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Is thoracolumbar fascia shear-wave modulus affected by active and passive knee flexion?

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Abstract

The purpose of this study was to examine the effect of passive and active knee flexion efforts on the stiffness of the thoracolumbar (TLF), semitendinosus (STF), and semimembranosus fascia (SMF). Fourteen young healthy males participated in this study. Using ultrasound shear-wave elastography, fascia elastic modulus was measured at rest (passive condition) and during submaximal isometric knee flexion efforts (active condition) with the hip at neutral position and the knee flexed at 0°, 45°, and 90°. Analysis of variance designs indicated that when the knee was passively extended from 90° to 0°, shear modulus of the TLF, SMF, and STF increased significantly ($p < 0.05$). Similarly, active knee flexion contractions caused a significant increase in TLF, SMF, and STF shear modulus ($p < 0.001$). Compared to hamstring fascia, the TLF showed greater thickness but a lower shear modulus ($p < 0.05$) while STF modulus was greater compared that to SMF during active contraction ($p < 0.05$). These results indicate that exercising the hamstring muscles can remotely influence the stiffness of the fascia which surrounds the lumbar area.

KEYWORDS
elastography, in vivo, myofascial path, semimembranosus, semitendinosus, spine, stiffness

1 | INTRODUCTION

The thoracolumbar fascia (TLF) is composed of several layers of connective tissue, which are arranged such that adjacent layers can glide over one another (Benjamin, 2009). TLF's role is to transmit and absorb loads during trunk movements and to maintain body posture (Willard et al., 2012). This fascia maintains strong linkages with various tissues, such as the erector spinae, the gluteus medius, and the sacrotuberous ligament (Vleeming et al., 1995; Willard et al., 2012). Theoretical estimations have shown that during trunk flexion, the TFL can exert high extension torque at the level of the fourth and fifth lumbar ligament (Gracovetsky, 2008). Exertion of forces from muscles or bones or ligaments changes the mechanical properties of the TLF which, in turn, transmit these forces to neighboring tissues (Findley et al., 2012), suggesting the important role of TLF conditioning for trunk stability and function.

It has been clinically proposed that myofascial connections between adjacent muscles form large chains that allow transfer of forces though these paths (Myers, 1997). One of these paths refers to a serial connection between the erector spinae, the sacrotuberous ligament, the hamstrings, the gastrocnemius, and the plantar fascia ("the superficial back line") (Myers, 1997). Research studies have confirmed connections between the TLF and the gluteus medius, the gluteus maximus and the sacrotuberous ligament (Vleeming et al., 1995; Willard et al., 2012), between the sacrotuberous ligament and the tendon of the semitendinosus (ST) and biceps femoris (Bierry et al., 2014; Vleeming et al., 1995) as well as between the hamstrings' and the gastrocnemius' fasciae (Cruz-Montecinos et al., 2015). In a cadaveric experiment, passive elongation of the biceps femoris caused displacement of the deeper lamina of the TLF, thus contributing to an effective load transfer between the spine, pelvis, legs, and arms (Vleeming et al., 1995). More recent studies have further supported the theory of serial connection between skeletal muscles and fasciae. For example, there is evidence that myofascial release of the lumbar muscles influences the flexibility of the hamstring musculature (Fauris et al., 2021). Similarly, myofascial release applied to the plantar surface of the foot influenced the sit-and-reach test score, which is a measure of hamstring and lumbar muscles' flexibility (Grieve et al., 2015). Further, passive or active tensioning of the latissimus dorsi impacted hip range of motion and stiffness (Carvalhais et al., 2013). Others reported significant associations between pelvic range of motion and displacement of the gastrocnemius fascia (Cruz-Montecinos et al., 2015) or between passive changes in ankle position and the length of the semimembranosus (SM) muscle (Wilke et al., 2020).

It has been proposed that alteration of fasciae stiffness during joint movements accompanies transfer of loads from the fasciae to adjacent tissues (Findley et al., 2012; Krause et al., 2016; Willard et al., 2012). Advances in ultrasound (US) imaging have allowed the reliable in vivo measurement of muscle tissue stiffness using shear-wave elastography (SWE) (Drakonaki et al., 2009; Nicholls et al., 2020). This technique allows measurement of the shear modulus by measuring the tissue displacement produced by shear waves, which are transverse waves generated in tissue by periodic displacement by the US waves produced from a linear US transducer. Shear-wave velocity can be calculated by measuring the tissue displacement, and the SWE modulus can be subsequently calculated as the ratio of shear stress to shear strain taking into considerations the material density for soft tissue that is approximately close to that of water ($1\,g/cm^3$) (Nicholls et al., 2020; Taljanovic et al., 2017). Using commercially available US scanners, color-coded quantitative SWE modulus maps (named "elastograms") are presented in real time on the US scanner screen side by side to the B-mode US images, displaying shear-wave velocities (in meters/second) or tissue elasticity (in kilopascals; kPa) and allowing the estimation of tissue stiffness in a chosen region of interest (ROI).

