3-30-2007

Dynamic Mechanical Properties of Human Cervical Spine Ligaments Following Whiplash

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

By

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Class of 2006

DYNAMIC MECHANICAL PROPERTIES OF HUMAN CERVICAL SPINE LIGAMENTS FOLLOWING WHIPLASH.

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ABSTRACT

The purpose of this study is to quantify the dynamic mechanical properties of human cervical ligaments following whiplash. Cervical ligaments function to provide spinal stability, propioception, and protection during traumatic events to the spine. The function of cervical ligaments is largely dependant on their dynamic biomechanical properties, which include force and energy resistance, elongation capability, and stiffness. Whiplash has been shown to injure human cervical spine ligaments, and ligamental injury has been shown to alter their dynamic properties, with potential clinical consequences such as joint degeneration and pain. In this study we quantified the dynamic properties of human lower cervical ligaments following whiplash and compared their properties to those of intact ligaments. Whiplash simulation was performed using biofidelic whole cervical spine with muscle force replication (WCS-MFR) models. Next, ligaments were elongated to failure at a fast elongation rate and peak force, peak elongation, peak energy, and stiffness values were calculated from non-linear force-elongation curves. Peak force was highest in the ligamentum flavum (LF) and lowest in the intraspinous and supraspinous ligaments (ISL+SSL). Elongation was smallest in middle-third disc (MTD) and greatest in ISL+SSL. Highest peak energy was found in capsular ligament (CL) and lowest in MTD. LF was the stiffest ligament and ISL+SSL least stiff. These findings were similar to those found in intact ligaments. When directly comparing ligaments following whiplash to intact ligaments in a prior study it was found that the anterior longitudinal ligament (ALL) and CL had altered dynamic properties that were statistically significant, providing direct evidence that whiplash may alter the dynamic properties of cervical ligaments. These findings may contribute to the understanding of whiplash injuries and the development of mathematical models simulating spinal injury.

Acknowledgements

I would like to thank all of the current and former members of the Yale biomechanics research laboratory, who contribute to the team-based approach to research facilitated by Dr. Manohar Panjabi. Contributors to the work included in this paper include Yasuhiro Tominaga, Anthony B. Ndu (YMS- IV), Marcus P. Coe (YMS IV), Paul C. Ivancic, Shigeki Ito, Wolfgang Rubin, Erik J. Carlson (YMS II), and Manohar M. Panjabi. I am especially indebted to Paul Ivancic and Dr. Panjabi for mentorship and editing of the corresponding manuscript prepared for publication, to Dr. Jonathan Grauer, our departmental thesis advisor; and to Gwen Henderson and Edward Teng (YMS IV), my lab mates and partners in the daily pursuit to learn and contribute something.

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Introduction

Although the term whiplash was coined by Crowe in 1928(1), the definition of what constitutes whiplash injury remains controversial.(2) The classic definition is that of a hyperextension injury of the soft tissues of the neck following a rear-impact injury, though it is clear that similar injuries may occur following front or side impacts.(1-3) Other terms used to describe the phenomenon include neck sprain, neck strain, soft-tissue injury, and acceleration-deceleration injury. A landmark study on whiplash by the Quebec task force(4) suggested the term whiplash associated disorders (WAD) to encompass the common associated symptoms including neck pain, headache, visual symptoms, and dizziness. In this study the task force reviewed over 10,000 publications on whiplash and found only 346 "worthy of consumption"(5). In the wake of confusion caused by the new medical phenomenon of whiplash, Dr. Crowe himself expressed regret at having invented the phrase 34 years earlier, citing misapplication of the term by physicians, patients, and lawyers.(6, 7) What is clear is that whiplash is a phenomenon with a profound societal impact that shows no signs of waning. Whiplash injuries have an estimated annual incidence of one million in the United States,(1, 8) resulting in societal costs up to $29 billion after factoring in diagnosis, treatment, insurance, and litigation.(9-12) Many of these costs are related to the long-term morbidity associated with whiplash. Most estimates of chronic disability related to whiplash fall within the range of 14 to 18%, with some studies reporting up to 52% of patients continue to suffer whiplash-associated symptoms after one year.(1, 13-18) Between 5 and 8% of whiplash patients develop chronic symptoms severe enough to diminish their work capacity.(4, 8, 19, 20) Costs and morbidity related to whiplash are expected to continue rising in Western nations due to increases in traffic, litigation, and mandatory seat-belt use; though few would argue that the life-saving benefits of seatbelt use outweigh associated increases in cervical spine injuries.(17, 18) Despite substantial resources and research

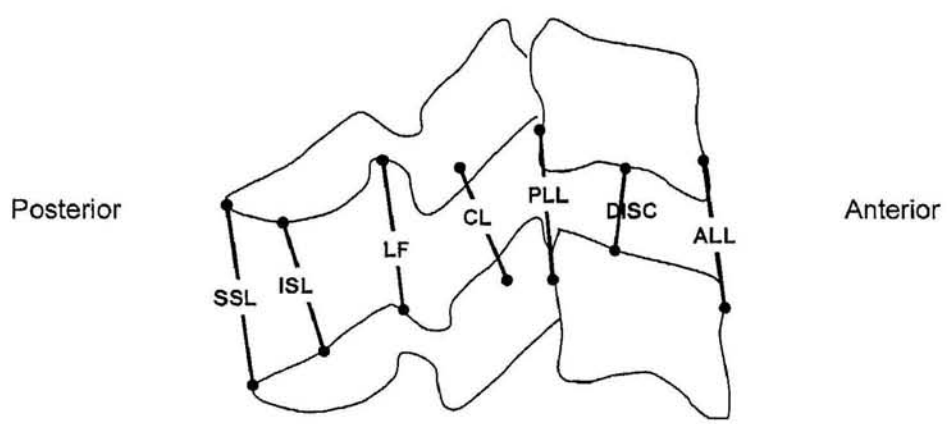
devoted to elucidating whiplash associated disorders (WAD), mechanisms of morbidity remain largely unknown and contested(21); with theories ranging from psychosocial factors to anatomical injury as root causes of whiplash symptoms.(22-30) Present knowledge remains incomplete regarding specific anatomical components injured during whiplash and causes of resulting chronic symptoms.(17, 30-33)

Clinical evidence linking whiplash forces and injury to chronic neck pain was first provided by Barnsley et al (1995) in a study investigating the role of zygapohyseal (facet) joints as a source of pain. (34) Zygapohyseal joints had been suspected of playing a role in whiplash pain due to shearing forces from intervertebral translation during whiplash that is resisted by tension within the zygapophyseal joint capsule and the anterior annulus of the intervertebral disc.(2) Barnsley et al were able to relieve whiplash-associated neck pain with anesthetic blocks of the cervical facet joints providing clinical evidence implicating the facet joints and associated cervical ligaments in whiplash pain. They found cervical facet joint pain to be the primary cause of chronic whiplash pain in 54% of patients suffering long-term neck pain. Lord et al (1996) subsequently used percutaneous radio-frequency neurotomy to denature facet joint nerves and relieve whiplash associated neck pain.(35) Clinical, MRI, and autopsy studies have further identified cervical ligament, facet joint, and intervertebral disc injuries in whiplash victims.(2, 36-46) Biomechanical studies(30, 46-51) of whiplash injury mechanisms have demonstrated that the spine assumes an s-shaped curve with upper cervical spine (UCS) flexion and lower cervical spine (LCS) hyperextension at approximately 70 to 110 milliseconds after rear-impact, with hyperextension peaking in both UCS and LCS shortly thereafter. It is thought that during the initial phase of s-shaped curvature anterior elements of the spine including the anterior longitudinal ligament (ALL) stretch due to the hyperextension moment, while posterior elements

experience shear, tension, flexion, extension, and compressive forces leading to facet joint compression, capsular ligament (CL) stretching, and intervertebral disc disruption.(49, 52) Biomechanical experiments have since demonstrated whiplash-induced injuries to ALL, CL, and intervertebral discs.(30-32, 46, 50, 53-58)

The cervical spinal ligaments (**Figure 1**) serve several functions including spinal stability, mechanoreceptor functions, and energy absorption during trauma.(59-62)

Figure 1. ***Schematic of lateral view of two cervical vertebrae and ligaments.*** Ligaments included: anterior longitudinal ligament (ALL), middle-third disc (disc), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligaments (ISL+SSL).



