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The 3D *in vivo* Achilles' Tendon Moment Arm, Quantified during Active Muscle Control and Compared Across Sexes

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Abstract

The Achilles' tendon moment arm (ATma) is a critical quantity in that it defines the triceps surae's ability to generate a moment on the calcaneus, which is then transferred to the foot. This measure has been primarily acquired two-dimensionally in small male populations. Thus, the primary purpose of this study was to establish the first *in vivo* three-dimensional measures of the ATma, measured non-invasively during dynamic activity in a large normative population, inclusive of both males and female subjects (n=20). Subjects were each placed supine in a 1.5T MRI and asked to repeat a simulated toe-raise while a full sagittal-cine-phase contrast (dynamic) MRI dataset was acquired. From these data, the 3D and 2D ATma was calculated. The ATma was scaled by the distal tibial width, based on a correlation analysis. The 2D ATma overestimated its 3D counterpart by 3.1 mm, on average. The scaled ATma was no different between the male and female cohorts, but the scaled Achilles' tendon area was smaller in the male cohort. The magnitudes of the ATma were most similar to previously reported values when variations in ankle angle were taken into account. The results of this study have important implications for the applicability of ATma data to both clinical questions and modelling. Any future studies should adapt the ATma based on subject size and/or sex, ensure compatibility between the manner in which the ankle angle is defined and the data being used, and account for the influence that muscle force has on the 3D ATma.

Keywords

ankle; MRI; dynamic; moment arm; lever arm; soleus; gastrocnemius

1. INTRODUCTION

With the incidence of Achilles' tendon rupture, along with other ankle injuries on the rise (Nyyssonen et al., 2008; Suchak et al., 2005), it is imperative that the function of the Achilles' tendon in both the healthy and the diseased state be better understood. Yet, to date, there remains a paucity of 3D, *in vivo* data on the dynamic properties of this tendon. Although non-invasive techniques are being developed (Pourcelot et al., 2005), Achilles' tendon force is typically estimated through musculoskeletal models (Xiao and Higginson,

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CONFLICT OF INTEREST

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2008), due to the invasive nature of current measurement techniques (Finni et al., 1998; Ravary et al., 2004). With the rapid advancement of 3D dynamic musculoskeletal models there is an urgent need for accurate, 3D, *in vivo* measures of Achilles' tendon moment arm (ATma).

Unfortunately, current ATma data (Fath et al., 2010; Klein et al., 1996; Leardini and O'Connor, 2002; Lee and Piazza, 2009; Maganaris, 2004; Rugg et al., 1990) remain two-dimensional, are based on small male populations, and are most often acquired passively; leaving the true, dynamic 3D ATma with active muscle control undefined. This latter point may be crucial, as it has been shown that active muscle control alters the ATma (Maganaris, 2004). Comparing data across studies can be difficult for three reasons. First, the definition of ankle angle varies across studies and is often not reported. Second, three unique 2D methods have been used to calculate the ATma: **1)** the center of rotation (COR) method (Fath et al., 2010; Maganaris, 2004), **2)** the tendon excursion (TE) method (Fath et al., 2010; Klein et al., 1996; Lee and Piazza, 2009; Maganaris, 2004) and **3)** modeling (Leardini and O'Connor, 2002). Lastly, the relationship between the subject's size, weight, and/or foot size to the ATma has not been investigated, except recently in children (Waugh et al., 2011). Thus, variation in subject size is likely an unexplained source of variability in previous studies.

Therefore, the primary purpose of this study was to provide the first *in vivo* 3-dimensional measures of the ATma, measured during dynamic activity in a relatively large normative population, inclusive of both male and female participants. The ATma was calculated as the shortest distance from the Achilles tendon to the instantaneous helical axis (IHA) of the calcaneal-tibial joint. This was most similar to the previous implementations of the COR methodology, but geometric interpretation errors (Shibanuma et al., 2004) were minimized by using cine-phase contrast (PC) magnetic resonance (MR) velocity data to analytically track the musculoskeletal kinematics of the rear-foot during a volitional movement. The relationship between the ATma and measures of subject size (*e.g.*, height, weight, tibial width, foot length, Achilles tendon cross-sectional area, Achilles' tendon length) was explored in order to scale the ATma. Next, the ATma was simplified to a 2D calculation in order to provide insights into the errors associated with calculating the 3D ATma two-dimensionally. Since Achilles' tendon rupture is most prevalent in men (Cretnik and Frank, 2004; Nyyssönen et al., 2008), knowledge of the difference in the 3D ATma between male and females could help elucidate the etiology of these ruptures by enabling more accurate estimates of Achilles' tendon force across the sexes. Unfortunately, to date, only two female subjects have been included within past studies. Thus, the hypothesis that the scaled ATma was not significantly different in males than in females was tested.

