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Short communication

A biomechanical model for fibril recruitment: Evaluation in tendons and arteries

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ABSTRACT

Simulations of soft tissue mechanobiological behaviour are increasingly important for clinical prediction of aneurysm, tendinopathy and other disorders. Mechanical behaviour at low stretches is governed by fibril straightening, transitioning into load-bearing at recruitment stretch, resulting in a tissue stiffening effect. Previous investigations have suggested theoretical relationships between stress-stretch measurements and recruitment probability density function (PDF) but not derived these rigorously nor evaluated these experimentally. Other work has proposed image-based methods for measurement of recruitment but made use of arbitrary fibril critical straightness parameters. The aim of this work was to provide a sound theoretical basis for estimating recruitment PDF from stress-stretch measurements and to evaluate this relationship using image-based methods, clearly motivating the choice of fibril critical straightness parameter in rat tail tendon and porcine artery. Rigorous derivation showed that the recruitment PDF may be estimated from the second stretch derivative of the first Piola-Kirchoff tissue stress. Image-based fibril recruitment identified the fibril straightness parameter that maximised Pearson correlation coefficients (PCC) with estimated PDFs. Using these critical straightness parameters the new method for estimating recruitment PDF showed a PCC with image-based measures of 0.915 and 0.933 for tendons and arteries respectively. This method may be used for accurate estimation of fibril recruitment PDF in mechanobiological simulation where fibril-level mechanical parameters are important for predicting cell behaviour.

1. Introduction

The complex microstructural arrangement of collagen and elastin fibrils endows highly loaded soft tissues such as blood vessels and tendons with their unique mechanical properties. These fibrils commonly have high tortuosity (“crimp”) at low stretches which is removed as the tissue stretch is increased, resulting in a stiffening behaviour as observed with in situ loading experiments using multiple imaging modalities: Atomic Force Microscopy (AFM) (Rigozzi et al., 2011), Optical Coherence Tomography (OCT) (Hansen et al., 2002), X-ray scattering (Fratzl et al., 1998), polarizing microscopy (Diamant et al., 1972). As each fibril uncrimps at its recruitment stretch it begins to bear load, so the distribution of recruitment stretches for a tissue is an important determinant of the tissue mechanical property.

The recruitment stretch plays a significant role in the static and dynamic fatigue properties of these tissues – more highly loaded fibrils may fail first (Fung et al., 2009). Recruitment is also important in the mechanobiology of the tissue, as the local mechanical environment for mechanosensitive cells is profoundly altered with fibril uncrimping (Hill et al., 2012; Watton et al., 2004; Wren et al., 1998) and modulating cell behaviour through extracellular matrix stiffness is proposed as a novel target for multiple diseases (Lampi and Reinhart-King, 2018).

Many constitutive material models for fibrillar soft tissues have been proposed, either phenomenologically representing the tissue stiffening effect of recruitment (Fung, 1967; Holzapfel et al., 2000) or explicitly representing fibril recruitment with a distribution of recruitment stretches (Frisén et al., 1969). A variety of different probability distribution functions (PDF) for recruitment have been proposed and fitted to experimental data (Hill et al., 2012). The simplest, a triangular PDF has three parameters (start of recruitment, maximum rate of recruitment, end of recruitment) that are readily identifiable from imaging data (Aparicio et al., 2016).

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Direct measurement of fibril recruitment from microscopy has used the concept of straightness ratio, S and a critical value, S_c such that a fibril may be considered to be load bearing if $S > S_c$. Various different values for S_c have been used, apparently arbitrarily.

It has been proposed that the second derivative of the stress stretch relationship is proportional to the fibril recruitment PDF (Bontempi, 2009) but without rigorous derivation and evaluation against imaging data of fibril recruitment. The aims of this work were to provide a rigorous proof of the fibril recruitment stretch PDF relationship to the second derivative of stress with respect to stretch, to evaluate this relationship experimentally in rat tail tendons and porcine arteries and explore its sensitivity to critical straightness ratios. The hypothesis was that the stress stretch derivative parameter was correlated to image-based measures of the recruitment PDF. In situ microscopy was used to obtain measures of fibril recruitment and stress stretch behaviour in the same specimen under tensile load. Tendon and artery were chosen as having both strong clinical and scientific interest and widely differing stress-stretch behaviour to test this relationship.

2. Materials and methods

2.1. Theoretical development

We use the term fibril to refer to a quasi-crystalline filament of collagen 50 – 500 nm in diameter, as defined for tendon (Handsfield et al., 2016). Let the macroscale tissue stretch of a fibrillar tissue undergoing a tensile test be $\lambda = 1 + e$ where e is engineering strain. Both are λ and e are work conjugate to the first Piola-Kirchhoff stress. Any individual fibril bears no load until $\lambda = \lambda_R$, where λ_R is the recruitment stretch for that fibril. The macroscale tissue strain energy density function $W_t(\lambda)$ is the integral of the energies of individual fibrils $W_f(\lambda, \lambda_R)$ over all recruitment stretches $\lambda_R \leqslant \lambda$ weighted by the probability density function $g(\lambda_R)$ that an individual fibril has a given recruitment stretch λ_R. Using rule of mixtures and assuming tensile load is carried only by one family of fibrils, we include the fibril area fraction ϕ for this family. This gives:

$$W_t(\lambda) = \phi \int_1^{\lambda} g(\lambda_R) W_f(\lambda, \lambda_R) d\lambda_R$$

