● *Original Contribution*

VISUALIZING TENDON ELASTICITY IN AN *EX VIVO* PARTIAL TEAR MODEL

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Abstract—Supersonic shear imaging (SSI) is evaluated as a means of visualizing changes in regional tendon elasticity caused by partial tears in a porcine model. Thirty digital flexor tendons were cut to 25% (n = 10), 50% (n = 10) and 75% (n = 10) of the tendon thickness along the deep surface. Tendon elasticity was mapped left of, centered on and right of the tear site before and after tearing from 0% to 2% strain. Shear wave speed increased at 1% ($p < 0.05$) and 2% ($p < 0.001$) strain for all regions. Deep surface shear wave speed decreased in the 25%, 50% and 75% tears ($p < 0.05$ and $p < 0.001$). Computational tendon tear models were also created to investigate regional changes in strain resulting from a tear. In the computational models, strain on the deep surface decreased progressively with increasing tear size. Visualization of tendon shear wave speed was achieved in normal and partially torn tendons, indicating the potential of SSI to add tendon shear wave speed to traditional morphologic assessment of partial tears, which may improve assessment of tendon health. (E-mail: dewall@wisc.edu or KLee2@uwhealth.org)   © 2014 World Federation for Ultrasound in Medicine & Biology.

Key Words: Partial tear, Shear wave imaging, Tendon rupture, Ultrasound elastography.

INTRODUCTION

Tendons facilitate movement, stabilize joints and absorb impact, but these functions can be impaired with partial tears, which commonly occur in the rotator cuff and Achilles tendon (Clayton 2008; Tashjian 2012). Partial tears weaken tendons and can progress to rupture. Tendon rupture is often preventable with physical therapy or surgery, but there are currently no quantitative metrics to guide treatment strategy. The "50% rule" is a semi-objective practice that states that surgical repair is generally required for a 50% or greater partial-thickness tear (Pedowitz et al. 2011), but has little scientific basis. Tendon mechanical properties (*e.g.,* tendon strain) can be measured with B-mode ultrasound (US) or magnetic resonance imaging by tracking dynamic tendon motion (Maganaris 2002; Reeves 2006; Reeves et al. 2005), which may provide additional information for assessing tendon health. However, when evaluating tendons in the resting state (*i.e.,* statically), B-mode US and magnetic resonance imaging assess only tendon morphology.

Quantitative parameters such as tendon elasticity, a tissue mechanical property, may provide more relevant and reliable information for guiding treatment strategies that is not provided by morphologic assessment alone. Some studies have used mechanical testing techniques to quantify tendon strain in normal and damaged states (Mazzocca et al. 2008). However, the invasive nature of many mechanical testing methodologies precludes their use in a clinical setting.

Studies have used US to non-invasively quantify tendon mechanics *in vivo*. With the use of B-mode US imaging, tendon motion has been tracked during tendon stretch to measure local tendon mechanics. By tracking the motion of the musculotendinous junction, local measurements of tendon strain have been obtained *in vivo* (Maganaris 2002). This method has revealed changes in tendon mechanical properties with age and tendon loading conditions (*e.g.,* bed rest, resistance training) (Reeves 2006; Reeves et al. 2005).

Tendon mechanical properties have also been measured with US elastography. Strain imaging, one form of US elastography (Ophir et al. 1991), measures tissue strain by comparing US images before and after tendon compression, typically using the US transducer as the compression device. Studies have used this technique to measure strain in patients with lateral

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epicondylitis and Achilles tendinopathy, revealing tendon softening in patients relative to healthy control patients (De Zordo et al. 2009, 2010).

Supersonic shear imaging (SSI), a form of shear wave imaging (Bercoff et al. 2004), may also hold promise. SSI pushes the tissue of interest with focused US beams, generating a shear wave. This wave is tracked with ultra-fast US data acquisition to estimate its speed (v_s), which is proportional to the shear modulus (μ) of the tissue, a measure of tissue elasticity ($\mu = \rho v_s^2$, where ρ is material density and is assumed constant). Thus, SSI can measure tissue elasticity without applied tendon stretch or compression. SSI has shown promise in breast (Athanasiou et al. 2010), liver (Bavu et al. 2011) and prostate (Barr et al. 2012) applications, but the utility of this technique has not been thoroughly investigated in tendons.

