Published in final edited form as:

Foot (Edinb). 2008 June ; 18(2): 61–67. doi:10.1016/j.foot.2007.12.002.

Effect of metatarsal phalangeal joint extension on plantar soft tissue stiffness and thickness

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Abstract

Background—Plantar soft tissue stiffness and thickness are important biomechanical variables to understand stress concentrations that may contribute to tissue injury.

Objective—The purpose of this study was to determine the effects of passive metatarsal phalangeal joint (MPJ) extension on plantar soft tissue stiffness and thickness.

Methods—Seventeen healthy participants (7 male, 10 female, mean age 25.3 y, SD 4.4 y, mean BMI 24.7 kg/m², SD 3.2 kg/m²) were tested. Plantar soft tissue stiffness and thickness were measured at the metatarsal heads, midfoot and heel using a custom-built indenter device and an ultrasound machine.

Results—Indicators of soft tissue stiffness (K_1 values) at the metatarsal heads and midfoot showed increases in stiffness of 81–88% (SD 20–33%) in the MPJ extension position compared with the MPJ neutral position. Soft tissue thickness measures at the metatarsal heads with the MPJ in neutral ranged from a mean of 8.9–13.5 mm and decreased, on average, by 8.8% (SD 2.9%) with MPJ extension.

Conclusions—MPJ extension has a profound effect on increasing forefoot plantar soft tissue stiffness and a consistent but minimal effect on reducing soft tissue thickness. These changes may help transform the foot into a rigid lever at push-off consistent with the theory of the windlass mechanism.

Keywords

Windlass mechanism; Material properties; Plantar tissue; Soft tissue thickness; Soft tissue stiffness; Plantar fascia

1. Introduction

The plantar aponeurosis arises at the plantar surface of the calcaneus and inserts into the plantar surface of the phalangeals. According to the “windlass mechanism” described by Hicks, when the metatarsal phalangeal joints (MPJ) are extended, the plantar aponeurosis is subsequently pulled distally around the metatarsal heads (MTHs), developing tension in the plantar

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aponeurosis [1] and simultaneously raising the medial arch height [2,³]. This passive tension developed within the plantar aponeurosis when the MPJs are extended, purportedly helps to convert the foot into a rigid structure. The plantar aponeurosis also has multiple connections to the surrounding soft tissue. Besides raising the arch height, we speculate that increased tension on the plantar aponeurosis via MPJ extension will also increase soft tissue stiffness (STS) and decrease soft tissue thickness (STT). We believe that the biomechanics involved in the windlass mechanism apply whether the foot is loaded in weight bearing or unloaded in open chain toe extension.

STS and STT are important variables required for computational models (i.e., finite element analysis) to identify stress concentrations that may lead to tissue injury or breakdown [4–7]. These computational models, if valid, can help design optimal treatments to distribute high localized stresses that may contribute to injury or skin breakdown. In general, a single estimated value of STS and STT has been used in these computational models regardless of the phase of walking. STS and STT values typically are taken from the literature [7] or estimated using in vivo testing [4,⁶]. Evidence exists, however, to indicate that STS and STT can vary across individuals and populations; i.e., diabetics compared to non-diabetics [8,⁹] and we propose STS and STT can vary depending on MPJ position.

The purpose of this study, therefore, was to determine the effect of passive MPJ extension on plantar STS and STT. Given the principle of the windlass mechanism, it was hypothesized that the plantar STS would increase and STT would decrease when the MPJs are placed in an end-range extended position as compared with the MPJs in a neutral position.

2. Methods and materials

2.1. Participants

Seventeen healthy participants, 7 males and 10 females, between the ages of 21 and 35 years (mean age 25.3 y, SD 4.4, mean BMI 24.7 kg/m², SD 3.2) and with no known foot impairments were tested. Healthy subjects were studied to establish clearly the relationship in those without impairments before other populations are studied. All subjects were recruited from the general student population in the Program in Physical Therapy, Washington University School of Medicine. The exclusion criteria consisted of any active disease state that might affect soft tissue stiffness and thickness including diabetes, fibromyalgia, scleroderma, scar tissue at the sites of investigation, or history of neuropathy. Subjects gave their informed consent in accordance with the Washington University School of Medicine's policy on research with human subjects.

