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Perspective Article

Hylonome

In vivo human tendon mechanical properties: Effect of resistance training in old age

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Abstract

Recent advances in ultrasound scanning have made it possible to obtain the mechanical properties of human tendons *in vivo*. Application of the *in vivo* method in elderly individuals showed that their patellar tendons stiffened in response to a 14-week resistance training program by ~65% both structurally and materially. The rate of muscle torque development increased by ~27%, indicating faster contractile force transmission to the skeleton. The present findings suggest that strength training in old age can at least partly reverse the deteriorating effect of ageing on tendon properties and function.

Keywords: Collagen, Ultrasound, Exercise, Ageing, Muscle

Functional implications of tendon viscoelasticity

Although the primary role of tendons is to transmit contractile forces to the skeleton to generate joint movement, tendons do not behave as rigid bodies. Instead, they exhibit a viscoelastic behaviour on tension, which has important implications for whole-body function for reasons that include a) modulation of the length and force of the in-series muscle[1], b) influence of joint-position control[2], c) reduction of metabolic energy during locomotion[3].

Testing methodologies

The mechanical properties of tendons have been studied mostly using tensile testing methodologies, in which isolated tendon specimens are stretched by an external force, while both the specimen deformation and the applied force are recorded[4-6]. Such methodologies are considered to mimic adequately the way that loading is imposed on several tendons in real life. However, questions are raised as to whether the results of *in vitro* methods can be directly extrapolated to infer *in vivo* function. This is because: 1) The forces exerted by maximal tendon loading under *in vivo* conditions may not reach the "linear" region in which stiffness and Young's modulus measurements are taken under *in vitro* conditions. 2) To perform a tensile test *in vitro*, clamping of the specimen is necessary. Fixing a fibrous structure with clamps is inevitably associated with fibre slippage and/or stress concentration that may result in premature rupture. 3) Many *in vitro* experiments have been performed using preserved tendons, which may have altered properties[7]. Some of the above problems have been circumvented by testing animal tendons *in situ* after the animal has been killed or anaesthetized[8]. The *in situ* muscle preparation is made to contract by electrical stimulation, thus pulling on the tendon, which lengthens as a function of the contractile force applied on it in a fashion similar to that obtained when the actuator of a tensile machine pulls an isolated specimen. Clearly, such *in situ* protocols are inapplicable to humans. However, adapting similar principles to those used under *in situ* material testing has recently allowed the development of a non-invasive method for assessing the mechanical properties of human tendons *in vivo* (see representative studies in Table 1). The *in vivo* method is based on real-time ultrasound scanning of a reference point along the tendon during static contraction of the in-series muscle. The limb is fixed on the load cell of a dynamometer to record changes in joint moment and estimate changes in muscle force occurring during activation and subsequent relaxation in a maximal isometric contraction. The muscle forces generated pull the tendon proximally and cause a longitudinal deformation, which is measured by the recorded displacement of the reference land-

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Tendon	Structure	Stiffness ($N \cdot mm^{-1}$)	Young's modulus (GPa)	Ref.
Tibialis anterior	Tendon	161	1.2	9
»	Tendon-aponeurosis	32	0.5	10
Gastrocnemius	Tendon	150	1.2	11
»	Tendon-aponeurosis	467-494	1.0-1.5	12
»	Tendon-aponeurosis	26-33	-	13
Triceps surae	Tendon	760	0.8	14
Vastus larteralis	Tendon-aponeurosis	68-106	-	15
»	Tendon-aponeurosis	161-770	-	16

Table 1. Typical examples of ultrasound-based mechanical property values in human tendons.

mark in the tendon. The force-elongation plots obtained can be reduced to the respective stress-strain plots by normalization to the dimensions of the tendon, which can also be measured using ultrasonography. From these curves, the stiffness ($N \cdot mm^{-1}$) and Young's modulus (GPa) of the tendon can be calculated. The above principles have been applied to several human tendons and tendon-aponeurosis systems, mostly in young adults, yielding typical values of stiffness and Young's modulus illustrated in Table 1. In this report, we will focus on elderly human tendons and examine the changes they undergo in response to strength training. A more comprehensive presentation of these results has appeared elsewhere[17].

Effect of resistance training on elderly human tendon

The mechanical properties of the patellar tendon in a group of older adults before and after 14 weeks of knee extensor strength training ($n=9$, mean age: 74 years) and in a non-training control group ($n=9$, mean age: 67 years) tested at baseline and after 14 weeks of normal activity. Training was performed 3 times per week using isotonic resistance leg-extension and leg-press devices, at an intensity of 80% of the 5-repetition maximum.

