CONTROLLED MOTION STRAIN MEASUREMENT USING LATERAL SPECKLE TRACKING IN ACHILLES TENDONS DURING HEALING

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ABSTRACT

Ultrasound elastography imaging is being increasingly used to indicate the functional properties of musculoskeletal pathologies. Usually, this involves a real-time free-hand acquisition which has questionable consistency and does not provide a material deformation that is representative of the material function.

This paper presents a novel method for acquiring quantitative ultrasound lateral strain images of in-vivo Achilles tendons in a controlled manner using a computer controlled foot articulation device and a novel 2D multiscale, multi-compression, sub-pixel precision block matching algorithm. The method has been applied to a longitudinal study of 21 patients with Achilles tendon rupture which have repeated data points taken over 24 weeks of therapy.

The controlled lateral motion results are compared to the more conventional freehand axial-decompression method, and also against clinical measures.

The results show that the controlled lateral motion strain images have a smaller standard deviation of average value than the freehand decompression motion (σ_{lat} = 0.77%, σ_{comp} =1.74%) and a tighter fitting to a strain improvement model which followed the same form as clinical measures of patient healing over the group (R^2_{lat} = 0.78, R^2_{comp} = 0.34).

Index Terms— *Ultrasound, elastography, strain, block matching, tendon, Achilles, rupture*

1. INTRODUCTION

The Achilles tendon (AT) is the largest tendon in the human body and is the primary energy storage flexor tendon used during walking and running. Damage to the AT can be at the cellular level or can be a macroscopic tear or full rupture, usually as an activity-related injury.

Due to low vasculature in the tendon, healing of full rupture takes a long time, often over six months from rupture to return to sport. During this healing period, there are several different clinical desires to be able to accurately monitor the progress of tendon repair: for measuring the response of groups of patients in clinical trials of new treatments[1]; for evaluating the response of an individual patient to a treatment; and for diagnosis of conditions, particularly in elite athletes. Existing non-invasive techniques to assess the AT healing progress are to take spatial measurements of the tendon geometry in ultrasound images, assess functionality based on a regime of motions and surveys of patients on their capabilities and pain response such as the Victorian Institute of Sports Assessment-Achilles (VISA-A) score[2].

Freehand ultrasound (US) elastography has been used in non-quantitative ways to aid clinicians in assessment of AT damage as an indicative visual tool. Attempts have also been made to make quantitative assessments by comparison of the measured strain values in surrounding fat or muscle material using 'live' elastography functions existing in commercial systems[3]. These measurements are unsatisfactory as the forcing and displacement is arbitrary, are of low, not functionally representative magnitude, and strains are only measured in the direction of the transducer axis which is not the functional direction of the highly anisotropic tendon (see Figure 1)

Other methods for measuring lateral strain *in-vivo* include displacement tracking of the muscle tendon junction (MTJ) with US images[4] and the implanting of tantalum beads during surgery (for rupture patients) which can then be tracked in three dimensions using CT imaging[5]. Both of these studies were conducted under known loading conditions. These methods also lack potential to be developed much further as a sophisticated technique as they only provide point displacement measurements and thus only average tendon strain.

This paper reports on the potential viability of a method to perform two dimensional ultrasound strain imaging of damaged AT strained in the transducer-lateral direction under controlled motion and using a multiscale, multiframe block matching displacement tracking algorithm with sub-pixel precision. The displacement is then used to produce 2D strain images and the average strain of the pathological area is calculated. This process is repeated at intervals to confirm that the measured average strain decreased over the treatment period of 24 weeks. The consistency of these lateral strain measurements has been compared against those taken with a more conventional manual decompression motion. These results were also compared with the clinical VISA survey results of the patients.

