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Elastography Based Assessment of Myofascial Stiffness of the Lower Extremities in Healthy Adolescent Soccer Players

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Abstract

Background:
Sport-related muscle injuries are common and often caused by overuse. These injuries can develop without notice until symptoms suddenly arise, including pain, muscle stiffness, early fatigue, and cramping. Therapists may find reduced joint range of motion and increased stiffness in muscles and fascias related to myogeloses or myofascial trigger points. Reducing stiffness through manual or physical approaches is a common technique for reducing the risk of overuse injuries, despite a lack of scientific evidence to support this practice. Ultrasound elastography is an emerging tool in sports medicine that can measure stiffness qualitatively, semi-quantitatively, and quantitatively, and can be used to monitor the efficacy of therapy and possibly determine if stiffness changes are related to overuse injuries.
Objectives:
This study aimed to identify intraindividual stiffness differences within the lower limb muscles and fascias of young, competitive soccer players with shear wave elastography and strain elastography and correlate these findings with range of motion, static alignment, and pedobarography. The study hypothesized that soccer players would show intraindividual stiffness differences with higher stiffness values in the dominant leg and that decreased range of motion, compromised static alignment, and pedobarography would correlate with increased stiffness and side differences.
Material and Methods:
20 male soccer players (age 14.6 $\pm$ 0.5 years, BMI 20.8 $\pm$ 1.4 $\frac{kg}{m^2}$) participated in this study. Large area shear wave elastography and strain elastography were used to measure stiffness at 26 points in 22 lower limb muscles and fascias. In addition, range of motion, postural assessments, and pedobarography assessments were conducted.
Results:
Six measurement points showed intraindividual differences ΔE, ranging from 6 to 40 kPa while no significant difference was found between the dominant and nondominant leg. Strain and shear wave elastography revealed limited comparability. Range of motion was decreased in the low back, the hip and the ankle without any leg dominance. The heel-buttock distance was equally increased in both legs. Posture was normal, but forefoot and rearfoot pressure distribution was shifted towards the forefoot without a preference for any leg. No significant correlation was found between myofascial stiffness, range of motion, static alignment, and pedobarography.
Conclusion:
Large area shear wave elastography showed intraindividual stiffness differences in six muscles/fascias of the lower limbs. These differences, combined with abnormal range of motion and foot pressure distribution, may suggest the presence of myofascial dysbalances that could increase the likelihood of overuse injuries. How-

ever, there was no significant correlation between myofascial stiffness and range of motion, static alignment, or pedobarography. Further research is necessary to determine whether stiffness assessment using shear wave elastography could serve as a marker for predicting the risk of overuse injuries in future studies.

Zusammenfassung

Hintergrund:
Sportbedingte Muskelverletzungen sind häufig und oft durch Überlastung verursacht. Diese Verletzungen können sich unbemerkt entwickeln, bis plötzlich Symptome wie Schmerzen, Muskelsteifigkeit, frühe Ermüdung und Krämpfe auftreten. Therapeuten können eine reduzierte Gelenkbeweglichkeit und erhöhte Steifigkeit in Muskeln und Faszien im Zusammenhang mit Myogelosen oder myofaszialen Triggerpunkten feststellen. Eine Reduktion der Verhärtungen durch manuelle oder physikalische Ansätze ist ein häufiger Ansatz zur Verringerung des Risikos von Überlastungsverletzungen, jedoch gibt es wenig Evidenz hierfür. Als neuartige, objektive und quantifizierbare Methode zur Steifigkeitsbestimmung rückt die Ultraschall-Elastografie zunehmend in den Fokus der Sportmedizin. Sie ermöglicht es, die Gewebesteifigkeit qualitativ, semi-quantitativ und quantitativ zu messen. Ferner kann sie und zur Überwachung der Wirksamkeit von Therapien und zur Bestimmung von Steifigkeitsveränderungen im Zusammenhang mit Überlastungsverletzungen verwendet werden.
Ziele:
Diese Studie hatte zum Ziel, intraindividuelle Steifigkeitsunterschiede innerhalb der Muskulatur der unteren Extremitäten junger Leistungs-Fußballspieler mit Scherwellen-Elastografie und Strain-Elastografie zu identifizieren und diese Befunde mit der Gelenkbeweglichkeit, Körperstatik und Pedobarographie zu korrelieren. Es wurde angenommen, dass Fußballspieler intraindividuelle Steifigkeitsunterschiede mit höherer Steifigkeit im dominanten Bein zeigen. Ferner, dass sich eine eingeschränkte Beweglichkeit sowie Abweichungen in der Körperstatik und Pedobarographie zeigen, welche mit einer erhöhten Steifigkeit und intraindividuellen Seitendifferenzen korrelieren.
Material und Methoden:
20 männliche Fußballspieler (Alter 14.6 $\pm$ 0.5 Jahre, BMI 20.8 $\pm$ 1.4 $\frac{kg}{m^2}$) nahmen an dieser Studie teil. Scherwellen Elastografie und Strain Elastografie wurden an 26 Messpunkten in 22 Muskeln und ihren Faszien der unteren Extremitäten angewandt. Ferner wurden Beweglichkeitstestungen, Körperstatik und Pedobarographie Messungen erhoben.
Ergebnisse:
Sechs Messpunkte zeigten intraindividuelle Unterschiede ΔE, von 6 bis 40 kPa reichend, während kein signifikanter Unterschied zwischen dem dominanten und nicht-dominanten Bein festgestellt wurde. Die Strain und Scherwellen Elastografie zeigten eine begrenzte Vergleichbarkeit. Die Gelenkbeweglichkeit war im unteren Rücken, der Hüfte und dem Sprunggelenk reduziert, ohne Seitendominanz. Der Ferse-Gesäß-Abstand war in beiden Beinen erhöht. Die Körperstatik war normal, jedoch zeigte sich die Vorfuß-Rückfuß Druckverteilung in Richtung des Vorfußes verschoben, ohne

Seitendominanz. Es wurde keine signifikante Korrelation zwischen myofaszialer Steifigkeit, Gelenkbeweglichkeit, Körperstatik und Pedobarographie gefunden.
Schlussfolgerung:
Die Scherwellen-Elastografie zeigte intraindividuelle Steifigkeitsunterschiede in sechs Muskeln/Faszien der unteren Extremitäten. Diese Unterschiede, kombiniert mit abnormaler Gelenkbeweglichkeit und Fußdruckverteilung, könnten auf das Vorhandensein von myofaszialen Dysbalancen hinweisen, die das Risiko für Überlastungsverletzungen erhöhen könnten. Es ließ sich jedoch keine Korrelation zwischen myofaszialer Steifigkeit und Gelenkbeweglichkeit, Körperstatik oder Pedobarographie nachweisen. Es sind weitere prospektive Studien mit Scherwellen-Elastografie notwendig, um die Rolle von myofaszialer Steifigkeit als möglichen Marker für das Risiko von Überlastungsverletzungen zu untersuchen.

Contents

Acronyms

ADD	Adductor muscles
ADF_a	Active ankle dorsiflexion
ADF_p	Passive ankle dorsiflexion
BF	Biceps femoris muscle
BMI	Body mass index
BROM	Back range of motion
FFD	Finger-floor distance
FIFA	Féderation Internationale de Football Association
GL	Lateral head of gastrocnemius muscle
GM	Medial head of gastrocnemius muscle
GMAX	Gluteus maximus muscle
GMED	Gluteus medius muscle
HBD	Heel-buttock distance
HIP_{ER}	External rotation of the hip
HIP_{IR}	Internal rotation of the hip
ICC	Intraclass correlation coefficient
IQR	Interquartile range
LA-SWE	Large area shear wave elastography
LLF	Lumbar lateral flexion
LRO	Lumbar rotation
MRI	Magnetic resonance imaging
MTrPs	Myofascial trigger points
OIs	Overuse injuries
PF	Frontal position of the pelvis
PL	Peroneus longus muscle
RF	Rectus femoris muscle
RFdi	Distal rectus femoris muscle
RFpr	Proximal rectus femoris muscle
ROI	Region of interest
ROM	Range of motion
SD	Standard deviation
SE	Strain elastography
SLR	Straight leg raise test
SR	Strain ratio
SWE	Shear wave elastography
SWR	Shear wave ratio
TA	Tibialis anterior muscle
TFL	Tensor fasciae latae muscle

UE	Ultrasound elastography
VL	Vastus lateralis muscle
VM	Vastus medialis muscle
VP	Frontal position of vertebra prominens

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1 Introduction

1.1 Injuries in Soccer Players

Overuse injuries (OIs) in sports are an ever-present complication which cause an average time loss of four weeks for affected professional football athletes during the season (Ekstrand et al., 2020). OIs can occur when the load-limit of the myofascia is exceeded, which can affect any athlete. Traditional clinical measures to detect potential overload of the musculoskeletal system include joint range of motion, pedobarography, and static alignment. These measures can deviate from normal ranges due to joint pathologies, central nervous system issues, or structural abnormalities like myofascial trigger points (MTrPs) (Bagcier, Yurdakul, Üşen, & Bozdag, 2022). MTrPs in muscles and fascias can cause a disbalance between synergistic and antagonistic muscles, leading to the mentioned abnormalities in clinical tests (Watson, 1995). Currently, several risk factors for overuse injuries in athletes have been identified, including previous injury, muscle imbalances, muscle weakness, and poor flexibility (Owen et al., 2013), but no specific marker has been defined. Even though the Féderation Internationale de Football Association (FIFA) has launched a warm-up program that has been shown to reduce the rate of OIs by up to thirty percent, it is rarely used in professional football (Bizzini, Junge, & Dvorak, 2013; Finch, 2006). OIs are usually addressed reactively when pain or dysfunction arises, rather than proactively through prevention measures. Rehabilitation of injured players focuses on reducing myofascial stiffness, which plays a role in OIs (Owen et al., 2015), but the pathophysiology leading to microtrauma is not fully understood. Wilke and colleagues (2019) suggest that altered myofascial force transmission may play a role, while another perspective regards OIs as a result of myofascial pain syndrome due to MTrPs. MTrPs are defined as hypersensitive palpable nodules in a taut band of skeletal muscles and are associated with a hyperirritable spot. In healthy individuals, latent MTrPs are found, while symptomatic individuals exhibit both latent and active MTrPs. Diagnosis of MTrPs used to rely solely on manual palpation using criteria such as a tender spot within a taut band, referred pain, and a local twitch response (Simons, Travell, & Simons, 1999). An animal study of Mense et al. (2003) found that repetitive peripheral nerve stimulation led to the development of sarcomere contractures and muscle fiber tears in rats, suggesting that MTrPs may result from overuse of myofascial structures. According to Bagcier et al. (2022), latent MTrPs that do not produce pain can still be harmful as they can reduce muscle strength and lead to muscle imbalances. Ge et al. (2014; 2012) found that these MTrPs are associated with increased intramuscular electromyographic activity during synergistic muscle activation and accelerated muscle fatiguability. Therefore, the identification and treatment of latent MTrPs could be crucial in injury prevention. Typically, manual palpation is used to identify MTrPs before various manual therapies or physical modalities are applied to eliminate them (Ramon, Gleitz, Hernandez, & Romero, 2015). However, palpation is subjective and highly dependent on the examiner's experience (Hsieh et al., 2000; Bron, Franssen, Wensing, & Oostendorp, 2007). A more objective approach is the use of ultrasound elastography (UE) (Ophir, Cespedes, Ponnekanti, Yazdi, & Li, 1991; Haser et al., 2017) or Magnetic Resonance Elastography, which can image and quantify regional stiffness such as MTrPs in the fascia and muscle (Chen et al., 2016). UE can be regarded as an extension of manual palpation, providing stiffness quantification that is not available through manual palpation. The use of UE is becoming increasingly popular among pain specialists and medical professionals to identify stiff areas such as MTrPs or stiff fascias and to monitor the effectiveness of therapeutic interventions (Haser et al., 2017; Bauermeister, 2015; Bauermeister & Raßmann, 2017). While Magnetic Resonance Elastography is primarily a research tool, UE is readily available in conventional ultrasound equipment and is used by several elite soccer clubs to objectively locate stiff areas and monitor changes in stiffness resulting from various therapeutic modalities (Haser et al., 2017). UE can be used to determine the appropriate treatment modality for reducing myofascial stiffness. In a recent study, UE was applied to monitor the effects of physical exercise in a special police force, and it showed significant stiffness changes in the thoracolumbar fascia after following a structured mobility routine (Slomka, 2022). Currently, there is a lack of prospective studies correlating the extent and number of sites with increased stiffness to physical load. This study aims to establish a foundation for future research to examine the role of MTrPs and regional myofascial stiffness as potential markers for evaluating an individual's propensity for OIs.

1.2 Ultrasound Elastography

UE was first described in 1991 as a noninvasive modality for measuring the mechanical properties of tissue (Ophir et al., 1991). In 1999, Pesavento et al. (1999) developed real-time elastography for clinical use. UE is commonly used to diagnose liver fibrosis, tumors in the breast, liver, thyroid, and prostate (Asteria et al., 2008; Cho et al., 2008; Evans et al., 2012; Frulio & Trillaud, 2013). It is also suitable for musculoskeletal applications (Klauser, Faschingbauer, & Jaschke, 2010).

1.2.1 Types of elastography

Strain elastography

Strain elastography (SE) is based on the principle of vertical axial displacement (compressive strain) of the tissue caused by an external compressive force. The strain is then calculated by comparing the echo responses before and after the compression. The ratio of the change in thickness (Δl) to the initial thickness (l) of an object defines compressive strain, and this ratio is dimensionless (Ophir et al., 1991).

$$Compressive\ Strain = \frac{\Delta l}{l} \tag{1.1}$$

Through rhythmic compression and decompression of the ultrasound probe, stress is applied to the tissue. The resulting strain is displayed in a color-coded elastogram with high stiffness represented typically in red (see figure 1.1) (Klauser et al., 2014; Sigrist, Liau, Kaffas, Chammas, & Willmann, 2017). High strain values correspond to soft tissue, and low strain values correspond to stiff tissue. The pixel count of each color can be used for a semi-quantitative analysis to compare identical anatomic regions within the same individual or to monitor treatment effects (Wu, Chang, Mio, Chen, & Wang, 2011; Sánchez-Infante, Bravo-Sánchez, Jiménez, & Abián-Vicén, 2021; Xu et al., 2017). Another semi-quantitative application is calculating the strain ratio (SR) of two areas at the same depth (Cho et al., 2008) or to calculate SR in different depths using a phantom of defined stiffness lying on the skin (Ariji, Nakayama, Nishiyama, Nozawa, & Ariji, 2015). For the evaluation of the mechan-

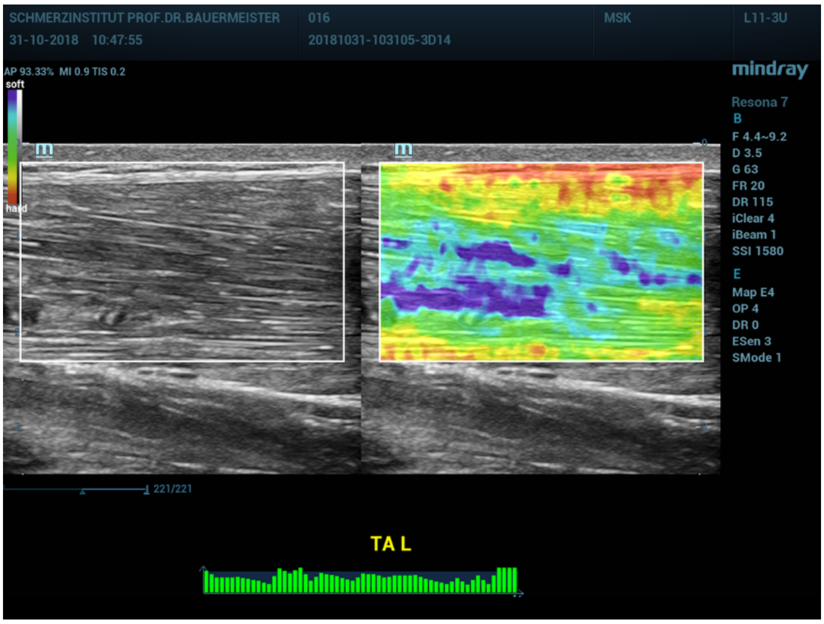


Figure 1.1: Strain elastogram of the tibialis anterior muscle. Red pixels indicate stiff tissue and blue pixels indicate soft tissue. The superficial layer is stiffer than the deeper muscle parts.

ical properties of the musculoskeletal system, SE has been regarded as a promising modality (Drakonaki, Allen, & Wilson, 2012; Klauser et al., 2010; Whittaker et al., 2007). SE has been used to evaluate the Achilles tendon in healthy individuals (De Zordo et al., 2009), those with unilateral Achilles tendon pain (Sconfienza, Silvestri, & Cimmino, 2010), surgically repaired complete ruptures (Tan et al., 2012), and cadavers with histological control showing accordance between histological and sonoelastographic findings (Klauser et al., 2013). SE has also been shown to be capable of detecting changes in the stiffness of the biceps brachii muscle after eccentric contraction compared to a durometer (Niitsu, Michizaki, Endo, Takei, & Yanagisawa, 2011). Inami et al. (2017) reported a nonlinear relationship between the stiffness of the gastrocnemius muscle measured with SE and the intensity of isometric contraction, which represents stiffness changes in response to contraction. Patients with inflammatory myositis showed changes in the elasticity of the affected muscles, and there was a correlation between the SE findings and elevated serum markers (Botar-Jid et al., 2010). A case study showed exact correlation between SE and magnetic resonance imaging (MRI) findings in a case of congenital Bethlem myopathy where SE detected changes not visible in B-scan sonography and MRI

(Drakonaki & Allen, 2010). The inter- and intrarater reliability of SE in the musculoskeletal application is documented in a few studies. For the examination of the gastrocnemius muscle, good reliability (intrarater reliability intraclass correlation coefficient (ICC)=0.89-0.77; interrater reliability ICC=0.89) was reported (Chino, Akagi, Dohi, Fukashiro, & Takahashi, 2012). Alsiri et al. (2020) reported excellent interrater reliability for the rectus femoris muscle (ICC=0.95-0.90) and the gastrocnemius muscle (ICC=0.93-0.90). Interrater reliability ranged from poor to good for biceps femoris muscle (BF) (ICC=0.4- (-0.3)), medial head of gastrocnemius muscle (GM) (ICC=0.7-0.3), and rectus femoris muscle (RF) (ICC=0.7-(-0.2)) (Böttner, Böhm, & Bauermeister, 2018).