2. METHODS

Experimental Data Collection

Twenty healthy subjects (14 males and 6 females, Table 1) participated in this IRB approved study and gave informed consent. Subjects were excluded if they had any contraindications to MR imaging, reported previous foot impairment, pathology, pain, or surgery.

Subjects were placed supine in a 1.5T magnet (LX-9.1M4; GE Medical Systems, Milwaukee, WI, USA) with the hip and knee maintained in full extension (Figure 1). The subject's sock-covered foot was placed into a custom-built ankle-loading-device (Sheehan et al., 2007) so that the calcaneal-tibial axis of rotation aligned with the axis of rotation of the ankle-loading-device, as evidenced by a lack of tibial motion during plantarflexion (PF) and dorsiflexion (DF). In addition, the subject's foot was strapped to the freely moving foot-pedal, which allowed three rotational degrees freedom at the plantar surface of the forefoot

(“ball of the foot”) and had a base plate extending to the mid-calcaneus. Lastly, dual transmit-receive phased array coils were fixed to the ankle-loading-device so that they were centered over the malleoli, medial and lateral to the foot. Plastic stops placed in the ankle-loading-device limited foot-pedal rotation to a comfortable, repeatable range, typically 1–5° less than the subject’s maximum calcaneal-tibial DF to PF range. A resistive load was applied in calcaneal-tibial PF by an external weight system that included a 2.3kg (51b) weight, which hung freely outside the MR imager. A cam, integral to the ankle-loading-device, created a fixed moment arm from the rope to the ankle-loading-device’s center of rotation. The loading level was selected such that all subjects could smoothly and comfortably perform the task without reporting fatigue at the end of the trial (based on a preliminary analysis).

A full fast-PC MR image set (x,y,z velocity and anatomic images over 24 time frames) was acquired while the subjects cyclically repeated simulated single-leg toe-raises with partial-weight support. Specifically, subjects were asked to push the foot-pedal down and release it back to the beat of an auditory metronome (cycle rate=35 cycles/minute with 2 beats/cycle). Imaging time was approximately 4 minutes. The foot-pedal allowed the rear-foot joints to move in three degrees of rotation; internal-external and inversion-eversion, as well PF-DF. Prior to data collection, subjects practiced the task until they could comfortably repeat the motion. In order to acquire three-dimensional high resolution static MR images (sagittal and axial), the plastic stops were moved so that the rear foot was held in a stationary position as close to anatomically neutral as possible. The 3D static images were reviewed by a musculoskeletal radiologist to confirm the absence of foot pathology.

Calculation of the ATma

Unlike other imaging techniques, cine-PC MR imaging acquires velocity data directly. Thus, the orientation and displacement of the tibia and calcaneus were individually quantified by integrating the cine-PC MR velocity data. Being independent of the need to visually track anatomical points of interest, cine-PC MR provides excellent accuracy {< 0.5 mm (Sheehan et al., 1998)} and subject repeatability {repeatability of calcaneal-tibial kinematics across two trails <1.8° and <1.5mm, (Sheehan et al., 2007)} for tracking *in vivo* 3D musculoskeletal kinematics. From the orientation data, the angular velocity of the calcaneus relative to the tibia was quantified as a unit vector ($\hat{\mathbf{u}}$), which defined the direction of the calcaneal-tibial instantaneous-helical-axis {IHA, (Sheehan, 2010)}. In a similar manner, the most distal point on the soleus myotendinous junction ($\mathbf{p}^{\text{SOL}}$) was tracked through integration of the cine-PC MR data (Figure 2). Using calcaneal kinematics, the 3D displacement of the Achilles’ tendon insertion into the calcaneus ($\mathbf{p}^{\text{CAL}}$) was analytically tracked. $\mathbf{p}^{\text{SOL}}$ and $\mathbf{p}^{\text{CAL}}$ defined the Achilles’ tendon line-of-action ($\mathbf{r}^{\text{OQ}}$).