Despite their important functions, there have been relatively few studies describing basic anatomy and function of the cervical ligaments.(63) The anatomy and function of the seven cervical ligaments which are the focus of the present study were described in two landmark papers by Johnson et al (1975) and Panjabi et al (1980).(64, 65) Generally,

ligaments are uniaxial structures comprised of varying combinations of collagen and elastin for the purpose of resisting tension, while intervertebral discs contain large amounts of proteoglycans in the nucleus capable of withstanding compressive loads.(59) The disc nucleus is surrounded by a collagenous annulus fibrosus that resists tension, shear, and torsion.(59, 64) The anterior longitudinal ligament (ALL) is a thin, translucent structure that extends without interruption over the anterior vertebral bodies. It is thought to assist in limiting spinal extension, but its small thickness in the lower cervical region suggests it plays a secondary role.(59, 64, 65) The intervertebral disc, which we call middle-third disc (MTD) in this study due to the region of disc studied, is a dense fibrous structure in the intervertebral space. It is composed of three components: annulus fibrosus, nucleus pulposus, and cartilaginous end plates. The nucleus pulposus is capable of withstanding great compressive forces and injuries to the intervertebral disc usually involve tension, torsion, and shear forces to the annulus fibrosus.(64) The posterior longitudinal ligament (PLL) is a thick band of dense tissue that runs continuously over the posterior side of the vertebral bodies. It appears to limit flexion, intervertebral distraction, and possibly protects the spinal cord from posterior protrusion of disc material.(65) The capsular ligaments (CL) are thick, dense, and fibrous structures located between the facet joints at a 90 degree angle to the plane of the joints. They function to stabilize facet joint motion.(65) The ligamentum flavum (LF) runs between the laminae and has one of the highest elastin contents of any tissue in the body. Its function is unclear, but it may assist in spinal extension.(64, 65) The intraspinous and supraspinous ligaments (ISL+SSL) are smaller posterior spinal ligaments that may contribute to limitation of spinal flexion.(59, 65) Size and biomechanical observations led Johnson et al to conclude that the annulus fibrosus, PLL, and CL were the primary ligamental stabilizers of the lower cervical spine, while the other ligaments play more specialized or secondary roles.(65)

Previous biomechanical studies have documented dynamic ligament strains above physiological limits and mechanical spinal instability due to simulated rear impacts of a whole cervical spine model with muscle force replication and surrogate head.(32, 53, 55, 56) Dynamic strains in anterior longitudinal ligaments(53) and annular fibers(55) beyond physiological limits were observed at the middle and lower cervical spine. Following whiplash, the ligaments were classified as having sustained no macroscopic rupture, partial rupture with no visible damage to the underlying annulus, or complete rupture with visible anterior annulus tears. Intervertebral levels with completely ruptured ligaments sustained significantly greater dynamic intervertebral extension, peak ligament strain, and joint laxity as compared to uninjured intervertebral levels. Capsular ligaments at C5-C6 and C6-C7 were found to be at risk for subfailure injury due to excessive dynamic strains.(56) The aforementioned results were supported by Ito et al (2004) who documented increased joint laxity at the middle and lower cervical spine based upon flexibility tests performed prior to and following each impact.(32) No previous studies have documented the dynamic mechanical properties of human cervical spine ligaments following whiplash.

Given their important role in the stabilization and resistance to forces transmitted to the spinal column, there is a relative paucity of data on the biomechanical properties of intact cervical ligaments.(63) In mathematical models simulating intervertebral disc annular and nucleus degeneration such as that occurring in degenerative conditions or after trauma Yoganandan et al (2001) predicted that compared to intact discs degenerated discs had increased stiffness and transmitted less load distribution to the facet joints.(59) While disc bulge, annulus stress, and fiber strain decreased in degenerated discs, stress and strain density increased in bony cortex which the authors predicted could cause bony changes and osteophytes. This study suggests that

degenerated or damaged cervical ligaments may have dysfunctional dynamic mechanical properties leading to clinical consequences. Biomechanical evidence of macroscopic intervertebral disc damage leading to changes in dynamic properties has been documented in several studies of disc degeneration.(66, 67) Thompson et al (2004) documented the mechanical effects of concentric tears, radial tears, and rim lesions to the anterior annulus in lumbar sheep intervertebral discs and found that anterior rim lesions caused decreased ability to resist motion in the disc.(66) Other studies have shown that altered disc biomechanics were a result of disc degeneration in general rather than from the lesions induced for study, reinforcing the need to differentiate between specific lesion types in the face of disc degeneration to determine the cause of altered dynamic properties.(67-69) While whiplash causes cervical ligament injury that could theoretically alter their dynamic properties; we are unaware of any studies documenting the dynamic properties of cervical ligaments with lesions specific to whiplash.

Because common cervical injuries are associated with dynamic forces, quantifying the dynamic properties of cervical spine ligaments are necessary for understanding their physiological functioning under traumatic conditions.(70) The dynamic properties most commonly examined in prior studies have been determined from force-deformation or stress-strain curves. These curves may be used to determine the most common fundamental mechanical properties of ligaments including stiffness, elongation capabilities, and resistance to force, and energy absorbed. To date, most studies testing the dynamic properties of cervical ligaments have done so at quasistatic, or slow elongation rates that are more similar to forces encountered at sub-injury loads.(63, 70-73) There have been few studies performed on the mechanical properties of cervical ligaments at loads simulating traumatic conditions. Yoganandan et al (1989) tested the

biomechanical properties of the anterior longitudinal ligament (AL) and ligament flavum (LF) at 8.89, 25.0, 250.0, and 2,500 mm/sec and found that the mechanical parameters of force, stiffness, and peak energy at failure increased with increasing load rates.(70) Their findings demonstrated that dynamic properties vary with loading rate and further underscored the need to quantify these properties at fast rates to understand how cervical ligaments function under dynamic conditions. Elongation before failure did not vary with loading rate and ligaments failed once a critical stretching point was reached regardless of load rate, indicating that ligaments are deformation sensitive.(59) Panjabi et al (1998) tested the mechanical properties of alar and transverse cervical ligaments at slow (0.1 mm/s) and fast (920 mm/s) elongation rates; finding that strain and energy absorbed decreased while stiffness increased for both ligaments as elongation rate was increased.(74) These results illustrated that upper cervical ligament dynamic mechanical properties vary as a function of loading rate, again emphasizing the importance of understanding cervical ligament function at fast and slow loading rates. Only one study has examined the biomechanical properties of all the major lower cervical ligaments at fast loading rates. Ivancic et al (2005) tested peak force, peak elongation, peak energy absorbed, and stiffness at 722. mm/s for the anterior longitudinal ligament (AL), posterior longitudinal ligament (PL), middle-third disc (MTD), capsular ligament (CL), ligamentum flavum (LF), and intraspinous and supraspinous ligaments (ISL+SSL).(71) When comparing their results to the dynamic properties at slow rates reported in prior studies they found that during higher speed elongations cervical ligaments tend to be stiffer, absorb less energy, and fail at a higher peak force and smaller peak elongation. To our knowledge, no study has examined the biomechanical properties of cervical ligaments following whiplash injury.

Statement of Purpose and Hypothesis

The purpose of this study is to determine the biomechanical properties of human cervical ligaments following whiplash, and to determine if these properties are altered when compared to their intact counterparts(71). Ligaments tested include the anterior longitudinal ligament (ALL), middle-third disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and intraspinous and supraspinous ligaments (ISL+SSL). Due to substantial data documenting whiplash injury to cervical ligaments, particularly to ALL(52, 53), MTD(48, 55), and CL(31, 56); we hypothesize that cervical ligaments that have undergone whiplash will exhibit altered biomechanical properties compared to intact ligaments.