Shear modulus is associated with muscle elongation or force in lower extremity muscles (Drakonaki et al., 2009; Koo et al., 2014; Taljanovic et al., 2017). Using SWE on human cadavers, studies revealed that fascia can externally compress the muscles and, hence, increase their stiffness (Hatta et al., 2015; Koo et al., 2014). An in vivo experimental study has also reported that evaluation of TLF SWE modulus shows high reliability and that TLF modulus increases as the trunk flexion angle increases (Chen et al., 2020). Moreover, there is evidence that the SWE modulus of the posterior thigh muscles can be influenced by pelvic movement which, in turn, can affect TLF function (Nakamura et al., 2016) while isometric plantar flexion influences the stiffness of the paraspinal and gastrocnemius muscles, which implies that there is a connection between fascia and muscles of the "superficial back line" (Chen et al., 2022). To the best of our knowledge, the changes in lumbar fascia and posterior thigh fascia stiffness during passive knee stretches using SWE have not been previously examined.

Since fascia stiffness displays varying anatomy as it spreads along various tissues, muscles, and tendons (Otsuka et al., 2021; Willard et al., 2012), it would be interesting to compare stiffness properties between two distinct regions, the lumbar and the posterior thigh area. Experiments in canine tissue have shown that the fascia that encapsulates the posterior thigh muscles (fascia lata) is stiffer than TLF (Henderson et al., 2014) but it is unclear whether these results can be generalized to humans. Further, if myofascial continuity between then TLF and thigh fascia exists, then it could be hypothesized that knee flexion movement should influence both hamstring fascia and TLF function. Such mechanism may assist in explaining why therapeutic interventions, such as myofascial release or passive stretching exercises, which are applied on the posterior thigh area can influence lumbar myofascial function and vice versa (Fauris et al., 2021; Grieve et al., 2015; Joshi et al., 2018; Krause et al., 2016; Wilke et al., 2017). Furthermore, such mechanism could help explain why some athletes experience pain in the posterior thigh area (often termed "spine-related" injury), which has been attributed to a potential connection between hamstring muscle tendon injuries and spinal/lumbopelvic disorders (Fuller et al., 2007). Force transfer via myofascial chains takes place not only during passive joint movements but also during active muscle exercises or stiffening of the muscles (Krause et al., 2016). It has been shown that hamstring muscle stiffness is greater during active contractions compared to passive conditions while hamstring muscle or TLF SWE modulus increases as body position changes (Chen et al., 2021; Evangelidis et al., 2021; Mendes et al., 2018). This implies that therapeutic modalities which involve dynamic muscle contractions at different joint angles may also result in force transmission to adjacent fascia tissues and, hence, influence function of adjacent and distant body areas. To the best of our knowledge, the influence of changes in knee angle and knee flexion contractions on stiffness of the posterior thigh fascia and TLF has not been previously examined.

The aim of this study was to investigate the changes in stiffness of TLF, semitendinosus (STF), and SM fascia (SMF) using SWE during passive and active knee flexion. It was hypothesized that passive changes in knee angle as well as active knee flexion contractions would influence stiffness of the TLF, STF, and SMF. Further, we expected that stiffness would differ between the three types of fasciae.

2 | METHODS

2.1 | Participants

A priori power analysis using the G power (version 3.1.9.4, Heinrich-Heine-Universität Düsseldorf) was performed to determine the minimum sample size required to achieve acceptable statistical power. For a repeated measures design which compared the effect of within-subject variables on each dependent measurement, a minimum sample of 12 individuals was predicted using the following as inputs: power = 0.80, effect size = 0.30, a type 1 error = 5%, and correlation among repeated measurements = 0.30. For sample recruitment, we contacted a total of 93 amateur soccer players who played in the regional soccer amateur league (six amateur clubs). To be included in this study, the participants should have had no history of musculoskeletal low back and lower limb injuries or surgery, neurological or soft tissue inflammatory disease, drug intake or muscle soreness and they would have reported no episodes of low back pain for the past year. Sixty-four players responded positively and agreed to participate. Of these, 26 athletes did not meet the inclusion criteria and 15 withdrew prior to the first measurement session. The remaining nine athletes participated in the preliminary screening session, but they did not conclude all measurement sessions, so they were excluded from the study. Hence, a total of 14 healthy active males (age: 23.7 ± 7.31 years; mass 78.9 ± 8.02 kg; height 181 ± 9.71 cm) participated in this study voluntarily. The participants gave their informed written consent, and the protocol was approved by the Institutional Ethics Committee (ERC/018/2022) in accordance with the Helsinki Declaration.

2.2 | Instrumentation

US and SWE was performed using a GE system (LOGIQ E9, R5 version, Chicago, USA). A ML6-15 (4–15 MHz) linear array transducer was used for B-mode imaging and fascia thickness measures. SWE was performed using a 9 L (2–8 MHz) 2D linear transducer utilizing the "comb brush" excitation technique that transmits multiple focused acoustic radiation force impulses simultaneously (Song et al., 2014). Shear-wave velocity (in m/s) is measured using time-interleaved shear wave tracking and the system built-in software automatically calculates the Young's elastic modulus in kilopascals (kPa), using the equation $E = \rho V^2$, where E is Young's modulus, ρ is tissue density (assumed to be 1 g/cm^3), and V is the velocity of shear waves (Drakonaki, 2012).

