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Acute Surgical Injury Alters the Tensile Properties of Thoracolumbar Fascia in a Porcine Model

Recent work utilizing ultrasound imaging demonstrated that individuals with low back pain (LBP) have increased thickness and decreased mobility of the thoracolumbar fascia (TLF), an indication that the TLF may play a role in LBP. This study used a porcine injury model (microsurgically induced local injury)—shown to produce similar results to those observed in humans with LBP—to test the hypothesis that TLF mechanical properties may also be altered in patients with LBP. Perimuscular TLF tissue was harvested from the noninjured side of vertebral level L3-4 in pigs randomized into either control ($n = 5$) or injured ($n = 5$) groups. All samples were tested with a displacement-controlled biaxial testing system using the following protocol: cyclic loading/unloading and stress relaxation tests at 25%, 35%, and then 45% of their resting length. Tissue anisotropy was also explored by comparing responses to loading in longitudinal and transverse orientations. Tissues from injured pigs were found to have greater stretch–stretch ratio moduli (measure of tissue stiffness), less energy dissipation, and less stress decay compared to tissues from control pigs. Responses across these variables also depended on loading orientation. Clinical significance: these findings suggest that a focal TLF injury can produce impairments in tissue mechanical properties away from the injured area itself. This could contribute to some of the functional abnormalities observed in human LBP.
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Introduction

Low back pain (LBP) is a common musculoskeletal disorder in North America with 27.8% of people over the age of 18 reporting an episode of LBP each year [1] and 80% of all people experiencing LBP during their lifetime [2]. Most individuals with LBP (85%) do not have detectable damage to spinal structures, using current diagnostic imaging techniques, and are given a diagnosis of nonspecific LBP (ns-LBP) [3]. Underlying mechanisms for ns-LBP are not fully understood, contributing to relatively poor outcomes and wide disparity between treatment approaches [4]. While much focus has been directed at the intervertebral disk, nerve roots, facet joints, and associated spinal musculature, findings have not been conclusive for identifying structural or mechanical changes in people with ns-LBP unrelated to trauma or other pathologies [3].

People with ns-LBP often exhibit increased trunk muscle activation [5,6], decreased mobility [7], and stiffer postural control [8,9]. It is well understood that ns-LBP is a multifactorial issue with potential contributions from mechanistic and/or pathoanatomic causes (e.g., lesions in spinal structures) [10], psychological/behavioral issues (e.g., fear avoidance, hypokinesia) [11], and altered movement strategies [12]. While instructive, these paradigms have not fully explained many of the observed impairments in people with ns-LBP. A pathophysiological model has been proposed incorporating connective tissue, such as perimuscular fascia, and inflammatory processes as contributors to both pain and biomechanical changes in people with ns-LBP [13].

Thoracolumbar fascia (TLF) is an anatomical structure covering the superficial aspect of the dorsal trunk musculature. While historically considered less essential than the more specialized organs and muscles it envelopes [14], fascial tissue such as TLF is now gaining recognition for its role in many important functions. These functions include sustaining loads and stabilizing the spine, providing attachment for trunk and extremity muscles, potential proprioceptive function, and pain [14,15]. Ultrasound cine recordings during passive trunk flexion have demonstrated reduced shear strain between adjacent layers of the TLF in human subjects with chronic LBP compared to control subjects without LBP [15]. Recent work also demonstrated that a unilateral surgical injury to the TLF in a porcine model induces tissue changes on the side contralateral to the injury that were similar to those observed in human LBP patients (increased thickness and decreased shear

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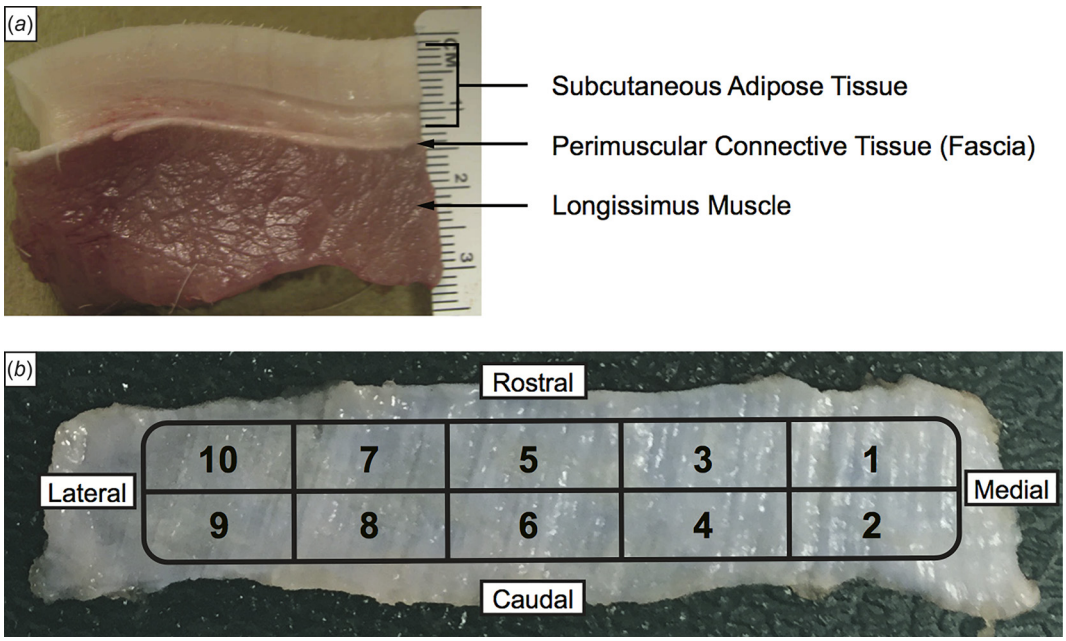

