) Cord Compression Profile (H1116) b) Radiograph Image at Maximum Compression

DISCUSSION

We measured cord compression in six C0-T2 specimens under simulated head-first impact that produced common clinically-observed spinal column fractures. As a biofidelic cord response was desired, a surrogate spinal cord with *in vivo*-like mechanical properties was used. A follower preload system was also incorporated into the model to simulate the influence of musculature on the kinematic

response of the cervical spine in head to ground impact. The cord deformation varied with the injury. In all cases the maximum compression was transient and it was greater than the final compression which in all cases was zero. However, since the cord did not fill the entire canal, the final measure of canal occlusion will be found by diagnostic radiograph and dissection.

In the past, cord compression has been extrapolated from bony canal occlusion data (Carter, 2000). Our method of measuring cord compression addresses several of the issues raised by the use of a transducer such as the biofidelity and the lack of a continuous deformation profile. The former issue has been addressed by Pintar and colleagues (1995) with their pressure-sensitive, biofidelic surrogate cord. However their method did not measure the degree of compression or provide a complete continuous profile. Our method addresses these issues. However our surrogate cord has only been matched to *in vivo* spinal cord material property data under tension, although transverse compression properties have also been characterized.

Qualitative analysis of the kinematic response during our impacts showed a lack of the higher order buckling pattern as seen by Nightingale (1996). Our results indicate that "snap through buckling" does not occur in the presence of a follower load which supports our hypothesis. This is the first use of the follower load system in an impact test. The goal of the follower load is to simulate the musculature of the cervical spine. However, this is a complex system which we have simplified to a single force vector.

CONCLUSION

This is the first report of cord compression based on high speed direct visualization of a biofidelic spinal cord under axial impact. We feel that this approach promises to provide clinically relevant information on the biomechanics of cord deformation that can be used to optimise animal models of spinal cord injury so that they are as consistent to human injuries as possible. In the future this approach can be used to evaluate injury prevention measures from the perspective of cord deformation.

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