The 3D ATma was defined as equal to the perpendicular distance between and the IHA (Sheehan, 2007), such that multiplying it by the magnitude of the Achilles’ tendon force produced the magnitude of the moment created by the Achilles’ tendon force about the point on the IHA nearest to the Achilles’ tendon (Appendix A):

$$ATma = \frac{(\hat{\mathbf{F}} \times \hat{\mathbf{u}})}{|\hat{\mathbf{F}} \times \hat{\mathbf{u}}|} \bullet \mathbf{r}^{\text{OQ}} \quad (\text{eq1})$$

Bolds, “^”, and non-bolded characters denote vectors, unit vectors and magnitudes, respectively.

$\mathbf{r}^{\text{OQ}}$ = The position vector from any point on the IHA to any point on $\hat{\mathbf{F}}$

$\hat{\omega}$ = The direction of the IHA

As a point of comparison the 2D ATma (ATma_{2D}) was calculated by assuming that the IHA was directed purely in the medial-lateral direction and that $\hat{\mathbf{f}}$ was isolated to the sagittal plane of the tibial coordinate system (Figure 2). This calculation, as opposed to a simple projection of the ATma into a 2D plane, was selected to best mimic previous 2D COR calculations of the ATma (Appendix A). Both the ATma and the ATma_{2D} were reported relative to the tibial shaft-to-foot (tib-foot) angle, defined to approximate the clinical ankle angle. Negative tib-foot angles indicated DF, positive angles indicated PF, and zero was the anatomically neutral position.

Statistical Analysis

To create population averages, all data were interpolated to single degree tib-foot angle increments. Due to variations in flexibility, not all subjects could reach the extremes of PF and DF. Thus, reported averages were restricted to ankle angles where there were more than three subjects represented in each cohort. In an attempt to account for potential size variations across subjects four variables were measured, using the 3D static sagittal and axial MR images {Figure 3, foot_length, Achilles' tendon cross-sectional area (AT_area), distal tibial width (Twidth), and Achilles' tendon free length (AT_length)}. These four variables along with subject height and weight were correlated (Pearson's) with the value of the ATma at 20° tib-foot angle. This angle was selected, as it was the mid-range of motion and an angle in which all subjects were represented. In addition, a step-wise linear regression was performed to determine if the ATma could be predicted from these six variables. Since Twidth demonstrated the strongest, significant correlation with the ATma (Table 2), it was used to scale the ATma:

$$ATma(scaled) = ATma(unscaled) * \frac{\text{average T Width}}{\text{individual T width}} \quad (\text{eq2})$$

average Twidth = average Twidth over the 20 subjects (49.2mm, Table 1)

individual Twidth = subject specific Twidth

An a priori power analysis determined that 5 subjects were needed per group ($\alpha = 0.05$ and $\beta = 0.9$) to test the hypothesis that the ATma(*scaled*) was smaller in females than in males. Since previous data were not available for female subjects, it was assumed that the variance in both populations would be similar to that reported previously, 4.0mm (Maganaris, 2004) and that difference in ATma between genders would be half the difference seen between previously between maximum DF and PF (5.0 mm). Comparisons between groups were completed using a 2-tailed Student's t-test, assuming equal variances. To limit the number of statistical tests, the comparison between the male and female cohorts was completed at a tib-foot angle of 20° only, matching the correlation analysis.

3. RESULTS

The ATma(*scaled*) rose from the dorsiflexed to neutral position and then dropped slightly throughout plantarflexion (Table 3 and Figure 4). On average the ATma_{2D}(*scaled*) overestimated the ATma(*scaled*) by 3.1mm (~6.1%), with the largest differences occurring near the end ranges of motion.

The ATma(*scaled*) was no different between the male and female cohorts ($\Delta ATma = 3.9\text{mm}$, a 7.2% difference, $p = 0.24$, Figure 5 and Table 3), negating the null hypothesis. The

importance of scaling was evidenced by the fact that without scaling a significant difference in the ATma between cohorts did exist ($\Delta\text{ATma}(\text{unscaled}) = 8.3\text{mm}$, a 15.0% difference, $p=0.03$).

