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Structural and material properties of human foot tendons

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ABSTRACT

Backgrounds: The aim of this study was to assess the mechanical properties of the main balance tendons of the human foot *in vitro* reporting mechanical structural properties and mechanical material properties separately. Tendon structural properties are relevant for clinical applications, for example in orthopedic surgery to elect suitable replacements. Tendon material properties are important for engineering applications such as the development of refined constitutive models for computational simulation or in the design of synthetic materials.
Methods: One hundred uniaxial tensile tests were performed to obtain the mechanical response of the main intrinsic and extrinsic human foot tendons. The specimens were harvested from five frozen cadaver feet including: Extensor and Flexor tendons of all toes, Tibialis Anterior and Posterior tendons and Peroneus Brevis and Longus tendons.
Findings: Cross-sectional area, load and strain failure, Young's modulus and ultimate tensile stress are reported as a reference of foot tendon mechanical properties. Two different behaviors could be differentiated. Tibialis and Peroneus tendons exhibited higher values of strain failure compared to Flexor and Extensor tendons which had higher Young's modulus and ultimate tensile stress. Stress–strain tendon curves exhibited proportionality between regions. The initial strain, the toe region and the yield point corresponded to the 15, 30 and 70% of the strain failure respectively.
Interpretation: Mechanical properties of the lesser-studied human foot tendons are presented under the same test protocol for different engineering and clinical applications. The tendons that work at the inversion/eversion plane are more deformable at the same stress and strain rate than those that work at the flexion/extension plane.

1. Introduction

Tendon is a white fibrous tissue that connects the extremes of the muscles to the bones. Its biomechanical role is to transmit the contraction of the muscles to the skeleton in order to produce force or movement of the body. This tissue is evolutionary mechanoadapted to work axially, analogous to a string, with a large length compared to its section.

Mechanical properties of tendons have been studied previously, particularly the Achilles tendon. It is one of the biggest tendon of the human body and very relevant clinically due to its high incidence of injury. Furthermore, its location and structure facilitate the measures *in vivo*. There is an extensive bibliography about this tendon which in certain situations is extrapolated to estimate the properties of other tendons, as in the case of other foot tendons where the information available is scarce and incomplete (Sharkey and Hamel 1998; Thordarson et al. 1995).

The material properties reported for tendons have a great variability. For example, the Young's modulus varies in an order of magnitude from 0.2 to 2 GPa (Ker 2007; Maganaris et al., 2008; Wang 2006). There are two main reasons for this disparity of results: one is the natural biological variation and the other is the different procedures used to assess the properties. To reduce the influence of the first factor some authors prefer to test animal specimens where the history of the subject can be controlled, although for clinical applications human material is frequently required. The second factor could be compensated applying the same methodology to calculate the properties, but there is no agreement about a proper method to evaluate mechanical tendon properties yet. Furthermore, different methodologies are needed depending on the objective pursued. For instance, in the field of simulation, the

Abbreviations: CSA, cross-sectional area; EDB, extensor digitorum brevis; EDL, extensor digitorum longus; EHL, extensor hallucis longus; FDB, flexor digitorum brevis; FDL, flexor digitorum longus; FHL, flexor hallucis longus; PB, peroneus brevis; PL, peroneus longus; TA, tibialis anterior; TP, tibiales posterior.

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characterization of a tissue can be approximated by linear, hyperelastic or viscoelastic models, which requires different mechanical parameters. The positive aspect of the use of different techniques to assess the mechanical properties is that it prevents bias.

The mechanical properties of human foot tendons are valuable information in different fields. In orthopedic surgery, the use of tendon grafts is common to repair tendons and ligaments (Giannini et al. 2008; Sebastian et al. 2007; Zhao and Huangfu 2012). Among other characteristics, the mechanical structural properties of the potential graft are one of the prerequisites that surgeons evaluate in the election of a suitable replacement. Detailed information of structural properties of every foot tendon would help surgeons in the decision making process. Computational biomechanics is another field in which experimental data of actual behavior of human foot tendons would provide a significant advance. From the engineering perspective, the human foot is a complex structure of small bones supported by strong ligaments and controlled by a network of tendons and muscles. Considering that the current barrier in foot computational simulation is the inclusion of these musculotendinous structures in the models (Morales-Orcajo et al., 2015), a detailed description of their material properties will help in the definition and adjustment of foot tendon material models.

The purpose of this study is to assess the mechanical properties of the human foot tendons responsible for the stabilization of the ankle joint and control motion of toes. One hundred samples of these lesser-studied foot tendons were tested *in vitro*. Particular effort was made to proportionate a refined description of their hyperelastic feature. As outcome, a dataset of experimental values for engineering and clinical applications is provided.