2.2. Plantar Tissue Testing

The participants were positioned prone on a padded table, with the foot of the subject allowed to extend beyond the table edge. The knee was positioned in full extension and the ankle in a resting plantar flexed position with the toes relaxed. The first, second, third and fifth MTHs were palpated, and the central portion of each of the MTHs was marked with a permanent marker. The MTHs were chosen because they have been recognized as areas of high plantar pressures and thus at risk of ulceration [10]. An indentation site on the central plantar heel and distal midfoot (2 and 4 cm proximal to the second MTH) were also marked, as shown in Fig. 1. The heel was selected to determine if MPJ extension affected this area, which is located away from the primary MPJ movement. The distal midfoot was selected to represent the area where a metatarsal pad might be placed to relieve plantar pressure on the MTHs [11]. The dorsal surface of the foot was stabilized using a Versa Form foam-stabilizing pillow (Tumble Forms, Jackson, MI, USA). The foam-stabilizing pillow minimized the skeletal motion of the metatarsal bones during the indentation procedure. STS and STT measures were obtained with

the subjects' MPJs in a relaxed neutral position, then with the MPJs passively extended to their end-range. MPJ extension simulates the position of the MPJ during the late stance phase of walking. One of the investigators held the toes in a maximally extended but comfortable position, while another investigator obtained the measurements. Subjects were instructed to relax their foot and leg as much as possible so MPJ motion would be passive. The foot chosen for testing was alternated between subjects, and the same investigator performed all measurements. STS measurements preceded STT measurements, with a five minute delay between measures to clean and re-calibrate equipment. Participants were tested over a one-hour period.

2.3. Soft Tissue Stiffness Assessment

Plantar STS was defined as the relationship between the applied force and resultant displacement (F/D) as an indenter device (Fig. 1) was pushed perpendicularly into the soft tissue of the plantar foot. This stiffness curve is nonlinear; i.e. stiffness is relatively low with initial indentation and stiffness values increase with increased force as illustrated in Fig 2. A parallel three-element viscoelastic mechanical model [8] was used to characterize the stiffness curve. Variables K_1 and K_2 were indicators of the slope of the initial toe region and the final steep portion of this stiffness curve, respectively (Fig. 2) [8,12].

Plantar STS was measured using a custom-built indenter device (Fig. 1) [12] capable of determining force and displacement during soft tissue compression. The device, and accuracy and reliability testing using the device, have been described previously in detail [12]. Briefly, the device consisted of a load cell (MLP-10, Transducer Techniques, Temecula, CA, USA) mounted on a 3-dimensional Metrecom measurement device (Metrecom 2000, Faro Technologies, Lake Mary, FL, USA). A cylindrical stylus with a diameter of 7.9 mm and length of 7 cm was machined and fitted in the load cell. The load cell had a range of 0 to 10 lb (0 to 44.48 N). The analog output from the load cell was amplified and conditioned with a standalone amplifier (TMO-2, Transducer Techniques, Temecula, CA, USA), which was converted to a digital signal with an analog-to-digital converter card installed in a computer. The signal was sampled at 40 Hz and analyzed by software developed in our laboratory. A picture of the Metrecom with the load cell, stylus, and amplifier is shown in Fig. 1.

The indenter device was calibrated using methods previously described [12]. The device was then used to measure the F/D characteristics of the plantar surfaces over the first, second, third and fifth MTHs, 2 and 4 cm proximal to the second MTH, and the heel (seven locations total, Fig. 1) using the protocols approved by the institutional review board. Three to four preconditioning indentations were applied to each site before the measurements were made to minimize initial viscoelastic changes to indentation. Each test consisted of three perpendicular indentations that met the following criteria at each site: the force range was constrained to 2 to 9 N, the force/time slope range was limited to 20 to 30 N/s, and the velocity R^2 was required to be greater than .95. These criteria were chosen to attempt to enhance reproducibility of measures and represent forces observed during gait as described previously [12]. Our previous work indicated good reliability of stiffness values (intra- and inter-rater reliability as measured by ICC=.99)[12]. Indenter values also were accurate as indicated by close agreement with F/D measurements taken with an Instron device (poron values were 5.34 and 5.52 N/mm, plastizote values were .98 and 1.04 N/mm, for indenter and Instron devices, respectively) [12].