2.3 | Experimental protocol

All examinations were performed with the volunteer in the prone position with the hip in neutral position and the hands lying next to the body. An elastic strap was used to maintain pelvic position.

2.4 | Measurement of maximum isometric strength

A warm-up consisting of static stretches of hamstrings followed by three to four submaximal trials at 0° (=full extension), 45°, and 90° knee flexion angles were performed. Maximum voluntary contractions (MVCs) of knee flexors were then performed in each knee flexion angle. Participants were asked to exert maximum knee flexion contractions with resistance which was provided by a hand-held force sensor (K-Force muscle controller, sampling rate 75 Hz, Kinvent Biomecanique, Montpellier, France). The experimenter-held device was placed 4–5 cm above the lateral malleolus. We then measured the distance between the lateral knee epicondyle and the force sensor position (lever arm). Subsequently, the registered force was multiplied by the lever arm to estimate knee joint torque (in N m). In each joint position, the participant was requested to perform three MVC efforts of the knee flexors, of 5 s each. Each individual performed three trials and the maximum value was considered as the MVC torque.

2.5 | US and SWE imaging protocol

Before starting any of the measurements, the deep fascia was preliminary identified using B-mode US as the echogenic striated linear structure lying deep to the subcutaneous layer and just superficially to the muscle of interest (Wilke & Tenberg, 2020). The experimenter adjusted the zoom of the device such that the clear images of the fascia appear in the field of view. All measurements were taken from the deep fascia at three independent locations: lumbar, posterior thigh SM muscle, and posterior thigh ST muscle. The precise locations were first identified on the left body side and marked on the skin. For the TLF, the fourth lumbar (L4) vertebra spinal process was identified, and the transduce was then shifted horizontally to the left side at 2.5 cm away from the L3 spinous process at the L3–L4 level (Figure 1). For the STF, the probe was positioned in the middle of the ST muscle belly, approximately at 40% of the distance between the medial condyle and the ischial tuberosity. The SMF was identified more medially than STF, approximately at 40% of the SM length, defined as the distance between the medial condyle and the ischial tuberosity. In each location, the probe was first positioned axially to identify the respective anatomical level and then turned longitudinally to visualize the fascia and enable SWE measurements. US gel was applied liberally to the areas of imaging to ensure good sonic coupling between the transducer and skin.

The maximum thickness of the fascia was first measured in axial plane using the caliper tool available by the system built-in software in the zoom setting to allow precise caliper placement. Three measurements were acquired to calculate the mean value of fascia thickness.

SWE measurements were acquired with the participant at rest from knee flexion angles of 0°, 45°, and 90° (passive condition). An experimenter held the knee of the participant at each joint angle for a period of 5 s during SWE acquisition and until the elastogram appeared on the screen. Measurements took place separately for each probe location (TLF, SMF, and STF) with a 10-min rest period between them. The order of tests was randomized across subjects.

Following the passive test condition, SWE measurements were taken while the participant performed submaximal contractions

