MEASUREMENT OF THE SERIES ELASTIC STIFFNESS IN ANKLE DORSIFLEXORS

Mark de Zee[1], Michael Voigt[1,2] and Thomas Sinkjær[1]
1) Center for Sensory-Motor Interaction, Aalborg University, Denmark
2) Institute of Exercise and Sports Science, University of Copenhagen, Denmark
E-mail: mdz@miba.auc.dk

Abstract – *Reliable measurements have to be made of the series elastic stiffness (SES) in vivo in humans in order to evaluate the functional significance of SES on the control and economy of movement. We measured the SES in the human ankle dorsiflexors using the quick release method with a custom designed high-pressure hydraulic actuator in nine subjects. The subjects produced isometric dorsiflexion torques before the release which had an angular velocity of 15 rad.s^{-1}. The measured SES was 294.8±81.0 Nm.rad^{-1} (mean ± S.D.). With the present method it was possible to obtain the total stiffness of all elastic tissue in series with the contractile part of the dorsiflexors. The results showed that the SES of the human dorsiflexors are likely to work within the toe region of the stress-strain curve in vivo, which is in line with literature. Due to the low in vivo stresses in the dorsiflexors the possibility of storing elastic energy is not very important in these muscles.*

Keywords: Series elastic stiffness, Quick release, Dorsiflexors, Elastic energy.

1. Introduction

The series elastic stiffness (SES) of active muscles has a significant influence on both the control and the economy of movement. For making a quantitative evaluation of the functional significance of SES during human movement, reliable quantitative measures of SES in vivo in humans have to be made.

The SES is located in all the elastic tissue in series with the contractile part of the muscle. The SES is mainly located in the tendon and aponeurosis. Therefore, it is not enough to measure only tendon stiffness like it has been done in vitro (1;13) and in vivo (9), because there are indications of a difference between tendon and aponeurosis stiffness (10), although Scott and Loeb (12) found no difference in the stiffness of the cat soleus tendon and aponeurosis.

Ultrasonography is a recent method to study in vivo muscle architecture, tendon and aponeurosis stiffness in static and dynamic situations (3;4;9;11), but is limited to one superficial muscle at the time and measures only on one end of the muscles (normally the distal end). During natural movement, however, joint torque is generated by more than one muscle. Hence, more elastic tissue lying parallel is involved in the movement. For example a dorsiflexion torque in natural circumstances is not solely generated by the superficial m. tibialis anterior, but also by the deeper m. extensor digitorum longus and the m. extensor hallucis longus.

To get a functional measure of SES during a certain task/movement we need to incorporate all elastic tissue in series with the contractile part of the different muscles active. A possible method is the quick release method. In this method the muscle is shortened at a high but constant speed. The movement should be completed before the first reflexes arrive and the speed should be above the maximum shortening speed of the muscle. By recording the decline in torque as a function of shortening the series elastic component (SEC) can be measured and expressed in Nm.rad^{-1}. Hof (6) used the quick release method to measure the series elasticity release curve of the human triceps surae muscle using an angular velocity of 14 rad.s^{-1}.

The goal of this paper is to quantify the SES in the human ankle dorsiflexors using the quick release method with a custom designed high-pressure hydraulic actuator.

2. Methods

A PID controlled hydraulic actuator (MTS-systems Corporation 215.35) was used for the quick release experiments (14). The foot of the subject was firmly strapped to a custom-made foot adapter with large cable ties. The position of the foot adapter was adjusted such that the anatomical axis of the ankle coincided as well as possible with the center of rotation of the actuator. The releases were imposed between 20 degrees plantar-flexion and 10 degrees dorsi-flexion with an angular velocity of 15 rad.s^{-1}. The foot adapter was instrumented with load cells (Kistler, Slimline) and accelerometers (Kistler, K-SHEAR Piezotron). The position of the rotary actuator was monitored with an angular displacement transducer (Transtek DC ADT series 600). Surface electromyograms were recorded from the tibialis anterior and soleus muscles. The signals were sampled at 4000 Hz.