2.4. Soft Tissue Thickness Assessment

A two-dimensional clinical ultrasound machine (Acuson 128XP10, Mountain View, California, USA) was used to measure plantar STT. Measurements were made with a 10.0-MHz linear array transducer (aperture size 26 mm) of a computed sonography system (Acuson

128XP10, Mountain View, California, USA). Using an ultrasound wand, approximately 8.9 cm by 1.3 cm, the aforementioned test locations were identified on the ultrasound monitor, with the exception of the midfoot sites (2 and 4 cm proximal to the second MTH). Accurate STT measures at the midfoot were hampered by the oblique angle of the metatarsal shaft. The distance from the external surface of the skin to the external surface of bone defined STT. STT at the first MTH was obtained by measuring the distance from the skin's surface to the bony surface of the metatarsal between the medial and lateral sesamoid bones. The ultrasound wand was pushed perpendicular to the plantar surface of the foot over each marked site. To ensure clear identification of the sites on the monitor, slight force was applied through the wand to the skin of the foot. Once the site was identified, the investigator reduced the force through the wand to the point where the wand was just resting on the skin but with enough pressure for the site to still be visible on the monitor. A cursor on the ultrasound monitor was then used to mark the most superficial aspect of the bony prominence and measure STT from that point.

2.5. Statistical Analysis

A paired t-test was used to compare the STS and STT between the toes neutral and toes extended positions at each test location. All statistical analyses were performed using the SYSTAT computer program package, version 10.0 (SYSTAT, Inc., Evanston, IL, USA), with the level of significance set at $p \leq 0.05$.

3. Results

Summary statistics (mean and standard deviation) are provided to describe plantar STS and STT values for all subjects (Table 1).

3.1. Soft Tissue Stiffness

Plantar STS values (K_1 and K_2) at the forefoot and midfoot were significantly greater with MPJ extension compared to when the MPJs were in a neutral position. When the MPJs were extended, the mean K_1 and K_2 values at the MTHs and midfoot showed an increase in STS by 81.4% (SD 20.1%) and 87.5% (SD 33.2%), respectively. The mean K_1 and K_2 values were significantly different ($p \leq 0.005$) between the MPJ neutral and MPJ extended trials at all test sites, except at the heel (Fig. 3). Although STS increased at all test sites, the largest increase in STS (both in K_1 and K_2) was observed at the first MTH (Fig. 3).

3.2. Soft Tissue Thickness

Plantar STT decreased at the MTHs ($p \leq 0.002$) but not at the heel when the MPJs were extended as compared to when the MPJs were in a neutral position (Fig. 4). STT measures at the MTHs with the MPJ in neutral ranged from a mean of 8.9–13.5 mm and decreased, on average, by 8.8% (SD 2.9%) with MPJ extension (Table 1).

4. Discussion

These results demonstrate plantar soft tissue under the MTHs and midfoot to be stiffer (80%) and thinner (9%) with MPJ extension compared with MPJ neutral position. MPJ extension did not have an effect on STT or STS at the heel.

4.1. Soft Tissue Stiffness

The finding that plantar STS was significantly greater with MPJ extension as compared with MPJ neutral position is consistent with the proposed windlass mechanism theory described by Hicks [2]. After observing that passive extension of the first MPJ resulted in tightening of the plantar aponeurosis, Hicks believed the arch-rising effect, i.e. the windlass mechanism, served

to provide a stiffer, more rigid foot that does not yield to the increasing forces experienced during the push-off phase of walking. We believe the increase in stiffness of the plantar soft tissue with MPJ extension may be attributed to the increased tension developed within the plantar aponeurosis and plantar soft tissue beneath the MTHs. Tension in the plantar fascia was not measured in this study. We speculate, however, that the increased tension in the plantar fascia and surrounding plantar soft tissues resulted in a generalized increase in stiffness that we measured using these methods. As noted previously, these STS values are important for developing computational models of the foot.

The largest increase in STS (~100%, Table 1) occurred at the first MTH with MPJ extension compared with MPJ neutral. Likewise, Zheng et al. found the first MTH to have the greatest increase (66%) in stiffness with varying ankle positions in a small sample ($n = 4$) of young healthy subjects (mean age 22 y) as compared to other locations on the foot [9]. This posture-dependence of stiffness was not demonstrated in the elderly patients with diabetes mellitus. The study by Zheng et al. investigated STS using active ankle dorsiflexion and plantarflexion, whereas this present study used passive MPJ extension. Zheng et al. speculated that the tissues became stiffer when the underlying muscles were activated [9]. In both studies, younger populations were found to have the largest change in STS at MTH 1, as predicted by Hicks who described the greatest windlass effect seen at the first ray compared to other rays (i.e., plantar flexion of the first metatarsal with extension of the first MPJ) [2].