Fig. 1 Samples were extracted from each strip of tissue (*a*) and labeled according to the region they were taken from for consistency (*b*)

strain during passive trunk flexion) [16]. Both of these studies examined mobility between structural layers in the fascial complex, including the shear plane between deep subcutaneous and perimuscular layers [15,16]. The current research seeks to investigate mechanical properties of the deep perimuscular fascia layer in isolation using this porcine model.

As with all connective tissue, TLF is a biphasic material with both solid and fluid components that contribute to its viscoelastic behavior. Viscoelastic behavior of a material can be studied by investigating creep or stress relaxation. Measurement of creep requires a load-controlled mechanical testing approach, where the load can be applied and held constant while deformation is measured. Stress-relaxation response can be experimentally achieved using a displacement-controlled approach, where the material is deformed by a known displacement and the resulting force is measured and used to calculate stress. In this study, cyclic loading/unloading and single stepped displacement protocols were used to characterize the cyclic relaxation and stress-relaxation properties, respectively. From these two tests, tissue stiffness, energy storage and dissipation, and ability of the tissue to distribute stress can be estimated [17].

Purpose and Hypotheses. The primary purpose of this work was to investigate changes in mechanical properties of porcine TLF in response to a localized, contralateral fascia injury. Specific variables of interest included stress–stretch ratio (SS) modulus (a measure similar to elastic modulus, describing tissue stiffness), strain energy, hysteresis, and stress relaxation response (SRR). It was hypothesized that local fascia injury would result in detrimental changes to the mechanical properties of the contralateral TLF. A secondary purpose was to investigate the anisotropy of TLF by comparing responses to loading in longitudinal and transverse orientations under uniaxial and biaxial loading conditions.

Methods

This research was conducted utilizing porcine tissues from previous work at the University of Vermont [16]. Animal housing and treatment conditions are described elsewhere [16]. Briefly, castrated male domestic swine ($n = 20$), between the ages of 4 and 6 weeks were randomized into one of four groups: injury, movement restriction, injury plus movement restriction, and control ($n = 5$ per group). Efforts were made to control for potential variability among the pigs by using a matching approach where the study protocol was initiated on four pigs from the same litter at a time, with each littermate assigned to one of the four groups. The current study evaluated tissue samples from pigs in the injured ($n = 5$) and control ($n = 5$) groups. Therefore, each pig in the injured group had a matched littermate in the control group. Pigs in the injured group underwent a unilateral fascia injury in the dorsal trunk at L3-4 under isoflurane inhalation anesthesia. Blunt and microsurgical dissection tools were used to detach the perimuscular fascia from adjacent deep subcutaneous tissue, producing a 4×4 cm injury centered 2 cm lateral to the midline. The perimuscular fascia itself was not incised. The incision was closed with five interrupted nylon skin sutures. All animals were euthanized after 8 weeks, and $1 \times 5 \times 3$ cm transversely oriented tissue blocks were excised dorsally from the L3-4 level contralateral to the injury site. Tissue blocks were immediately frozen in liquid nitrogen and stored at $-80\,^{\circ}$C. Samples were shipped to University of Waterloo for the testing described in this work. Pigs in the control group underwent isoflurane anesthesia without injury.

Sample Preparation. Tissue blocks were thawed and the TLF (perimuscular fascia) was carefully dissected from underlying muscle. Several 5×5 mm test samples were taken from each tissue block for testing and numbered according to region of extraction (Fig. 1). Tissue blocks ranged in size from 1 to 3 cm. Attempts were made to extract the maximum number of samples from each tissue block. Samples were tested immediately following dissection.