2. Methods

2.1. Tendon specimens

A total of one hundred tendons samples were taken from five male elder donors, with the approval of the ethical committee of clinical research of the Hospital Clínico San Carlos in Madrid. A sample of each tendon was cut from the most relative uniform cross-sectional area (CSA) removing all the soft tissue around the tendon. After the dissection, the samples were frozen and kept at a temperature of −20 °C (Devkota and Weinhold 2003; Schechtman and Bader 1997; Sebastian et al. 2007; Zhao and Huangfu 2012) until the day of testing (8–12 months) (Vergari et al. 2011).

The tendons included in the experiments were sorted in two groups: the long tendons involved in flexion and extension of the toes, on the one hand and the thick tendons intervening on the inversion and eversion of the ankle, on the other hand. The former includes the Extensor Digitorum Brevis (EDB) and the Extensor Digitorum Longus (EDL) which extent lesser toes, the Extensor Hallucis Longus (EHL) which extends the great toe, the Flexor Digitorum Brevis (FDB) and the Flexor Digitorum Longus (FDL) which flex the four lateral toes and the Flexor Hallucis Longus (FHL) which flexes the hallux. The latter involves the Tibialis Anterior (TA) and the Tibialis Posterior (TP) which invert the foot, and the Peroneus Brevis (PB) and Peroneus Longus (PL) which evert the foot (Fig. 1). These tendons enable us to stay balanced in upright position.

2.2. Testing procedure

Specimens were gradually thawed and kept hydrated until the time of testing at room temperature (~25 °C). The CSA was measured right before testing taking the average of three measures along the longitudinal axis of the sample. The maximal and minimal diameters of the tendon were measured with a digital caliper to calculate CSA by approximating it as an ellipse (Giannini et al. 2008; Vergari et al. 2010). A pair of screw lock clamps was specifically designed to perform the tests. The inner sides of the stainless steel clamps were milled with small holes to improve the grip. No cycle of tissue preconditioning was applied to the samples (Butler et al. 1984; Giannini et al. 2008; Schechtman and Bader 1997; Zhao and Huangfu 2012).

A universal testing machine (Instron Ltd., U.K., model 5548) was used to perform the tests (Fig. 2). An initial stretch of 1 MPa was applied to remove any slack in the samples. Then, a displacement was applied at a rate of 0.1 mm/s to failure. This rate corresponds with approximate 10–20% s^{-1} depending on length sample. Strain was measured using clamp-to-clamp displacement. The trials with evidences of slipping or initial damage were discarded.

2.3. Tendon stress–strain curve

The stress–strain curve is the normalized curve of the load–displacement graph provided by the test machine. This curve represents the main material parameters. The x coordinate indicates the strain failure, the y coordinate indicates the ultimate tensile stress and the Young's modulus is the slope of the linear region of the curve. Stress (σ) is

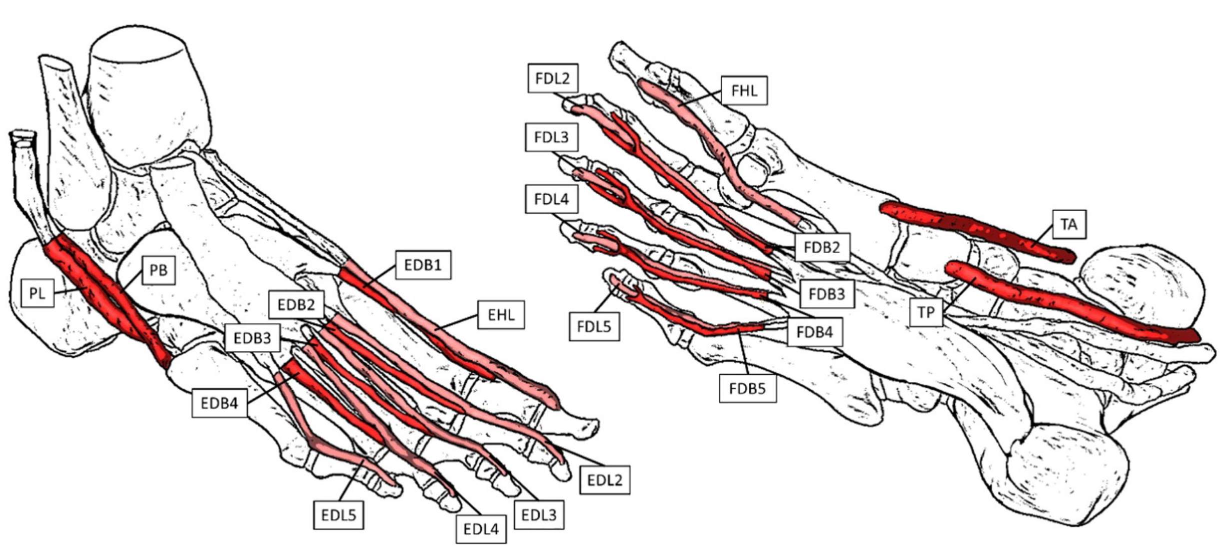

Fig. 1. Schematic view of the anatomical position of the specimens tested.