Shear wave elastography

Shear wave elastography (SWE) was first described in 1995 (Sarvazyan et al., 1995) as a quantitative method for stiffness measurement. An acoustic radiation force generates shear waves that result in the displacement of tissue particles, leading to the generation of orthogonal shear waves in all dimensions (Nightingale, Soo, Nightingale, & Trahey, 2002; Bercoff, Tanter, & Fink, 2004). The ultrasound probe that emits the push beam records the propagated shear waves. The shear wave velocity C_s is higher in stiff tissue and lower in soft tissue (Gennisson, Catheline, Chaffaï, & Fink, 2003). Stiffness can be measured using C_s (in $\frac{m}{s}$), Shear Modulus G (in kPa) or Young's modulus of elasticity E (in kPa). The corresponding variables can be calculated by

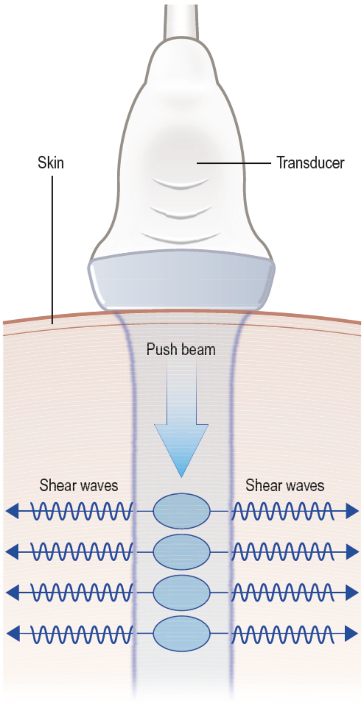
$$G = \rho C_s^2 \tag{1.2}$$

$$G = \frac{E}{2(1+\nu)} \tag{1.3}$$

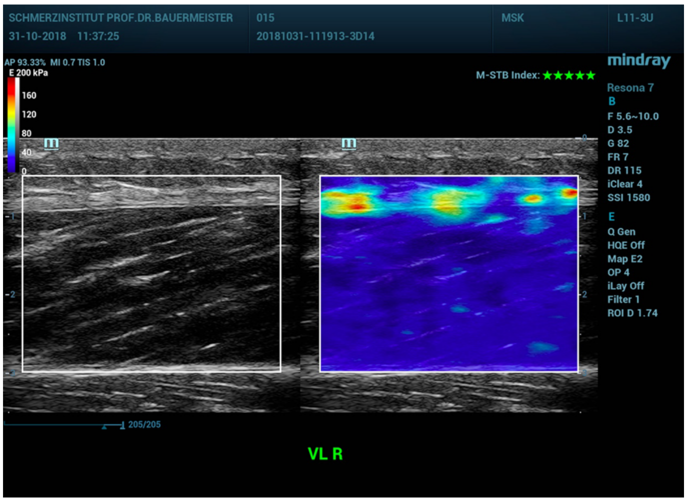
$$C_s = \sqrt{\frac{G}{\rho}} = \sqrt{\frac{E}{2(1+\nu)\rho}} \tag{1.4}$$

$$E = 3G \tag{1.5}$$

where ρ (in $\frac{kg}{m^3}$) is the tissue density and ν is the Poisson's ratio (Parker, Fu, Graceswki, Yeung, & Levinson, 1998). In musculoskeletal application of SWE, ρ is assumed 1000 $\frac{kg}{m^3}$ and $\nu \approx 0.5$ (Creze et al., 2017; Eby et al., 2015) resulting in equation 1.5. SWE is an emerging technique in the field of musculoskeletal ultrasound. It strongly correlated with histology in a mouse model evaluating skeletal muscle fibrosis (Martins-Bach et al., 2021). Various examination and evaluation techniques have been published in several studies on SWE, mainly on lower limb muscles (Creze et al., 2018). Additionally, some studies have evaluated the applicability of SWE to the deep fascia (Luomala, Pihlman, Heiskanen, & Stecco, 2014; Otsuka, Shan, & Kawakami, 2019) and a few publications have focused on the plantar fascia (Kapoor et al., 2010; Lee et al., 2014; Wu et al., 2011) and thoracolumbar fascia (Blain et al., 2019; Chen, Zhao, Liao, Zhang, & Liu, 2020; Wakker, Kratzer, Schmidberger, Graeter, & Group, 2020). Wu and colleagues (2020) reported excellent interrater reliability for RF (ICC 0.987), vastus medialis muscle (VM) (ICC 0.963), vastus lateralis muscle (VL) (ICC 0.952), BF (ICC 0.981), GM (ICC 0.953)



(a) Principle of shear wave elastography: The ultrasound probe emits a push beam, displacing tissue particles. This generates shear waves propagating orthogonally, which are subsequently recorded by the same ultrasound probe. (Schleip, Wilke, & Baker, 2021)



(b) Shear wave elastogram of the vastus lateralis muscle. The color-coded bar on the left indicates the values for Young's modulus of elasticity where red represents high stiffness and blue represents low stiffness.

Figure 1.2: Shear wave elastography principle

and lateral head of gastrocnemius muscle (GL) (ICC 0.968). Shear wave properties of VM and GM, assessed by Dubois et al. (2015) at rest and during passive stretching, were reliable and, especially at rest, well reproducible (interrater reliability ICC=0.91-0.87; intrarater reliability ICC=0.94-0.91). The stretched muscles showed significantly higher shear moduli compared to the relaxed state.
Most of the inter-day experiments reported good to excellent inter-day reliability. A recent study on ten lower limb muscles reported excellent test-retest reliability (ICC=0.99-0.94) between five operators (Liu et al., 2020). For the RF the intra-rater reliability (ICC=0.93–0.94), inter-day reliability (ICC=0.81–0.91), and inter-rater reliability (ICC=0.95) were excellent (Taş, Onur, Yılmaz, Soylu, & Korkusuz, 2017). Excellent inter-day reliability was documented for the hamstrings (ICC=0.96-0.82) (Ichihashi et al., 2016). Excellent to good inter-day reliability was reported in lower leg muscles (ICC=0.96-0.90 for GM, GL, tibialis anterior muscle (TA), peroneus longus muscle (PL)) (Le Sant et al., 2017). Baumer et al (2017) reported fair to moderate inter-day repeatability with SWE applied on the upper limb. There is limited research on using SWE in soccer players. One study by Tas et al. (2019) investigated the stiffness of knee muscles and tendons in 17 male professional soccer players with a median age of 30. Results showed that, compared to the control

group (n=22, median age 28 years), soccer players had higher stiffness in the RF, but no significant difference in the VM stiffness was observed. Another study examined the elastic properties of the GM in 40 male professional soccer players with a mean age of 25 years in relaxed and contracted conditions comparing both legs. The study found no left-to-right differences (Minafra, Alviti, Giovagnorio, Cantisani, & Mazzoni, 2020). To date, there are no cut-off values for shear wave speed or E that indicate a pathologically increased stiffness level. Therefore, a side-by-side comparison of the stiffness values is used to identify abnormalities or asymmetries. Moreover, results of SE and SWE, acquired by the same ultrasound device, have not yet been directly correlated or tested for comparability. As strain measurements are dimensionless, strain ratios between the fascia and the muscle are calculated to ensure comparability to shear wave measurements (see 2.4.2 for more details).

Comparability of SWE measurements

Several factors can affect the accuracy of SWE measurements, including age and gender. While some studies have not reported significant gender differences in stiffness measurements (Akagi & Kusama, 2015; Botanlioglu et al., 2013; Souron et al., 2016; Alfuraih, Tan, O'Connor, Emery, & Wakefield, 2019), others have reported contradictory findings. For instance, higher shear moduli were found in males than females in the GM (Yoshida et al., 2016). The menstrual cycle was found to have no significant effect on overall passive muscle stiffness (Taş & Aktaş, 2020), although a significant influence was observed in the contracted muscle condition (Ham, Kim, Choi, Lee, & Lee, 2020). Additionally, for the biceps brachii muscle, increasing shear moduli in the elderly (>60 years) and higher shear moduli in females were observed (Eby et al., 2015), while higher shear moduli in the RF and GL were reported in younger subjects (<60 years) (Akagi & Kusama, 2015). Moreover, a study of 26 young (20–35 years), 21 middle-aged (40–55 years), and 30 elderly (77–94 years) subjects revealed a gradual reduction in resting muscle stiffness in the lower limb muscles (RF, VM, VL, vastus intermedius, BF, semitendinosus, semimembranosus), as well as in the biceps brachii muscle, with increasing age. This decline in stiffness correlated with lower muscle strength and mass (Alfuraih et al., 2019). Similarly, another study of 50 subjects from four different age groups showed a decrease in muscle stiffness in older subjects in the RF, VM, VL, BF, gracilis, semitendinosus, semimembranosus, GM, GL and tendons and ligaments around the knee (Wu et al., 2020).

Muscle activation, temperature, blood circulation, and tissue heterogeneity are in-vivo factors affecting muscle stiffness. Studies have reported a linear relationship between increasing shear wave speed and progressive isometric contraction in the biceps brachii muscle (Nordez, Gennisson, Casari, Catheline, & Cornu, 2008; Yoshitake, Takai, Kanehisa, & Shinohara, 2014; Yavuz et al., 2015), possibly due to increased cross-bridge formation (Yoshitake et al., 2014). Static stretching has been found to decrease stiffness in lower limb muscles, including the GM (Nakamura et al., 2014) and GL muscle (Akagi & Takahashi, 2013; Taniguchi, Shinohara, Nozaki, & Katayose, 2015) and the hamstrings muscles (semitendinosus, semimembranosus and BF) (Umegaki et al., 2015). Long-term stretching over four to five weeks showed similar effects (Akagi & Takahashi, 2013; Dubois et al., 2015; Ichihashi et al., 2016). Skeletal muscle tissue is heterogeneous due to the various layers and structures present, including nerves, arteries, veins, and lymphatic vessels (Davis, Baumer,

Bey, & Holsbeeck, 2019). Shear wave speed has been found to be higher around vessels in the GM, and muscle fiber orientation can also influence SWE as measurements in the transverse orientation to the fibers gave lower values than in the longitudinal plane (parallel to the muscle fibers) and the images were more stable in the longitudinal plane (Chino, Kawakami, & Takahashi, 2017). Transducer alignment along the muscle fibers is generally recommended for more stable images (Dorado Cortez et al., 2016; Miyamoto, Hirata, Kanehisa, & Yoshitake, 2015; Davis et al., 2019).
Ex-vivo factors include the depth of the region of interest (ROI), pressure applied by the probe, and manufacturer-related factors. The depth of the ROI has been found to impact stiffness measurement values, with minimal changes observed at 3-6 cm depth in a phantom study with a predefined stiffness of 1.90 $\frac{m}{s} \approx 10.83kPa$ - but accurate measurements only obtainable up to 5 cm depth in a phantom with a higher stiffness (3.97 $\frac{m}{s} \approx 47.28kPa$). However, since muscles usually do not reach such high stiffness values in the passive state, SWE can be used for assessing passive muscle stiffness (Davis et al., 2019). It is recommended to use minimal probe pressure to obtain reliable measurements since stiffness values increase with increased pressure (Creze et al., 2018). Additionally, the measurement values can be influenced by the manufacturer (Shin, Kim, Kim, Roh, & Lee, 2016), presets, and acoustic methods used (Ates et al., 2015; Kot, Zhang, Lee, Leung, & Fu, 2012; Creze et al., 2018). Therefore, identical body positioning of the subject is essential to obtain reliable measurements (Creze et al., 2018; Davis et al., 2019). In previous studies, the ROI for elastography was typically set to uniform muscle areas while avoiding areas with variable stiffness (Mendes et al., 2018; Lacourpaille, Hug, Bouillard, Hogrel, & Nordez, 2012; Siracusa et al., 2019; Cortez et al., 2016; Eby et al., 2015). The size of the ROI typically ranged from 100 to 300 mm^2, with one study analyzing the entire muscle depth while excluding areas with abnormally high or low stiffness (Mendes et al., 2018). To address the limitations of small area elastography, some studies used pixel counts of the colored elastogram, assigning E-values to specific colors throughout the ROI (Dubois et al., 2015; Mendes et al., 2018). However, the E-values may exceed the range of the colored elastogram, leading to an inaccurate median E-value. The device used in this study allowed manual tracing of the ROI, covering significantly larger areas of the muscle, up to 8.25 cm^2. Areas with abnormally high or low stiffness were included in the analysis to reflect the actual physical properties of the myofascia. Measurements with a ROI depth of 5cm had increased variance (Davis et al., 2019), so a maximum depth of 4.5cm was used to ensure accurate measurements.

1.3 Range of Motion

Restricted range of motion (ROM) of the lower limbs is considered a potential risk factor for lower extremity injuries, as revealed by various studies (Ekstrand & Gillquist, 1982; Knapik, Bauman, Jones, Harris, & Vaughan, 1991; Söderman, Alfredson, Pietilä, & Werner, 2001). Soccer players reportedly have lower ROM of the hips, knees, and ankles compared to non-athletic individuals (Ekstrand & Gillquist, 1982; Hattori & Ohta, 1986; Bradley & Portas, 2007). Sarcomere contractures, as found in MTrPs (Travell & Simons, 1983), are a possible etiology for a decreased ROM. Furthermore, densification or fibrosis in the fascial layers can also restrict the gliding ability and thus affect the ROM (Luomala et al., 2014; Ling & Slocumb, 1993). Clinically, decreased flexibility or increased stiffness of the soft tissue can also lead to reduced ROM (Myers, Laudner, Pasquale, Bradley, & Lephart, 2006; Wilson, Wood, & Elliott, 1991). These conditions can result in myofascial imbalances of varying degrees (Weishaupt, Obermueller, & Hofmann, 2000).

1.4 Static Alignment and Pedobarography

Alterations in the body's static alignment can occur due to differences in bone length caused by fractures or surgeries such as total hip or knee replacement (Lehnert-Schroth, 1992). However, a one-sided shortening of the hip abductors or thigh adductors is commonly considered a possible cause (Kopecky, 2004), although there is limited scientific validation of this hypothesis. Currently, there is little scientific evidence linking specific changes in static alignment to sports injuries. Watson (1995) found a correlation between muscle injuries and changes in static alignment, and a prospective study involving 210 high school basketball players demonstrated a sevenfold increase in ankle injuries among those with increased postural sway compared to those without (McGuine, Greene, Best, & Leverson, 2000). In soccer players, increased postural sway has been found to be associated with a higher risk of ankle and leg injuries (Tropp, Ekstrand, & Gillquist, 1984; Söderman et al., 2001).

1.5 Objectives

Previous research has identified various factors that increase the risk of injury in soccer players, including muscle stiffness, imbalances, decreased ROM, and changes in static alignment and foot pressure distribution, as mentioned in 1.1. However, there has been limited research on myofascial stiffness in soccer players. UE is a promising tool for assessing myofascial stiffness and evaluating muscles individually. To date, there have been no studies using UE on soccer players aged 14-16 years. This study aims to assess the stiffness of muscles and fascias of the lower extremities of young soccer players using two UE modalities, SWE and SE. One aim is to investigate, wether SE and SWE deliver consistent results. Due to a lack of standard values, elastography findings between the individuals legs are compared. A systematic comparison between the left and the right leg, and between the dominant and the nondominant leg is performed in order to investigate a possible influence of specific repeated motion patterns when kicking the ball. The study will also evaluate ROM, static alignment, and foot pressure distribution to compare the findings to UE. Another intention is to establish baseline data for future studies that will track changes in the physical condition of the fascias and muscles of young soccer players over time and correlate it with injury occurrence. The UE region of interest will be set to the largest technically possible size (8.25 cm^2 for SWE and 8.16 cm^2 for SE) to maximize the amount of information obtained.

1.5.1 Hypotheses

(1) The first hypothesis was that in soccer players, stiffness values of the muscles and fascias of the lower limbs show an intraindividual difference (ΔE).

(2) The second hypothesis assumed that there would be a stiffness difference between the dominant and nondominant as well as between the left and the right leg, acquired by SE and SWE. Further, it was hypothesized that E would be higher in the dominant (shooting) leg compared to the non-dominant (standing) leg.

(3) The third hypothesis assumed that stiffness measurements, obtained as strain ratios and shear wave ratios, yield comparable results.

(4) The fourth hypothesis postulated that a decreased range of motion of the lower extremities would be observed and that these findings correlate with increased stiffness.

(5) The fifth hypothesis suggested that an impaired static alignment and altered foot pressure would be related to stiffness.

2 Material and methods

2.1 Study Design and Subjects

2.1.1 Subject selection

A priori power analysis was conducted to determine the sample size required for the study using G*Power statistical power analysis software (Version 3.1.9.3) (Faul, Erdfelder, Buchner, & Lang, 2009). The analysis assumed an alpha error of 0.05, a statistical power of 0.80, and an effect size of 0.60, based on a previous study using SWE (Miyamoto, Hirata, Miyamoto-Mikami, Yasuda, & Kanehisa, 2018). The critical sample size was estimated to be 19. 20 subjects were included in this cross-sectional study.
The team physician recruited 20 adolescent competitive soccer players (mean age 14.6 ± 0.5 years) from the FC Bayern Academy (Munich, Germany). The inclusion criteria were male gender, age 14-16, and similar training loads of 8-12 hours per week. Exclusion criteria were a history of severe musculoskeletal injuries or current minor infections (obtained through medical history). No subjects were excluded.

2.1.2 Assessment methodology

Ultrasound elastography, range of motion testing, body static alignment, and pedobarography were performed the same day before the training session. The last training session took place one day before to ensure similar conditions in all subjects (see fig. 2.1). The concept of the study was developed by the doctoral candidate. The ultrasound examinations were conducted by Prof. W. Bauermeister, who has over 25 years of experience in musculoskeletal ultrasound in elastography for more than 19 years. The team physiotherapist performed the static and ROM measurements. The coordination of the examinations, as well as the statistical analysis and discussion, were carried out by the doctoral candidate.

2.2 Patient Information and Informed Consent

Before the study, all participants received written and verbal information about the study goals and the examination conducted by the team physician. Both the participants and their authorized caretakers provided written informed consent. The study was approved by the institutional ethics committee (Reference 438/18 s).