Recently, some studies have used SSI to measure the elasticity of tendons. *In vivo* Achilles tendon elasticity has been measured (Arda et al. 2011; Aubry et al. 2013) and, in one study, was shown to be dependent on ankle angle (Aubry et al. 2013). SSI has also been used to investigate differences in Achilles tendon elasticity in patients with a complete rupture and in healthy control patients (Chen et al. 2013). Few controlled *ex vivo* studies have been performed. The repeatability of SSI in normal deer tendons through a range of loads has been reported (Peltz et al. 2013). However, no studies have investigated SSI in an *ex vivo* tendon injury model. In this study, we apply this technique to an *ex vivo* porcine tendon tear model at different levels of strain to visualize regional changes in shear wave speed caused by partial tendon tears.

METHODS

Specimen preparation

Thirty digital flexor tendons were dissected from porcine lower limbs acquired from a local abattoir. The tendons were cleaned of muscle and connective tissue and removed from the distal bone insertion site. After dissection, the tendons were wrapped in gauze soaked in phosphate-buffered saline solution (40 parts NaCl, 5.75 parts Na_2HPO_4, 1 part KCl, 1 part KH_2PO_4), covered in aluminum foil and sealed in plastic bags that were stored in a $-30°C$ freezer (Duenwald et al. 2009). Tendons were brought to room temperature before testing. Specimens were kept hydrated in physiologic saline throughout preparation.

Experimental procedure

Before testing, the length of the tendon was measured (mean = 72.18 ± 6.05 mm), and an indelible mark was made at the tendon midpoint. The width and thickness of the unloaded specimen were measured three times at this point with an electronic digital caliper and averaged. The average tendon dimensions were 6.18 ± 0.48 mm for width and 4.04 ± 0.39 mm for thickness. Three groups of tendons were imaged in the normal and cut states. The following cut depths were tested: 25% (n = 10), 50% (n = 10) and 75% (n = 10) of tendon thickness. To make the cut, the cut depth was marked on a Thomas blade using digital calipers and a marker. The cut was then carefully made to the specified depth along the deep surface of the tendon.

Specimens were loaded into wedge grips in a tensile mechanical testing system (ESM301 L, Mark-10, Copiague, NY, USA). The distal and proximal ends of the tendon were gripped by the top and bottom grips, respectively (Fig. 1). Load was measured with a 50-lb load cell (M5-50, Mark-10), and displacement was controlled by the ESM301 L test stand. Tendons were slowly lengthened until a 1.25-N pre-load was maintained. The grip-to-grip length was measured and used as the initial length (L_0) for testing, or 0% strain (L/L_0).

It is well known that the stress-strain curve of tendons contains two distinct nearly elastic regions (Fung 1993). The toe region (typically less than 5% strain) exhibits strain-stiffening behavior as the collagen fibers uncrimp. After the collagen fibers are stretched and aligned, the tendon stress-strain curve enters the linear region, where stiffness remains relatively constant with further elongation (*i.e.,* increases in strain) (Hansen et al. 2002). Maganaris and Paul (1999) concluded that *in vivo* physiologic loading for the tibialis anterior tendon lies within the toe region. We therefore applied small strains (1% and 2%) to stay within the toe region. A given tendon was preconditioned from 0% to 2% strain at 20 mm/min for 10 cycles and rested for 1000 s to allow for viscoelastic recovery (Duenwald et al. 2009). The acrylic box surrounding the test stand was then filled with phosphate-buffered saline solution, and the tendon was imaged through the acoustic window, as illustrated in Figure 1.

Shear wave speed maps were acquired with SSI on an Aixplorer US scanner using an SL 15-4 linear array transducer (Supersonic Imagine, Aix-en-Provence, France). Shear wave speed was estimated from two shear waves propagating in a 10×10-mm region of interest (ROI), with one wave generated on the left side and one on the right side of the ROI, which allowed shear wave speed to be estimated on either side of the cut. A directional filter was used to remove reflected waves (Deffieux et al. 2011).

First, normal tendons were imaged with SSI at 0% strain (L_0) at the following settings: shear wave pulse = penetration, spatial smoothing = 7 and temporal averaging (persistence) = low. SSI data were acquired in a ROI centered on the indelible mark made at the tendon midpoint. After the ROI was interactively positioned under