Methods

Overview

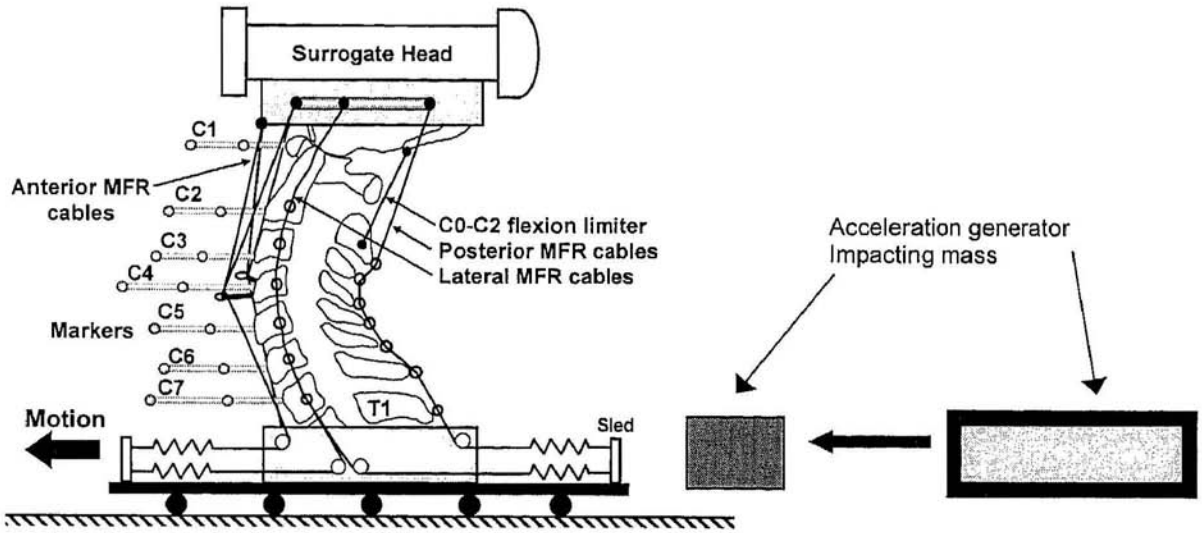
First, whiplash simulation was performed using the biofidelic whole human cervical spine model with muscle force replication and surrogate head. Next, the ligaments were elongated to failure at a fast elongation rate and non-linear force-elongation curves were recorded. Peak force, peak elongation, peak energy, and stiffness values were calculated. Peak elongation was compared with average physiological elongation computed using a simple mathematical model. Present results were compared with previously reported data of intact ligaments.(71)

Specimen Preparation for Whiplash Simulation

Six fresh-frozen human whole cervical spine specimens (occiput-T1) were dissected of all non-osteoligamentous structures and mounted in resin at the occiput and T1 in normal neutral posture. The average age of the specimens was 70.8 years (range, 52 to 84 years) and there were four male and two female donors. Apart from typical age

related changes, the specimens did not suffer from any disease that could have affected the osteoligamentous structures. A surrogate head, attached to the occipital mount, and the spine were stabilized using compressive muscle force replication (MFR) and a C0-C2 flexion limiter cable to model the effect of chin-sternum contact on cervical flexion. The model, described in detail elsewhere, has been validated against in vivo simulated whiplash data.(75) Whiplash simulation was performed using a previously developed bench-top sled apparatus (**Figure 2**)(76) at nominal T1 horizontal accelerations of 3.5, 5, 6.5 and 8 g.(77) The sled apparatus consisted of a sled with attached WCS-MFR specimen, impacting mass, power springs, and a braking system. The impacting mass was released from the power springs and struck the sled from the rear.

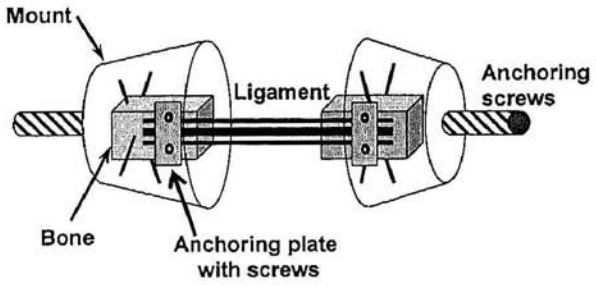
Figure 2. ***Schematic of whiplash apparatus with the WCS-MFR model attached to sled.*** Acceleration generation system consisting of power springs and a piston power the impacting mass to strike the sled from the rear at 3.5, 5, 6.5 and 8 g.



Cervical Ligament Preparation

After final trauma of 8 g, the 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 sectioned at the pedicles. Anterior elements were sectioned coronally into thirds to create anterior longitudinal ligament (ALL), middle-third intervertebral disc (MTD), and posterior longitudinal ligament (PLL) bone-ligament bone preparations. Posterior elements were sectioned to create capsular ligament (CL), ligamentum flavum (LF), and intraspinous and supraspinous ligament (ISL+SSL) preparations. Left and right CLs from each intervertebral level were prepared separately, while left and right LFs from each level were prepared as a single unit. Each preparation was then mounted for mechanical testing (**Figure 3**).

Figure 3. ***Schematic of bone-ligament-bone preparation.*** Anchoring plates ensured mid-substance tears during elongation.



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 testing apparatus. To increase fixation of ALLs and PLLs to bone, plastic plates were glued atop ligament attachments and rigidly secured with machine screws. In total, 98 bone-ligament-bone specimens were prepared (**Table 1**).

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). Last column labelled *sum total* shows total number of samples for all ligaments combined.

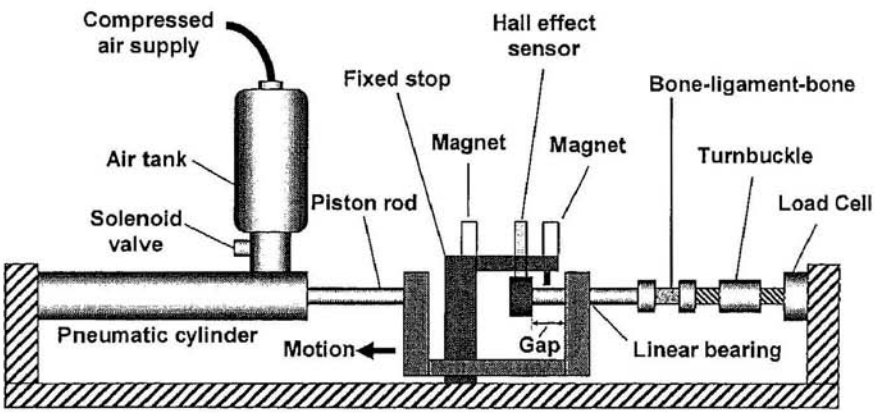
	ALL	MTD	PLL	CL	LF	ISL+SSL	Sum total
C2-C3	3	1	3	6	3	3	19
C3-C4	3	1	2	4	3	2	15
C4-C5	3	0	3	6	3	2	17
C5-C6	2	0	3	6	2	3	16
C6-C7	2	0	3	6	2	2	15
C7-T1	3	1	2	6	2	2	16
Totals	**16**	**3**	**16**	**34**	**15**	**14**	**98**

Experimental Apparatus

A custom apparatus was constructed to generate high-speed elongation of the bone-ligament preparations (**Figure 4**).(71) 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 velocity prior to the onset of ligament elongation. Force was measured with 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, MA) positioned between two magnets (13 x 13 x 5 mm, part no. PR28ES4187B, Dexter Magnetic, Billerica MA).

Figure 4. ***Schematic of experimental apparatus.*** Air flow controlled via the solenoid valve caused movement of the piston rod and therefore ligament elongation. Force was measured by the load cell and elongation by the Hall effect sensor positioned between two magnets.

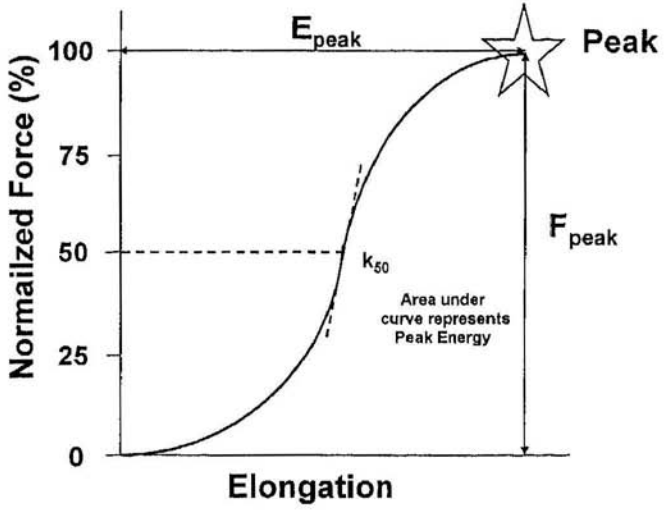


Prior to testing, the ligament was preloaded to 5 N of tension. Force and elongation data were sampled at 6.3 kHz up to complete ligament rupture. The average peak ligament elongation rate was 725 mm/s.

Data Analyses

Peak force was defined as the maximum force attained, while peak elongation was the elongation at the peak force. Peak energy was calculated by integrating the force between zero and peak elongation. To obtain ligament stiffness values, each force-elongation curve (**Figure 5**) was fitted to a second order polynomial and its derivative evaluated at 25, 50, 75% of peak force. Average (SD) r^2 was 0.97 (0.03).

Figure 5. ***Schematic of ligament peak force-peak elongation curves.*** Load-deformation curves spinal ligaments are nonlinear due to high flexibility of ligaments at low loads and increasing stiffness at higher loads. Area under the curve represents peak energy.



