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Mechanical Linkage between Achilles Tendon and Plantar Fascia Accounts for Range of Motion of Human Ankle–Foot Complex

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ABSTRACT

SHIOTANI, H., K. TAKAHASHI, Y. HONMA, K. TOMARI, H. HAYASHI, N. SADO, and Y. KAWAKAMI. Mechanical Linkage between Achilles Tendon and Plantar Fascia Accounts for Range of Motion of Human Ankle–Foot Complex. *Med. Sci. Sports Exerc.*, Vol. 55, No. 1, pp. 66–73, 2023. **Purpose:** The human ankle–foot complex possesses a passive range of motion (ROM) through changes in tibiocalcaneal (θ_{cal}) and foot arch (θ_{arch}) angles. Based on the anatomical linkage between the Achilles tendon (AT) and plantar fascia (PF), we hypothesized that AT and PF with different mechanical properties conjointly modulate the passive ROM of the human ankle–foot complex. We examined the association of AT and PF stiffness with passive ankle–foot ROM and further addressed differences between sexes. **Methods:** A series of sagittal magnetic resonance images of the foot and passive ankle plantar flexion torque were obtained for 20 men and 20 women with their ankle–foot passively rotated from 30° of plantar flexion to 20° of dorsiflexion. Based on the measured changes in AT and PF lengths, θ_{cal}, θ_{arch}, and passive torque, AT and PF stiffness were determined. **Results:** Upon passive ankle dorsiflexion, AT and PF were lengthened; their length changes were inversely correlated. Men showed a stiffer AT, more compliant PF, less calcaneal rotation, and greater foot arch deformation compared with women. Furthermore, we found inverse correlations between AT stiffness and ROM of θ_{cal}, and between PF stiffness and ROM of θ_{arch} in men and women. **Conclusions:** Passive AT and PF extensibility counter each other. AT and PF stiffness and passive ROM of ankle–foot components were countered between sexes; however, associations between stiffness and passive ROM of the ankle–foot complex were consistent between sexes. Our findings support the notion that the balanced mechanical interaction between the AT and PF can account for the passive ROM of the human ankle–foot complex *in vivo*, and the differences between sexes. **Key Words:** JOINT FLEXIBILITY, MEDIAL LONGITUDINAL ARCH OF THE FOOT, TRICEPS SURAE MUSCLE–TENDON UNIT, SEX DIFFERENCE, STIFFNESS, MAGNETIC RESONANCE IMAGING

Passive joint range of motion (ROM), assessed in various clinical tests, is a major indicator of joint flexibility. It is known that reduced ankle joint dorsiflexion ROM impairs force attenuation during running and landing (1,2), increasing the risk of injury (3,4). Therefore, physical fitness and rehabilitation programs incorporate various modalities to improve ankle joint ROM for enhanced exercise performance and injury prevention (5–8). However, excessive ankle joint ROM (e.g., hypermobility) is a potential injury risk (9). Thus, insufficient and excessive ankle joint ROMs are possible causes of injury and impairment of exercise performance. Hence, clarifying the mechanisms that modulate passive ankle joint ROM remains a critical challenge in athletic and clinical settings.

Ankle joint ROM is generally evaluated based on the angle between the shank and the entire foot (typically represented as the direction of the sole of the foot). However, the human foot has a unique arch-shaped structure formed by the forefoot and rearfoot bones (Fig. 1) that can change their relative positions (foot arch deformation) with passive ankle–foot rotation (10,11). Thus, foot arch deformability contributes to the

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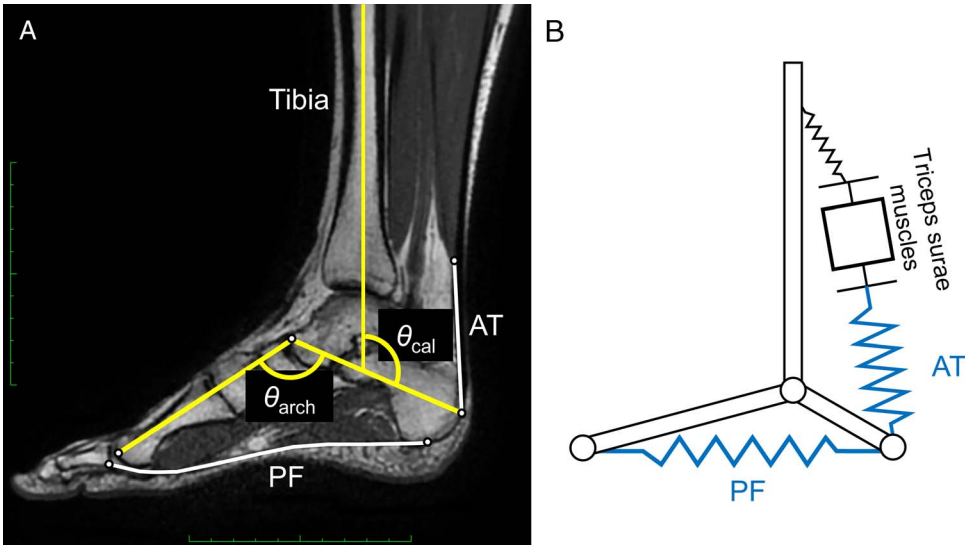