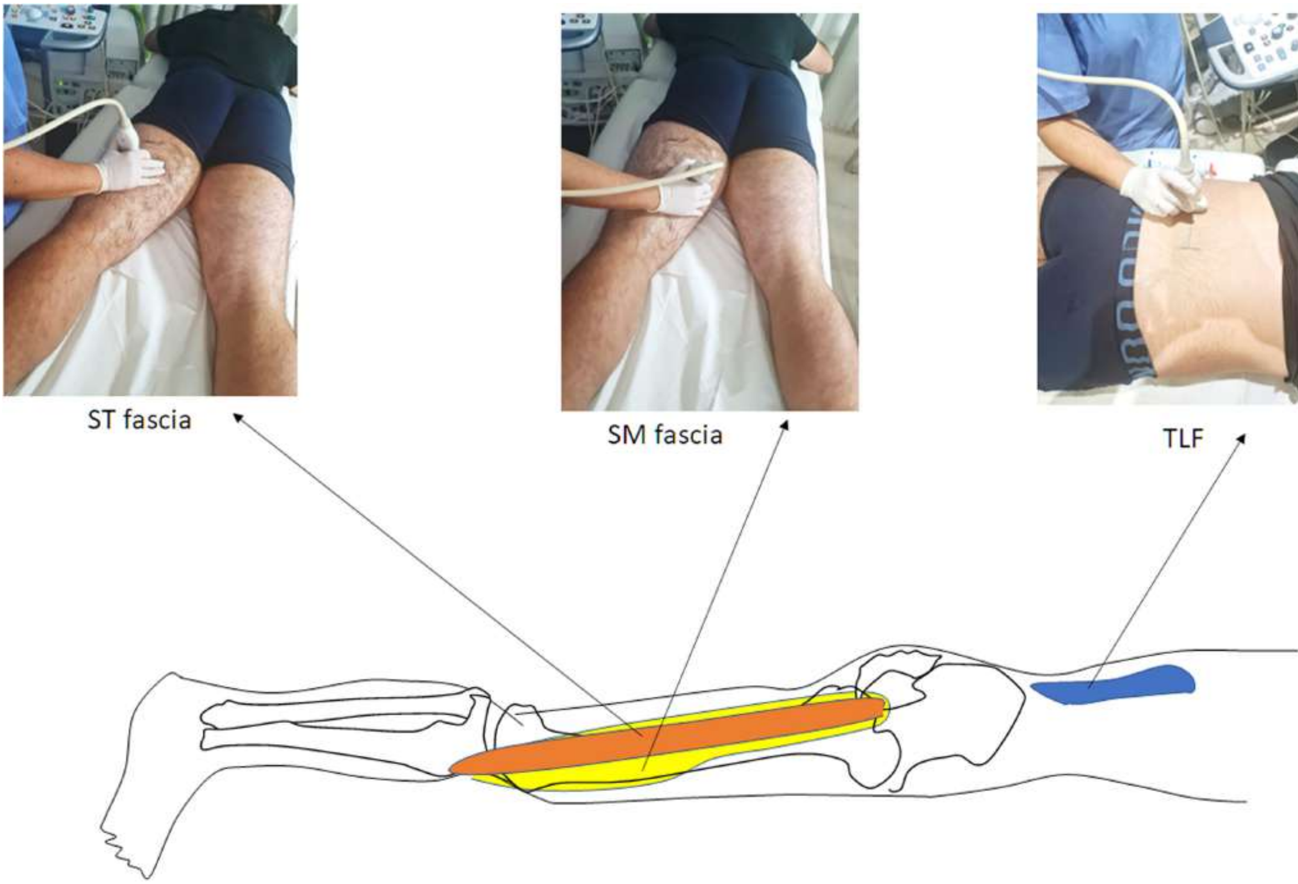

FIGURE 1 Experimental setup. The participant assumed the prone position and the knee was passively moved at three knee joint positions. The ultrasound probe was placed 2.5 cm away from the L3 spinous process at the L3–L4 level to visualize the thoracolumbar fascia. The corresponding position for the semitendinosus fascia was in the middle of the muscle belly, approximately at 40% of the distance between the medial condyle and the ischial tuberosity. The SM fascia was identified more medially than ST fascia. SM, semimembranosus fascia; ST, semitendinosus fascia; TLF, thoracolumbar fascia.

(70% MVC) at each angle against resistance provided by a tester using the handheld dynamometer (active condition). The participant was asked to gradually exert more force (for about 2 s) and then maintain the target force for approximately 5 s. Three trials per condition were recorded.

The SWE modulus was measured using the US system default standard musculoskeletal setting (Figure 2). The SWE modulus of the tissues was presented as a rectangular color-coded box over-imposed on each B-mode image (elastogram) including the subcutaneous fat, the fascia, and the superficial area of the muscle (about 1cm deeper than the superficial fascia). The color spectrum ranged from blue (very low shear modulus) to green (low) and yellow (intermediate) to red (high). The SWE modulus was measured by applying manually selected circular ROIs on each fascia. The ROIs were determined by the same examiner to include only the hyperechoic fascia layer, excluding subcutaneous fat and muscle. The mean SWE modulus values (in kPa) in each ROI and the mean values of the three ROIs acquired in each of the three trials were calculated. Higher SWE modulus indicates a stiffer tissue. All examinations were performed by the same radiologist with >18 years of experience in US and US elastography.

2.5.1 | Statistical analysis

Normal distribution of the raw SWE modulus data was confirmed using Kolmogorov–Smirnov tests. A three-way analysis of variance (ANOVA) was used to examine the effect of joint angular position (3 angles), condition (passive, active contraction), and type of fascia (TLF, SMF, and STF) on SWE modulus. A one-way ANOVA was used to examine the difference in thickness between the three types of fasciae. Significant main or interaction effects were followed by post hoc Tukey test comparisons between pairs of means. Statistical significance was set at $p < 0.05$.

3 | RESULTS

Table 1 presents the SWE modulus values for each condition. The ANOVA indicated a non-significant three-way interaction effect on SWE modulus ($p > 0.05$). There was a statistically significant condition X angular position effect ($p = 0.005$). Post hoc Tukey tests showed that during the passive condition, SWE modulus (averaged for all tissues) was greater at 0° than at 45° and 90° ($p > 0.05$). In