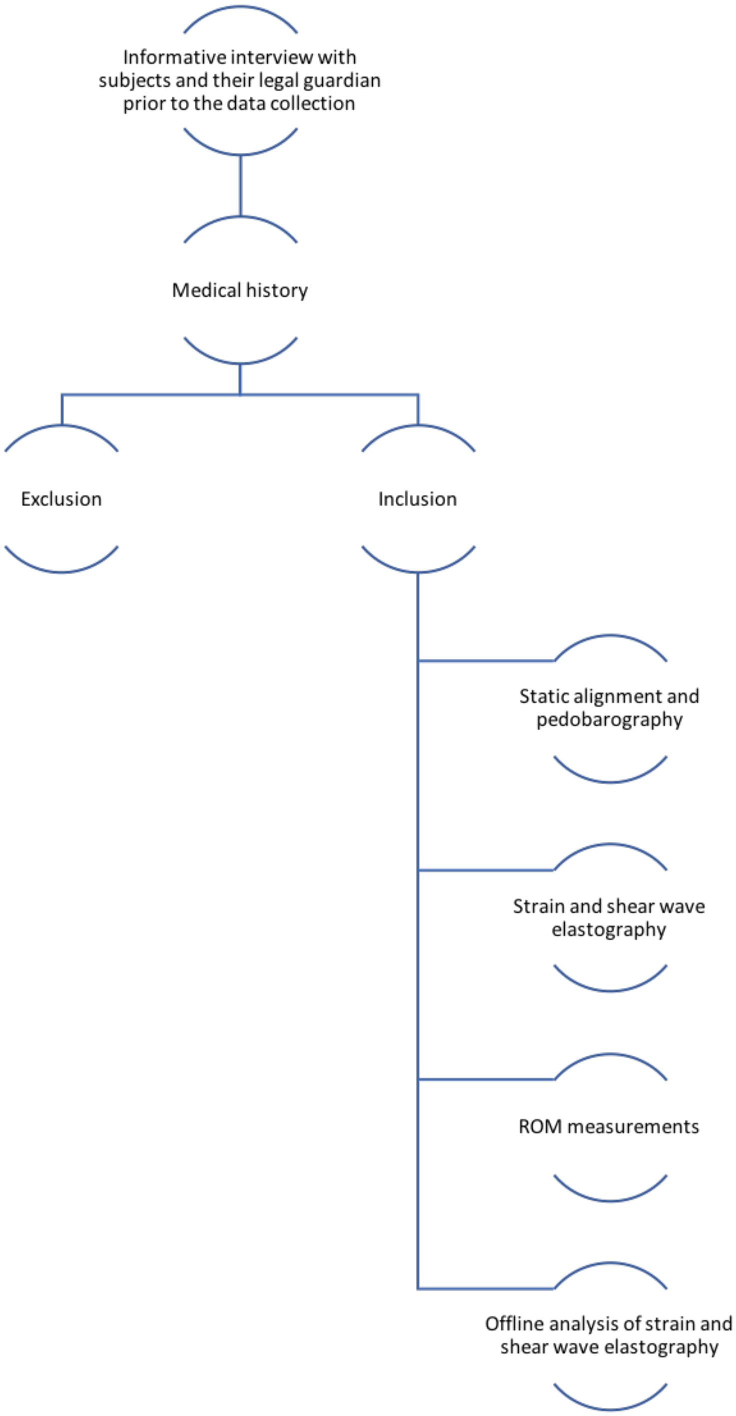


Figure 2.1: Study schedule

2.3 Data Acquisition

A questionnaire was administered to gather demographic data such as age, gender, height, weight, occupation, leisure sports, competitive sports, training age, shooting leg, time of the last training session, current complaints, past injuries, and surgeries. Body mass index (BMI) was calculated using weight and height information. All collected data were recorded and saved in a digital format in an anonymous manner.

2.4 Ultrasound Elastography Examination

2.4.1 Examination protocol

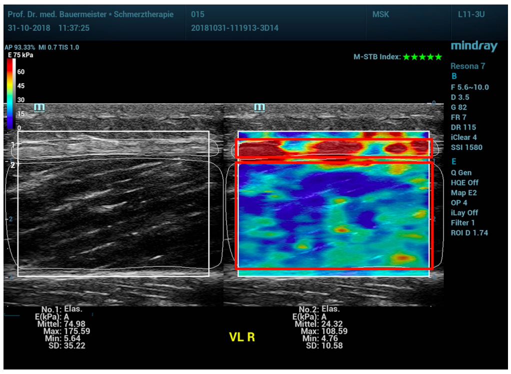
The ultrasound elastography assessment was performed with a Mindray Resona 7 ultrasound device (Mindray Bio-Medical Electronics Co., Shenzhen, China). A linear 11-3 MHz transducer was used for the strain and shear wave elastography assessment. The device settings for SWE were as follows: Push-beam Q-Gen with a frequency range of 5.6-10.0 MHz, PB-Impulse-frequency 1 Hz, depth 3.5cm for tensor fasciae latae muscle (TFL), proximal rectus femoris muscle (RFpr), distal rectus femoris muscle (RFdi), VM, VL, TA, PL and 5cm for gluteus medius muscle (GMED), gluteus maximus muscle (GMAX), adductor muscles (ADD), BF, GM and GL. For SE the frequency range was 4.4-9.2 MHz and the same depth was used as for SWE. UE measurements were recorded without minimum preload, in the longitudinal plane. A water-soluble transmission gel Aquasonic 100 (Parker Laboratories Inc., USA) was applied and the participants were examined in prone and supine position with the legs fixated to prevent motion (see Figure 2.2a). In supine position UE elastograms of TFL, RFpr, RFdi, VM, VL, TA, PL were recorded. In prone position UE elastograms of GMED, GMAX, ADD, BF, GM and GL were recorded. The measurements were performed at the muscle belly where typically electromyography recordings are made (Murray, 1995).

Large area shear wave elastography

To achieve optimal SWE results, minimizing tissue pressure and motion of both the probe and the tissue being examined (Eby et al., 2013) is crucial. The measurements were taken with the probe being fixed in an articulated arm (Maquet, Germany) to ensure motion stability (see Figure 2.2a). The ultrasound device used also provided a motion stability index, which ranged from “0” (poor quality) to “5” (maximum quality) and indicated any interference caused by factors such as subject respiration, arterial pulsation, or movements of the transducer during free-hand examination. Measurements were only taken with a rating of five stars to ensure high-quality results. The elastogram was set at maximum size with approx. 3.3cm x 2.5cm for large area shear wave elastography (LA-SWE). The elastogram was displayed in “dual mode” with the B-scan and the elastogram side by side, with the elastogram color-coded from blue (soft) to red (stiff) to indicate tissue stiffness (see Figure 1.2b). Stiffness calculations for Young’s modulus E in kPa were provided for the entire ROI, including mean, maximum, and minimum values. It would be ideal to report all stiffness measurements as shear wave velocity C_s, but due to patent restrictions in Germany, the elastography device used in this study only reported

Young's modulus of elasticity E in kPa. The images were stored as raw data to allow later analysis.

(a) Set-up for shear wave elastography assessment. The subject's legs are secured to prevent motion, and the ultrasound probe is attached to an articulated arm.



(b) SWE tracing of the vastus lateralis muscle. The fascia (iliotibial tract) and the muscle were traced (red boxes), with the results for E appearing below.

Figure 2.2: Shear wave elastography assessment

Strain elastography

SE requires rhythmic manual compression of the tissue with 80-100 beats per second. The elastogram was set at maximum size with approx. 3.4cm x 2.4cm. Similar to SWE, the exam was conducted in "dual mode" with the B-Scan and the elastogram side by side with the color code for the elastogram ranging from blue, indicating soft, to red, indicating stiff tissue (see Figure 1.1). The most stable elastogram was saved for offline analysis.

2.4.2 Data evaluation

Large area shear wave elastography

The offline analysis was conducted from the stored raw data. The fascia and muscle layers were outlined, and the mean, maximum, minimum, and standard deviation values for Young's modulus of elasticity (E) in kPa were obtained (see Figure 2.2b). Mean values were used for comparisons across limbs. The shear wave ratio (SWR) was calculated by dividing Young's modulus of the fascia by Young's modulus of the muscle (according to equation 2.1). The results were documented in an Microsoft Excel sheet (Microsoft Corporation, Redmond, WA, USA) for further statistical analysis.

Strain elastography

Offline analysis of the fascias and the muscles was done by calculating SR as the ratio of the strain of the muscle divided by the strain of the fascia (equation 2.2).

$$Shear\ Wave\ Ratio\ (SWR) = \frac{E_{Fascia}}{E_{Muscle}} \tag{2.1}$$

$$Strain\ Ratio\ (SR) = \frac{Strain_{Muscle}}{Strain_{Fascia}} \tag{2.2}$$

Strain elastography only provides information about the stiffness ratios between two regions of interest, in this case, the muscle and fascia, without giving absolute values. Low strain implies high elasticity, whereas, in shear wave elastography, a high Young's modulus indicates high tissue elasticity. To ensure comparability, the muscle was used as the denominator in the SWR calculation and as the numerator in the SR calculation. A ratio above 1 indicates that the fascia is stiffer than the muscle, while a ratio below 1 that the muscle is stiffer than the fascia.

2.5 Range of Motion

2.5.1 Lumbar range of motion

Lumbar ROM was measured with the back range of motion (BROM) Instrument (Performance Attainment Associates, Roseville, Minnesota, USA) with two-degree grading. Lumbar rotation (LRO) was measured using a compass goniometer, while the built-in gravity-goniometer was used to measure lumbar lateral flexion (LLF) (see figures 2.4a, 2.3) (Paul). The ROM measurements were taken by the team physiotherapist. The intra- and inter-rater reliability of the BROM varied from excellent to poor. Breum et al. (1995) found good intra- and inter-rater reliability for LLF (ICC=0.91 and 0.85) and moderate to poor for LRO (ICC=0.57 and 0.36). Good intra-rater reliability could be confirmed by Kachingwe and Phillips (2005) for LLF (ICC range 0.85-0.83), intra-rater reliability for LRO however was fair to poor (ICC range 0.76-0.58). Madson et al. (1999) found excellent intra-rater reliability for LLF to the left (ICC=0.91), LLF to the right (ICC=0.95) and LRO to the right (ICC=0.93) and good for LRO to the left (ICC=0.88). Atya et al. (2013) documented good intra-rater reliability for LRO (ICC range, 0.86-0.88) and for LLF (ICC range, 0.81-0.82).

Lumbar rotation and lateral flexion measurements

The BROM was positioned at the T12 level, stabilizing the compass needle with a magnetic yolk at the S1 level. The subject was seated with the hip and knee bent at approximately 90°and the feet flat on the floor. The subject was instructed to rotate the trunk to the left slowly, ensuring it goes full range, and afterwards to the right. For the lateral flexion with the gravity inclinometer the subject was asked to do smooth steady lateral flexion to the left and to the right (see Figure 2.3).
Published values for LRO were predominantly reported between 12.8 to 16.6 degrees (Van Herp, Rowe, Salter, & Paul, 2000; Peach, Sutarno, & McGill, 1998; Hindle, Pearcy, Cross, & Miller, 1990), so in this study for statistical analysis, 15 degrees were considered the typical value. For LLF reported values ranged between 30.2 and 37.1 degrees (Stubbs, Fernandez, & Glenn, 1993), the American Academy of Orthopaedic Surgeons declares 35 degrees as a typical value which was indeed used as a reference value in this study (Burrows, 1965).

Figure 2.3: Lumbar rotation and lateral flexion assessment. (Prof. Dr. med. W. Bauermeister, Munich, Germany)

2.5.2 Joint measurements

For the testing of the extremities' ROM, the Deluxe Inclinometer (Performance Attainment Associates, Roseville, Minnesota, USA) with one-degree grading was used, which employs a small fluid-dampened oscillation-free ball to indicate inclination and provide a parallax-free reading (see Figure 2.4b). The Inclinometer was utilized in this study to measure internal rotation of the hip (HIP_{IR}), external rotation of the hip (HIP_{ER}) and straight leg raise test (SLR).
Furness et al. (2015) reported the inclinometer to be a reliable and valid tool for clinical use with excellent inter-rater reliability for the ROM examination of the shoulder rotation (ICC=0.98 and 0.99). Crowell et al. documented excellent intra-rater reliability for pelvic inclinometer (ICC=0.95) from a different manufacturer (Crowell, Cummings, Walker, & Tillman, 1994). Intra-rater reliability was excellent for the assessment of shoulder ROM (ICC=0.99 and 0.97) using a handheld inclinometer from a different manufacturer (Kolber, Saltzman, Beekhuizen, & Cheng, 2009).

Hip internal and external rotation

The subject was in a prone position to assess the external and internal rotation of the hip. The knee was passively flexed to 90°. The inclinometer was placed posteriorly at the lower third of the lower leg with the thighs adducted. The examiner assured that compensatory movements like hip abduction were avoided. The sacrum was stabilized by the examiner's hand to recognize its motion when reaching the limit of the hip rotation. Typical values for HIP_{IR} range from 47.3 degrees (Boone &

Azen, 1979) to 32.5 degrees (Roaas & Andersson, 1982), the American Academy of Orthopaedic Surgeons suggests 45 degrees (Burrows, 1965) which was used in this study. HIP_{ER} ranged from 47.2 degrees (Boone & Azen, 1979) to 33.7 degrees (Roaas & Andersson, 1982), the American Academy of Orthopaedic Surgeons reports 45 degrees (Burrows, 1965).

Straight leg raise

The subject was supine with the legs extended for the passive SLR. The inclinometer was placed laterally on the distal third of the lower leg. The leg was raised to a point just before pelvic co-movement would occur. Reported values ranged from 60.6 to 89.0 degrees (Sweetman, Anderson, & Dalton, 1974) and 70 to 86 degrees (Elson & Aspinall, 2008). In this study, a reference value of 80 degrees was used.

Ankle dorsiflexion

Active ankle dorsiflexion (ADF_a) was assessed with the knee extended in a supine position, passive ankle dorsiflexion (ADF_p) was measured in a prone position with the knee in 90° flexion. The BROM unit was placed on the foot's plantar surface, and the readings were made from the gravity inclinometer.
Values of ADF_a 12.0 o 17.4 degrees (Stubbs et al., 1993), 12.6 degrees (Boone & Azen, 1979), 15.3 degrees (Roaas & Andersson, 1982) and 20 degrees (Burrows, 1965) were reported. In this study, 20 degrees was considered a typical value for ADF_a, and 30 degrees for ADF_p as proposed by the American Academy of Orthopaedic Surgeons (Burrows, 1965).

2.5.3 Distance measurements

The measurement of finger-floor distance (FFD) combines the evaluation of hamstring, gluteal, and back muscle tightness. A distance above 0 cm is considered a positive (i.e. not physiological) test result (Czaprowski et al., 2013; Gauvin, Riddle, & Rothstein, 1990). FFD test showed reproducible results at an interval of six months (Biering-Soerensen, 1984). Gauvin et al. documented it as highly reliable regarding intra- and inter-rater reliability (ICC=0.98 and 0.95) (Gauvin et al., 1990). One study reported poor reproducibility (Merritt, McLean, Erickson, & Offord, 1986).
Heel-buttock distance (HBD), the distance between the heel and the buttocks, assesses the knee flexion range (Theiler et al., 1996). According to Theiler et al. (1996), the inter-rater reliability of the HBD was relatively poor. However, it is a simple, fast, often clinically used method to get a rough impression of the flexibility of the RF.

Finger-floor distance

The FFD was measured barefoot with a ruler in a standing position, feet at shoulder width, and knees straight. The distance between the tip of the middle finger and the floor was measured in cm. Typically, FFD was 0 cm (Czaprowski et al., 2013; Gauvin et al., 1990).

Heel-buttock distance

The HBD was tested passively in the prone position with a ruler. The knee was bent until a significant resistance was felt. The distance between the heel and the buttock was measured in cm. HBD above 0 cm indicates tightness of the quadriceps (Theiler et al., 1996), however some authors consider values above 15 cm as pathological (Keays, Mason, & Newcombe, 2015).

(a) Back Range of Motion Instrument (BROM) with a compass goniometer (upper part) and a gravity inclinometer (lower part).

(b) Deluxe Inclinometer.

Figure 2.4: ROM measurement tools

2.6 Static Alignment and Pedobarography

To obtain postural and foot pressure data, the ABW BodyMapper 4D (SinfoMed GmbH, Frechen, Germany) was used in this study (see Figure 2.5). This allows a complete body analysis applying raster-stereography and pedobarography. Measurements were obtained from the frontal and sagittal plane using body markers and were performed by the team physiotherapist. A recent meta-analysis on 19 studies using raster-stereography estimated satisfactory reliability and validity in detecting spinal posture (Krott, Wild, & Betsch, 2020). In the sagittal plane, inter-day and inter-week reliability were high (ICC 0.94–0.99) (Schroeder, Reer, & Braumann, 2015). The projected stripe pattern was recorded with a video camera. For static

Figure 2.5: Set-up for static alignment and pedobarography assessment.

alignment measurements, the subjects stood on a pedobarography-plate with six markers on their back, and the projected stripe pattern was recorded with a video camera. A three-dimensional image was then constructed using a triangulation technique with a depth resolution of 1/100mm, and the measurement error was assumed to be less than 1 mm (Ohlendorf et al., 2012). In a previous study a maximum reproducibility error of the measurement technique of 2% was found (Ohlendorf et al., 2012). Parameters such as frontal position of vertebra prominens (VP) and frontal position of the pelvis (PF) were assessed to estimate vertebral and pelvic alignment in the frontal plane. Additionally, foot balance was assessed using the foot plate to measure total force, foot pressure distribution between the left and right as well as the dominant and nondominant feet, and between the fore- and rearfoot.

2.7 Statistical Analysis

Statistical analysis was performed using Microsoft Excel (Microsoft Corporation, Redmond, WA, USA) version 16.37, R Studio (R Studio, Inc., Boston, MA, USA) version 3.6.1, and Orange Data Mining (Bioinformatics Lab, University of Ljubljana, Slovenia) version 3.25. The initial level of significance was set at $P<0.05$. Due to multiple testing, Bonferroni correction was applied to reduce the risk of type I error. Data were tested for normality using Shapiro-Wilk test. Standard descriptive statistics consisted of means and standard deviation (SD) as well as medians and the interquartile range (IQR) for the continuous variables and counts and percentages for the categoric variables. Results were expressed as mean ± SD or median (IQR). The SWE, SE, ROM and pedobarography values were grouped in dominant and nondominant as well as left and right side. For the intraindividual stiffness difference in SWE between the harder and softer side, independent of leg dominance, ΔE was computed by subtracting the Young's modulus E of the leg with the lower (E_{low}) from the leg with the higher E value (E_{high}), $\Delta E = E_{high} - E_{low}$. Due to the small sample size (n=20), before testing for differences, Shapiro Wilk test was used to check wether differences of the pairs follow a normal distribution. The Wilcoxon Signed-Rank test was then used for non-normally distributed data, while the paired t-test was used for normally distributed data to evaluate differences between groups. SR and SWR were analyzed for differences using the previously described methods. Additionally, the concordance between SR and SWR was assessed by comparing the difference between their respective ratios (muscle to fascia, for details see 2.4.2) and a hypothetical ratio of 1, which represents equal stiffness distribution between the muscle and fascia.
Differences to reference values in ROM were calculated using either one-sample Wilcoxon Signed-Rank test or one-sample t-test. Correlation between UE and ROM and between UE and body static alignment and pedobarography was tested using Spearman rank correlation. Effect sizes were calculated using Cohen's d for parametric data and effect size r ($r = \frac{z}{\sqrt{N}}$) for nonparametric data, with values of 0.2, 0.5, and 0.8 classified as small, moderate, and large for Cohen's d effect sizes (Cohen, 1990), and values of 0.1, 0.3, and 0.5 classified as small, moderate, and large for r effect sizes (Rosenthal, Cooper, & Hedges, 1994).

3 Results

3.1 Descriptive Data of Subjects

All recruited subjects were included in the study (n=20) with a mean age of 14.6 ± 0.5 years, a mean BMI of 20.8 ± 1.4 $\frac{kg}{m^2}$, an average of 8.8 ± 2.4 years of soccer activities, and 9.2 ± 1.9 hours of soccer training per week. 16 participants had no complaints, and four had minor complaints that did not interfere with the training, such as knee, hip, and groin pain and one had a toe injury. Ten participants reported minor injuries in their past medical history such as muscle fiber tears (n=3), wrist fractures (n=2), arm fracture (n=1), Morbus Osgood Schlatter (n=1), patella bipartita (n=1), clavicle injury (n=1), and ankle injury (n=1). 17 participants did not have surgeries in the past, while one had wrist surgery, one had an appendectomy and one had a tonsillectomy (see table 3.1).

