1

Imaging Studies of the Mechanical and Architectural Characteristics of the Human Achilles Tendon in Normal, Unloaded and Rehabilitating Conditions

Shantanu Sinha[1] and Ryuta Kinugasa[2,3]
[1]Department of Radiology,
University of California, San Diego, California
[2]Department of Human Sciences,
Kanagawa University, Kanagawa
[3]Organ/Whole Body-Scale Team,
Computational Science Research Program,
RIKEN, Saitama
[1]USA
[2,3]Japan

1. Introduction

Tendons are known to have a profound impact on the overall function of the musculoskeletal system in their role as a structural link and force transmitter between muscle and bone. Their unique viscoelastic response under tension allows efficient use and recycling of stored energy during stretch involved in locomotion, modulating joint position control, and providing protection from muscle injuries through reduction of mechanical oscillation and shock. The study of the mechanical behavior of the Achilles Tendon, defined as the thick or external tendon running from the calcaneus insertion to the distal part of the soleus muscle, is of particular clinical importance, since it is known to be the most likely site of tendon rupture and tear in humans. This chapter will review the known mechanical, architectural and biochemical characteristics of this tissue, vital for the transmission for force generated by the muscle fibers to the bone.

1.1 Imaging studies

Real-time ultrasonography has become popular for in vivo measurement of human force-length relationships under uniaxial mechanical stress, an important parameter for the assessment of mechanical properties of biological tissues. Recent studies using this technique have shown that, similar to muscles, the mechanical properties of tendon tissue undergo substantial changes in response to both increased and decreased levels of physical loading and with aging and disuse and importantly, that the effects are partly mitigated by resistance training. Such changes in mechanical properties of tendons will significantly affect the overall musculoskeletal performance. However the exact causative mechanisms remain unclear. This chapter will explore the architectural and mechanical characteristics of the tendon that are likely to be modified as a result of chronic unloading and cause reduced force production of the limb and, how these changes can be reversed with physical rehabilitation.

Velocity encoded phase-contrast magnetic resonance imaging (VE-PC MRI) is another imaging technique used to noninvasively measure Achilles tendon strain and changes in its force-displacement relationship concomitant with chronic unloading and subsequent recuperation. This technique will be reviewed in terms of its ability to quantify the Achilles tendon Young's modulus (MPa) from a stress-strain curve. Higher spatial resolution, high tissue contrast and large field of view (FOV) afforded by MRI also allow one to clearly define and segment the two ends of the Achilles tendon; such capabilities are important for elimination of undesirable strain contributed by exogenous tissues and for consistent monitoring of the same anatomic landmarks over the duration of several months as required in longitudinal studies. We will review the results of several studies using the VE-PC MRI technique that provide evidence that tendinous tissues exhibit spatially non-uniform strain patterns and that this heterogeneity of the mechanical behavior of tendinous tissues is altered depending on the contraction type and loading condition.

1.2 Role of Achilles tendon in musculo-skeletal dynamics

A confounding feature of musculoskeletal system is the relationship between the calcaneus excursion (~30 mm) in the human ankle and the length of human soleus muscle fibers (~35 – 45 mm). In order to find the answer to the puzzle, theoretical approaches have been made to understand possible mechanisms which may explain how the human soleus muscle-tendon complex generates movements at the calcaneus that are almost equal to the length of its muscle fibers. Preliminary evidence suggests that some mechanical gain may arise at the ankle where the Achilles tendon is pulled towards the ankle center of rotation, creating an easily observed curvature in the human Achilles tendon. Curvature of the tendon even during relatively high-force contractions suggests that posterior movement of the Achilles tendon is constrained, even though there does not appear any definitive anatomical structure equivalent to the crural ligaments acting on the tendons of the dorsiflexors. Such a constraint would increase the proximo-distal excursion of the calcaneus relative to excursion of the Achilles tendon above the ankle. Elucidation of such mechanisms will be reviewed since these aspects influence tendon moment arm estimations and are critical when predicting mechanical behavior of muscle from joint performance or vice versa. With further understating of structure-function relationship, geometric muscle model will serve as heuristic purposes as well as accurate prediction of muscle function.

2. Mechanical and architectural properties of tendon and aponeurosis

2.1 Stress-strain characteristics of Achilles tendon

Muscle and tendinous structures (aponeurosis and tendon) make up a functional unit, the so called muscle-tendon complex. The in-series, morphological arrangement of tendinous structures within a muscle-tendon complex imposes a force-transmitting role on the structures and also take advantage of their elastic as well as the viscoelastic properties (Rigby et al. 1959; Scott & Loeb 1995; Zuurbier & Huijing 1992) supplementing the passive force transmission with energy storage and recycling. These mechanisms enhance joint performance and efficient power production (Alexander & Bennet-Clark 1977; Cavagna et al. 1964; Hof et al. 2002; Morgan et al. 1978). Therefore, the interaction between muscle and tendinous structures in a muscle-tendon complex has a direct impact on the performance and control of the involved joint(s). In addition, it has been suggested that tendinous structures within a muscle may possess active functions which includes resisting to stretch by producing contractile force, sensing mechanical load, generating mechanical signal and propagating the signal via the gap junction network which probably modulates collagen synthesis (McNeilly et al. 1996; Purslow 2002; Ralphs et al. 2002). The potential consequences for disruption of muscle function by changes in tendon seem almost unrecognized, perhaps in part due to our poor understanding of the structural integration between muscle and tendon and the degree to which their operation may be 'detuned' by changes in mechanical properties of either muscle or tendon. The medical consequences of operating outside of the normal boundaries of a well tuned musculo-tendinous system are also poorly understood, although clearly recognized in the persistent atrophy experienced in microgravity despite rigorous exercise programs.

Muscle force generation is length and velocity sensitive. The process is repetitive in the sense that muscles will always generate force based on their length-tension and force-velocity properties, causing tendon deformation. The magnitude of tendon deformation will depend on its own mechanical properties such as stress-strain relationship. Stress-strain curves are an extremely important graphical measure of a material's mechanical properties. Stress is defined by the ratio of tendon force to tendon cross-sectional area, and strain is defined by the amount of tendon deformation relative to its resting length.

Inspection of the shape of the stress-strain relationship in a soft tissue such as tendon provides insight into the unique properties of most soft tissues compared with traditional materials such as steel or wood. Typical engineering materials usually exhibit linear, elastic, homogeneous, and isotropic properties, which is reflected as a linear stress-strain curve for loads below the elastic limit. In contrast, biological materials often exhibit non-linear, inelastic, inhomogenous and anisotropic behavior. There are three distinct regions of stress-strain (Figure 1: stress-strain curve). The toe region typically lies below 3% strain, a region in which specimen elongation is accompanied by very low stress. This low initial stiffness of tendon in the toe region is thought to be caused, in part, by the straightening of the collagen crimp (Rigby et al. 1959) or shearing action between the collagen fibrils and the ground substance of the tendon (Hooley et al. 1980). The linear region is evident beyond approximately 2 to 3% tensile strain. The slope of this linear portion of the curve has been used to define the "Young's modulus" of the tendon. This region of linear strain extends to about 4 to 5% (Wainwright et al. 1982). The Young's modulus for rat tendon is approximately 1.0 GPa (Rigby et al. 1959). Permanent deformation occurs beyond the region of linear or reversible strain. The ultimate or failure strain of tendon is about 8 to 10% (Rigby et al. 1959). There is a considerable yield region in which tendon deformation is accompanied by very little increase in stress. However, although numerous measurements of tendon stress-strain properties were made historically (Woo et al. 1982), few were made under physiological conditions. This is because the in vivo tendon properties are more

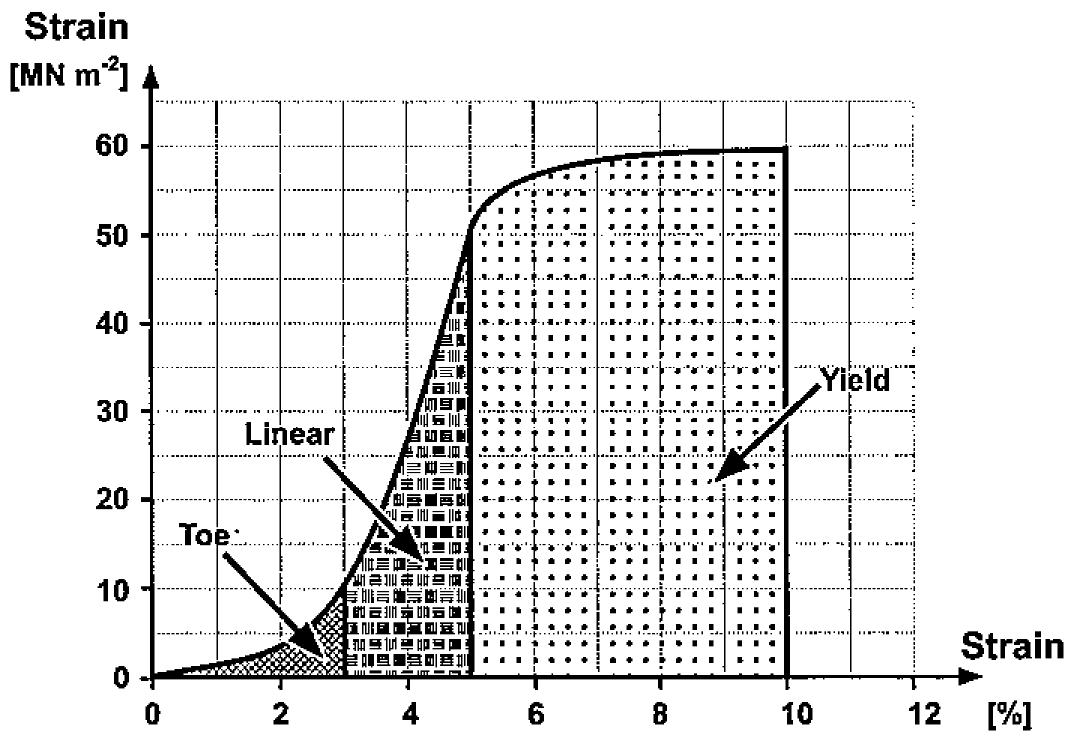


Fig. 1. Typical stress-strain curve for tendon (from Herzog et al. 2007).