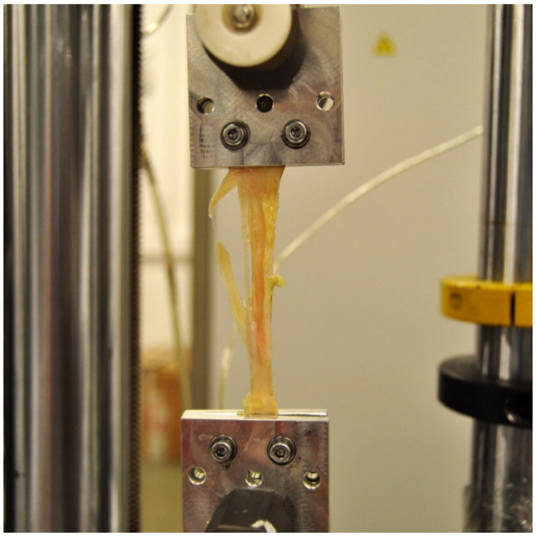

Fig. 2. Peroneus Brevis sample in the universal testing machine during test.

calculated dividing the load (F) by the initial CSA (engineering stress) $\sigma = F/CSA_0$ and the strain ($\boldsymbol{\varepsilon}$) is calculated dividing the displacement (δ) by the initial length (L_0) of the sample $\varepsilon = \delta/L_0$. Three distinct regions are identified: the toe region characterized by a non-linear strain with small tension, the linear region where the stress increases linearly with the strain and the yield region where macroscopic failure appears. The toe region, in turn, is possible to divide in two parts: the noteworthy initial strain and the non-linear transition previous to the linear region (Fig. 3).

A criterion to determine these regions never has been stipulated and it is subjected to the visual perception of the slope of the curve. This demarcation although is qualitatively worthy could be misleading in quantifying accurately the actual range of each region. Therefore, we propose a criterion to quantify those regions taking as reference the Young's modulus of each test. Hence, the initial strain is defined as the part of the curve from its beginning to a deviation greater than 20% of the Young's modulus (Point A Fig. 3); and the linear region is defined as the part of the curve with the curvature less than 20% of the Young's modulus, thus delimiting toe and yield region as well (Point B and C

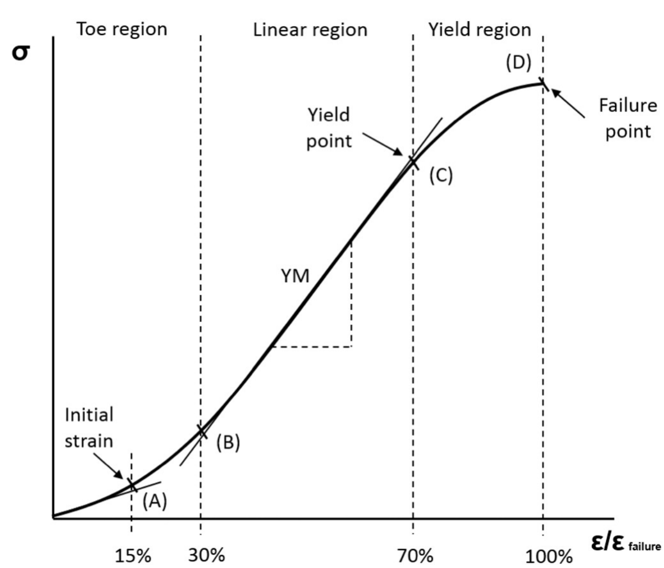

Fig. 3. *Scheme of typical stress–strain curve for tendon tissue normalized to strain failure, identifying regions and proportionality ratios. YM - Young's modulus.*

Fig. 3). Visually, the resulting regions fit with the description in the literature (Abrahams 1967).

2.4. Statistical analysis

The variables of interest for the evaluation of the mechanical response of foot tendons that work at different planes were the ultimate tensile stress and the Young's modulus. Differences were determined using two tailed impaired t-tests and the significant level was set at $p < 0.05$.

3. Results

3.1. Tendon mechanical structural properties

In Table 1 are summarized the structural properties of each foot tendon. Ranges are reported together with mean and standard deviation in order to facilitate comparison with other works. Failure load values are the maximum load recorded by the load cell of the testing machine. After that value, load decreased drastically. Note that values reported of extensor and flexor digitorum correspond to the toe-branches of the tendons (Fig. 1). Hence, to estimate the failure load of the whole tendon, the maximal load measured in these experiments should be multiplied by the number of branches of the tendon.

Due to the length of the samples tested do not correspond with the total physiological length of the tendons, structural properties of the tendon such as stiffness and energy to failure cannot be reported.