Table 3.1: Demographic characteristics

Characteristics	**Study Group**[a]
Age (years)	14.6 ± 0.5
Gender, male, n (%)	20 (100%)
Height (cm)	177.5 ± 6.3
Weight (kg)	65.7 ± 6.5
BMI($\frac{kg}{m^3}$)	20.8 ± 1.4
Years of training	8.8 ± 2.4
Training units per week (hours)	9.2 ± 1.9
Free leg (left/right)	11 (55%)/8 (40%)
Current complaints	
No	16 (80%)
Minor[b]	4 (20%)
Injuries in the past	
No	10 (50%)
Minor[c]	10 (50%)
Surgeries in the past	
No	17 (85%)
Minor[d]	3 (15%)

[a] Values are presented as mean ± SD or as counts and percentages.
[b] Slight pain of the knee (n=1), slight pain of the hip (n=1), slight pain of the groin (n=1), toe injury (n=1).
[c] Muscle fiber tear (n=3), wrist fractures (n=2), arm fracture (n=1), Morbus Osgood Schlatter (n=1), patella bipartita (n=1), clavicle injury (n=1), ankle injury (n=1).
[d] Wrist surgery (n=1), appendectomy (n=1), tonsillectomy (n=1).

3.2 Results of Ultrasound Elastography

3.2.1 Large area shear wave elastography

Significant intraindividual stiffness differences ΔE were noted in fascia of tensor fasciae latae muscle (TFL_f), fascia of vastus lateralis muscle (VL_f), tibialis anterior muscle (TA_m), fascia of peroneus longus muscle (PL_f), fascia of medial head of gastrocnemius muscle (GM_f), fascia of lateral head of gastrocnemius muscle (GL_f) after Bonferroni correction. Ranges from 40.40 kPa (VL_f) to 6.00 kPa (TA_m were observed (see table 6.1, figures 3.1, 3.2).
No significant difference was found for stiffness between the dominant and nondominant legs muscles and fascias (see table 6.2). These findings were also reflected in the pooled mean values (see table 3.2).
No significant difference was found for stiffness between the left and right legs muscles and fascias (see tables 3.2, 6.3).

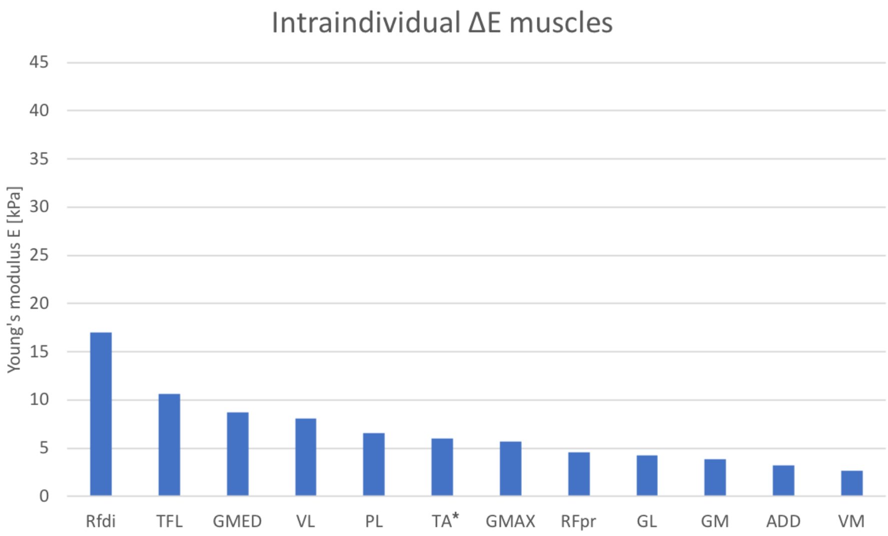


Figure 3.1: Intraindividual muscle stiffness ΔE (n=20).
* significant difference (p<0.002).

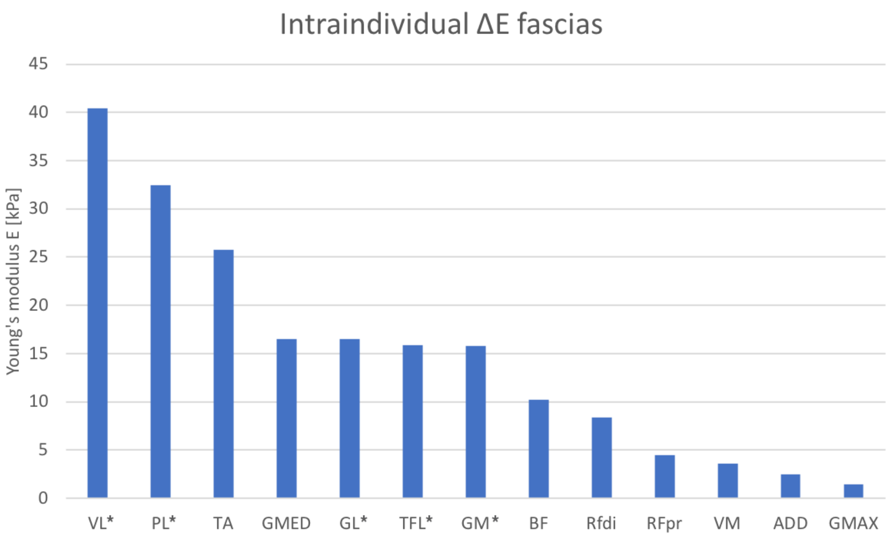


Figure 3.2: Intraindividual fascia stiffness ΔE (n=20).
* significant difference (p<0.002).

Table 3.2: Pooled mean values of Young's modulus

	Dominant leg[a]	**SEM**	**Nondominant leg[a]**	**SEM**	**P**	**Effect size**
Muscle	24.3 ± 2.6	0.571	24.3 ± 2.8	0.619	0.967[c]	0.011[d]
Fascia	45.7 ± 7.9	1.766	43.2 ± 6.5	1.442	0.220[c]	0.335[d]
P	<0.001*[c]		<0.001*[c]			
Effect size	3.638[d]		3.809[d]			

	Left leg[a]	**SEM**	**Right leg[a]**	**SEM**	**P**	**Effect size**
Muscle	24.9 ± 2.8	0.631	22.6 (4.0)	0.520	0.054[b]	0.477[d]
Fascia	45.4 ± 7.8	1.746	43.5 ± 6.7	1.488	0.342[c]	0.260[d]
P	<0.001*[c]		<0.001*[c]			
Effect size	3.485[d]		3.974[e]			

[a] E (kPa).Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] r effect size.
* Significant difference (level of significance after Bonferroni correction <0.002.)

3.2.2 Strain elastography - strain ratios muscle/fascia

There was no significant difference for SRs between the dominant and nondominant and between the left and right leg, see tables 6.4, 6.5 and figure 3.3. Shear wave ratios neither showed any difference between the dominant and nondominant and the left and right leg (tables 6.4, 6.5).

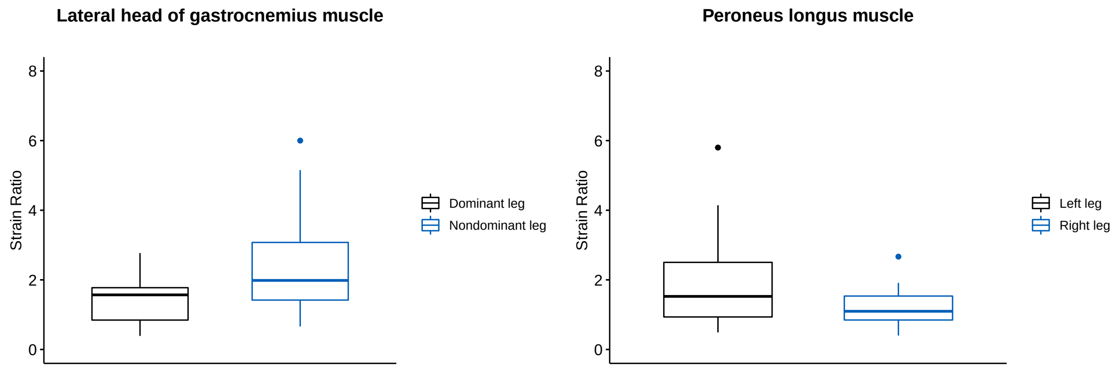


Figure 3.3: Strain Ratios in the dominant and nondominant and in the left and right leg. After the Bonferroni correction, no significant difference was found.

Ratio comparison between strain and shear wave elastography

After the Bonferroni correction in 7 of 13 muscles, there was a significant difference between the SR and SWR. Those were RFpr, VM, TA (right and nondominant leg), PL (right and dominant leg), GMAX, ADD (except for nondominant leg), GL (only in the dominant leg), see tables 6.4, 6.5. These differences were not reflected in the pooled mean values of SR and SWR (see table 3.3).
SRs of muscle to fascia were above 1 in 8 out of 13 muscles (TFL, RFpr, VM, VL, ADD (except in the nondominant leg), BF (only in the left leg), GM, GL). The calculated SWRs were above 1 in 8 out of 13 muscles (TFL, RFpr, VL, TA, PL, BF, GM and GL), see tables 6.4, 6.5, 6.6. Strain and shear wave ratios above 1 indicate higher fascia stiffness.

Table 3.3: Pooled mean values of SR and SWR

	SR[a]	**SEM**	**SWR[a]**	**SEM**	**P**	**Effect size**
Dominant leg	1.93 ± 0.62	0.137	1.83 ± 0.26	0.058	0.517[c]	0.225[d]
Nondominant leg	1.90 ± 0.64	0.143	1.73 ± 0.23	0.052	0.313[c]	0.343[d]
P	0.523[c]		0.126[c]			
Effect size	0.057[d]		0.388[d]			
Left Leg	1.10 ± 0.61	0.135	1.79 ± 0.28	0.062	0.457[c]	0.240[d]
Right leg	1.92 ± 0.65	0.146	1.76 ± 0.23	0.050	0.282[c]	0.325[d]
P	0.846[c]		0.592[c]			
Effect size	0.017[d]		0.137[d]			

[a] E (kPa).Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] r effect size.
[*] Significant difference (level of significance after Bonferroni correction <0.002.)

3.3 Results of Range of Motion Measurements

3.3.1 Side comparison

No significant difference in dominant to non-dominant comparison was found (see table 6.7). No significant difference in ROM of the left and right leg was found (see table 6.8).

3.3.2 Comparison to published typical values

Compared to published typical values, LLF, HIP_{IR}, ADF_a, and ADF_p were decreased, and HBD was increased. LRO, HIP_{ER}, SLR and FFD were were consistent with the typical values, see tables 6.7, 6.8.

3.4 Results of Static Alignment and Pedobarography

3.4.1 Descriptive statistics

The static alignment with the vertebra prominens and the pelvis as points of reference for the frontal plane was normal (see table 6.9). The pedobarography showed no difference in total foot pressure between the dominant and nondominant and between the left and right leg (see figure 3.4). The distribution of the fore- and rearfoot pressure differed significantly from the typical distribution in the dominant and nondominant, as well as in the left and right leg (see figure 3.5, tables 6.10, 6.9).

3.4.2 Correlation between myofascial stiffness, range of motion and foot pressure

There was no significant correlation between myofascial stiffness, ROM, and foot pressure after the Bonferroni correction. Considering a more liberal p-value (0.05 instead of 0.01), HBD and E of distal rectus femoris muscle ($RFdi_m$) correlated significantly in the non-dominant leg ($\rho = 0.490$), see table 6.12, figure 3.6. The forefoot pressure correlated significantly with E of fascia of distal rectus femoris muscle ($RFdi_f$) in the non-dominant leg (ρ=0.533), see table 6.14, figure 3.7.

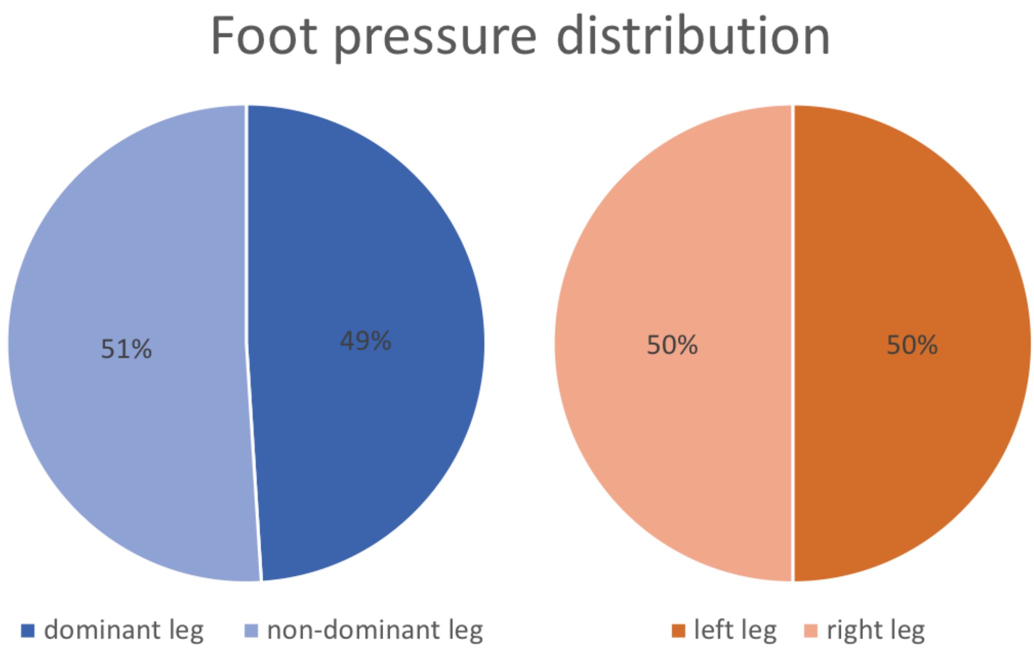


Figure 3.4: Total Foot pressure distribution in % between the dominant and non-dominant leg and between the left and right leg.

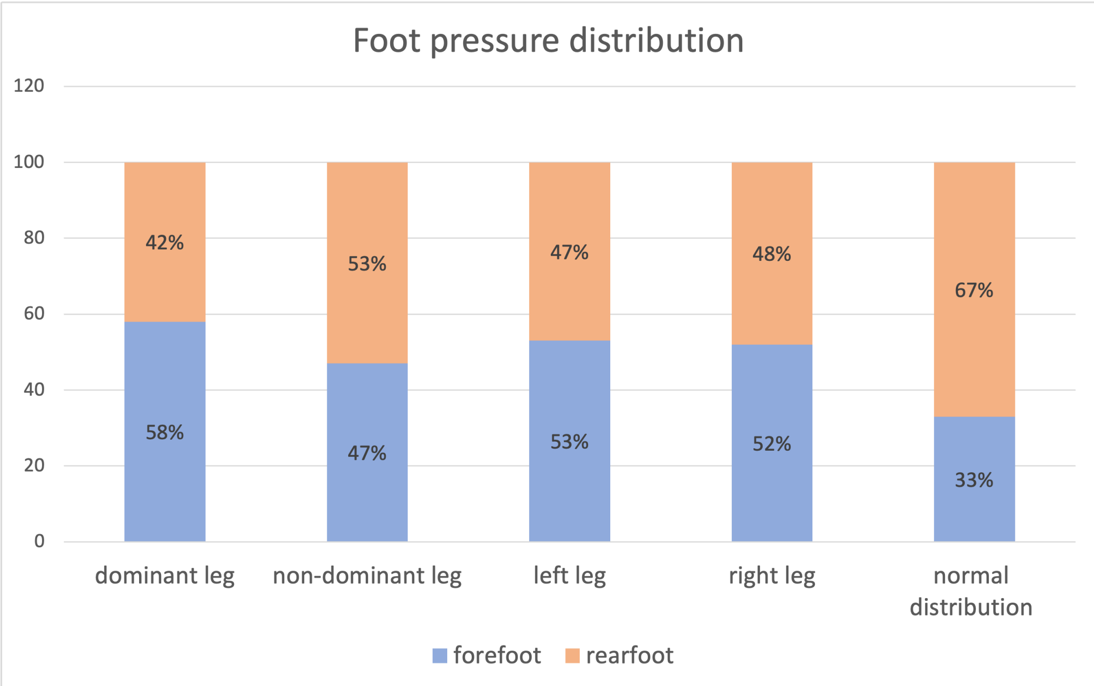


Figure 3.5: Forefoot-Rearfoot pressure distribution in % for the dominant and non-dominant leg and in the left and right leg.

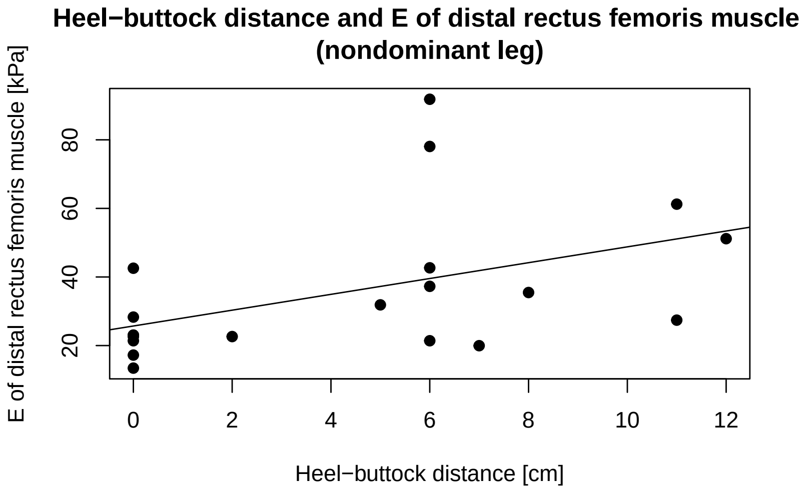


Figure 3.6: Correlation between heel-buttock distance and E of distal rectus femoris muscle in the non-dominant leg.

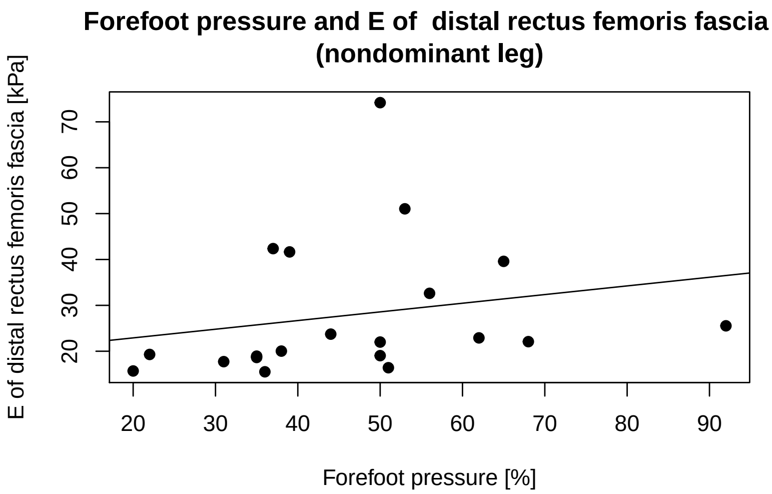


Figure 3.7: Correlation between forefoot pressure and E of distal rectus femoris fascia in the non-dominant leg.