difficult to estimate than the simple material properties of tendon tested in isolated condition. Several approaches have been used to define tendon in vivo properties. The examination of tendon properties under in vivo conditions necessitates the use of cadaver specimens. In one approach by Zajac (1989), estimates of tendon strain during muscle contraction were about 3% based on a literature values. Direct measurement of tendon strain during passive loading of a muscle-tendon unit (Lieber et al. 1991) and during muscle contraction (Leiber et al. 2000) yielded approximately similar results in that tendon strain about 3% at muscle maximum titanic tension.

2.2 Experimental determination of stress-strain properties of the Achilles tendon

Real time ultrasound has become popular for in vivo assessment of human tendon stress-strain relationship. The same general principles of in vivo tendon testing have often been applied with the aim of characterizing the mechanical behavior of the human tendon in different in-vivo situations and conditions. The results obtained vary greatly (Arampatzis et al. 2005; Bojsen-Møller et al. 2004; Kubo et al. 2002, 2004; Maganaris and Paul 2002; Muramatsu et al. 2001; Reeves et al. 2005). In young sedentary adults, for example, the tendon stiffness, Young's modulus and mechanical hysteresis values are ~ 17-760 Nm/mm, 0.3-1.4 GPa, and 11-19%, respectively (for a review see Maganaris et al. 2008). In addition, VE-PC MRI is potentially an alternative and supplementary in vivo technique. Tissue velocity measures of the tendinous tissues using VE-PC MRI enable us to estimate the Young's modulus during a submaximal contraction.

Using this VE-PC-MRI based approach, a new method was developed to characterize in-vivo and non-invasively, the mechanical (elastic) properties of the human Achilles tendon (Shin et al, 2008b). Achilles tendon force and calcaneus-movement-adjusted displacement were measured during a submaximal isometric plantarflexion in 4 healthy subjects, 4 repeated trials each. The measured force-length (F-L) relationship was least-squares fitted to a cubic polynomial. The curves were best fitted to a third-order polynomial, with non-linear "toe-region" at smaller forces followed by a linear elastic region. Typical error was calculated for tendon displacement at multiple force levels, stiffness from the "linear region", and transition point from the displacement point separating the linear and non-linear parts of the curve. Elastic constants of human Achilles tendon determined from these force-displacements curves, showed excellent correlation coefficient of each repeat set with the "average" curve, ranging from 0.89-0.99, 0.98-0.99, 0.97-0.99, and 0.88-0.99, respectively (Figure 2A). Qualitatively, individual differences were observed in the force-length profile in proportion to their level of physical activity (Figure 2B). The method yielded Force-Length relationships, stiffness and transition point values that showed good within and day-to-day repeatability. The technique compared well with the more conventional one using ultrasonography. Its reliability indicates potential for measuring tendon structural changes following an injury, disease, and altered loading. Both of these compliance related properties of the tendon have tremendous implications in the transmission of force arising from muscle.

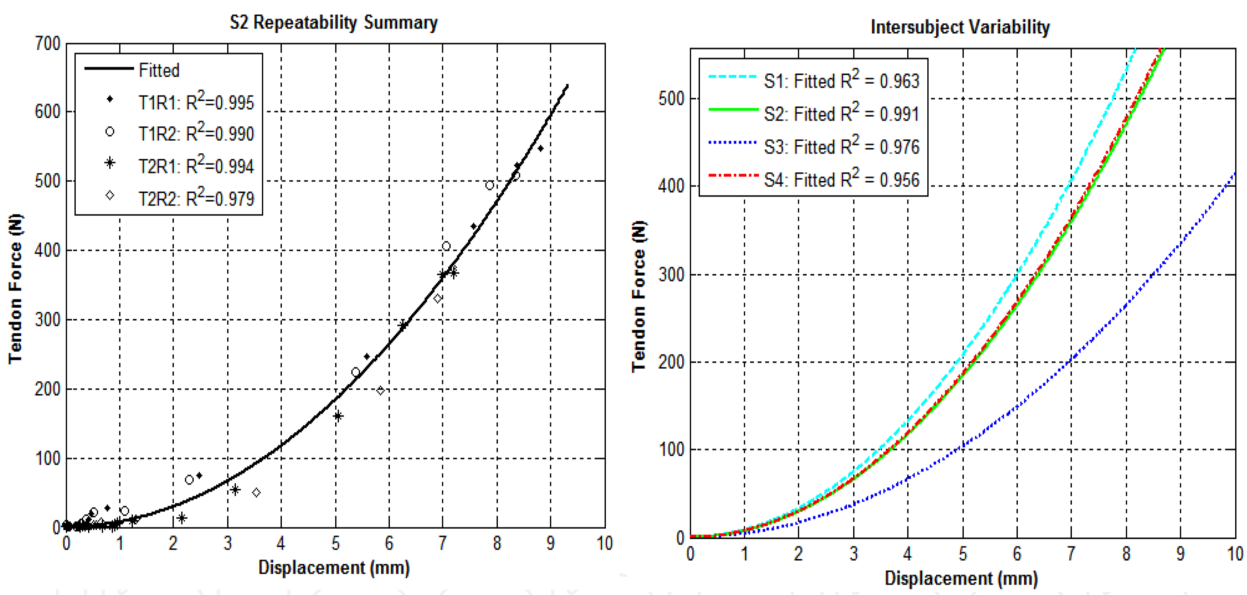


Fig. 2. Variability in Force-Displacement Curves for the Achilles Tendon using VE-PC MRI: (A) Interexam variability in one subject. Results from different trails are shown in different symbols. (B) Intersubject variability shown for 4 different subjects, with each subject show in different colors.

In a typical skeletal muscle-tendon unit, tendinous tissue commonly consists of an external free tendon, which is typically referred to as tendon, and an internal tendon, which is typically referred to as aponeurosis. The tendon connects the muscle proper to bone. The aponeurosis provides the attachment area for the muscle fascicles. In a recent attempt the separation of the mechanical behavior of the aponeurosis from that of the free tendon could be determined during ankle plantarflexion contraction using VE-PC MRI. One of the strengths of this technique compared with ultrasound technique is that it allows one to visualize the entire length of the aponeurosis, tendon, and even calcaneus. This allowed to the comparison of the stiffness between different tendinous tissues such as Achilles tendon and even regional differences along the aponeurosis. In a recent MRI study our group has also shown difference in stiffness between the Achilles tendon and distal aponeurosis of the medial gastrocnemius muscle (Kinugasa et al. 2010).

2.3 Correlation of structure of the Achilles tendon with its function

The primary role of tendon is to transmit the force of its associated muscle to bone. As such, the tendon needs to be relatively stiff and strong under tension. Herzog (2007) have stated that mechanical properties, such as fiber-bundle organization of the tendon allow for the maintenance of high tensile strength, with considerable flexibility in bending, in the same way that a wire rope maintains high tensile strength and flexibility as compared to an equal cross-section of solid steel (e.g., Alexander (1988a); Alexander (1988b) and Wainwright et al. (1982)). The significance of the observed tensile properties can be appreciated by considering the tendon function. Tendon must be sufficiently stiff and strong to transmit muscle force to bone. Ker et al. (1988) studied that relative size of muscle and tendon dimensions. The thin tendons require long muscle fiber, which allow for significant changes in length, to compensate for tendon deformation during muscle contraction (Ker et al. 1988). In contrast, the thick tendons deform less than thin tendons, and may not need extra-long fibers, indicating that the tendon dimension could have an impact on its mechanical properties.

