Journal of Biomechanics 49 (2016) 507–513

Contents lists available at ScienceDirect

Journal of Biomechanics

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

In vivo passive mechanical properties estimation of Achilles tendon using ultrasound

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ARTICLE INFO

Article history:
Accepted 21 October 2015

Keywords:
Achilles tendon
Tangent modulus
Ultrasound
Passive

ABSTRACT

A methodology is proposed for estimating Achilles tendon tangent modulus *in vivo*, to account for its large deformations and non-linear behavior. *True* stress is found dividing the axial force by the tendon *true* cross-sectional area (CSA), whose shrinking caused by axial tension is estimated with Poisson's coefficient. The *true* strain is calculated as the integral of incremental deformations along the tendon length change. *Triceps surae* tendon CSA and ankle moment arm, with the foot at relaxed equilibrium position, are estimated from subject-personalized data. Healthy males ($N=19$) volunteered for the study. The test consisted of passive ankle mobilization at the dynamometer with 5°/s velocity, from 30° of plantar flexion to the limit of dorsiflexion. Ultrasound was used to track myotendinous junction (MTJ) and tendon elongation, with the probe oriented over the medial gastrocnemius. Non-linear tendon stiffness pattern was observed during the joint range of motion, reaching 200 N/mm peaks for the subjects with greater amplitudes of maximum dorsiflexion. The maximum values of modulus of elasticity, calculated from usual engineering stress and strain, (188.56 ± 99.19 MPa) were smaller than those reported in the literature for active maximum voluntary contractions tests. Maximum values for tangent modulus from true stress and strain were 312.38 ± 171.95 MPa. Such differences are likely to increase in large deformations.

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1. Introduction

In vivo passive mechanical properties of human *triceps surae* (TS) muscle–tendon unit (MTU) are usually estimated from the passive torque–angle relationship, measured by dynamometry (Gajdosik, 2002; Magnusson, 1998). Some authors define the slope of the linear part of this curve as the MTU passive stiffness index (PSI) (Abellaneda et al., 2009; Nakamura et al., 2012, 2011). This approach has been applied to study the acute and chronic adaptations of the TS MTU mechanical properties due to different interventions, such as hypertrophy or stretching protocols (Morse et al., 2008; Nakamura et al., 2011).

Advances in imaging techniques, especially the ultrasonography (US), allowed the non-invasive and real-time study of the human MTU structure under different activation conditions (Fukunaga et al., 2001, 1997; Loram et al., 2006). Similarly, US has been used in studies that examine the muscle and tendon plasticity in response to a variety of conditions such as aging, pathology, sports and rehabilitation (Blazevich, 2006; Kawakami, 2005). Some authors improved PSI calculation by measuring the MG myotendinous junction (MTJ) displacement by US (Abellaneda et al., 2009; Kato et al., 2013, 2005; Nakamura et al., 2011; Samukawa et al., 2011).

However, in order to compare tendon condition among individuals with different tendon sizes, an intensive material property such as the Elasticity or Young Modulus (E) should be more meaningful than the overall structure stiffness. A few studies estimated Achilles tendon (AT) modulus of elasticity *in vivo*, most of them using maximal voluntary plantar flexion contractions at neutral position (0°) to load the tendon, while monitoring the MTJ displacement. The results are controversial, with E values varying from 220 to 1671 MPa, for different groups and similar methodologies (Zhao et al., 2009; Arya and Kulig, 2010). However, considering fixed AT cross-sectional area and TS moment arm (r) can provide inaccurate E estimates. In addition, undesired ankle movement is present during large plantar flexion isometric efforts (Maganaris et al., 2000; Muraoka et al., 2004).

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http://dx.doi.org/10.1016/j.jbiomech.2015.10.033
0021-9290/Published by Elsevier Ltd.

