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Dynamic Mechanical Properties of Intact Human Cervical Spine Ligaments

A Thesis Submitted to the
Yale University School of Medicine
in Partial Fulfillment of the Requirements for the
Degree of Doctor of Medicine

By
Anthony Ndu

2007

Dynamic Mechanical Properties of Intact Human Cervical Spine Ligaments.

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Most previous studies have investigated ligaments mechanical properties at slow elongation rates of less than 25 mm/s. The purpose of this study was to determine the tensile mechanical properties, at a fast elongation rate, of intact human cervical anterior and posterior longitudinal, capsular, and interspinous and supraspinous ligaments, middle-third disc, and ligamentum flavum.

A total of 97 intact bone-ligament-bone specimens (C2-C3 to C7-T1) were prepared from six cervical spines (average age: 80.6 years, range, 71 to 92 years) and were elongated to complete rupture at an average (SD) peak rate of 723 (106) mm/s using a custom-built apparatus. Non-linear force vs. elongation curves were plotted and peak force, peak elongation, peak energy, and stiffness were statistically compared ($P < 0.05$) among ligament. A mathematical model was developed to determine the quasi-static physiological ligament elongation.

Highest average peak force, up to 244.4 and 220.0 N in the ligamentum flavum and capsular ligament, respectively, were significantly greater than in the anterior longitudinal ligament and middle-third disc. Highest peak elongation reached 5.9 mm in the interspinous and supraspinous ligaments, significantly greater than in the middle-third disc. Highest peak energy of 0.57 J was attained in the capsular ligament, significantly greater than in the anterior longitudinal ligament and middle-third disc. Average stiffness was generally greatest in the ligamentum flavum and least in the interspinous and supraspinous ligaments. For all ligaments,

peak elongation was greater than average physiological elongation computed using the mathematical model.

Comparison of the present results with previously reported data indicated that high-speed elongation may cause cervical ligaments to fail at a higher peak force and smaller peak elongation and may be stiffer and absorb less energy, as compared to a slow elongation rate. These comparisons may be useful to clinicians for diagnosing cervical ligament injuries based upon the specific trauma.

Acknowledgments:

This thesis project was undertaken as part of a large wide-ranging project in which there were many collaborators. I was involved in development of the study design, specimen dissection and preparation, testing, data collection and some data analysis. Paul Ivancic was involved in all aspects of the study. Yasuhiro Tominaga was involved in specimen preparation. Erik Carlson worked on data analysis and manuscript preparation. Wolfgang Rubin worked on the protocol and testing apparatus used in the study. Manohar Panjabi oversaw the entire process.

I would like to thank Paul Ivancic and Dr. Panjabi for their guidance, advice and teaching throughout my time in the lab and afterwards.

Background and Introduction

Intact spine ligaments provide passive stability to the spinal column as well as protect the spinal cord from injury (1). This is especially important in the cervical spine with its required wide range of motion for head positioning.

Injuries to the cervical spine can take the form of bony fractures, subluxations, cord compressions or ligamentous injury. Fractures, subluxations and cord compressions can often be diagnosed with modern imaging modalities. Unless a ligament fails completely or ruptures, evaluating ligament injury is a bit more challenging, as it requires expensive imaging modalities such as MRI to pick up the visual cues of ligament subfailure. The focus of the current study is to analyze cervical spine ligaments and their properties at failure.

The primary function of ligaments is to resist tensile loads, and they do this most effectively with loads in the same direction as their fibers. The second function of spinal ligaments, protecting the spinal cord, is accomplished in two ways. First, the ligaments restrict the range of motion of the spine within specific limits to prevent any compromise of the cord through hyperextension, hyperflexion, translation or rotation. Secondly, spinal ligaments absorb high amounts of energy to protect the cord in traumatic situations where high loads are applied at fast speeds (2).

Cervical Spine Ligament Structure and Function

The properties of each cervical spine ligament are dependent upon its specific anatomical location, the orientation of its fibers, geometry, and unique material composition. The main cervical spine ligaments inferior to the C2 vertebra, besides the intervertebral disc, include the

anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL).

The ALL and PLL, spanning the anterior and posterior aspects of vertebral bodies, respectively, have similar composition. The ALL originates in the basioccipital region and attaches to the anterior surface of all the vertebral bodies and intervertebral discs of the spine down to the sacrum. Occupying this position along the anterior surface of the vertebral column allows the ALL to perform its primary function of limiting extension of the spine. The PLL also originates in the basioccipital region, but unlike ALL, PLL runs along the posterior aspect of the spine down to the coccyx, attaching to the vertebral bodies and discs. The PLL consists of approximately 67% collagen and 6% elastin fibers (3-6). PLL is also thicker than its anterior counterpart, ALL. PLL appears to limit flexion and intervertebral distraction, while also protecting the spinal cord from posterior protrusion of disc material.

Located between vertebral bodies, the intervertebral disc consists of a central nucleus pulposus, which is viscous and jelly-like in the young, encased by annulus fibrosis fibers, having a high collagen composition with an increasing percentage of elastin near the cartilaginous endplates (1, 7). The fibers of the annulus fibrosis are arranged in concentric lamellae at 60° angles to one another.

This orientation allows the annulus fibrosis to resist multidirectional shear force between vertebrae, as a shear force in any direction will increase tension in the same direction as some portion of the annulus fibrosis fibers, allowing these fibers to resist the load.

The CLs, which encase facet joints, have an increasing percentage of elastin posteriorly near the LF (8). The fibers of the capsular ligaments are aligned perpendicularly to the plane of

the facet joints. This orientation allows the CLs to limit intervertebral translation and thus resist any shearing forces the spine may encounter.

The LF, the most elastic tissue in the human body, attaches at adjacent laminae bilaterally and consists of approximately 80% elastin and 20% collagen (9, 10). It has been found to have tension present in situ when the spine is in the neutral position. The value of this resting tension has been found to decrease with age from about 18N in subjects under 20 yrs old to about 5N in subjects over 70 yrs old (11). It has been proposed that this resting tension prevents protrusion of the ligament into the spinal canal during extension of the neck (11). Because the ligament is under tension in the neutral position, when the neck is extended, the tensile forces are reduced but the ligament does not go completely slack and thus does not encroach into the spinal canal. This resting tension also results in a resting compression of the intervertebral disc, which is believed to add some stability to the spine (11). As a posterior spinal ligament, the LF also functions to limit flexion of the spine.