Reconstruction of the lower limb muscle connective tissue from axial anatomical MR images (Iwanuma et al. 2011; Kinugasa et al. 2010) and from the Visible Human data (Hodgson et al. 2006) reveals a complex and somewhat consistent internal structure. The Achilles tendon is rather flat near its broad insertion to the calcaneus but becomes oval in the mid-region and then sheet-like as it courses proximally over the posterior surface of the soleus muscle (Figure 3, Hodgson et al. 2006; Kinugasa et al. 2010). The overall length of human tendon is approximately 68 mm and its width becomes larger as the region shifts more distally from the insertion of the soleus muscle (approximately 13 mm) to the calcaneus (approximately 28 mm) (Iwanuma et al. 2011). A small portion of the tendon and aponeurosis forms a ridge that protruds into the distal portion of the soleus and often reached the anterior surface of the soleus muscle. This structure is generally referred to as the median septum (Oxorn et al. 1998). The median septum extends toward the origin of the soleus for about 70% of the muscle length and is located in the anterior compartment of the muscle as this compartment becomes apparent in the proximal portion of the soleus (Hodgson et al. 2006). The posterior aponeurosis of the soleus muscle also remains very clear over the distal 60–70% of the soleus muscle and continues as a thinner epimyseal sheet in the proximal muscle.

A few studies have examined changes in tendon and aponeurosis dimensions under a human voluntary force exertion condition. In the transition from rest to maximal contraction, the length (superior-inferior direction) and width (medial-lateral direction) of the aponeurosis increases by 7% and 21%, respectively (Maganaris et al. 2001). However, the change in aponeurosis width is variable depending on the measurement regions. Figure 4 shows a 3-dimensional reconstructed image of the entire medial gastrocnemius and axial morphological MR images at 30%, 50%, and 90% locations along the proximal-distal direction under rest, at 20% maximal voluntary contraction (MVC), and at 40% MVC from

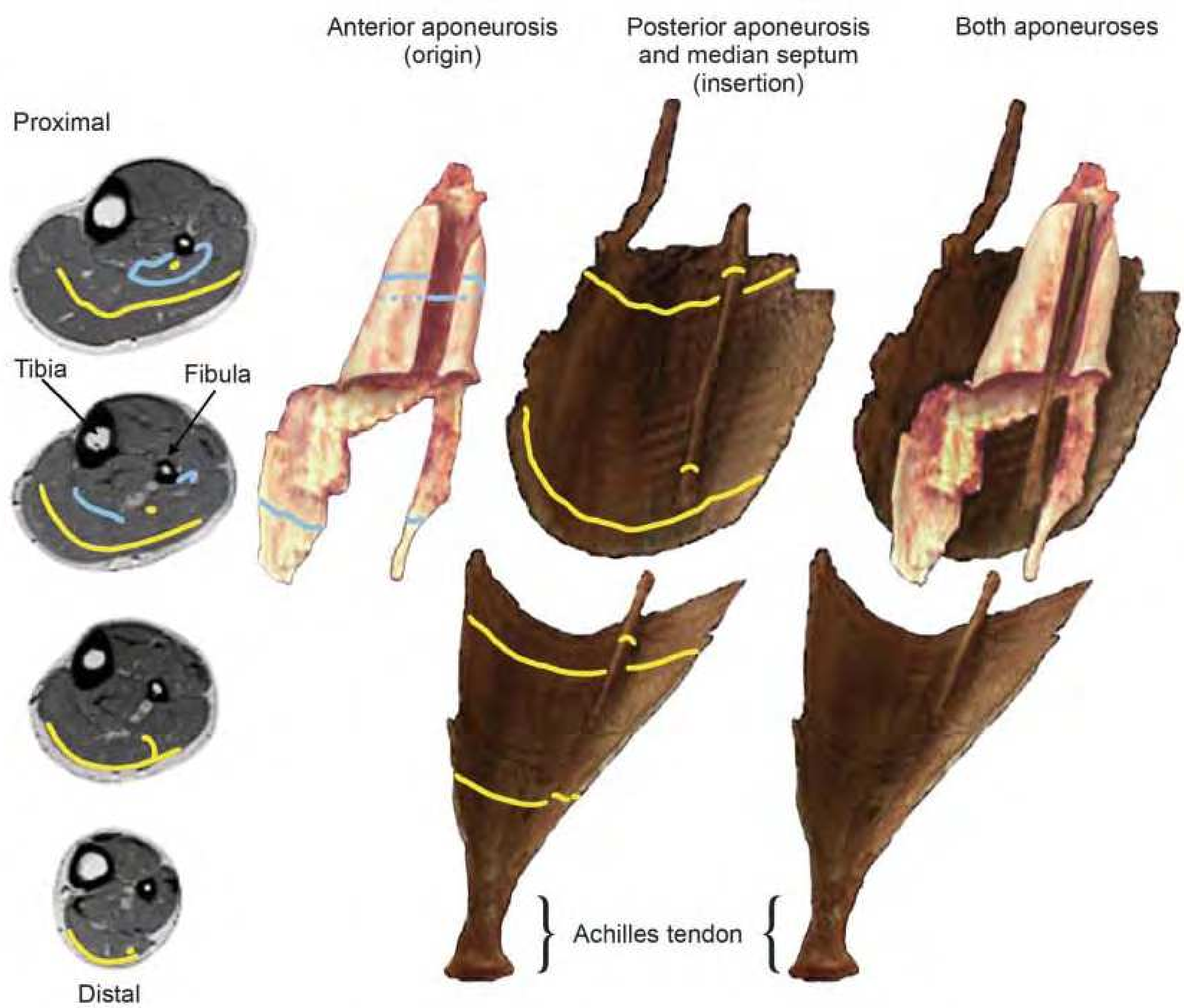


Fig. 3. Three-dimensional digital dissection of the soleus aponeurosis of origin and insertion from MRI images: These are views from the anterior. The gap in the middle of the aponeurosis of insertion indicates that the reconstruction was combined from proximal and distal images of the lower leg. The axial images on the left show the relative location of the origin (blue lines) and insertion (yellow lines). The colored lines in the 3D structures correspond to those in the axial images. A small band of the aponeurosis of insertion did not have a clear termination within the soleus but was contiguous with the gastrocnemius muscle (from Hodgson et al. 2006).

one subject (Kinugasa et al. 2008). At the 50% location, the deep aponeurosis exhibited greater sinuosity in the cross section as force levels increased, which resulted in a significantly greater segment length for 40% MVC (Kinugasa et al. 2008). In contrast, the cross-sectional segment length of the deep aponeurosis at the 90% location decreases significantly with increasing force levels. The contracted muscle is shorter with greater thicknesses in the proximal and middle regions. Presumably, the deep aponeurosis expanded in the medial-lateral axis. The distal region of the contracted muscle is thinner and the axial segment length of the aponeurosis decreases, accompanied by aponeurosis stretch along the proximal-distal axis.

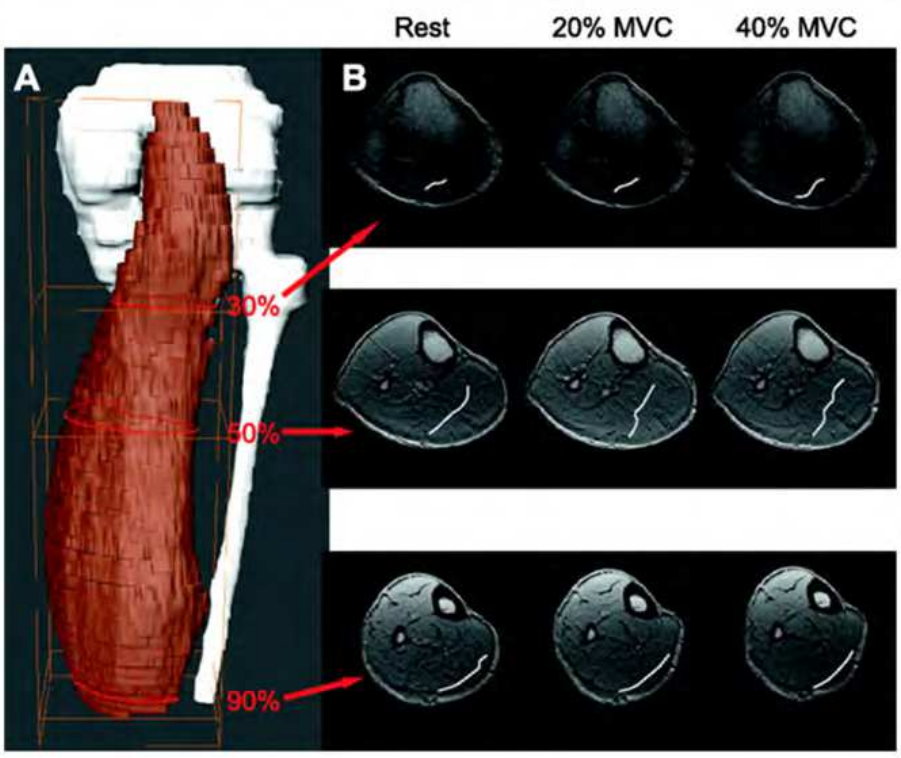


Fig. 4. Changes in cross-sectional segment shape and length of the deep aponeurosis. *A*: 3D reconstructed images of the MG, tibia, and fibula from a stack of axial MR images in 1 representative subject. Red lines correspond to positions at 30%, 50%, and 90% locations along the proximal-distal axis. *B*: axial morphological images at 30%, 50%, and 90% locations as indicated in the 3D image at rest, 20% MVC, and 40% MVC from 1 subject. The deep aponeurosis is indicated by white lines in these axial images and reveals changes in shape and cross-sectional segmental length after force production (from Kinugasa et al. 2008).