Mechanical Testing Protocol. Each sample was assigned to a condition of uniaxial longitudinal (UL), uniaxial transverse (UT), or biaxial loading. Sample thickness was measured using a photoelectric sensor (ZX-LD40 L Smart Sensor, Omron Canada, Inc., Toronto, ON, Canada). Samples were mounted in a displacement controlled biaxial mechanical testing system (BioTester 5000, CellScale, Waterloo, ON, Canada) with tungsten rake attachments (tine diameter = 0.25 mm; tine spacing = 0.70 mm). Samples were oriented with the longitudinal axis parallel to, and the transverse axis perpendicular to the primary fiber direction (Fig. 2). For

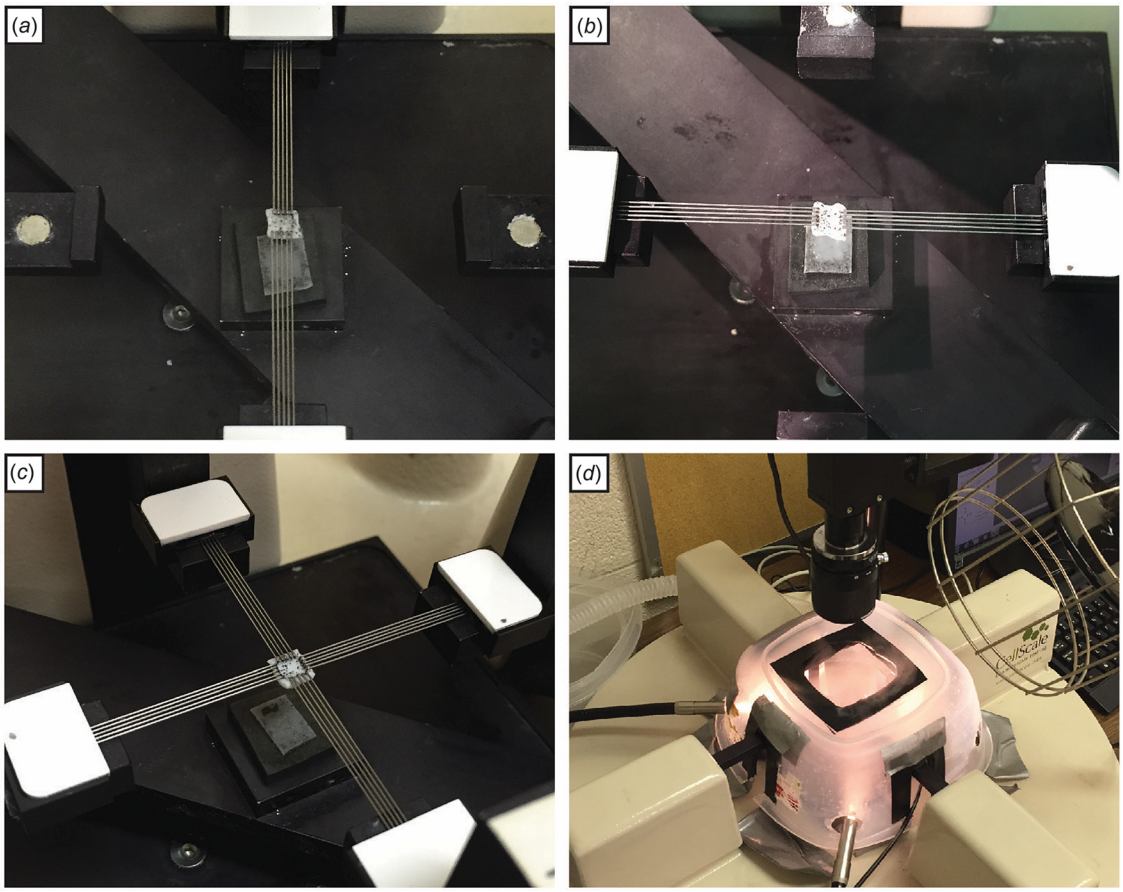

Fig. 2 TLF samples mounted for testing in the uniaxial longitudinal (*a*), transverse (*b*), and biaxial (*c*) conditions. Once mounted the testing apparatus was enclosed in a temperature and humidity controlled environment to ensure consistent hydration (*d*).

uniaxial test conditions, only rakes for the relevant axis were inserted. During testing, samples were housed in a temperature ($28 \pm 1\,^{\circ}C$) and humidity ($90 \pm 5\%$) controlled environment to ensure consistent hydration [18]. Samples were pretensioned to approximately 10 mN in each axis to remove tissue slack. Initial rake width was set at 3500 μm and displacement controlled as a percentage of starting rake width. Rake displacement, referred to herein as "stretch ratio," describes the amount of stretch relative to initial rake width. Samples were preconditioned with five cycles of loading/unloading to 10% stretch ratio at a rate of 6%/s. This stretch rate was deemed to be within expected ranges of physiological movement rates during voluntary lumbar spine flexion [15]. Following preconditioning, samples were returned to the zero position for a 60 s rest period. All samples followed the same test protocol, with stretch applied uniaxially in either longitudinal or transverse orientations, or simultaneously in both axes for biaxial loading.