FIGURE 1—Example image of θ_{cal} and θ_{arch} angles, AT, and PF (A). Schematic of AT and PF as series elastic components (B).

ROM of the ankle–foot complex. In other words, the passive ROM of the ankle–foot complex can be broken down into calcaneal rotation and foot arch deformation.

Passive ROM is thought to be modulated by passive tension applied to tissues surrounding the joint. During passive ankle dorsiflexion, the Achilles tendon (AT) contributes to more than half of the elongation of the triceps surae muscle–tendon unit (12–14). Based on its anatomy, the AT may modulate calcaneal rotation, although its mechanical properties, such as stiffness, are considered to determine passive ankle joint dorsiflexion ROM (15,16). The existing literature suggests that the plantar fascia (PF), a dense connective tissue stretching from the calcaneus to the proximal phalanges, provides a primary supporting base for the foot arch (17–19). Thus, the AT and PF may modulate calcaneal rotation and foot arch deformation, respectively. However, the associations between AT stiffness and calcaneal rotation, and between PF stiffness and foot arch deformation have not been clarified *in vivo*.

The AT and PF in humans are anatomically linked (20–22), indicating that these tissues are series elastic components extending from the lower leg to the forefoot (Fig. 1). Previous *ex situ* studies have demonstrated that the AT and PF exhibit uneven mechanical properties (23,24). The anatomical linkage between the AT and PF leads to the hypothesis that these tissues counterbalance their length changes based on their mechanical properties, modulating the relative position of the calcaneus in the human ankle–foot complex. This may result in changes in tibiocalcaneal and foot arch ROM at the same ankle joint angle. However, there is no *in vivo* evidence to support this hypothesis. Previous studies have reported differences in human AT (25) and PF (26) stiffness between sexes and individuals. Notably, the relationships in stiffness do not correspond between sexes, despite the anatomical connection between the AT and PF; that is, male individuals exhibit a stiffer AT (25) and more compliant PF (26) compared with female individuals. Measuring AT and PF behaviors simultaneously and considering their associations and differences between sexes will enable researchers to develop a theory that integrates these pieces of evidence.

The aim of this study was to examine the associations between AT and PF stiffness and passive ROM of the ankle–foot complex, and the differences between sexes. We hypothesized that 1) the AT and PF exhibit a balanced mechanical interaction during passive ankle–foot motion, 2) AT and PF stiffness are associated with relative calcaneal rotation and foot arch deformation, and 3) AT and PF stiffness and passive ROM of the ankle–foot complex differ between sexes.

METHODS

Participants. Forty young participants, comprising 20 men (mean ± SD: age, 22.8 ± 2.4 yr; height, 1.68 ± 0.04 m; and body mass, 63.5 ± 7.1 kg) and 20 women (age, 23.3 ± 3.5 yr; height, 1.59 ± 0.05 m; and body mass, 53.6 ± 5.1 kg) were recruited for this study. They were all healthy and mildly active, with no history of lower limb or foot injuries within 12 months or surgery in the past. The participants were asked to refrain from strenuous exercise 24 h before the measurements. Referring to data from our preliminary results (n = 12), *a priori* power analysis (G*Power v3.1; Heinrich Heine-Universität Dusseldorf) with an assumed type 1 error of 0.05 and statistical power of 0.80 was conducted to determine statistically significant correlations between maximal length changes in AT and PF. The estimated critical sample size was 16. Thus, 20 men and 20 women were recruited to consider possible attrition. This study was approved by the Human Research Ethics Committee of Waseda University (reference number: 2019-279) and conducted according to the Declaration of Helsinki. Written informed consent was obtained from all participants before data collection.