3.2. Tendon stress–strain curve

A total of six trials were discarded. The rest showed the characteristic stress–strain curve of tendon tissue (Fig. 3). Using the criterion defined in the Section 2.3 the regions of the each stress–strain curve were accurately quantify. In Table 2 are presented the mean strain values of the four points that delimit the regions. Inversion/eversion tendons showed strain values nearly twice that of flexor/extensor toes tendons for each region.

Based on these values, strain proportionality within regions was detected when normalized to strain failure, understanding failure point as the point of the curve that supports the maximum stress. The initial strain, the toe region and the yield point corresponded to 15, 30 and 70% of the strain failure respectively. These delimiting points and the region proportionality are shown schematically in Fig. 3.

3.3. Tendon mechanical material properties

In Fig. 4, the summary of all trial curves is depicted. The points represent the failure point of the stress–strain curve of each sample tested. They are connected to the origin with the characteristic S-shaped

Table 1
Structural properties of human foot tendons.

Tendon	Cross-sectional Area [mm^2]		Failure load [N]	
	Mean (SD)	(Range)	Mean (SD)	(Range)
EDB[1]	3.09 (SD 0.95)	(1.53–4.56)	37 (SD 14)	(12–62)
EDL[1]	4.81 (SD 1.25)	(2.83–7.34)	127 (SD 84)	(52–237)
FDB[1]	2.72 (SD 1.02)	(1.48–5.18)	65 (SD 48)	(13–179)
FDL[1]	4.87 (SD 2.12)	(1.54–9.03)	74 (SD 30)	(37–138)
EHL	8.53 (SD 1.35)	(6.91–10.41)	316 (SD 88)	(247–443)
FHL	15.77 (SD 2.64)	(12.53–18.69)	525 (SD 200)	(356–770)
TA	26.40 (SD 2.85)	(23.75–29.85)	458 (SD 155)	(306–669)
TP	24.02 (SD 3.69)	(20.29–33.18)	475 (SD 60)	(397–528)
PB	11.87 (SD 2.01)	(10.13–15.22)	239 (SD 99)	(138–392)
PL	16.59 (SD 3.38)	(13.22–21.17)	346 (SD 196)	(130–629)

[1] These measurements correspond to the toe-branches of distal part of the tendon and not to the common proximal part. SD — Standard deviation.

Table 2
Strain values of the point that delimit each region of the stress–strain curve.

		EDB	EDL	FDB	FDL	Average
Initial strain (%)	(A)	1.0	1.4	1.4	1.6	1.4
Toe region (%)	(B)	1.4	2.9	2.5	1.9	2.2
Yield point (%)	(C)	4.3	6.2	5.2	5.0	5.2
Failure point (%)	(D)	6.0	8.7	7.1	7.2	7.2
		TA	TP	PB	PL	Average
Initial strain (%)	(A)	2.5	3.6	1.5	2.1	2.4
Toe region (%)	(B)	6.4	6.4	2.6	4.5	5.0
Yield point (%)	(C)	12.1	12.9	9.2	9.5	10.9
Failure point (%)	(D)	17.0	16.3	14.2	13.7	15.3
		EHL		FHL		Average
Initial strain (%)	(A)	1.6		2.0		1.8
Toe region (%)	(B)	3.9		4.3		4.1
Yield point (%)	(C)	9.8		8.8		9.3
Failure point (%)	(D)	11.7		12.3		12.0

tendon curve although most of the curves have been removed in order to clarify the visualization. Strain failures are summarized for each tendon in Table 2. Young's modulus and ultimate tensile stress of each foot tendon is reported in Table 3.

Two different behaviors were distinguished with a significant difference of $p < 0.001$. On the one hand, flexor/extensor tendons were characterized by lower strain failure and higher Young's modulus. The median curve of these tendons is graphed with a black solid line (Fig. 4). On the other hand, inversion/eversion tendons presented larger strain limits and lower Young's modulus. The median stress–strain curve of these tendons is graphed with a red solid line (Fig. 4). The flexor/extensor tendons presented an average Young's modulus of 390 MPa (SD 175) against 195 MPa (SD 83) for inversion/eversion tendons.

There was no significant difference in Young's modulus and ultimate tensile stress between flexor and extensor tendons neither interdonor variability.

4. Discussion

Mechanical properties of human tissues are relevant for many practical issues. Frequently, the measurement of these properties is concentrated in particular components with high clinical interest