3. Adaptations of tendon and aponeurosis under unloading and rehabilitation conditions

3.1 Experimental determination of Achilles tendon compliance after unloading

Muscle atrophy is the process of loss of skeletal muscle tissue typically from disuse or unloading, which can arise from a variety of clinical condition including immobilization during use of cast, prolonged bed rest, and micro-gravity during space-flight. One intriguing characteristic of muscle atrophy is that the resultant decline in force per unit cross sectional area (CSA) is disproportionately larger than the accompanying decrease in muscle volume from atrophy. Our laboratory has reported, for example, muscle volume reduction of 6% accompanied by a force decline of 48% resulting from a 4 week unilateral limb suspension (Lee et al. 2006). Although reduced neural drive and a decrease in single fiber-specific tension are known to contribute to this phenomenon, recent evidence, obtained by MRI and ultrasound, suggests that changes in tendon mechanical properties as well as musculoskeletal architecture also play a role. Tendon and aponeuroses are passive elements of the musculoskeletal system whose main function is to transmit forces and displacement generated by muscle fibers to the bone. If the tendon stiffness is significantly altered, the mechanical consequence is that the passive elements will be required to undergo more strain in order to transmit the same force output compared to the pre-atrophy level.

Several studies have shown that unloading can have a significant negative effect on the mechanical properties of human tendon. Previously, the reduction in stiffness was reported mainly from ultrasound studies in humans (Kubo et al. 2002, 2004; Maganaris et al. 2006; Reeves et al. 2005) as well as in some animal studies (Ameida-Silveira et al. 2000; Woo et al. 1982). Recently, the tendon mechanical properties were measured with the stress-strain relationship established from the VE-PC MRI studies. For these VE-PC MRI studies, the Young's modulus showed a 17% reduction for the Achilles tendon (Figure 5, Shin et al. 2008a) and 29% reduction for the distal aponeurosis (Kinugasa et al. 2010) after 4-wk unloading. The extent of decline found in these studies is similar to that reported by Kubo et al. (2002, 2004), who studied humans subjected to 20 days of bed rest (-28%), but it is much more moderate than the percent decline reported in studies of 90 days simulated microgravity (-58%, Reeves et al. 2005) and paralysis (-59%, Maganaris et al. 2006). This suggests that the extent of stiffness decrease may have a direct correlation to the duration of unloading.

3.2 Physiological implications of changes in tendon compliance

A decrease in Achilles tendon stiffness results in a clear functional disadvantage. Decreased stiffness results in less efficient transfer of contractile force produced by the muscle to the bone, which leads to delayed motor behavior (Proske & Morgan 1987). Additionally, the compliant tendon results in a leftward shift of the force-length curve of the muscle fiber, resulting in a decline in force production for a given amount of fiber shortening (Shin et al. 2008a).

The exact causative mechanism for the reduction in tendon mechanical stiffness remains unclear. It is generally attributed to material deterioration, since the tendon CSA remains unchanged with chronic unloading in humans (de Boer et al. 2007; Kubo et al. 2004; Reeves et al. 2005; Shin et al. 2008a). However, this conclusion necessitates several major presumptions. The results of animal experiments characterizing the effect of chronic unloading on tendon dimension are inconsistent. For example, in animal models, tendon size has been shown to decrease (Schulze et al. 2002), as well as not to change (Almeida-Silveira et al. 2000; Heinemeier et al. 2009; Matsumoto et al. 2003), and even to increase (Kotani et al. 1998; Tsuchida et al. 1997) in response to chronic unloading. Interestingly, if the Achilles tendon of a rabbit is cut, glycosaminoglycan content and fibroblast number increase, and the number of small collagen fibers increase (Flint 1982).

Several human studies (de Boer et al. 2007; Kubo et al. 2004; Reeves et al. 2005; Shin et al. 2008a) have indicated that chronic unloading does not lead to significant change in tendon size, but dimensional measures were limited to a very restricted fraction of the length. Recently, in fact, human tendon size was shown to increase by 5% after 4-wk unloading based on relatively high-resolution data and sensitive procedure to test for significance (Figure 6, Kinugasa et al. 2010). One explanation may arise from further analysis showing that the entire length of the Achilles tendon and distal aponeurosis and median septum remained unchanged in cross-sectional segment lengths. This possibly indicates a slight increase in thickness rather than changes in the overall dimensions within a cross section.

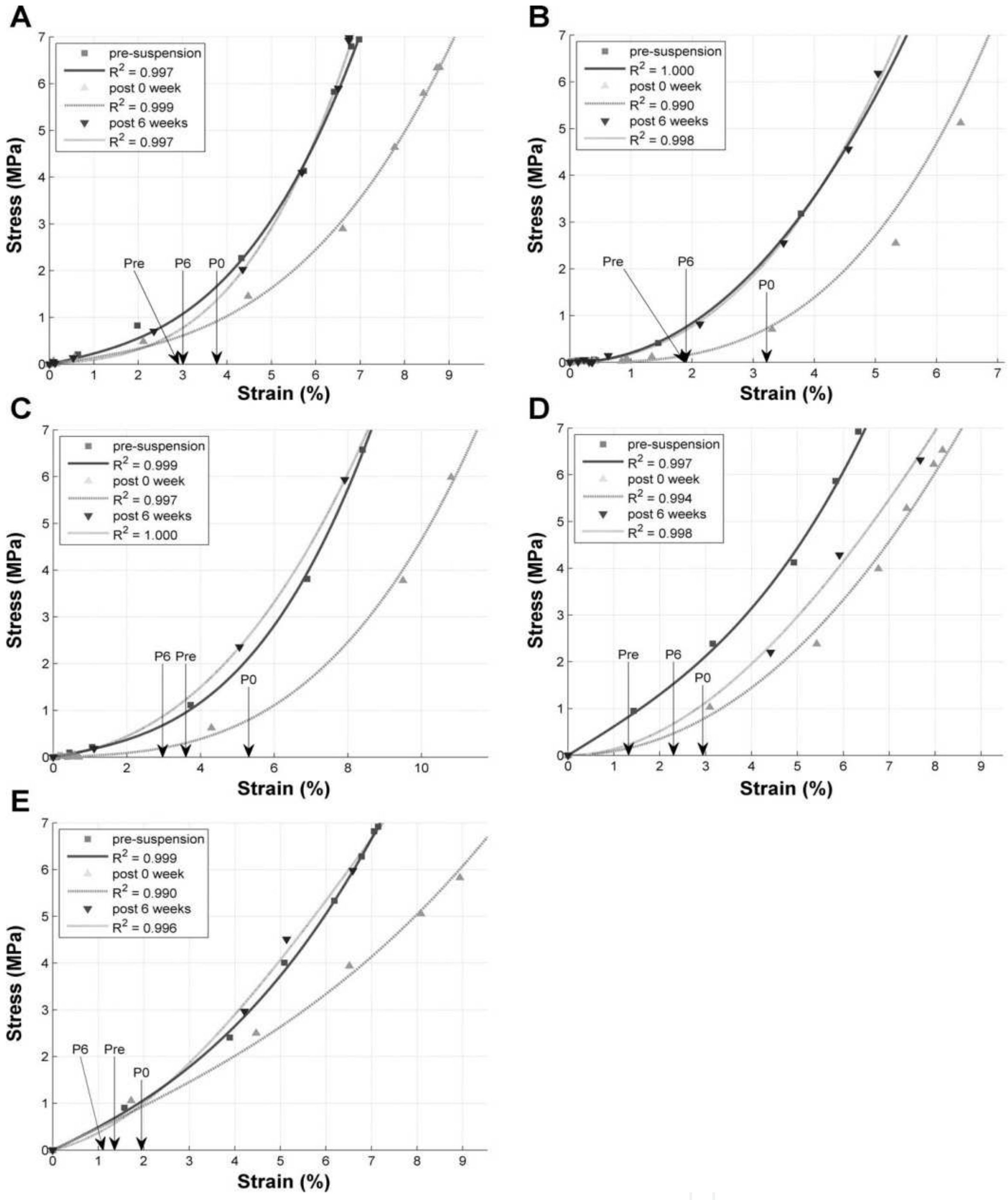


Fig. 5. Effect of Unloading on Stress-strain curve of the Achilles tendon: at pre-suspension (Pre), at post-suspension (P0), and after 6 wk of physical rehabilitation (P6) for five subjects (from Shin et al. 2008a).