The ISL, not present in all adult cervical spines, is composed of approximately 5% to 20% elastin (3, 12). The SSL, the most posterior cervical ligament, has been described as histologically similar to and continuous with ISL, making it difficult to identify as a separate structure (13, 14). ISL and SSL function together to limit flexion of the spine and restrict anterior horizontal displacement of the vertebrae. The specific composition of each ligament, together with its cross-sectional area and length, determines its mechanical properties.

Cervical Ligament Mechanical Properties

Ligaments are viscoelastic materials and therefore have a time dependent component to their properties. Because of this, loading ligaments at different rates, or applying the same force

over a different period of time, will lead to different biomechanical properties in the ligaments. Several previous biomechanical studies have investigated the tensile mechanical properties of human cervical spine ligaments at slow elongation rates, under 10 mm/s, and have reported peak force, peak elongation, stiffness, and energy absorbing capacity.

Chazal et al (14) studied ALL, PLL, LF, ISL and SSL from fresh cadavers and living subjects at an elongation rate of 0.01mm/sec. Interspinous and supraspinous ligaments were dissected from living subjects as surgical specimens during various surgical procedures such as repairs of herniated lumbar discs, or cord decompression surgeries. They observed the highest peak force and elongation in the PLL. Pintar et al (15) and Myklebust et al (16) tested ligaments at a rate of 10mm/sec and reported the average peak force and peak elongation of ALL, PLL, LF, CL, and ISL. CL achieved the highest peak force and elongation, ISL the lowest peak force, and ALL and PLL the lowest peak elongations. Exploring the results of these two studies we see that the 1000 fold increase in elongation rate from 0.01mm/sec to 10mm/sec affected which ligament achieved the higher peak force. Yoganandan et al (17) found that the peak stress in ALL, PLL, and CL was generally higher than in LF and ISL when they tested the ligaments at elongation rates of 10mm/sec. The peak strain in CL, LF, and ISL was generally higher than in ALL and PLL. Lastly, Przybylski et al (18) found the average peak force, stiffness, and energy of PLL were greater than those of ALL at an elongation rate of 0.33mm/sec.

Few biomechanical studies have documented the tensile mechanical properties of human cervical spine ligaments at fast elongation rates (19-21). Yoganandan et al (20) studied ALL and LF at 8.9, 25, 250 and 2500 mm/s. Peak force, stiffness, and energy absorbing capacity were found to increase with increased elongation rate, however no clear pattern emerged for peak elongation.

Panjabi et al (19) investigated the mechanical properties of alar and transverse bone-ligament-bone preparations at 0.1 and 920 mm/s. Increased stiffness and decreased elongation and energy absorption were observed at 920 mm/s, as compared to 0.1 mm/s.

Shim et al (21) studied alar and transverse ligaments and ALL, PLL, CL, LF, and ISL quasi-statically and at fast elongation rates between 10,000 and 12,000 mm/s. They found that the higher elongation rate caused increased ligament strength, but reduction in peak elongation.

There has also been some work to suggest that the site of ligament failure is dependent on displacement rate. Yoganandon et al (22) found that at slow displacement rates the ligament failed by avulsing from the vertebrae while at fast elongation rates failure was achieved through ligament substance tears.

Cervical Spine Trauma

Cervical spine ligaments work intimately with the bony structures of the spinal column to prevent instability, deformity, and injury to the neural elements. Many accidents have been associated with soft tissue injuries of the neck which can often be described by the term whiplash associated disorders (WAD). This is often a diagnosis of exclusion in symptomatic patients, once fractures, dislocations and subluxations are ruled out. Because the symptoms of the soft tissue injury are not always confined to the neck, the term WAD was suggested by the Quebec Task Force. Symptoms of WAD include, but are not limited to neck pain, spine stiffness, headache, shoulder and back pain, numbness, dizziness, sleeping difficulty, fatigue, and memory and cognitive deficits (23).

Recent work has implicated the zygapophyseal (facet) joints in WAD. Manchikati et al evaluated 106 patients suffering from neck pain at an interventional pain management practice using diagnostic blocks of lidocaine and bupivacaine injected into the facet joints. 70% of their patients reported a response to the lidocaine blocks. Using bupivacaine as a confirmatory block, 60% of their patients reported a response to the double block of lidocaine, followed by bupivacaine 2 to 4 weeks later. From this data they reported a prevalence rate of facet joint pain in chronic neck pain of 60% (24). Chen et al used nerve conduction studies in goats to evaluate the existence of pain receptors in the facet joint. They reported that there are indeed receptors in the facet joint capsule which serve pain and proprioceptive sensory functions (25). While the facet joints, which are spanned by the capsular ligaments are the only ligaments that have been directly linked to the pain of whiplash, many ligaments are stretched with WAD and may also play a significant role in the sequelae that patients report.

Facet joint dislocation represents another form of cervical ligament injury. In a retrospective analysis of 30 cervical spine injury subjects with bilateral facet dislocation Carrino et al found on MRI, ALL disruption in 27% of the cases, disc herniation or disruption in 90%, and PLL disruption in 40% (26). They grouped LF, ISL and SSL together as the posterior column complex and found disruption in this area in 97% of cases. A subset of facet joint dislocation is termed “interlocking”. Interlocking occurs when the articular process of the upper vertebrae slips forward, over the articular process of the inferior vertebrae. According to Braakman et. al, the CL must rupture (i.e fail) for this type of injury to occur (27).