The testing protocol consisted of five repetitions of cyclic loading/unloading and one stress-relaxation test at the following three stretch ratios: 25%, 35%, and 45%. For each test, samples were stretched at 6%/s, with 60 s rest periods between each condition (Table 1). Force and displacement data were sampled at 30 Hz.

Data Analysis. Surface dimensions of each sample were determined by digitizing its edges on images captured by an overhead camera (Sony XCD-910, Sony Electronics, Inc., Tokyo, Japan) using LABJOY image analysis software (LabJoy v10.24, CellScale, Waterloo, ON, Canada). The images used for this calculation were captured immediately prior to preconditioning. Surface dimensions were combined with measured thickness values to calculate cross-sectional area (CSA) and volume.

Time-varying force and actuator displacement data were processed in MATLAB (version R2015a, Mathworks, Natick, MA). Signal noise was attenuated using a fourth-order, zero-lag, low-pass Butterworth filter ($fc = 3$ Hz). Normal engineering stress was calculated as measured tensile force divided by initial cross-sectional area, which included 35% of the apron width (i.e., region of sample outside rake boundary) [19]. Stress versus stretch ratio curves were produced using displacement measured from the BioTester's linear actuators.

Stress–Stretch Ratio Modulus During Cyclic Loading. For each loading axis, SS moduli, which provided an intrinsic measure of stiffness, were computed as slopes of the linear region of stress–stretch ratio curves for each stretch level (25%, 35%, and 45%). The linear region was defined as the central section of a region bounded by the upper limit of the nonlinear toe region and the endpoint of the loading curve (Fig. 3). Each curve was visually

Table 1 Test protocol used for each sample included preconditioning followed by cyclic loading and static hold conditions at incrementally increasing stretch ratios

No. of cycles	Stretch ratio (%)	Hold duration (s)
5 (preconditioning)	10	0
5	25	0
1	25	240
5	35	0
1	35	240
5	45	0
1	45	240

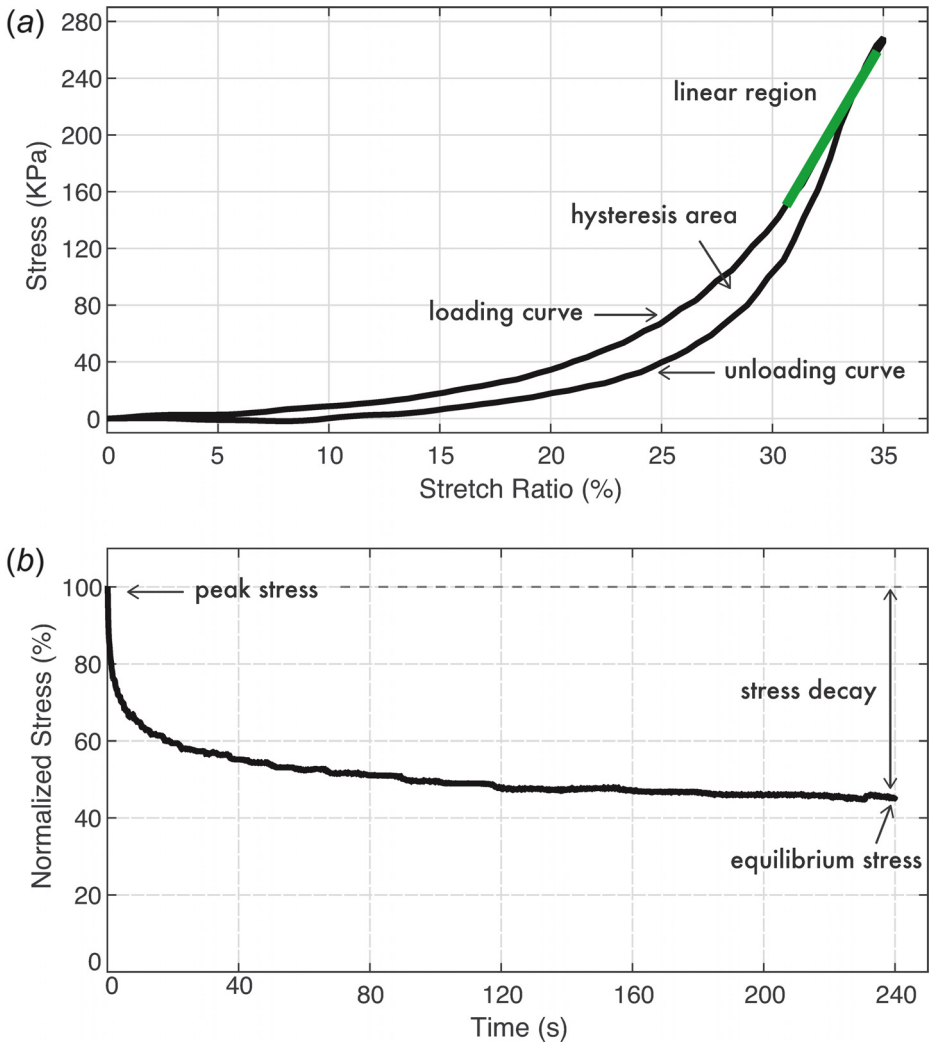

Fig. 3 Representative data of a control sample undergoing UL tension in (*a*) cyclic loading (fourth cycle shown) and (*b*) stress-relaxation. Both tests are depicted for the 35% stretch condition.

inspected to ensure the linear region was appropriately defined. Stress–stretch ratio moduli values were averaged from fourth and fifth cycles as loading responses were most stable during these cycles.