The torque signal had to be corrected for inertia and passive stiffness of the antagonists. The correction method used is a modified version of the method applied by Hof (5). The correction for inertia was based on the recording of the angular acceleration plus two additional recordings. The first extra recording was a slow release with an angular velocity of 0.1 rad.s^{-1} with the muscle passive, which gave the passive moment-angle curve. The second extra recording was a fast release (15 rad.s^{-1}) also with the muscle passive. From this recording a transfer function H was obtained with the angular acceleration as input and the torque as output. The recorded data was first corrected for inertia by using the acceleration signal. Subsequently the passive moment was subtracted from the total moment.

Three females and six males (see for subject data table 1) gave their informed consent to participate in the experiments. They were sitting in a chair with the knees slightly bent. After the extra recordings necessary for correction, twenty quick releases were performed. Before each release the subject produced a steady level of dorsiflexion torque. This initial torque was varied between 10 and 90 % MVC.

Values in the text are presented as mean ±S.D.

Subject	Sex	Age [yr]	Height [cm]	Body mass [kg]	$M_{0,max}$ [Nm]	SES [Nm/rad]
1	f	43	162	53	42	140
2	f	29	174	60	46	225
3	f	31	169	87	59	308
4	m	44	187	86	74	363
5	m	34	173	102	80	256
6	m	33	167	60	70	273
7	m	35	180	88	89	383
8	m	25	198	73	74	319
9	m	30	181	70	64	387
Mean		33.8	176.8	75.4	66.5	294.8
S.D.		6.3	11.1	16.3	15.3	81.0

Table 1: Subject data. $M_{0,max}$ is the maximum isometric dorsiflexion moment, recorded just before the release.

3. Results

The subjects produced a maximum dorsiflexion joint moment $M_{0,max}$ just before the release of 66.5±15.3 Nm. Figure 1A presents 20 releases of subject 3, which are corrected for inertia and passive stiffness, with different initial torques. Since the individual releases start at different initial torque levels, but at the same joint angle, the torque and angle are shifted between the measurements. Therefore the submaximal release curves were angle shifted so the initial torque level corresponded to the torque level on the trial with the highest ($M_{0,max}$) initial torque (see figure 2). The results of the shift are presented in figure 1B. Finally, an average of the curves in panel 1B is presented in 1C.

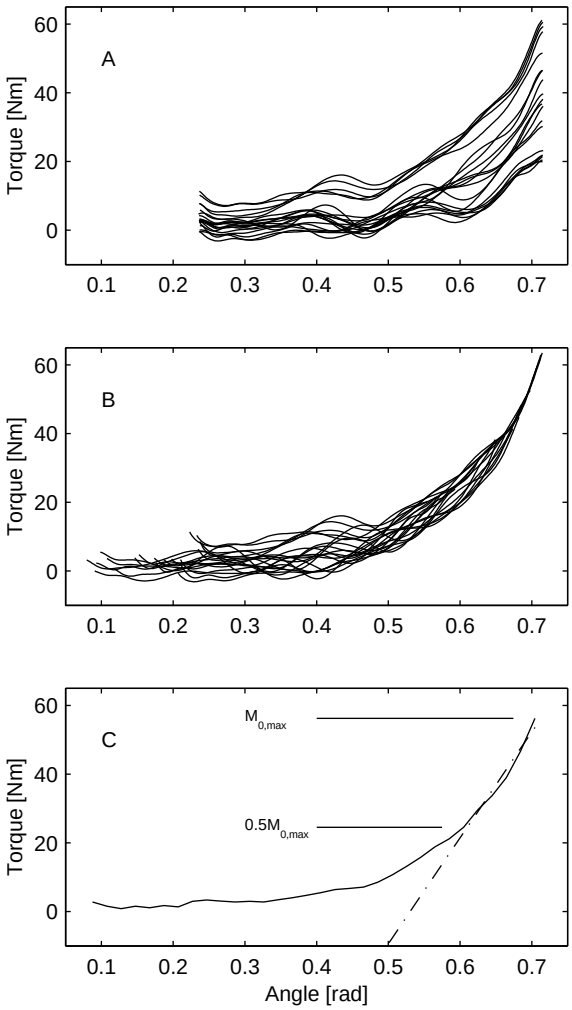

Fig. 1: A) Twenty release curves of subject 3 with different initial torques. B) The release curves in A are shifted to the left in such a way that they coincide with the release curve with the highest initial torque. C) An average of the curves in B. A linear fit was made between the two horizontal lines.