An alternative way to find AT modulus of elasticity is imposing a controlled angular displacement to the ankle without muscle contraction. This maneuver causes ankle passive reaction torque that can be measured by a dynamometer. Assuring that the muscles crossing the ankle are fully relaxed, the reaction torque is caused by stretching the musculotendon units, to which is summed the effect of compressing periarticular tissues and ligaments (Riener and Edrich, 1999). At slow movements, and at the central region of the joint range of motion (ROM), the role of passive structures (such as fascia, ligaments and joint capsule) and viscoelastic effects are hypothesized to be small (Hufschmidt, 1982). Nevertheless, at the limits of ROM, such effects can become relevant; however, in our knowledge, no quantitative data on the contribution of these structures to the ankle passive torque is available in the literature.

Small forces can cause relatively large tendon deformation close to slack length, what is known as the toe region of the force–length curve (Maganaris and Paul, 1999). On the opposite side of the same curve, large tendon deformations and non-linear mechanical behavior are normally observed in axial testing machines *in vitro* experiments (Vergari et al., 2011). Therefore, a single value of E cannot represent the overall stress–strain characteristics of the tendon. In such cases, the local derivative of the stress with respect to the strain, the *instantaneous*, or *tangent modulus* should be used (Popov, 1990). Considering a structure with a bar-like shape and positive Poisson's ratio, such as the Achilles tendon, large longitudinal deformations imply shrinkage of the cross-sectional area. The so-called *true* stress is found dividing the axial force by the *true* area, directly measured or calculated from Poisson's coefficient definition. The classical *engineering strain* definition, i.e., the deformation divided by the non-deformed length, is no longer valid for large deformations. In this case, the *natural* or *true* strain is more appropriate and is calculated as the integral of incremental strains along the specimen length change.

This study aims to propose a methodology to estimate AT tangent modulus *in vivo*. Ankle angular displacement is imposed by a dynamometer that also measures passive torque. Ultrasound image data is used to track MTJ displacement, AT insertion and cross-sectional area (CSA). This data is used to find the true stress throughout the test. Variable tendon CSA and *triceps surae* ankle moment arm are estimated from subject-specific data, as an effort to find the true stress and, finally, the tangent modulus curve.

2. Methods

2.1. Tangent modulus calculus

Tangent modulus of a mechanical structure (E_{true}) is defined as the local derivative of the axial true stress (σ_{true}) relatively to the axial true strain (ϵ_{true}) (Oomens et al., 2009):

$$E_{true} = \frac{d\sigma_{true}}{d\epsilon_{true}} \tag{1}$$

Accordingly, the engineering tangent modulus (E_{eng}) can be calculated from engineering stress (force dived by a fixed CSA) and engineering strain (deformation dived by initial non-deformed length).

Consider a small tendon axial incremental change of length $\Delta L^T = L^T - L_0^T$, where L^T is the actual and L_0^T the initial (reference) tendon length. True strain is found integrating the differential $d\epsilon_{true} = dL^T/L^T$:

$$\epsilon_{true} = \int_{L_0^T}^{L^T} d\epsilon_{true} = \ln(1+\epsilon_{eng}) \tag{2}$$

where the *engineering* strain is given by $\varepsilon_{eng} = \left(L^T - L_0^T\right)/L_0^T$.

Tendon cross-sectional true stress is given by:

$$\sigma_{true} = \frac{F}{CSA_{true}} \tag{3}$$

such that F is the tendon force and CSA_{true} the true cross-sectional area. Tendon force is found as $F = M/r(AA)$, where M is the ankle torque, r the AT moment arm with respect to ankle joint center and AA the ankle angle.

Tendon *true* cross-sectional area CSA is estimated by considering it has an elliptical shape with semi-major and minor axis a and b, being $CSA = \pi ab$. Initial (reference) tendon cross-sectional area CSA_0 is calculated directly from ultrasound measurements (see Section 2.6), and could be approximated by an ellipse with semi-axis a_0 and b_0. Actual CSA can be determined from strain measurements, considering the definition of Poisson's ratio:

$$\nu = -\frac{\varepsilon_{true}}{\varepsilon_{t-true}} \tag{4}$$

Where ε_{t-true} is the true transversal strain, associated to the cross-section shrinkage caused by tendon stretching. Similarly to the true axial strain, the true transversal strain is related with the engineering transversal strain as $\epsilon_{t-true} = \ln(1+\epsilon_{t-eng})$. Thus, eq. (4) can be expressed as (Vergari et al., 2011):

$$\nu = -\frac{\ln(1+\epsilon_{t-eng})}{\ln(1+\epsilon_{eng})} \tag{5}$$

$$1+\varepsilon_{t-eng} = (1+\varepsilon_{eng})^{-\nu} \tag{6}$$

Tendon Poisson's ratio $\nu = 0.55$ was adopted (Vergari et al., 2011). Let dx be the dimension of an actual (deformed) element in the same plane of the cross-sectional area, with initial dimension dx_0. Engineering transversal strain is expressed as $\epsilon_{t-eng} = (dx - dx_0)/dx_0$. Substituting this term in eq. (6), the actual elementary element dimension is:

$$dx = dx_0\left(1+\varepsilon_{eng}\right)^{-\nu} \tag{7}$$

and the ellipse semi-axis $a = a_0(1+\varepsilon_{eng})^{-\nu}$ and $b = b_0(1+\varepsilon_{eng})^{-\nu}$. True cross-sectional area can be calculated by the formula:

$$CSA_{true} = CSA_0\left(1+\epsilon_{eng}\right)^{-2\nu} \tag{8}$$

If the relationship $\sigma_{true} \times \epsilon_{true}$ can be modeled though a nth order polynomial,

$$\sigma_{true} = a_1\varepsilon_{true}^n + a_2\varepsilon_{true}^{n-1} + \ldots + a_n\varepsilon_{true} + a_{n+1} \tag{9}$$

tangent modulus follows from eq. (1),

$$E_{true} = na_1\varepsilon_{true}^{n-1} + (n-1)a_2\varepsilon_{true}^{n-2} + \ldots + 2a_{n-1}\varepsilon_{true} + a_n \tag{10}$$

2.2. Tendon deformation

To solve the equations derived in the previous section, L^T and L_0^T should be determined. Musculotendon length is the sum tendon and muscle lengths ($L^{MT} = L^T + L^M$). L^{MT} can be found from regression curves (Grieve et al., 1978). To find muscle initial and actual lengths, the following procedure has been used.

a) The foot was fastened to the dynamometer at 30° of plantar flexion position ($PF_{initial}$). The ultrasound probe was held by the researcher such that the MTJ was visualized approximately in the center of the image frame. The horizontal distance from the MTJ to the right border of the image frame was measured ($\Delta MTJ_{initial}$), as shown in Fig. 1. More details about the experimental procedure are described in Sections 2.4– 2.6;
b) The ankle was moved by the dynamometer from $PF_{initial}$ to maximal dorsiflexion at a constant angular velocity of 5°/s, without moving the US probe. Since MTJ moved to the right, the distance from MTJ to the right border of the image (ΔMTJ) decreased, and was measured at intermediate joint angles with 5° steps;

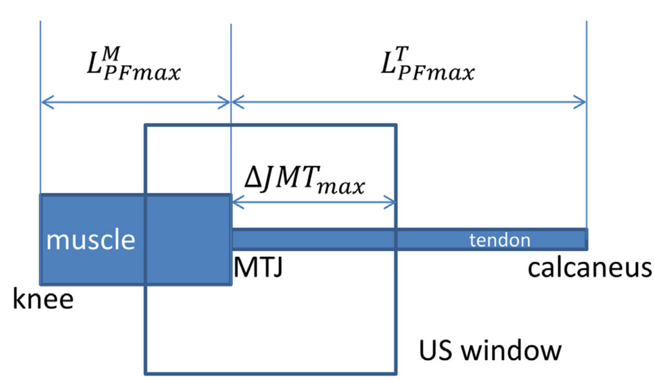

Fig. 1. Diagram showing how the initial tendon reference displacement (ΔJMT_{max}) is measured, from the myotendinous junction (MTJ) to the border of the ultrasound window.