Statistics. Data from each cervical level, C2-C3 to C7-T1, were combined for each ligament. Single-factor, non-repeated measures ANOVA ($P<0.05$) and 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. Students un-paired t-tests ($P<0.05$) were performed to determine differences between present data and previously reported data for intact ligaments.(71)

Physiological Ligament Elongations

To enable comparison of present peak elongation to physiological elongation, a simple mathematical model was constructed using previously reported *in vitro* quantitative anatomy of cervical spine ligaments and vertebrae(61, 78), average *in vivo* normal intervertebral centers of rotation (CoRs), and average *in vitro* physiological intervertebral rotations.(32) For each FSU in neutral posture, the average CoR and ligament origin and insertion coordinates were identified in an anatomical coordinate system, fixed to the inferior endplate of the lower vertebral body. The upper vertebra was rotated about the CoR in flexion and extension to peak physiological rotations. Physiological 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. Average physiological elongations were obtained for each ligament by averaging data from C2-C3 to C7-T1.

Results

Average physiological elongations are shown in **Table 2**.

Table 2. ***Physiological ligament elongations (mm).***

Average (± 2 SD range) elongation, obtained using mathematical model, for 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). Rotation direction used to obtain 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

Dynamic force-elongation curves up to peak force, physiological elongation average, and range are shown in **Figures 6A-6F** (please see appendix). Peak ligament elongation was generally greater than the physiological range, with the exception of LF and ISL+SSL, which generally had peak elongation in excess of average physiological elongation.

Average peak force, elongation, and energy differed among ligaments (please see **Table 3A** on page 17). Highest peak force of 204.6 N was attained in LF, followed by 177.5 N in CL. Peak force in LF was significantly greater than in MTD and ISL+SSL. Highest peak elongation of 4.9 mm was observed in ISL+SSL, followed by 4.5 mm in CL and 3.9 mm in PLL. Peak elongation in ISL+SSL was significantly greater than in ALL, MTD, and LF. Peak energy ranged from 0.04 J in MTD to 0.44 J in CL. Peak energy in CL was significantly greater than in ALL, MTD, and ISL+SSL.

LF was generally the stiffest ligament, while ISL+SSL was generally the least stiff (please see **Table 3B** next page). At 25% of peak force, greatest stiffness of 77.2 N/mm was attained in LF, significantly greater than in ISL+SSL. LF stiffness was 96.5 N/mm and 112.1 N/mm at 50 and 75% of peak force, respectively; significantly greater than ALL, PLL, and ISL+SSL.

Table 3. ***Mechanical properties of human cervical ligaments at 725 mm/s following whiplash.*** 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. 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 column *Significant*. Blank entry indicates no significant difference was observed.

A) Peak Force, Elongation, and Energy

	Peak Force	Significant	Peak Elongation	Significant	Peak Energy	Significant
ALL	132.2 (70.1)		3.0 (0.8)	CL,ISL+SSL	0.21 (0.19)	CL
MTD	71.6 (59.6)	LF	1.4 (0.1)	PLL,CL,ISL+SSL	0.04 (0.30)	CL
PLL	149.4 (54.1)	ISL+SSL	3.9 (1.2)	MTD	0.29 (0.13)	
CL	177.5 (73.1)	ISL+SSL	4.5 (1.4)	ALL,MTD,LF	0.44 (0.25)	ALL,MTD,ISL+SSL
LF	204.6 (85.1)	MTD,ISL+SSL	3.4 (0.8)	CL,ISL+SSL	0.27 (0.11)	
ISL+SSL	67.6 (52.3)	PLL,CL,LF	4.9 (1.3)	ALL,MTD,LF	0.19 (0.16)	CL

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

	Stiffness at 25%	Significant	Stiffness at 50%	Significant	Stiffness at 75%	Significant
ALL	57.2 (34.5)	ISL+SSL	59.3 (35.5)	LF,ISL+SSL	60.2 (38.5)	LF,ISL+SSL
MTD	61.8 (48.9)		85.9 (72.3)	ISL+SSL	104.2 (90.5)	ISL+SSL
PLL	55.2 (21.1)	ISL+SSL	55.4 (23.9)	LF	53.8 (30.1)	LF
CL	60.7 (29.3)	ISL+SSL	57.6 (28.6)	LF,ISL+SSL	52.0 (32.2)	
LF	77.2 (38.8)	ISL+SSL	96.5 (51.0)	ALL,PLL,CL,ISL+SSL	112.1 (61.8)	ALL,PLL,ISL+SSL
ISL+SSL	21.7 (10.1)	ALL,PLL,CL,LF	19.1 (9.4)	ALL,MTD,CL,LF	15.7 (9.4)	ALL,MTD,LF

Discussion

The present study determined tensile mechanical properties, at a fast elongation rate, of human anterior longitudinal ligament (ALL), middle-third intervertebral disc (MTD), posterior longitudinal ligament (PLL), capsular ligament (CL), ligamentum flavum (LF), and interspinous and supraspinous ligament (ISL+SSL) following final whiplash simulation at 8 g. The ligaments were elongated to failure and peak force, peak elongation, peak energy, and stiffness values were determined from force-elongation curves. Highest average peak force of 204.6 N occurred in LF, significantly greater than lowest peak force in ISL+SSL. Highest average peak elongation reached 4.9 mm in ISL+SSL while highest average peak energy of 0.44 J was attained in CL. LF was the stiffest ligament, while ISL+SSL was least stiff. Peak elongation was generally above the physiological elongation range in most ligaments.

Experimental design of the present study has several important advantages over previous methods of whiplash simulation(52, 79) and tensile mechanical testing(63, 70, 72, 81). The whole cervical spine with muscle force replication (WCS-MFR) model for whiplash simulation has been validated elsewhere(75, 76), and advantages include enhanced biofidelity and the ability to investigate osteoligamentous injury mechanisms. Prior models used to simulate whiplash injury have included human volunteers(50, 79, 82), mathematical models(83-85), anthropometric test dummies(80, 86, 87), whole cadavers(49, 88, 89), and isolated whole cervical spine (WCS) models(52, 58, 90). Of these models only the WCS has proven able to quantify soft tissue injury both before and after whiplash simulation. The WCS+MFR model was introduced to overcome the limitations of the lack of muscle force simulation in WCS models. Bone-ligament-bone preparations for tensile testing optimized usage of scarce human cadaveric specimens. Rigid fixation of ligament attachments within mounts ensured mid-tissue tears during

elongation, preventing ligament avulsion from bone. Average peak ligament elongation rate of 725 mm/s was consistent with prior studies(70, 71, 74) and likely similar to that occurring in traumatic injury.

The present study has limitations that should be discussed before data interpretation. The average age of the present specimens was 70.8 years due to limitations in availability of young cadaveric material, making ligament specimens susceptible to age-related changes (91-94). Because only six cervical spines were available, ligament sample sizes were maximized by combining spinal levels for bone-ligament-bone preparations for data analyses to enable testing of hypothesis (**Table 1**). Prior mechanical studies demonstrating injury to ligaments following whiplash have shown that injuries are most likely to occur at C4-C5 through C6-C7 in ALL, CL, and MTD(53, 55, 56). Combining upper and lower intervertebral levels could potentially have influenced our results. Analyses comparing ALL, CL, and MTD ligaments by vertebral level showed no statistically significant differences. However, sample sizes for each level were very small, possibly obscuring real differences. Despite our efforts in maximizing samples for testing, difficulties in specimen preparation and complete rupture of some ligaments during whiplash simulation led to decreases in sample sizes for some ligaments. The exclusion of ligaments completely ruptured during whiplash simulation and the particularly small sample size of viable MTD specimens could have biased present study results. Additionally, ligament elongation along the direction of its fibers was difficult for MTD, CL, and ISL+SSL, due to anatomical constraints. Limitations of the WCS+MFR model have been discussed elsewhere(32, 75, 76, 95), and include fixation of T1 to the trauma sled and lack of active muscle force simulation. Lack of active muscle force contraction in WCS-MFR models is not likely to significantly affect results since reflexive muscle contraction of an unaware occupant after rear-impact

collision takes approximately 150 ms, whereas most damage from whiplash takes place within 100 ms after impact during s-curvature of spine.(47)

How do present ligament mechanical properties, obtained following whiplash, compare with previously published data of intact ligaments? Using similar methodology, Ivancic et al (2005) elongated intact cervical ligaments to complete rupture at 723 mm/s and reported peak force, peak elongation, peak energy, physiological elongation, and stiffness at 25, 50, and 75% of peak force for ALL, MTD, PLL, CL, LF, and ISL+SSL. Results of the present study are within range of those found in Ivancic et al (**Table 3A-3B**). For all six parameters maximum and minimum values occurred in the same ligament for both intact ligaments and ligaments following whiplash. In both studies highest peak force occurred in LF and was followed by CL, PLL, ALL, MTD, and ISL+SSL. Similarly, intact and whiplash ligaments had greatest peak elongation occur in ISL+SSL, followed by CL, PLL and LF, ALL, and MTD. In general for both studies, LF was the stiffest ligament followed by MTD, while ISL+SSL was the least stiff ligament across all three stiffness parameters. Following whiplash, there was greater variability between ligaments for stiffness at 25 and 50% of peak force compared to intact ligaments. At 25% of peak force following whiplash, there were four significant differences between ISL+SSL and ALL, CL, LF, and PLL, compared to one significant difference in intact group between ISL+SSL and CL. Similarly, at 50% of peak force following whiplash there seven significant differences involving all ligaments tested, compared to one significant difference between ISL+SSL and LF in intact study.

