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Assessment of Passive Stiffness of Medial and Lateral Heads of Gastrocnemius Muscle, Achilles Tendon, and Plantar Fascia at Different Ankle and Knee Positions Using the MyotonPRO

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Data Interpretation D
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Background: The aim of this study was to assess the passive stiffness of the medial and lateral gastrocnemius (MG and LG), Achilles tendon (AT), and plantar fascia (PF) at different ankle and knee positions.





Material/Methods: Stiffness was assessed using a portable hand-held device (MyotonPRO). In 30 healthy participants (15 males, 15 females) with the knee fully extended or flexed 90°, stiffness of the MG, LG, AT, and PF was measured at 50° plantar flexion, 0° (neutral position), and 25° dorsiflexion (not for AT) of the ankle joint by passive joint rotation.

Results: With the knee fully extended, passive dorsiflexion caused significant increase in muscle stiffness ($P < 0.001$), whereas AT and PF stiffness increased with passive ankle dorsiflexion regardless of knee position ($P < 0.001$). Increased stiffness was observed in MG compared to LG ($P < 0.001$) and at the 3-cm site of AT compared to the 6-cm site ($P < 0.05$). Stiffness was greater in LG compared to MG at -50° plantar flexion ($P < 0.001$) and was greater in MG compared to LG at 25° dorsiflexion ($P < 0.05$). Stiffness of AT increased in a distal-to-proximal pattern: 0 cm > 3 cm > 6 cm ($P < 0.001$).

Conclusions: Stiffness assessed by use of the MyotonPRO was similar assessments using other techniques, suggesting that the MyotonPRO is capable of detecting the variations in stiffness of MG, LG, AT, and PF at different ankle and knee positions.

MeSH Keywords: **Achilles Tendon • Elasticity • Fascia • Muscle, Skeletal**

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Background

In the field of biomechanics, stiffness is the property of soft tissue that characterizes the resistance to an external force or to a contraction that deforms its initial shape [1]. During contraction or stretching, the stiffness of soft tissue increases [2]. Soft tissue is dynamic; therefore, accurate assessment of soft tissue stiffness throughout its functional range is crucial for establishing the physiological level of stiffness and improving physiological functions [3].

In recent years, the stiffness of soft tissue has been quantitatively assessed using various techniques. Magnetic resonance elastography (MRE) has been used to assess the stiffness of the medial head of the gastrocnemius (MG) by Green et al. to investigate the anisotropic properties of muscle stiffness, but only MG and a relaxed condition have been investigated [4]. Shear wave elastography (SWE) has been used in the triceps surae and Achilles tendon (AT) during the entire ankle functional range [6], and at different knee positions [5]. However, the widespread use of MRE and SWE is limited by the expensive equipment and the lack of portability [6].

The MyotonPRO (Myoton AS, Estonia) is a relatively inexpensive, portable, hand-held device that can be used to assess different soft tissue stiffness. The basic principal of the MyotonPRO device is that exerts a quickly released, single mechanical impulse (time 15 ms, force 0.4 N) under constant precompression force (0.18 N) of the subcutaneous tissue layer above the muscle or tendon being assessed [1]. Compared with the single-axis accelerometer in the Myoton-2 and Myoton-3 devices, the MyotonPRO device has a tri-axial accelerometer, making it more versatile in terms of application [7]. Investigators in various fields have used the device, including in sport medicine [8,9], rehabilitation [10], and spaceflight [11]. Previous studies have shown that the MyotonPRO can reliably assess stiffness of MG, the lateral head of the gastrocnemius (LG), AT, and plantar fascia (PF) [12–14]. A study verified that the correlation of MyotonPRO and SWE in infraspinatus, erector spinae, and gastrocnemius was significant; both “elastic modulus” and “stiffness” values can reflect the relative stiffness properties of the soft tissue, although they are not really equivalent to the true modulus of elasticity obtained by biomechanical testing *in vitro* [15]. Compared with SWE, the acquisition costs for MyotonPRO are low and the examiners do not require special training. Therefore, it is meaningful to establish a baseline of normal stiffness values at various ankle and knee positions using the MyotonPRO. However, to the best of our knowledge, stiffness of MG, LG, AT, and PF at different ankle and knee positions has not been assessed using the MyotonPRO, so the ability of the MyotonPRO to detect the variations in stiffness of soft tissue at different ankle and knee positions remains unknown, and often only 1 site of AT has been investigated.

The purpose of the present study was to assess the variations in stiffness of MG, LG, AT, and PF at different ankle joint angles with the knee fully extended or flexed 90° by using the MyotonPRO. We hypothesized that the MyotonPRO is sensitive enough to detect the same changes in stiffness that have been measured with SWE.

