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Viewpoint

On the hysteresis in the human Achilles tendon

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On hysteresis in the human Achilles tendon

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Elastic hysteresis is a property of tendon and describes the energy dissipated due to material viscosity. The amount of tendon hysteresis is important for efficiency of locomotion. Higher hysteresis is associated with greater energy dissipation as heat, and thus less energy can be recoiled to propel our movements. Classical papers report hysteresis of about 7% in the plantaris tendon of sheep (9) and about 10% in tendons of different mammals (3, 24). While greater hysteresis values have been presented especially in human studies in vivo (e.g. 10-20), several authors have suggested that the low hysteresis values are likely to be realistic because it ensures greater elastic recoil and minimizes heat damage (1, 3, 9).

Since the 1990's ultrasound imaging has become a popular tool when assessing *in vivo* tendon properties in humans. It is possible to measure tendon properties from isometric loading-unloading cycles (e.g. 20, 25) and even during natural locomotion such as hopping (18). The most often reported tendon property is stiffness, a very relevant parameter regarding the potential to store elastic energy. However, the amount of energy dissipation that occurs after storage (i.e. hysteresis) also affects efficiency of our locomotion. This raises questions about why there are far more

studies reporting stiffness than hysteresis. For example, the PubMed search term ‘tendon stiffness’ returns 1689 hits compared to just 69 for ‘tendon hysteresis’.

This viewpoint was stimulated by two observations: 1) the statistical skewness whereby numerous articles have reported tendon stiffness and Young’s modulus, but far fewer have reported tendon hysteresis; 2) *in vivo* human studies seem very often to report hysteresis values greater than 10%, suggesting either that there are methodological differences between in vivo and in vitro studies, or that human tendons *in vivo* have a much poorer capacity to store and reutilize elastic energy. In this article we focus on the healthy human Achilles/gastrocnemius tendon (AT) since it has an important locomotor function, and clearly a low AT hysteresis would allow elastic recoil for efficient locomotion (1, 27).

Figure 1 shows the mean hysteresis values from selected animal studies and from the majority of human studies in the last 30 years. Two observations are evident from the figure: 1) animal studies report smaller values than human studies; 2) in the human data there is a very large range of hysteresis values. The variability in human studies may be explained by several methodological factors. Firstly, the definition of tendon length and assessment of length change both vary. For example, the tendon can also include parts of the aponeurosis, and not only the “free” external tendon, which appears to have lower hysteresis (30) than the gastrocnemius tendon (Fig. 1). In the literature there are about five different ways that have been used to assess tendon length change during voluntary contractions by ultrasonography:

1) Movement of a medial gastrocnemius (MG) muscle fascicle-tendon cross point is traced using ultrasonography. The displacement of this point is taken

as the change in tendon length (e.g. 12-14). This early method has disadvantages that the following methods account for partly or completely: a) "tendon length" includes aponeurosis with differing properties, b) it does not account for displacement at the insertion of the tendon, and c) absolute tendon length is not assessed.

2) Movement of both MG muscle-tendon junction and calcaneus are tracked. The difference between the displacements of these points denotes the change in tendon length (e.g. 20). Free AT length change has been obtained similarly by tracking the soleus muscle-tendon junction but video analysis was used instead of ultrasonography to track calcaneal movement (30).
3) MG muscle-tendon junction is tracked with corrections including calcaneal rotation that has been determined during passive movement (e.g. 7, 8, 21).
4) MG muscle-tendon junction is tracked using ultrasonography with motion analysis recording of both the heel and the ultrasound probe positions, and the linear distance between the tendon origin and insertion is calculated (5, 18).
5) The same as in 4 but including the curvature of the tendon (e.g. 2, 28).

Secondly, measurements without tendon preconditioning may be one source of the greater hysteresis (4, 19). Thirdly, tendon force measurements contain uncertainties that arise from the estimations and assumptions required in calculating the forces. A common assumption is that all of the plantar flexor moment is transmitted via the Achilles tendon, although contributions from other muscles (synergistic and agonistic) are likely to occur (6). Further, the moment arm values used can affect the force values considerably, and mean values from the literature may distort the individual variability.

Uncertainties in tendon length and force measurements affect not only hysteresis but also other measures of tendon properties, raising the important question of why hysteresis values are not reported as often as stiffness values. One may consider that hysteresis is generally so small that it can be ignored while stiffness is the primary property affecting muscle-tendon function. From a practical perspective, measurements of hysteresis are more demanding *in vivo* in humans, where the smoothness of the unloading phase is much more difficult to control than the loading phase (22). This difficulty in force control is characterized by fluctuations of the curve during the unloading phase, from which the raw data are rarely presented (Fig 2) (5, 18). This difficulty to control the relaxation phase may well add to the variability between individuals. For example, hysteresis ranges of 2-45% (mean 17% (5)), 17-35% (mean 26% (18)), 4-40% (mean 19% (16)), 10-37% (mean 22% (17)), and 4-36% (mean 17% (14)) have been reported. Farris et al. (5) speculated that this large individual variation places some people at greater risk of thermal damage.

Is this large variability due to individual differences or methodological uncertainty? In the literature, Young's modulus in particular shows much less variability than hysteresis in human studies *in vivo*. Stiffness within a given tendon, which is associated with muscle strength, also shows less relative variation (SD/mean) than hysteresis (21% vs. 55%) in the same studies that present a large range in hysteresis values (5, 12, 18).

