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Viscoelastic properties of the cervical spinal ligaments under fast strain-rate deformations

Scott R. Lucas [a,*], Cameron R. Bass [a], Robert S. Salzar [a], Michelle L. Oyen [b], Chris Planchak [c], Adam Ziemba [d], Barry S. Shender [e], Glenn Paskoff [e]

[a] *Center for Applied Biomechanics, University of Virginia, 1011 Linden Avenue, Charlottesville, VA 22902, USA*
[b] *Cambridge University Engineering Department, Cambridge CB2 1PZ, UK*
[c] *Naval Surface Warfare Center Carderock Division, 2600 Tarawa Ct, Norfolk, VA 23521, USA*
[d] *Triodyne Inc., 666 Dundee Road, Northbrook, IL 60062, USA*
[e] *Naval Air Systems Command, 48110 Shaw Road, Patuxent River, MD 20670, USA*

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Abstract

The mechanical response of ligaments under fast strain-rate deformations is a necessary input into computational models that are used for injury assessment. However, this information frequently is not available for the ligaments that are routinely injured in fast-rate loading scenarios. In the current study, experiments were conducted at fast strain rates for the cervical spinal ligaments: the anterior longitudinal ligament, the posterior longitudinal ligament and the ligamentum flavum. Bone–ligament–bone complexes at three spine levels were harvested for mechanical testing. Displacement-controlled sub-failure uniaxial tensile tests were performed in both load–relaxation and sinusoidal conditions. A nonlinear (separable) viscoelastic model was used to examine the experimental data. An unexpected result of the modeling was that the instantaneous elastic functions could be approximated as linear for these strain rates. A five-parameter model was sufficient to characterize the ligament viscoelastic responses and had good predictive capacity under different applied loading conditions.

Keywords: Spine; Ligament; Viscoelasticity; Strain-rate dependence; Biomechanics

1. Introduction

The cervical spinal ligaments serve important functions for stabilization and motion during normal neck function. However, these ligaments can be subject to excessive mechanical loading in several different injury scenarios. Whiplash, or neck hyperextension, is debilitating, with large societal costs and a widespread prevalence in automotive crashes [1]. It has been shown that cervical spinal ligament injury is possible during both rear impacts [2] and frontal impacts [3,4]. Injuries to the anterior longitudinal ligament (ALL) have been clinically documented in whiplash cases [5]. Military events which may cause ligamentous spinal injury include aircraft ejection [6], repetitive loading in high-speed boats and tanks [7], and aircraft crashes [8–10].

Because of the morbidity associated with cervical spine ligament injury, there is an increasing interest within both the automotive industry and the military to ascertain the material properties – both deformation and failure – of cervical spinal ligaments. Computational models of the spine are effective tools to investigate and to develop ways to mitigate these injuries. In order to produce effective computational models, it is imperative to have accurate material properties of the model constituents that are valid within the rates of an injury scenario. In aircraft ejection, cervical spinal ligaments may exhibit fast strain-rate deformations

* Corresponding author. Tel.: +1 434 296 7288; fax: +1 434 296 3453.
E-mail address: slucas@virginia.edu (S.R. Lucas).

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greater than 10 s^{-1}. There is currently a lack of information regarding the fast-rate deformation and failure properties of the cervical spinal ligaments, which prevents the development of accurate computational models for neck injury. The three primary cervical spine ligaments are the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL) and the ligamentum flavum (LF). The ALL is located on the anterior aspect of the vertebral column, the posterior longitudinal ligament (PLL) is located on the posterior aspect of the vertebral column and the ligamentum flavum (LF) extends vertically from adjacent superior to inferior vertebral lamina. The primary function of the ALL and PLL is to provide stability of the head and neck. The LF resists abrupt flexion in the vertebral column and prevents injury to the intervertebral discs. If these ligaments are compromised, it is more likely that vertebral column instability or intervertebral disc injury will result [11].

The biomechanical properties of cervical spinal ligaments have been studied at quasi-static strain-rate deformations ($\sim$0.005–1 s^{-1}) [12,13] and fast strain-rate deformations (greater than 10 s^{-1}) [14–16]. These studies did not include viscoelastic deformation characterization of the cervical spinal ligaments. For accurate computational modeling of an impact, it is imperative to include the ligament fast-rate relaxation properties. There exists a phenomenological viscoelastic model of the intact cervical spine in torsion [17]; however, to the authors' knowledge, there are no existing viscoelastic models for human cervical spinal ligaments in any loading conditions.