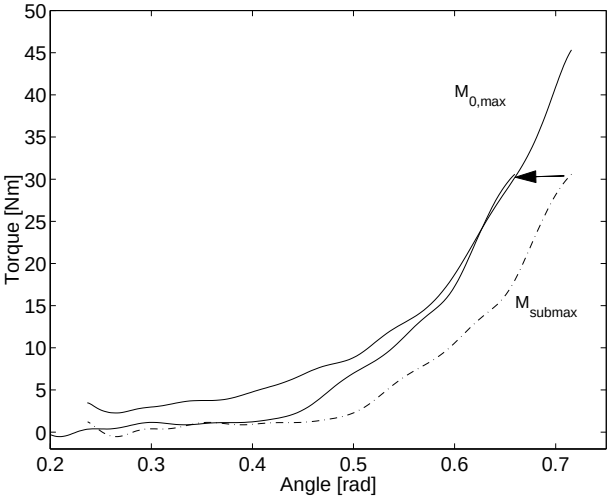

Fig. 2: Shift of a submaximal release curve to the left so the curve coincides with the $M_{0,max}$ curve (example of subject 2).

The SES of every subject was calculated as the slope of the curve in 1C between 50 % and 100 % of the maximum torque $M_{0,max}$ of that particular subject (see figure 1C). The calculated SES was 294.8±81.0 $Nm.rad^{-1}$. See for individual values table 1.

Discussion

In the present study the series elastic stiffness (SES) of the human dorsiflexors was obtained in vivo by measuring the torque and angle during a quick release. The release was controlled by a custom designed high-pressure actuator.

With the quick release method in the present study it was possible to obtain the total stiffness of all elastic tissue in series with the contractile part of the dorsiflexors. The method can not be used to obtain tendon stiffness and aponeurosis stiffness separately. The advantage, however, is that the quick release method quantifies the functional SES during a certain task/movement including the deeper elastic tissues. That is useful for discussing the storing and releasing of elastic energy during human movement. In principal it is possible to measure SES of the knee flexors and/or extensors, but the inertial effects will be much higher.

The shift of the curves to the left as showed in figure 1B is only justified if the amount of elastic tissue involved is the same for the different levels of initial torques. The aponeurosis is already at low forces totally involved, because of the lateral shear transmission of force (7) and the muscle fibres active at low forces are not concentrated in one part of the muscle but are distributed (2). We see that the release curves are curvinlinear, which indicates that during maximum isometric dorsiflexions the serial elastic component (SEC) is still operating in the toe region. This is in line with the study by Maganaris and Paul (10), where they concluded the same for the m. tibialis anterior tendon and aponeurosis. Also the study by Ker et al. (8) indicates that the human dorsiflexor tendons operates well below 30 MPa within the toe region, which explains the high safety factor (around 4) for these tendons.

The increase in stiffness with torque is to our conviction due to the fact that the SEC is operating in the toe region over the range of torques examined. Hof (6) noticed also an increase in stiffness with torque in his quick release experiments on the human triceps surae. His explanation, however, differs from ours. Hof proposed that with a higher torque more aponeurosis strands parallel with each other is engaged and consequently that leads to an increase of stiffness. This proposal does not take into account for any lateral shear transmission of force. It might be that also the SEC of the human triceps surae works in the toe region with isometric torques up to voluntary maximum.

Although the SEC of the dorsiflexors are working in the non-linear toe region we made a linear fit of the curve as in figure 1C between 50 % and 100 % of $M_{0,max}$. In that part the non-linearity is small and the obtained SES values can be compared with the K1 values obtained by Hof (6) (Table 2, page 797). There is a large variation between subjects in both studies. Our SES values range from 140 to 387 $Nm.rad^{-1}$. The K1 values in Hof's study range from 258 to 405 $Nm.rad^{-1}$. Note that Hof calculated the K1 values at a fixed torque of 100 Nm, while our SES value is an average over the torque levels from 50 % to 100 % of $M_{0,max}$. Most probably the K1 value would be higher for a maximal isometric contraction in the triceps surae. These preliminary results indicate that the SES of the dorsiflexors is lower than the SES of the plantarflexors under maximum isometric contractions. This is not surprising having in mind that the cross-sectional area of the tendons of the plantarflexors is 39 % larger than the cross-sectional area of the dorsiflexors while there is only a slight difference in length and moment arm (15).