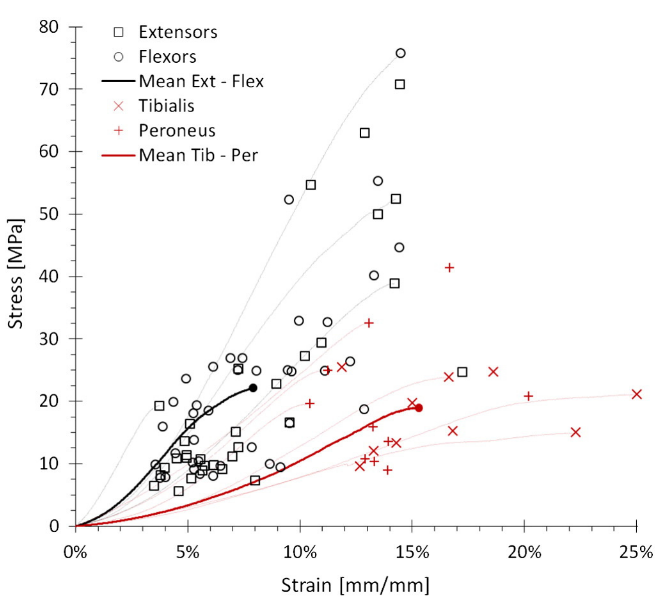

Fig. 4. Stress–Strain graph including all samples tested sorted by muscle function.

Table 3
Material properties sorted by tendon.

Tendon	Young's Modulus [MPa] Mean (SD)	Ultimate tensile stress [MPa] Mean (SD)
EDB	294 (SD 137)	14 (SD 9)
EDL	395 (SD 180)	26 (SD 21)
FDB	505 (SD 172)	28 (SD 19)
FDL	337 (SD 138)	16 (SD 7)
EHL	448 (SD 183)	39 (SD 18)
FHL	440 (SD 119)	26 (SD 10)
TA	165 (SD 73)	17 (SD 6)
TP	187 (SD 54)	19 (SD 5)
PB	203 (SD 94)	20 (SD 10)
PL	227 (SD 116)	20 (SD 12)

SD – Standard deviation.

disregarding others less critical. This leads to the estimation of the properties of those other components by extrapolation. That is the case of tendon tissue where most of the studies measure the properties of the Achilles tendon, the patellar tendon or the supraspinatus tendon. In the present study, the tensile response of the lesser-studied foot tendons was tested *in vitro*. Structural and material properties are reported as reference under the same test protocol. Moreover, a criterion for quantitative description of the stress–strain curve has been defined for the first time.

The major finding of the experiments was the significant difference of the mechanical response between tendons which work at different planes. Flexor/extensor toe tendons showed twice the Young's modulus than inversion/eversion tendons. In practical terms, this means that tibialis and peroneus tendons undergo higher deformation for the same tensile stress. In computational simulation for linear models, the Young's modulus is the main parameter to describe the mechanical behavior of the material. In the case of tendon tissue the description of the fully mechanical behavior by a linear model is simplistic. Tendon tissue performs a hyperelastic behavior, thus, the typical material parameters of Young's modulus, ultimate tensile stress and strain failure are not sufficient. The quantitative description of the stress–strain curve provides a more refined characterization.

A qualitative description of the stress–strain curve has been defined previously (Schechtman and Bader 1997; Spyrou and Aravas 2011; Wang 2006). Generally three regions are identified: toe region, linear region and yield region. These regions are described based on the visual observation of a representative typical stress–strain curve, but do not consider the wide variability of each specimen. When we analyzed the result of the present work we were conscious of that description do not well-fit our curves and the delimitation of the regions were subjective to the perception of the analyst. In order to standardize this division the criterion described in Section 3.2 is proposed. Furthermore, we found important to include another subdivision to characterize the transition between the initial strain and the beginning of the linear region. This region was also identified by Abrahams (1967). The histological studies describe this second region at the stage where the collagen fibers reach the final parallel orientation that governs the linear behavior after the stretch out of the wave-pattern that occurs initially Abrahams (1967).

During testing process some experimental difficulties were found. First of all, the tendons tested are considerably small compared with the rest of the tendons in the human body. Measure the CSA of these small tendons was critical because of the sensibility of the measurement. Little variations in the CSA had high impact in the calculus of the material properties. Caliper was chosen because of its ease of use and its sufficient level of accuracy and repeatability. This method is acceptable to measure tendons with CSA greater than 1.5 mm^2 and constant along the longitudinal axis. Lumbrical tendon tests were discarded for their tiny CSA; with diameters of 0.9 mm and a precision of 0.1 mm the CSA of the lumbricals varied up to 20%. Although the caliper have higher accuracy, the sensitivity of the human eye to detect

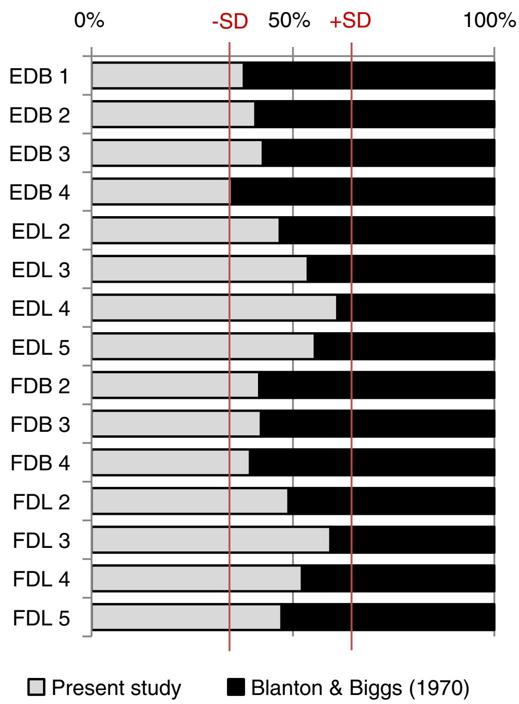

Fig. 5. Comparison of the CSA measured in the present study with the CSA reported by Blanton and Biggs (1970).