Force-elongation curves (**Figures 6A-6F**) in intact and whiplash studies were similar. In both studies peak elongations for ALL, MTD, PLL, and CL all occurred beyond physiological ranges of elongation calculated by mathematical model. LF and ISL+SSL

had peak elongations fall within physiological range in both studies, although in general they were above physiological elongation average. Rate of failure within physiological range was slightly higher in LF and ISL+SSL following whiplash. Following whiplash, 60% of LFs and 79% of ISL+SSLs failed within physiological range, compared to 40% of intact LFs and 67% of intact ISL+SSLs. LF and ISL+SSL failures within physiological range could suggest a risk of cervical spine instability should similar failures occur in vivo. We do not believe this to be true. Ivancic et al (2005) discussed this issue at length and found that ALL, MTD, PLL, and CL lie closer to the intervertebral centers of rotation necessitating a greater role in spinal column stabilization and protection than posterior ligaments that are further away from intervertebral centers of rotation. These findings are supported by a classic biomechanical study(96), which demonstrated smaller increases in spinal mobility with sequential transection from posterior to anterior compared to transection beginning with anterior components. These findings indicate that spinal stabilization is conducted primarily by the anterior cervical ligaments with LF and ISL+SSL playing a less significant role.(71)

Intact ligaments from Ivancic et al (2005) and ligaments following whiplash from present study were directly compared in **Table 4**. Intact ligaments had greater peak force, peak elongation, and peak energy at rupture than corresponding ligaments following whiplash. Correspondingly, the ranges of values for these parameters were greater in intact ligaments due to larger maximum values for each parameter. A few of these relationships approached significance. Intact CLs had an average peak force of 220.0 N at rupture, significantly higher than the 177.5 N in CLs following whiplash ($P< 0.05$). Intact ALLs had an average peak elongation of 4.0 mm at rupture, significantly greater than the average of 3.0 mm in ALLs after whiplash ($P< 0.05$). LF peak elongation approached significance with an average of 4.2 mm in intact ligaments and 3.4 mm

following whiplash ($P < 0.10$). CL approached significance for peak energy with an average of 0.57 J in intact ligaments and 0.44 J following whiplash ($P < 0.10$). No stiffness parameter comparisons approached significance.

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Table 4. ***Comparison of present mechanical properties of cervical ligaments obtained following whiplash with previously published data of intact ligaments.(71)*** Average (SD) peak force (N), peak elongation (mm), peak energy (J), and stiffness (N/mm) at 25%, 50%, and 75% of peak force. 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 ($P < 0.05$) between intact and whiplash ligaments are indicated with bold brackets, while trends ($P < 0.10$) are indicated with thin brackets.

		ALL	MTD	PLL	CL	LF	ISL+SSL
Peak Force	Intact	137.9 (111.5)	115.6 (79.9)	163.7 (80.2)	220.0 (83.7)	244.4 (143.0)	85.5 (67.6)
	Whiplash	132.2 (70.1)	71.6 (59.8)	149.4 (54.1)	177.5 (73.1)	204.6 (85.1)	67.6 (52.3)
Peak Elongation	Intact	4.0 (1.0)	2.1 (0.9)	4.2 (1.5)	4.9 (1.4)	4.2 (1.5)	5.9 (2.9)
	Whiplash	3.0 (0.8)	1.4 (0.1)	3.9 (1.2)	4.5 (1.4)	3.4 (0.8)	4.9 (1.3)
Peak Energy	Intact	0.25 (0.15)	0.12 (0.15)	0.33 (0.18)	0.57 (0.30)	0.36 (0.25)	0.33 (0.39)
	Whiplash	0.21 (0.19)	0.04 (0.30)	0.29 (0.13)	0.44 (0.25)	0.27 (0.11)	0.19 (0.16)
Stiffness at 25%	Intact	46.9 (38.8)	61.3 (38.8)	71.6 (49.7)	69.4 (34.3)	72.7 (44.0)	22.1 (12.7)
	Whiplash	57.2 (34.5)	61.8 (48.9)	55.2 (21.1)	60.7 (29.3)	77.2 (38.8)	21.7 (10.1)
Stiffness at 50%	Intact	49.9 (45.2)	81.4 (61.0)	63.6 (41.4)	65.1 (29.2)	98.8 (65.4)	21.3 (11.7)
	Whiplash	59.3 (35.5)	85.9 (72.3)	55.4 (23.9)	57.6 (28.6)	96.5 (51.0)	19.1 (9.4)
Stiffness at 75%	Intact	50.4 (53.4)	96.0 (79.1)	53.0 (33.5)	57.9 (28.9)	118.4 (82.9)	19.9 (11.7)
	Whiplash	60.2 (38.5)	104.2 (90.5)	53.8 (30.1)	52.0 (32.2)	112.1 (61.8)	15.7 (9.4)

The present study indicates that ALL and CL ligaments that have undergone whiplash may have altered biomechanical properties compared to their intact counterparts. Prior biomechanical studies of whiplash injury mechanisms have demonstrated that whiplash induces anterior longitudinal ligament (ALL) injury due to a hyperextension moment while shear and compressive forces lead to facet joint compression, capsular ligament (CL) stretching, and intervertebral disc disruption.(30, 31, 46, 48, 50-56) These studies suggest that whiplash-induced alterations to cervical ligament biomechanical properties would be most likely to occur in ALL, CL, and MTD. Results of the present study are consistent with these results. Although we found no statistically significant indications of MTD damage in the present study, the small sample size of only three MTDs may have limited power.

What are the potential implications of altered or dysfunctional dynamic properties in ALL and CL cervical ligaments? Because spinal ligaments contribute to spinal stability through direct resistance to forces on spine and by providing feedback to muscle stabilizers, it has been hypothesized that mechanical injury to ligaments could decrease their load-bearing capabilities causing instability. This would require compensation of load distribution by other structures such as facet joints or stabilizing muscles that could cause structural degeneration over time.(59, 62) Spinal ligaments are suspected to heal poorly, and this type of clinical scenario could lead to a repetitive cycle of mechanical injury altering dynamic properties and leading to instability, degeneration, and chronic pain.(53) Panjabi (2005) suggested a mechanism of chronic whiplash pain in humans whereby subfailure injuries damage spinal ligament mechanoreceptors and cause muscle dysfunction due to corrupted transducer signals from injured mechanoreceptors.(97) The resulting muscle dysfunction could cause ligament and facet joint stress leading to inflammation of neural tissue capable of causing whiplash pain.

The presence of mechanoreceptors in spinal ligaments has been documented in PLL, intervertebral discs, ISL+SSL, facet joints, and LF in numerous human and animal models.(98-102) Whiplash has been shown to alter propioceptive function in patients suffering chronic symptoms, giving further credence to mechanoreceptor disruption theory as a cause of chronic whiplash pain.(103, 104)

Recent neurophysiological and biomechanical studies have shed light on prior studies linking capsular ligament (CL) injury to whiplash pain. Kallakuri et al (2004) demonstrated the presence of pain-associated neuropeptides substance P and calcitonin gene-related peptides in human cervical facet joint capsule (FJC) nerve fibers, supporting the hypothesis of facet joint related whiplash pain.(105) Lu et al (2005) quantified the responses of FJC receptors in goat cervical spines to capsular deformation and found that subcatastrophic capsular injury could cause activation of FJC nociceptors and contribute to facet joint pain.(106) Lee et al (2004) used a rodent neck pain model to demonstrate behavioral hypersensitivity indicative of pain in rats subjected to tensile cervical facet joint loading, providing direct evidence of pain produced by facet joint injury.(107) Lee et al (2005) sought to characterize failure and subfailure mechanical properties of rat cervical facet capsular ligaments in tension and found that subfailure injuries could lead to nociceptive physiological changes in the spine leading to pain despite a lack of gross ligament injury(108). Results of the present study may contribute to the understanding of subfailure injuries to capsular ligament and their potential relationship to altered dynamic properties and symptoms resulting from spinal instability. Further studies are needed to elucidate the potential link between subfailure capsular ligament injury and whiplash pain.