Statement of purpose, specific hypothesis, and specific aims of thesis:

While previous studies have investigated the mechanical properties of human cervical spine ligaments, primarily at slow elongation rates, no study has comprehensively investigated all ligaments at each spinal level at fast elongation rates. The purpose of the present study was to determine the tensile mechanical properties, at a fast elongation rate, of intact human cervical ALL, intervertebral disc, PLL, CL, LF, and ISL+SSL tested to failure. Understanding the high-speed ligament mechanical properties may help facilitate diagnosis, treatment and outcomes of injury.

Materials and Methods

Overview

First, human cervical spine ligaments were tested in tension to failure at a fast elongation rate and non-linear force-elongation curves were generated. Peak force, peak elongation, peak energy, and stiffness were calculated. Subsequently, a simple mathematical model was developed to determine the quasi-static physiological ligament elongation.

Definitions:

Below are definitions of terms used throughout this thesis.

- **Stiffness** is the resistance offered by a structure when it is subjected to external loads (28). Since stiffness is a measure of force over distance, it is reported in units of Newton/meters (N/m).
- **Deformation energy** is the amount of work done by the elongating load (28) and is reported in units of joules (J).
- **Ligament failure** occurs if the applied force exceeds the limits of what the ligament can withstand. Ligament failure is the stretching of the ligament until it ruptures, either mid-substance, defined as a tear in the ligament fibers, or as an avulsion from the surrounding bone, such that the bone-ligament-bone complex is no longer continuous (28)
- **Viscoelasticity** is the time dependent property of a material (28). Most organic materials exhibit viscoelastic properties. Examples of viscoelastic properties are the phenomena known as creep and relaxation. Ligament mechanical property data are needed at fast elongation rates because ligaments are viscoelastic substances. They display different

properties when subjected to fast elongation versus when they undergo slow elongation, i.e changes in stiffness and energy absorption.

Mathematical Model for Average Physiological Ligament Elongations

To enable comparison of the present peak elongation to physiological elongation, a simple mathematical model was constructed to determine the physiological ligament elongation. These data were necessary to determine if the peak ligament elongation during high-speed mechanical testing exceeded the physiological limit.

Previously reported in vitro quantitative anatomy of cervical spine ligaments and vertebrae (29, 30) average in vivo normal intervertebral centers of rotation (CoRs) (31), and average in vitro physiological intervertebral rotations (32) were used to determine the physiological ligament elongation. Panjabi et al. (30) reported the lengths and coordinates of the origins and insertions of the of cervical spine ligaments. For each FSU in neutral posture, the average ligament origin and insertion coordinates were identified in the anatomical coordinate system. Ito et al. (32) defined the physiologic flexion and extension rotations of each spinal level using flexibility testing.

The upper vertebra was rotated about the CoR, as defined by Dvorak et al. (31), in flexion and extension to peak physiological rotation (32). Physiologic ligament elongations were calculated as differences in ligament lengths at maximum flexion for PLL, CL, LF, and ISL+SSL and at maximum extension for ALL and MTD, relative to neutral posture lengths.

Data for each ligament was averaged across all levels. The physiological ligament elongation range was defined as the average physiological elongation +/- 2 SD.

Specimen Preparation

Six intact human cervical spine specimens (C2 to T1) were carefully dissected of all non-osteoligamentous soft tissues. The average age of the specimens was 80.6 years (range: 71 to 92 years). The specimens had no history of cervical spine injury or disease that could have affected the osteoligamentous structures. Specimens were divided into two equal groups: the first group was dissected into C2-C3, C4-C5, and C6-C7 functional spinal units (FSUs) while the second group was dissected into C3-C4, C5-C6, and C7-T1 FSUs. Each FSU was frozen and then sectioned at the pedicles using an electric saw. The specimens were frozen in small blocks of ice to minimize trauma to the soft tissue during the cutting process. Anterior elements were sectioned coronally into thirds to create ALL, middle-third disc (MTD), and PLL bone-ligament bone preparations, while posterior elements were appropriately sectioned to create CL, LF, and ISL+SSL preparations (**Figure 1**). While the anterior and posterior thirds of the intervertebral disc were contained on the ALL and PLL specimens respectively, the disc material was transected to ensure that only ALL or PLL were intact to resist the axial load of testing. Left and right CLs from each intervertebral level were prepared separately. Using manual dissection all the soft tissue, except for the ligament to be tested, was removed from the bone-ligament-bone specimens and any tissue that could not be removed was transected to ensure the axial load was applied only to the ligament being tested.

Each preparation was then mounted for mechanical testing (**Figure 2**). To ensure rigid anchoring of bone within quick setting bondo mounts (Evercoat Z-Grip, Fibre Glass-Evercoat, Cincinnati, OH), two perpendicular thru-holes were drilled into each bone in which 19 gauge needles were inserted. Each mount contained an anchoring screw for subsequent attachment to the experimental apparatus. To increase the fixation of ALLs and PLLs to the bone and ensure

mid-substance tears, plastic plates were glued atop the ligament attachments and rigidly secured with machine screws. During preliminary trials ALL and PLL were found to avulse from the bone and thus required greater fixation. Mid-substance tears were achieved in the other ligaments without the plates. In total, 97 bone-ligament-bone specimens were prepared **(Table 1)**.