contact between the caliper and the soft tissue is not better than 0.1 mm. Vergari et al. (2010) found comparable results of the elliptical approximation against more sophisticated techniques. In the literature, only one study that report CSA of these smaller tendons were found. Blanton and Biggs (1970) used a planimeter over a thirty magnifications projected area to determine tendon CSA and the values reported were similar (Fig. 5)

Clamping the samples at the testing machine is also a common difficulty testing soft tissues. The firmness with which the sample is fastened has to balance the sufficient strength to avoid the slip out of the sample, but not too much to brake collagen fibrils. Similar clamping systems, such as the used in this study, have been employed in previous uniaxial tensile tests of tendon tissue (Abrahams 1967; Schechtman and Bader 1997). Due to the impact of the clamping system, stress concentrations were expected in the samples. Therefore, in physiological condition, values of tendon failure may be underestimated (Ker et al. 2000; Maganaris et al., 2008). This underestimation could be accentuated by the fact of reporting engineering stress instead of true stress. Engineering stress at tendon failure has been found around 10% smaller than the corresponding true stress (Vergari et al. 2011). From a clinical point of view the failure values reported can be considered as conservative.

Other parameter of the testing process, which may have an impact in the results, is the storage method, but the literature is inconsistent in this regard. Different studies conclude in conflicting resolutions about the influence of freezing storage in the mechanical properties. Giannini et al. (2008) found important changes in frozen–thawed tendons compared to controls in a histological and mechanical analysis. Similarly, Smith et al. (1996) found significant differences in mechanical properties comparing distinct freezing methods. However, Huang et al. (2011) did not register mechanical differences until the fifth freezing–thawing cycle. No alterations in the tensile properties in other human and animal tendon studies have been also reported (Ho and Meng 2002; Jung et al. 2011; Vanbrocklin and Ellis 1965). These contradictory results are also present in ligament literature (Moon et al. 2006; Viidik and Lewin 1966). Nevertheless, for comparison purposes the unconfirmed impact of storage method is counteracted by the fact that most of the *in vitro* studies of human tendon used freezing methods to preserve the samples until the testing day.

Conversely, it has been observed in previous studies that tendons and ligaments exhibit strain rate sensibility (Abrahams 1967; Noyes et al. 1974; Vanbrocklin and Ellis 1965; Wren et al., 2001). As a viscoelastic tissue, tendons became stiffer at higher strain rates. The rate used in this study can be considered slow rate, thus in quick events such as ankle sprain the failure will occur later. The rate of strain has to be taken into account when comparing results of mechanical tests.

From the perspective of reconstruction surgery, the mechanical parameter used to evaluate a potential graft is the failure load. Previous studies in this matter have studied the structural properties of different foot tendons for reconstruction of Achilles tendon and knee ligaments. Zhao and Huangfu (2012) reported strength values of anterior half of the PL of 322.4 N (SD 63.2) comparable to our measurement of 346 N (SD 196). Other *in vitro* studies have informed a failure load of 333.1 N (SD 137.2) and 348.8 N (SD 124.9) for PB (Datta et al. 2006; Sebastian et al. 2007), somewhat higher than 239 N (SD 99) that we measured. In the same studies, failure loads of FHL of 511 N (SD 164.3) and 241.5 N (SD 82.2) were also reported. In that case, inferior to the 525 N (SD 200) currently measured. Bigger differences were found with the TP failure loads reported by Giannini et al. (2008) which showed fourth times more strength than the samples tests in the current investigation.

Regarding material properties, the most complete dataset reported about these lesser-studied foot tendons was the investigation carried out by Blanton and Biggs (1970). They investigated the ultimate tensile stresses depending on the position and function of the muscle. They reported lower extremity extensor and flexor tendon values separately. The stress failure ranged between 9–55 MPa for both groups being 26.2 MPa the average for extensor tendons and 31.2 MPa for flexor tendons. These results are similar to the range of 6–76 MPa obtained in the present study and the mean values of 22.3 MPa and 22.1 MPa for extensor and flexor tendons respectively. These results do not represent a significant difference between extensor and flexor tendons in accordance with Cronkite (1936) findings. However, Benedict et al. (1968) found a slightly higher difference between extensor and flexor tendons. Extensor tendons failed at an average stress of 92.3 MPa while flexors only reached 75.4 MPa. Schechtman and Bader (1997) obtained a similar range, around 100 MPa of ultimate tensile stress for EDL tendons. In that study Young's modulus were not reported, but it can be inferred from their typical stress–strain curve, mean ultimate tensile stresses and mean strain failure results that they obtained values that doubled our measurements as happens with the Young's modulus reported by Benedict et al. (1968) and Giannini et al. (2008).