Material and Methods

Participants

Thirty healthy subjects (15 males, 15 females) were recruited from Guangzhou University of Chinese Medicine. Individuals were excluded if they had: 1) any lower-limb injuries or plantar heel pain before the experiment; 2) a history of lower-limb surgery; 3) any gait anomalies; 4) had been taking any medication that affects muscle tone; 5) any other nerve, muscle, or orthopedic disease; or 6) performed strenuous exercise within 48 h of testing. The present study was approved by the Guangzhou University of Chinese Medicine. The experimental procedures were explained to the participants, and they all signed an informed consent. The present study was approved by the Human Subjects Ethics Committee of the Clinical Medical College of Acupuncture, Moxibustion, and Rehabilitation, Guangzhou University of Chinese Medicine. Before testing, all participants were completely informed about the experimental procedures and they all provided signed informed consent.

Experimental procedure

Laboratory experiments were done at the Guangzhou University of Chinese Medicine. Before testing, height and weight were measured and body mass index (BMI) was calculated. The dominant limb was regarded as the one subjects used to kick a ball [16,17]. Passive joint rotation was performed by an experienced physical therapist. Angular displacement was measured with a goniometer.

MyotonPRO examination

A MyotonPRO device (Myoton AS, Estonia) was used to assess stiffness. The measurement site of MG was located at 70% of the lower leg length measured from the lateral malleolus to the popliteal fossa, where cross-sectional areas of the gastrocnemius are almost maximum [18,19]. LG was measured at 1/3 of the leg length from the fibular head to the heel [20]. The AT was assessed at 3 successive levels: 0, 3, and 6 cm from bone insertion (calcaneus) [21]. The PF was measured between the first and second metatarsal bone on the anterior edge of the inferior calcaneal border in the posterior section of the PF [22,23].

Measurement of muscle, tendon, and fascia stiffness

The details of the experimental process were: 1) before testing, subjects were asked to relax for 5 min to ensure muscles and tendons were relaxed; 2) measurement sites were marked with a skin marker; 3) during the tests, subjects lay prone and fully relaxed on the examination table; and 4) stiffness was assessed with the MyotonPRO positioned on the skin markers of the MG, LG, and PF at -50° , 0° , and 25° (0° representing neutral position, -50° representing 50° plantar flexion, and 25° representing 25° dorsiflexion) of ankle joint rotation with the knee fully extended, and then flexed to 90° . Due to the saturation limit of stiffness for the MyotonPRO, the AT was only measured at -50° and 0° . In order to ensure restoration of original musculotendon properties, stiffness at the 3 ankle angles were measured at 1-min intervals [24,25]. All assessments were made 3 times, and mean stiffness values were used for data analysis. Stiffness (N/m) was calculated by the MyotonPRO system based on the equation: $S = \alpha_{max} \cdot m \cdot probe / \Delta l$ (m =the mass of the testing end, α_{max} =maximum deformation acceleration of the tissue, Δl =maximum displacement of the tissue) [26].

Statistical analysis

SPSS software (SPSS version 22.0, IBM, USA) was used for statistical analysis. The level of significance was set at $P < 0.05$. For the stiffness data on muscles, separate 3-way analyses of variance (ANOVAs) (knee angle \times ankle angle \times muscle/tendon) with repeated measures were performed. If there was an interaction among knee angle \times ankle angle \times muscle/tendon, the simple effects were investigated and Bonferroni correction was performed. Since muscle/knee angle/ankle angle of tendons have only 2 levels, a paired t test was used to analyze the difference between muscles, between knee angles of muscles/tendons, and between ankle angles of tendons. A 2-way ANOVAs with repeated measures was performed for the stiffness data of PF. Stiffness data are reported as means \pm SD. Figures were created using GraphPad Prism 5; however, the SD and error bars reported are not representative of the statistics due to the paired nature of the analyses. Therefore, the effect size was also calculated using Cohen's d [27] considering 0.2, 0.5, and 0.8 as small, medium, and large effects, respectively.

Results

Participant demographic data

The average age of subjects was 22.9 ± 3.8 years. Average height was 165.4 ± 9.2 cm and body mass was 60 ± 10 kg.

