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The effects of ligamentous injury in the human lower cervical spine

P. Devin Leahy [a], Christian M. Puttlitz [a,b,c,*]

[a] Orthopaedic Bioengineering Research Laboratory, Department of Mechanical Engineering, Colorado State University, Fort Collins, CO 80523, USA
[b] School of Biomedical Engineering, Colorado State University, Fort Collins, CO 80523, USA
[c] Department of Clinical Sciences, Colorado State University, Fort Collins, CO 80523, USA

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ABSTRACT

Damage is often sustained by the anterior longitudinal ligament (ALL) and ligamentum flavum (LF) in the cervical spine subsequent to whiplash or other cervical trauma. These ligaments afford substantial cervical stability when healthy, but the ability of the ALL and LF to stabilize the spine when injured is not as conclusively studied. In order to address this issue, the current study excised ALL and LF tissues from cadaveric spines and experimentally simulated whiplash-type damage to the isolated ligaments. Stiffnesses and toe region lengths were measured for both the uninjured and damaged states. These ligamentous mechanical properties were then inputted into a previously-validated finite element (FE) model of the cervical spine and the kinematic effects of various clinically relevant combinations of ligamentous injury were predicted. The data indicated three and five-fold increases in toe region length for the LF and ALL injury variants, respectively. These toe length distensions resulted in FE predictions of supra-physiologic ranges of motion, and these motions were comparable to spines with no ligamentous support. Finally, a set of cadaveric cervical spine ligament-sectioning experiments confirmed the FE predictions and supported the finding that partial injury to the relevant ligaments produces equivalent cervical kinematic signatures to spines that have completely compromised ALL and LF tissues.

1. Introduction

Cervical spine ligaments are frequently damaged during head impact and/or other cervical trauma, potentially leading to spinal instability (Ivancic et al., 2004; Panjabi et al., 2004). A common injury mechanism is whiplash, which is capable of damaging both the posterior and/or anterior ligamentous structures (Tominaga et al., 2006). Depending on the load application vector, whiplash injuries may involve cervical hyperextension and/or hyperflexion (Foreman and Croft, 2002). Hyperextension places the anterior longitudinal ligament (ALL) and facet capsule (FC) ligaments at risk, and hyperflexion motion often damages the flaval (LF) and interspinous ligaments (ISL) (Ivancic et al., 2004; Panjabi et al., 2004). Previous *in-vitro* whiplash simulations have measured the magnitude of ligamentous elongation during these loading events, and have also found total quasi-static cervical flexion and extension range of motion (ROM) after a simulated traumatic whiplash event to be supra-physiologic (Ito et al., 2004; Ivancic et al., 2004; Panjabi et al., 2004). However, these studies examined the cervical spine in its entirety, where it is difficult to discern the mechanical contribution of specific ligaments relative to auxiliary structural tissues (such as the intervertebral discs).

Maintaining proper cervical stability is critical, since supra-physiologic ROM can produce nervous tissue impingement and injuries to other peri-spinous soft tissues (White and Panjabi (1990); Ivancic et al., 2004; Hogan et al., 2005; Dickerman et al., 2006). It is unknown what capability the injured ligaments have in limiting cervical ROM below non-damaging levels. Thus, determining the mechanical properties of these injured ligaments is a vital step in understanding how to clinically treat and stabilize severe whiplash injuries, thereby minimizing the potential for permanent neural impairment (Dickman et al., 1991; Hogan et al., 2005; Panjabi et al., 2006).

In response, the goal of the current study was to measure the effect of post-traumatic damage on the mechanical properties of cervical ligaments, and model the kinematic alterations experienced by the cervical spine as a result of these specific ligamentous injuries. The initial aim was accomplished by excising individual ligaments from the cervical spine, wherein the intact (uninjured) and post-injury mechanical properties of these tissues could be consecutively measured. Testing of ligamentous properties focused on the ALL and LF. The ALL and LF ligaments were chosen as they: (1) have been found to greatly influence spinal mechanics; (2) have vastly different ratios of collagen to

* Corresponding author at: Colorado State University, Department of Mechanical Engineering, Orthopaedic Bioengineering Research Laboratory, 1374 Campus Delivery, Fort Collins, CO 80523-1374, USA. Tel.: +1 970 491 0956.
E-mail addresses: puttlitz@engr.colostate.edu, christian.puttlitz@colostate.edu (C.M. Puttlitz).