Strain Energy and Hysteresis During Cyclic Loading. Strain energy was calculated as area under the loading portion of the force–displacement curve (Fig. 3), normalized to the initial volume of each tissue sample. This provided a measure of strain energy density, expressed in microjoules per unit volume, in millimeters cubed (μJ/mm^3). Hysteresis represented energy dissipated in the load/unload cycle, and was calculated as the difference between loading and unloading energies, expressed as percentage of loading energy. Hysteresis and strain energy values were also averaged from the fourth and fifth cycles at each stretch level.

Stress Relaxation Response During Static Stretch. Stress relaxation response was calculated by normalizing equilibrium tensile stress (after 240 s of static stretch) to peak stress (when displacement was first applied) (Fig. 3) and subtracting this value from 100%, expressed as percentage reduction from peak stress. Higher SRR values reflect greater stress decay.

Statistical Analyses. Mechanical properties were calculated for each of the four loading conditions: UL, UT, biaxial longitudinal (BL) and biaxial transverse (BT) at each of the three stretch levels. Tissue sample dimensions were compared between injury groups with *t*-tests. Differences in mechanical properties between injury groups, loading condition, and stretch level were explored with a $2 \times 4 \times 3$ mixed model ANOVA (SPSS, Inc., Chicago, IL). Relevant findings included main effects and injury group interactions showing differing responses to loading and/or stretch conditions between injured and control pigs. *T*-tests with Bonferroni adjustments were used for post hoc testing when appropriate. An alpha level of 0.05 was used for all tests.

Results

The number of samples each tissue block yielded ranged from 3 to 9 for a total of 80 (41 from five control pigs; 39 from five injured pigs). Sample dimensions are summarized in Table 2.

Table 2 Mean (SD) sample dimensions. There were no significant differences ($p > 0.05$) in thickness or CSA between injured and control group samples.

	Thickness (μm)	CSA—axial (mm^2)	CSA—transverse (mm^2)
Control	908.4 ± 354.0	2.72 ± 2.19	2.19 ± 2.08
Injured	943.2 ± 331.5	2.58 ± 2.21	2.35 ± 2.10
p-value	0.63	0.77	0.73