Magnetic resonance imaging. The participants were asked to rest in a supine position with the knee fully extended on the bed of a 3T magnetic resonance imaging (MRI) scanner (SIGNA Premier 3.0T; GE Healthcare, Chicago, IL). Each

participant's dominant leg and foot were firmly secured to a custom-made fixture (VINE, Tokyo, Japan; Fig. 2A). The longitudinal direction of the foot and leg was carefully aligned to the longitudinal direction of the footplate and leg frame of the fixture, respectively. The fixture was set on a 60-channel bed coil and covered with a 30-channel air coil (GE Healthcare). The participants first performed five slow passive ankle plantar/dorsiflexion cycles from 30° to −20° (0° reflects neutral position, wherein the footplate is perpendicular to tibial axis, and positive/negative values reflect plantar flexion/dorsiflexion, respectively) to account for a possible conditioning effect (27,28). After preconditioning, the leg and foot were secured at footplate angles of 30°, 20°, 10°, 0°, −10°, and −20° in this order, with lower limb and foot muscles fully relaxed. Care was taken to ensure that the foot was in contact with the footplate to avoid unnecessary inversion/eversion or adduction/abduction and that the leg was in the same position during measurements.

Based on previous MRI studies in which AT and PF length changes were measured during passive ankle rotation (10,11), a series of oblique sagittal magnetic resonance (MR) images with a proton density-weighted sequence were obtained for each footplate angle. The scan parameters were as follows: repetition time, 1400 ms; echo time, 20 ms; slice thickness, 1.5 mm; interslice gap, 0 mm; number of slices, 13; field of view, 300 mm × 300 mm; matrix, 512 × 512 pixels; and duration, 45 s. The scanning plane of the oblique sagittal image was carefully adjusted to scan PF and AT lengths (Fig. 2C).

Passive torque measurement. Participants were asked to rest in a supine position with the knee fully extended on the examination bed in an indoor biomechanical laboratory adjacent to the MRI room. Their dominant leg and foot were firmly secured to a custom-made dynamometer (VTF-002; VINE; Fig. 2B) at footplate angles of 30°, 20°, 10°, 0°, −10°, and −20°, with lower limb and foot muscles fully relaxed. Following the same preconditioning as for the MRI measurement, the passive torque was measured for at least 45 s (same as the MRI scan) for each footplate angle, and their mean values were calculated for further analysis. In a pilot test ($n = 6$), we confirmed that the ankle joint angles measured using an electrogoniometer (SG110/A; Biometrics, Newport, United Kingdom) were comparable between the fixture and dynamometer with a regression equation $y = 1.002x + 0.033$ ($R^2 = 0.996$), wherein the intercept was not significantly different from 0 ($P = 0.864$). Therefore, we assumed that the data obtained from these two sessions could be combined.

Data analysis. MR images were processed and analyzed using image analysis software (Osirix MD; Pixmeo, Geneva, Switzerland) based on previous studies by our group (10,11). The lines connecting the anterior vertex of the talus to the anterior tip of the first metatarsus and calcaneal tuberosity defined the forefoot and rearfoot segments orientations, respectively. The tibiocalcaneal angle (θ_{cal}) was measured as the orientation of the rearfoot segment relative to the longitudinal direction of the tibia. The foot arch angle (θ_{arch}) was defined as the relative orientation of the forefoot and rearfoot segments. Here, Figure 2C shows MR images and analysis examples. The angular displacements of θ_{cal} and θ_{arch} were calculated by subtracting the angle measured with a footplate angle of 30° from those measured at 20° to −20°. The ROM values of θ_{cal} and θ_{arch} were calculated by subtracting the angle measured with a footplate angle of 30° from the angle measured at −20°.

The AT moment arm was determined using the center of rotation (COR) method, described in detail elsewhere (29,30). The instantaneous COR between the tibia and talus was determined using the Reuleaux method (31). The instantaneous COR values at footplate angles of 20°, 10°, 0°, and −10° were derived from rotations of 30° → 10°, 20° → 0°, 10° → −10°, and 0° → −20°, respectively. Owing to the design of the