4 Discussion

4.1 Intraindividual Stiffness Differences

The first hypothesis was that in soccer players, stiffness values of the muscles and fascias of the lower limbs show an intraindividual difference (ΔE). Significant intraindividual stiffness differences ΔE in the muscles and fascias were noted in $\mathrm{TFL_f}$, $\mathrm{VL_f}$, $\mathrm{TA_m}$, $\mathrm{PL_f}$, $\mathrm{GM_f}$ and $\mathrm{GL_f}$ after Bonferroni correction with ranges for ΔE from 40.40 kPa ($\mathrm{VL_f}$) to 6.00 kPa ($\mathrm{TA_m}$), see table 6.1. Before adjusting for multiple comparisons with Bonferroni correction, the level of significance was 0.05 and significant differences were observed in every measurement point (muscle and fascia).

From a practitioner‘s perspective, stiffness differences are used as a guide for therapeutic interventions targeting treatment towards the stiffest area. Generally, examiners have to rely on manual palpation to identify stiff and tender myofascial areas. This subjective approach requires years of experience and is difficult to quantify objectively to be considered in follow-up examinations (Hsieh et al., 2000; Gerwin, Shannon, Hong, Hubbard, & Gevirtz, 1997; Simons et al., 1999). Other techniques that can be used to assess myofascial stiffness functionally include the vertical hop test (Pruyn et al., 2012), free-oscillation technique (Watsford et al., 2010), and myometry (Viir, Laiho, Kramarenko, & Mikkelsson, 2006). The issue of leg stiffness is a matter of debate within the field, with some researchers suggesting that high stiffness can improve athletic performance, while others attribute it to an increased risk of injury (Butler, Crowell III, & Davis, 2003). For instance, Arampatzis et al. (1999) have suggested that high stiffness can be useful for runners' performance, whereas other authors have argued that high stiffness may increase the risk of injury (Ekstrand & Gillquist, 1982; Pruyn et al., 2012). These techniques however do not allow to specifically measure an individual muscle or they only measure surface muscle structures. Various therapeutic interventions are applied to reduce stiffness differences and improve myofascial function. These include trigger-point injections (Travell, Rinzler, & Herman, 1942), dry needling (Maher, Hayes, & Shinohara, 2013), various forms of manual therapy (Gao, Caldwell, Zhang, & Park, 2020; Weiss, 2001), shockwave therapy (Zhang, Duan, Liu, & Zhang, 2019), repetitive peripheral magnetic stimulation (Okudera et al., 2015; Prucha, Socha, Sochova, Hanakova, & Stojic, 2018) and other modalities (Janssens, 1992). All therapies aim to reduce stiffness differences, which in turn reduces pain and improves function (Hsieh et al., 2007). However, due to the lack of standard stiffness values, the patient serves as their own control, aiming to equalize the stiffness values in the corresponding anatomical area. SWE allows for an objective assessment of myofascial stiffness and monitoring of treatment effects.

As shown in this study, stiffness differences between limbs can be identified by using SWE. This study is the first to investigate the absolute intraindividual leg stiffness difference (ΔE) assessed by SWE, as previous studies either compared the left and right leg (Botanlioglu et al., 2013; Minafra et al., 2020) or the dominant and nondominant leg (Feng, Li, Liu, & Zhang, 2018; Taş et al., 2019; McPherson et al., 2020).
When comparing the stiffness of the muscle with the respective fascia, there were significantly higher E-values for the fascias for TFL, VL, TA, PL, BF, GM and GL, regardless of leg dominance, with small to large effect sizes (see table 6.1). However, the fascia stiffness was lower than the muscle stiffness in GMAX. It is worth noting that in the analysis of GMAX, the fascia was difficult to distinguish from the muscle due to its thin collagenous layer.
The muscle and the fascia form an anatomical and functional unit, providing force transmission across the joints (Wilke et al., 2019). As a result, the stiffness of the fascia may contribute to the development of overuse injuries. The intraindividual difference in stiffness (ΔE) may be an important parameter for injury risk and prevention, regardless of which leg is dominant. This study seeks to establish a baseline for future research and monitoring. Athletes may develop areas of greater stiffness over time, which could be associated with a higher risk of injury (Wik et al., 2021).

4.2 Leg Dominance

In this study, the second hypothesis assumed that there would be a stiffness difference between the dominant and nondominant as well as between the left and the right leg, acquired by SE and SWE, and that E would be higher in the dominant leg. For the results of SE, see the next section 4.2.1. We found no significant difference in E between the dominant and the nondominant, as well as between the left and the right leg after multiple testing correction (see tables 6.2, 6.3). Only in the GMED_m, with the non-dominant leg being stiffer, a moderate effect size was observed when using a more lenient p-value of 0.05. It is plausible that the dominant leg, which is used more frequently for kicking and generating higher ball velocity, is more susceptible to overuse injuries (DeLang et al., 2021). Thus, it was initially hypothesized that soccer players would have higher stiffness values in the muscles and fascias of the lower limb in the dominant leg. However, this study failed to support this hypothesis. Instead, the GMED_m had a higher E value in the non-dominant leg, with statistical significance before after applying the Bonferroni correction. Similar findings were reported for leg strengths in 41 English soccer players with higher values for knee flexion in the non-dominant limb (Rahnama, Lees, & Bambaecichi, 2005). While kicking a ball, the contralateral hip abductors stabilize the pelvis, so GMED is under high demand. This may functionally explain the higher stiffness value in the nondominant leg. However, GMED_m only showed a side difference before adapting the p-value, and one would expect to find more asymmetries in highly demanded muscles like the quadriceps femoris if leg dominance had a systematic influence on stiffness patterns. In line with the findings of this study, RF and VM muscles showed no difference in stiffness between the dominant and nondominant leg of 17 professional soccer

players with a median age of 30 years (Taş et al., 2019). Due to its bi-articular nature and high mechanical demand, the RF is a common injury site in young soccer players (Ekstrand, Haegglund, & Walden, 2011; Garrett Jr, 1996; Hawkins, Hulse, Wilkinson, Hodson, & Gibson, 2001; Lewin, 1989; McGregor & Rae, 1995) (Jacobs, Bobbert, & van Ingen Schenau, 1996). There was a high ΔE of 17.03 kPa in the $\mathrm{RFdi_m}$, which could suggest an increased susceptibility to injury in the population of this study, but further research with a larger cohort is needed to confirm this finding. Hamstring injuries are a common occurrence among elite soccer players (Ekstrand et al., 2011; Garrett Jr, 1996; Hawkins et al., 2001; Lewin, 1989; McGregor & Rae, 1995), but in this particular study, the $\mathrm{BF_m}$ showed a low ΔE of 2.58 kPa. However, there was a significant intra-individual difference of 10.18 kPa in the fascia, and the role of the fascia in the context of injury risk has yet to be studied extensively.
The $\mathrm{VL_f}$, which includes the deep fascia and iliotibial tract, exhibited the highest stiffness values with 133.88 kPa on the stiffer side and 93.48 kPa on the other side, resulting in an ΔE of 40.40 kPa. The iliotibial tract is a fibrous reinforcement of the fascia lata and has high stiffness values in healthy individuals, typically around 50 kPa (Otsuka et al., 2019). However, the ΔE of the $\mathrm{VL_f}$ in this study's population was noticeably high. There was a slight tendency for the dominant leg to have higher $\mathrm{VL_f}$ stiffness, but this trend did not reach statistical significance.

4.2.1 Strain elastography

The third hypothesis postulated that stiffness measurements, obtained as strain ratios and shear wave ratios, yield comparable results. Strain elastography is a semi-quantitative method to assess tissue stiffness by measuring strain. No significant SR difference was found between the dominant and nondominant leg and between the left and right leg. SWR did neither show any side differences (see tables 6.4, 6.5). SRs and SWRs were consistently above 1 in TFL, RFpr, VL, BF, GM and GL indicating higher stiffness in the fascia than in the muscle. SRs and SWRs were not in agreement in VM, ADD, TA and PL (see table 6.6). These results show a discrepancy between the two measuring methods (SE and SWE), which could be explained by the different physical principles underlying the two methods. Conventionally SRs are calculated between areas of similar depth but not between layers like fascia to muscle which are under different compressive stress due to the depth difference (Havre, Waage, Gilja, Ødegaard, & Nesje, 2011). In the published literature SRs have been calculated as the ratio between two muscles (Kwon, Park, Lee, & Chung, 2012) and between the ROI and a coupling agent of defined stiffness lying on the surface (Ariji et al., 2015). In this study the relative stiffness relation between the muscle and the fascia was evaluated. Moreover, it has to be considered that the compressive surface of the transducer used in this study was very narrow (5,5cm x 1cm) leading to an early decay of the applied strain (Cosgrove et al., 2013). This might account for the fact that the SR measurements between fascia and muscle are inaccurate since the highest compression was applied to the fascia. The surface of the transducer can be enlarged by using a footprint extender to facilitate a uniform distribution of strain within the ROI (Doyley, Bamber, Fuechsel, & Bush, 2001), which was not used in this study. Therefore SWE ratios can be estimated to be

more reliable as this method is not directly affected by the size of the transducer's surface.
Another comparability limiting factor is the form of stress generation (Kim, Park, & Lee, 2016). Manual rhythmic compressions for SE done by the operator are usually not standardized in amplitude and frequency. This can be compensated by using computer-controlled shakers (Pesavento et al., 1999), which, however, was not applied in the current study. In clinical practice, SE is a widely available tool for semi-quantitative estimation of muscle and fascia stiffness. Depending on the field of application, SWE holds more advantages, especially for clinical use, as measurements are quantitative and more precise, but SE is more frequently available (Ewertsen, Carlsen, Christiansen, Jensen, & Nielsen, 2016). Depending on the clinical question and the examiner's experience, elastography examinations with SE can be performed within a short period of time and therefore be useful in routine use.

4.3 ROM and Myofascial Stiffness

The fourth hypothesis of this study proposed that a reduction in the ROM of the lower extremities is associated with increased stiffness. However, no side differences were observed within individuals. The study found that LLF, HIP_{IR}, ADF_a, ADF_p and HBD had significantly decreased ROM compared to reported values of healthy subjects in previous studies (Czaprowski et al., 2013; Gauvin et al., 1990; Paul; Burrows, 1965; Stubbs et al., 1993), see tables 6.7, 6.8. It should be noted that the reported values were not specifically based on soccer players but rather on individuals without a competitive sports background.

Lumbar rotation and lateral flexion
The literature indicates that LRO for males aged 20 to 30 years varies from 12.8° (Van Herp et al., 2000) to 16.6° (Peach et al., 1998), with an average of approximately 15° (Hindle et al., 1990). This study found no significant difference in LRO compared to the average value of 15°. A decrease in the lumbar range of motion indicates reduced flexibility in the contralateral long back extensor muscles and the ipsilateral multifidi and rotatores muscles, as noted in previous studies (Barclay, De Forest, & Stam, 1971; Dvorak, Vajda, Grob, & Panjabi, 1995; Fitzgerald, Wynveen, Rheault, & Rothschild, 1983).
The American Academy of Orthopaedic Surgeons (1965) defines a reference value of 35° for LLF. This study showed a significant decrease in LLF compared to the typical value of 35°. Another survey of 104 healthy male subjects aged 20-70 years, with a mean age of 40 years, found a LLF of 34.8 ± 6.4° on the left and 36.2 ± 5.3° on the right in the 20-29 age group (Dvorak et al., 1995). Stubbs et al. (1993) also investigated LLF in 55 male subjects, 15 of whom were in the age group of 25-34 years and had a LLF of 35.70 ± 7.99° to the left and 37.10 ± 8.59° to the right.

Hip internal and external rotation
According to the American Academy of Orthopaedic Surgeons (1965), 45° is considered the standard reference value for HIP_{IR} and HIP_{ER}. It is important to note that hip ROM can vary depending on the subject's position during the assessment. A study by Han et al. (2015) found that hip rotation was significantly higher when assessed in the prone position compared to the sitting position. In this study, mea-

surements were taken in the prone position, and a significant decrease in HIP_{IR} was observed in both legs with strong effect sizes. However, HIP_{ER} was not significantly decreased.
Compared to a study of 53 healthy male subjects under the age of 20, which reported HIP_{IR} and HIP_{ER} values of 50.3 ± 6.1° and 50.5 ± 6.1° respectively in the sitting position (Boone & Azen, 1979), the subjects in this study had lower hip ROM. This limited ROM has been linked to an increased risk of injury in previous studies (Ekstrand & Gillquist, 1982; Knapik et al., 1991). Although it is expected that the study population will experience overuse injuries at some point due to their limited hip ROM, an exact prediction cannot be made based on these results alone.

Straight leg raise test
The shortening of the hamstring muscles can be determined by the SLR test, as indicated by previous research (de Lucena, dos Santos Gomes, & Guerra, 2011; Santonja Medina, Sainz De Baranda Andujar, Rodriguez Garcia, Lopez Minarro, & Canteras Jordana, 2007). In this study, the SLR was lower in the nondominant leg but did not significantly deviate from the typical value of 80°. According to Boyd et al. (2012), inter-limb differences of up to 11 degrees are considered normal in healthy individuals. This suggests that the results of this study reflect a physiologically normal range of motion in the SLR.
According to the SWE results, there was no significant difference between the dominant and nondominant leg in the biceps femoris muscle, and there was no correlation between Young's modulus of BF and SLR. These findings suggest that both modalities agree in not finding a side difference. Rose (1991) reported SLR values of 73.7 ± 15.9° in the left leg and 74.1 ± 16.1 ° in the right leg for 15 female and 3 male participants aged 19.5 ± 4.6 years. Mitani and colleagues (2015) reported SLR values of 68.7 ± 10.4° in the left leg and 71.3 ± 9.7° in the right leg of 15 male subjects aged 25.4 ± 5.4 years. The SLR of the current study population was higher compared to the available data. This might be explained by regular stretching and flexibility training, which could compensate for the high mechanical demand of the hamstring muscles (Al Attar, Soomro, Sinclair, Pappas, & Sanders, 2017).

Active and passive ankle dorsiflexion
In this study, ADF_a was measured with the subjects in a supine position with the knee straight, and a general decrease in flexibility was observed. ADF_a evaluates the flexibility of the triceps surae muscle, including the gastrocnemius muscle, soleus muscle, and peroneus longus muscle (Bradley & Portas, 2007). A value below 20° is considered as an indication of the shortening of the triceps surae muscle (Barclay et al., 1971; Boone & Azen, 1979; Krivickas & Feinberg, 1996; Burrows, 1965). ADF_p was measured with the subjects in a prone position with the knee bent at 90° to evaluate the soleus muscle in isolation. ADF_p below 30 ° is indicative of muscle shortening (DiGiovanni et al., 2002; Soucie et al., 2011). A previous study on professional soccer players aged 25.6 ± 4.7 years reported ADF_a values of 18.4 ± 4.2° in uninjured players, and 18.7 ± 7.3° in injured players (Bradley & Portas, 2007). In the current study, the subjects showed lower ROM for both ADF_a and ADF_p. Since ankle dorsiflexion may be a potential risk factor for ankle sprains (Ekstrand & Gillquist, 1983), it is suggested to consider this parameter for screening purposes in follow-up examinations of injuries.

Heel-buttocks distance
The study found that HBD measurements were above 0cm in both legs, which is considered non-physiological and may indicate shortening of the rectus femoris muscle (Krivickas & Feinberg, 1996; Okamura et al., 2014). Okamura and colleagues (2014) examined 192 athlete skaters (92 male, 100 female, mean age 15.4 ± 1.8 years) and found a higher incidence of injury in skaters with increased HBD, FFD or decreased SLR. In line with that, Krivckas and Feinberg (1996) found an increased injury risk in 201 college athletes with a median age of 19.8 years (males) and 19.6 years (females) with lower ROM of the lower limb. Mitani et al. (2015) reported similar results with HBD of 3.6 ± 3.4cm in the right leg and 4.5 ± 4.3 cm in the left leg of 15 male subjects aged 25.4 ± 5.4 years.
HBD measurements in this study were not controlled by a strain gauge sensor, which may have led to variability in the data due to differences in pressure application. To compensate for this, the same examiner conducted all the measurements. Future studies should consider using pressure control to reduce the potential for variability in the data.

Finger-floor distance
The results showed no significant increase in FFD. FFD values above 0 cm are generally considered elevated, indicating reduced flexibility in the hamstring, hip, and back muscles (Czaprowski et al., 2013; Gauvin et al., 1990; Broer & Galles, 1958). The lack of significant increase in FFD, together with the almost normal SLR, suggests that the subjects may not have significant stiffness in their hamstring muscles at this point in time, which is also in accordance with SWE findings in BF.

4.4 Altered Static Alignment and Foot Pressure Distribution

The fifth hypothesis proposed that soccer players would have altered foot pressure and impaired static alignment related to myofascial stiffness. However, the results did not support the hypothesis of imbalances in the static alignment associated with the frontal plane (see table 6.9), which is consistent with the SWE and ROM findings that showed no significant differences between the dominant and nondominant leg (see tables 6.2, 6.4). On the other hand, the hypothesis that pedobarography would reveal a disbalance was confirmed. The pressure distribution in the forefoot and rearfoot showed a significant difference from the typical distribution, with the pressure on the forefoot being higher than on the rearfoot in our population (see tables 6.9, 6.10). The reference values were assessed in 106 healthy women (mean age 25 years, mean BMI 21 $\frac{kg}{m^2}$) and were in accordance with previously reported findings (Ohlendorf et al., 2019; Lalande, Vie, Weber, & Jammes, 2016). The association between postural and plantar asymmetries and ankle injuries is a topic of ongoing research, and the findings have been mixed. Azevedo et al. (2017) found plantar pressure asymmetries in soccer players with higher pressure in the nondominant foot, and suggested that these were adaptations to the mechanical demands of soccer practice. Some studies, such as Trop et al. (1984) and McGuine et al. (2000), have suggested that abnormal postural sway or balance may be a marker for increased susceptibility to ankle sprains in soccer and basketball players respectively.

However, other studies, such as Beynnon et al. (2001), have found no difference in postural sway between injured and uninjured athletes. Overall, it is still unclear if postural and plantar asymmetries are reliable predictors of ankle injuries, and more research is needed to understand the relationship between these factors.

4.5 Correlation between functional tests and Ultrasound Elastography

E of $RFdi_m$ correlated positively with HBD in the nondominant leg but failed to reach significance after the Bonferroni correction (see table 6.12). The higher the stiffness in $RFdi_m$, the higher was the HBD. SWE and ROM findings show a tendency towards increased RF stiffness in the population. ROM measurements showed a decrease in lumbar, hip, and ankle flexibility, without side differences and without a significant correlation to findings in SWE. ROM assessment is a traditional and widely available tool for evaluating the musculoskeletal system. However it remains questionable if this modality is sufficient in musculoskeletal assessment. ROM can indicate the condition of muscle groups, but it is rarely possible to draw conclusions about individual muscles. The study found a correlation between forefoot pressure and E of $RFdi_f$ in the nondominant leg, although it did not reach statistical significance after Bonferroni correction (see table 6.14). Forefoot pressure tends to increase when the body is shifted forward in the sagittal plane. Additionally, E of $RFdi_m$ was found to be higher in the nondominant leg, but again, without statistical significance. It is possible that the higher E of $RFdi_m$ in the nondominant leg resulted in a forward shift of the body. However, no correlation was found between E of $RFdi_m$ and forefoot pressure. Similar to ROM, the static alignment and foot pressure distribution assessment can be regarded as an extension to ultrasound elastography based evaluations but the results of this study did not show a correlation between postural assessment and SWE, and specific muscles or muscle groups can be more precisely evaluated by ultrasound elastography. Since this study did not measure any variables in the sagittal plane, it remains unclear whether muscle shortening results in a measurable shift in the sagittal plane.