2. METHODOLOGY

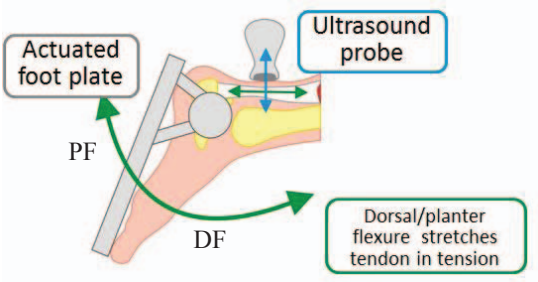


Figure 1. Foot motion types conducted during ultrasound analysis of damaged tendon for strain analysis.

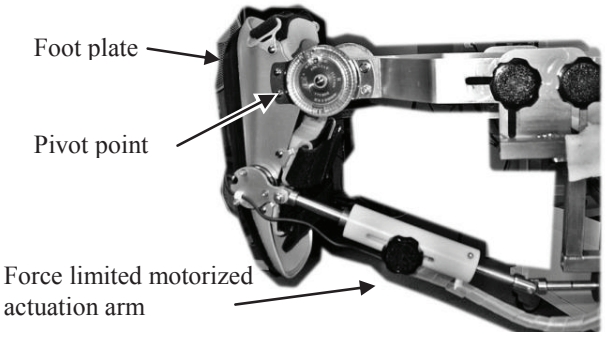


Figure 2. Photograph of foot articulation device

2.1 Patient group

A total of 21 patients were recruited for this study, of which 10 were treated with surgery and the remainder without. The clinical study's primary purpose was to investigate the effect of platelet rich plasma (PRP) on tendon healing, but for the purposes of this paper patient treatments were considered similar. All patients were in the age range 20-50 and had complete AT rupture due to sport related injury. All patients were otherwise fit and well. Ethics approval has been granted and can further details can be viewed by the International Standard Randomised Controlled Trial Number Register ISRCTN93608625.

2.2 Data Acquisition Method

In order to generate a repeatable and controlled transducer-lateral motion in the AT a computer controlled foot clamp was used to manipulate each patient's foot (Figure 2). This allowed a set amount of motion with less than 5mm heel motion to be guaranteed and also had a force-measured safety release to ensure patients were not over exerted.

Each patient had the rupture area manually located using US B-mode imaging in the sagittal plane by the clinician. Each patient was subjected to three types of motion for strain analysis: manual (uncontrolled) compression and decompression with the US probe in the probe-axial direction (pressing down and releasing); lateral motion by moving the foot in dorsal flexure (DF), equivalent to forced toe-lifting from a neutral position; and pushing the foot down in plantar flexure (PF) equivalent to forced toe-pointing as shown in Figure 1.

Each patient had two US sequences captured for each motion type collected. The patients had subsequent sequences captured at intervals throughout their treatment regime, notionally at weeks 0, 1, 3, 6, 8, 12 and 24 since rupture. In addition to strain measures each patient filled out a VISA questionnaire to assess their progress at each time point. This accumulated to over 800 ultrasound image sequences.

2.3 Equipment and data

The ultrasound imaging system was a Zonare z.one (ZONARE Medical Systems, Inc, Mountain View, CA) with a L14-5 linear array transducer with 55mm wide field of view and central frequency of 12MHz. The probe characteristics were such as to generate pixel resolutions of approximately 0.04mm x 0.2mm (depth x width).

The data was stored in a demodulated in-phase and quadrature (IQ) format from which the RF (or A-line) signal from each transducer element could be reconstructed. The decompression sequences were typically 5-8 frames long and the lateral sequences 16 frames long, sampling at approximately 30fps.

2.4 Displacement imaging algorithm

The displacement tracking method used was a Matlab implementation using the same algorithm developed by this group for use of 2D displacement tracking for strain imaging for breast cancer detection over an ultrasound image sequence [6]. It comprises of the following features:

Block matching: 2D image windows 1mm deep by 3mm wide of the first (reference) frame were taken and correlation found over a search range in the subsequent (search) frame. The position with the least mean squared error was found for each direction and the displacement recorded. This was then repeated to cover the whole image area. The displacements were found with a 90% window overlap in both directions so that the resultant motion and strain images are at a lower resolution than the original ultrasound image. This aided computation time with an acceptably small loss of information.