FSU

DASHED LINES REPRESENT PLANES OF SECTIONING

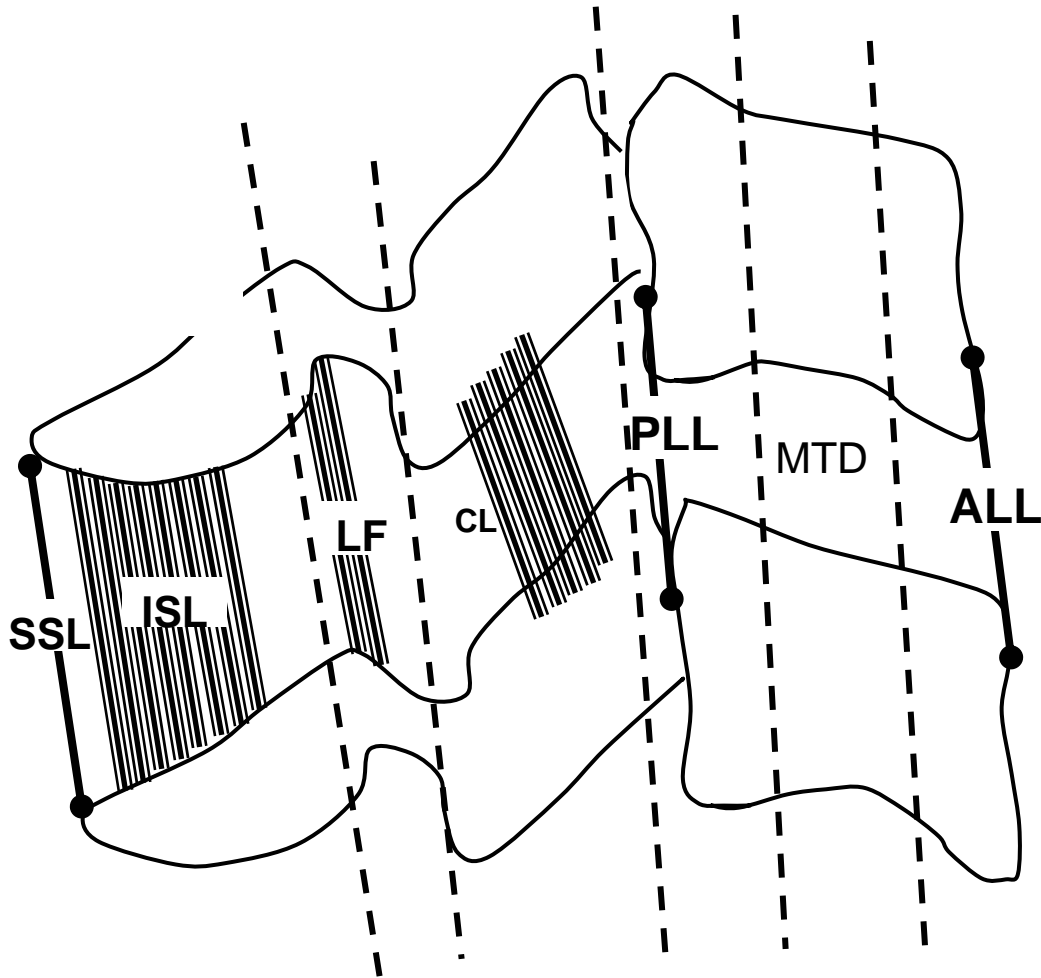


Figure 1: Sagittal view of Functional Spinal Unit (FSU). Dashed lines represent planes of sectioning. From anterior to posterior, the ligaments represented are: Anterior Longitudinal Ligament (ALL), Middle Third Disc (MTD), Posterior Longitudinal Ligament (PLL), Capsular Ligament (CL), Ligamentum Flavum (LF), Interspinous ligament (ISL), Supraspinous Ligament (SSL).

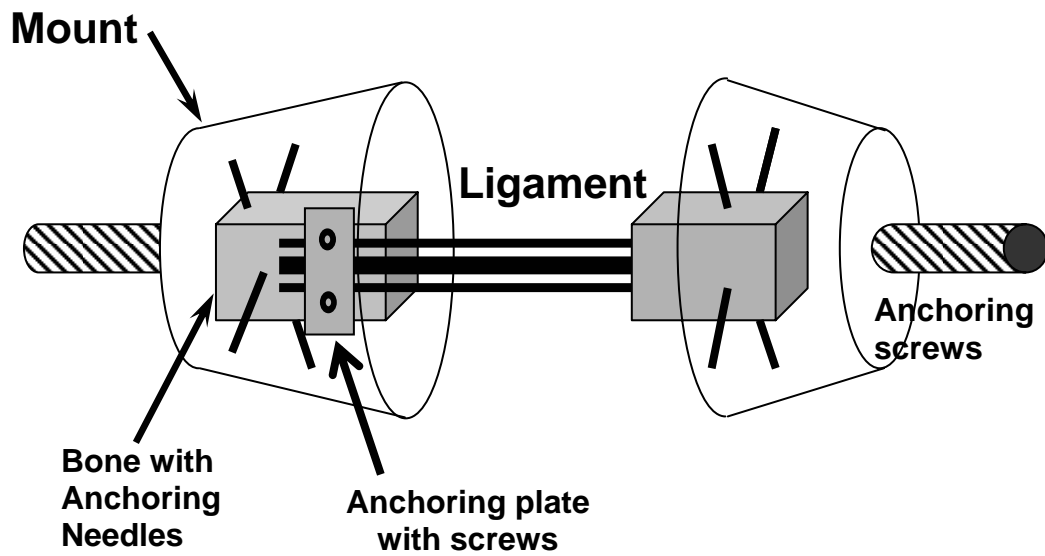


Figure 2. Schematic of bone-ligament-bone preparation. Anchoring plates ensured mid-substance tears during elongation.

Table 1. *Sample sizes for bone-ligament-bone preparations.* Cervical ligaments included: anterior longitudinal ligament (ALL), middle-third disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL).^a

	ALL	MTD	PLL	CL	LF	ISL+SSL
C2-C3	2	3	3	3	2	1
C3-C4	3	1	2	6	3	2
C4-C5	2	2	3	5	2	1
C5-C6	3	2	3	6	2	1
C6-C7	1	1	3	6	3	2
C7-T1	3	2	2	6	3	2
Totals	14	11	16	32	15	9

^a Difficulties in specimen preparation and mounting combined with a lack of ISL and SSL in some spines resulted in decreased sample sizes for some ligament

Experimental Apparatus

A custom apparatus was constructed to generate high speed elongation of the bone-ligament preparations (**Figure 3**) (33). The apparatus consisted of a pneumatic cylinder (model 1.5 x 5 Allenair, Minneola, NY) supplied with compressed air via an air tank. Air flow from the tank to the pneumatic cylinder was controlled by a solenoid valve. A controlled gap in the system permitted the pneumatic piston to achieve sufficiently high speed prior to the onset of ligament elongation. Force was measured with a uni-axial load cell (667 N capacity, model LCCA-150, Omega, Stamford, CT). Elongation was measured using a Hall effect sensor (A3506LU, Allegro Microsystems, Worcester, Mass.) positioned between two magnets (13 x 13 x 5 mm, part no. PR28ES4187B, Dexter Magnetic, Billerica Mass.). The accuracy of the Hall effect sensor was 0.025 mm (34). Prior to testing, the ligament was preloaded to 5 N tension, and this was defined as zero elongation. The force and elongation data were sampled at 6.3 kHz up to complete ligament rupture. The average (SD) peak ligament elongation rate was 723 (106) mm/s. This elongation rate was selected to simulate high energy trauma.