4.6 Limitations

Currently, there are no universally accepted standardized examination protocols or technical specifications for push beam generation and signal recording in musculoskeletal ultrasound elastography. This lack of standardization could explain the divergent E-values reported in the literature. The present study aimed to minimize the impact of variables affecting stiffness measurements to increase the reproducibility of elastography results. However, there are a few important points that should be taken into account.
(1) The selection of anatomical positions for UE in this study was based on the motor endplate in the muscle belly, a method commonly used in electromyography (Murray, 1995) and other research studies (Taniguchi et al., 2015; Lee, Kim, & Lee, 2021; Taş et al., 2019). However, some researchers have questioned whether the stiffness values obtained in the muscle belly area alone accurately reflect the overall

stiffness of the muscle (Zhou et al., 2020). In upcoming research, it may be beneficial to conduct measurements across the entire muscle length rather than just at one position.
(2) Prior studies have typically measured stiffness in small, homogeneous areas using a ROI of a few square millimeters. The reliability of obtaining representative measurements for the entire muscle and its fascia from a tiny area is debatable. In contrast, this study analyzed both homogeneous and inhomogeneous areas using a ROI of several square centimeters to capture all possible stiffness levels in a significantly larger portion of the muscle. Therefore, comparisons with previous studies may not be valid.
(3) Different push beam generation methods and shear wave recording approaches are other factors to consider, as the impact on stiffness measurements still needs to be understood.
(4) Young's modulus of elasticity E is used to measure tissue elasticity. However, the equations that derive E from C_s assume the medium isotropic and incompressible, whereas myofascial tissue is anisotropic and partly compressible. As a result, estimated values of ρ and ν can lead to measurement inaccuracies, especially because the density ρ, and Poisson's Ratio ν for that matter, of muscle and fascia may differ.
(5) ROM measurements are not standardized and have a wide range of non-sport-specific standard values. Specific adaptations of the myofascial system occur in each sports activity, resulting in higher or lower ROM, making it difficult to determine if the ROM findings are pathological.
(6) Pedobarography is not commonly used, and there is a lack of data to determine if the pressure distribution between the forefoot and rearfoot, as suggested by the manufacturer, is a reliable reference value. The published reference values were only evaluated in women.

5 Conclusion

LA-SWE can be considered an extension of manual palpation because it provides an objective and more detailed assessment of stiffness through quantification. It has high inter- and intrarater reliability and requires minimal training since generating and recording shear waves is automated. The stiffness measurements obtained can be recorded and used to track changes over time. LA-SWE was found to be effective in evaluating stiffness differences within the same individual across six different muscles/fascias, regardless of leg dominance. Monitoring the short and long-term effects of therapeutic interventions and preventive measures could be facilitated through the use SWE. As more studies are conducted using the same technology and assessment protocols, it may become possible to establish normal stiffness values for muscles and fascias for specific age groups. Follow-up studies may also help to identify threshold stiffness values that predict the risk of injury.

The findings obtained with LA-SWE and SE were in partial agreement, possibly due to a certain degree of operator-dependency and technological differences in stress generation in SE. In order to fully utilize the potential of the more cost-effective and widely available SE technology, these factors need to be optimized. Currently, LA-SWE appears to be a valid and objective tool for follow-up studies to gain a better understanding of the role of myofascial stiffness in injury.

Tools such as ROM, postural assessment, and pedobarography are already established in clinical practice for evaluating injury risk, even without agreed-upon typical values for specific sports. However, since there were no correlations found between myofascial stiffness and these tools, it raises questions about whether measuring only one area of the muscle is representative enough of overall stiffness. Future studies that assess stiffness along the entire length of the myofascia may yield different results with a higher correlation between ROM, static alignment, pedobarography, and elastography measurements. Further research is needed to determine whether assessing stiffness using shear wave elastography can serve as a marker for predicting the risk of injury in future studies

6 Appendix

Table 6.1: Comparison of intraindividual Young's modulus

	Higher E[a]	**Lower E[a]**	**Delta**	**P**	**Effect size**
Tensor fasciae latae					
TFL_m	36.89 (13.45)	26.28 (9.31)	**10.61**	0.002[b]	-0.461[e]
TFL_f	55.21 (25.67)	39.32 (8.96)	**15.89**	<0.001*[b]	-0.539[e]
P	<0.001*[c]	<0.001*[c]			
Effect size	1.143[d]	1.557[d]			
Proximal rectus femoris					
$RFpr_m$	21.08 (5.54)	16.53 (4.08)	**4.55**	0.005[b]	-0.411[e]
$RFpr_f$	24.34 (14.58)	19.85 (8.60)	**4.49**	0.007[b]	-0.391[e]
P	0.014[b]	0.024[c]			
Effect size	-0.346[e]	0.744[d]			
Distal rectus femoris					
$RFdi_m$	38.53 (36.22)	21.50 (20.16)	**17.03**	0.009[b]	-0.376[e]
$RFdi_f$	27.21 (21.31)	19.36 (7.47)	**8.35**	0.013[b]	-0.351[e]
P	0.120[b]	0.324[b]			
Effect size	-0.186[e]	-0.072[e]			
Vastus medialis					
VM_m	14.61 (3.14)	11.98 (2.03)	**2.63**	0.003[b]	-0.436[e]
VM_f	14.75 (5.20)	11.17 (3.52)	**3.58**	0.003[b]	-0.436[e]
P	0.860[b]	0.915[c]			
Effect size	0.171[e]	0.004[d]			
Vastus lateralis					
VL_m	33.42 (9.28)	25.34 (9.60)	**8.08**	0.003[b]	-0.438[e]
VL_f	133.88 ± 34.78	93.48 ± 26.29	**40.40**	<0.001*[c]	1.311[d]
P	<0.001*[c]	<0.001*[c]			
Effect size	3.837[d]	3.454[d]			
Tibialis anterior					
TA_m	37.75 ± 5.40	31.75 ± 4.50	**6.00**	<0.001*[c]	1.208[d]
TA_f	96.71 ± 32.74	70.94 ± 18.88	**25.77**	0.006[b]	-0.396[e]
P	<0.001*[c]	<0.001*[c]			
Effect size	2.513[d]	2.856[d]			
Peroneus longus					
PL_m	31.09 (9.62)	24.56 (5.87)	**6.53**	0.005[b]	-0.411[e]
PL_f	96.82 ± 19.85	64.38 ± 16.70	**32.44**	<0.001*[c]	1.768[d]
P	<0.001*[c]	<0.001*[c]			
Effect size	3.886[d]	3.024[d]			

Table 6.1: Comparison of intraindividual Young's modulus

	Higher E[a]	Lower E[a]	Delta	P	Effect size
Gluteus medius					
$GMED_m$	24.86 (12.44)	16.11 (6.53)	**7.24**	0.004[b]	-0.415[e]
$GMED_f$	31.17 (50.60)	14.65 (10.91)	**16.52**	0.002[b]	-0.451[e]
P	0.060[b]	0.598[b]			
Effect size	-0.246[e]	0.039[e]			
Gluteus maximus					
$GMAX_m$	23.56 (5.87)	17.88 (6.62)	**5.68**	0.008[c]	0.886[d]
$GMAX_f$	10.55 (2.74)	9.07 (1.98)	**1.48**	0.035[b]	-0.286[e]
P	<0.001*[c]	<0.001*[c]			
Effect size	2.900[d]	2.404[d]			
Adductor longus & adductor magnus					
ADD_m	17.88 (3.50)	14.65 (4.05)	**3.23**	0.007[b]	-0.390[e]
ADD_f	16.36 (5.42)	13.88 (4.97)	**2.48**	0.049[b]	-0.261[e]
P	0.268[c]	0.660[c]			
Effect size	0.356[d]	0.140[d]			
Biceps femoris					
BF_m	19.98 (4.98)	17.40 (2.70)	**2.58**	0.040[b]	-0.420[e]
BF_f	40.64 ± 15.73	30.46 ± 13.37	**10.18**	0.033[c]	0.698[d]
P	<0.001*[c]	<0.001*[c]			
Effect size	1.632[d]	1.302[d]			
Medial head of gastrocnemius					
GM_m	21.79 ± 5.39	17.97 ± 4.40	**3.82**	0.017[b]	-0.336[e]
GM_f	47.82 ± 17.49	32.00 ± 9.13	**15.82**	0.001*[b]	-0.471[e]
P	<0.001*[c]	<0.001*[b]			
Effect size	2.012[d]	-0.724[e]			
Lateral head of gastrocnemius					
GL_m	21.24 ± 4.32	16.99 ± 3.86	**4.25**	0.002[b]	-0.447[e]
GL_f	51.31 ± 15.95	34.83 ± 11.52	**16.48**	<0.001*[c]	1.184[d]
P	<0.001*[c]	<0.001*[c]			
Effect size	2.573[d]	2.076[d]			

[a] E (kPa). Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] r effect size.
* Significant difference (level of significance after Bonferroni correction <0.002).

Table 6.2: Young's modulus in dominant and non-dominant leg

	Dominant leg[a]	Non-dominant leg[a]	P	Effect size
Tensor fasciae latae				
$\mathrm{TFL}_{\mathrm{m}}$	32.54 (14.20)	30.82 (10.60)	0.806[c]	0.078[d]
$\mathrm{TFL}_{\mathrm{f}}$	43.10 (20.72)	44.08 (16.23)	0.484[c]	0.224[d]
P	<0.001*[b]	0.001*[c]		
Effect size	-0.506[e]	1.119[d]		
Proximal rectus femoris				
$\mathrm{RFpr}_{\mathrm{m}}$	17.48 (4.96)	20.12 (7.09)	0.930[c]	0.028[d]
$\mathrm{RFpr}_{\mathrm{f}}$	21.36 (9.17)	22.96 (12.71)	0.640[b]	0.057[e]
P	0.057[c]	0.028[b]		
Effect size	0.619[d]	-0.301[e]		
Distal rectus femoris				
$\mathrm{RFdi}_{\mathrm{m}}$	30.51 (38.68)	27.86 (20.27)	0.600[c]	0.169[d]
$\mathrm{RFdi}_{\mathrm{f}}$	26.79 (18.77)	22.05 (15.50)	0.896[c]	0.042[d]
P	0.262[b]	0.102[b]		
Effect size	-0.101[e]	-0.201[e]		
Vastus medialis				
VM_{m}	13.53 (1.98)	13.79 (4.55)	0.828[c]	0.069[d]
VM_{f}	12.10 (6.30)	13.42 (5.08)	0.925[b]	0.228[e]
P	0.379[b]	0.735[b]		
Effect size	-0.049[e]	0.099[e]		
Vastus lateralis				
VL_{m}	29.75 (12.62)	28.35 (7.89)	0.632[c]	0.153[d]
VL_{f}	122.61 ± 41.20	104.75 ± 29.93	0.125[c]	0.496[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	3.016[d]	3.354[d]		
Tibialis anterior				
TA_{m}	35.90 ± 6.81	33.59 ± 4.38	0.210[c]	0.403[d]
TA_{f}	89.05 ± 35.32	78.60 ± 21.74	0.267[c]	0.356[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	2.090[d]	2.869[d]		
Peroneus longus				
PL_{m}	26.67 (4.44)	29.49 (11.26)	0.387[b]	-0.045[e]
PL_{f}	75.69 (41.54)	81.93 (31.39)	0.445[c]	0.244[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	3.000[d]	2.414[d]		
Gluteus medius				
$\mathrm{GMED}_{\mathrm{m}}$	18.31 ± 5.46	23.38 ± 8.61	0.034[c]	0.670[d]
$\mathrm{GMED}_{\mathrm{f}}$	19.62 (11.87)	26.53 (55.31)	0.277[b]	-0.094[e]
P	0.561[b]	0.409[b]		
Effect size	0.024[e]	-0.036[e]		
Gluteus maximus				
$\mathrm{GMAX}_{\mathrm{m}}$	19.91 ± 5.30	20.73 ± 5.05	0.618[c]	0.159[d]
$\mathrm{GMAX}_{\mathrm{f}}$	9.72 ± 1.87	10.30 ± 3.77	0.538[c]	0.196[d]
P	<0.001*[c]	<0.001*[b]		
Effect size	2.563[d]	-0.733[e]		

Table 6.2: Young's modulus in dominant and non-dominant leg

	Dominant leg[a]	**Non-dominant leg[a]**	**P**	**Effect size**
Adductor longus & adductor magnus				
ADD_m	15.73 (3.19)	17.28 (5.04)	0.178[c]	0.434[d]
ADD_f	15.10 (5.12)	15.53 (6.55)	0.857[c]	0.057[d]
P	0.981[c]	0.171[c]		
Effect size	0.007[d]	0.441[d]		
Biceps femoris				
BF_m	18.54 (3.54)	19.35 (6.31)	0.524[c]	0.203[d]
BF_f	34.24 ± 14.36	36.86 ± 16.46	0.594[c]	0.170[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	1.425[d]	1.371[d]		
Medial head of gastrocnemius				
GM_m	20.79 ± 5.58	18.97 ± 4.81	0.276[c]	0.350[d]
GM_f	41.76 (21.91)	31.56 (16.13)	0.105[c]	0.525[d]
P	<0.001*[c]	<0.001*[b]		
Effect size	2.003[d]	-0.693[e]		
Lateral head of gastrocnemius				
GL_m	19.75 ± 4.17	18.48 ± 4.98	0.385[c]	0.278[d]
GL_f	46.30 ± 16.80	39.83 ± 15.02	0.207[c]	0.406[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	2.169[d]	1.908[d]		

[a] E (kPa). Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] r effect size.
[*] Significant difference (level of significance after Bonferroni correction 0.002).

Table 6.3: Young's modulus in left and right leg

	Left leg[a]	Right leg[a]	P	Effect size
Tensor fasciae latae				
TFL_m	30.82 (13.65)	32.03 (9.63)	0.462[c]	0.235[d]
TFL_f	43.33 (17.75)	43.85 (20.06)	0.897[c]	0.041[d]
P	0.014[c]	<0.001*[b]		
Effect size	0.813[d]	-0.613[e]		
Proximal rectus femoris				
$RFpr_m$	20.62 ± 8.34	17.36 ± 3.56	0.053[b]	-0.255[e]
$RFpr_f$	24.19 (8.91)	20.46 (9.02)	0.060[b]	-0.246[e]
P	0.019[c]	0.022[c]		
Effect size	0.776[d]	0.756[d]		
Distal rectus femoris				
$RFdi_m$	26.46 (38.98)	29.51 (22.73)	0.490[c]	0.220[d]
$RFdi_f$	22.46(20.35)	22.91 (14.12)	0.683[c]	0.130[d]
P	0.239[b]	0.127[c]		
Effect size	-0.112[e]	0.493[d]		
Vastus medialis				
VM_m	13.67 (3.04)	13.43 (3.60)	0.304[c]	0.330[d]
VM_f	13.73 (4.70)	11.53 (5.60)	0.035[b]	-0.286[e]
P	0.882[b]	0.728[c]		
Effect size	0.187[e]	0.111[d]		
Vastus lateralis				
VL_m	28.65 (11.03)	30.34 (9.52)	0.854[c]	0.059[d]
VL_f	111.52 ± 37.12	115.84 ± 37.04	0.714[c]	0.117[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	3.011[d]	3.066[d]		
Tibialis anterior				
TA_m	34.86 ± 5.21	34.63 ± 6.41	0.903[c]	0.039[d]
TA_f	91.28 ± 34.2	76.37 ± 22.16	0.110[c]	0.517[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	2.306[d]	2.558[d]		
Peroneus longus				
PL_m	27.12 (8.57)	27.94 (8.57)	0.685[b]	0.076[e]
PL_f	82.64 ± 27.23	78.56 ± 21.83	0.604[c]	0.166[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	2.413[d]	3.112[d]		
Gluteus medius				
$GMED_m$	21.75 ± 7.81	19.95 ± 7.54	0.464[c]	0.234[d]
$GMED_f$	23.10 (38.73)	17.46 (27.66)	0.327[b]	-0.071[e]
P	0.020[b]	0.968[b]		
Effect size	-0.134[e]	0.292[e]		
Gluteus maximus				
$GMAX_m$	20.33 ± 4.75	20.31 ± 5.60	0.989[c]	0.004[d]
$GMAX_f$	9.38 (2.33)	10.28 (2.88)	0.570[c]	0.181[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	2.871[d]	2.125[d]		

Table 6.3: Young's modulus in left and right leg

	Left leg[a]	**Right leg[a]**	**P**	**Effect size**
Adductor longus & adductor magnus				
ADD_m	15.79 (3.21)	17.24 (4.12)	0.808[c]	0.077[d]
ADD_f	13.98 (4.73)	15.86 (5.73)	0.308[c]	0.327[d]
P	0.279[c]	0.695[c]		
Effect size	0.347[d]	0.125[d]		
Biceps femoris				
BF_m	18.41 (4.59)	19.35 (4.44)	0.415[c]	0.260[d]
BF_f	31.99(14.39)	32.57 (19.35)	0.990[c]	0.004[d]
P	<0.001*[b]	<0.001*[c]		
Effect size	-0.746[e]	1.204[d]		
Medial head of gastrocnemius				
GM_m	20.42 ± 6.24	19.52 ± 4.11	0.860[b]	0.171[e]
GM_f	33.70 (16.04)	38.27 (22.14)	0.698[c]	0.124[d]
P	<0.001*[b]	<0.001*[b]		
Effect size	-0.693[e]	-0.746[e]		
Lateral head of gastrocnemius				
GL_m	18.92 ± 4.34	19.31 ± 4.91	0.795[c]	0.083[d]
GL_f	42.22 ± 17.55	43.91 ± 14.85	0.745[c]	0.104[d]
P	<0.001*[c]	<0.001*[c]		
Effect size	1.823[d]	2.224[d]		

[a] E (kPa).Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] r effect size.
[*] Significant difference (level of significance after Bonferroni correction 0.002).