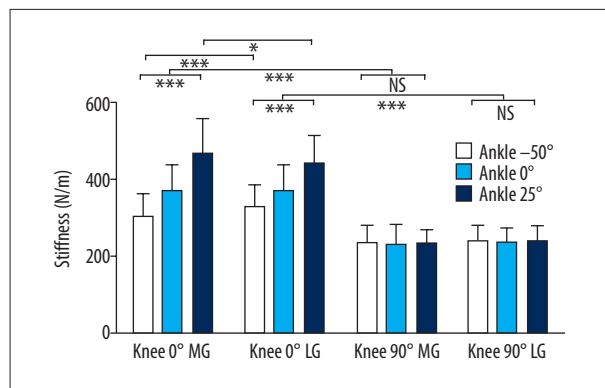


Figure 1. Variations in stiffness of the MG and LG at -50° (white), 0° (blue), and 25° (deep blue) of ankle joint with the knee fully extended or flexed 90° . * $P < 0.05$; *** $P < 0.001$; NS – non-significant at $P < 0.05$.

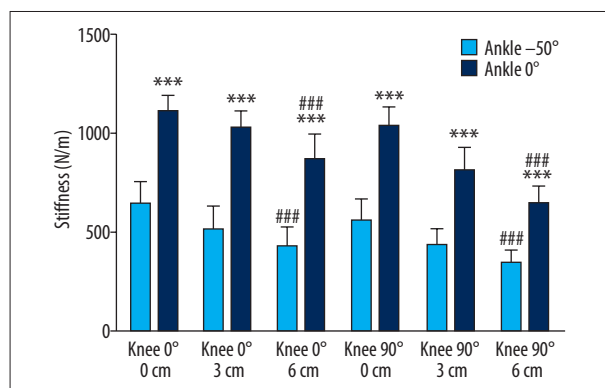


Figure 2. Changes in stiffness of the 3 sites of AT at -50° (blue) and 0° (deep blue) of ankle joint with different knee positions. *** Significant difference between -50° and 0° of ankle joint ($P < 0.001$). ### Significant difference among the 3 levels ($P < 0.001$).

Effect of ankle/knee rotation on stiffness

The relationship between muscle stiffness and passive ankle/knee rotation angle is shown in Figure 1. A 3-way repeated-measures ANOVA illustrated a significant 3-way interaction (knee angle \times ankle angle \times muscle, $P = 0.001$). Further, Bonferroni post hoc analysis revealed that the stiffness of the MG and LG with the knee fully extended was significantly greater than those performed with the knee flexed 90° ($P < 0.001$, $1.36 \leq d \leq 3.45$). The stiffness of the MG and LG increased as the ankle was passively dorsiflexed only when the knee was fully extended ($P < 0.001$). From -50° to 25° of ankle joint rotation, the increase in stiffness was larger in the MG compared to the LG (MG = 163 ± 81 N/m and LG = 111 ± 62 N/m, $P < 0.001$, $d = 0.72$).

The relationship between AT stiffness and passive ankle/knee rotation angle is depicted in Figure 2. A 3-way ANOVA showed a significant knee angle \times ankle angle \times tendon level

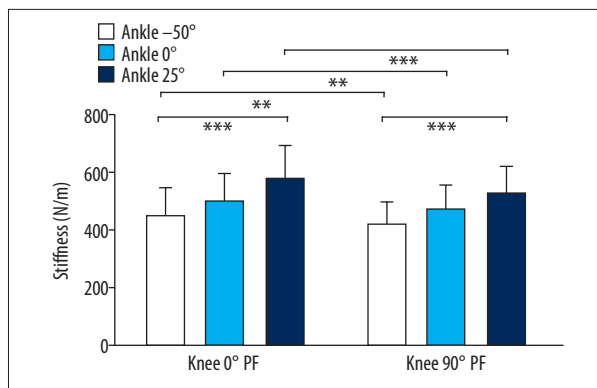


Figure 3. Variations in stiffness of the PF at -50° (white), 0° (blue), and 25° (deep blue) of ankle joint angle with the knee fully extended or flexed 90° . ** $P < 0.01$; *** $P < 0.001$.

interaction ($P < 0.001$). Regardless of knee angle, the stiffness at the 3 AT sites were significantly larger at 0° (when the ankle was in a more dorsiflexed position) ($P < 0.001$). Further analysis indicated that stiffness at the 3 sites was larger with the knee fully extended than flexed ($P < 0.001$, $0.76 \leq d \leq 2.24$). From -50° to 0° of ankle joint rotation, the significant increase in stiffness was only observed at the 3-cm site of AT compared to the 6-cm site when the knee was fully extended (3-cm = 523 ± 121 N/m; 6-cm = 442 ± 119 N/m, $P < 0.05$, $d = 0.67$).