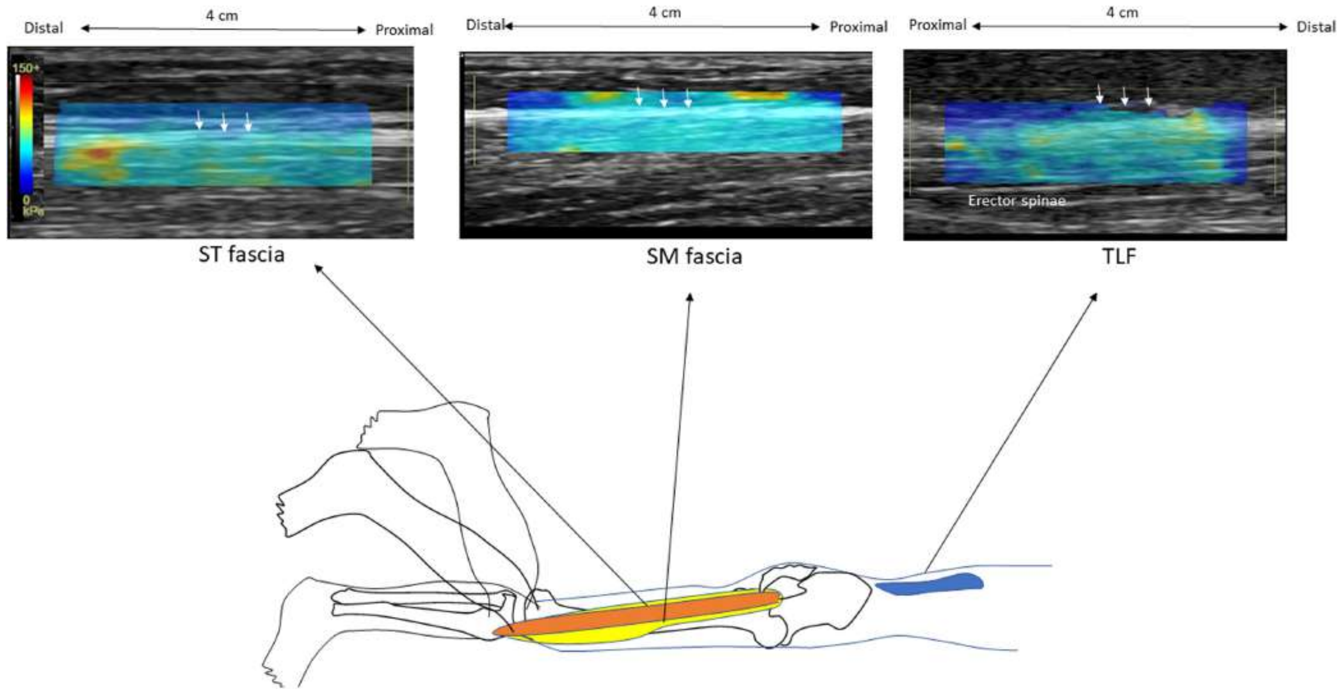

FIGURE 2 Example elastograms from the semitendinosus, the semimembranosus, and thoracolumbar fasciae. First, elastographic data were obtained using a rectangular color-coded box to isolate the measurement area. Then, circles of smaller diameter were drawn along each fascia (positions are indicated with vertical arrows) and the software provided the shear modulus. The color scale was extracted from the software and is enlarged so that the measurement scale is easily visible. SM, semimembranosus fascia; ST, semitendinosus fascia; TLF, thoracolumbar fascia.

TABLE 1 Mean (±SD) shear-wave modulus of the thoracolumbar fascia (TLF), semimembranosus fascia (SM), and semitendinosus fascia (ST) during different exercise durations.

Knee angle	TFL	STF	SMF
0	3.51 ± 1.73*	5.57 ± 1.94	6.64 ± 2.48
45	2.70 ± 1.41*	4.33 ± 1.59	4.31 ± 1.59
90	2.29 ± 1.03*†	3.63 ± 1.27†	3.59 ± 1.29†
	Contraction		
0	8.43 ± 3.73*	16.16 ± 7.84	14.16 ± 5.58
45	10.27 ± 4.66*	14.31 ± 4.84	13.00 ± 4.35
90	10.88 ± 5.73*	16.21 ± 7.23	15.30 ± 5.39

*Significantly different compared to STF and SMF, $p < 0.05$.
†Significantly different compared to 0°, $p < 0.05$.

contrast, during contraction, no differences in SWE modulus between angles were observed. Furthermore, the condition X tissue effect was also statistically significant ($p = 0.03$). Post hoc analysis showed that TLF modulus was lower than STF and SMF modulus at all conditions (Figure 3, $p < 0.05$). During the passive condition, no differences between STF and SMF modulus were observed. In contrast, during the active condition, STF modulus was greater than that of SMF ($p < 0.05$). All fasciae SWE modulus values were lower at rest compared to contraction ($p < 0.01$).

Mean group transverse thickness values of TLF, SMF, and STF are presented in Figure 4. The ANOVA showed that there was a significant difference in thickness between the three types of fasciae ($p = 0.02$). Post hoc Tukey tests showed that TLF thickness was greater than SMF and STF thickness ($p < 0.05$). No difference between SMF and STF was observed ($p > 0.05$).

4 | DISCUSSION

The main findings of this study were first, that TLF, SMF, and STF SWE modulus increased from more flexed to more extended knee joint angles, second, that fasciae modulus increased significantly during submaximal contractions and, finally, there were differences in SWE modulus between the TLF, SMF, and STF. To the best of our knowledge, changes in stiffness of TLF and hamstrings' fascia as a function of changes in knee flexion angle and contraction have not been previously examined.

