Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Quantifying Achilles tendon force *in vivo* from ultrasound images

Taylor J.M. Dick [a,*,1], Allison S. Arnold [b,1], James M. Wakeling [a,1]

[a] Department of Biomedical Physiology and Kinesiology, Simon Fraser University, Burnaby, BC, Canada
[b] Concord Field Station, Harvard University, Bedford, MA, USA

ARTICLE INFO

Article history:
Accepted 28 July 2016

Keywords:
Ultrasound
Tendon stiffness
Muscle force
Ankle moment

ABSTRACT

This study evaluated a procedure for estimating *in vivo* Achilles tendon (AT) force from ultrasound images. Two aspects of the procedure were tested: (i) accounting for subject-specific AT stiffness and (ii) accounting for changes in the relative electromyographic (EMG) intensities of the three triceps surae muscles. Ten cyclists pedaled at 80 rpm while a comprehensive set of kinematic, kinetic, EMG, and ultrasound data were collected. Subjects were tested at four crank loads, ranging from 14 to 44 N m (115 to 370 W). AT forces during cycling were estimated from AT length changes and from AT stiffness, which we derived for each subject from ultrasound data and from plantar flexion torques measured during isometric tests. AT length changes were measured by tracking the muscle-tendon junction of the medial gastrocnemius (MG) relative to its insertion on the calcaneus. Because the relative EMG intensities of the triceps surae muscles varied with load during cycling, we divided subjects' measured AT length changes by a scale factor, defined as the square root of the relative EMG intensity of the MG, weighted by the fractional physiological cross-sectional areas of the three muscles, to estimate force. Subjects' estimated AT forces during cycling increased with load ($p < 0.05$). On average, peak forces ranged from 920 ± 96 N (14 N m, 115 W) to 1510 ± 129 N (44 N m, 370 W). For most subjects, ankle moments derived from the ultrasound-based AT strains were 5–12% less than the net ankle moments calculated from inverse dynamics ($r^2 = 0.71 \pm 0.28$, RMSE $= 8.1 \pm 0.33$ N m). Differences in the moments increased substantially when we did not account for changes in the muscles' relative EMG intensities with load or, in some subjects, when we used an average stiffness, rather than a subject-specific value. The proposed methods offer a non-invasive approach for studying *in vivo* muscle-tendon mechanics.

1. Introduction

The purpose of this study was to evaluate a procedure for estimating Achilles tendon (AT) forces from ultrasound-based measurements of AT length changes. Over the past decade, B-mode ultrasound has emerged as a useful tool for quantifying *in vivo* muscle-tendon parameters during movement. For example, ultrasound images of muscle fascicles have been recorded during walking, running (Lichtwark and Wilson, 2006; Farris and Sawicki, 2012), and jumping (Kurokawa et al., 2001) and have been used to quantify fascicle strains. Ultrasound has also been used, with motion capture, to measure mechanical properties of tendons under isometric conditions (Kongsgaard et al., 2011) and during walking (Lichtwark and Wilson, 2006) and hopping (Lichtwark and Wilson, 2005). These and other studies (*e.g.*, Maganaris, 2003) have shown that ultrasound-based approaches can provide measures of muscle-tendon mechanics that cannot be obtained from motion capture alone.

Ultrasound-based measures of AT length changes, combined with information about the tendon's stiffness and slack length, could provide information about AT forces during dynamic tasks. For example, Lichtwark and Wilson (2006) estimated the average AT force generated by six subjects during walking by multiplying the subjects' average AT length changes, measured *via* ultrasound, by an average tendon stiffness that was estimated in a prior study (Lichtwark and Wilson, 2005). However, determining the forces transmitted by a complex, composite tendon such as the AT – in individual subjects – remains challenging for several reasons. First, AT stiffness has been shown to vary across subjects (Magnusson et al., 2001; Kubo et al., 2003; Lichtwark and Wilson, 2005; Morrison et al., 2015), and whether an average stiffness is sufficient to estimate force, or whether stiffness must be estimated on a per subject or per muscle basis, remains unknown. Second, three large muscles insert into the AT, and the relative contributions of the medial and lateral gastrocnemius (MG, LG) and soleus (SOL) to AT force may change, depending on the task (*e.g.*, Wakeling and Horn, 2009; Wakeling et al., 2010). Thus, strains measured at a single muscle-tendon junction (MTJ) may not be representative of strains throughout the tendon (Franz et al., 2015).