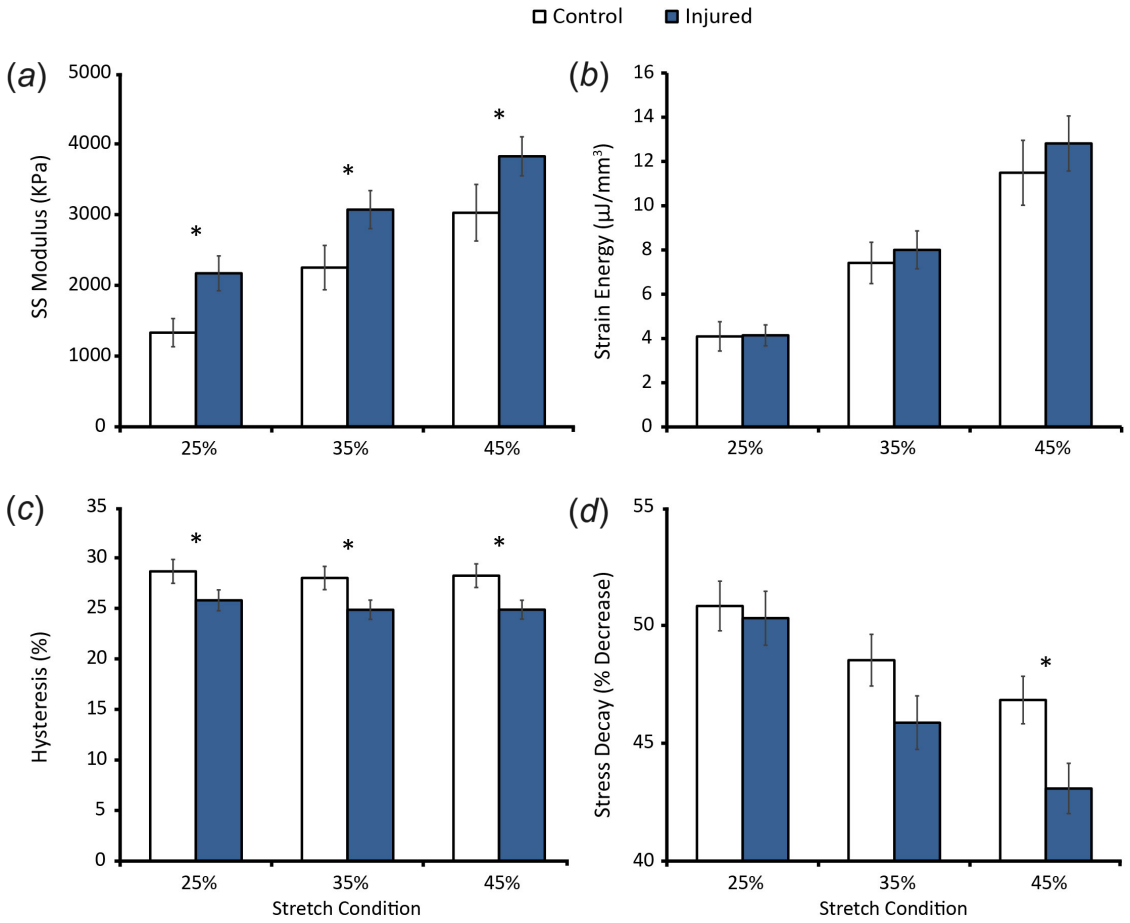

Fig. 4 Tensile testing results: (*a*) SS modulus, (*b*) strain energy, (*c*) hysteresis, and (*d*) stress decay compared between injured and control tissues at each stretch level. Average values depicted for each measure are collapsed across all orientations and loading conditions. Error bars represent the standard error associated with each mean. Significant differences between injured and control values are denoted by "*".

Stress–Stretch Ratio Modulus During Cyclic Loading. Samples from one control pig were not included in comparisons of SS moduli as this sample was significantly thinner (464.0 ± 108.2 μm, $p < 0.001$) than the other control pigs (908.4 ± 354.0 μm). Therefore, 71 tissue samples (32 from four control pigs and 39 from five injured pigs) were included for this analysis.

There were significant main effects of injury group ($F_{1,63} = 5.11$, $p = 0.027$), loading condition ($F_{3,63} = 5.07$, $p = 0.003$), and stretch ($F_{2,126} = 135.1$, $p < 0.001$). Thoracolumbar fascia samples from the injured pig group had higher SS moduli than those from control animals (3021.7 ± 1756.8 kPa versus 2203.0 ± 1869.5 kPa, respectively). Regardless of injury group, SS moduli also depended on loading condition ($F_{3,63} = 5.1$, $p = 0.003$) and stretch level ($F_{2,126} = 135.5$, $p < 0.001$). Overall, tissues had greater moduli values when loaded transversely versus longitudinally (BT = 3750.3 ± 2101.3 kPa, UT = 2596.8 ± 1741.5 kPa, BL = 2278.5 ± 1545.4 kPa, UL = 1970.6 ± 1425.1 kPa); however, only the BT condition was significantly different from the others on post hoc testing (*t*-tests with Bonferroni adjusted alphas). Thoracolumbar fascia moduli values also increased with stretch level (25% stretch: 1792.3 ± 1403.6 kPa; 35% stretch: 2701.4 ± 1733.1 kPa; 45% stretch: 3464.5 ± 1981.0 kPa), with all three stretch levels being significantly different from each other on post hoc comparisons. There were no significant interactions involving injury group. Figure 4(*a*) shows average SS moduli for injured and control groups at each stretch level.

Strain Energy and Hysteresis During Cyclic Loading. There was a significant main effect of stretch ($F_{2,144} = 173.4$, $p < 0.0001$), with strain energy increasing at higher stretch levels (25% stretch = 4.1 ± 3.4 μJ/mm^3, 35% stretch = 7.7 ± 5.3 μJ/mm^3, and 45% stretch = 12.1 ± 8.0 μJ/mm^3), and all stretch conditions differing significantly from each other on post hoc comparison ($p < 0.001$). There was also a main effect of loading condition ($F_{3,72} = 10.23$, $p < 0.0001$), with tissue samples having higher strain energy during transverse compared to longitudinal loading conditions (UL = 4.9 ± 3.7 μJ/mm^3, BL = 5.8 ± 4.2 μJ/mm^3, UT = 10.2 ± 7.4 μJ/mm^3, and BT = 11.7 ± 8.3 μJ/mm^3). There were no significant main effects or interactions that included injury group for strain energy (Fig. 4(*b*)).

There were significant main effects of injury group ($F_{1,72} = 5.9$, $p = 0.018$) and loading condition ($F_{3,72} = 7.0$, $p < 0.0001$) for hysteresis. Tissue samples from control pigs (28.3 ± 7.4%) dissipated more energy than those from injured pigs (25.2 ± 6.0%). Tissue samples dissipated more energy (i.e., higher hysteresis) when loaded longitudinally versus transversely (UL = 30.7 ± 7.2%, BL = 27.5 ± 6.0%, UT = 26.2 ± 5.9% and BT = 22.4 ± 5.5%). Stretch level did not affect energy dissipation ($F_{1,144} = 2.16$, $p = 0.119$), and there were no interactions with injury group. Figure 4(*c*) shows average hysteresis for injured and control groups at each stretch level.