The results support the hypothesis that lengthening (by extending the knee joint) and active isometric contraction of the hamstrings influences stiffness of the fasciae that surround the posterior thigh and lumbar muscles (Table 1). To the best of our knowledge, the simultaneous changes in fasciae stiffness of the lumbar and the posterior thigh areas have not been previously examined. Hence, it is difficult to compare our results directly with the results of previous studies. A mechanism that is responsible for the present results is the existence of a mechanical path, which allows force transmission from the posterior thigh fascia to the TLF (Krause et al., 2016; Myers, 1997). Two types of evidence support this mechanism. First, there are anatomical connections

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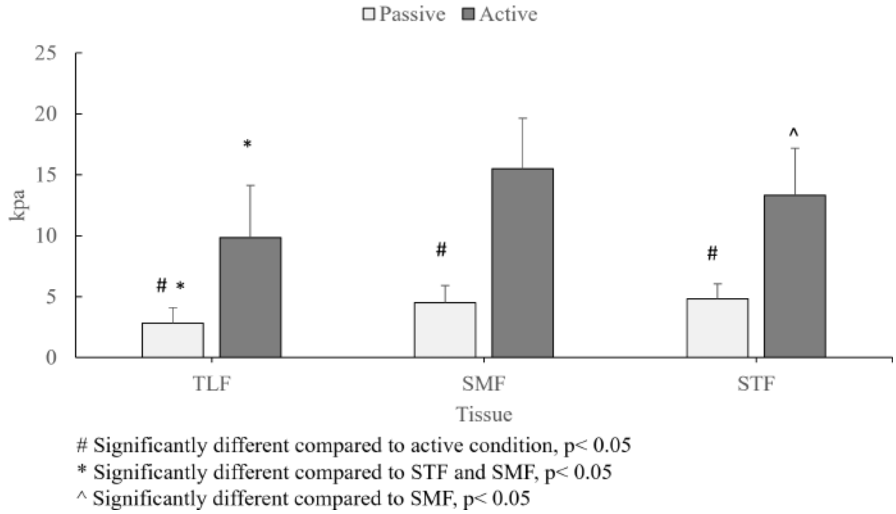

FIGURE 3 Mean group shear modulus values of the thoracolumbar (TLF), semimembranosus (SMF) and semitendinosus (STF) fascia at rest and contraction. Bars indicate average values from all angular positions. Error bars indicate standard deviation. #Significantly different compared to active condition, $p<0.05$. *Significantly different compared to STF and SMF, $p<0.05$. ^Significantly different compared to SMF, $p<0.05$.

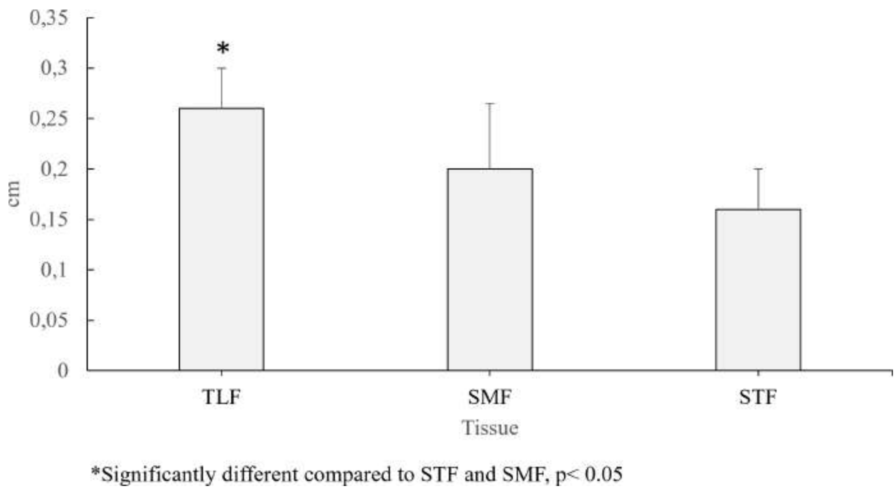

FIGURE 4 Mean group transverse thickness values of the thoracolumbar (TLF), semimembranosus (SMF) and semitendinosus (STF) fascia which were recorded while the participant relaxed in the prone position with the knee in full extension (error bars indicate standard deviation). *Significantly different compared to STF and SMF, $p<0.05$.

between tissues which are arranged in series along the posterior surface of the body (Bierry et al., 2014; Cruz-Montecinos et al., 2015; Vleeming et al., 1995). In particular, the TLF maintains linkages with the glutei muscles and the common tendon of biceps femoris and ST via the sacrotuberous ligament (Bierry et al., 2014; Vleeming et al., 1995). In addition, anchor points that connect the sacrotuberous ligament, the posterior thigh fasciae and muscles up to the triceps surae and the plantar fascia have been identified (Cruz-Montecinos et al., 2015). Second, previous studies have found that modification of myofascial tissue properties through manipulation or exercise which is applied at a specific location can remotely affect the properties of distant myofascial structures along the superficial back line (Cruz-Montecinos et al., 2015; Vleeming et al., 1995; Wilke et al., 2020; Wilke & Tenberg, 2020). In particular, Vleeming et al. (1995) observed that application of pulling force to the common (proximal) tendon between ST and biceps femoris resulted in displacement of the deep lamina of the TLF up to the level of L5–S1. Another study reported that pelvic movement causes displacement of the gastrocnemius fascia (Cruz-Montecinos et al., 2015). The authors suggested that the pelvic muscle contraction or stretch transmits forces to the posterior thigh fasciae which, in turn, are transmitted to the gastrocnemius fascia (Cruz-Montecinos et al., 2015). Finally, a recent study found a significant association between ankle dorsiflexion and caudal displacement of the SM muscle and encapsulated fascia (Wilke et al., 2020). Another mechanism that may have contributed to these findings is a possible increase in neural activation of the lumbar muscles during hamstring exercise which, in turn, can increase TLF stiffness. This increase may be due to (mild) isometric contractions of the lumbar musculature which might have taken place during active contraction of the hamstrings. Unfortunately, our experimental protocol cannot verify this suggestion.