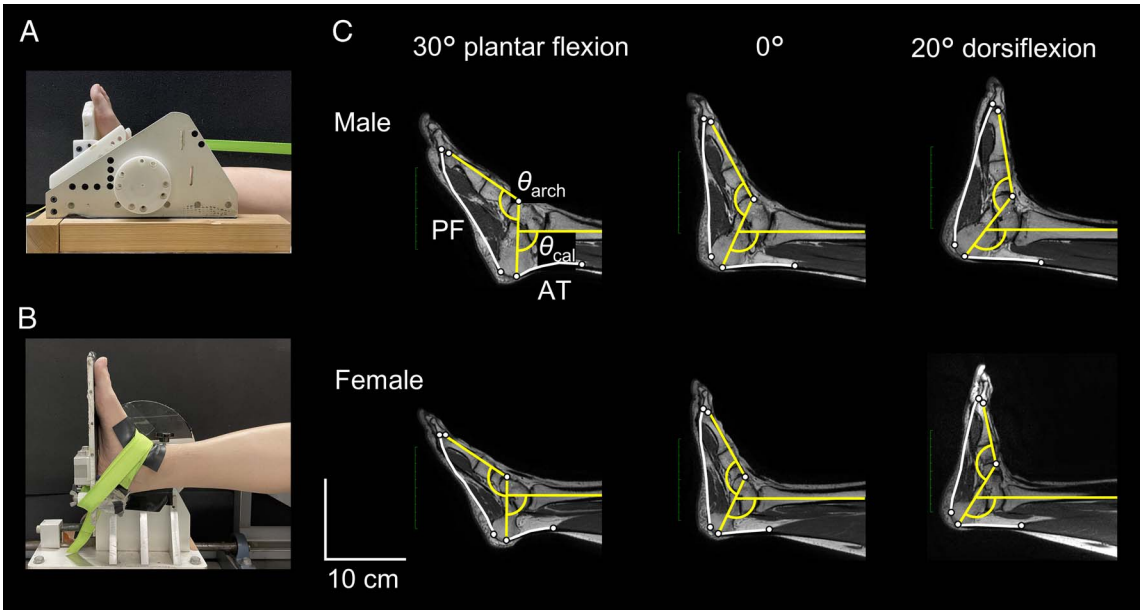

FIGURE 2—Custom-made fixture (A) and dynamometer (B) used in the present study. Representative MR images with a footplate angle of 30°, 0°, and −20° in male and female participants (C).

fixture, we could not obtain MR images with a footplate angle of −30°. Thus, we assumed that the coordination of the instantaneous COR at −20° was the same as that at the footplate angle of −10°. The AT line of action was determined for each footplate angle as a straight line between the distal end of the soleus muscle belly and the insertion site to the calcaneal tuberosity. Subsequently, the AT moment arm was determined as the shortest (perpendicular) distance from the COR to the AT line of action. The AT force (F_{AT}) was calculated using the following equation:

$$F_{AT} = TQ \times MA^{-1} \quad [1]$$

where TQ is the passive ankle plantar flexion torque, and MA is the AT moment arm. In this study, PF force (F_{PF}) was calculated using a previously proposed equation (32):

$$F_{PF} = 0.474 \times F_{AT} + 0.041 \times BW \quad [2]$$

where BW is body weight.

The AT length was manually traced between the distal end of the soleus muscle belly and calcaneal tuberosity. The PF length was manually traced between the medial process of the calcaneal tuberosity and posterior base of the first proximal phalanx. The relative length changes of each component (ΔL_{AT} and ΔL_{PF}) were calculated by subtracting the lengths measured with a footplate angle of 30° from those measured at 20° to −20°. Strains for each component were calculated as the length changes divided by the length measured with a footplate angle of 30°.

The relationships between force and length changes (F_{AT}–ΔL_{AT} and F_{PF}–ΔL_{PF}) were divided into the toe region (where changes in forces for length changes were nonlinear) and linear region (where changes in forces for length changes were proportional) to calculate the AT and PF stiffness values. We combed for the changes in tissue elongation for each footplate angle in descending order (i.e., −20° → −10°, −10° → 0°, etc.); subsequently, we determined the range over which the slope of the linear fit between the change in tissue elongation and force was closest to zero as the linear region (33,34). The slopes of the regression lines in the linear region were calculated as AT and PF stiffness.

Statistical analysis. The normality of the data was assessed using a Shapiro–Wilk test. After normality was confirmed, the ROM of θ_{cal} and θ_{arch} and ΔL_{AT}, ΔL_{PF}, AT and PF strains, passive torque, F_{AT}, and F_{PF} with the ankle–foot in the maximally dorsiflexed position, and AT and PF stiffness between male and female participants were compared using an unpaired t-test. Cohen's d was calculated as a measure of the effect size. Pearson correlation coefficients between ΔL_{AT} and ΔL_{PF}, with the ankle in the maximally dorsiflexed position, were calculated for male and female participants, respectively. Pearson correlation coefficients between AT stiffness and ROM of θ_{cal} and between PF stiffness and ROM of θ_{arch} were calculated for male and female participants to analyze the relationship between tissue stiffness and passive ankle–foot motion. The correlation coefficients were interpreted as weak ($|r| < 0.30$), moderate ($0.30 \leq |r| < 0.50$), and strong ($|r| \geq 0.50$) (35). The slopes of the regression lines between male and female participants were compared using analysis of covariance.