Conclusion

The present study determined the tensile mechanical properties of cervical ligaments at a fast elongation rate after whiplash simulation. Ligaments studied included ALL, MTD, PLL, CL, LF, and ISL+SSL. Peak force was highest in LF, which was significantly greater than peak force in MTD and ISL+SSL. Peak elongation was lowest in MTD and highest in ISL+SSL. MTD elongation was significantly smaller than CL, PLL, and ISL+SSL while ISL+SSL had elongation significantly greater than LF, ALL, and MTD. Highest peak energy was found in CL, which was significantly greater than ALL, MTD, and ISL+SSL. LF was stiffest ligament and ISL+SSL was least stiff. When compared to intact ligaments, CL and ALL after whiplash had significantly lower peak force and peak elongation, respectively. Present data may be useful in understanding the effects of whiplash on dynamic mechanical properties of cervical ligaments, contribute to understanding subfailure cervical ligament injuries following whiplash, and contribute to the development of mathematical models of the spinal column following injury.

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Appendix

Figure 6. Ligament force vs. elongation, including average physiological elongation, shown by vertical dashed line, with two standard deviation range, indicated by grey shading. 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 (ISL+SSL).

Figure 6A

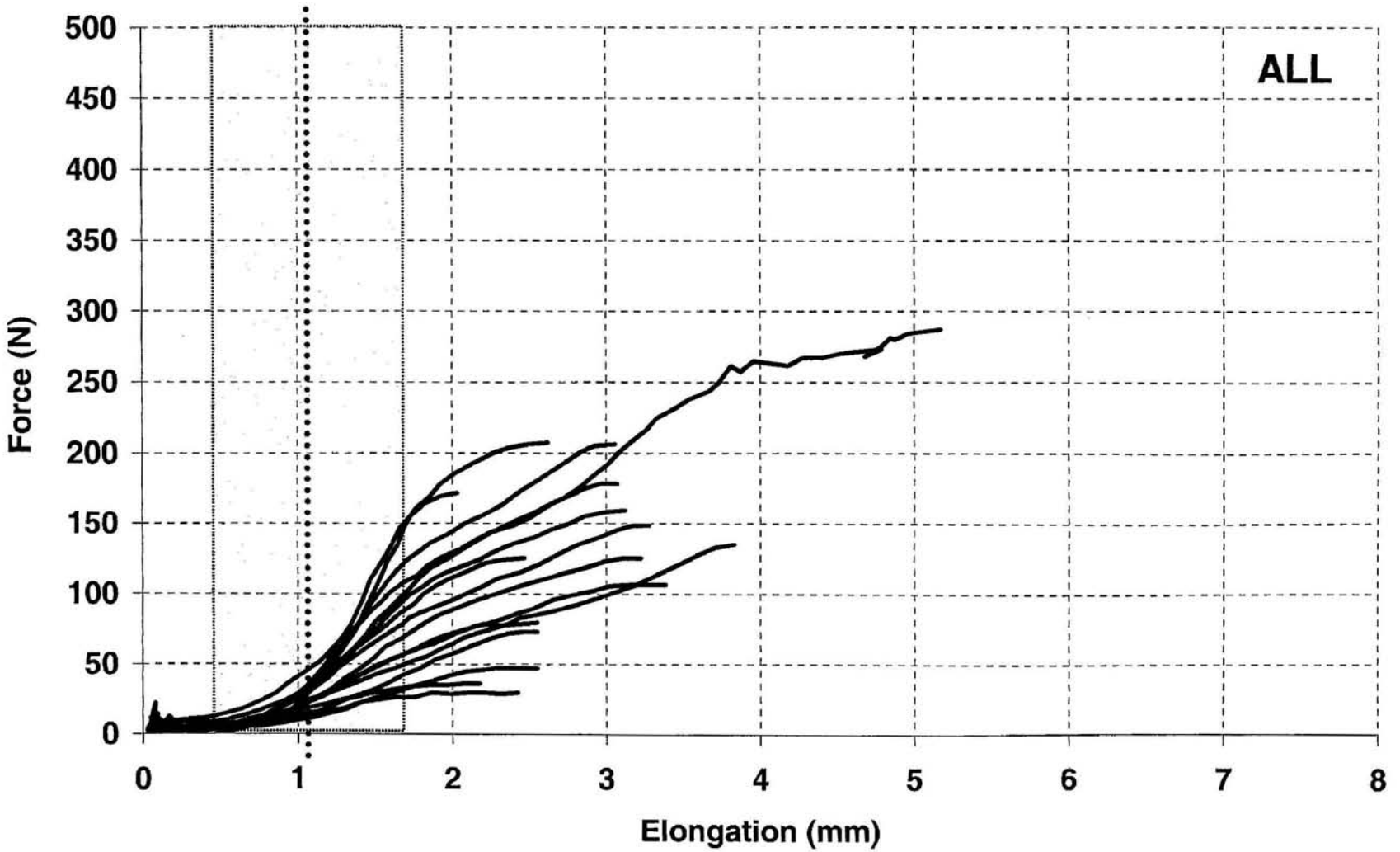

ALL
Force (N)
500
450
400
350
300
250
200
150
100
50
0
0
1
2
3
4
5
6
7
8
Elongation (mm)

Figure 6B

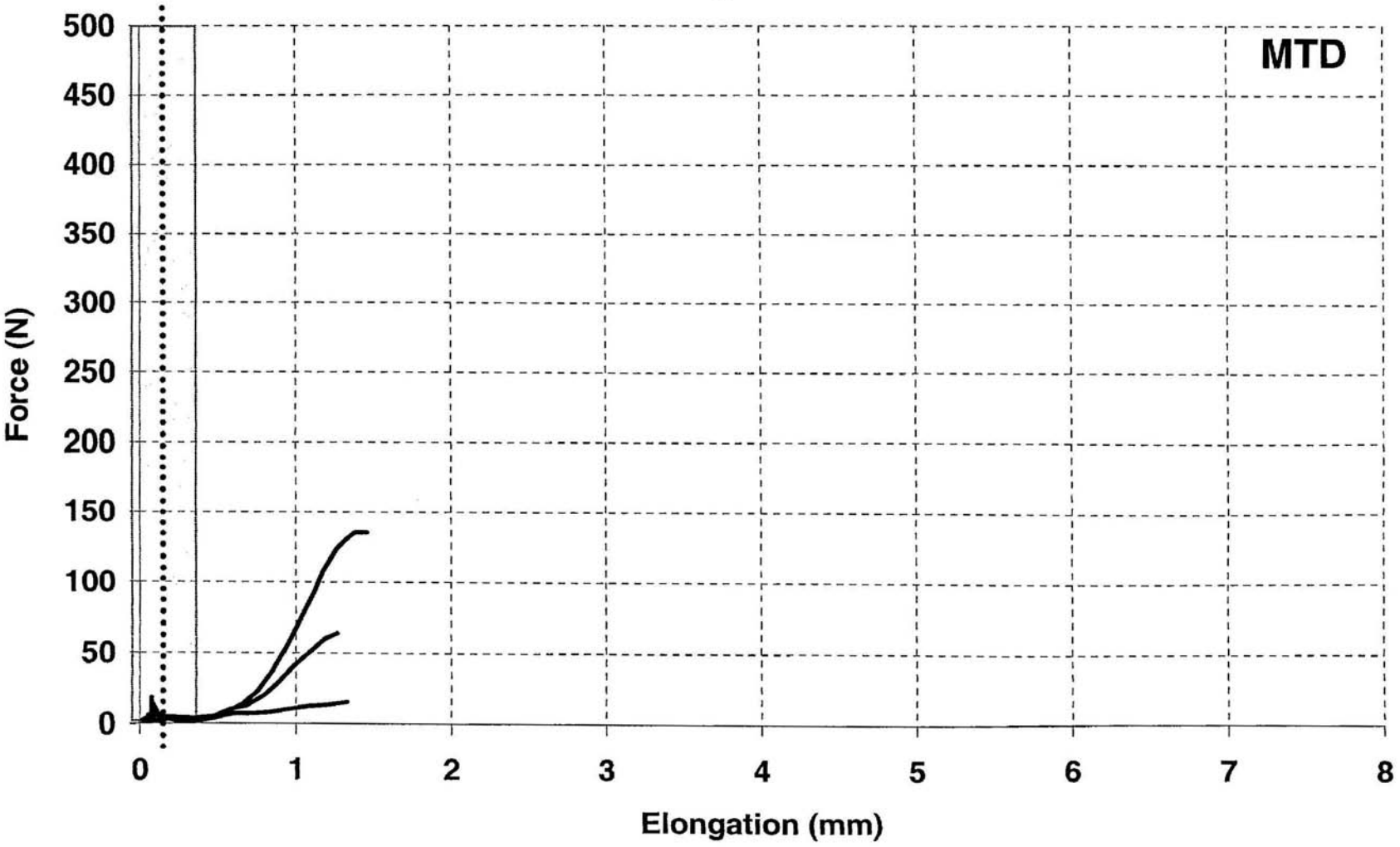

MTD
500
450
400
350
300
250
200
150
100
50
0
Force (N)
0
1
2
3
4
5
6
7
8
Elongation (mm)