Even though the ST and SM muscles have common actions about the hip and the knee, they do not have the same attachments and architecture (Kellis et al., 2012). ST muscle shares a common proximal tendon with biceps femoris, and they are directly connected to TLF via the sacrotuberous ligament, the SM inserts proximally to the ischial tuberosity (Kellis et al., 2012). Since the SMF does not have a direct connection with the sacrotuberous ligament, one would expect that a change in SM stiffness in response to passive and active exercise conditions would not influence the stiffness of TLF. Huijing et al., (2007) found that during passive or active contractions of a muscle group, force transmission takes place between synergists within this muscle group. Hence, we can hypothesize that changes in stiffness of all hamstring components (not just the long head of the biceps femoris) could influence paraspinal muscle and fascia stiffness.

Very few studies report SWE modulus values of the TLF in healthy individuals (Chen et al., 2020, 2021, 2022). The present TLF SWE modulus values at rest were much lower than those reported in a series of studies by Chen et al. (2020, 2021, 2022) (~40 kPa). This is because we did not multiply the shear-wave velocity by a factor of 3, while Chen et al. (2020, 2021, 2022) used the values obtained by the device. Further, Chen et al. (2020, 2021, 2022) quantified stiffness by placing the probe axially relative to the long axis of the body, whereas we acquired images in the longitudinal orientation with the probe along the long axis of the body. It is known that elastography measures largely depend on the orientation of the probe relative to the fibers of the underlying tissue (Gennisson et al., 2010). Further, lumbar fascia does not have constant morphology throughout its course (Willard et al., 2012). Hence, rotating the probe by 90° may change shear-wave transmission not only because the underlying tissue has a different orientation relative to the probe but also because it has different thickness and attachments to adjacent tissues.

The results also showed that the SWE modulus differed between the hamstrings and the paraspinal muscles as well as between the hamstring muscles themselves (Table 1). Particularly, the SWE modulus values of SMF and STF were greater than TLF modulus values in the prone-relaxed position (Table 1, Figure 3). No previous studies could be identified that compared stiffness of these fasciae in vivo. One may suggest that in the prone position, the paraspinal muscles and the TLF are almost relaxed (Blain et al., 2019) while the hamstrings are lengthened (Kellis & Blazevich, 2022) which may explain the difference in stiffness between those tissues. Nevertheless, even in the most shortened hamstring position (knee angle of 90°), TLF stiffness is approximately 30% lower than the corresponding STF and SMF stiffness values (Table 1). Hence, one cannot exclude the possibility that morphological differences between various fasciae can lead to corresponding differences in stiffness. This is supported by experimental measurements in canine tissue which indicated that the elastic modulus and maximum load of fascial lata are greater than the corresponding values of the TLF (Henderson et al., 2014). These previously reported (Henderson et al., 2014) measurements, however, were performed in animal tissue, which was removed and tested and, hence, they are not directly comparable to in vivo measurements in humans. Nevertheless, these findings support that the fascia that surrounds the trunk area (TLF) has different stiffness properties than the fascia that surrounds the thigh muscles. Speculatively, a factor that may have contributed to this result may be that the TLF is subjected to forces from various directions (Vleeming et al., 1995) while the SMF and STF may be subject to forces in the longitudinal and transverse direction (Otsuka et al., 2021; Henderson et al., 2014).

Being less stiff, we would expect that the TLF shows lower thickness than the SMF and STF thickness; however, the results showed the opposite (Figure 4). The reason for this finding is unclear. One factor may be that SWE modulus was assessed in the longitudinal direction while thickness was quantified in the transverse direction. Since tissues develop stiffness in various directions, it is possible that a greater thickness in the transverse direction does not necessarily indicate a greater stiffness in the longitudinal direction. Further research could examine fascia stiffness by measuring SWE modulus in more than one location for each tissue and at different body positions.