Stress Relaxation Response During Static Stretch. There was a significant injury group by stretch interaction ($F_{2,144} = 5.87$, $p = 0.004$), with samples from the injured group having significantly lower stress decay during relaxation tests at 45% stretch only ($t = -2.56$, $p = 0.012$) (Fig. 4(*d*)). Regardless of injury group, SRR was significantly affected by loading condition

($F_{3,72} = 44.6$, $p < 0.0001$). Specifically, SRR was lower in transverse versus longitudinal loading conditions (BT = 42.3 ± 5.2%, UT = 43.1 ± 5.8%, BL = 48.8 ± 4.5%, and UL = 54.7 ± 5.4%). Stress relaxation decreased with increasing stretch levels ($F_{2,144} = 58.5$, $p < 0.0001$), with SRR at 25% stretch (50.6 ± 6.9%) being higher than both 35% (47.2 ± 7.1%, $p = 0.003$) and 45% (45.0 ± 6.8%, $p < 0.001$) stretch levels.

Discussion

The primary purpose of this work was to investigate changes in mechanical properties of porcine TLF following a localized, contralateral fascia injury. The hypothesis that surgical TLF injury would result in detrimental effects to mechanical properties of uninjured fascia away from the site of the injury was primarily supported. There were several detrimental effects of a contralateral injury on perimuscular fascia. Samples from injured pigs were found to have greater SS moduli, lower energy dissipation, and less stress decay during prolonged stretch.

The surgical fascia injury utilized in this research was meant to simulate a relatively mild soft tissue injury to the low back, involving fascia with subsequent inflammatory response but no structural damage that would impact low back mechanics [16]. To our knowledge, this is the first investigation of fascia mechanical properties utilizing an induced injury model in vivo. All mechanical properties investigated in this study, with the exception of energy storage, were found to differ between injured and control pigs, even though the perimuscular fascia itself was not incised in the surgical protocol, and injury was induced contralateral to the side tested tissues were extracted from.

Tissue samples from injured pigs were found to have higher tensile SS moduli than those from control pigs at physiological stretch rates. This result is consistent with findings of decreased tissue mobility in both people with ns-LBP and living pigs from which tissues were harvested for this study. Langevin et al. [15] found decreased shear strain mobility between fascial layers in people with LBP compared to healthy controls when subjected to passive flexion. This finding was replicated in living pigs experiencing the surgical fascia injury described in this work [16]. Our results, taken in conjunction with ultrasound findings observed in living pigs and humans, suggest that changes in mechanical properties of perimuscular fascia could play a role in mobility limitations observed in people with LBP.

While the impact of a controlled contralateral injury on measures of perimuscular fascia stiffness have not—until now—been studied, this increase in stiffness (as quantified herein by SS moduli) may have occurred as a result of chronic inflammation or biochemical changes similar to aging. Hammer et al. [20] found older donors had significantly stiffer iliotibial tract tissues compared to young donors. Although no biochemical tests were conducted, the authors hypothesized these findings could be attributed to decreased elastin and water content and a shift from collagen-III to collagen-I in older subjects. Similar increases in stiffness with aging have been documented in deep temporal fascia [21].

Corresponding to increased SS moduli observed in injured pigs, hysteresis (energy dissipation) decreased compared to control pigs. Previous work by Shadwick [22] described similar changes in digital flexor tendons with maturation in pigs and considered these to be a normal developmental response correlated with specific functional requirements of this structure. The author attributed the observed changes to morphological and biochemical factors: primarily higher collagen content and larger fibril diameter, and noted these changes are similar to processes observed during wound healing [22].

Tissue from injured pigs also exhibited decreased relaxation response when exposed to constant stretch. Stress relaxation is a time-dependent viscous response that allows biological tissue to return to its relaxed state and potentially minimizes total stress in the tissue [17]. Significant group differences were detected only at the highest (45%) stretch condition. These results are similar to findings from studies investigating biomechanical responses to injury in porcine skin. Corr and Hart [23] reported greater compliance in axial compared to transverse loading of uninjured skin, similar to our findings of greater relaxation response in the axial loading condition regardless of injury group. These authors also reported greater relaxation responses and higher low load stiffness in scarred skin, consistent with our findings in the fascia from injured pigs. Assuming a normal relaxation response helps to minimize stress in tissue, this finding could be a potential mechanism for recurrence of LBP episodes and/or chronicity of LBP following a single fascia injury. Full trunk flexion is an important spinal motion for functional activities of daily living such as dressing and lifting. While the amount of stretch the TLF undergoes during full trunk flexion in humans has not been quantified, it is possible that 45% elongation applied in this study produces TLF stretch similar to what would occur near the end range of functional trunk flexion in humans, a stretch level at which fascia injury may occur. Liao and Vesely [17] found differences in SRR between three types of mitral valve chordae; however, there were no observed differences in collagen, water, or elastin content between the three tissues. They did, however, find differences in glycosaminoglycan (GAG) content, with highest GAG levels being associated with lowest amount of relaxation. This finding has been replicated in bovine tendons whereby selective knockout of GAG molecules resulted in greater SRR [24]. Other work reported increased water content in injured ligaments corresponding to higher stress-relaxation behavior [25], with the hypothesis that increased fluid content creates greater exudation potential, resulting in increased viscous behavior. Biochemical analyses could help explain whether morphological and/or compositional changes that might impact proteoglycan, and resulting fluid, content have occurred in these tissues and provide insight into underlying mechanisms.