This assumption is supported by some relevant findings including an increase in rat collagen fiber proportion (Binkley & Peat 1986) and CSA of ewe spinal ligament (Kotani et al. 1998) as a result of chronic unloading. However, animal data on the effects of chronic unloading on collagen fibril size, density, and number are conflicting. Human studies have demonstrated that, during 2 wk of unloading, there are either no changes or a downregulation of collagen I and III mRNA (Heinemeier et al. 2009) and collagen synthesis

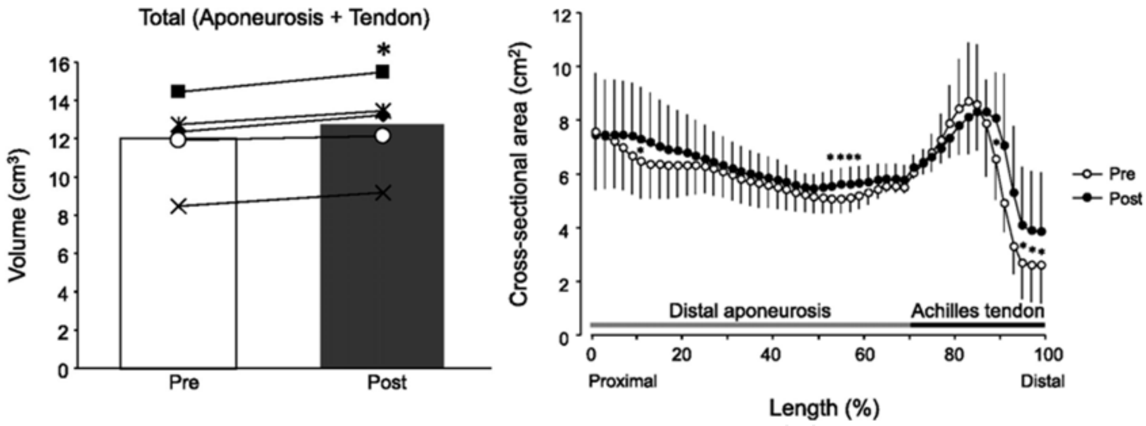


Fig. 6. Effect of 4-wk unilateral lower limb suspension: Changes in volume (left) and cross-sectional area (right) along the entire length of Achilles tendon and distal aponeurosis following 4-wk unilateral lower limb suspension (from Kinugasa et al. 2010).

(Christensen et al. 2008), indicating that the ultrastructure of collagen fibril might not alter with chronic unloading. An increase in the water content in extracellular space may therefore provide a possible explanation for tendon hypertrophy. It is possible that extracellular space could be increased in response to chronic unloading (Kotani et al. 1998, Tsuchida et al. 1997). Although the tendon hypertrophy observed may be expected to compensate for the reduction in the tendon stiffness, the absence of any significant correlation between the magnitude of tendon hypertrophy and reduced Young's modulus (Figure 7) seen in Kinugasa's study (2010) suggests that dimensional factors are not critical to the elastic properties. The literature seems to indicate that the altered tendon elastic modulus is largely due to material deterioration. Changes in the structure and packing of the collagen fibers (Danielsen & Andreassen 1988), such as loss of transverse bands of collagen fiber (Paavola et al. 2002), increased collagen fiber crimping (Patterson-Kane 1997), and reduction in the covalent intramolucular cross-links (Bailey 2001), may generally be considered to be factors in the alteration of tendon material properties.

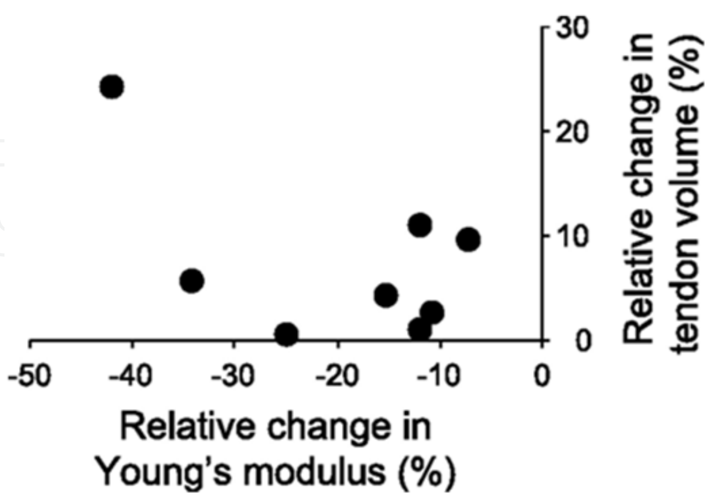


Fig. 7. Relationship between relative changes in volume and Young's modulus of Achilles tendon and distal aponeurosis after 4 wk of unilateral lower limb suspension. The regression line is not shown since the relationship was found to be insignificant (from Kinugasa et al. 2010).

The effects of unloading or disuse were demonstrated by reduction of the muscle force. The recovery of muscle volume and force with a chronic rehabilitation are well-documented in literature (Berg et al. 1991; Lee et al. 2006; Maclntyre et al. 2005), but to our knowledge, there is only one study which investigated changes in tendon mechanical properties with physical rehabilitation. Shin et al. (2008a) investigated the change in the Young's modulus in response to 6 wk of physical rehabilitation (six exercise; 1) warm-up, 2) strength, 3) balance, 4) stretching, 5) cool-down, 6) post-evaluation) after 4 wk of limb suspension. The Young's modulus was decreased by 17% following 4-wk unloading and returned to the pre-loading level at the end of 6-wk of physical rehabilitation (Figure 5).

4. Amplification of ankle rotation by deformation of Achilles tendon

4.1 Experimental determination of amplification factor

In-vivo measurements of muscle shortening during plantarflexion / dorsiflexion movements of the ankle show that the distance moved by the calcaneus exceeds the shortening of the muscle fibers (Hodgson et al. 2006). Thus, mechanisms must exist between the muscle fiber and calcaneus which amplify the muscle fiber length changes. One such system appears to be the internal mechanics of the muscle which prevents changes in aponeurosis separation as the muscle lengthens and shortens. We have hypothesized a second system operating at the ankle. The observed curvature of the Achilles tendon under load indicates the presence of a mechanical constraint close to the ankle which limits posterior movement of the tendon as the ankle rotates. Mechanical analysis of such a constraint suggests that it would modify the relationship between muscle shortening and ankle rotation, adding more amplification to the translation of muscle fiber shortening to ankle rotation (Hodgson et al. 2006). Although no structure is readily apparent, the observation of tendon curvature indicates that a force component perpendicular to the tendon load axis is present to displace the tendon from the linear orientation which it would

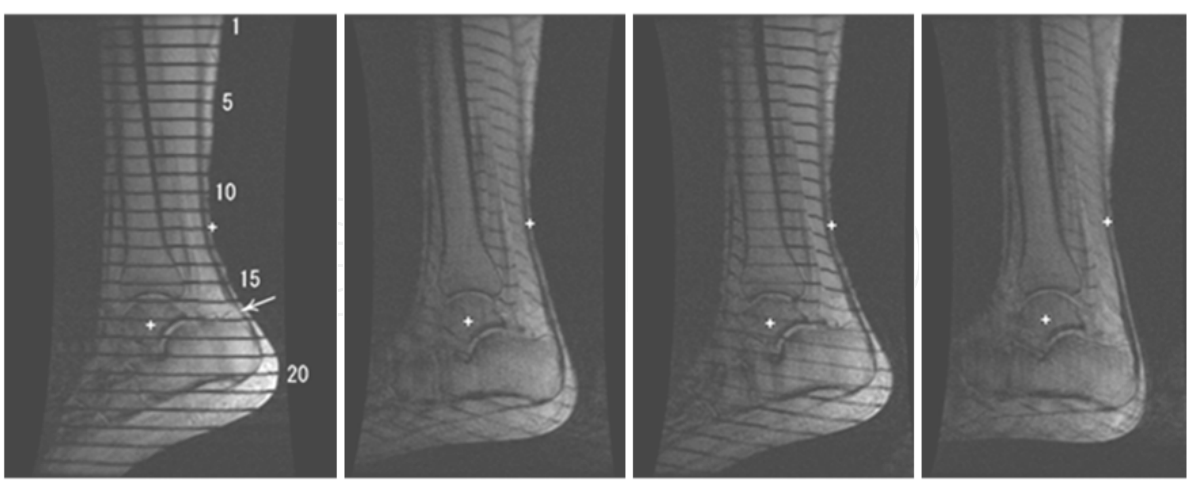


Fig. 8. Spin-tagged sagittal images showing the rotation of the foot during passive dorsiflexion of the foot (left to right): White crosses indicate the ankle center of rotation and the region above the ankle where posterior movement of the tendon appears to be prevented. The first panel identifies the tag-line numbers and the arrow indicates the point where the calcaneus meets the Achilles tendon. Movement of the Achilles tendon and aponeurosis was measured by noting its intersection with each tag line in several frames.

adopt with tensile loading alone. This prediction has been tested by comparing the displacement of the aponeuroses above the ankle and the movement at different points along the length of the Achillles tendon and calcaneous as the subject's foot was moved through ~30^o by the motor at a cycle rate of 30^o s^{-1}, with the subject exerting 40% MVC. On an oblique sagittal image of the lower leg in which the entire length of the aponeurosis and tendon could be visualized, the intersection of spin tag lines in the soleus muscle and the Achilles tendon / aponeurosis was detected, and these points of intersection were tracked in successive frames of the cine-MRI data (Figure 8). Additionally, the location of the apparent deviation of the Achilles tendon from a straight line was noted on each frame.