Figure 3 shows the effect of joint angle on PF stiffness. There was a significant effect of the 3 ankle joint angles on PF stiffness; stiffness increased with increasing ankle dorsiflexion at both knee angles ($P < 0.001$) and the knee angle \times ankle angle interaction was not significant ($P > 0.05$). Additionally, PF stiffness with the knee fully extended was significantly greater than those with the knee flexed 90° ($P < 0.01$)– 0.001 , $0.51 \leq d \leq 0.54$).

Stiffness differences among muscles/tendons

Further analysis showed a significant difference between MG and LG with the knee fully extended; stiffness was greater in LG than MG at -50° ($P < 0.001$, $d = 0.51$) and was greater in MG than LG at 25° ($P < 0.05$, $d = 0.51$), which were not observed at the 90° flexion of knee joint (Figure 1). As for the 3 levels of AT, regardless of knee and ankle rotation, stiffness increased in a distal-to-proximal pattern of 0 cm $>$ 3 cm $>$ 6 cm ($P < 0.001$, $0.81 \leq d \leq 2.20$) (Figure 2).

Discussion

The present study assessed variations in stiffness of muscles, fascia, and tendons at different ankle and knee joint angles induced by passive ankle rotation. The major findings were: 1) passive dorsiflexion induced a larger increase in MG and

LG stiffness only when the knee was fully extended, whereas AT and PF stiffness increased irrespective of knee position; 2) AT stiffness at the 3 sites and PF stiffness was less with the knee was flexed than extended; 3) the effects of passive dorsiflexion on stiffness was largest in the MG and the AT at the 3-cm site; 4) stiffness was greater in LG than MG at -50° plantar flexion and was greater in MG than LG at 25° dorsiflexion when the knee was fully extended; 5) regardless of ankle and knee position, AT stiffness increased in a distal-to-proximal pattern of 0 cm $>$ 3 cm $>$ 6 cm. The present findings suggest that MG and LG can be effectively stretched by passive joint rotation only when the knee is fully extended, whereas passive joint rotation influences stiffness of the AT and PF similarly, regardless of the knee position.

Effects of ankle and knee rotation on muscle stiffness

Passive joint rotation has a considerable effect on stiffness in muscles when sufficient mechanical stress is imposed [28,29]. The present study shows that the stiffness of MG and LG was smaller with the knee flexed rather than extended (Figure 1). In addition, with the knee flexed, the MG and LG exhibited little increase in stiffness when the ankle was dorsiflexed, indicating the muscles were under little stretch (tension) in this position. Other authors reported that the MG and LG are stiffer than other plantar flexor muscles (i.e., soleus, flexor digitorum longus, flexor hallucis longus, tibialis posterior, and the lateral peroneus longus) with the knee fully extended, whereas the soleus is the stiffest with the knee flexed 90° [5].

We found that the increase in stiffness with ankle dorsiflexion (-50 to 25°) was greater in the MG than LG. This result indicates that passive joint rotation with the knee fully extended has the most effect on MG, which is consistent with the larger slack angle (the angle beyond which soft tissue begin to develop passive stiffness) of the MG (i.e., the MG becomes slack at greater degree of plantar flexion compared to the LG) [29]. Therefore, during the ankle plantar flexion, the MG is the last of the triceps surae muscles to become slack. The evidence for the apparently greater stiffness in the MG is equivocal and may be related to anatomical factors, including insertional orientation of the muscles into the AT. Anatomical data reported by Edama et al. show that unlike LG, which has an oblique attachment to the AT, MG has a more linear attachment [30]. Given the more linear attachment, most of the tension during a passive joint rotation is more likely to act on the MG. Thus, it is conceivable that the greater sensitivity of the MG to ankle rotation is related to differences in anatomy that affect slack length and inherent tissue properties.

Effects of ankle and knee rotation on AT and PF stiffness

In contrast to MG and LG, AT stiffness increased at all 3 sites with ankle dorsiflexion, irrespective of the knee position (Figure 2).

This finding is consistent with others who found that the shear modulus of the AT, measured by shear wave elastography, increased with the ankle dorsiflexion when the knee was fully extended [16,31]. From the finding of the present study, with the knee flexed 90°, MG and LG exhibited little increase in stiffness regardless of the ankle position (Figure 1). This finding might be explained by the mechanical effects of the soleus. The shear modulus in the soleus significantly increased with ankle dorsiflexion when the knee was flexed 90°, as shown by shear wave elastography [5]. Hence, the force created by passive joint rotation in the flexed knee position may reflect greater stiffness in the soleus-AT unit. Thus, during the dorsiflexion process, AT may be stretched even though the knee was flexed 90°.