A secondary aim was to investigate the anisotropy of TLF through comparing properties when tensile loading occurred along longitudinal versus transverse fiber orientations under biaxial and uniaxial loading conditions. Thoracolumbar fascia was found to be highly anisotropic with responses across all variables depending on loading direction. Previous studies have also found fascia to be anisotropic; however, there is variability in these findings depending upon which structure the fascia was harvested from. The finding of greater SS modulus in the transverse orientation is consistent with that of Henderson et al. [26] who found dorsal–ventral (analogous to "transverse" in this work) orientation of canine thoracolumbar fascia to have an elastic modulus greater than three times that of cranial–caudal orientation. Human cadaver studies testing abdominal [27], plantar [28], and crural fascia [29] all observed tissue anisotropy to be in the opposite direction, with longitudinal exhibiting higher moduli than transverse orientation. This is likely an example of structure following function as these tissues all have greatly different functional requirements. Both canines and pigs are quadruped animals and this could explain the difference in directionality of these findings in comparison to the human abdominal fascia, as differences in gravitational loading likely lead to different mechanical requirements of the tissue. The TLF also transmits large forces to abdominal musculature through the lateral raphe and therefore requires the ability to resist stress transversely [14–30].

This study had several notable limitations. It is unknown to what degree findings from the porcine model transfer to human tissues. Although the observed reduction in shear plane fascia mobility of people with ns-LBP can be reproduced using the current porcine injury model, mechanisms underlying ns-LBP may depend on additional factors. As the tested samples were extracted from a very small region, it is unknown if our findings can be generalized to the whole tissue. While the pigs received the same care and handling throughout the course of the study, with the exception of the surgical injury, it is possible there were differences among individual pigs that could have impacted our findings.

Attempts were made to control for variability in the pigs by using a matched approach where littermates were assigned in pairs to control and surgical conditions. It must also be noted that the control pigs were not a true "control" group in that they were not subjected to any surgical procedure, whereas a "true control" condition would have used a sham surgical condition. Additionally, variability of ligament mechanical properties have been demonstrated between different breeds of pigs [31], and this is a potential limitation for generalization of findings from an animal model. Because of the limited number of pigs in the parent study, we needed to take multiple tissue samples from each pig in order to have a sufficient sample size to randomize to the various experimental conditions. That is, for the current study, tissue sample, not animal was considered the unit of analysis. Moderate correlation was observed between tissue samples within pigs (ICC = 0.54). The estimated between-pig standard deviation ($\sigma_{\text{between pig}} = 1101$) and between-tissue sample within-pig standard deviation ($\sigma_{\text{within-pig}} = 1025$) were quite similar. In order to confirm our findings, a future study with a larger number of animals is needed to confirm that our findings hold when individual pigs are used as the experimental unit. Without a biochemical analysis, we were unable to determine if inflammatory mediators responsible for morphological changes (i.e., collagen synthesis, fibroblast formation) were present following injury. Such analyses may help explain mechanical changes observed in injured TLF.

Overall, findings of this study support the hypothesis that TLF mechanics are impacted through soft tissue injury, even away from the injured area itself, and changes in these mechanics may play a role in the underlying mechanisms for ns-LBP. Force transmission, specifically, can be altered through injury-induced changes to mechanical properties of tissue regions even away from the injury site. As connective tissues such as perimuscular fascia play a substantial role in force transmission between active and passive structures, their mechanical properties are important to whole body mobility. Nevertheless, until recently, fascia tissue has been overlooked as a potential contributor to LBP. While beyond the scope of the current study, there is a potential interaction between fascia stiffness and transmission of mechanical forces to nociceptors that might modify ascending sensory input to peripheral pain pathways. This potential mechanism in combination with altered nociceptive thresholds in chronic pain conditions could be of interest for future studies of musculoskeletal pain syndromes. Given the growing evidence for both mechanosensory and nociceptive function in fascia tissue [32–34], relationships between biomechanical changes, such as those observed in this work, and development of pain should be explored in future studies.

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