* Corresponding author. Fax: +778 782 3040.
E-mail address: taylor_dick@sfu.ca (T.J.M. Dick).

[1] All authors were directly involved in the preparation and execution of this study. All authors contributed to the writing of the manuscript. The material in this manuscript will not be submitted for publication elsewhere.

http://dx.doi.org/10.1016/j.jbiomech.2016.07.036
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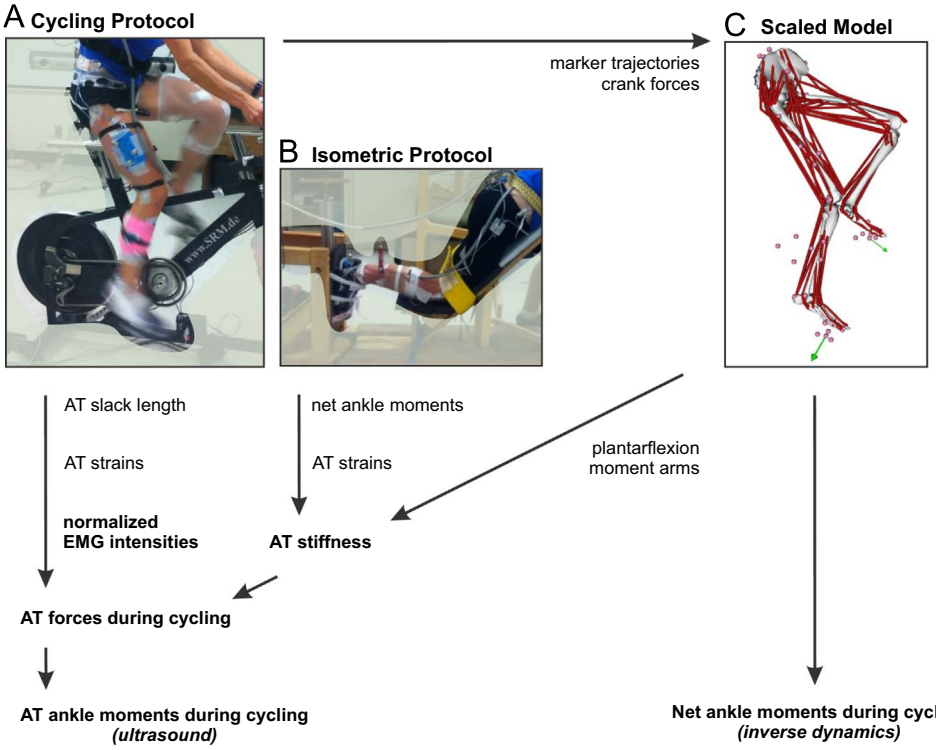

Fig. 1. Approach for estimating and evaluating *in vivo* Achilles tendon (AT) forces during cycling from tracked ultrasound images. During the cycling protocol (A), subjects pedaled on a stationary bike while we measured AT length changes, 3D marker trajectories, crank reaction forces, and surface EMG. A trigger from the ultrasound system was used to synchronize all data, and we confirmed synchronization by identifying the start of each crank cycle from markers on the left pedal. During the isometric protocol (B), subjects generated ramped plantar flexion contractions while we measured AT length changes and plantar flexion torques; these data were used to estimate AT stiffness. A musculoskeletal model (C) was scaled to each subject and was used to calculate the muscles' plantar flexion moment arms and net ankle moments during cycling. For each subject, we compared ankle moments derived from the AT forces (left) to the net ankle moments calculated *via* inverse dynamics (right). We examined whether the AT moments changed when we used an average stiffness, rather than the measured stiffness, or when we neglected to account for relative differences in the EMG intensities of the triceps surae muscles. Experimental data from a representative subject are provided in Supplementary materials (Suppl. Fig. S1).

Third, AT strains that are calculated from measured length changes depend, in part, on the assumed slack length of the composite tendon; however, determining the AT's *in vivo* length at the start of force transmission is often not straightforward.