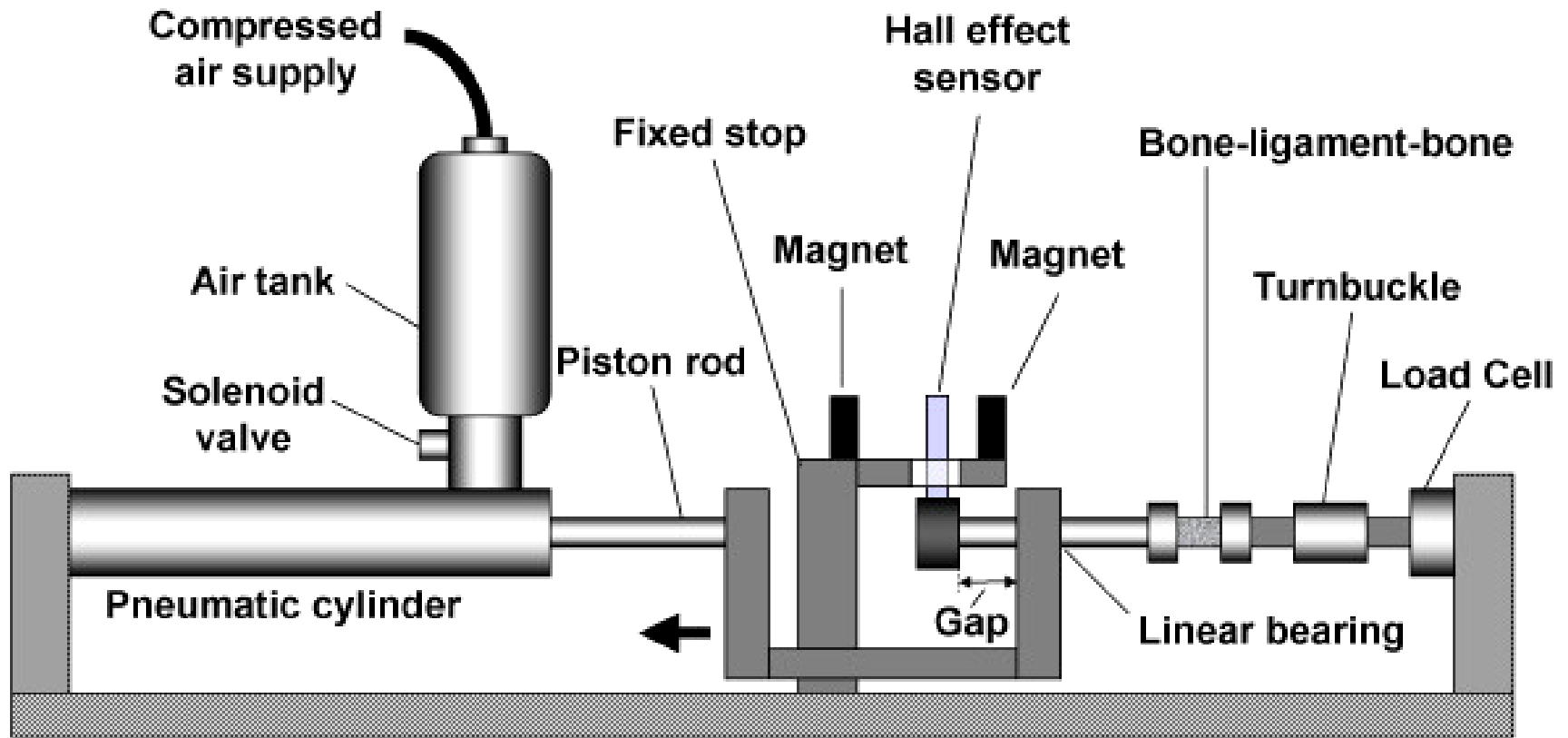


Figure 3. Schematic of the experimental apparatus. Air flow was controlled via a solenoid valve and caused movement of the piston rod, and therefore ligament elongation. Force was measured by a load cell and elongation by a Hall effect sensor positioned between two magnets.

Data Analyses

No filtering of data was performed. Peak force was defined as the maximum force attained, while peak elongation was the elongation at the peak force (**Figure 4**). Peak energy was calculated by integrating the force between zero and peak elongation. To obtain ligament stiffness, each force-elongation curve was fitted to a second order polynomial and its derivative evaluated at 25, 50 and 75% of peak force. Average (SD) r^2 was 0.97 (0.04).