The elongation of the patellar tendon was assessed during a ramp isometric knee extension performed at a 90 deg knee angle (full extension = 0 deg), with the hip angle placed at 85 deg (supine position = 0 deg) on an isokinetic dynamometer (Cybex NORM). A 7.5 MHz linear ultrasound probe (ATL-HDI 3000) was held by a custom-built external fixation device on the skin along the patellar tendon. On the scans recorded during contraction (30 Hz), the displacement of the apex of the patella was measured at 10 % intervals of maximal torque and was assumed to represent the elongation of the patellar tendon (Figure 1). To estimate the absolute tendon forces corresponding to the above torque values, the following procedures were followed. First, the measured knee extension torque was corrected for the effect of antagonistic muscle action. This was achieved by taking measurements of electromyographic (EMG) activity from the biceps femoris muscle during the isometric knee extension test used to measure tendon elongations and during an isometric knee flexion test at the same joint configuration. The antagonistic torque of the knee flexors during a knee extension contraction was calculated from the EMG-torque relationship of the biceps femoris muscle when acting as an agonist. The addition of this torque to the measured knee extension torque yielded the torque produced by the quadriceps muscle group alone. By dividing this torque value by the patellar tendon moment arm length, the respective patellar tendon force was calculated. The patellar tendon moment arm length was quantified from magnetic resonance images (E-scan, Esaote Biomedica) as the distance between the tibio-femoral contact point and the patellar tendon line of pull. The tendon forces estimated were combined with the tendon elongations measured, and the slope of the curve produced over the interval 90-100% of maximal force was calculated. The values produced gave the stiffness of the patellar tendon. The force-elongation curves were transformed to stress-strain curves by normalizing force to the tendon's cross-sectional area and stress to the tendon's original length. The tendon's cross-sectional area was measured on axial-plane sonographs taken at 25, 50 and 75% of the tendon's length. The tendon's length was measured on longitudinal sonographs as the length from the apex of the patella to the superior aspect of the tibial tuberosity. The slope of the stress-strain curve over the interval 90-100% of maximal stress was the Young's modulus of the tendon.

Measurement of the rate of torque development (RTD) was also taken to assess whether any training-induced changes in tendon mechanical properties would affect the speed of