Several points on the foot were also tracked and used to determine the center of rotation of the ankle. In Fig. 9, which demonstrates the geometry of ankle rotations, Point X is the MG musculo-tendinous junction, Point Y → point where posterior movement of the Achilles tendon is prevented, Point Z → the junction between Achilles tendon and calcaneus, Point C → the ankle center of rotation; Line CZ → lever arm upon which the Achilles tendon acts, Line CB → the Achilles tendon moment arm which is different from CZ because the Achilles

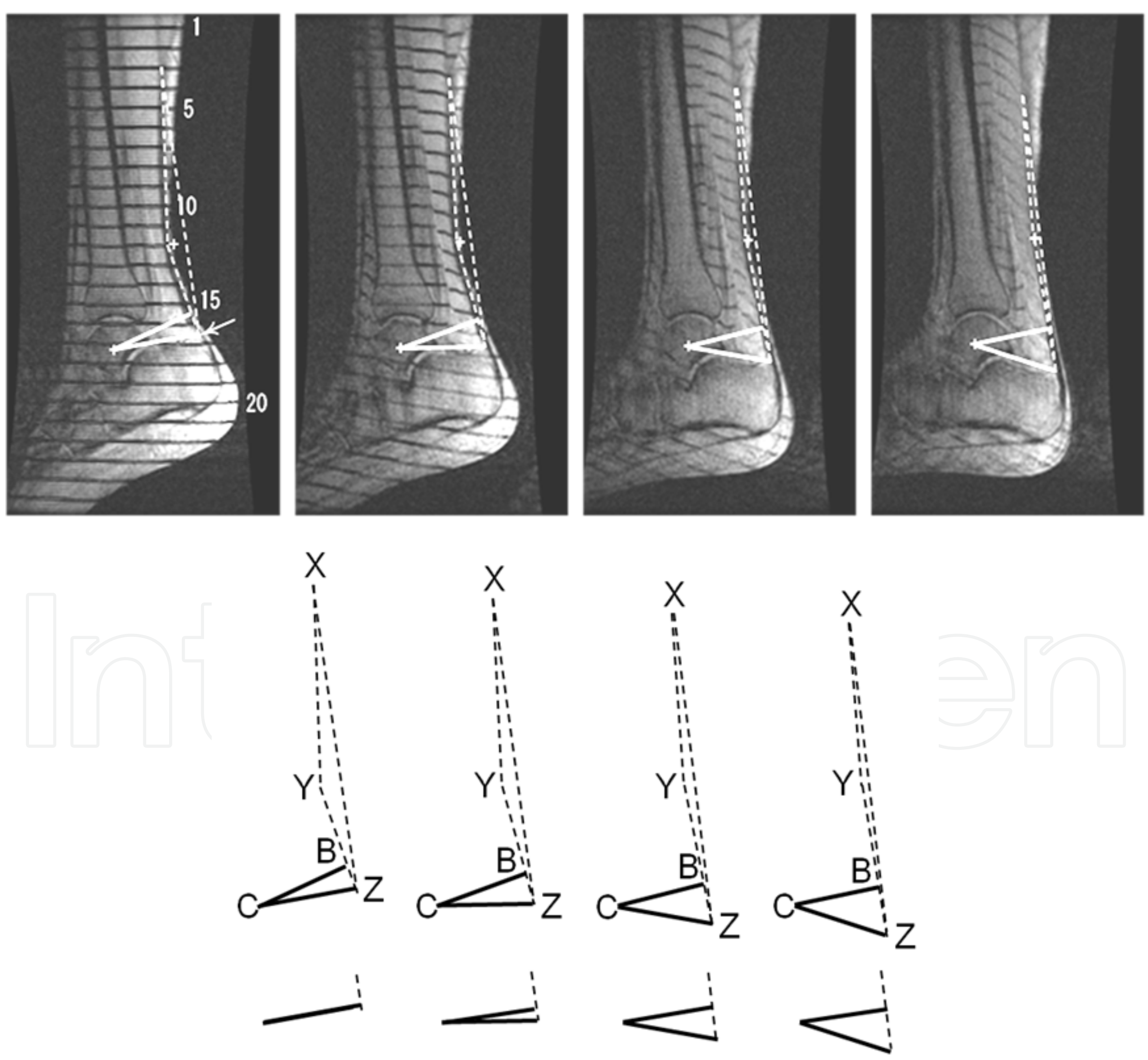


Fig. 9. Geometry of ankle rotation due to restriction of posterior movement of the Achilles tendon.

tendon pulls at an angle to CZ, Line XZ → the distance between the MG musculo-tendinous junction and calcaneus, Lines XY and YZ indicate the path of the Achilles tendon, Length XY+YZ represents the length of the Achilles tendon. The geometry of triangle XYZ implies that line XY lengthens as the ankle dorsiflexes (left to right in figure). Thus point Z moves a greater distance than point X, even if tendon length remains constant. Moment arm BC equates to lever arm length CZ multiplied by Cos angle BCZ. Angle BCZ increases with dorsiflexion, therefore the Achilles tendon moment arm must decrease with dorsiflexion. The lower row of diagrams on the right illustrates the greater change in moment arm which would occur without the Achilles tendon constraint.

4.2 Physiological implications of amplification factor

Figure 10 illustrates that the constraint to the posterior movement of the Achilles tendon reduces the tendon moment arm at the ankle over a wide range of ankle angles. A shorter moment arm increases the ankle rotation for a given change in muscle length. The range of moment arm change is also reduced significantly, possibly easing the challenge of controlling ankle torque at different ankle angles. The calculated lever arm length from the ankle center of rotation to the calcaneo-tendinous junction was 53.1 +/- 3.8 (SD, range 50.4 – 58.6) mm. The actual moment arm was always less than this value due to the anatomical relationship between the ankle and the Achilles tendon. Restriction of the outward movement of the Achilles tendon always resulted in an acute angle between the direction of tendon pull and the line between the ankle center of rotation and the calcaneo-tendinous junction. The moment arm is modified by the sine of this angle and is correctly measured by the minimum distance between a line through the Achilles tendon and the ankle center of rotation (Rugg et al. 1990).

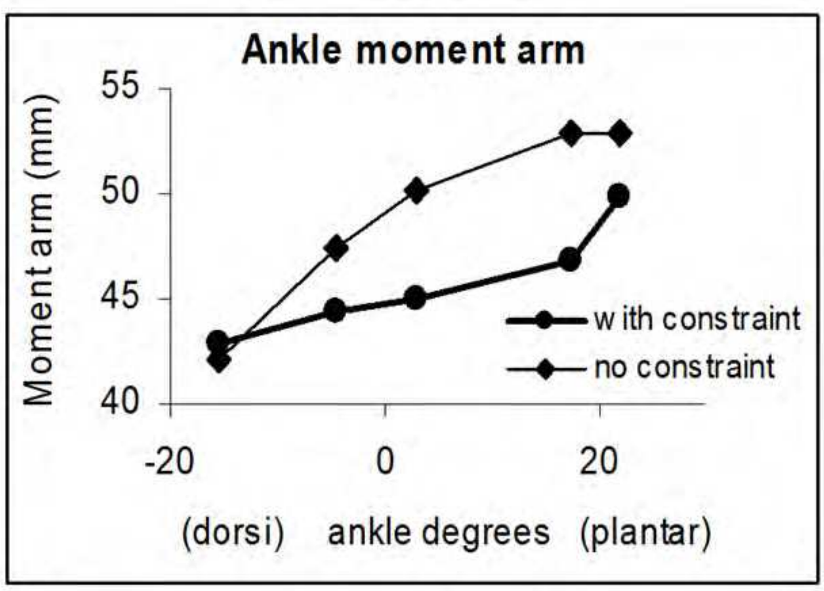


Fig. 10. Achilles tendon moment arm: The moment arm measured from MRI data (with constraint) is shown along with a theoretical moment arm based upon the Achilles tendon lever arm and a constant direction of pull set to the direction measured at full dorsiflexion (no constraint). The lack of constraint would also require the Achilles tendon to move posteriorly by ~1.5 cm.

Figure 11 demonstrates that the tags on the calcaneus have greater displacement relative to the tags on the aponeurosis, illustrating the amplification of tendon movement initially postulated. An unstrained posterior aponeurosis would result in all tag lines displacing by the same amount. Experimental observations confirmed previous observations that the aponeurosis does not move uniformly, exhibiting displacements which suggest some regions of the aponeurosis stretch while other regions undergo compression (Figure 10). Furthermore, some tags on the calcaneus showed a greater displacement than the displacement of the aponeurosis, consistent with the well recognized phenomenon that Achilles tendon stretches under load.