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elastin content; and (3) are commonly injured from excessive strains experienced during whiplash-type trauma (Panjabi et al., 1975; Ivancic et al., 2004; Mow and Huiskes, 2005). Assessment of the effects of ligamentous injury on cervical flexibility was accomplished by inputting the measured ligamentous mechanical properties into finite element (FE) models of the cervical spine.

2. Materials and methods

The methodology involved three phases: (1) physical experimentation on ligaments to determine the alterations in their mechanical properties due to damage; (2) FE predictions of the cervical kinematics that result from this ligamentous damage; and (3) cadaveric cervical spine experimentation to demonstrate the predictive accuracy of the FE model.

2.1. Ligament damage experimentation

Five male and two female cadaveric C0–C7 spines (average age: 53.1 years) were denuded of their musculature and other extraneous tissues with care taken to preserve the ligamentous structures. Bone-ligament-bone specimens were extracted from the spines with a diamond-bladed bandsaw (Exakt model 30/736, Exakt Apparetebau GmbH & Co., Norderstedt, Germany). The total yield of the 7 spines included 14 ALL (C2–C3: $n=6$, C4–C5: $n=6$, C5–C6: $n=1$, C6–C7: $n=1$) and 12 LF (C2–C3: $n=7$, C4–C5: $n=4$, C6–C7: $n=1$) specimens. The bony portions of the specimens were potted in poly-methyl-methacrylate (PMMA). Self-tapping screws were inserted into the osseous tissues to increase purchase and reduce the possibility of slippage within the PMMA. Visual inspection of the bony tissues' position within the PMMA was made both before and after the testing procedure to ensure slippage did not occur.

Tensile testing of the individual ligaments was accomplished with a servo-hydraulic loadframe (Mini Bionix II, model 858, MTS, Eden Prairie, MN; Fig. 1). Displacement was measured via a crosshead-mounted linear variable differential transformer (LVDT), and the associated force was measured by an inline load-cell (5 kN capacity, Model 661.19-01, MTS, Eden Prairie, MN). Specimens were housed within a heated, saline-filled tank which maintained physiologic temperature (37 °C) and hydration. Reference position (displacement$=$0 mm) was defined by the extension of the resting ligaments when weighted by the upper potting box (approximately 300 g) while submerged in saline. The loadframe actuator was lowered until a hole in the actuator fixture aligned with a hole in the upper potting box. A shear pin could only be inserted when the two components were vertically aligned within $\pm$0.01 mm with no preload force, ensuring an equal starting position and tension for all ligaments. The lower potting box was mounted on a custom-made, biaxially-translating table to enable horizontal alignment. An automated testing sequence was developed that required no user intervention beyond zeroing the LVDT and load-cell at the beginning of the test. The testing sequence was accomplished using the following protocol: (1) increase displacement to induce a 5 N tensile load, hold displacement at this level for 10 min for tissue relaxation; (2) apply 120 cycles of sinusoidal displacement (0.0 to 0.4 mm relative to the displacement at the beginning of the step) at 1 Hz for preconditioning, and quasi-statically (0.2 mm/s) ramp from zero displacement to 40 N to determine initial stiffness; (3) dwell 10 min; (4) apply 120 cycles of sinusoidal displacement for preconditioning and quasi-statically ramp from zero displacement to 40 N for a duplicate initial stiffness measurement; (5) induce partial ligament damage (detailed below); (6) apply 120 cycles of sinusoidal displacement for preconditioning and quasi-statically ramp from zero displacement to 40 N to determine the final stiffness at 10, 30, and 90 min after the damage step. The 40 N maximum force was chosen as it tensioned the ligaments throughout the extension ranges typically experienced during normal, physiologic cervical motion (Womack et al., 2011).

The partial ligament damage protocol (Step 5) was executed by quasi-statically preloading the ligaments to 10 N to remove appreciable slack in the tissues and testing mechanism, rapidly tensioning the ligaments at 50 mm/s, and immediately reversing the actuator at 35 mm/s when the load-cell detected a specified drop in force that indicated initial tearing of the ligaments. Approximately 0.05 s were required to accelerate the actuator to 50 mm/s from the 0.2 mm/s rate immediately preceding damage. The force drop values were set to 1% for the ALL specimens and 3% for LF specimens. These magnitudes were determined from pilot experiments to consistently induce damage without completely compromising the ligaments. The loading rate was modeled after a strain rate of 10/s, which has been reported for impact trauma events (Lucas et al., 2008). Force and displacement data were recorded at 205 Hz for the quasi-static loads and 1024 Hz for the high-rate loads.