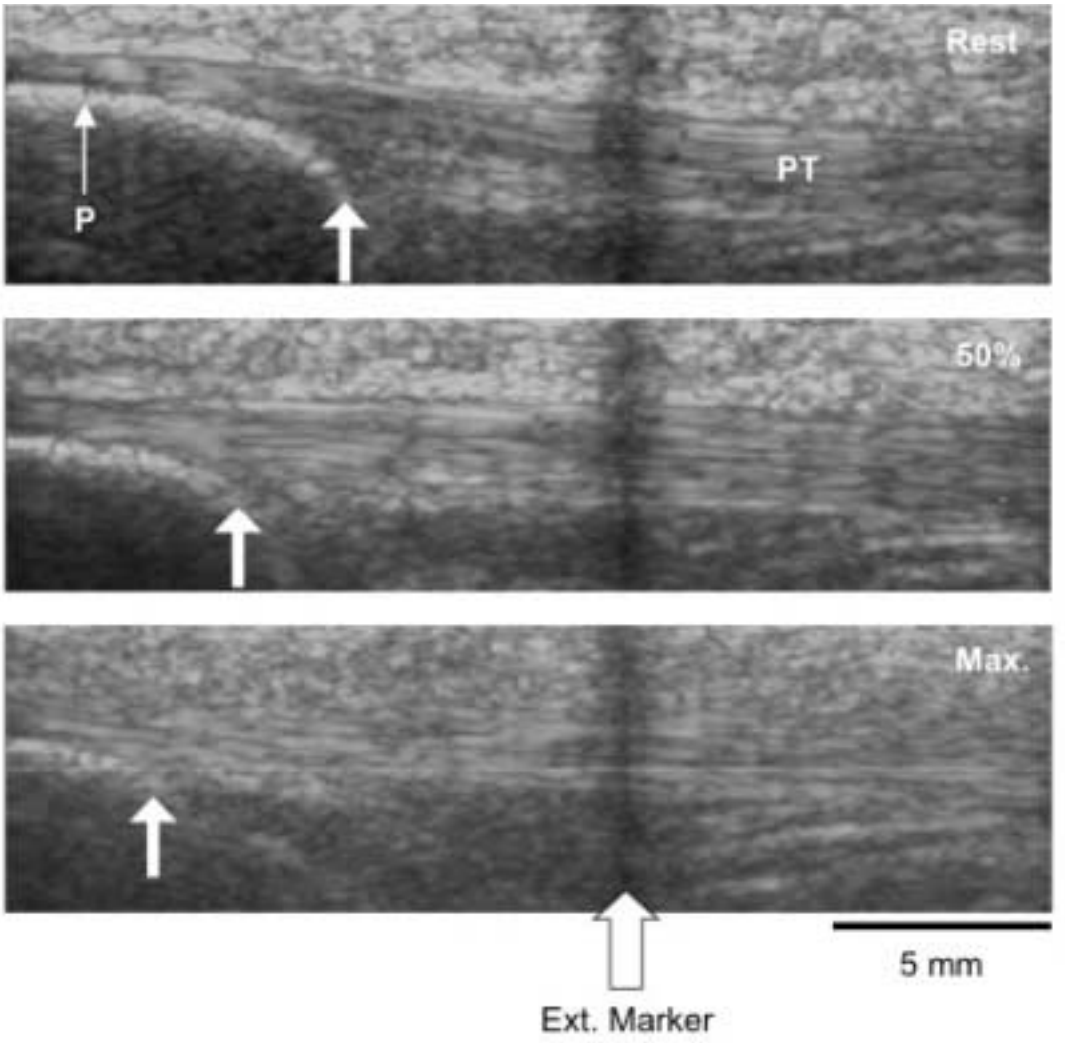

Figure 1. Sagittal-plane scans of the patellar tendon (PT) at rest, during isometric contraction (top) at 50% of maximal force (middle) and at maximal tendon force (bottom). Arrows indicate the apex of the patella (P). Reprinted with permission from Reeves et al.[17].

contractile force transmission to the skeleton. For this measurement, the subjects were instructed to reach their maximal contraction torque as rapidly as possible. The RTD was calculated from the gradient of the torque-time relationship over the first 100 ms after the onset of torque development.

Measurements taken on separate days showed good agreement: the intraclass correlation coefficients were 0.99 for tendon cross-sectional area, tendon length, sub-maximal and maximal tendon elongations. Typical error was 1.5 mm^2 for tendon cross-sectional area, 0.6 mm for tendon length, and 0.1 mm for sub-maximal and maximal tendon elongations.

Results and discussion

The patellar tendon force-elongation and stress-strain curves had the established curvilinear pattern, with larger elongations and strains at lower loads and *vice versa*. The values of the main variables examined are illustrated in Table 2. As seen, the stiffness and Young's modulus of the control and training groups were very similar at baseline. In the control group, the 14-week period of normal activity had no effect on the tendon mechanical properties. In marked contrast, the 14-week period of strength training increased the stiffness and Young's modulus of the tendons in the intervention group by ~65%. The RTD increased by ~27 % after training, but remained unaltered in the control group. The finding of training-induced increases in structural stiffness agrees with previous findings in animal studies[18-20] and indicates that training in old age can at least partly reverse the deteriorating effect of ageing on tendon mechanical properties[21-23]. The similarities in stiffness and Young's modulus increase indicates that training stiffened the material the tendon is made of, rather than inducing hypertrophy. The exact mechanisms involved, however, are not apparent from

	Training Group		Control Group	
	Pre	Post	Pre	Post
Stiffness ($N \cdot mm^{-1}$)	2187	3610	2247	2255
Young's modulus (GPa)	1.3	2.2	1.3	1.3
Tendon CSA (mm^2)	84	84	87	87
Tendon length (mm)	48	48	48	48
RTD ($Nm \cdot s^{-1}$)	483	613	502	437

Table 2. The mean values of the main variables examined. CSA, cross-sectional area; RTD, rate of torque development.

the present study. Based on findings from animal studies, it is suggested that both collagen turnover and the packing density of collagen fibrils may have increased[19,24,25]. Also, alterations in the crimp angle of collagen fibrils have been reported to occur following exercise training in rat tendons[20]. An increased tendon collagen content and decreased crimp angle would increase the tendon's material stiffness. Changes in the tendon's water content might also contribute to the increases in tensile stiffness[26]. The 27% increase in the RTD after training is consistent with the increased tendon stiffness and indicates faster transmission of contractile forces to the skeleton. This has implications for the rapid execution of motor tasks in old age, such as accelerating or decelerating the body and reacting to a fall. Another implication of the changes in tendon stiffness relates to the behavior of the in-series muscle. The increased patellar tendon stiffness observed after training could result in a reduced muscle fibre shortening in the quadriceps femoris muscle group, which would then cause a shift in the optimal angle for force generation. This would suggest that training-induced changes in tendon stiffness might alter the force-length relation of muscle, irrespective of any changes in muscle size and neural activation.

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