The disparity of results between experiments is likely caused by methodological differences such as storage method, criteria of exclusion, clamping system, preconditioning, preload configuration or strain rate coupled with the natural variability of living tissues (Cook et al. 2014). Maganaris et al. (2008) argued that this discrepancy is substantially reduced when comparing mechanical properties using the same methodology. The properties assessed in this study constituted a reference of the mechanical properties of the balance foot tendons calculated on the basis of the same methodology.

As anecdotal note, during testing in the laboratory room was possible to hear the soft sound of the tendon fiber breakage which reminds the guitar strings break noise. Benedict et al. (1968) pointed out this sound as rapidly break apart of fiber bundles.

This study assessed *in vitro* the mechanical properties of the main balance tendons of the human foot reporting structural and material properties sorted by tendon. Detailed information of the methodology employed during experiments was provided in order to frame the results for future applications. Although the number of specimens per tendon is limited, the potential of the study lies in the comparison between different foot tendons under the same test protocol. The results obtained can be utilized in different engineering and clinical applications. The material properties together with the quantitative description of the stress–strain curve will help in the design of synthetic materials and in the development of refined constitutive models for computational simulation. The structural properties provide worthy information to practitioners of eligible tendons for reconstruction surgeries and for the estimation of tendon injuries.

5. Conclusions

The *in vitro* study of the mechanical properties of the main balance foot tendons revealed that flexor/extensor toes tendons have higher Young's modulus and ultimate tensile stress while inversion/eversion tendons have larger strain failure. In other words, Tibialis and Peroneus tendons are more deformable at the same stress and strain rate. A criterion to quantify the regions of the stress–strain curve was defined. Stress–strain tendon curves exhibit proportionality between regions corresponding the initial strain, the toe region and the yield point to the 15, 30 and 70% of the strain failure respectively.

Conflict of interest

There is no conflict of interest.

Acknowledgments

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References

Abrahams, M., 1967. Mechanical behaviour of tendon in vitro. Med. Biol. Eng. 5, 433–443.

Benedict, J.V., Walker, L.B., Harris, E.H., 1968. Stress–strain characteristics and tensile strength of unembalmed human tendon. J. Biomech. 1, 53–63. http://dx.doi.org/10.1016/0021-9290(68)90038-9.

Blanton, P.L., Biggs, N.L., 1970. Ultimate tensile strength of fetal and adult human tendons. J. Biomech. 3, 181–189.

Butler, D., Grood, E., Noyes, F., Brackett, K., 1984. Effects of structure and strain measurement technique on the material properties of young human tendons and fascia. J. Biomech. 1, 579–596.

Cook, D., Julias, M., Nauman, E., 2014. Biological variability in biomechanical engineering research: significance and meta-analysis of current modeling practices. J. Biomech. 47, 1241–1250. http://dx.doi.org/10.1016/j.jbiomech.2014.01.040.

Cronkite, A.E., 1936. The tensile strength of human tendons. Anat. Rec. 64, 173–186. http://dx.doi.org/10.1002/ar.1090640205.

Datta, B., Salleh, R., Mafulli, N., Neil, M., Butler, A., Walsh, W., 2006. Mechanical properties of human flexor hallucis longus, peroneus brevis and tendo achilles tendons. 52nd Annual Meeting of the Orthopaedic Research Society Paper no : 1889, p. 1889.

Devkota, A.C., Weinhold, P.S., 2003. Mechanical response of tendon subsequent to ramp loading to varying strain limits. Clin. Biomech. 18, 969–974. http://dx.doi.org/10.1016/S0268-0033(03)00168-2.

Giannini, S., Buda, R., Di Caprio, F., Agati, P., Bigi, A., De Pasquale, V., Ruggeri, A., 2008. Effects of freezing on the biomechanical and structural properties of human posterior tibial tendons. Int. Orthop. 32, 145–151. http://dx.doi.org/10.1007/s00264-006-0297-2.

Ho, N., Meng, C., 2002. The effect of post mortem freezing storage on the tensile properties of tendon. Proceedings of the IEEE 28th Annual, pp. 53–54.

Huang, H., Zhang, J., Sun, K., Zhang, X., Tian, S., 2011. Effects of repetitive multiple freeze–thaw cycles on the biomechanical properties of human flexor digitorum superficialis and flexor pollicis longus tendons. Clin. Biomech. 26, 419–423. http://dx.doi.org/10.1016/j.clinbiomech.2010.12.006.