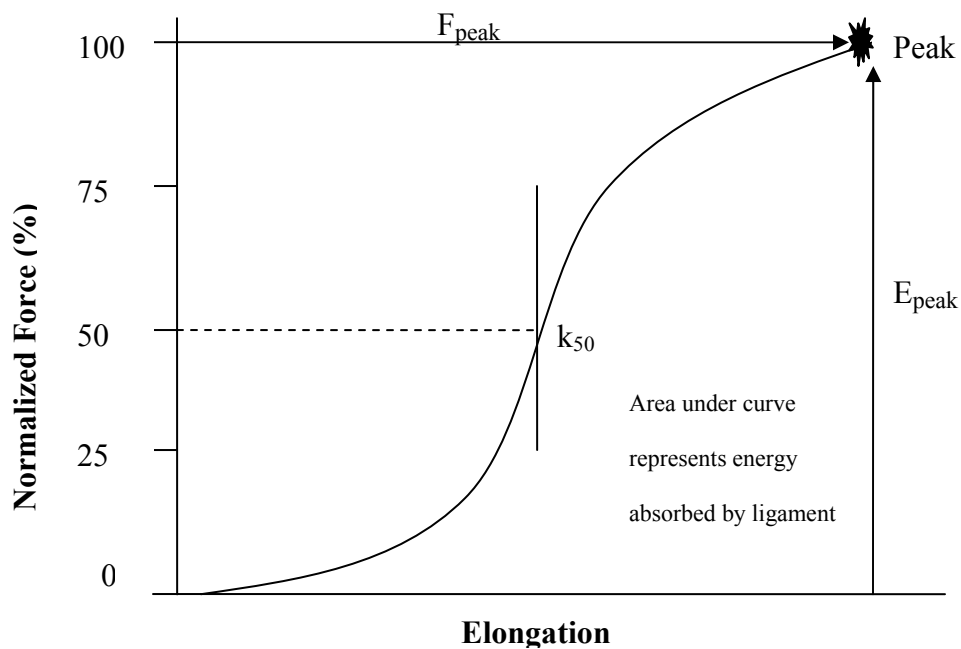


Figure 4: Schematic of ligament peak force-peak elongation curves.

Statistics.

Data from each spinal level, C2-C3 to C7-T1, were combined for each ligament (**Table 1**). Single-factor, non-repeated measures ANOVA ($P < 0.05$) and pair-wise Bonferroni post hoc tests were used to determine differences among ligaments in peak force, peak elongation, peak

energy, and stiffness at 25, 50, and 75% of peak force. Adjusted P-values were computed based upon the number of post-hoc tests performed.

Results

The average physiological ligament elongations calculated using the mathematical model were largest at ISL+SSL, 3.6 mm, and LF, 2.3 mm (**Table 2**). MTD had the smallest physiological elongation of 0.3 mm.

Table 2. Physiological ligament elongations (mm). Average physiological elongation and physiological elongation range (average ± 2 SD) obtained using a mathematical model. The ligaments included the anterior longitudinal ligament (ALL), middle-third disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL). The rotation direction used to obtain the physiological elongation is given for each ligament. See Methods for further details.

	Physiological Elongation	Rotation Direction
ALL	1.2 (0.6, 1.8)	Extension
MTD	0.3 (0.1, 0.5)	Extension
PLL	0.8 (0.2, 1.4)	Flexion
CL	1.2 (0.4, 2.0)	Flexion
LF	2.3 (0.9, 3.7)	Flexion
ISL+SSL	3.6 (1.2, 6.0)	Flexion

The dynamic force vs. elongation curves up to peak force displayed varying trends among ligaments (**Figures 5A to 5F**). For comparison, the average physiological ligament elongation with ± 2 SD range is shown on each graph as a vertical dotted line representing the average and the standard deviation range represented by the grey shading. The peak ligament

elongation was generally greater than the physiological range, with the exception of LF and ISL+SSL, which had peak elongation in excess of the average physiological elongation.

The average peak force, elongation, and energy varied by ligament (**Table 3A**). The highest peak force of 244.4 N was attained in LF, followed by 220.0 N in CL. The peak forces in LF and CL were significantly greater than those in ALL, MTD, and ISL+SSL. The highest peak elongation of 5.9 mm was observed in ISL+SSL, followed by 4.9 mm in CL and 4.2 mm in LF and PLL. MTD had significantly less peak elongation than all ligaments. The highest peak energy of 0.57 J was attained in CL, followed by PLL, LF, and ISL+SSL, with values ranging from 0.33 to 0.36 J. The peak energy in CL was significantly greater than in ALL and MTD, while the peak energy in PLL significantly exceeded that in MTD.

LF was generally the stiffest, while ISL+SSL was generally the least stiff (**Table 3B**). At 25% of peak force, the highest stiffness of 72.7 N/mm was attained in LF. CL stiffness of 69.4 N/mm significantly exceeded ISL+SSL stiffness. At 50% of peak force, LF stiffness of 98.8 N/mm was significantly greater than in ISL+SSL. At 75% of peak force, LF stiffness of 118.4 N/mm was significantly greater than in ALL, PLL, CL, and ISL+SSL. MTD stiffness of 96.0 N/mm significantly exceeded ISL+SSL stiffness.