The range of forces experienced in vivo is very different between dorsiflexors and plantarflexors. The SEC in the plantarflexors experience high excentric forces during natural movements like running, which will increase the stiffness of the SES and makes it possible to store a high amount of elastic energy. The SEC in the dorsiflexors will rarely experience forces higher than the maximal voluntary contraction. Hence, the possibility of storing elastic energy in the dorsiflexors is not very important.

Acknowledgements

The Danish National Research Foundation and The Danish Foundation of Physical Disability (Vanførefonden) supported this work.

References

1. Benedict, J.V., L.B. Walker, and E.H. Harris. Stress-strain characteristics and tensile strength of unembalmed human tendon. *J.Biomech.* 1: 53-63, 1968.
2. Burke, R.E., D.N. Levine, M. Salcman, and P. Tsairis. Motor units in cat soleus muscle: physiological, histochemical and morphological characteristics. *J.Physiol.(Lond.)* 238: 503-514, 1974.
3. Fukashiro, S., M. Itoh, Y. Ichinose, Y. Kawakami, and T. Fukunaga. Ultrasonography gives directly but noninvasively elastic characteristic of human tendon in vivo. *Eur.J.Appl.Physiol.* 71: 555-557, 1995.
4. Fukunaga, T., M. Ito, Y. Ichinose, S. Kuno, Y. Kawakami, and S. Fukashiro. Tendinous movement of a human muscle during voluntary contractions determined by real-time ultrasonography. *J.Appl.Physiol.* 81: 1430-1433, 1996.
5. Hof, A.L. Correcting for limb inertia and compliance in fast ergometers. *J.Biomech.* 30: 295-297, 1997.

6. Hof, A.L. In vivo measurement of the series elasticity release curve of human triceps surae muscle. *J.Biomech.* 31: 793-800, 1998.
7. Huijing, P.A., G.C. Baan, and G.T. Rebel. Non-myotendinous force transmission in rat extensor digitorum longus muscle. *J.Exp.Biol.* 201: 683-691, 1998.
8. Ker, R.F., R.M. Alexander, and M.B. Bennet. Why are mammalian tendons so thick? *J.Zool, Lond.* 216: 309-324, 1981.
9. Maganaris, C.N. and J.P. Paul. In vivo human tendon mechanical properties. *J.Physiol.(Lond.)* 521: 307-313, 1999.
10. Maganaris, C.N. and J.P. Paul. Load-elongation characteristics of in vivo human tendon and aponeurosis. *J.Exp.Biol.* 203: 751-756, 2000.
11. Narici, M. Human skeletal muscle architecture studied in vivo by non-invasive imaging techniques: functional significance and applications. *J.Electromyogr.Kinesiol.* 9: 97-103, 1999.
12. Scott, S.H. and G.E. Loeb. Mechanical properties of aponeurosis and tendon of the cat soleus muscle during whole-muscle isometric contractions. *J.Morphol.* 224: 73-86, 1995.
13. van Brocklin, J.D. and D.G. Ellis. A study of the mechanical behaviour of toe extensors tendons under applied stress. *Arch.Phys.Med.Rehabil.* 46: 369-373, 1965.
14. Voigt, M., M. de Zee, and T. Sinkjaer. A fast servo-controlled hydraulic device for the study of muscle mechanical and reflex properties in humans. *International Society of Biomechanics XVIIth Congress, Calgary, Canada* 1999.(Abstract)
15. Yamaguchi, G.T., A.G.U. Sawa, D.W. Moran, M.J. Fessler, and J.M. Winters. A survey of human musculotendon actuator parameters. In: *Multiple muscle systems,* edited by J.M. Winters and S.L. Woo. New York: Springer-Verlag, 1990, p. 717-773.