Figure 6C

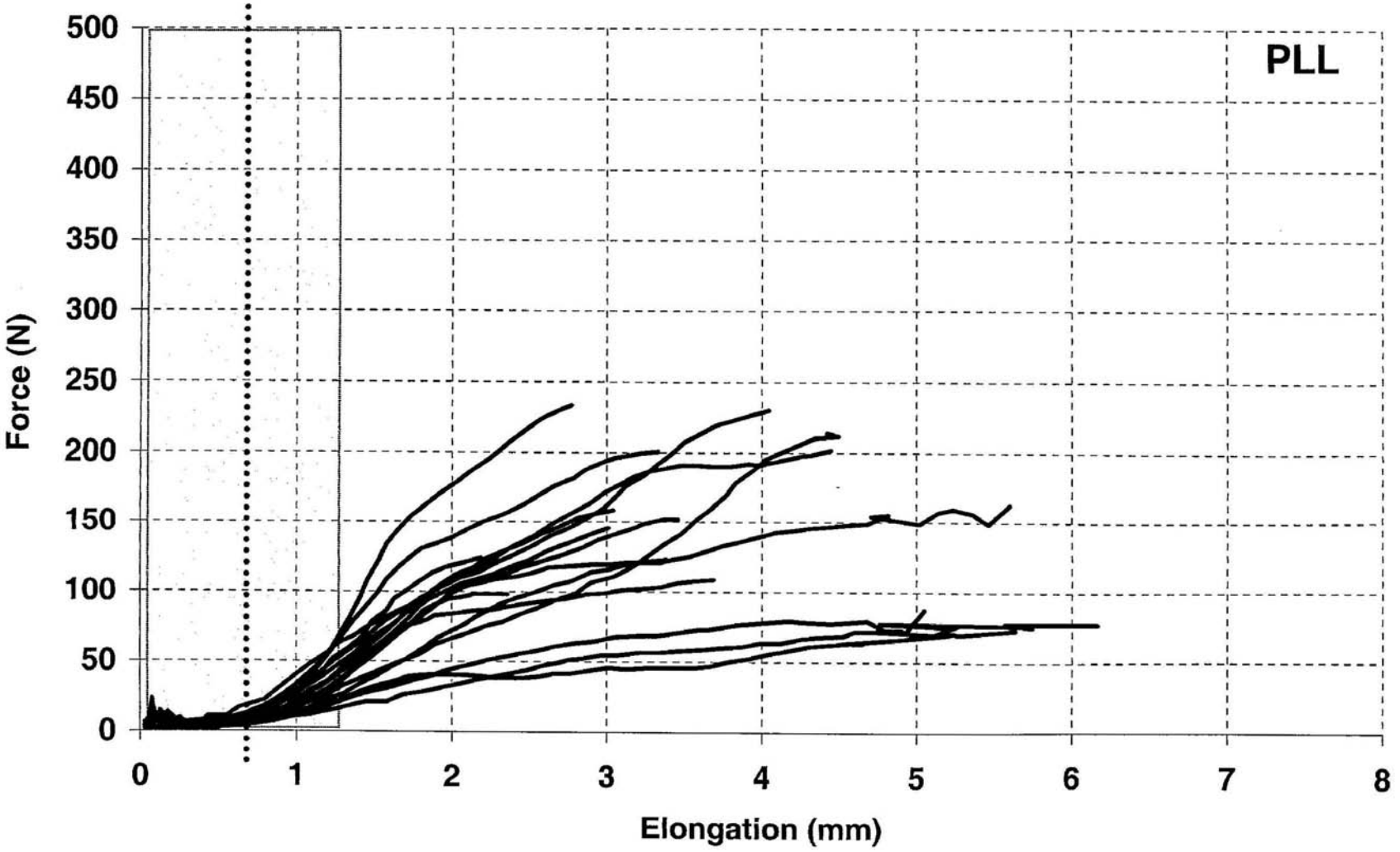

PLL
Force (N)
500
450
400
350
300
250
200
150
100
50
0
0
1
2
3
4
5
6
7
8
Elongation (mm)

Figure 6D

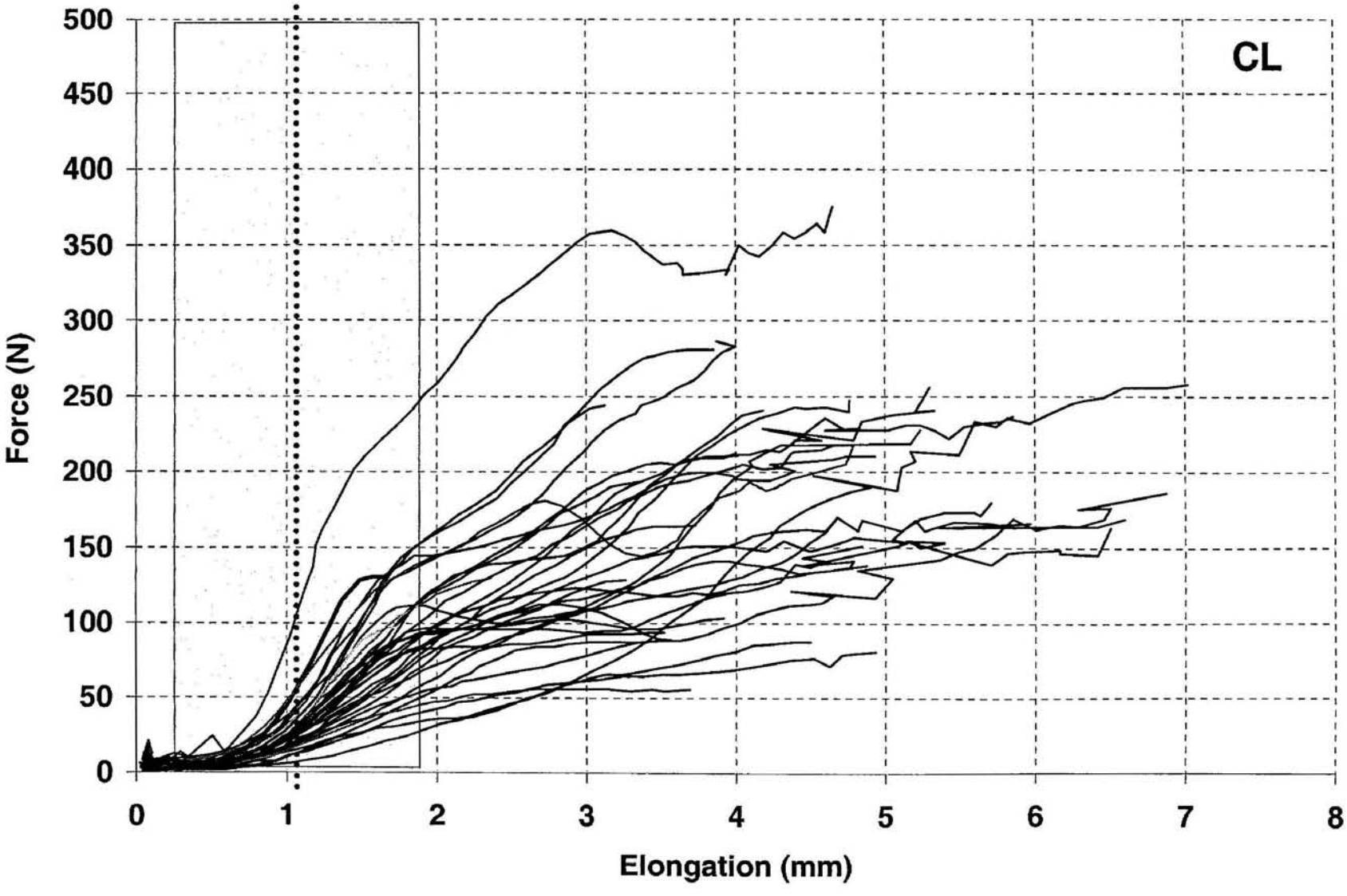

CL
Force (N)
500
450
400
350
300
250
200
150
100
50
0
0
1
2
3
4
5
6
7
8
Elongation (mm)