To examine the intrarater and interrater reliabilities of the measurements, an examiner (H. S.) repeated the analyses three times for each participant. The intraclass correlation coefficients ($ICC_{1,3}$), SEM, and coefficient of variation were calculated (36). The $ICC_{2,1}$ values of the measurements were calculated to assess the interrater reliability between the two examiners (H. S. and Y. H.). ICC was interpreted as good ($0.75 < ICC \leq 0.90$) or excellent ($ICC > 0.90$) (37). Statistical analyses were performed using SPSS software (SPSS Statistics 28; IBM, Armonk, NY). The statistical significance level was set at $\alpha = 0.05$.

RESULTS

The ROM of θ_{cal} and θ_{arch} in response to footplate angle changes significantly differed between male and female participants (Fig. 3). The ROM of θ_{arch} accounted for approximately 8%–37% of total ankle–foot motion (13%–37% in men and 8%–29% in

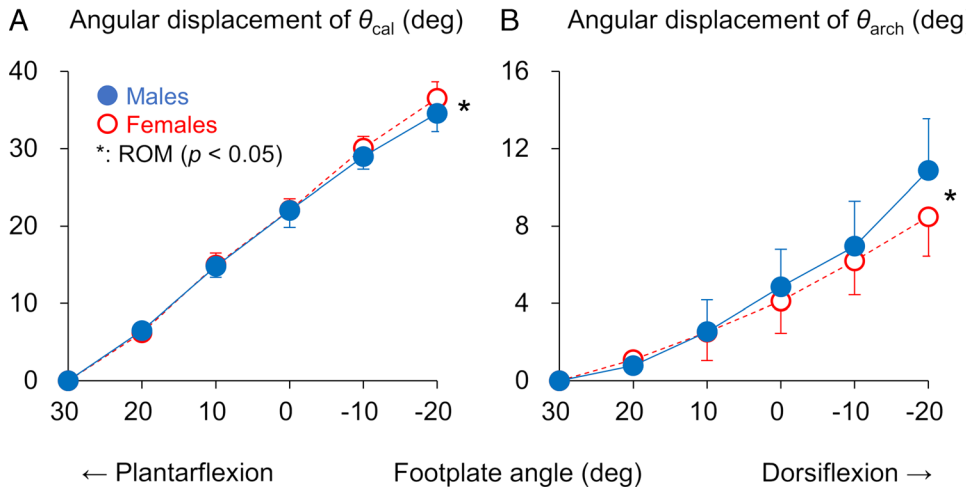

FIGURE 3—Angular displacements of θ_{cal} (A) and θ_{arch} (B) with passive ankle–foot dorsiflexion. *Significant difference in the ROM between male and female participants ($P < 0.05$).

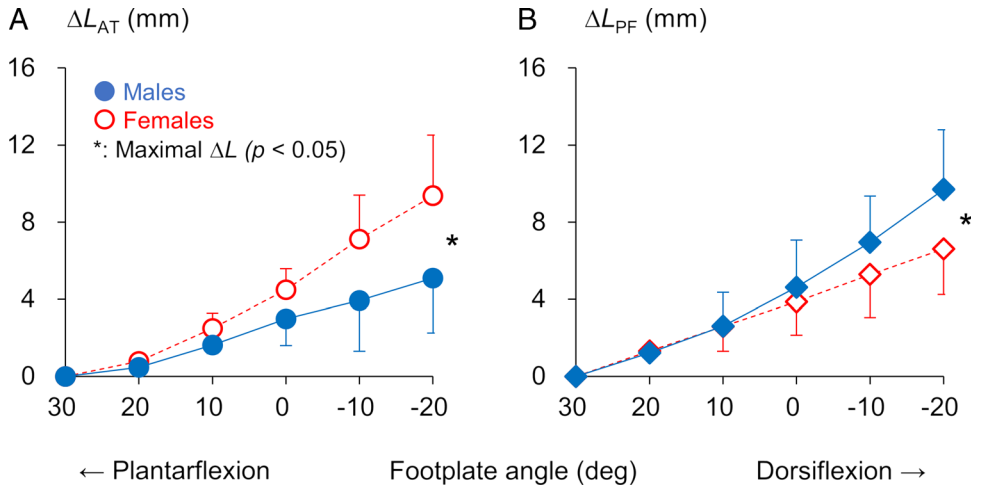

FIGURE 4—ΔL_{AT} (A) and ΔL_{PF} (B) with passive ankle–foot dorsiflexion. *Significant difference in the maximal length changes between male and female participants ($P < 0.05$).