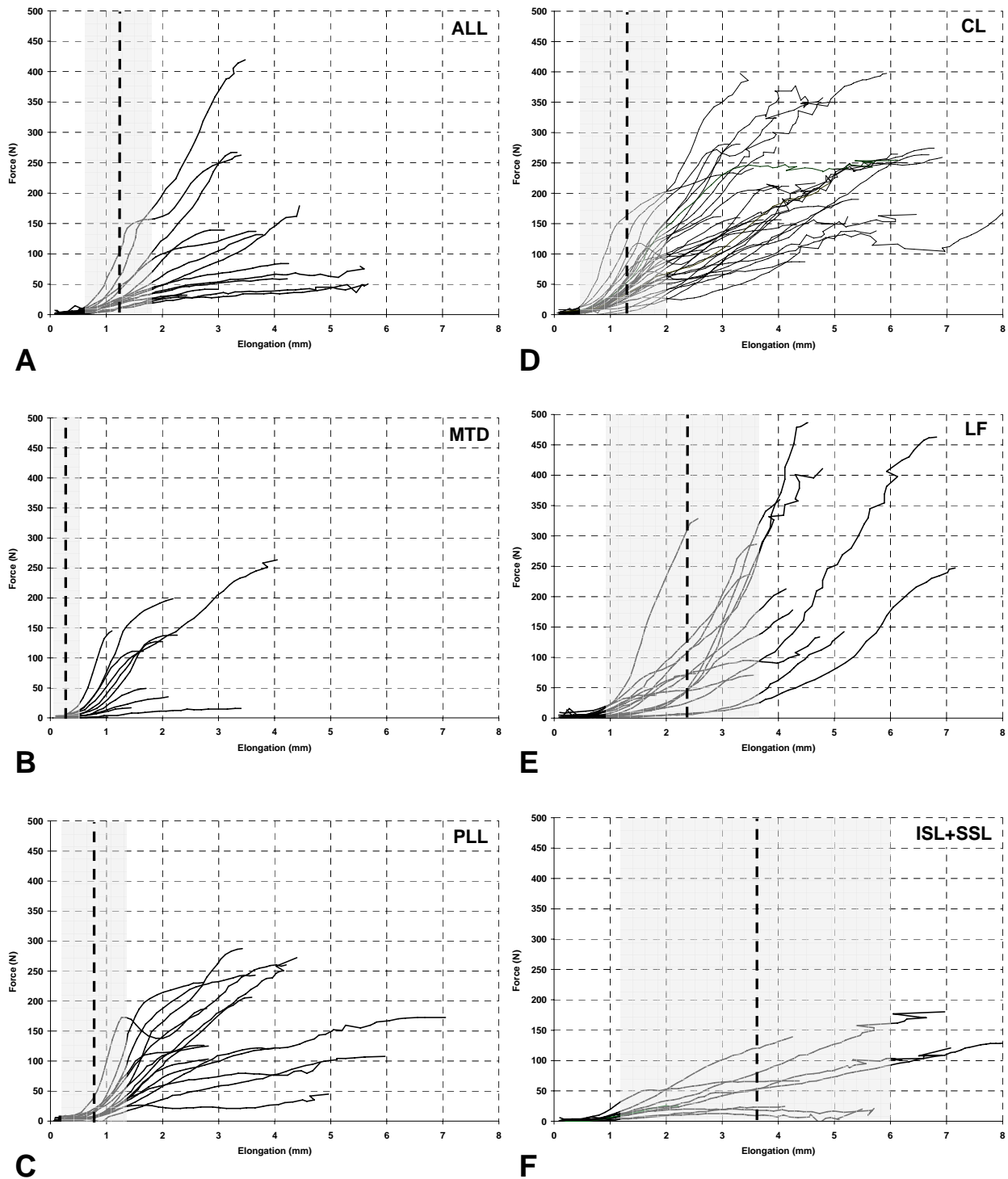


Figure 5. Ligament force vs. elongation including the average physiological elongation, shown by a vertical dashed line, and the physiological ligament elongation range (average ± 2 SD), indicated by grey shading. The ligaments included A) anterior longitudinal ligament (ALL), B) middle-third disc (MTD), C) posterior longitudinal ligament (PLL), D) capsular ligament (CL), E) ligamentum flavum (LF), and F) interspinous and supraspinous ligaments

Table 3. Mechanical properties of human cervical ligaments at 723 mm/s. Average (SD) **A)** peak force (N), peak elongation (mm), peak energy (J), and **B)** stiffness (N/mm) at 25, 50, and 75% of peak force. The ligaments included anterior longitudinal ligament (ALL), middle-third disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL). Significant differences ($P < 0.05$) among ligaments are indicated in the column *Significant*. A blank entry indicates that no significant difference was observed.

A) Peak Force, Elongation, and Energy

	Peak Force	Significant	Peak Elongation	Significant	Peak Energy	Significant
ALL	137.9 (111.5)	CL,LF	4.0 (1.0)	MTD	0.25 (0.15)	CL
MTD	115.6 (79.9)	CL,LF	2.1 (0.9)	Every ligament	0.12 (0.15)	PLL,CL
PLL	163.7 (80.2)		4.2 (1.5)	MTD	0.33 (0.18)	MTD
CL	220.0 (83.7)	ALL,MTD,ISL+SSL	4.9 (1.4)	MTD	0.57 (0.30)	ALL,MTD
LF	244.4 (143.0)	ALL,MTD	4.2 (1.5)	MTD	0.36 (0.25)	
ISL+SSL	85.5 (67.6)	CL,LF	5.9 (2.9)	MTD	0.33 (0.39)	

B) Stiffness at 25%, 50%, and 75% of Peak Force

	Stiffness at 25%	Significant	Stiffness at 50%	Significant	Stiffness at 75%	Significant
ALL	46.9 (38.8)		49.9 (45.2)		50.4 (53.4)	LF
MTD	61.3 (38.8)		81.4 (61.0)		96.0 (79.1)	ISL+SSL
PLL	71.6 (49.7)		63.6 (41.4)		53.0 (33.5)	LF
CL	69.4 (34.3)	ISL+SSL	65.1 (29.2)		57.9 (28.9)	LF
LF	72.7 (44.0)		98.8 (65.4)	ISL+SSL	118.4 (82.9)	ALL,PLL,CL,ISL+SSL
ISL+SSL	22.1 (12.7)	CL	21.3 (11.7)	LF	19.9 (11.7)	MTD, LF

Discussion

The present study determined the tensile mechanical properties, at a fast elongation rate, of the anterior longitudinal ligament (ALL), middle-third disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SLL). The ligaments were elongated to complete rupture and peak force, peak elongation, peak energy, and stiffness were determined from the force vs. elongation curves. The experimental design of the present study has several important advantages over those of previous studies (15-17, 20). The present experimental methodology, using bone-ligament-bone preparations, optimized usage of scarce human cadaveric specimens. Rigid fixation of the ligament attachments within the mounts ensured mid-substance tears during elongation, avoiding ligament avulsion from the bone.

The limitations of the present study must be considered before interpreting the results. Since availability of young human cadaveric material is limited, the average age of the present specimens was 80.6 years. Although only six cervical spines were used, bone-ligament-bone preparations from all spinal levels were combined to maximize the sample size for each ligament (**Table 1**). The maximum number of samples per spinal level was six for CL and was three for all other ligaments, however difficulties in specimen preparation and mounting and lack of ISL+SSL in some spines (3) resulted in decreased sample sizes for some ligaments, as seen in Table 1. In addition, every attempt was made to elongate each ligament along the direction of its fibers. However, this was difficult for MTD, CL and ISL+SSL due to anatomical constraints.