PF stiffness was also increased by ankle dorsiflexion regardless of the knee position (Figure 3). Erdemir et al. demonstrated that with increased AT force, the tension in the plantar fascia also increased [32]. In the present study, stiffness in AT was also increased by ankle dorsiflexion (Figure 2). Therefore, stiffness in AT and PF were increased similarly by ankle dorsiflexion. The PF stiffness in the present study is similar to the mean in one report (446.4 ± 46.17 Nm) [33], but less than another (518.89 ± 72.33 N/m) [20], possibly due to the different measurement site or ankle angle.

The present study has shown that the stiffness of the 3 levels of AT and PF performed with the knee flexed was significantly smaller than when the knee was fully extended (Figures 2, 3, respectively). Le Sant et al. suggested that the passive torque with the knee flexed 90° was smaller, indicating a smaller total stiffness in the lower leg muscles [10]. The present study indicates that MG and LG are under very little tension with the knee flexed 90° (Figure 1). Hence, it is reasonable that AT stiffness at the 3 sites and PF stiffness with the knee flexed were smaller than stiffness with the knee fully extended.

The stiffness increase at the 3-cm site was higher than at the 6-cm site when the knee was fully extended. The AT acts as a mechanical buffer to absorb energy of motion and prevent the muscle from damage resulting from a suddenly increased load. If the AT is too stiff and hence not being able to absorb sufficient energy, it is more likely to be injured [34]. Additionally, too much stiffness in the tendon can reduce the time of torque transmission and thus increase the risk of injury [35]. Since dorsiflexion and contraction increase the tendon stiffness, in this case, the AT is prone to injury and rupture. Thus, the 3-cm site may be prone to injury and rupture. From the perspective of vascular distribution, the 3-cm site is one of the least perfused areas [36]. Therefore, this lack of vascular perfusion and the greater sensitivity of the 3-cm site to ankle rotation may put the AT at high risk of injury and poorly able to heal.

Inter-muscle differences in passive stiffness

In the present study, LG stiffness was greater compared with MG at the -50° ankle angle only with the knee fully extended. However, MG stiffness was greater compared with LG at 25° angle of the ankle joint (Figure 1). Muscle fiber type is one possible factor that may affect stiffness, given that stiffness of type I muscle fibers is greater than that of type II muscle fibers [37,38]. However, this factor cannot explain the greater MG stiffness, as fiber type percentage is similar in the 2 heads of the gastrocnemius [39]. Another view is that the perimysium and endomysium could affect muscle stiffness [40]. A previous report found that the shear modulus in LG was greater compared to MG at -30° of plantar flexion, using shear wave elastography (MG=27.6±7.3kPa; LG=33.5±6.3kPa) [41]. As mentioned above, the slack angle of MG is more plantar-flexed than the LG when the knee is fully extended. Additionally, reduction of stiffness in MG might be greater during plantar flexion because of a more linear attachment of the MG to the AT [30]. In addition, no significant stiffness difference was found between MG and LG when the ankle was flexed at 0° (Figure 1). Considering these factors, it is reasonable that the stiffness in MG was smaller than in LG at a -50° ankle joint angle and MG was higher than LG at a 25° ankle joint angle. Using shear wave elastography, Hirata et al. had suggested that the shear modulus in MG was also greater than LG at 25° ankle joint angle [6]. Therefore, this may help explain the greater MG stiffness value at a 25° ankle joint angle.

Intra-tendon difference in stiffness

AT stiffness differed between the 0-cm, 3-cm, and 6-cm sites, and increased in a distal (6 cm cm)-to-proximal (0 cm) pattern (Figure 2). This finding was in agreement with previous studies [42,43]. We also found that the AT compliance increased in a distal-to-proximal pattern.

The present study evaluated the changes in muscle and tendon stiffness using the MyotonPRO device at different ankle and knee joint angles. Excessive deformation of the tissues at 25° dorsiflexion may change the stiffness properties to such a degree that they cannot fully recover within 1 min [24]; therefore, to avoid this effect, randomization of the ankle angle order was not performed.

A limitation of the present study is that the passive stiffness in the soleus, which is an important part of the triceps surae, was not evaluated. Another limitation is that stiffness assessment was performed at only 1 point for each target. Therefore, it is not clear whether the present findings hold true for other regions. In addition, the effect of skin on the MyotonPRO measurement when the nearby joint is stretched remains unknown. Finally, the findings from our study apply only to healthy adults who are free of injury or disease.

Conclusions

The present study established stiffness values of the MG, LG, AT, and PF in healthy humans at different ankle and knee positions by using the MyotonPRO. Stiffness assessed by the MyotonPRO shows similar changes as with other techniques, suggesting that the MyotonPRO is capable of detecting the

variations in stiffness of MG, LG, AT and PF at different ankle and knee positions.

Conflict of interest

None.

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