women). Upon passive ankle–foot dorsiflexion, AT and PF were lengthened differently for male and female participants (Fig. 4). Differences in the ROM of θ_{cal} and θ_{arch}, length change and strain of the AT and PF, and passive tension with the ankle–foot in a maximally dorsiflexed position are listed in Table 1. The force–length change relationships of AT and PF are shown in Figure 5. Men exhibited a significantly stiffer AT (140.1 ± 57.9 N·mm^{-1}) and a more compliant PF (22.9 ± 14.7 N·mm^{-1}) than the women (72.8 ± 33.5 N·mm^{-1}, $P < 0.001$, $d = 1.423$, and 46.1 ± 29.8 N·mm^{-1}, $P = 0.035$, $d = 0.987$).

We found significant moderate–strong correlations ($|r|$ ranged from 0.494 to 0.573) between ΔL_{AT} and ΔL_{PF}, between AT stiffness and ROM of θ_{cal}, and between PF stiffness and ROM of θ_{arch} (Fig. 6). The slopes of the regression lines were not significantly different for male and female participants in the relationships between ΔL_{AT} and ΔL_{PF} ($F = 0.321$, $P = 0.574$), between AT stiffness and ROM of θ_{cal} ($F = 1.222$, $P = 0.276$), and PF stiffness and ROM of θ_{arch} ($F = 2.728$, $P = 0.107$). Furthermore, we confirmed that the intrarater and interrater reliabilities of the measurements were excellent for all measured variables, with ICC > 0.90 (Table 2).

DISCUSSION

To the best of our knowledge, this is the first study to clarify the associations between AT and PF stiffness and passive ROM of the ankle–foot components, and their sex differences.

TABLE 1. Sex differences in angular displacement of θ_{cal} and θ_{arch}, length change and strain of AT and PF, and passive tension.

Variables	Male ($n = 20$)	Female ($n = 20$)	P	Cohen's d
θ_{cal} (°)	34.6 ± 3.4	36.5 ± 2.4	0.042	0.646
θ_{arch} (°)	10.9 ± 3.3	8.5 ± 2.4	0.013	0.832
ΔL_{AT} (mm)	5.1 ± 3.0	9.4 ± 3.6	<0.001	1.298
AT strain (%)	6.4 ± 3.9	12.6 ± 4.8	<0.001	1.418
ΔL_{PF} (mm)	9.7 ± 3.7	6.6 ± 3.0	0.006	0.920
PF strain (%)	5.7 ± 2.2	4.1 ± 1.7	0.017	0.814
Passive torque (N·m)	21.1 ± 4.7	16.6 ± 5.4	0.008	0.889
F_{AT} (N)	434.4 ± 117.3	351.5 ± 112.6	0.023	0.721
F_{PF} (N)	231.4 ± 56.6	188.2 ± 53.9	0.014	0.782

Data are shown as mean ± SD. The presented values are with the ankle–foot in a maximally dorsiflexed position.

As hypothesized, we found correlations between AT and PF extensibility, indicating that there is a balanced mechanical interaction between AT and PF upon passive ankle–foot dorsiflexion. Furthermore, for both sexes, stiffer AT and PF were associated with less calcaneal rotation and foot arch deformation, respectively, thereby suggesting commonality of these relationships in humans, regardless of sex. Although previous studies have reported that AT mechanical properties can determine passive ankle joint ROM (15,16), this study provide a novel finding that the balance between AT and PF stiffness can account for breakdown of passive ROM of the ankle–foot complex (i.e., calcaneal rotation and foot arch deformation).