The present results may be compared with previously reported mechanical properties of human cervical spine ligaments obtained at *fast* elongation rates. We could find only two previous studies (**Table 4A**). Yoganandan et al (20) tested only ALL and LF at elongation rates of 250 and 2500 mm/s. Shim et al (21) tested alar and transverse ligaments and ALL, PLL, CL, LF, and ISL at elongation rates between 10,000 and 12,000 mm/s, however peak elongation and peak force data were presented only for CL, and alar and transverse ligaments. At both rates reported by Yoganandan et al (20), the peak force, elongation, and energy exceeded the corresponding present data (**Table 3A**), with the exception of the peak force in LF at 250 mm/s. Comparisons between our CL data (**Table 3A**) and the average data from Shim et al (21) indicated that peak force increased, while peak elongation decreased with increased elongation rate.

The present results may be compared with previously reported results obtained at *slow* elongation rates of less than 25 mm/s. Numerous studies have reported the peak force, elongation, and energy (**Table 4B**) using various experimental methodologies (14-18, 20, 35-37). Comparisons between the present fast rate results to the slow rate results from the literature indicate that the average peak force increased while the peak elongation and energy absorbing capacity decreased with increased elongation rate (**Tables 3A and 4B**). However, exceptions included the peak force of MTD and the peak energy of PLL and ISL+SSL. Panjabi et al (19) observed similar peak elongation and energy vs. rate relationships, while testing cervical alar and transverse ligaments at slow (0.1 mm/s) and fast (920 mm/s) rates. Additionally, these researchers observed increased ligament stiffness due to increased elongation rate. Shim et al (21) documented increased ligament strength and decreased peak elongation with increased

elongation rate. Thus, these cumulative findings suggest that during high speed elongation, cervical ligaments may fail at a higher peak force and smaller peak elongation and may be stiffer and absorb less energy, as compared to the slow elongation speed.

A few LF and ISL+SSL specimens of the present study failed within the physiological ligament elongation range (**Figures 5E and 5F**). If these failures were to occur in vivo, we believe that the cervical spine would not become unstable. The average peak elongations of LF and ISL+SSL (**Table 3A**) were between 1.6 and 1.8 times the corresponding average physiological elongations (**Table 2**). In contrast, the peak elongations of ALL, MTD, PLL, and CL were between 3.3 and 7.0 times the average physiological elongations. Thus, these ligaments which lie close to the intervertebral centers of rotation have a greater factor of safety and play a critical role in stabilizing the spinal column and protecting the neural tissues from injury. These findings are supported by the results of a classic biomechanical study, which investigated increases in cervical spine mobility due to transection of ligamentous components (38). *Large* increases in mobility were observed due to extension loading applied following sequential transection of components from anterior to posterior, indicating the importance of ALL and MTD in stabilizing the spine. In contrast, relatively *small* increases in mobility were observed due to flexion loading applied following sequential transection of components from posterior to anterior, indicating a less significant role for LF and ISL+SSL. LF and ISL+SSL contain a greater proportion of elastin fibers to enable large elongations during neck flexion, as compared to those in the vicinity of the intervertebral centers of rotation. The primary role of LF may be to protect the spinal cord from impingement during neck motion, while its contribution to spinal stability may be secondary.

Table 4. Previously reported mechanical properties of human cervical ligaments. Average (range) peak force (N), elongation (mm), and energy (J) at **A**) fast (≥ 250 mm/s) and **B**) slow (≤ 25 mm/s) elongation rates. The ligaments included anterior longitudinal ligament (ALL), intervertebral disc (IVD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL). The averages and ranges were calculated using average data from: a) Yoganandan et al (20); b) Myklebust et al (16); c) Pintar et al (15); d) Przybylski et al (18); e) Siegmund et al (35); f) Winkelstein et al (36); g) Chazal et al (14); h) Yoganandan et al (37); i) Yoganandan et al (17); j) Shim et al (21).

A) Fast elongation rates (≥ 250 mm/s)

		Peak Force	Peak Elongation	Peak Energy
ALL	250 mm/s	166.4 ^a	6.4 ^a	0.65 ^a
	2500 mm/s	349.5 ^a	6.3 ^a	1.23 ^a
LF	250 mm/s	181.5 ^a	6.3 ^a	0.63 ^a
	2500 mm/s	335.1 ^a	8.0 ^a	1.32 ^a
CL	10,000 to 12,000 mm/s	259.7 (140.5, 526.9) ^j	2.6 (1.8, 3.6) ^j	

B) Slow elongation rates (≤ 25 mm/s)

	Peak Force	Peak Elongation	Peak Energy
ALL	117.3 (8.0, 140.0) ^{b,c,d,f,g}	5.8 (2.4, 8.1) ^{b,c,f,g,i}	0.43 (0.14, 0.68) ^{d,f,i}
IVD	569.0 ^h	11.1 ^h	4.00 ^h
PLL	114.7 (82.0, 186.0) ^{b,c,d,g}	4.9 (2.4, 6.3) ^{b,c,g,i}	0.22 (0.13, 0.31) ^{d,i}
CL	130.2 (88.4, 224.0) ^{a,b,c,e}	8.3 (5.8, 10.8) ^{a,b,c,e,i}	1.50 ⁱ
LF	136.2 (81.0, 215.0) ^{b,c,f}	7.8 (5.7, 8.9) ^{b,c,f,i}	0.51 (0.35, 0.70) ^{f,i}
ISL+SSL	33.1 (32.0, 34.2) ^{b,c}	7.0 (6.5, 7.3) ^{b,c,i}	0.16 ⁱ

The present study investigated the tensile mechanical properties, at a fast elongation rate, of human cervical ligaments including ALL, MTD, PLL, CL, LF, and ISL+SSL. These data may be used by clinicians to determine the strongest and stiffest cervical ligaments and by engineers for improving the biofidelity of mathematical models of the cervical spine. Comparison of the present results with previously reported data indicated that high speed elongation may cause cervical ligaments to fail at a higher peak force and smaller peak elongation and may be stiffer and absorb less energy, as compared to a slow elongation rate. These comparisons may be useful to clinicians for diagnosing cervical ligament injuries based upon the specific form of trauma.

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Abbreviation for Binding:

Properties of Cervical Spine Ligaments