Twidht, AT_length, height, and weight were all larger in the male versus the female cohort (Table 1). The AT_area was no different between the two cohorts (0.80mm^2 , $p = 0.86$). Yet, the AT_area(*scaled*) was smaller in males than in females (-11.3mm^2 , $p = 0.02$). The scaling factor for the AT_area was weight, as this was the one variable that moderately correlated with the AT_area. Foot_length and Twidht were both moderately correlated with ATma, Twidht, and weight. Foot length and Twidht were strongly and moderately correlated with height, respectively. The measurement of AT_length was highly variable across subjects and demonstrated no correlations with the other variables of interest. The stepwise linear regression was able to explain 52.5% ($p=0.003$) of the variance in the ATma using just two variables ($\text{ATma} = 10.472 + 1.255*\text{Twidht} - 0.344*\text{AT_area}$).

If the data from past studies were centered at the mid-tib-foot-angle (18°) of the current study (Figure 6), the majority of data fell within one standard deviation of the current ATma(*scaled*). After the shift, the data reported by Fath and colleagues (2010) was nearly identical to the current study. Similarly, the current data fell between the values reported for passive rotation and maximum isometric contraction (Maganaris, 2004).

4. DISCUSSION

This study is the first to quantify the 3D Achilles' tendon moment arm non-invasively and *in vivo* during loaded PF, using a population at least twice the size of most past studies. The functional movement evaluated in the current study can be related to a subject standing with the balls of their feet on a beam raising and lowering partial body weight through a range of motion, just shy of the maximum PF and DF, under their own muscular control. Being that Cine-PC MRI is the only available technique that can simultaneously track 3D muscular and skeletal kinematics *in vivo*, it removes the potential errors associated with simplifying a 3D motion (Sheehan et al., 2007) to 2D, as required for the 2D static COR methodology. Calculating the ATma in two-dimensions leads to an overestimation of the ATma (Appendix A) and exposes the analysis to potential geometric errors (Shibanuma et al., 2005). Cine-PC MRI also enables the evaluation of a volitional task, requiring active muscle control, which is not feasible using the TE methodology, as the required assumption of no work being done, as the ankle progresses through its range of motion, is violated.

After accounting for the variation in the ankle angle definition across studies, there was a surprising similarity across results near the anatomically neutral position, but not at larger angles of PF. Since it has been previously shown that the ATma increases with increased Achilles' tendon force (Maganaris, 2004), it is likely that the previous studies overestimated the ATma under passive conditions. This overestimation can be attributed to two likely factors. The first is the simple fact that the ATma_{2D} overestimated its 3D counterpart due to the non-sagittal plane motion. This overestimation may be larger than the currently reported differences in the ATma and ATma_{2D} (Figure 4), as the ATMA_{2D} did not contain the potential geometric errors associated with tracking 3D bone movement in a 2D image (Shibanuma et al., 2004). The sole focus on male subjects of most previous studies, without accounting for subject size would have resulted in a larger ATma than the one reported currently.

One major difference in the current data, as compared to previously reported data (Maganaris, 2004; Rugg et al., 1990), was the slight decrease in ATma during plantarflexion. This cannot be attributed to the fact that the current study calculated the

ATma from calcaneal-tibial IHA, and not talocrural IHA, as the location of the calcaneal-tibial and talocrural IHAs do not differ greatly from each other in the axial plane (Sheehan, 2010). Since all subjects demonstrated a similar relationship between the ATma and tib-foot angle, the variability in range-of-motion across subjects was likely not a contributing factor in the slope of the ATma versus tib-foot angle. Therefore, the variation in slope from the current study to those in the past may be due to the fact that these previous studies standardized the ankle angles at which images of the rear foot were acquired and did not allow subjects to select their own natural movement. This could have easily resulted in an unnatural rear-foot alignment, particularly at the end ranges of motion as the talus became locked into the mortise.