Figure 6E

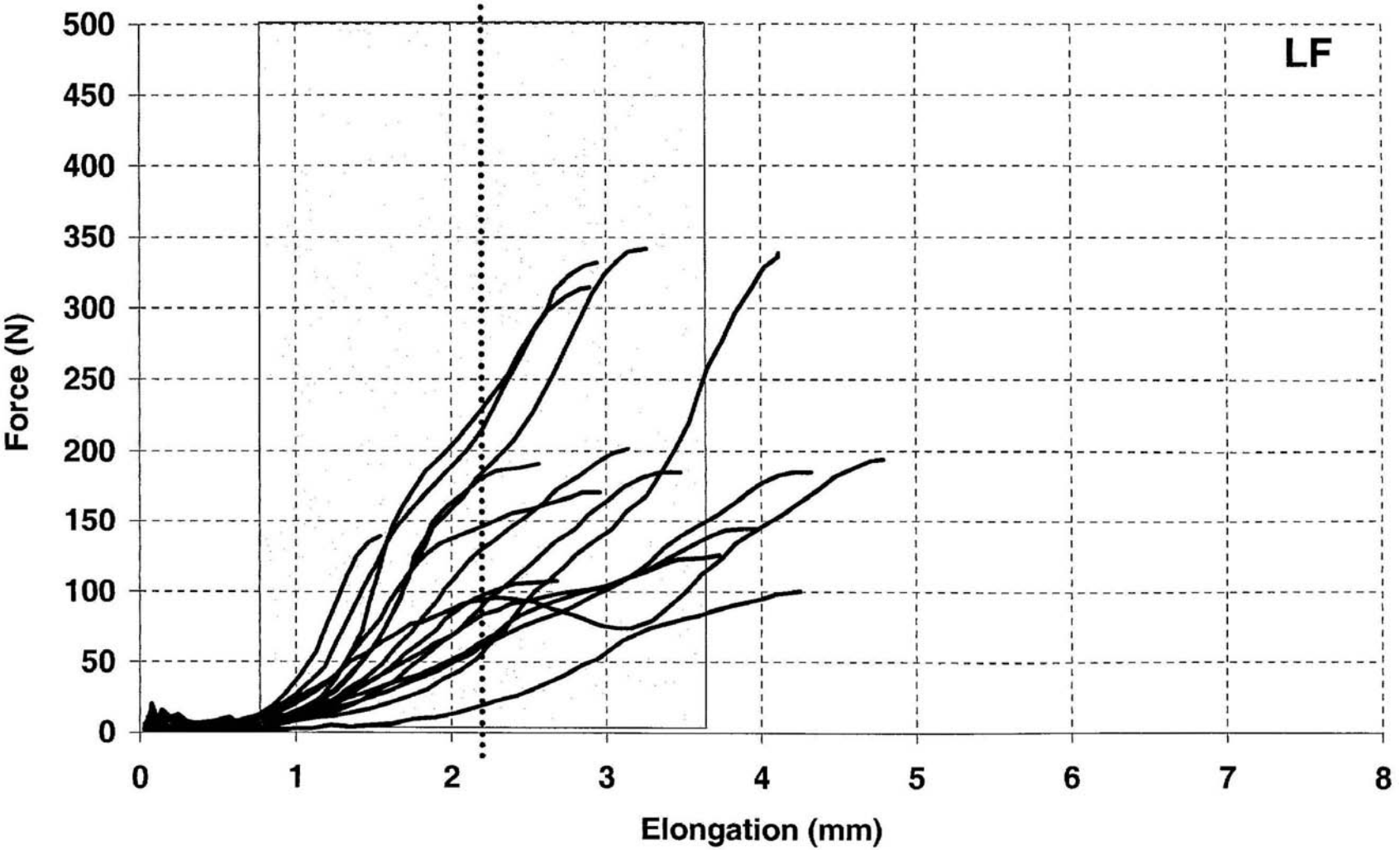
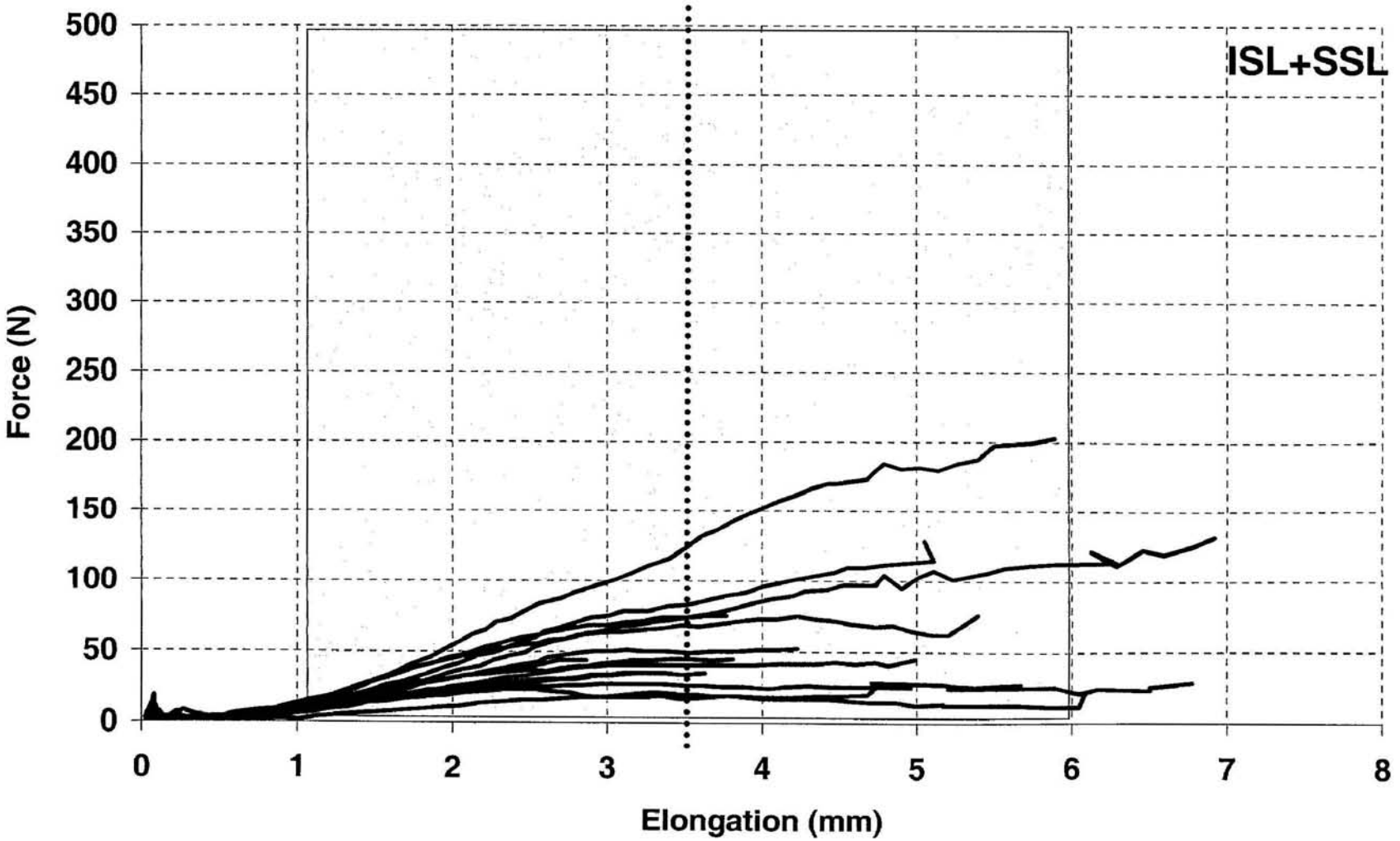

LF
Force (N)
500
450
400
350
300
250
200
150
100
50
0
0
1
2
3
4
5
6
7
8
Elongation (mm)

Figure 6F


ISL+SSL
Force (N)
500
450
400
350
300
250
200
150
100
50
0
0
1
2
3
4
5
6
7
8
Elongation (mm)

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