We choose a simple strain energy function (SEF) for a fibril (Hill et al., 2012) using $\lambda_f = \lambda/\lambda_R$ as the fibril stretch:

$$W_f = E\frac{(\lambda_f - 1)^2}{2} = E\frac{(\lambda - \lambda_R)^2}{2\lambda_R^2}$$

so the whole tissue SEF is

$$W_t(\lambda) = \phi E \int_1^{\lambda} g(\lambda_R) \frac{(\lambda - \lambda_R)^2}{2\lambda_R^2} d\lambda_R$$

where E is the elastic modulus of a fibril, assumed constant for all fibrils.The first Piola-Kirchhoff whole tissue stress, P_t, is given:

$$P_t = \frac{dW_t}{de} = \frac{dW_t}{d\lambda}\frac{d\lambda}{de} = \frac{d}{d\lambda}\left[\phi E \int_1^{\lambda} g(\lambda_R) \frac{(\lambda - \lambda_R)^2}{2\lambda_R^2} d\lambda_R\right]$$

This can be simplified using the Leibniz integral rule, which has standard form as follows (Protter and Morrey, 1985):

$$\frac{d}{d\lambda}\left[\int_{a(\lambda)}^{b(\lambda)} f(\lambda, \lambda_R) d\lambda_R\right] = f(\lambda, b(\lambda))\frac{db}{d\lambda} - f(\lambda, a(\lambda))\frac{da}{d\lambda} + \int_{a(\lambda)}^{b(\lambda)} f_{\lambda}(\lambda, \lambda_R) d\lambda_R$$

Here we set $f(\lambda, \lambda_R) = g(\lambda_R)(\lambda - \lambda_R)^2/2\lambda_R^2$ and confirm that $f(\lambda, \lambda_R)$ and f_{λ}, its partial derivative with respect to λ, are continuous in the domain of interest, and the integral limits $a(\lambda) = 1$ and $b(\lambda) = \lambda$ and their derivatives with respect to λ are also continuous in this domain.

Therefore we obtain

$$P_t = \phi E \int_1^{\lambda} g(\lambda_R) \frac{(\lambda - \lambda_R)}{\lambda_R^2} d\lambda_R$$

applying the rule a second time with the same continuity assumptions gives

$$\frac{dP_t}{d\lambda} = \phi E \int_1^{\lambda} g(\lambda_R) \frac{1}{\lambda_R^2} d\lambda_R$$

and applying a third time:

$$\frac{d^2P_t}{d\lambda^2} = \phi E\, g(\lambda)\frac{1}{\lambda^2}$$

This motivates the definition of a stress stretch parameter, $\zeta = \lambda^2 \frac{d^2P_t}{d\lambda^2} = \phi E g(\lambda)$ with units of Pa, to estimate the recruitment stretch PDF $g(\lambda)$ from mechanical tests.

2.2. Experimental work

Rat tail tendon fascicles (n = 7) were collected from 10-week old rats sacrificed in unrelated work. Tendon fascicles were dissected from the distal end of each tail. The mean cross-sectional area of each specimen assuming circularity was calculated from three optical micrometer (Keyence, Milton Keynes, UK; precision 10^{-5} mm) measurements of diameter under a tare load of 0.05 N. Samples were stored at −20 °C for up to one week before use. Prior to imaging and testing samples were removed from the freezer and allowed to warm in phosphate buffered saline (PBS) to room temperature for two hours.

Porcine aortic arteries (n = 5) were dissected within an hour of sacrifice from 6 month old Landrace pigs' hearts obtained from a local abattoir. Samples from the aortic arch region were carefully excised, cut longitudinally and then into circumferential strips. Width and height dimensions of samples were obtained as an average of three measurements again using an optical micrometer under a tare load of 0.05 N. Samples were then stored in phosphate buffer solution (PBS) at −20 °C for over 24 h. Prior to imaging and testing samples were removed from the freezer and allowed to warm to room temperature for two hours.

Mechanical testing was carried out using a custom-built uniaxial micro-tensile testing machine (linear actuator resolution 1 μm, T–NA0A850, Zaber; load cell resolution 0.033 N, WMC–5, Interface). The tissue was held in custom clamps and suspended in a PBS bath on a 0.13 mm thick glass cover slip over the inverted microscope objective. The artery strips were extended in the circumferential direction. Zero strain was defined with a tare load of 5 mN. A mean stress-strain curve was obtained for each specimen from four displacement ramp loadings at 0.2% s^{-1} to 4% (tendon) or 100% (artery) strain, data sampled at 10 Hz with 5 min rest between each test. The second derivatives of the stress-stretch curves for each tissue sample were obtained using cubic spline fits and numerical differentiation in Matlab.

The testing machine was mounted on an LSM 710 inverted confocal laser-scanning microscope (Zeiss, UK) fitted with a tunable pulsed femtosecond titanium-sapphire Chameleon laser (Coherent, UK) operating at 860 nm. SHG signal was collected at 430 nm. A stepped displacement protocol with increments of 0.5% strain (tendon) or 7% strain (artery) was used to apply up to 4% strain (tendon) or 100% (artery). After a 2 min equilibration time at each

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