MECHANICAL PROPERTIES OF INJURED HUMAN CERVICAL SPINE LIGAMENTS AND CORRESPONDING EFFECT ON SPINAL KINEMATICS

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INTRODUCTION
The assessment of cervical spine instability following traumatic injury is controversial [1, 4, 5, 8]. Typical definitions of cervical instability are based on the presence of several key detectable injuries using simple radiographs, computed tomography (CT), and magnetic resonance (MR) imaging. Although these imaging modalities have been shown to be relatively reliable for detection of fractures and very large soft tissue injuries, they are largely deficient for determining the presence of smaller soft tissue injuries, such as hyperstrengthened ligaments [1, 3]. Soft tissue injuries of this nature may be revealed with dynamic range of motion (ROM) assessment, such as a flexion and extension test with radiography. However, these tests are currently inadequate for determining the existence of specific injuries. Cervical soft tissue injuries demand further analysis, given the risk of severe and permanent neurological impairment that may accompany these injuries [2, 5].

This study specifically focused on partial injury to the anterior longitudinal ligament (ALL) and the ligamentum flavum (LF). These two ligaments are susceptible to injury in whiplash (rapid flexion-extension) loading [6, 9]. The injuries were simulated between the C5 and C6 vertebrae, a commonly injured region in whiplash [6].

Experimental data resulting from tensile testing of healthy and injured ligaments were utilized to modify a previously validated and converged finite element model of the intact lower cervical spine (C3-C7) [12]. Intact ligament properties at the affected (C5-C6) level were modified to reflect partially injured and completely injured cases. Resultant ROM due to application of a 0.75 N-m moment in the axial rotation, flexion+extension, and lateral bending directions was measured.

MATERIALS AND METHODS

Experimental Setup
Fresh-frozen cadaveric human cervical spines (n=3) that exhibited no evidence of degeneration were harvested. Musculature, intervertebral discs, facets, and other extraneous tissue were removed, with care taken to preserve the ALL and LF. Saline hydration was utilized at 15 minute intervals during preparation of the specimens. Bone-ligament-bone specimens were extracted from the cervical complex with a diamond-bladed bandsaw. Total yield of the spines included 6 ALL (C2-C3 n=2, C4-C5 n=2, C5-C6 n=1, C6-C7 n=1) and 6 LF (C2-C3 n=3, C4-C5 n=2, C6-C7 n=1) specimens. Specimens were potted in poly-methyl-methacrylate (PMMA). Self-tapping screws were inserted into the bone tissue to increase purchase, reducing the possibility of specimen slippage within the PMMA.

Tensile testing was accomplished with a servo-hydraulic load frame (Mini Bionix II, model 858, MTS, Eden Prairie, MN). Displacement was measured via a crosshead-mounted linear variable differential transformer (LVDT). Force was measured by an inline-mounted 5 kN load cell (Model 661.19-01, MTS, Eden Prairie, MN). Specimens were housed within a heated saline filled tank, replicating physiological temperature (37°C) and hydration. Reference position (displacement = 0 mm) was determined by the position of the resting ligament weighted by the upper potting apparatus in saline (300g). An automated testing program was developed, requiring no user intervention beyond zeroing the LVDT and load-cell at the beginning of the test. The testing sequence included the following steps: 1) increase displacement to induce 5 N tensile load, hold displacement at this level for 10 minutes, 2) apply 120 cycles of sinusoidal displacement (0.0 to 0.4 mm relative to displacement at 4 N of tension) for preconditioning and quasi-statically (0.2 mm/s) ramp from zero displacement to 40 N to determine initial stiffness, 3) dwell 10 minutes, 4) repeat Step 2, 5) induce partial ligament damage, 6) repeat
Step 2 at 10, 30, 90, and 270 minutes after damage step to measure final stiffness. Partial ligament damage (Step 5) was executed by preloading the ligaments to 10 N, 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, indicating initial tearing of the ligament. The force drop values were set to 1% for ALL and 3% for LF specimens, which were determined from pilot experiments to repeatedly induce damage without completely tearing the ligaments. The loading rates were modeled after a strain rate of 10/s, reported to exist in impact trauma [7].

**Statistical Analysis**

Stiffness values were measured at discrete force intervals, between 10-20 N, 20-30 N, and 30-40 N. This method accounted for the permanent yield experienced by the specimen, as well as the nonlinear mechanical behavior seen in ligaments. Displacement at the 10 N and 40 N force levels were also compared, to quantify the amount of permanent deformation.

These stiffness and displacement data were implemented in the FE model. Three ligamentous partial injury models were created at the C5-C6 level: 1) ALL and facet capsule (FC) ligament injury, 2) LF and interspinous (ISL) ligament injury, and 3) ALL, FC, ISL, and LF injury. Fitting whiplash injury pathology, the FC ligament was modeled to sustain corollary damage along with the ALL in spinal extension [9]. The FC properties were modeled to undergo a reduction in stiffness and increase in yield displacement in equivalent percentage to the property alterations experienced by the ALL. The ISL was assumed to have completely torn during damage to the LF [6]. For comparative purposes, the FE model was also modified to reflect uninjured and completely sectioned ALL, FC, ISL, and LF ligaments.

**RESULTS**

The experimental ligament data show an increase in toe length (displacement at 10 N) after damage for all specimens (increase of 2,000 mm = 699.87% for ALL, 3,236 mm = 335.61% for LF, Figure 1). Stiffness within the 10-40 N range experienced only a minor reduction for LF specimens (3.84%), but a much larger reduction for ALL specimens (62.44%).

Incorporation of these injured ligament properties into the FE model resulted in increased rotation over the intact case (Figure 2). The ALL+FC injuries yielded a larger increase to extension ROM over intact than LF+ISL injuries (0.83 vs 0 degrees). This effect was reversed for flexion loadings (0.77 vs 1.90 degrees).

**DISCUSSION**

Injuries to the ALL+FC resulted in additional extension ROM, while injuries to the posterior ligaments (LF+ISL) demonstrated increased flexion ROM. These data are congruent with previous findings [10]. Noting that LF+ISL injuries gave nearly identical flexion-extension ROM to ALL+FC injuries (10.27 vs 10.02 degrees), it is difficult to specify which ligamentous structure is injured, without knowing the dividing point between flexion and extension. However, both lateral bending and axial rotation showed a more substantial difference in ROM between LF+ISL and ALL+FC injuries (lateral: 6.20 vs 10.48 degrees, respectively; axial: 7.22 vs 10.22, respectively). This suggests ROM tests should not be limited to only flexion and extension, as is often the case [11].

**REFERENCES**


Figure 1: Displacement at 10 N (toe region) for intact and injured ligaments.

Figure 2: Intervertebral rotation for the C5-C6 level at 0.75 Nm loading for various injury conditions.