Jung, H.-J., Vangipuram, G., Fisher, M.B., Yang, G., Hsu, S., Bianchi, J., Ronholdt, C., Woo, S.L.-Y., 2011. The effects of multiple freeze–thaw cycles on the biomechanical properties of the human bone-patellar tendon-bone allograft. J. Orthop. Res. 29, 1193–1198. http://dx.doi.org/10.1002/jor.21373.

Ker, R., 2007. Mechanics of tendon, from an engineering perspective. Int. J. Fatigue 29, 1001–1009. http://dx.doi.org/10.1016/j.ijfatigue.2006.09.020.

Ker, R., Wang, X., Pike, A., 2000. Fatigue quality of mammalian tendons. J. Exp. Biol. 1327, 1317–1327.

Maganaris, C.N., Narici, M.V., Maffulli, N., 2008. Biomechanics of the Achilles tendon. Disabil. Rehabil. 30, 1542–1547. http://dx.doi.org/10.1080/09638280701785494.

Moon, D.K., Woo, S.L.-Y., Takakura, Y., Gabriel, M.T., Abramowitch, S.D., 2006. The effects of refreezing on the viscoelastic and tensile properties of ligaments. J. Biomech. 39, 1153–1157. http://dx.doi.org/10.1016/j.jbiomech.2005.02.012.

Morales-Orcajo, E., Bayod, J., Barbosa de Las Casas, E., 2015. Computational Foot Modeling: Scope and Applications. Arch. Comput. Methods Eng. http://dx.doi.org/10.1007/s11831-015-9146-z.

Noyes, F.R., DeLucas, J.L., Torvik, P.J., 1974. Biomechanics of anterior cruciate ligament failure: an analysis of strain-rate sensitivity and mechanisms of failure in primates. J. Bone Joint Surg. Am. 56, 236–253.

Schechtman, H., Bader, D., 1997. In vitro fatigue of human tendons. J. Biomech. 30, 829–835.

Sebastian, H., Datta, B., Maffulli, N., Neil, M., Walsh, W.R., 2007. Mechanical properties of reconstructed Achilles tendon with transfer of peroneus brevis or flexor hallucis longus tendon. J. Foot Ankle Surg. 46, 424–428. http://dx.doi.org/10.1053/j.jfas.2007.07.003.

Sharkey, N.a., Hamel, A.J., 1998. A dynamic cadaver model of the stance phase of gait: performance characteristics and kinetic validation. Clin. Biomech. 13, 420–433.

Smith, C.W., Young, I.S., Kearney, J.N., 1996. Mechanical properties of tendons: changes with sterilization and preservation. J. Biomech. Eng. 118, 56–61. http://dx.doi.org/10.1115/1.2795946.

Spyrou, L.a., Aravas, N., 2011. Muscle and tendon tissues: constitutive modeling and computational issues. J. Appl. Mech. 78, 041015. http://dx.doi.org/10.1115/1.4003741.

Thordarson, D.B., Schmotzer, H., Chon, J., Peters, J., 1995. Dynamic support of the human longitudinal arch — a biomechanical evaluation. Clin. Orthop. Relat. Res. 165–172.

Vanbrocklin, J.D., Ellis, D.G., 1965. A study of the mechanical behavior of toe extensor tendons under applied stress. Arch. Phys. Med. Rehabil. 46, 369–373.

Vergari, C., Pourcelot, P., Holden, L., Ravary-Plumioën, B., Laugier, P., Mitton, D., Crevier-Denoix, N., 2010. A linear laser scanner to measure cross-sectional shape and area of biological specimens during mechanical testing. J. Biomech. Eng. 132, 105001. http://dx.doi.org/10.1115/1.4002374.

Vergari, C., Pourcelot, P., Holden, L., Ravary-Plumioën, B., Gerard, G., Laugier, P., Mitton, D., Crevier-Denoix, N., 2011. True stress and Poisson's ratio of tendons during loading. J. Biomech. 44, 719–724. http://dx.doi.org/10.1016/j.jbiomech.2010.10.038.

Viidik, A., Lewin, T., 1966. Changes in tensile strength characteristics and histology of rabbit ligaments induced by different modes of postmortal storage. Acta Orthop. Scand. 37, 141–155.

Wang, J.H.-C., 2006. Mechanobiology of tendon. J. Biomech. 39, 1563–1582. http://dx.doi.org/10.1016/j.jbiomech.2005.05.011.

Wren, T.a., Yerby, S.a., Beaupré, G.S., Carter, D.R., 2001. Mechanical properties of the human achilles tendon. Clin. Biomech. 16, 245–251.

Zhao, J., Huangfu, X., 2012. The biomechanical and clinical application of using the anterior half of the peroneus longus tendon as an autograft source. Am. J. Sports Med. 40, 662–671. http://dx.doi.org/10.1177/0363546511428782.