MPJ extension did not have an effect on STS at the heel. The location of testing at the heel was proximal to the origin of the plantar aponeurosis and therefore, the increased stiffness of the aponeurosis from MPJ extension did not affect plantar soft tissue material properties at the heel.

The device and methods described in this paper have been used in a previous study to measure plantar STS [8]. K_1 values in a neutral MPJ position for a group of 20 subjects with a mean age 60 years (SD 11 y) were similar but slightly lower (0.69–0.81 N/mm) in the Klaesner et al. study compared to the K_1 values in the same position for this study (0.99–1.08 N/mm) using younger subjects [8]. K_2 values were similar but slightly higher (1.37–2.28 N/mm) in the Klaesner et al. study compared to the K_2 values in this study (0.96–1.45 N/mm). Additional research is needed to understand better the physiological correlates of K_1 and K_2 spring constants illustrated in Fig. 2, how age, gender, activity level and disease processes affect these material properties of plantar tissue, and how these properties relate to soft tissue injury or breakdown.

4.2. Soft Tissue Thickness

Plantar STT under the metatarsal heads and midfoot decreased 8.8% (SD 2.9%) with MPJ extension compared with MPJ neutral. The decrease in thickness of the plantar soft tissue when the toes are extended may be attributed to the tension and subsequent distortion of tissues across the plantar skin, aponeurosis and subcutaneous tissue. Interestingly, Gefen used radiographic fluoroscopy to visualize the longitudinal distortion on the plantar fascia during the gait cycle and reported ranges of distortion to be 9–12% [1], consistent with the finding in our paper.

The STT values during MPJ neutral position reported in this study are similar to previous results presented by Gooding et al. [13] and Zheng et al. [9] and slightly greater than that reported by Mueller et al. [14]. Using high resolution ultrasound, Gooding et al. measured the thickness of the heel and MTHs 1–5 in control subjects to be 18.6 mm, 12.9 mm, 14.2 mm, 13.6 mm, 12.9 mm and 11.5 mm, respectively [13]. Zheng also found similar STT results at the heel (16 mm), MTH 1 (12 mm) and MTH 2 (14 mm) in young control subjects (mean age 22 y) [9]. The reported values by Gooding and Zheng were in a similar range to the results of the plantar tissues with toes neutral in our study (heel: 13.6 mm, MTH 1: 13.4 mm, MTH 2:

12.6 mm, MTH 5: 8.9 mm). Mueller et al. reported the thickness value of skin and subcutaneous tissue at similar sites in older control subjects (mean age 55 y, SD 9) as 14.5 mm (MTH 1) and 7.2 mm (MTH 5) [14]. In general, the magnitude of STT was higher in our study than those presented by Mueller in which computed tomography was used to measure plantar STT with the foot in a loaded position, which would cause some tissue compression [14]. MPJ extension did not have an effect on STT, again we are attributing the lack of change in STT to the proximal location of testing at the heel and the more distal origin of the plantar aponeurosis.

The increased tension on the plantar soft tissues resulted in a relatively small decrease in thickness with a substantial generalized increase in midfoot and forefoot plantar soft tissue stiffness. Clearly, the relationship between deformation and force is non-linear and would appear to be beneficial because it allows a small change in tissue thickness to result in a substantial increase in plantar soft tissue stiffness helping the foot to become more rigid in preparation for the push-off phase of walking.

4.3. Limitations of the study and directions for the future

The approach used in this study has limitations to consider in regard to testing. The study included a single group of young healthy individuals. Plantar STS and STT measures as they relate to the windlass mechanism should be investigated using groups of subjects with diabetes mellitus and peripheral neuropathy to better understand how these measurable parameters contribute to skin breakdown and plantar pressures. One research group has documented that the plantar fascia was thickened and MPJ motion was reduced in people with diabetes and peripheral neuropathy compared to control subjects which contributed to altered onset of the windlass mechanism and increased loading on the forefoot [15,16]. In particular, loss of motion at the MPJs would seem to compromise the ability of the foot to vary its soft tissue material properties during the late stance phase of walking and could contribute to injury.

Any measurement approach has potential for error. In this study all data were collected using the same two devices by the same investigator to minimize variability and measurement error. Variation in subject positioning from the MPJ neutral to MPJ extended position may have an effect on individual subject STS and STT values but the one tester used standardized methods to extend the MPJs of every subject until a firm joint end-feel could be felt or until the patient was no longer able to tolerate the change in position. Furthermore, the same amount and angle of force and velocity was applied to each subject with STS and STT measurements. Those measurements that were outside the range set for force, velocity and angle of STS measurements were automatically rejected, and the tester was required to make another indentation until the satisfactory criteria were met. Prior work has shown our device and methods to be accurate and reliable [12].

