Cervical spine instability following axial compression injury: A biomechanical study

P.C. Ivancic*

Biomechanics Research Laboratory, Department of Orthopaedics and Rehabilitation, Yale University School of Medicine, 333, Cedar Street, P.O. Box 208071, New Haven, CT 06520-8071, USA

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**ABSTRACT**

Background: Axial compression injuries of the cervical spine occur during contact sports, automobile collisions, and falls. The objective of this study was to use flexibility tests to determine biomechanical instability of the cervical spine due to simulated axial compression injuries.

Hypothesis: We hypothesized that the axial compression injuries cause severe biomechanical instability throughout the cervical spine.

Materials and methods: The injuries were simulated using 2.4 m/s head-first impacts of a cadaveric cervical spine model (n = 10) mounted horizontally to a torso-equivalent mass on a sled and carrying a surrogate head in protruded posture. Intact and post-impact flexibility tests were performed up to 1.5, 3, and 1.5 Nm in flexion-extension, axial torque, and lateral bending, respectively. Instability parameters of range of motion (RoM) and neutral zone (NZ) were determined for injured spinal levels and statistically compared (P < 0.05) between intact and post-impact.

Results: The sagittal instability parameters indicated extension-compression injuries at the upper and middle cervical spine and flexion-compression injuries at the lower cervical spine. Increases in extension RoM were 14.9° at the upper cervical spine and 24.9° (P < 0.05) at the middle cervical spine and in flexion RoM at C7/T1 were 25.6°. RoM and NZ increases in axial rotation and lateral bending were nearly symmetric among left and right.

Discussion: Multidirectional instability of the upper cervical spine caused by atlas and dens fractures was evidenced by increases between 36% and 53% in RoM and NZ due to the impacts. The sagittal RoM of injured spinal levels of the middle and lower cervical spine exceeded a proposed threshold for clinical instability by between 67% and 114%. The instability documented throughout the cervical spine was consistent with clinical observations of cord injuries and paralysis in patients.

Level of evidence: Level IV, controlled laboratory investigation.

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1. Introduction

High-energy axial compression can cause catastrophic injuries to the cervical spine during automobile collisions, contact sports, and falls [1–4]. A recent case study reported a left C6 facet fracture that extended into the laminar junction in a American football player due to a high-energy, head-down tackling maneuver [5]. In this maneuver, the neck is forcefully compressed between the head and torso masses and is subjected to significant axial load. Previous biomechanical studies have simulated axial compression neck injuries by vertically dropping human head-neck specimens affixed to an equivalent torso mass in an inverted posture [6–8]. These studies have observed neck buckling, consisting of extension at the middle cervical spine and flexion at the lower spinal levels, C6/7 and C7/T1, in addition to higher-order neck buckling modes. Fracture-dislocation injuries and ligamentous lesions produced in the biomechanical studies are consistent with real-life neck injuries reported clinically due to head-first impacts [9–11]. Axial compression causing neck buckling can cause clinical instability, defined as “the loss of the ability of the spine under physiological loads to maintain its pattern of displacement so that there is no initial or additional neurological deficit, no major deformity, and no incapacitating pain” [12].

Mechanical instability inherent in the definition of clinical spinal instability may be quantified, in vitro, using flexibility testing of spinal segments [13] by determining and comparing pre- and post-trauma flexibility parameters. This approach has been used to document the intervertebral level, severity, and progression of spinal injury due to simulated trauma of cadaveric spinal segments [14–16]. It has also been used to evaluate the effectiveness of spinal implants in preventing excessive spinal motion and laxity [17,18]. Although previous studies have determined the...
biomechanical time-history responses of the head and neck during simulated head-first impacts of cadaveric specimens [6–8,19–21], we are unaware of previous research that has quantified the resultant biomechanical instability of the cervical spine. Quantification of biomechanical instability due to axial compression injuries is needed to help improve our understanding of neck injury severity and differences in injury severity throughout the cervical spine.