2D Spline representation: In order to obtain sub-pixel precision, the search region in the comparison frame was regenerated as a series of continuous spline functions (one function for each line, either horizontal or vertical depending on search direction) as described in [7]. The minimum least square error was then found efficiently using the standard Matlab numerical solver functions (the Levenberg-Marquardt algorithm was used in this implementation), reducing the quantization error. This has been of great significance for lateral tracking where the horizontal pixel size is determined by the transducer element size and spacing rather than the US frequency.

Multiscale: The displacement was estimated at 3 different image resolutions for the vertical displacement estimation- two down sampled B-mode resolutions and one using the radio frequency (RF) signal- and at two B-mode resolutions for the lateral estimation.

For the lower resolutions, the B-mode envelope of the RF signal was generated, blurred with a 2D Gaussian filter and every N horizontal pixel lines taken, where N is the down sample factor.

The calculated distances in the lower resolution analyses were then used as initial estimates in the numerical solver of the higher resolution images. Not only did this aid computation speed, but it also allowed the solver to find solutions with increased stability. Particularly in deeper tissue, correlation of the fine level RF signal becomes more unlikely as there is a higher signal to noise ratio. The lower resolution images allow a solution based more upon the larger ultrasound features to be found. This is similar to the process described in [8].

Multiframe Compression: As the displacements generated are in the order of millimeters, it was not possible to do a simple comparison between the last and first frames in the sequence. Instead, the frame to frame displacement over each subsequent frame and sum the displacement image stack calculated to obtain the overall displacement.

Outlier removal: finally, a median filter was used to remove outliers from the displacement image.

2.5 Strain image algorithm

The strain was estimated by calculating the point value of the gradient of the displacement image. This was achieved by taking a user specified two dimensional window (3mm x 3mm) of the displacement and fitting a one-dimensional polynomial, with the regression points weighted by the normalised mean square error. This was performed in both lateral and axial directions to get the independent strain in each of these directions.

2.6 Segmentation of tendon strain

Finally, the areas of damaged tendon seen in the ultrasound image were identified by manual segmentation of the images by a clinician and stored in a database. This allowed average values to be calculated for the damaged area.

3. RESULTS & DISCUSSION

3.1 Images

B-mode images were of a consistent and good quality with the tendon, fat, muscles, damaged area and other features usually very clearly visible. In some cases there were significant shadows caused by the blood clot in the healing tendon or the knot of surgical suture. In a few cases (<1%) there was poor contact with the skin or obvious large amounts of decorrelation of the speckle pattern. In cases where this caused artifacts in the displacement and strain images, the analysis was discarded.

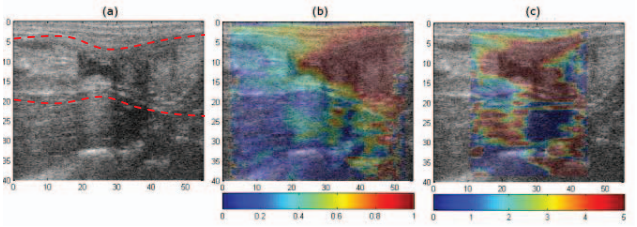


Figure 3. Example of images output from ruptured tendon moving in dorsal flexure lateral motion. a)B-mode grayscale (with dashed outline showing tendon boundary), b)displacement (mm), c) strain (%)

The PF motion gave inconsistent images, as toe-pointing motion from neutral over-slackens the AT. The DF and manual decompression motions gave visually more consistent displacement and strain images. Examples of the lateral motion B-mode, displacement and strain images can be seen in Figure 3.

3.2 Average strain consistency

As there were multiple analyses for each motion type for a given patient and week, the consistency of the average strain values for each motion type could be calculated. These are summarized in Table 1.