This study provides some evidence that the plantar soft tissue becomes considerably stiffer and slightly thinner with MPJ movement from a neutral position to an extended position. We believe the results of the soft tissue material properties investigated in this study can be integrated into developing accurate computational models of the foot. The data collected in this project can help to refine the computational models that are being used to optimize the design of orthotic devices in an effort to prevent skin breakdown in people with diabetes and peripheral neuropathy.

5. Conclusions

The results of this study indicate MPJ extension has a profound effect on increasing plantar STS and a consistent but minimal effect on reducing STT. These changes may help transform the foot into a rigid lever at push-off consistent with the theory of the windlass mechanism. Furthermore, the results of this study provide estimates of STS and STT values, which may be

used in computational models (i.e., finite element analysis) of the foot simulating the push-off phase of walking.

Acknowledgments

This study was funded by the National Center of Medical Rehabilitation and Research, National Institutes of Health (RO1 HD 36895).

This study was funded by the National Center of Medical Rehabilitation and Research, National Institutes of Health (RO1 HD 36895). The authors would like to thank Mr. Chris Hoy for his assistance in data collection.

References

1. Gefen A. The in vivo elastic properties of the plantar fascia during the contact phase of walking. *Foot Ankle Int* 2003;24(3):238–44. [PubMed: 12793487]
2. Hicks JH. The mechanics of the foot. II. The plantar aponeurosis and the arch. *J Anat* 1954;88(1):25–30. [PubMed: 13129168]
3. Kappel-Bargas A, Woolf RD, Cornwall MW, McPoil TG. The windlass mechanism during normal walking and passive first metatarsalphalangeal joint extension. *Clin Biomech* 1998;13(3):190–4.
4. Actis RL, Ventura LB, Smith KE, Commean PK, Lott DJ, Pilgram TK, et al. Numerical simulation of the plantar pressure distribution in the diabetic foot during the push-off stance. *Med Biol Eng Comput* 2006;44(8):653–63. [PubMed: 16937207]
5. Cheung JTM, Zhang M. A 3-dimensional finite element model of the human foot and ankle for insole design. *Arch Phys Med Rehab* 2005;86(2):353–8.
6. Erdemir A, Saucerman JJ, Lemmon D, Loppnow B, Turso B, Ulbrecht JS, et al. Local plantar pressure relief in therapeutic footwear: design guidelines from finite element models. *J Biomech* 2005;38(9):1798–1806. [PubMed: 16023466]
7. Gefen A. Plantar soft tissue loading under the medial metatarsals in the standing diabetic foot. *Med Eng Phys* 2003;25(6):491–9. [PubMed: 12787987]
8. Klaesner JW, Hastings MK, Zou DQ, Lewis C, Mueller MJ. Plantar tissue stiffness in patients with diabetes mellitus and peripheral neuropathy. *Arch Phys Med Rehab* 2002;83(12):1796–1801.
9. Zheng YP, Choi YK, Wong K, Chan S, Mak AF. Biomechanical assessment of plantar foot tissue in diabetic patients using an ultrasound indentation system. *J Ultrasound in Med and Biol* 2000;26(3):451–56.
10. Kelly VE, Mueller MJ, Sinacore DR. Timing of peak plantar pressure during the stance phase of walking. A study of patients with diabetes mellitus and transmetatarsal amputation. *J Am Pod Med Assoc* 2000;90(1):18–23.
11. Mueller MJ, Lott DJ, Hastings MK, Commean PK, Smith KE, Pilgram HK. Efficacy and mechanism of orthotic devices to unload metatarsal heads in people with diabetes and a history of plantar ulcers. *Phys Ther* 2006;86(6):833–42. [PubMed: 16737409]
12. Klaesner JW, Commean PK, Hastings MK, Zou D, Mueller MJ. Accuracy and Reliability Testing of a Portable Soft Tissue Indentor. *IEEE Trans Neural Syst Rehab Eng* 2001;9(2):232–40.
13. Gooding GA, Stess RM, Graf PM, Moss KM, Louie KS, Grunfeld C. Sonography of the sole of the foot. Evidence for loss of foot pad thickness in diabetes and its relationship to ulceration of the foot. *Invest Radiology* 1986;21(1):45–8.
14. Mueller MJ, Hastings MK, Commean PK, Smith KE, Pilgram T, Robertson DD, et al. Forefoot structural predictors of plantar pressures during walking in people with diabetes and peripheral neuropathy. *J Biomech* 2003;36(7):1009–17. [PubMed: 12757810]
15. D'Ambrogio E, Giacomozzi C, Macellari V, Uccioli L. Abnormal foot function in diabetic patients: the altered onset of Windlass mechanism. *Diabet Med* 2005;22(12):1713–19. [PubMed: 16401317]
16. Giacomozzi C, D'Ambrogio E, Uccioli L, Macellari V. Does the thickening of Achilles tendon and plantar fascia contribute to the alteration of diabetic foot loading? *Clin Biomech* 2005;20(5):532–9.