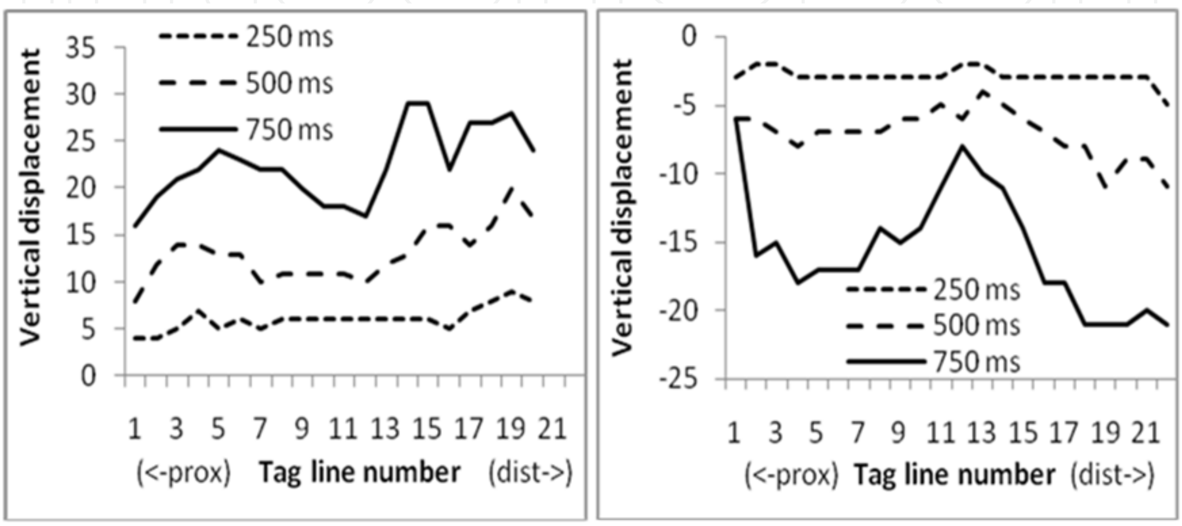


Fig. 11. Vertical displacement of tags at various stages of the plantarflexion (right column) / dorsiflexion cycle (left column). Note the greater displacement of tags on the calcaneus relative to tags on the aponeurosis, illustrating the amplification of tendon movement. Note also the uneven displacement along the aponeurosis, indicating regions of stretch and compression. The line style identifies the delay from the beginning of the movement and shows a progressive displacement of each ROI from the rest position (0). The most proximal region of the muscle is on the left of each graph and the calcaneus locations are on the right.

The alternative explanation for the differences between calcaneus and aponeurosis displacement may be an amplification mechanism due to the constraint on posterior movement of the Achilles tendon. These MRI experiments suggest that the major point of action of this constraint is 61.6 +/- 5.1 (SD, range 52.7 -65.6) mm superior to and 33.5 +/- 6.9 mm posterior to the ankle center of rotation.

The mechanism amplifies the influence of muscle shortening upon ankle rotation and decreases the Achilles tendon moment arm over much of the ankle range of movement. The data indicate that attempts to measure Achilles tendon length changes by measuring the distance between the musculo-tendinous junction and the calcaneus must also take into account the curvature of the tendon and any changes in curvature which arise from a change in ankle angle. Computations of the potential amplification by this configuration suggests a maximum gain of ~1.1 (Hodgson et al. 2006). While apparently quite small, this could have a significant impact on experimentally observed Achilles tendon strain. For example, ultrasound measurements of aponeurosis excursion during a maximum voluntary contraction typically report a displacement of 20 – 30 mm. If a significant proportion of this movement involves rotation of the ankle, this could account for a significant portion of the observed strain.

5. Conclusion

It is widely recognized that our muscular system changes its properties in response to altered mechanical loading. Previous studies have documented deleterious changes in muscle following chronic unloading, while recent studies provide information of significant changes in tendinous structures within the muscle. Recent work in our laboratory (Finni et al. 2003) in addition to work of others (Gans 1982) has demonstrated a significance of muscular structural organization. We believe the integrated structure of muscle and tendon is specifically tuned to the normal operating environment and normal physiological state. A better understanding of this interaction between muscle and tendon and their co-dependence in maintaining normal physiological function is essential, as is a better understanding of how unloading and atrophy disrupt normal function. One of the main objectives of our group's research has been to investigate the effects of chronic unloading on the mechanical properties of tendinous structures in in-vivo human skeletal muscles and furthermore, to gain insight into the possible mechanism(s) of undesirable adaptations of human muscular-tendinous structures occurring secondary to decrease in mechanical loading.

6. Acknowledgement

This study was supported in part by National Institute of Arthritis and Musculoskeletal and Skin Diseases Grant 2R01-AR-53343-05A1, USA.

7. References

Alexander, RM, Bennet-Clark HC. Storage of elastic strain energy in muscle and other tissues. *Nature* 265: 114–117, 1977.

Alexander RM. The spring in your step: the role of elastic mechanisms in human running. Biomechanics XI-A (eds. De Groot G, Hollander AP, Huijing PA, and van Ingen Schenau GJ). Free University Press, Amsterdam, 17–25, 1988a.

Alexander RM. Elastic Mechanisms in Animal Movement. Cambridge University Press, Cambridge, 1988b.

Arampatzis A, Stafilidis S, DeMonte G, Karamanidis K, Morey-Klapsing G, Brüggemann GP. Strain and elongation of the human gastrocnemius tendon and aponeurosis during maximal plantarflexion effort. *J Biomech* 38: 833–841, 2005.

Almeida-Silveira MI, Lambertz D, Perot C, Goubel F. Changes in stiffness induced by hindlimb suspension in rat Achilles tendon. *Eur J Appl Physiol* 81: 252–257, 2000.

Bailey AJ. Molecular mechanisms of ageing in connective tissues. *Mech Ageing Dev* 122: 735–755, 2001.

Berg HE, Dudley GA, Haggmark T, Ohlsen H, Tesch PA. Effects of lower limb unloading on skeletal muscle mass and function in humans. *J Appl Physiol* 70: 1882–1885, 1991.

Binkley JM, Peat M. The effects of immobilization on the ultrastructure and mechanical properties of the medial collateral ligament of rats. *Clin Orthop Relat Res* 203: 301–308, 1986.

Bojsen-Møller J, Hansen P, Aagaard P, Svantesson U, Kjaer M, Magnusson SP. Differential displacement of the human soleus and medial gastrocnemius aponeuroses during isometric plantar flexor contractions in vivo. *J Appl Physiol* 97: 1908–1914, 2004.

Cavagna GA, Saibene FP, Margaria R. Mechanical work in running. *J Appl Physiol* 19: 249–256, 1964.

Christensen B, Dyrberg E, Aagaard P, Kjaer M, Langberg H. Short-term immobilization and recovery affect skeletal muscle but not collagen tissue turnover in humans. *J Appl Physiol* 105: 1845–1851, 2008.

Danielsen CC, Andreassen TT. Mechanical properties of rat tail tendon in relation to proximal-distal sampling position and age. *J Biomech* 21: 207–212, 1988.

de Boer MD, Maganaris CN, Seynnes OR, Rennie MJ, Narici MV. Time course of muscular, neural and tendinous adaptations to 23 day unilateral lower-limb suspension in young men. *J Physiol* 583: 1079–1091, 2007.

Finni T, Hodgson JA, Lai AM, Edgerton VR, Sinha S. Mapping of movement in the isometrically contracting human soleus muscle reveals details of its structural and functional complexity. *J Appl Physiol* 95: 2128–2133, 2003.

Flint M. Interrelationships of mucopolysaccharide and collagen in connective tissue remodelling. *J Embryol Exp Morphol* 27: 481–495, 1972.

Gans C. Fiber architecture and muscle function. *Exer Sports Sci Rev* 10: 160–207, 1982.

Heinemeier KM, Olesen JL, Haddad F, Schjerling P, Baldwin KM, Kjaer M. Effect of unloading followed by reloading on expression of collagen and related growth factors in rat tendon and muscle. *J Appl Physiol* 106*:* 178–186*,* 2009.

Herzog W. Tendon/Aponeurosis. Biomechanics of the Musculo-skeleltal System (eds. Nigg BM and Herzog W). Wiley, England, 146–168, 2007.

Hodgson JA, Finni T, Lai AM, Edgerton VR, Sinha S. Influence of structure on the tissue dynamics of the human soleus muscle observed in MRI studies during isometric contractions. *J Morphol* 267: 584–601, 2006.

Hodgson JA, Shin D, Nagarsekar G, Edgerton VR, Sinha. Amplification of Achilles Tendon Displacement by a Pivot-like Restriction Amplifies Final Displacement of Calcaneous and Rotation of Ankle with Possible Impact on Measured Strain. #6871, Presented at 17th Intl. Soc Mag Res Med Mtng. in Hawaii, 2009.

Hof AL, Van Zandwijk JP, Bobbert MF. Mechanics of human triceps surae muscle in walking, running and jumping. *Acta Physiol Scand* 174: 17–30, 2002.

Hooley CJ, McCrum NG, Cohen RE. The viscoelastic deformation of tendon. J *Biomech* 13: 521–528, 1980.

Iwanuma S, Akagi R, Kurihara T, Ikegawa S, Kanehisa H, Fukunaga T, Kawakami Y. Longitudinal and transverse deformation of human Achilles tendon induced by isometric plantar flexion at different intensities. *J Appl Physiol* 110: 1615–1621, 2011.

Ker RF, Alexander RM, Bennett MB. Why are mammalian tendons so thick? *J Zool* 216: 309–324, 1988.