The goal of this study was to determine biomechanical instability of the cervical spine due to simulated axial compression injuries. Head-first impacts were performed with the head initially protruded to simulate a “ducking” posture that is common prior to impact [5,22]. We hypothesized that the axial compression injuries cause severe biomechanical instability throughout the cervical spine.

2. Materials and methods

2.1. Overview

The whole cervical spine specimen was used for three-plane flexibility testing while intact and following head-first impact with initial head protrusion. The model used to simulated head-first impacts consisted of the neck specimen mounted horizontally to a torso-equivalent mass on a sled and carrying a surrogate head. Dynamic time-history response data were reported in a parallel study [20].

2.2. Specimen and model preparation

Ten fresh-frozen human osteoligamentous whole cervical spine specimens were mounted in resin at the occiput and T1 vertebra. The average age of the specimens was 81.4 years (range: 76–90 years) with five male and five female donors. Apart from typical age related degenerative changes, the specimens did not suffer from any disease or injury that could have affected the osteoligamentous structures.

To prepare the cervical spine for flexibility testing, motion tracking markers were rigidly fixed to the anterior aspects of each cervical vertebra, with the exception of those that sustained bony fracture during impact. The flags, each with three non-collinear markers, were rigidly fixed onto plastic supports, which were fixed to the vertebrae. Additional flags were rigidly secured to the occipital and T1 mounts. A loading jig was applied to the occipital mount, while the T1 mount was fixed. The combined weights of the loading jig and occipital mount were counterbalanced to ensure that they would not apply artifact loads throughout flexibility testing.

To prepare the cervical spine for head-first impact (Fig. 1), the three-dimensional motion tracking markers were removed. A modified Hybrid III surrogate head (4.6 kg mass; 0.0214 kg m² sagittal moment of inertia; Humanetics Innovative Solutions, Plymouth, MI, USA) was rigidly attached to the occipital mount in anatomical position. The T1 mount of the specimen was rigidly fixed to the front of a torso-equivalent mass, 55.5 kg, which was fixed to an impact apparatus. When mounted to the impact apparatus, the average anterior tilt of the T1 vertebra was 27.1° (SD 5.7°), consistent with in vivo thoracic kyphosis [23,24]. An upward force was used to counterbalance the head weight and achieve head protrusion immediately prior to impact. Muscle forces were not simulated.

2.3. Three-plane flexibility testing

Three-plane flexibility testing was performed on each cervical spine specimen while intact and post-impact. Pure moments were applied to the occipital mount in 3 equal steps up to peak loads of 1.5, 3, and 1.5 Nm in flexion-extension, axial torque, and lateral bending, respectively (Fig. 2a). A custom-built loading apparatus was used for automated flexibility testing, which applied equal and opposite forces to the loading jig using cables attached to pistons which were loaded via a vacuum pump (Fig. 2b). At each moment step, the loading was held constant for 30 seconds to allow for viscoelastic creep, after which time kinematic data were recorded. Two preconditioning cycles were performed and data were recorded on the third loading cycle. A custom-built loading apparatus was used for automated flexibility testing. The kinematic data were measured using the Optotrak three-dimensional motion measuring system (Optotrak 3020, Northern Digital, Waterloo, Ontario, Canada). For the upper cervical spine, motions of the occiput were expressed relative to and in the coordinate system of the C2 or C3 vertebra. For the subaxial cervical spine, motions of each vertebra were expressed relative to and in the coordinate system of the directly inferior vertebra. Each vertebral coordinate system, which was fixed to and moved with the vertebra, had its z-axis horizontal and positive forward, y-axis vertical and positive upward, and x-axis horizontal and positive to the left and was initially aligned with the ground coordinate system in neutral posture. The Euler angles were calculated at each load increment for each spinal level in the sequence Rx, followed by Ry and Rz [26,27]. Flexion, left axial rotation, and right lateral bending were positive while extension, right axial rotation, and left lateral bending were negative. Average (SD) system errors, as determined in a separate study [28] were −0.05° (0.05°), −0.03° (0.04°), and −0.01° (0.02°) for rotations around the x, y, and z axes, respectively.