Stiffness values were calculated via linear regression of the force/displacement data at discrete force intervals: between 10–20 N, 20–30 N, and 30–40 N. The calculation of stiffness at discrete force intervals was undertaken to account for the permanent yield expected after inducing damage as well as to capture the nonlinear mechanical behavior typically observed in ligaments. Displacement at the 10 N (which defined the initial toe region length) and 40 N force levels were also compared before and after injury to quantify the amount of permanent distension due to the damage protocol. Percentage change values were calculated relative to the pre-damage state.

Statistical analyses were performed to determine significant differences between the stiffness and initial toe length readings for each of the five measurement timepoints specified in the testing protocol. A square root transformation was used to normalize the residuals, and therefore these values were independent of the means (SAS V9.2, Cary, NC). Analyses were conducted with a randomized block design, blocking on ligament type (ALL or LF) and measurement time as a fixed effect. P-values less than 0.01 were considered significant.

2.2. Computational modeling

A previously-developed C3–C7 FE model (ABAQUS V6.9-EF2, Dassault Systèmes Simulia Corp, Providence, RI) of the human cervical spine was modified to simulate the mechanical effects of hyperextension and hyperflexion injuries at the most common level of whiplash injury, C5–C6 (Ivancic et al., 2004). This model has been previously validated and converged for intact behavior (Womack et al., 2011). In brief, the anatomic geometry was generated from a single computed-tomography (CT) scan of an average-sized cadaveric spine. Validation was accomplished by pure-moment testing of cadaveric (C3–C7) osteoligamentous spines ($n=6$) where intervertebral ROM, facet contact pressure, cortical strain in the lamina, nucleus pulposus pressure, and annulus fibrosus bulge were measured during quasi-static loadings. These loads consisted of $\pm$2 Nm moments applied to the C3 vertebra in the axial rotation, lateral bending, and flexion and extension directions via a custom-designed, force-feedback robotic testing arm. C7 was rigidly fixed to a 6-degree of freedom load cell. Convergence was achieved by

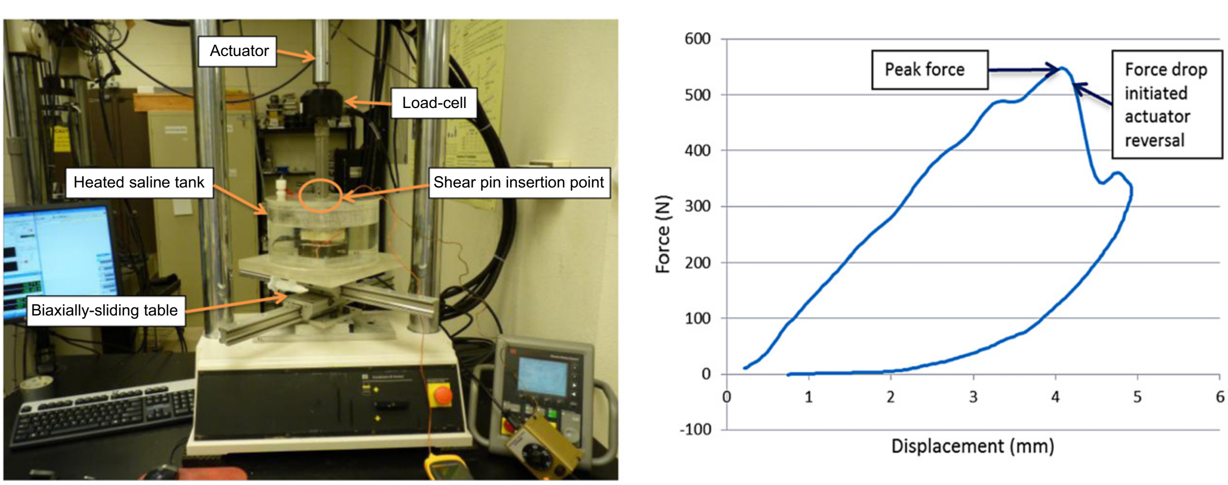

Fig. 1. (Left) The ligament tensile-testing apparatus. (Right) A typical damage step force/displacement curve at 50 mm/s loading. It can be noted that displacement continued to briefly increase after the force drop before the actuator decelerated and reversed, indicating system lag (average of 23 ms).