Male individuals exhibited a stiffer AT and more compliant PF than female individuals, which supports the current understanding from previous ultrasonography studies (25,26) and coincides with results that male individuals exhibited less calcaneal rotation and more foot arch deformation than female individuals for comparable ankle–foot motion. These findings

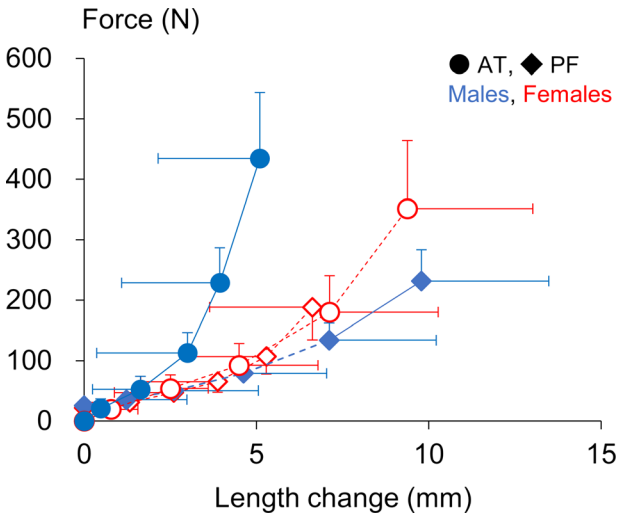

FIGURE 5—Relationship between force and length change in AT (*circles*) and PF (*rhombuses*). Male and female participants are represented with filled blue and opened red symbols, respectively.

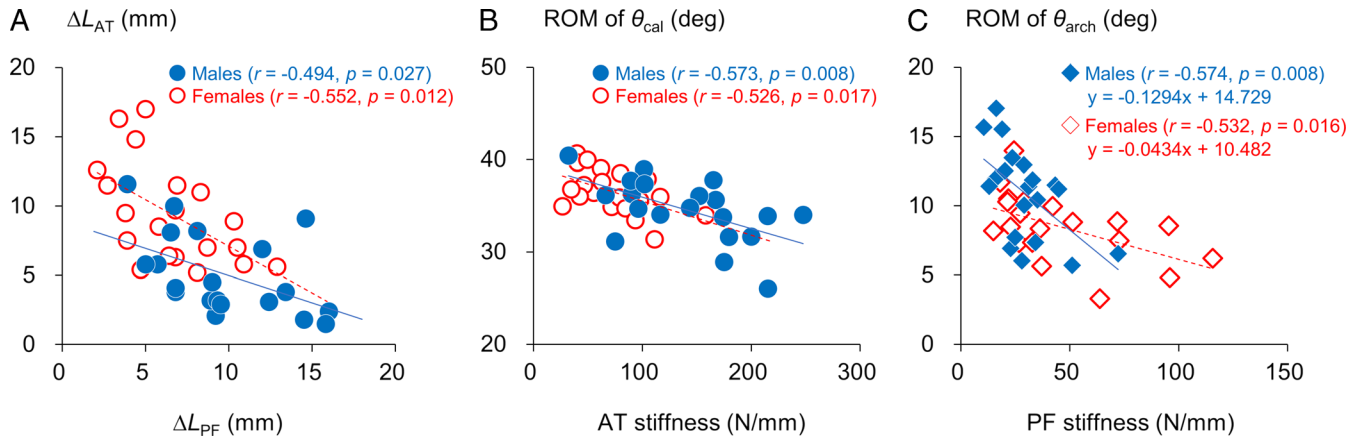

FIGURE 6—Relationships between ΔL_{AT} and ΔL_{PF} (A), between AT stiffness and ROM of θ_{cal} (B), and between PF stiffness and ROM of θ_{arch} (C).

indicate differences in passive ROM of the ankle–foot components between sexes. In the present study, male participants tend to have longer AT and PF (80.9 ± 16.0 and 166.9 ± 7.1 mm) than female participants (70.1 ± 12.3 and 158.2 ± 9.2 mm). However, the strain of AT and PF, a measure that accounts for the differences in the initial length, showed significant differences between sexes, which was consistent with the results for length changes. Thus, it is considered that sex differences in body size did not affect the present results. The slope of regression lines for ΔL_{AT}–ΔL_{PF} and AT stiffness–ROM of θ_{cal} relationships was matched between sexes (Figs. 6A, B), but that for PF stiffness–ROM of θ_{arch} was approximately three times greater in male than in female participants (Fig. 6C). Thus, the latter relationship implies that it may be a factor that differentiates the passive mechanics of the human ankle–foot complex. Furthermore, it has been reported that Achilles tendinopathy seems to be more prevalent in male individuals, and plantar fasciitis more prevalent in female individuals (38–40). Connective tissues, including AT and PF, exhibit adaptability to chronic mechanical loading (e.g., increases in the stiffness and cross-sectional area) (41–44). Therefore, the differences in AT and PF stiffness and passive ankle–foot motion between sexes may reflect the distribution of mechanical loading on AT and PF during daily exercise and relate to pathogenesis of these tissues.