Table 6.4: Strain and shear wave ratios in the dominant and nondominant leg

	Strain Ratio[a,f]	SW Ratio[a,g]	P	Effect size
Tensor fasciae latae				
TFL dominant leg	1.80 (1.12)	1.49 (0.54)	0.343[d]	0.304[e]
TFL nondominant leg	1.83 (1.63)	1.51 (0.35)	0.142[b]	-0.170[c]
P	0.665[b]	0.758[b]		
Effect size	0.067[c]	0.111[c]		
Proximal rectus femoris				
RFpr dominant leg	2.88 (1.37)	1.26 ± 0.27	<0.001*[b]	-0.786[c]
RFpr nondominant leg	2.56 (2.10)	1.31 ± 0.26	<0.001*[b]	-0.717[c]
P	0.445[b]	0.547[d]		
Effect size	-0.022[c]	0.192[e]		
Distal rectus femoris				
RFdi dominant leg	1.15 (0.48)	0.82 (0.42)	0.003[b]	-0.436[c]
RFdi nondominant leg	1.01 (0.40)	0.87 (0.24)	0.018[b]	-0.332[c]
P	0.534[b]	0.718[b]		
Effect size	0.013[c]	0.091[c]		
Vastus medialis				
VM dominant leg	2.36 ± 0.95	0.96 (0.33)	<0.001*[d]	1.877[e]
VM nondominant leg	1.89 ± 0.85	0.95 (0.29)	<0.001*[d]	1.163[e]
P	0.105[d]	0.758[b]		
Effect size	0.526[e]	0.111[c]		
Vastus lateralis				
VL dominant leg	3.07 (1.87)	3.96 ± 1.17	0.135[d]	0.483[e]
VL nondominant leg	2.10 (0.68)	3.57 ± 1.09	0.005[d]	0.951[e]
P	0.117[d]	0.289[d]		
Effect size	0.507[e]	0.340[e]		
Tibialis anterior				
TA dominant leg	1.17 (1.31)	2.22 (1.36)	0.027[d]	0.729[e]
TA nondominant leg	1.04 (1.05)	2.23 (0.98)	<0.001*[d]	1.600[e]
P	0.598[b]	0.989[b]		
Effect size	0.039[c]	0.364[c]		
Peroneus longus				
PL dominant leg	1.38 (1.20)	3.16 ± 1.12	<0.001*[d]	1.486[e]
PL nondominant leg	1.22 (0.78)	2.70 ± 1.11	0.002[d]	1.033[e]
P	0.902[d]	0.209[d]		
Effect size	0.039[e]	0.404[e]		
Gluteus medius				
GMED dominant leg	1.16 (0.46)	1.19 (0.79)	0.776[b]	0.120[c]
GMED nondominant leg	1.18 (0.56)	1.41 (1.73)	0.249[d]	0.371[e]
P	0.946[b]	0.315[d]		
Effect size	0.254[c]	0.322[e]		
Gluteus maximus				
GMAX dominant leg	1.14 (0.43)	0.48 (0.16)	<0.001*[d]	2.322[e]
GMAX nondominant leg	1.16 (0.41)	0.50 (0.16)	<0.001*[b]	-0.931[c]
P	0.630[d]	0.478[b]		
Effect size	0.154[e]	-0.009[c]		

Table 6.4: Strain and shear wave ratios in the dominant and nondominant leg

	Strain Ratio[a,f]	**SW Ratio[a,g]**	**P**	**Effect size**
Adductor longus & adductor magnus				
ADD dominant leg	1.50 (0.94)	0.99 (0.23)	<0.001*[b]	-0.680[c]
ADD nondominant leg	1.35 (0.83)	0.85 (0.34)	0.003[b]	-0.436[c]
P	0.181[d]	0.190[d]		
Effect size	0.431[e]	0.422[e]		
Biceps femoris				
BF dominant leg	1.94 ± 1.23	1.76 ± 0.54	0.547[d]	0.192[e]
BF nondominant leg	1.86 ± 1.05	1.80 ± 0.50	0.818[d]	0.073[e]
P	0.825[d]	0.802[d]		
Effect size	0.070[e]	0.080[e]		
Medial head of gastrocnemius				
GM dominant leg	1.51 (1.02)	1.90 (0.92)	0.554[d]	0.189[e]
GM nondominant leg	1.58 (0.77)	2.23 (1.92)	0.144[d]	0.472[e]
P	0.183[d]	0.475[d]		
Effect size	0.429[e]	0.228[e]		
Lateral head of gastrocnemius				
GL dominant leg	1.46 ± 0.65	2.34 ± 0.72	<0.001*[d]	1.279[e]
GL nondominant leg	2.40 ± 1.47	2.15 ± 0.55	0.465[d]	0.233[e]
P	0.012[d]	0.353[d]		
Effect size	0.831[e]	0.297[e]		

[a] Values are presented as median (IQR) or mean ± SD .
[b] From Wilcoxon signed-rank test.
[c] r effect size.
[d] From paired t-test.
[e] Cohen's d effect size.
[f] Strain ratio calculated as strain of the muscle divided by strain of the fascia.
[g] Shear wave ratio calculated as Young's modulus of the fascia divided by Young's modulus of the muscle.
[*] Significant difference (level of significance after Bonferroni correction 0.002).

Table 6.5: Strain and shear wave ratios in the left and right leg

	Strain Ratio[a,f]	**SW Ratio[a,g]**	**P**	**Effect size**
Tensor fasciae latae				
TFL left leg	1.83 (1.56)	1.46 (0.47)	0.043[d]	0.662[e]
TFL right leg	1.80 (1.16)	1.52 (0.48)	0.425[d]	0.255[e]
P	0.482[b]	0.471[d]		
Effect size	-0.007[c]	0.230[e]		
Proximal rectus femoris				
RFpr left leg	2.85 (1.15)	1.33 ± 0.26	<0.001*[d]	1.777[e]
RFpr right leg	2.69 (2.33)	1.24 ± 0.27	<0.001*[b]	-0.786[c]
P	0.862[b]	0.278[d]		
Effect size	0.172[c]	0.348[e]		
Distal rectus femoris				
RFdi left leg	1.05 (0.56)	0.89 (0.25)	0.006[d]	0.918[e]
RFdi right leg	1.10 (0.36)	0.81 (0.40)	0.004[b]	-0.421[c]
P	0.797[b]	0.904[b]		
Effect size	0.131[c]	0.206[c]		
Vastus medialis				
VM left leg	1.89 ± 0.85	1.02 (0.25)	0.001*[d]	1.056[e]
VM right leg	2.36 ± 0.95	0.92 (0.30)	<0.001*[d]	2.010[e]
P	0.109[d]	0.252[b]		
Effect size	0.519[e]	-0.116[c]		
Vastus lateralis				
VL left leg	2.41 (1.88)	3.62 ± 1.00	0.034[d]	0.694[e]
VL right leg	2.41 (1.79)	3.91 ± 1.26	0.044[d]	0.658[e]
P	0.787[b]	0.432[d]		
Effect size	0.126[c]	0.251[e]		
Tibialis anterior				
TA left leg	1.15 (1.12)	2.73 ± 1.27	0.005[d]	0.937[e]
TA right leg	1.07 (1.16)	2.24 ± 0.65	<0.001*[d]	1.155[e]
P	0.518[d]	0.131[d]		
Effect size	0.206[e]	0.488[e]		
Peroneus longus				
PL left leg	1.52 (1.57)	2.96 ± 1.28	0.014[d]	0.813[e]
PL right leg	1.10 (0.69)	2.90 ± 0.97	<0.001*[d]	2.124[e]
P	0.048[d]	0.876[d]		
Effect size	0.645[e]	0.050[e]		
Gluteus medius				
GMED left leg	1.05 (0.42)	1.42 (0.92)	0.090[d]	0.550[e]
GMED right leg	1.27 (0.48)	1.09 (1.12)	0.841[b]	0.158[c]
P	0.310[b]	0.843[d]		
Effect size	-0.078[c]	0.063[e]		
Gluteus maximus				
GMAX left leg	1.14 (0.40)	0.47 (0.14)	<0.001*[d]	2.570[e]
GMAX right leg	1.15 (0.46)	0.51 (0.19)	<0.001*[d]	2.09[e]
P	0.674[d]	0.183[b]		
Effect size	0.134[e]	-0.143[c]		

Table 6.5: Strain and shear wave ratios in the left and right leg

	Strain Ratio[a,f]	**SW Ratio[a,g]**	**P**	**Effect size**
Adductor longus & adductor magnus				
ADD left leg	1.40 (0.72)	0.92 ± 0.16	<0.001*[b]	-0.627[c]
ADD right leg	1.62 (0.83)	0.98 ± 0.24	0.001*[d]	1.095[e]
P	0.490[d]	0.350[d]		
Effect size	0.220[e]	0.300[e]		
Biceps femoris				
BF left leg	1.81 (1.82)	1.84 ± 0.43	0.294[d]	0.336[e]
BF right leg	1.64 (1.33)	1.72 ± 0.59	0.754[d]	0.100[e]
P	0.150[d]	0.473[d]		
Effect size	0.464[e]	0.229[e]		
Medial head of gastrocnemius				
GM left leg	2.15 (1.91)	1.82 (0.94)	0.347[d]	0.301[e]
GM right leg	1.73 (1.21)	1.89 (0.84)	0.994[d]	0.002[e]
P	0.215[d]	0.654[d]		
Effect size	0.399[e]	0.143[e]		
Lateral head of gastrocnemius				
GL left leg	1.54 (0.88)	2.08 (0.85)	0.046*[d]	0.651[e]
GL right leg	1.76 (1.71)	2.25 (0.65)	0.114[b]	-0.190[c]
P	0.262[d]	0.476[d]		
Effect size	0.360[e]	0.227[e]		

[a] Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] r effect size.
[d] From paired t-test.
[e] Cohen's d effect size.
[f] Strain ratio calculated as strain of the muscle divided by strain of the fascia.
[g] Shear wave ratio calculated as Young's modulus of the fascia divided by Young's modulus of the muscle.
[*] Significant difference (level of significance after Bonferroni correction 0.002).

Table 6.6: Strain ratios and shear wave ratios

	SR = 1, P	**Effect size**	**SWR = 1, P**	**Effect size**
Tensor fasciae latae				
TFL dominant leg	<0.001*[b]	0.977[c]	<0.001*[a]	-1.033[d]
TFL nondominant leg	<0.001*[a]	-0.763[d]	<0.001*[b]	1.405[c]
TFL left leg	<0.001*[a]	-0.815[d]	<0.001*[b]	1.074[c]
TFL right leg	<0.001*[b]	0.907[c]	<0.001*[a]	-1.033[d]
Proximal rectus femoris				
RFpr dominant leg	<0.001*[a]	-1.033[d]	<0.001*[b]	0.963[c]
RFpr nondominant leg	<0.001*[a]	-1.001[d]	<0.001*[b]	1.191[c]
RFpr left leg	<0.001*[b]	1.550[c]	<0.001*[b]	1.308[c]
RFpr right leg	<0.001*[a]	-1.033[d]	<0.001*[b]	0.882[c]
Distal rectus femoris				
RFdi dominant leg	0.031[b]	0.521[c]	0.024[a]	-0.443[d]
RFdi nondominant leg	0.338[a]	-0.093[d]	0.006[b]	0.698[c]
RFdi left leg	0.135[b]	0.349[c]	0.002[a]	-0.644[d]
RFdi right leg	0.083[a]	-0.310[d]	0.024[a]	-0.442[d]
Vastus medialis				
VM dominant leg	<0.001*[b]	1.428[c]	0.644[b]	0.105[c]
VM nondominant leg	<0.001*[b]	1.046[c]	0.927[a]	0.326[d]
VM left leg	<0.001*[b]	1.045[c]	0.452[a]	-0.027[d]
VM right leg	<0.001*[b]	1.428[c]	0.554[b]	0.135[c]
Vastus lateralis				
VL dominant leg	<0.001*[a]	-1.001[d]	<0.001*[b]	2.524[c]
VL nondominant leg	<0.001*[a]	-0.791[d]	<0.001*[b]	2.358[c]
VL left leg	<0.001*[a]	-0.825[d]	<0.001*[b]	2.622[c]
VL right leg	<0.001*[a]	-0.921[d]	<0.001*[b]	2.300[c]
Tibialis anterior				
TA dominant leg	0.189[a]	-0.197[d]	<0.001*[a]	-1.033[d]
TA nondominant leg	0.224[b]	0.281[c]	<0.001*[b]	1.950[c]
TA left leg	0.189[a]	-0.197[d]	<0.001*[b]	1.361[c]
TA right leg	0.176[b]	0.314[c]	<0.001*[b]	1.898[c]
Peroneus longus				
PL dominant leg	0.056[a]	-0.355[d]	<0.001*[b]	1.927[c]
PL nondominant leg	0.056[a]	-0.357[d]	<0.001*[b]	1.534[c]
PL left leg	0.010[a]	-0.519[d]	<0.001*[b]	1.529[c]
PL right leg	0.154[b]	0.331[c]	<0.001*[b]	1.950[c]
Gluteus medius				
GMED dominant leg	0.067[a]	-0.334[d]	0.090[a]	-0.300[d]
GMED nondominant leg	0.123[a]	-0.259[d]	0.040[a]	-0.391[d]
GMED left leg	0.317[a]	-0.107[d]	0.012[a]	-0.504[d]
GMED right leg	0.027[a]	-0.432[d]	0.189[a]	-0.197[d]
Gluteus maximus				
GMAX dominant leg	0.053[a]	-0.361[d]	<0.001*[b]	4.029[c]
GMAX nondominant leg	0.005[a]	-0.578[d]	<0.001*[a]	-1.001[d]
GMAX left leg	0.017[a]	-0.473[d]	<0.001*[b]	4.770[c]
GMAX right leg	0.036[a]	-0.432[d]	<0.001*[a]	-1.033[d]

Table 6.6: Strain ratios and shear wave ratios

	SR = 1, P	**Effect size**	**SWR = 1, P**	**Effect size**
Adductor longus & adductor magnus				
ADD dominant leg	<0.001*[a]	-0.782[d]	0.813[b]	0.054[c]
ADD nondominant leg	0.006[a]	-0.557[d]	0.075[b]	0.422[c]
ADD left leg	<0.001*[a]	-0.721[d]	0.037[b]	0.500[c]
ADD right leg	<0.001*[a]	-0.714[d]	0.689[b]	0.091[c]
Biceps femoris				
BF dominant leg	0.003[b]	0.767[c]	<0.001*[b]	1.417[c]
BF nondominant leg	0.002[b]	0.826[c]	<0.001*[b]	1.594[c]
BF left leg	<0.001*[a]	-0.829[d]	<0.001*[b]	1.960[c]
BF right leg	0.005[b]	0.701[c]	<0.001*[b]	1.219[c]
Medial head of gastrocnemius				
GM dominant leg	<0.001*[a]	-0.738[d]	<0.001*[b]	1.368[c]
GM nondominant leg	<0.001*[b]	1.105[c]	<0.001*[a]	1.033[d]
GM left leg	<0.001*[b]	1.097[c]	<0.001*[a]	-1.033[d]
GM right leg	0.001*[a]	-0.690[d]	<0.001*[a]	-1.033[d]
Lateral head of gastrocnemius				
GL dominant leg	0.005[b]	0.708[c]	<0.001*[b]	1.866[c]
GL nondominant leg	<0.001*[b]	0.957[c]	<0.001*[b]	2.071[c]
GL left leg	<0.001*[b]	0.909[c]	<0.001*[b]	1.967[c]
GL right leg	<0.001*[a]	-0.714[d]	<0.001*[a]	-1.033[d]

[a] From Wilcoxon signed-rank test.
[b] From one-sample t-test.
[c] Cohen's d effect size.
[d] r effect size.
[*] Significant difference (level of significance after Bonferroni correction <0.002).

Table 6.7: Range of motion testing dominant vs. non-dominant side

	ROM value[a]	**Typical value**	**P**	**Effect size**
Lumbar rotation [°]				
LRO dominant side	11.0 (2.25)	15.0	0.014[b]	-0.492[f]
LRO nondominant side	10.0 (7.25)	15.0	0.057[b]	-0.354[f]
P	0.296[b]			
Effect size	0.269[f]			
Lumbar lateral flexion [°]				
LLF dominant side	30.0 (6.0)	35.0	<0.001*[b]	-0.738[f]
LLF nondominant side	30.5 (10.5)	35.0	<0.001*[b]	0.923[f]
P	0.887[d]			
Effect size	0.045[e]			
Hip internal rotation [°]				
HIP_{IR} dominant leg	34.3 ± 6.0	45.0	<0.001*[c]	1.797[e]
HIP_{IR} nondominant leg	34.6 ± 6.6	45.0	<0.001*[c]	1.609[e]
P	0.940[d]			
Effect size	0.024[e]			
Hip external rotation [°]				
HIP_{ER} dominant leg	41.4 ± 9.1	45.0	0.088[c]	0.402[e]
HIP_{ER} nondominant leg	39.4 ± 9.0	45.0	0.011[c]	0.626[e]
P	0.489[d]			
Effect size	0.221[e]			
Straight leg raise [°]				
SLR dominant leg	83.3 ± 7.3	80.0	0.057[c]	0.453[e]
SLR nondominant leg	79.6 ± 8.9	80.0	0.843[c]	0.045[e]
P	0.158[b]			
Effect size	0.455[f]			
Active ankle dorsiflexion [°]				
ADF_a dominant leg	11.5 (7.0)	20.0	<0.001*[b]	0.771[f]
ADF_a nondominant leg	12.0 (7.8)	20.0	<0.001*[b]	0.717[f]
P	0.615[d]			
Effect size	0.160[e]			
Passive ankle dorsiflexion [°]				
ADF_p dominant leg	23.9 ± 5.2	30.0	<0.001*[c]	1.173[e]
ADF_p nondominant leg	24.3 ± 6.8	30.0	0.001*[c]	0.840[e]
P	0.837[d]			
Effect size	0.066[e]			
Heel-buttock distance [cm]				
HBD dominant leg	3.0 (7.0)	0.0	0.002*[b]	0.628[f]
HBD nondominant leg	5.5 (6.3)	0.0	0.002*[b]	0.633[f]
P	0.737[b]			
Effect size	0.100[f]			
Finger-floor distance [cm]				
FFD	0.0 (0.0)	0.0	0.476[b]	-0.286[f]

[a] Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From one sample t-test.
[d] From paired t-test.
[e] Cohen's d effect size.
[f] r effect size.
[*] Significant difference (level of significance after Bonferroni correction <0.006).

Table 6.8: Range of motion testing left vs. right side

	ROM value[a]	Typical value	P	Effect size
Lumbar rotation [°]				
LRO left	11.0 (7.3)	15.0	0.049[b]	-0.371[f]
LRO right	10.0 (2.3)	15.0	0.018[b]	-0.471[f]
P	0.489[b]			
Effect size	-0.005[f]			
Lumbar lateral flexion [°]				
LLF left	30.5 (9.0)	35.0	<0.001*[b]	-0.802[f]
LLF right	30.0 (9.0)	35.0	<0.001*[b]	0.905[f]
P	0.849[d]			
Effect size	0.061[e]			
Hip internal rotation [°]				
HIP_{IR} left	35.1 ± 7.1	45.0	<0.001*[c]	1.402[e]
HIP_{IR} right	33.6 ± 5.3	45.0	<0.001*[c]	2.162[e]
P	0.437[d]			
Effect size	0.248[e]			
Hip external rotation [°]				
HIP_{ER} left	38.9 ± 8.9	45.0	0.006[c]	0.687[e]
HIP_{ER} right	41.8 ± 9.1	45.0	0.132[c]	0.352[e]
P	0.314[d]			
Effect size	0.323[e]			
Straight leg raise [°]				
SLR left	81.0 ± 7.2	80.0	0.544[c]	0.138[e]
SLR right	81.9 ± 9.3	80.0	0.373[c]	0.204[e]
P	0.735[d]			
Effect size	0.108[e]			
Active ankle dorsiflexion [°]				
ADF_a left	11.0 (3.8)	20.0	<0.001*[b]	-0.766[f]
ADF_a right	13.5 (6.8)	20.0	0.001*[b]	-0.694[f]
P	0.042[b]			
Effect size	-0.274[f]			
Passive ankle dorsiflexion [°]				
ADF_p left	24.1 ± 6.7	30.0	<0.001*[c]	0.889[e]
ADF_p right	24.1 ± 5.4	30.0	<0.001*[c]	1.093[e]
P	1.000[d]			
Effect size	0.0[e]			
Heel-buttock distance [*cm*]				
HBD left	5.5 (7.25)	0.0	0.002*[b]	-0.629[f]
HBD right	3.0 (6.0)	0.0	0.002*[b]	-0.629[f]
P	0.644[b]			
Effect size	0.058[f]			
Finger-floor distance [*cm*]				
FFD	0.0 (0.0)	0.0	0.100[b]	-0.286[f]

[a] Values are presented as median (IQR) or mean ± SD.
[b] From Wilcoxon signed-rank test.
[c] From one-sample t-test.
[d] From paired t-test.
[e] Cohen's d effect size.
[f] r effect size.
[*] Significant difference (level of significance after Bonferroni correction <0.006).