No previous study has examined the validity and utility of scaling to account for size differences across subjects. Scaling, as opposed to normalization, was used to provide direct comparison to past results. The correlation analysis revealed that both *Twidth* and *foot_length* were moderately correlated with ATma and thus, were appropriate scale factors. Scaling using *Twidth* had two primary effects, it reduced the variability of the ATma and it eliminated the difference between the male and female cohorts. Combining *Twidth* and *AT_area* enabled for 52.5% variability in the ATma to be predicted, which is slightly higher than that reported for predicting the ATma in children, 49% (Waugh et al., 2011). A combined *Twidth* and *AT_area* scaling factor was not used, as the simpler single scaling factor (*Twidth*) was more clinically applicable and relevant.

As the scaled ATma was no different between the male and female cohorts, this variable does not provide an explanation for the higher incidence of Achilles' tendon rupture in males. The differences seen in the ATma(*unscaled*) indicate that previous estimations of Achilles' tendon force within a female cohort would have erred on the side of being too low, if they were based on published data for male subjects, without proper scaling (Wunderlich and Cavanagh, 2001; Yu et al., 2008). This could have profound effects on the conclusions reached using modelling to evaluate female performance. Interestingly, unlike the ATma, the *AT_area* was moderately correlated with weight only. The ATma, scaled by body weight (Kongsgaard et al., 2005), was smaller in the male, as compared to the female, cohort. All things being equal, this decrease would result in higher tendon stresses, which could partially explain the higher prevalence of Achilles' tendon rupture in men (Cretnik and Frank, 2004; Nyssönen et al., 2008).

The primary limitation of this study was the fact that the ATma was quantified during emulated partial-weight bearing, instead of full-weight bearing. This was necessary in order to exclude fatigue and maintain volunteer comfort. As open MR imaging technology improves, experiments including full-weight bearing will become available. The non-invasive dynamic nature of this experiment, its ability to incorporate muscle control, and its excellent accuracy/subject-repeatability justify this potential limitation. Further, it was not possible to know the exact shift in ankle angle required to align previous estimations of the ATma with the current results. Centering the data appeared appropriate, as it created a consistency across studies. Although this study represents one of the largest cohorts used for the quantification of the ATma, it is still a relatively small population and future work should focus on quantifying the ATma in a larger female cohort, as well as in specific ethnic populations. For example, the ratio of Achilles' tendon rupture rate between African Americans and Caucasians within the US military is 2.9 (White et al., 2007). This larger database would also provide the power needed to expand the comparison between sexes from a single tib-foot angle to the full range of motion. As *AT_area* has been shown to both increase (Magnusson and Kjaer, 2003) or not increase (Urlando and Hawkins, 2007) with intermittent high levels of load and the overall numbers of females in the current study was

small, future work exploring the relationship of the AT_area with subject size, weight, and exercise level is warranted.

In conclusion, having accurate and reliable data for the 3D ATma is critical both clinically and experimentally as it defines how the force within the triceps surae is transformed into a moment on the calcaneus. The current study not only provides such data (acquired *in vivo* during dynamic activities and inclusive of both male and female subjects) it highlights four key points for future research requiring estimates of the ATma. Any future studies should account for the error associated with using a 2D ATma for a 3D analysis and ensure compatibility between the manner in which the ankle angle is defined and the data being used, as a shift in ankle angle greatly alters the compatibility of results across studies. Future studies should adapt the ATma relative to subject size. Twidth, foot_length, and AT_area are the best variables for scaling and predicting the ATma, based on the current and previous results (Waugh et al., 2011). Lastly, the relationship between muscle force and ATma should be taken into account when using ATma data in modelling studies.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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REFERENCES

