Journal of Biomechanics 48 (2015) 262–268

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Examining the feasibility of applying principal component analysis to detecting localized changes in mechanical properties

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ARTICLE INFO

Article history:
Accepted 24 November 2014

Keywords:
Principal component analysis
Skeletal muscle
Strain imaging
Ultrasound
Compression

ABSTRACT

Ultrasound elastography is being increasingly used to diagnose musculoskeletal pathologies. This technique measures strain distributions within the muscle tissue, which can be used to calculate local tissue stiffness. However, inaccuracies in the model assumptions, as well as operator variability during imaging, can cause significant error in the resulting stiffness measurements. Principal component analysis (PCA) offers a means to better account for the physiological differences and inter-operator error between subjects. In this study we explore the feasibility of using PCA to detect contusion injury in skeletal muscle. We used finite element analysis to simulate quasi-static transverse compressions of both healthy and injured biceps brachii muscles. Injuries were modeled as circular regions with dissimilar mechanical properties. The compression magnitude and direction, as well as the muscle material properties, were varied to account for user error and inter-muscle differences, respectively. Image noise was added to the resulting displacement maps to simulate measurement error. PCA was performed on the displacement images of the healthy muscle. The resulting principal components were compared against the displacement maps from injured muscle to identify local changes in mechanical properties. The results indicate that changes in mechanical properties of $\pm 5\%$ can be detected for regions as small as $\pm 5\%$ of the muscle depth. However, detection accuracy was greatly affected by image noise. We show that PCA of displacement images is a feasible tool for detecting and quantifying localized changes in mechanical properties. Additional work is needed before this technique can be applied to images generated using ultrasound elastography.

1. Introduction

Damage to skeletal muscle tissue, such as contusions, tears, and strains, accounts for approximately 90% of all sports related injuries (Beiner et al., 1999). Ultrasound is recognized as a viable alternative to magnetic resonance imaging for diagnostic imaging of soft tissues, due to its portability, affordability, and real-time imaging capabilities (Campbell et al., 2005; Martinson and Stokes, 1991). Ultrasound elastography (USE), or strain-imaging, is emerging as a potential tool for monitoring and diagnosing musculoskeletal pathologies (Drakonaki et al., 2012), however its high degree of operator-dependence has limited its widespread use.

The oldest and most prevalent USE technique is quasi-static elastography (QSE), where the ultrasound transducer is used to manually compress the tissue of interest, and record the subsequent deformation (strain) (Drakonaki et al., 2012; Ophir et al., 1991; Sarvazyan et al., 2012). Visual inspection of these strain maps against a healthy baseline image can identify local regions of increased or decreased deformation levels, which correspond to regions of decreased or increased stiffness, indicating possible abnormal tissue.

Tissue stiffness differences can be quantified by calculating the effective Young's modulus of the tissue. However, these calculations require mathematical models of the force-displacement relationship of the tissue, which use simplifying assumptions about the tissue properties; including material isotropy, linear elasticity, plane strain, or plane stress (Doyley et al., 2005; Gokhale et al., 2008; Sarvazyan et al., 2012; Shore et al., 2012). Since muscle tissue is both nonlinear and inhomogeneous, these assumptions introduce inaccuracies to the calculated stiffness values. For QSE techniques, stiffness calculations require either the assumption of constant uniform stress, or measurements of both the applied pressure and resulting displacement, which require additional equipment (Han et al., 2003; Niitsu et al., 2011).

Further inaccuracies are introduced into QSE strain measurements by operator error. Since the compressions are usually performed freehand, the resulting displacement measurements may be affected by variations in compression magnitude and direction, as well as out-of-plane motion and transducer shaking (Drakonaki et al., 2012; Kadour and Noble, 2009). These variations may cause image artifacts that can lead to misdiagnosis of tissue damage. Averaging displacement images from repeated compressions of a single subject has been shown

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http://dx.doi.org/10.1016/j.jbiomech.2014.11.037
0021-9290/

to reduce image noise and increase intra-subject repeatability (Leineweber et al., 2014), but inter-subject comparisons are still affected by operator error, as well as anatomical and physiological differences between subjects.

Principal component analysis (PCA) offers a means to better account for these physiological differences and inter-operator error between subjects. PCA is a statistical technique used to identify the primary features, or principal components, common among a set of similar images. These principal components can then be combined to reconstruct an approximation of any image exhibiting the major features. Cochran and Gao showed that PCA could be used to comprehensively represent a set of simulated elastograms of the patellar tendon (Cochran and Gao, 2013). They also introduced a metric called the *damage-displacement,* to directly compare elastograms of injured tissue and the healthy principal component model to locate the damaged areas and quantify their severity without the need for mathematical approximations of tissue stiffness. Therefore, with PCA and the damage-displacement technique, no additional information about applied pressure is necessary, and compressions can be performed without having to record force data.

In this study we explore the feasibility of using PCA to detect contusion injury in skeletal muscle. A finite element (FE) model of the relaxed muscle tissue subjected to force-controlled transverse compression was used to obtain the displacement distributions. Using the *damage-displacement* technique, we compare the healthy PCA set against individual injury FE models to explore the effects of injury size and severity on injury detectability with and without the presence of added displacement image noise.