2.4. Simulated head-first impact with head protrusion

Simulated head-first impacts with initial head protrusion (Fig. 1) were performed at an average impact velocity of 2.4 m/s (SD 0.1 m/s). Average (SD) initial head protrusion immediately prior to impact was 5.2 cm (1.1 cm), compression was 1.8 cm (1.3 cm), and flexion was 9.6° (5.5°) relative to normal lordotic neutral posture. This head protrusion posture caused extension at the upper cervical spine and flexion at the subaxial spinal levels. The sled containing the torso mass, neck specimen, and head was accelerated using an acceleration generation system consisting of a piston, high-energy compression springs, and computer controlled electromagnetic release. Axial compression neck injuries were caused by abrupt deceleration of the head into a padded barrier and continued forward torso momentum. Macroscopic neck injuries were determined by fluoroscopy and visual inspection following the impacts and prior to the post-impact flexibility tests.
Fig. 2. a: flexibility testing in which pure moments were applied in flexion, extension, axial torque, and lateral bending. Spinal motions were determined using vertebral motion tracking markers rigidly fixed to the anterior aspects of the cervical vertebrae; b: schematic drawing of the loading apparatus modified from Yamamoto et al. [25] which includes paired vacuum-operated, low-friction glass pistons which move as freely as the loading jig in order to produce pure moments. The directions of the loads applied by the pistons can be changed to enable pure moments applied in flexion-extension, axial rotation, or lateral bending.

Fig. 3. Flexibility testing protocol in which pure moments were applied in 3 equal steps up to peak loads of 1.5, 3, and 1.5 Nm in flexion-extension, axial torque, and lateral bending, respectively. Motion data were recorded on the third loading cycle. The rotation-moment curves, ranges of motion (RoMs), and neutral zones (NZs) were determined for the specimens while intact and post-impact.

2.5. Data analyses

Rotation-moment curves were plotted for the spinal levels that sustained macroscopically identifiable injuries. The biomechanical instability parameters of range of motion (RoM) and neutral zone (NZ) (Fig. 3) were computed from the rotation-moment curves for each motion direction of the intact and post-impact data. The RoM represents the total rotation of the spinal region while the NZ is the part of the RoM that is produced with minimal internal resistance and is the zone of high flexibility or laxity [13]. The RoMs and NZs were averaged separately for the upper, middle, and lower spine regions that sustained injuries. Paired t-tests ($P < 0.05$) were performed to determine significant differences in the average RoM and NZ between intact and post-impact. Adjusted $P$-values were computed based upon the number of statistical tests performed.

3. Results

The impacts caused hyperflexion at C6/7 and C7/T1 and hyperextension at the superior spinal levels. Distinct injuries were observed throughout the cervical spine, summarized in Table 1 from a parallel study [20]. The injuries consisted of atlas fractures (specimens 7 and 9) and type III dens fracture (specimen 9) at the upper cervical spine, complete injuries of the anterior longitudinal ligament and disc at the middle cervical intervertebral levels (specimens 1, 2, 4, 5, and 7), and complete injuries of the most ligaments at C7/T1 (all specimens). Eight specimens were grossly unstable at C7/T1 and required solid fusion at this spinal level prior to the post-impact flexibility tests, thus reducing the C7/T1 sample size to 2.

Specimen-specific plots of rotation vs. moment for the spinal levels that sustained injuries are presented in graphical form for the upper (Fig. 4a), middle (Fig. 4b), and lower (Fig. 4c) cervical spine. Post-impact RoM and NZ were consistently larger than intact. Biomechanical instability was observed at C0/C2 in axial rotation due to atlas fracture (27° increase in left plus right axial RoM; specimen 7; Fig. 4a) and at C0/3 in extension, axial rotation, and lateral bending due to combined atlas and type III dens fracture (27.4° RoM and 26.8° NZ increases in extension; specimen 9; Fig. 4a). Among injured intervertebral levels of the middle cervical spine, C2/3 through C4/5, extension RoM and NZ consistently increased...
Fig. 4. Rotation-moment curves of the spinal levels that sustained injuries: (a) upper cervical spine, (b) middle cervical spine, and (c) lower cervical spine, C7/T1. Intact (○) and post-impact (●) data are shown. Flexion, left axial torque/rotation, and right lateral bending are positive while extension, right axial torque/rotation, and left lateral bending are negative.
Table 1