- Cretnik A, Frank A. Incidence and outcome of rupture of the Achilles tendon. *Wien Klin Wochenschr.* 2004; 116 Suppl 2:33–38. [PubMed: 15506308]
- Fath F, Blazevich AJ, Waugh CM, Miller SC, Korff T. Direct comparison of in vivo Achilles tendon moment arms obtained from ultrasound and MR scans. *J Appl Physiol.* 2010
- Finni T, Komi PV, Lukkariniemi J. Achilles tendon loading during walking: application of a novel optic fiber technique. *Eur J Appl Physiol Occup Physiol.* 1998; 77(3):289–291. [PubMed: 9535592]
- Klein P, Mattys S, Rooze M. Moment arm length variations of selected muscles acting on talocrural and subtalar joints during movement: an in vitro study. *J Biomech.* 1996; 29(1):21–30. [PubMed: 8839014]
- Kongsgaard M, Aagaard P, Kjaer M, Magnusson SP. Structural Achilles tendon properties in athletes subjected to different exercise modes and in Achilles tendon rupture patients. *J Appl Physiol.* 2005; 99(5):1965–1971. [PubMed: 16081623]
- Leardini A, O'Connor JJ. A model for lever-arm length calculation of the flexor and extensor muscles at the ankle. *Gait Posture.* 2002; 15(3):220–229. [PubMed: 11983496]
- Lee SS, Piazza SJ. Built for speed: musculoskeletal structure and sprinting ability. *J Exp Biol.* 2009; 212(Pt 22):3700–3707. [PubMed: 19880732]
- Maganaris CN. Imaging-based estimates of moment arm length in intact human muscle-tendons. *Eur J Appl Physiol.* 2004; 91(2–3):130–139. [PubMed: 14685871]
- Magnusson SP, Kjaer M. Region-specific differences in Achilles tendon cross-sectional area in runners and non-runners. *Eur J Appl Physiol.* 2003; 90(5–6):549–553. [PubMed: 12905044]
- Nyyssonen T, Luthje P, Kroger H. The increasing incidence and difference in sex distribution of Achilles tendon rupture in Finland in 1987–1999. *Scand J Surg.* 2008; 97(3):272–275. [PubMed: 18812279]

- Pourcelot P, Defontaine M, Ravary B, Lematre M, Crevier-Denoix N. A non-invasive method of tendon force measurement. *J Biomech.* 2005; 38(10):2124–2129. [PubMed: 16084214]
- Ravary B, Pourcelot P, Bortolussi C, Konieczka S, Crevier-Denoix N. Strain and force transducers used in human and veterinary tendon and ligament biomechanical studies. *Clin Biomech (Bristol, Avon).* 2004; 19(5):433–447.
- Rugg SG, Gregor RJ, Mandelbaum BR, Chiu L. In vivo moment arm calculations at the ankle using magnetic resonance imaging (MRI). *J Biomech.* 1990; 23(5):495–501. [PubMed: 2373722]
- Sheehan FT. The 3D patellar tendon moment arm: quantified in vivo during volitional activity. *J Biomech.* 2007; 40(9):1968–1974. [PubMed: 17161841]
- Sheehan FT. The instantaneous helical axis of the subtalar and talocrural joints: a non-invasive in vivo dynamic study. *J Foot Ankle Res.* 2010; 3:13. [PubMed: 20626876]
- Sheehan FT, Seisler AR, Siegel KL. In vivo talocrural and subtalar kinematics: a non-invasive 3D dynamic MRI study. *Foot Ankle Int.* 2007; 28(3):323–335. [PubMed: 17371656]
- Sheehan FT, Zajac FE, Drace JE. Using cine phase contrast magnetic resonance imaging to non-invasively study in vivo knee dynamics. *J Biomech.* 1998; 31(1):21–26. [PubMed: 9596534]
- Shibanuma N, Sheehan FT, Lipsky PE, Stanhope SJ. Sensitivity of femoral orientation estimates to condylar surface and MR image plane location. *J Magn Reson Imaging.* 2004; 20(2):300–305. [PubMed: 15269957]
- Shibanuma N, Sheehan FT, Stanhope SJ. Limb positioning is critical for defining patellofemoral alignment and femoral shape. *Clin Orthop Relat Res.* 2005; 434:198–206. [PubMed: 15864053]
- Suchak AA, Bostick G, Reid D, Blitz S, Jomha N. The incidence of Achilles tendon ruptures in Edmonton, Canada. *Foot Ankle Int.* 2005; 26(11):932–936. [PubMed: 16309606]
- Urlando A, Hawkins D. Achilles tendon adaptation during strength training in young adults. *Med Sci Sports Exerc.* 2007; 39(7):1147–1152. [PubMed: 17596783]
- Waugh CM, Blazeovich AJ, Fath F, Korff T. Can Achilles tendon moment arm be predicted from anthropometric measures in pre-pubescent children? *J Biomech.* 2011; 44(10):1839–1844. [PubMed: 21561625]
- White DW, Wenke JC, Mosely DS, Mountcastle SB, Basamania CJ. Incidence of major tendon ruptures and anterior cruciate ligament tears in US Army soldiers. *Am J Sports Med.* 2007; 35(8):1308–1314. [PubMed: 17468380]
- Wunderlich RE, Cavanagh PR. Gender differences in adult foot shape: implications for shoe design. *Med Sci Sports Exerc.* 2001; 33(4):605–611. [PubMed: 11283437]
- Xiao M, Higginson JS. Muscle function may depend on model selection in forward simulation of normal walking. *J Biomech.* 2008; 41(15):3236–3242. [PubMed: 18804767]
- Yu J, Cheung JT, Fan Y, Zhang Y, Leung AK, Zhang M. Development of a finite element model of female foot for high-heeled shoe design. *Clin Biomech (Bristol, Avon).* 2008; 23 Suppl 1:S31–S38.