Motion Type	Standard deviation (%)
DF strain	0.77
PF strain	0.93
Manual strain	1.74

Table 1, summary of the repeatability of the average strain measured by different motion types.

The controlled DF motion provided substantially more repeatable strain measurements than the manual decompression motion.

3.3 Trend Data

For each motion type the average strain was plotted over time as shown in Figure 4. This was visually compared with the average clinical results shown in Figure 5. PF motion showed no real correlation, but the average scores of the DF and manual decompression motions fit much better.

In order to quantify the consistency of the data, an exponential decay of strain was modeled to fit each of the data and the R^2 was found as a measure of the data scatter, as shown in Table 2. This shows that there was a greater consistency of patients fitting the model used. While the model is not corroborated it should stand as a reasonable bench mark for analyzing variability of fitted points.

Motion Type	R^2
DF strain	0.78
PF strain	0.23
Manual strain	0.34

Table 2, Summary of R^2 values for DF, PF and manual decompression motions fitted to $f(x) = a.exp(-bx) + c$ type curves.

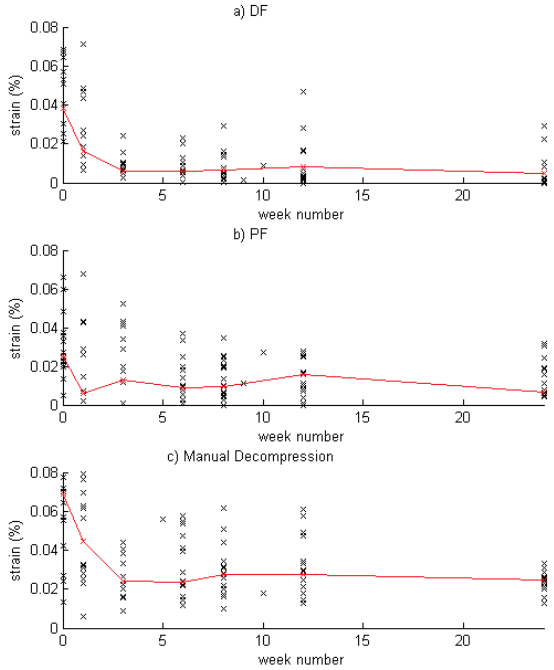


Figure 4. Average strain plotted for all patients over treatment period. The red line shows the mean.

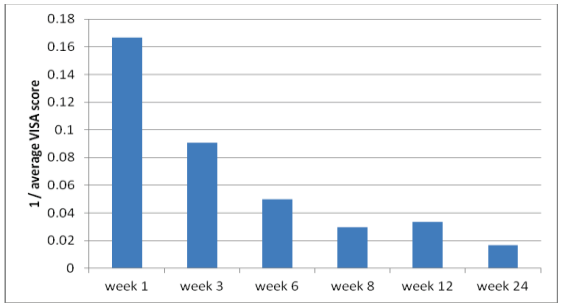


Figure 5. Average patient VISA improvement scores. (VISA is a score of positive ability, so 1/ VISA is presented here to show similarity to strain pattern)

5. CONCLUSIONS

This work has shown that there is potential value to further investigation of lateral strain imaging as a method of analyzing healing in Achilles tendon ruptures. By improving on the consistency of ultrasound sequence acquisition and using a multiscale, multicompression displacement tracking algorithm, the measurement of lateral strain can more consistent than with freehand probe-axial compression/ decompression motion and also still fit with clinical measures for group analysis.

This method also has potential to be improved upon. By measuring the force data and geometry of the patients, it should be possible to generate force-normalized strain, stiffness and Young's Modulus figures for each patient at each time point.

Further work is underway to corroborate the quantitative capabilities of lateral strain imaging in AT and the sensitivity of the technique to distinguish between treatment effectiveness for different patient groups and to establish its sensitivity for assessment of other pathologies such as tendinitis.

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