The current study was undertaken to address the lack of knowledge of cervical ligament deformation under fast strain rates. Experiments were conducted on the isolated ALL, PLL and LF bone–ligament–bone complexes from male and female human cervical spines under fast-rate deformations. A viscoelastic model was used to quantify responses; the model was validated using model predictions as compared with experimental data conducted under different loading conditions. An assessment of the influence of the number of viscoelastic coefficients was performed. Finally, a comparison of the viscoelastic coefficients was made between gender and among spinal levels.

2. Materials and methods

The cervical spines from six male and five female human cadavers were used in this study. Post-mortem human use at the Center for Applied Biomechanics is subject to review by the University of Virginia Cadaver Use Committee. The specimens were unembalmed and initially frozen. The average ($\pm$standard deviation (SD)) age, stature and mass for male specimens was 60 ± 8 years, 1749 ± 40 mm and 77 ± 17 kg and for female specimens was 58 ± 6 years, 1626 ± 40 mm and 62 ± 19 kg. Cervical spines were thawed and transected into the C3–C4, C5–C6, and C7–T1 functional spinal units (FSUs). The FSUs were segmented by cutting through the articular facets and removing any ligamentous structures holding them intact. Once the C3–C4, C5–C6 and C7–T1 FSUs were sectioned, the ALL, PLL and LF at each level could be isolated. A vertical band saw was used to isolate each bone–ligament–bone complex. The ALL was first isolated by cutting along the coronal plane at the midline of the adjacent vertebral bodies. The PLL was then isolated by cutting through the coronal plane of the adjacent pedicles. The LF was then isolated by cutting midway between the anterior and posterior aspect of the adjacent spinous processes. The remaining soft tissues around each bone–ligament–bone complex were left intact to preserve hydration. Additionally, saline-saturated gauze was wrapped around each specimen. Small (#4, 3/4 in.) wood screws were inserted into the bone on either side of the ligament and were used as an adherent to a two-part urethane casting resin (Fast Cast, Goldenwest, Cedar Ridge, CA). The bone–ligament–bone complexes were potted in aluminum cups, with the ligament spanning the gap between the cups, and the specimens were individually bagged and placed in a temperature-controlled water bath [14]. Hydration and temperature closely represented physiological conditions. The ligament isolation procedure resulted in a total of 99 ligaments as bone–ligament–bone complexes (3 ligaments at 3 FSUs for 11 specimens). Excluding ligaments that were damaged during preparation, 85 of the 99 ligaments were used in the current study. Of the 85 ligaments tested, 25 were ALLs, 31 were PLLs and 29 were LFs.

The potted bone–ligament–bone complexes were mounted in a universal test machine (Instron, Inc. # 8874 Canton, MA) for uniaxial tensile tests. Soft tissues remaining from the isolation procedure were then removed to completely isolate the bone–ligament–bone complex. Ligaments were aligned with an X–Y positioning table in a superior–inferior orientation that represented physiological conditions. The fixture was enclosed in an environmental chamber to maintain physiological temperature (37.2 ± 0.6 °C) and humidity (>90%). For a detailed schematic of the specimen orientation and test fixture, see the companion paper on ligament failure properties [14].

Mechanical tests were conducted within a framework of applied engineering strains, where engineering strain (ε_E) is defined as the ratio of input (gage) displacement, Δl, to initial ligament length, l_0. A zero strain state was defined by applying a 4 N tensile preload to the specimen (cf. [18,19]). After the preload, the initial ligament length (l_0) was measured using digital calipers. For the ALL and PLL, the initial length was defined as the distance between the distal endplate of the superior vertebral body and the caudal endplate of the inferior vertebral body with the specimen mounted into the test frame. For the LF, the initial length was defined as the distance between the distal surface of the superior lamina and the caudal surface of the inferior lamina. The average $\pm$1 SD of l_0 of the ALL, PLL and LF was 3.47 ± 0.81, 3.26 ± 0.85 and 3.87 ± 1.58 mm. For each test, the input displacement was determined by the necessary input engineering strain. Each ligament was preconditioned with a 10% ε_E sinusoidal input at 2 Hz for 120 cycles (P_{10}). Then each ligament was subjected to tensile ramp-hold inputs to engineer-