Figure 1. Dynamic MRI Setup

A custom-built ankle-loading-device (Sheehan et al., 2007) supported a dual transmit-receive phased array coil medial-lateral to the foot, helped guide the motion and enabled a force to be applied in either plantarflexion or dorsiflexion.

Figure 2. Key Anatomical Definitions

The insertion of the Achilles' tendon into the calcaneus (Pcal), the anterior edge of the tibia ($\mathbf{t_y}$), the line connecting the most medial to the most lateral edge of the tibia ($\sim \mathbf{t_z}$) in the dynamic cine axial image, and a line connecting the most distal aspect of the calcaneus and the (typically 3rd) metatarsal ($\mathbf{l_{foot}}$) were each identified in the image where the tibial shaft-to-foot (tib-foot) angle was as close to neutral as possible. The tibial coordinate system was established such that $\mathbf{t_y}$ was directed superiorly, $\mathbf{t_x}$ ($\mathbf{t_y} \times \sim \mathbf{t_z}$) was pointed anteriorly, and $\mathbf{t_z}$ ($\mathbf{t_x} \times \mathbf{t_y}$) was pointed medially. The tib-foot angle was defined as the 90° minus the angle between $\mathbf{t_y}$ and $\mathbf{l_{foot}}$, in order to approximate the externally defined clinical ankle angle. The tib-foot angle and Pcal were then tracked throughout the cycle based on the tibial and calcaneal kinematics, defined through integration of the cine-PC data. The origin of the Achilles' tendon (Psol) was tracked through integration of the soleus data.

Figure 3. Static Measures for correlations

All measures were acquired from the 3D static sagittal and axial images of the foot and tibia in a position as close to anatomical neutral as possible. The 3D light-box tool in MIPAV (Medical Image Processing and Visualization, NIH, Bethesda, MD) was used to visualize each point in all three cardinal planes, ensuring accurate identification of all points of interest. **Foot_length**: measured from the most posterior point on the calcaneus (Pdc, shown in 2 planes) to the most inferior point of the 1st metatarsal (P1m, shown in two planes).

AT_area (Achilles' tendon area): The area of the Achilles' tendon, measured in the image just inferior to the soleus' myotendinous junction. A single location was selected as a previous study found no variation in the cross sectional area of the Achilles' tendon along the length of the tendon (Magnusson and Kjaer, 2003). **Twidht** (Distal tibial width): The widest medial to lateral distance on the distal tibia, measured in the axial plane (typically at the level of the growth plate). **AT_length** (Achilles' tendon free length): The distance from the distal end of the soleus' myotendinous junction to the Achilles' tendon's insertion into the calcaneus.

Figure 4. Scaled Achilles' Tendon Moment Arm in two- and three-dimensions

The 3D ATma(*scaled*) is shown with star symbols and single standard deviation bars every 5 degrees. Its 2D counterpart, calculated assuming pure sagittal plane rotation (Appendix A), has single standard deviation bars provided every five degrees.

Figure 5. Achilles' Tendon Moment Arm

The average of all subjects is shown with a solid line (no symbols) with a surrounding grey area representing one standard deviation from the average. The female cohort is represented by open circles and the male population by open squares. Ranges where there were less than three subjects represented in the average were not displayed on the graph.