Cervical spine injuries. The injuries were due to head-first impact with initial head protrusion, summarized from a parallel study [20]. Complete and partial macroscopic injuries were documented in the following ligaments of the middle and lower cervical spine: anterior longitudinal ligament (A), anterior (a) and posterior (p) disc (D), posterior longitudinal ligament (P), right (r) and left (l) capsular ligament (C), ligamentum flavum (L), and interspinous (l) and supraspinous ligaments (S). Other injuries included 3-part atlas fractures (AF), Type III dens fracture (DF), spinous process fracture (SPF), and forward dislocation (FD).

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Upper cervical spine</th>
<th>Middle cervical spine</th>
<th>Lower cervical spine: C7/T1</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Complete injury</td>
<td>Partial injury</td>
<td>Fracture/dislocation</td>
</tr>
<tr>
<td>4</td>
<td>C3/2: A, Da</td>
<td>C3/2: Dp,P</td>
<td>D, P, CrL, L, S</td>
</tr>
<tr>
<td>6</td>
<td>AF</td>
<td>C4 SPF</td>
<td>Da</td>
</tr>
<tr>
<td>7</td>
<td>AF, DF</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

between 10.0° and 36.6° (Fig. 4b). Specimen 2 (C2/3) and specimen 7 (C4/5) demonstrated the largest increases in three-plane RoM and NZ as compared to the other injured levels of the middle cervical spine. At C7/T1, flexion RoM and NZ increased by 14.0° to 36.0° due to the impacts which demonstrated the flexion mode of injury at this spinal level (Fig. 4c).

Average RoMs and NZs of the upper, middle, and lower cervical spine from the intact and post-impact flexibility tests appear in Fig. 5. Among sagittal parameters, the increases due to the impacts were largest in extension at the upper and middle cervical spine and in flexion at the lower cervical spine. The impacts caused a significant increase in extension RoM at the middle cervical spine (27.9° vs. 3.0°, Fig. 5a). Small sample sizes precluded statistical testing of the biomechanical instability parameters of the upper and lower cervical spine regions. Average increases due to the impacts were 14.9° in extension RoM at the upper cervical spine and 25.6° in flexion RoM at C7/T1. While distinct instability modalities were observed in the sagittal plane among spinal regions (Fig. 5a), the average increases due to the impacts in axial rotation (Fig. 5b) and lateral bending (Fig. 5c) demonstrated near symmetric instability among left and right RoM and NZ.

4. Discussion

The present study quantified biomechanical instability of the cervical spine due to axial compression injuries. The injuries were simulated by head-first impacts with initial head protrusion using a model consisting of a human cadaver neck mounted horizontally to a torso-equivalent mass on a sled and carrying a surrogate head (Fig. 1) [20]. The axial load combined with bending moments caused flexion-compression injuries at the lower cervical spine and extension-compression injuries at superior spinal levels (Table 1). Intact and post-impact flexibility tests were used to quantify increases in the biomechanical instability parameters of RoM and NZ of the injured spinal levels. The sagittal injury modes determined from these parameters were consistent with the modes of dynamic loading throughout the neck. A significant increase of 24.9° in extension RoM was observed at the middle cervical spine due to the impacts (Fig. 5a). Among sagittal biomechanical instability parameters of the adjacent spinal regions, average increases of 14.9° in extension RoM at the upper cervical spine and 25.6° in flexion RoM at C7/T1 were observed. These data document gross biomechanical instability throughout the cervical spine due to head-first impacts, consistent with our hypothesis and with clinical observations of cord injuries and paralysis in patients [9–11].