Table 6.9: Results of static alignment and pedobarography in the left and right leg

	Measured value[a]	Typical value	P	Effect size
Position of VP in the frontal plane[e]				
	0.06 ± 20.2	0.0	0.990[b]	0.003[d]
Positon of pelvis in the frontal plane[e]				
	2.77 ± 15.83	0.0	0.443[b]	0.175[d]
Pressure [%]				
Left leg	50.10 ± 5.30	50.0	0.934[b]	0.019[d]
Right leg	49.90 ± 5.30	50.0	0.934[b]	0.019[d]
P	0.906[c]			
Effect size	0.038[d]			
Pressure rearfoot [%]				
Left leg	47.05 ± 17.27	67.0	<0.001*[b]	1.155[d]
Right leg	48.10 ± 14.92	67.0	<0.001*[b]	1.266[d]
P	0.838[c]			
Effect size	0.065[d]			
Pressure forefoot [%]				
Left leg	52.95 ± 17.28	33.0	<0.001*[b]	1.155[d]
Right leg	51.90 ± 14.92	33.0	<0.001*[b]	1.266[d]
P	0.838[c]			
Effect size	0.065[d]			
Total force [N]				
Left leg	274.90 ± 49.77			
Right leg	273.85 ± 50.50			
P	0.938[c]			
Effect size	0.021[d]			

[a] Values are presented as mean ± SD.
[b] From one-sample t-test.
[c] From paired t-test.
[d] Cohen's d effect size.
[e] Positive values indicate deviation to the right, negative values deviation to the left.
* Significant difference (level of significance after Bonferroni correction <0.008).

Table 6.10: Results of pedobarography in the dominant and nondominant leg

	Measured value[a]	Typical value	P	Effect size
Pressure [%]				
Dominant leg	48.80 ± 5.16	50.0	0.311[b]	0.233[d]
Nondominant leg	51.20 ± 5.16	50.0	0.311[b]	0.233[d]
P	0.149[c]			
Effect size	0.465[d]			
Pressure rearfoot [%]				
Dominant leg	41.85 ± 12.88	67.0	<0.001*[b]	1.953[d]
Nondominant leg	53.30 ± 16.93	67.0	0.002*[b]	0.809[d]
P	0.021[c]			
Effect size	0.761[d]			
Pressure forefoot [%]				
Dominant leg	58.15 ± 12.88	33.0	<0.001*[b]	1.953[d]
Nondominant leg	46.70 ± 16.93	33.0	0.002*[b]	0.809[d]
P	0.021[c]			
Effect size	0.761[d]			
Total force [N]				
Dominant leg	269.15 ± 57.40			
Nondominant leg	279.60 ± 40.93			
P	0.511[c]			
Effect size	0.210[d]			

[a] Values are presented as mean ± SD.
[b] From one-sample t-test.
[c] From paired t-test.
[d] Cohen's d effect size.
* Significant difference (level of significance after Bonferroni correction <0.013).

Table 6.11: Spearman correlation between E in the muscle of the dominant leg, ROM and foot pressure

	RFpr	**RFdi**	**BF**	**HBD**	**SLR**
RFdi	-0.216				
BF	0.186	-0.148			
HBD	0.390	0.041	0.010		
SLR	-0.399	-0.083	0.020	-0.227	
Forefoot pressure	0.089	0.213	-0.105	0.139	0.006

* $p<0.05$
** $p<0.01$

Table 6.12: Spearman correlation between E in the muscle of the nondominant leg, ROM and foot pressure

	RFpr	**RFdi**	**BF**	**HBD**	**SLR**
RFdi	0.342				
BF	0.160	-0.418			
HBD	0.156	0.490*	0.00		
SLR	0.035	-0.004	-0.171	-0.081	
Forefoot pressure	0.275	0.366	0.063	0.132	0.149

* $p<0.05$
** $p<0.01$

Table 6.13: Spearman correlation between E in the fascia of the dominant leg, ROM and foot pressure

	RFpr	RFdi	BF	HBD	SLR
RFdi	-0.045				
BF	0.274	-0.104			
HBD	0.249	-0.118	-0.328		
SLR	-0.433	0.289	0.046	-0.227	
Forefoot pressure	0.032	0.286	-0.172	0.139	0.006

* p<0.05
** p<0.01

Table 6.14: Spearman correlation between E in the fascia of the nondominant leg, ROM and foot pressure

	RFpr	RFdi	BF	HBD	SLR
RFdi	0.203				
BF	0.244	-0.198			
HBD	0.186	0.410	-0.035		
SLR	-0.135	0.292	-0.116	-0.081	
Forefoot pressure	0.161	0.533*	0.225	0.132	0.149

* p<0.05
** p<0.01

Technische Universität München

Patienteninformation zur Studie:

„Vergleich von Strain-Elastografie gegen Strain Ratio Bestimmung in symptomfreien, jugendlichen Fußballspielern

Comparision of Strain-Elastography vs. Strain-Ratio measurements in symptom free juvenile soccer players"

Sehr geehrte Probandin, sehr geehrter Proband,

die nachfolgenden Zeilen dienen Ihrer Information.

Wir möchten Sie einladen, an einer Studie teilzunehmen, um die Ultraschall Elastografie als diagnostisches Verfahren in der Prävention von Sportverletzungen zu beurteilen.
Die Ultraschall Elastografie ist ein bildgebendes Verfahren zur Untersuchung der Elastizität verschiedener Gewebe wie Muskeln und Faszien. Es sollen Messwerte in Korrelation zu den Ergebnissen einer Range Of Motion Testung, also einer Bestimmung des Bewegungsumfangs von Muskeln und Gelenken, sowie einer Statikvermessung bestimmt werden.
Bei den Messmethoden handelt es sich um gängige orthopädische Verfahren, die schmerzfrei sind und bei denen Sie nicht in Kontakt mit schädlicher Strahlung gelangen.

Zum Ablauf der Studie:
Sie erhalten vor dem Untersuchungstag einen Anamnesebogen zum Ausfüllen. Am Untersuchungstag wird zunächst die Körperstatik (Wirbelsäule und Becken) und die Beweglichkeit der Lendenwirbelsäule und Beine gemessen.
Anschließend erfolgt die Messung mittels der Elastografie an 30 Stellen der unteren Extremität links und rechts. Dabei wird der Ultraschallkopf auf die zu untersuchenden Stellen des Muskels mit Ultraschallgel aufgelegt und rhythmisch bewegt. Für die Untersuchungsdauer sollten Sie zwei Stunden einplanen.

Außerdem möchten wir Sie um ihr Einverständnis bitten, dass wir Ihre Daten pseudonymisiert und streng vertraulich zur anonymen Auswertung der Untersuchung verwenden dürfen.

Eine separate Einverständniserklärung zum Datenschutz wird Ihnen zur Unterschrift ausgehändigt. Es ist nicht mit Nebenwirkungen oder Risiken im Rahmen der Studie zu rechnen.

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Die Teilnahme an der Studie ist freiwillig und kann jederzeit ohne Angabe von Gründen widerrufen werden. Es entstehen Ihnen durch den Ausstieg aus der Studie keine Nachteile.

Sollten Sie minderjährig sein, muss die Einverständniserklärung zusätzlich von einem Erziehungsberechtigten unterschrieben werden.

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Comparision of Strain-Elastography vs. Strain-Ratio measurements in symptom free juvenile soccer players"

Patienteneinverständniserklärung

Ich bin über die im Rahmen dieser Studie geplanten Untersuchungen und deren Risiken ausführlich aufgeklärt worden. Die beiliegende Probandeninformation habe ich erhalten, gelesen und verstanden. Meine Fragen wurden umfassend und verständlich beantwortet.

◌ Ich erkläre meine Teilnahme.
◌ Ich nehme an der Studie nicht teil.

____
Name Proband(-in)

______________	____________________	__________________________
Datum und Ort	Unterschrift Proband(-in)	Unterschrift Erziehungsberechtigte(r)

______________	____________________
Datum und Ort	Unterschrift Projektleitung

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Einwilligungserklärung zur Datenerhebung

In dieser Studie ist Katharina Bauermeister (Technische Universität München, k.bauermeister@tum.de) für die Datenverarbeitung verantwortlich. Ihre Daten werden ausschließlich im Rahmen dieser Studie verwendet.
Dazu gehören personenidentifizierende Daten wie Name, Anschrift und sensible personenbezogene Gesundheitsdaten.

Alle unmittelbar Ihre Person identifizierenden Daten (Name, Geburtsdatum, Anschrift) werden durch einen Identifizierungscode ersetzt (pseudonymisiert). Dies schließt eine Identifizierung Ihrer Person durch Unbefugte weitgehend aus.

Ihre Daten werden auf einem PC mit Zugriffsbeschränkung gespeichert. Sie werden nach Ablauf nach Ablauf der gesetzlichen Löschfristen gelöscht.

Die Einwilligung zur Verarbeitung Ihrer Daten ist freiwillig, Sie können jederzeit die Einwilligung ohne Angabe von Gründen und ohne Nachteile für Sie widerrufen.

Sie haben das Recht, Auskunft über die Sie betreffenden Daten zu erhalten, auch in Form einer unentgeltlichen Kopie. Darüber hinaus können Sie die Berichtigung oder Löschung Ihrer Daten verlangen. Wenden Sie sich in diesen Fällen an Katharina Bauermeister, E-Mail k.bauermeister@tum.de, Telefon: 015253972634.

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Datum und Ort	Unterschrift Proband(-in)	Unterschrift Erziehungsberechtigte(r)

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Anamnese-Bogen

Alter: ______________________

Geschlecht: □ männlich □ weiblich

Körpergröße (in cm): __________________ Körpergewicht (in kg): _____________

Beruf: ____

Freizeitsport:

Sportarten: ____

Häufigkeit pro Woche: ____

Leistungssport:

Sportart: ____

Trainingsalter (seit wie vielen Jahren trainieren Sie in dieser Sportart): _______________

Trainingseinheiten pro Woche (in Stunden) : ______________________

Standbein: □ links □ rechts Spielbein: □ links □ rechts

Vor wie vielen Tagen haben Sie das letzte Training absolviert? _______________

Aktuelle Beschwerden:

Haben Sie derzeit Beschwerden?

□ keine

□ ja Welche? ____

Frühere Verletzungen:

z.B. Bandverletzung Knie/Sprunggelenk, Meniskusriss, Muskelfaser-/ Muskelbündelriss, Sehnenverletzung, Knochenbrüche, Gehirnerschütterung

Falls zutreffend, bitte mit angeben, wie lange die Verletzung her ist.

__________________________________ □ verheilt □ bleibende Beschwerden

__________________________________ □ verheilt □ bleibende Beschwerden

__________________________________ □ verheilt □ bleibende Beschwerden

__________________________________ □ verheilt □ bleibende Beschwerden

___________________________________ □ verheilt □ bleibende Beschwerden

Operationen:

□ nein

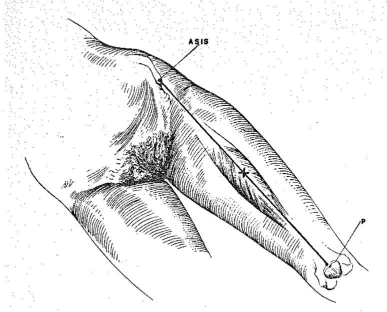
□ ja Welche? ____

!Bitte nicht ausfüllen!

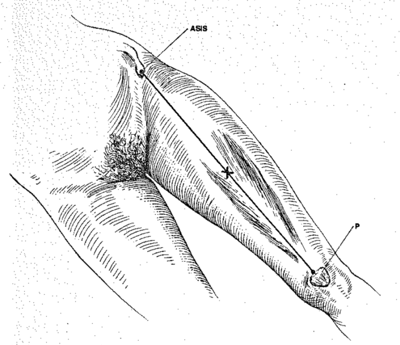
Probandencode	Probandennummer	Einschlussdatum

Measuring points for ultrasound elastography

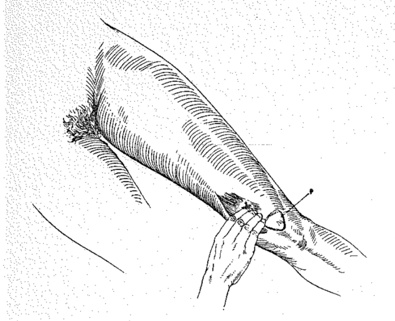
Supine position:



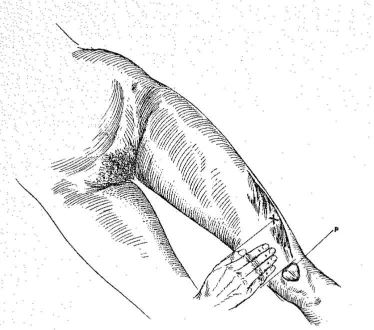
RECTUS FEMORIS



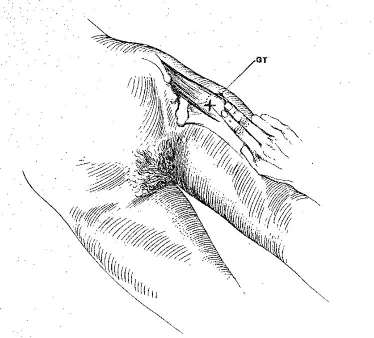
VASTUS INTERMEDIUS



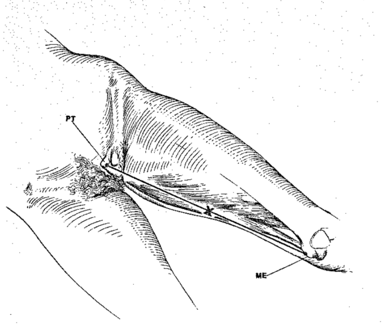
VASTUS MEDIALIS



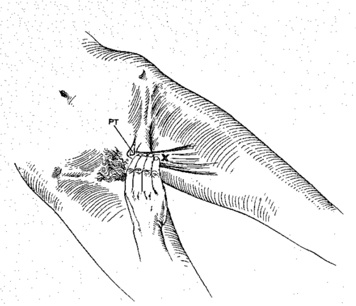
VASTUS LATERALIS



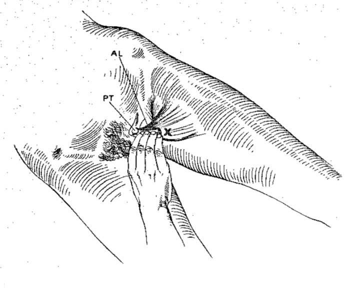
TENSOR FASCIE LATAE



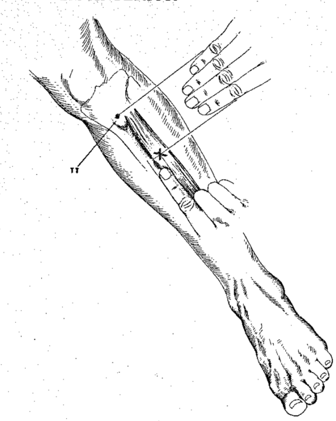
ADDUCTOR MAGNUS



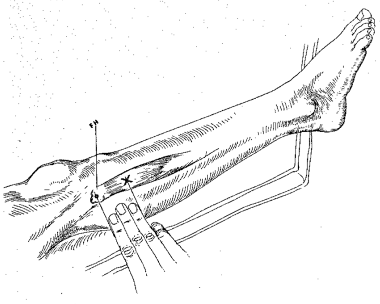
ADDUCTOR LONGUS



ADDUCTOR BREVIS



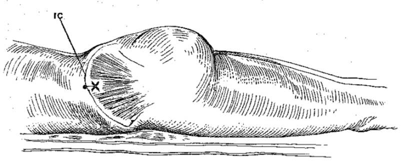
TIBIALIS ANTERIOR



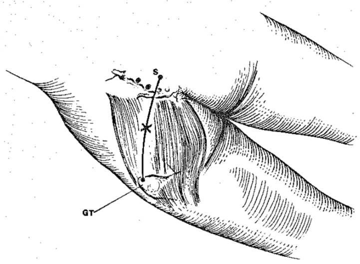
PERONEUS LONGUS

Prone position:

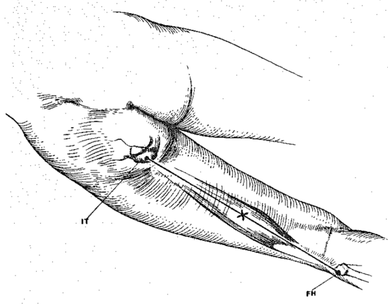
GLUTEUS MEDIUS



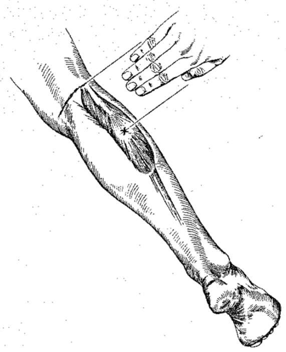
GLUTEUS MAXIMUS



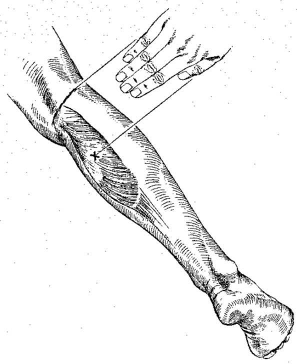
BICEPS FEMORIS: Long Head



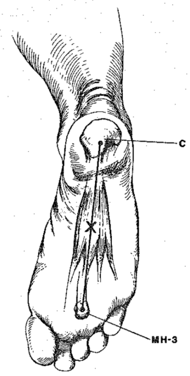
GASTROCNEMIUS: Lateral Head



GASTROCNEMIUS: Medial Head



FLEXOR DIGITORUM BREVIS



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ROM Measurements


1.1 Lumbar rotation

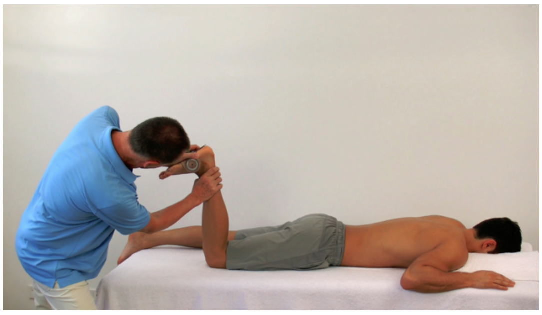

1.2 Lumbar lateral flexion


1.3 Hip internal rotation


1.4 Hip external rotation


1.5 Straight leg raise


1.6 Active ankle dorsiflexion


1.7 Passive ankle dorsiflexion


1.8 Heel-buttock distance


1.9 Finger-floor distance

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