Kinugasa R, Hodgson JA, Edgerton VR, Shin DD, Sinha S. Reduction in tendon elasticity from unloading is unrelated to its hypertrophy. *J Appl Physiol* 109: 870–877, 2010.

Kinugasa R, Shin D, Yamauchi J, Mishra C, Hodgson JA, Edgerton VR, Sinha S. Phase-contrast MRI reveals mechanical behavior of superficial and deep aponeuroses in human medial gastrocnemius during isometric contraction. *J Appl Physiol* 105: 1312–1320, 2008.

Kotani Y, Cunningham BW, Cappuccino A, Kaneda K, McAfee PC. The effects of spinal fixation and destabilization on the biomechanical and histologic properties of spinal ligaments. An in vivo study. *Spine* 23: 672–683, 1998.

Kubo K, Akima H, Ushiyama J, Tabata I, Fukuoka H, Kanehisa H, Fukunaga T. Effects of 20 days of bed rest on the viscoelastic properties of tendon structures in lower limb muscles. *Br J Sports Med* 38: 324–330, 2004.

Kubo K, Kawakami Y, Kanehisa H, Fukunaga T. Measurement of viscoelastic properties of tendon structures in vivo. *Scand J Med Sci Sports* 12: 3–8, 2002.

Lee HD, Finni T, Hodgson JA, Lai AM, Edgerton VR, Sinha S. Soleus aponeurosis strain distribution following chronic unloading in humans: an in vivo MR phase-contrast study. *J Appl Physiol* 100: 2004–2011, 2006.

Lieber RL, Leonard ME, Brown CG, Trestik CL. Frog semitendinosis tendon load-strain and stress-strain properties during passive loading. *Am J Physiol* 261: C86–92, 1991.

Lieber RL, Leonard ME, Brown-Maupin CG. Effects of muscle contraction on the load-strain properties of frog aponeurosis and tendon. C*ells Tissues Organs* 166: 48–54, 2000.

MacIntyre DL, Eng JJ, Allen TJ. Recovery of lower limb function following 6 weeks of non-weight bearing. *Acta Astronaut* 56: 792–800, 2005.

Maganaris CN, Kawakami Y, Fukunaga T. Changes in aponeurotic dimensions upon muscle shortening: in vivo observations in man. *J Anat* 199: 449–456, 2001.

Maganaris CN, Paul JP. Tensile properties of the in vivo human gastrocnemius tendon. *J Biomech* 35: 1639–1646, 2002.

Maganaris CN, Reeves ND, Rittweger J, Sargeant AJ, Jones DA, Gerrits K, De Haan A. Adaptive response of human tendon to paralysis. *Muscle Nerve* 33: 85–92, 2006.

Maganaris CN, Narici MV, Maffulli N. Biomechanics of the Achilles tendon. *Disabil Rehabil* 30: 1542–1547, 2008.

Matsumoto F, Trudel G, Uhthoff HK, Backman DS. Mechanical effects of immobilization on the Achilles' tendon. *Arch Phys Med Rehabil* 84: 662–667, 2003.

McNeilly CM, Banes AJ, Benjamin M, Ralphs JR. Tendon cells in vivo form a three dimensional network of cell processes linked by gap junctions. *J Anat* 189: 593–600, 1996.

Morgan DL, Proske U, Warren D. Measurements of muscle stiffness and the mechanism of elastic storage of energy in hopping kangaroos. *J Physiol* 282: 253–261, 1978.

Muramatsu T, Muraoka T, Takeshita D, Kawakami Y, Hirano Y, Fukunaga T. Mechanical properties of tendon and aponeurosis of human gastrocnemius muscle in vivo. *J Appl Physiol* 90: 1671–1678, 2001.

Oxorn VM, Agur AM, McKee NH. Resolving discrepancies in image research: the importance of direct observation in the illustration of the human soleus muscle. *J Biocommun* 25: 16–26, 1998.

Paavola M, Kannus P, Jarvinen TA, Khan K, Jozsa L, Jarvinen M. Achilles tendinopathy. *J Bone Joint Surg Am* 84*:* 2062–2076, 2002.

Patterson-Kane JC, Wilson AM, Firth EC, Parry DA, Goodship AE. Comparison of collagen fibril populations in the superficial digital flexor tendons of exercised and nonexercised thoroughbreds. *Equine Vet J* 29*:* 121–125, 1997.

Proske U, Morgan DL. Tendon stiffness: methods of measurement and significance for the control of movement. *J Biomech* 20: 75–82, 1987.

Purslow PP. The structure and functional significance of variations in the connective tissue within muscle. *Comp Biochem Physiol A Mol Integr Physiol* 133: 947–966, 2002.

Ralphs JR, Waggett AD, Benjamin M. Actin stress fibres and cell-cell adhesion molecules in tendons: organisation in vivo and response to mechanical loading of tendon cells in vitro. *Matrix Biol* 21: 67–74, 2002.

Reeves ND, Maganaris CN, Ferretti G, Narici MV. Influence of 90-day simulated microgravity on human tendon mechanical properties and the effect of resistive countermeasures. *J Appl Physiol* 98: 2278–2286, 2005.

Reeves NJ, Maganaris CN, Ferretti G, Narici MV. Influence of simulated microgravity on human skeletal muscle architecture and function. *J Gravit Physiol* 9: 153–154, 2002.

Rigby BJ, Hirai N, Spikes JD, Eyring H. The Mechanical Properties of Rat Tail Tendon. *J Gen Physiol* 43: 265–283, 1959.

Rugg SG, Gregor RJ, Mandelbaum BR, Chiu L. *J Biomech* 23: 495–501, 1990.

Schulze K, Gallagher P, Trappe S. Resistance training preserves skeletal muscle function during unloading in humans. *Med Sci Sports Exerc* 34*:* 303–313, 2002.

Scott SH, Loeb GE. Mechanical properties of aponeurosis and tendon of the cat soleus muscle during whole-muscle isometric contractions. *J Morphol* 224: 73–86, 1995.

Shin D, Finni T, Ahn S, Hodgson JA, Lee HD, Edgerton VR, Sinha S. Effect of chronic unloading and rehabilitation on human Achilles tendon properties: a velocity-encoded phase-contrast MRI study. *J Appl Physiol* 105: 1179–1186, 2008a.

Shin D, Finni T, Ahn S, Hodgson JA, Lee HD, Edgerton VR, Sinha S. In vivo estimation and repeatability of force-length relationship and stiffness of the human achilles tendon using phase contrast MRI. *J Magn Reson Imaging* 28: 1039–1045, 2008b.

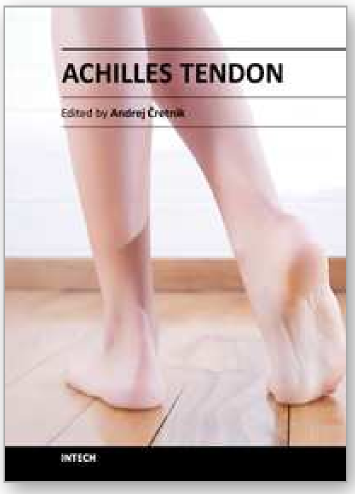
Tsuchida T, Yasuda K, Kaneda K, Hayashi K, Yamamoto N, Miyakawa K, Tanaka K. Effects of in situ freezing and stress-shielding on the ultrastructure of rabbit patellar tendons. *J Orthop Res* 15: 904–910, 1997.

Wainwright SD, Biggs WD, Currey JD, Gosline JM. Mechanical design in organisms. Princeton University Press, Princeton, NJ, 88–93, 1982.

Woo SL, Gomez MA, Woo YK, Akeson WH. Mechanical properties of tendons and ligaments. II. The relationships of immobilization and exercise on tissue remodeling. *Biorheology* 19: 397–408, 1982.

Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit Rev Biomed Eng* 17: 359–411, 1989.

Zuurbier CJ, Huijing PA. Influence of muscle geometry on shortening speed of fibre, aponeurosis and muscle. *J Biomech* 25: 1017–1026, 1992.



Achilles Tendon

Edited by Prof. Andrej Cretnik

ISBN 978-953-51-0264-9
Hard cover, 144 pages
Publisher InTech
Published online 08, January, 2012
Published in print edition January, 2012

Achilles tendon has always attracted a great attention. Its disorders include various problems from pain and swelling with bumps to functional impairment or even ruptures. Debates concerning aetiology and optimal treatment are still going on. A lot of efforts and research have already been put on to find the answers to unsolved problems and this book is an attempt to share (some of) these findings to the readers. If only one of the papers helps the therapists or patients in understanding and solving their problems, we will consider that the mission of the book was accomplished.

InTech Europe
University Campus STeP Ri
Slavka Krautzeka 83/A
51000 Rijeka, Croatia
Phone: +385 (51) 770 447
Fax: +385 (51) 686 166
www.intechopen.com

InTech China
Unit 405, Office Block, Hotel Equatorial Shanghai
No.65, Yan An Road (West), Shanghai, 200040, China
中国上海市延安西路65号上海国际贵都大饭店办公楼405单元
Phone: +86-21-62489820
Fax: +86-21-62489821