Figure 6. Comparison of previous and current work: ATma(scaled)

The representations of past work were typically approximated from graphical data. The 3D and not the 2D ATma was selected as the comparator, as it represents the true ATma and has a lower variability than the ATma_{2D}. **A.** Direct comparison to previous ATma data (COR methodology only). The Maganaris et al study (2004) included 6 male subjects and quantified the ATma during passive PF and during maximum voluntary contraction. Leardini et al. (2002) quantified the ATma based on a model, Rugg et al. (1990) quantified the ATma based on 10 male subjects while a triceps surae force was applied and Fath et al. (2010) quantified the ATma based on 7 male and 2 female subjects with passive rotation. **B** Identical data to **A**, but the ankle angle of the past studies was adjusted such that all data

were centered at the mid-tib-foot-angle (18°) of the current study. The majority of published data on the ATma reference an ankle angle that was measured externally (Fath et al., 2010; Lee and Piazza, 2009; Maganaris, 2004), likely making it smaller than the tib-foot angle. In contrast, the ankle angle, as defined by Rugg and colleagues (1990), was larger than the tib-foot angle. C. Past studies that quantified the ATma based on the TE method. Klein et al. (1996) and Lee et al. (2009) only reported the ATma for a single ankle angle using 10 male cadavers and 12 male subjects, respectively.

Demographics and key subject size measures. With the exception of the number of subjects each entry of the first three data columns provides the average value for the specific cohort along with one standard deviation ($XX \pm XX$). The fourth data column provides the difference between the male and female cohort. The final column provides the p-value of a Student's t-test (two-tailed, equal variance) comparing the values from the male and female cohorts. P-values less than 0.05 are bolded.

Table 1

	All		Females		Males		Diff	p-value
	Ave $\pm$ SD	SD	Ave $\pm$ SD	SD	Ave $\pm$ SD	SD	M-F	
Number	20		6		14			
Age (years)	25.5 $\pm$ 3.7		25.0 $\pm$ 4.2		25.6 $\pm$ 3.7		0.6	0.75
Height (cm)	174.3 $\pm$ 7.8		164.1 $\pm$ 5.3		178 $\pm$ 4.6		13.9	<0.001
Weight (kg)	70.1 $\pm$ 13.1		59.2 $\pm$ 8.7		74 $\pm$ 12.4		14.8	.02
Twidth (mm)	49.2 $\pm$ 3.4		46.1 $\pm$ 3.4		50.4 $\pm$ 2.7		4.3	.01
Foot_length (mm)	176.6 $\pm$ 10.5		163.2 $\pm$ 11.4		181.4 $\pm$ 4.4		18.2	<0.001
AT_length (mm)	48.3 $\pm$ 13.3		34.9 $\pm$ 13.1		53.1 $\pm$ 10.0		18.2	.004
AT_area (mm²)	53.9 $\pm$ 8.5		53.3 $\pm$ 9.8		54.1 $\pm$ 8.4		0.8	0.86

Table 2

Correlations: between 3D ATma and six measures of subject size (AT_area, Twidth, AT_length, Foot_length, Height, Weight).

	ATma	AT_area	Twidth	AT_length	Foot_length	Height
AT_area	-0.35					
Twidth	0.58**	0.14				
AT Free Length	0.23	0.15	0.25			
Foot_length	0.51*	-0.02	0.69**	0.17		
Height	0.38	-0.03	0.49*	0.35	0.81**	
Weight	0.22	0.58*	0.48*	0.22	0.50*	0.35

P < 0.001 is indicated by ** and *P* < 0.05 is indicated by *.

Scaled 3D Achilles' Tendon Moment Arm (mm): Entries that are greyed out represent tib-foot angles at which there were fewer than 3 subjects.

Table 3

Tib-Foot Angle	All		Male		Female	
	Average	SD	Average	SD	Average	SD
-5	46.8	4.8	46.8	4.8		
0	53.8	10.0	53.8	10.0		
5	55.5	8.1	54.0	7.0		
10	53.3	6.5	53.7	6.5		
13	55.3	9.4	53.6	6.1	60.7	16.6
15	55.5	7.0	54.3	5.2	59.7	11.3
20	53.4	6.3	54.4	5.5	50.5	8.2
25	52.3	5.7	52.6	6.4	51.6	4.1
30	49.7	4.7	51.5	5.1	47.3	3.0
33	47.5	7.5	45.9	9.2	50.1	3.2
34	51.0	5.5			51.3	4.9
35	49.8	5.6				
40	49.3	4.3				