In the present study, SMF and STF showed similar stiffness in the passive condition but there was a greater STF thickness during submaximal knee flexion contraction (Table 1). These differences were not influenced by joint angular position. To the best of our knowledge, there are no studies that examined stiffness of the fascia that surrounds the human hamstring muscles in vivo. In a recent study, Otsuka et al. (2021) reported a positive correlation between the deep STF thickness, ST (muscle) thickness, and knee flexion torque. This was attributed to the existence of linkages between the ST tendons and fascia lata, which allow transmission of the mechanical stress produced by the muscle to the overlying fascia lata (Otsuka et al., 2021). It has been suggested that the morphological properties of the deep fascia reflect the morphologies and functions of the neighboring tissues (e.g., subcutaneous adipose tissues and muscles) (Benjamin, 2009). Previous studies have shown that during passive stretch of the hamstrings, the SM muscle modulus is greater than that of ST (Berrigan et al., 2020; Ichihashi et al., 2016; Miyamoto et al., 2020; Nakamura et al., 2016; Nakao et al., 2018; Umegaki et al., 2015). During active contractions, however, ST muscle stiffness exceeds the corresponding SM stiffness (Evangelidis et al., 2021; Mendes et al., 2018). Hence, it seems that the inter-muscular differences in stiffness during contraction may partly explain the differences in stiffness of the fascia that surrounds these muscles.

There are some implications of this study. First, tension in the posterior thigh fascia area caused an increase in tension of the lumbar area fascia. Hence, individuals with low back myofascial disorders may experience pain in the posterior thigh area. This fits well with a previous consensus statement on hamstring muscle injuries according to which some athletes feel pain in the hamstring region which originates from the lumbar—pelvis area (spine-related injuries) (Fuller et al., 2007). Simulations of running movement have also shown that contraction of the erector spinae muscle is associated with stretch of the hamstring muscles (Thelen et al., 2006). Clinicians should examine the role of spine, pelvis, and sacrum in individuals who suffer from hamstring injuries. Further, the connection between thigh fascia and TLF modulus implies that individuals with increased tension of the hamstring myofascial tissues may also experience increased tension and, perhaps, pain in the low back region, as it has been previously suggested (Wilke et al., 2017). Second, our results support previous suggestions that therapeutic interventions that are applied in one body area can influence the mobility of neighboring and even more distant joints (Grieve et al., 2015; Joshi et al., 2018; Krause et al., 2016). Clinically, this means that individuals with low back disorders may see some benefits from therapeutic interventions that involve hamstring fascia manipulation (Wilke et al., 2017). For the same reason, TLF manipulation techniques may assist athletes to overcome pain problems in the posterior thigh area. Fauris et al. (2021) for example, have found an increase in hamstring flexibility following self-myofascial release application to the lumbar and the plantar fascia area. Third, exercise programs that include dynamic contractions of the hamstrings can increase TLF stiffness. Hence, such exercises could be an optional approach when an increase in lumbar myofascial stiffness is necessary, especially when trunk mobility is restricted (e.g., in cases of surgery). Alternatively, stretching hamstring exercises could increase hamstring flexibility and, through that, may lower TLF stiffness, when necessary. There is evidence, for example, that back extensor length has a positive association with hip extensor strength in individuals with low back pain (Arab et al., 2019).

This study has some limitations. First, the present findings are applicable to asymptomatic physically active healthy young males. Older people or individuals with low back pain symptoms or lower fitness levels may show different muscle function. Second, SWE modulus measurements depend on the measurement location (Drakonaki, 2012; Miyamoto et al., 2020). It is unclear whether similar results would be obtained from a different portion of TLF, SMF, and STF. Third, it is possible that changes in fascia stiffness is due to stretch or contraction of other muscles or joint movements. For example, it has been found that gluteus maximus contraction stretches the superficial lamina of the TLF (Vleeming et al., 1995) while interactions between synergists (especially the biceps femoris long head) might impact SMF, STF, and TLF stiffness. Pelvic movement has also been demonstrated to cause activation of paraspinal muscles (Takaki et al., 2016), and, speculatively, TLF stiffness. Despite our close attention in the present study to avoid pelvic tilt movement, its possible influence on fascia modulus measurement cannot be neglected. Finally, even though US measurements were taken using longitudinal US scans, the possibility of an angulation between the US probe and fascia exists, which may have influenced the SWE modulus measurements.

5 | CONCLUSION

The results showed considerable increase in TLF SWE modulus during passive knee flexion and active contractions of the knee flexors. SWE modulus during the relaxing prone position was lower in TLF than SMF and STF. No differences between STF and SMF modulus values were observed during passive stretch, but STF modulus was greater than that of SMF during active contraction. These results indicate that exercising the hamstring muscles can remotely influence the stiffness of the fascia which surrounds the lumbar area.

AUTHOR CONTRIBUTIONS

Eleftherios Kellis wrote the manuscript; conceived and designed the research. Eleftherios Kellis, Eleni E. Drakonaki, and Afxentios Kekelekis conducted experiments. Eleni E. Drakonaki analyzed the elastography data and Afxentios Kekelekis analyzed the force data. All authors read and approved the manuscript.

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CONFLICT OF INTEREST STATEMENT

The authors declare no conflict of interest.

DATA AVAILABILITY STATEMENT

The data that support the findings of this study are openly available in Mendeley V1, at https://doi.org/10.17632/mgjfhmkv7b.1.

ETHICS STATEMENT

This study was performed in line with the principles of the Declaration of Helsinki. Approval was granted by the Ethics Committee of University B (July 18, 2022/No. ERC018/2022).

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