Fig. 1.
(a) Demonstration of the indentation method used by our investigators; (b) The seven test sites: first, second, third and fifth metatarsal heads, 2 and 4 cm proximal to the second metatarsal head, and the heel.

$$Eq\ 1: F = K_1 X \quad \text{if } 0 \leq X \leq X_s$$

$$Eq\ 2: F = K_1 X + K_2 (X - X_s) + CX \quad \text{if } X_s < X \leq X_{MAX}$$

Fig. 2.

A mathematical model was used to characterize the stiffness curve (Klaesner et al., 2002). Variables K_1 and K_2 were indicators of the slope of the initial toe and final steep portions of the stiffness curve.

Fig. 3.

(a) A comparison of plantar soft tissue stiffness values between metatarsal phalangeal joints and toes in a neutral position (TN) and passively extended to end-range (TE): (a) K₁, (b) K₂. Abbreviation: Met, metatarsal. * Statistically different ($p \leq 0.005$, paired t-test).

Fig. 4.

A comparison of plantar soft tissue thickness values between metatarsal phalangeal joints and toes in a neutral position (TN) and passively extended to end-range (TE). Abbreviation: Met, metatarsal. * Statistically different ($p \leq 0.002$, paired t-test).

Table 1
Plantar soft tissue stiffness and thickness means and standard deviations for all subjects, n = 17

Site	Constant	MPJ Neutral		MPJ Extended		P-value
		Mean (SD)		Mean (SD)		
Heel						
	K ₁ (N/mm)	0.99 (0.18)		1.10 (0.23)		0.101
	K ₂ (N/mm)	1.44 (0.26)		1.36 (0.30)		0.161
	STT (mm)	13.61 (1.57)		13.61 (1.64)		0.987
Metatarsal 1						
	K ₁ (N/mm)*	0.98 (0.31)		1.98 (0.67)		0.000
	K ₂ (N/mm)*	1.37 (0.55)		2.64 (0.74)		0.000
	STT (mm)*	13.44 (1.86)		12.52 (1.51)		0.001
Metatarsal 2						
	K ₁ (N/mm)*	0.95 (0.24)		1.65 (0.35)		0.000
	K ₂ (N/mm)*	1.05 (0.24)		2.29 (0.45)		0.000
	STT (mm)*	12.61 (1.60)		11.14 (1.42)		0.000
Metatarsal 2 (2 cm)						
	K ₁ (N/mm)*	0.87 (0.49)		1.69 (0.18)		0.000
	K ₂ (N/mm)*	0.77 (0.22)		1.38 (0.43)		0.000
Metatarsal 2 (4 cm)						
	K ₁ (N/mm)*	0.80 (0.18)		1.62 (0.42)		0.000
	K ₂ (N/mm)*	0.81 (0.18)		1.04 (0.31)		0.005
Metatarsal 3						
	K ₁ (N/mm)*	1.07 (0.34)		1.69 (0.46)		0.000
	K ₂ (N/mm)*	1.04 (0.30)		1.98 (0.56)		0.000
	STT (mm)*	11.60 (1.45)		10.34 (1.36)		0.000
Metatarsal 5						
	K ₁ (N/mm)*	1.04 (0.30)		1.70 (0.42)		0.000
	K ₂ (N/mm)*	0.96 (0.29)		2.08 (0.56)		0.000
	STT (mm)*	8.92 (0.81)		8.41 (0.77)		0.002

Note. K₁ and K₂ values are indicators of soft tissue stiffness (STS).
Abbreviation: SD, standard deviation, STT, soft tissue thickness, MPJ, metatarsal phalangeal joint.

Statistically significant ($p \leq 0.05$, paired t-test).