Our model and results are limited by factors inherent to in vitro studies. The average age of our specimens, 81.4 years, was older than the younger population who are at risk of injury due to the}

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high-energy axial compression neck injuries. As the cervical soft tissues generally become stiffer and mobility decreases with age [29], greater neck motions of the younger population would be expected. However, biomechanical instability in the present older specimens due to the impacts is likely similar to that of the younger population. Cervical spine instability was determined due to head-first impacts with a specific initial posture: head protrusion. This impact configuration produced dynamic neck motions consistent with those observed in historic experiments [6-8]. We did not determine the effects of varying initial head-neck posture, head or torso masses, or impact speed on the biomechanical instability parameters. We used a torso equivalent rigid mass of 55.5 kg which represented that of a heavier athlete or obese male [30]. We assumed that the entire torso mass acted on the neck due to its forward momentum during head-first impact. This represented the worst-case scenario for causation of neck injuries. In real-life head-first impacts, neck loading due to head impact and forward torso momentum is decoupled. Muscle forces were not simulated and their effects on the resulting injuries and instability are not known. Due to a lack of young cadaveric material, our sample size was limited to older 10 specimens. Although 10 cervical spine specimens were studied, sample sizes of injured upper, middle, and lower cervical levels were less than 10 (Table 1). Although our model and results are limited by these factors, our study was successful in identifying the locations and severity of biomechanical instability in the cervical spine due to simulated head-first impacts.

Axial compression injuries of the upper cervical spine have been simulated in previous biomechanical experiments of human cadaveric segments. Panjabi et al. [31] and Oda et al. [32] applied high-energy weight drop to occupy through C3 specimens and observed primary atlas fractures and associated injuries including occipital condyle fractures and teardrop and traumatic spondylolisthesis of the axis. Comparison of intact and post-impact flexibility data indicated the greatest increases in sagittal biomechanical instability parameters: 90% in NZ and 40% in RoM, followed by 20% in lateral bending, and 10% in axial rotation [31,32]. In contrast, the present study simulated axial compression injuries to the whole cervical spine. In the 2 specimens that sustained upper cervical spine injuries (specimen 7: atlas fracture; specimen 9: atlas and dens fractures; Table 1), increases in RoM and NZ were 47% and 53% for total sagittal motion, 43% and 50% for axial rotation, and 37% and 36% for lateral bending, respectively (Fig. 5). These data indicate greater increases in sagittal RoM, axial rotation, and lateral bending, as compared to the previously reported data. Our results indicate multidirectional instability of the upper cervical spine evidenced by increases in instability parameters between 36% and 53% due to head-first impact.

The diagnosis of clinical instability is confounded by factors that may affect spinal motions measured from radiographic or MRI tests, such as muscle spasm, pain, and/or patient non-compliance [33,34]. White and Panjabi [12] proposed radiographic measures and neurologic observations for diagnosis of cervical spine instability in trauma patients. They proposed that sagittal rotation in excess of 20° at any spinal level of the middle or lower cervical spine indicates instability. The present average post-impact RoM data of injured intervertebral levels of the middle and lower cervical spine indicate gross instability. At the middle cervical spine, total sagittal RoM was 42.8°, more than double the proposed instability threshold. At the lower cervical spine, C7/T1, it was 33.4°, or 67% above the proposed threshold. While the increases in the axial rotation and lateral bending instability parameters were generally symmetric among left and right (Figs. 5b, c), increases in the sagittal parameters were largest in extension at the middle cervical spine and in flexion at the lower cervical spine (Fig. 5a).

5. Conclusions

Our study used flexibility testing to quantify biomechanical instability throughout the cervical spine due to simulated head-first impacts with initial head protrusion. In real-life, these impacts occur during contact sports, automobile collisions, and falls and may cause complete or incomplete cervical spine fractures, dislocations, or even death [1–5]. Although limited by a small sample size, our biomechanical results indicate potential instability throughout the entire cervical spine due to axial compression injuries, consistent with our hypothesis. Future work may investigate the effects of varying initial head-neck posture, head or torso masses, or impact speed, on the biomechanical instability parameters during simulated head-first impacts. Ultimately, these investigations provide data towards improving awareness of potential catastrophic neck injuries, and for development of injury prevention systems and improved training techniques in contact sports.

Disclosure of interest

The author declares that he has no conflicts of interest concerning this article.

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