Our procedure extends previous approaches in two ways: it characterizes AT stiffness on a subject-specific basis, and it accounts for changes in the relative electromyographic (EMG) intensities of the MG, LG, and SOL. To evaluate the procedure, we asked 10 competitive cyclists to pedal at a steady 80 rpm cadence at 4 different crank loads while we collected a comprehensive set of kinematic, kinetic, EMG, and ultrasound data. Cycling offers a unique paradigm for characterizing AT mechanics and testing new methods because loads can be varied to impose changes in the required joint moments without imposing large changes in the excursions of muscle-tendon units (MTUs). Here we estimate the forces and moments transmitted by the AT during cycling. We compared these estimates to subjects' net ankle moments calculated from inverse dynamics, and we assessed the sensitivity of these estimates to measured values of AT stiffness, AT slack length, and relative EMG intensities of MG, LG, and SOL. Our results provide important considerations for estimating AT forces from ultrasound images.

2. Methods

2.1. Acquisition of experimental data

Data were recorded from ten competitive female cyclists (age 28 ± 6 years; Suppl. Table S1) recruited from local cycling clubs. Each test session included a cycling protocol, during which subjects pedaled on a stationary bicycle (Indoor Trainer, SRM, Julich, Germany), and an isometric protocol, during which subjects generated ankle plantar flexion moments while secured in a custom frame (Figs. 1 and 2). All subjects gave informed consent, and protocols were approved by Institutional Review Boards at Simon Fraser University and Harvard University.

During the cycling protocol, we recorded ultrasound images of the MG MTJ, the 3D trajectories of 32 LED markers, reaction forces effective and ineffective (normal and radial) to the crank, and surface EMG patterns from 10 muscles (Fig. 1A and Suppl. Fig. S1). A B-mode ultrasound probe (7 MHz, 60 mm field-of-view; Echoblaster, Telemed, Vilnius, Lithuania) was secured over the distal MG MTJ on the right limb using a stretchy adhesive bandage, and an ultrasound gel pad (Parker Laboratories, NJ, USA) was placed at the probe-skin interface to enhance image quality and allow the muscles to bulge. We tracked the MTJs of both gastrocnemii in these experiments, but because the architecture of the MG is less complex than that of the LG (*e.g.*, Wolf et al., 1998), we measured length changes of the AT at the MG MTJ (Suppl. Video S1), consistent with previous studies (*e.g.*, Maganaris and Paul, 1999; Muramatsu et al., 2001). Markers were placed bilaterally over the greater trochanter, lateral epicondyle, lateral malleolus, calcaneus, and fifth metatarsal. Five markers were positioned on the pelvis, two were fixed on each pedal, and rigid marker triads were secured to the right thigh, shank, and ultrasound probe. Six "virtual" markers were defined based on the measured marker motions, segment lengths, and known bike dimensions; these markers located the subjects' hip centers (Siston and Delp, 2006), lateral epicondyles, and pedal centers of pressure. Markers were tracked at 100 Hz using an optical motion capture system (Certus Optotrak, NDI, Waterloo, Canada). Ultrasound images were recorded at 40 Hz, and prior calibration (Prager et al., 1998) determined the position of the ultrasound scanning plane relative to markers on the probe. While pedaling, subjects wore sport sandals with cleats that were secured to clipless instrumented pedals (Powerforce, Radlabor, Freiburg, Germany); the sandals had a stiff sole and allowed markers on the ankle and foot to be placed directly over bony landmarks on the skin. Reaction forces at the crank were recorded bilaterally at 2000 Hz. On the left limb, bipolar Ag/AgCl surface EMG electrodes (10 mm diameter, 21 mm spacing; Norotrode; Myotronics, Kent, USA) were placed over the mid-bellies of the MG, LG, SOL, tibialis anterior (TA), and six other muscles (not reported here). EMG signals were preamplified (gain 1000), band-pass filtered (bandwidth 10–500 Hz; Biovision, Wehrheim, Germany), and sampled at 2000 Hz as described elsewhere (*e.g.*, Wakeling and Horn, 2009; Blake and Wakeling, 2015).

Subjects were tested at 11 combinations of cadence and crank torque. For this study, we analyzed trials in which subjects pedaled at 80 rpm at crank torques of 14 (easy), 26, 35, and 44 N m (equivalent to a steep hill), corresponding to crank powers of 115, 220, 290, and 370 W. Sets of trials, each 15 s in duration, were

Please cite this article as: Dick, T.J.M., et al., Quantifying Achilles tendon force *in vivo* from ultrasound images. Journal of Biomechanics (2016), http://dx.doi.org/10.1016/j.jbiomech.2016.07.036