Foot arch deformation accounted for approximately 8%–37% of total ankle–foot ROM. Recent gait analyses have indicated misinterpretations from a conventional single rigid-body segment foot model regarding the angular displacement and generated mechanical power around the ankle joint and foot. Therefore, it is suggested to consider foot arch deformation using a multisegment foot model (45,46). Our results indicate that foot arch deformability has an impact on ankle–foot mechanics, even in a passive state.

Our findings have implications for clinical practice. Static stretching by passive ankle dorsiflexion is widely applied as a conservative treatment of Achilles tendinopathy and plantar fasciitis (47,48). Several clinical trials have examined the effects of different modalities and protocols to alleviate symptoms (49,50), but the optimal duration and intensity of stretching have not been established. Our study suggests that individual variability and sex differences in AT and PF passive extensibility and ankle–foot motion may diversify the static stretching effect. For instance, female individuals may be unable to adequately stretch their PF through ankle dorsiflexion alone, despite their potentially high incidence of plantar fasciitis (38,39). Obtaining individual characteristics will allow tailoring the optimal stretching intensity and protocol for patients.

This study had several limitations. First, all measurements in this study were performed in a relaxed state, but the equation for estimating F_{PF} was derived from a previous cadaver study examining force transmission between AT and PF under load (32). To our knowledge, there is no direct experimental evidence to provide a model of tensile force transmission between AT and PF without incorporating body weight. Thus, the effect of body weight on the force transmission model between AT and PF would need to be investigated in the future. In addition, AT and PF stiffness are not independent because F_{AT} and F_{PF} were calculated from the passive ankle plantar flexion torque. Hence, the relationship between the AT and PF stiffness remains unknown. However, AT and PF are anatomically linked (20–22), and their passive forces are shared (32). To independently measure their mechanical properties, surgical implantation of force transducers (51) or mechanical testing of dissected tissues (23) is required. Therefore, our measurements proposed a realistic approach for understanding the nature of AT and PF *in vivo*. Second, strictly speaking, the COR of the ankle (talocrural) joint moves inside the talus and is independent of θ_{cal}. However, in clinical practice, ankle joint ROM is generally assessed based on the angle between the shank

TABLE 2. Intrarater and interrater reliabilities of the measurements.

	Intrarater			Interrater
	$ICC_{1,3}$ (95% CI)	SEM (mm, deg, N·m)	CV (%)	$ICC_{2,1}$ (95% CI)
θ_{cal}	0.978 (0.936–0.994)	0.42	1.8	0.926 (0.875–0.956)
θ_{arch}	0.958 (0.881–0.989)	0.37	3.0	0.993 (0.989–0.996)
AT length	0.998 (0.997–0.999)	1.18	2.1	0.966 (0.943–0.980)
PF length	0.956 (0.933–0.972)	0.69	5.5	0.932 (0.866–0.959)
Passive torque	0.998 (0.997–0.999)	0.51	1.1	0.997 (0.994–0.998)

CI, confidence of interval; CV, coefficient of variation.

and entire foot. In this context, this study examined and clarified the determinant factor in the breakdown of passive ankle–foot motion. Third, this study considered cross-sectional observation without any intervention; therefore, it remains unclear whether the source of individual variability and sex differences is innate or acquired. Finally, the coefficients of determination (R^2) for the relationships between tissue stiffness and passive ankle–foot ROM ranged from 0.277 to 0.329, and thus, approximately 70% of their variability was unexplained. Other factors such as bony architecture (e.g., midtarsal locking mechanism) (52) and mechanical properties of ligaments, muscles, and nerves (53–55) may be involved. Further research is warranted to fully understand the mechanism that modulates ankle–foot mechanics *in vivo*.

CONCLUSIONS

We have demonstrated that passive AT and PF extensibility counter each other, thereby indicating balanced mechanical interaction between AT and PF in passive ankle–foot motion. AT and PF stiffness and passive ROM of ankle–foot components were countered between sexes; however, associations between their stiffness and passive ROM of the ankle–foot components were consistent between sexes. Our findings support the notion that a balanced mechanical interaction between the AT and PF can account for passive ROM of the human ankle–foot complex *in vivo* and its differences between sexes.

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The authors declare no conflicts of interest. The results of this study do not constitute an endorsement by the American College of Sports Medicine. The results of this study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation.

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