Therefore, we examined whether methodological issues contribute to the large variability of hysteresis relative to stiffness values. From our experience, although

modern ultrasound devices are equipped with possibilities to synchronize the images with the force data, there is a possibility for desynchronization due to computer processing time and because the ultrasound sampling frequency is usually much lower than that of other variables, e.g. force measurements.

To demonstrate the effect of desynchronization of force and ultrasound data (collected using method 4) on Achilles tendon hysteresis and stiffness in an isometric loading-unloading task, we purposefully offset the ultrasound-derived tendon length frame by frame (data from reference 23, N=12). With our recordings at 100 Hz, a shift of one frame (10 ms) caused hysteresis to decrease from 6% to -3%, and a further shift reduced it to -15%. The same shifts in the opposite direction increased hysteresis from 6% to 15% and 23%. Interestingly, the shift had a much smaller effect on stiffness, which decreased gradually from 220 Nm^{-1} (at -15% hysteresis) to 207 Nm^{-1} (-3%), 196 Nm^{-1} (6%), 187 Nm^{-1} (15%) and 180 Nm^{-1} (23%) (Fig. 2). Thus, while desynchronization of force and displacement by 10 ms increased hysteresis by 9-10%, stiffness only changed by 4-5%, illustrating that hysteresis is a much more sensitive measure than stiffness to desynchronization.

The large variability in hysteresis may be explained by a low sampling frequency of ultrasound images. For example, if the sampling frequency is 50 Hz, the maximum desynchronization is 20 ms corresponding to about 20% over/underestimation in hysteresis according to our data. While this may be treated by averaging multiple trials from each subject, it leaves an open question regarding the systematic trend toward higher hysteresis *in vivo* than *in vitro*. Can it be methodological or possibly a publication bias where authors have not reported hysteresis due to its large variability

which may also include negative values? This can only be resolved by a validation study where tendon hysteresis is first measured *in vivo* and then the same tendon is tested *in vitro*.

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Figure legends

Figure 1. Mean tendon hysteresis values from selected animal studies (filled symbols, various animal species) and Achilles/gastrocnemius tendon hysteresis in humans determined using ultrasonography (open symbols). The number beside the symbol refers to the number of reference.

Figure 2. An example from a representative subject demonstrating the effect of shift of tendon length data relative to the estimated tendon force by 10 ms. It caused stiffness and hysteresis to change from 222 Nm^{-1} and -16% (left panel) to 212 Nm^{-1} and -4%, to 203 Nm^{-1} and 6%, to 195 Nm^{-1} and 15% and to 188 Nm^{-1} and 23%. It is noteworthy that the loading phase of the curves displays a typical monotonic relationship while the unloading phase contains fluctuations. These fluctuations are likely due to errors in force calculations because the more demanding force control in the unloading phase may require differential use of synergistic and antagonistic muscles.

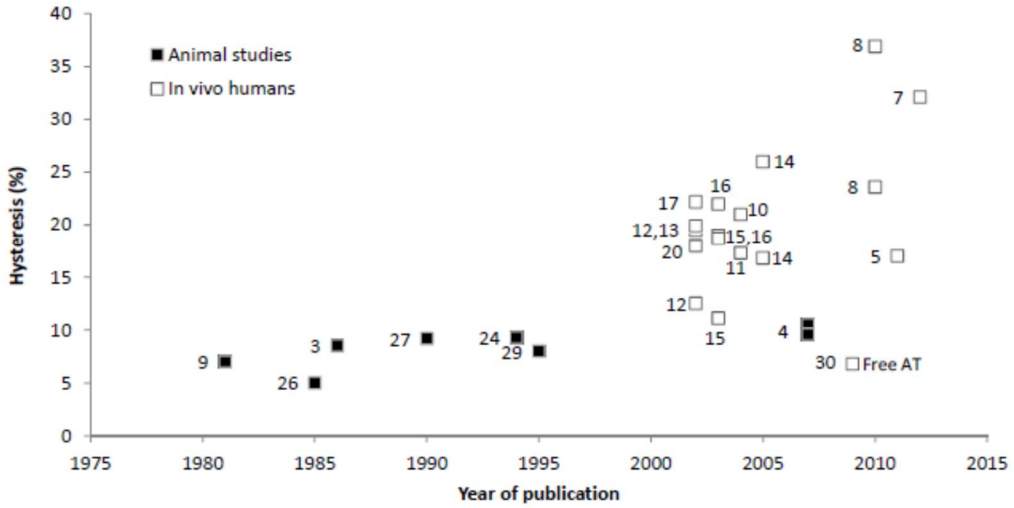

Figure 1.

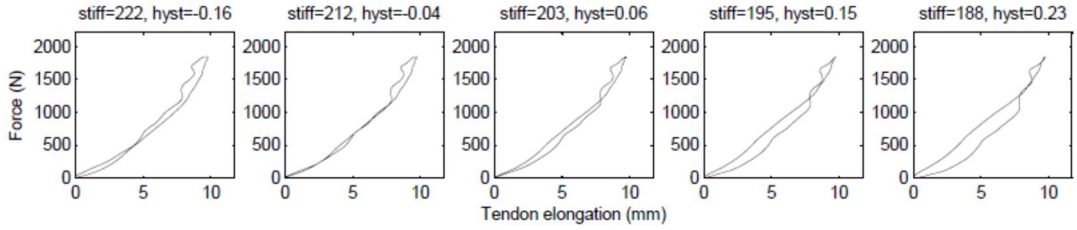

Figure 2.