2. Methods

Finite element methods were used to create a model of the *biceps brachii* muscle undergoing transverse compression, standard for quasi-static ultrasound elastography. FE models were created to represent both healthy and injured muscle tissue. A Monte-Carlo simulation was performed to select the input parameters to the healthy muscle model from a range of acceptable values. Principal component analysis (PCA) of the displacement maps generated by the FE analysis produced a representative set of normal healthy muscle variability. The injured muscle displacement maps were compared against these principal components to locate and quantify the extent of tissue damage

2.1. Finite element model

The FE analysis was performed in ABAQUS CAE (2013, Dassault Systemes, Paris, France). For the purposes of this study, the biceps muscle was approximated as a 2D rectangular structure subjected to plane stress conditions. The rectangle height was set to 15.9mm to match the average biceps depth measured in our previous studies (Leineweber et al., 2014). The rectangle length and depth were based on the measurements of the Terason 12L5 linear array transducer with a 38.1mm x 10mm cross-section (Terason, Burlington, MA, USA), which defines the width of the imaging window. To avoid end effects from the boundary conditions, the rectangle length was set to 152.4mm, four times the width of the imaging window (Fig. 1).

The muscle substructure was approximated as alternating layers of two hyperelastic materials with separate thicknesses, corresponding to the muscle fascicles and perimyseum. These layers span the entire length of the muscle, with

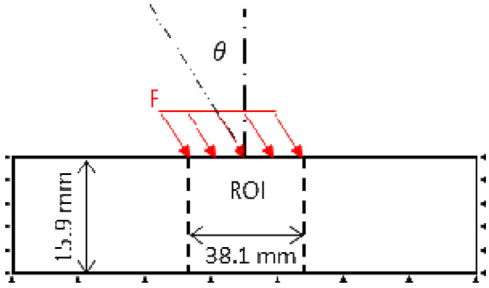

Fig. 1. Geometry and boundary conditions of the FE model. F represents the applied force magnitude and θ is the compression direction. θ was allowed to vary within $\pm 30^{\circ}$.

290 and 80μm approximate thicknesses for the fascicle and perimyseum layers, respectively (Zhang and Gao, 2012). The alternating layered pattern was created by assigning the corresponding muscle fascicle and perimyseum material properties to individual elements based on their position in the model.

A 774 x 400 element mesh (height x length) was applied to the rectangle, resulting in a 20.54μm x 381μm element size. This element size was chosen to ensure 14 and 4 elements, respectively, per muscle fascicle and perimyseum layer thickness, and is approximately four times the axial resolution of the musculoskeletal ultrasound system (Terason t3000, Terason, Burlington, MA, USA). Since the lateral strain was not as important to this study, a smaller element width was not necessary. All elements had a depth of 10mm to match the transducer thickness.

Both the muscle and perimyseum layers were treated as incompressible Mooney-Rivlin hyperelastic materials (Sharafi and Blemker, 2011; Teran et al., 2003; Zhang and Gao, 2012) with the strain energy function defined as

$$W = C_1(I_1 - 3) + C_2(I_2 - 3),$$

in which C_1 and C_2 are empirically determined material constants defined by ABAQUS, and I_1 and I_2 are the first and second principal invariants of the left Cauchy-Green tensor. The baseline material constants, C_1 and C_2, for the perimyseum and muscle fibers were taken from Zhang et. al. (2012). Muscle fascicles were approximated as transversely isotropic bundles of muscle fibers surrounded by the perimysium. New material constants for muscle fascicles were calculated using the Rule of Mixtures, assuming 90% of the fascicle volume was muscle fibers:

$$C_i = \frac{C_i^m C_i^e}{V_m C_i^e + V_e C_i^m}$$

where $i = 1, 2$, C_i is a material constant of the fascicle, C_i^m is the muscle fiber material constant, C_i^e is the perimyseum material constant, V_m is the volume fraction of muscle fibers, and $V_e = 1 - V_m$ is the volume fraction of perimyseum. The Mooney-Rivlin constants used in this study are listed in Table 1. The bulk modulus values, K are required as inputs in ABAQUS. To approximate tissue incompressibility, K is set to be very large relative to the corresponding C_1 and C_2 values.

Displacements in the lateral and axial direction were constrained to be zero on the bottom, left, and right edges of the rectangle. The top edge remained free except for a 38.1 mm wide region in the center of the rectangle, which was subjected to a constant compression. These compressions were modeled using an applied surface traction vector to the top row of elements, with a nominal magnitude 0.0117 N, which induced an approximate 2% mean strain in the model.

To incorporate the effects of inter-user variability, the magnitude and direction of the compressions were allowed to vary between 0.0105 N *and* 0.0129 N, representing a $\pm 10\%$ range from the baseline force magnitude. The compression direction was allowed to vary $\pm 10^{\circ}$ off-axis from a pure transverse compression.

Ultrasound imaging achieves its highest resolution in the direction of wave propagation. Therefore, most quasi-static elastography applications only measure displacements parallel to the beam-path, which is parallel to the compression. Although the FE model allows the analysis of displacement in both x and y directions, since the ideal compression in our model is orthogonal to the fiber direction, only the displacements transverse to the muscle fiber direction were considered for PCA analysis.

2.2. Simulating displacement noise

Once displacement maps were generated, displacement error was simulated by adding Gaussian white-noise to each displacement image. The white noise was modeled as zero-mean with a standard deviation equal to 0%, 1%, 2%, or 4% of the mean overall displacement of the healthy dataset (Cochran and Gao, 2013). The subsequent principal component analysis and *damage-displacement* techniques were independently applied to each of the four resulting image sets.

2.3. Developing the principal component model

A total of 100 FE models of healthy muscle were used. Monte-Carlo simulation was used to select the material constants, compression magnitude, and compression direction within a range of $\pm 10\%$ from the nominal values shown in Table 1. The resulting FE models were analyzed and the corresponding displacement maps from each model were saved. Only the displacements from the region of interest (ROI, 775 × 101 nodes, 774 × 100 elements) were used in the PCA.

Table 1
Healthy material constants.

	C1 (kPa)	C2 (kPa)	K (kPa)
Perimyseum	0.3	0.15	10
Fiber	3	1.5	100
Fascicle	2.46	1.23	100