

PASSIVE HUMAN MUSCLE PROPERTIES FOR FINITE ELEMENT HUMAN BODY MODELS FOR SAFETY

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ABSTRACT

The complexity of biological tissue presents significant difficulties for constitutive formulations for computational models for impact. The current study presents indentation tests on a volunteer's upper arm for the non-invasive investigation of passive skeletal muscle tissue mechanical properties. A novel Magnetic Resonance Imaging (MRI) technique and a custom designed MRI compatible indenter and force sensor were successfully used to record boundary conditions such as geometry, 3D tissue deformation and indentation force non-invasively. These, when combined with inverse finite element analysis (FEA), provide a means to derive parameters for constitutive formulations for human muscle tissue suitable for impact modelling.

The results demonstrate a complex anisotropic strain field arising from relatively simple indenter motion, with regions of tensile strain around the sidewall of the indenter as well as regions of compressive strain below the indenter. Furthermore, although not yet quantified, viscoelasticity and anisotropy of the tissue were qualitatively observed. This data can in the future be combined with FEA to improve the bio-fidelity of constitutive models of human muscle tissue.

Keywords: MRI, MUSCLE, COMPRESSION, HUMAN TESTS

Finite element models for safety require both geometric and constitutive input. Great advances have been made in the former, but the complexity of biological tissue presents significant difficulties for the latter. This is especially so for soft tissues, which exhibit nonlinearity, non-homogeneity, anisotropy and viscoelasticity. Skeletal muscle tissue accounts for almost half of the body's weight, so mechanical representation of passive muscle tissue is crucial for impact modelling. Determination of the constitutive properties of living human muscle tissue presents particular difficulties, and most studies have focused on *in-vitro* specimens (e.g. [1, 2]). Recent work on *in-vitro* porcine muscle samples showed that a strain-dependent transversely isotropic material model extended for viscoelasticity with a Prony series expansion can be effective for modelling time-dependent deformation behaviour of porcine muscle in compression [2-4]. Since validating such models for living human tissue presents significant difficulties and requires non-invasive methods (e.g. combined with inverse FEA), measurement of muscle deformation has so far been limited to animal tests and to small strain cases (e.g. [5, 6]). The current study presents indentation tests on the upper arm of a volunteer conducted inside a MRI scanner. A custom designed MRI compatible indenter (incorporating a novel optical force sensor) is used to apply compression and record the complex force history. The resulting non-linear 3D soft tissue deformation is measured using a novel MRI sequence. The methods presented allow, for the first time, the measurement of the boundary conditions required for inverse FEA based determination of the complex material parameters of living human skeletal muscle tissue.

METHODS

Using a custom designed MRI compatible indenter (figure 1) static ramp and hold indentation experiments (13.3mm for 3s at ~40mm/s) were conducted on the upper arm (figure 2A) of a healthy volunteer (female, age 24, height 1.65m, weight 65kg, ethical approval and informed consent obtained, Ethical Committee Academic Medical Centre, Amsterdam, the Netherlands) inside a 3T MRI scanner (Philips Intera, Philips Healthcare, Best, The Netherlands). The subject was placed in a supine position with the arm stretched and was instructed to relax and keep muscles passive during the indentation tests. Since electromyography is not generally possible in an MRI environment muscle activity was not monitored. However the brief and mild compression used in the current study should not result in activation of the muscle tissue.

The flat indenter head (manufactured from polyoxymethylene) is cylindrical (45mm in diameter) and its speed and depth can be varied via a computer controlled hydraulic master cylinder (figure 1A). The circularity of the indentation is also evident from the iso-surfaces shown in figure 2. The indenter is equipped with a novel, high time resolution (up to 1kHz) optical fibre based force sensor (figure 1C-D). The indenter force, derived through analysis of the strain and temperature dependent reflected peak wavelengths from two Fibre Bragg Gratings (FBG) (one of which is isolated from strain for temperature compensation) was calibrated using uniaxial testing (force error under 0.3% for 0~15N).

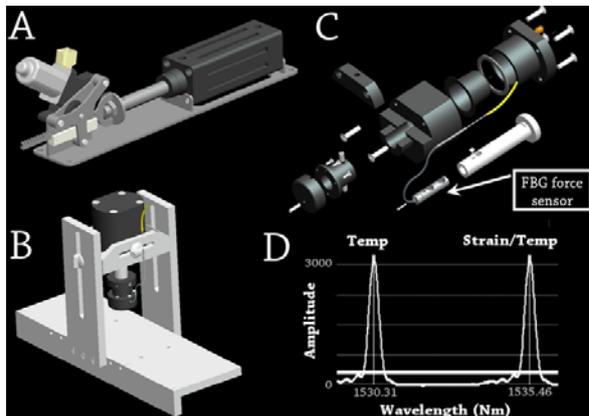


Figure (1): The computer controlled hydraulic master cylinder (A), the MRI compatible hydraulic indenter (B and C) the optical force sensor (C) and a sample force sensor output showing reflected FBG peaks from the two gratings (left just temperature, right also strain)

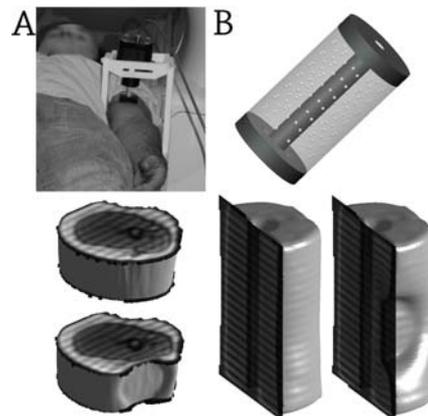


Figure (2): The volunteer test (A) and validation phantom containing markers (B). In addition SPAMM tagged MRI slices and iso-surfaces for the initial and deformed configurations are shown for the human (left middle and bottom) and phantom (below in middle and right)

The soft tissue deformation resulting from the indentation was recorded using a novel MRI technique based on SPATial Modulation of the Magnetization (SPAMM) [7]. In SPAMM tagged MRI contrasting surface (or tag) patterns can be temporarily introduced in the tissue (visible as lines in figure 2A-B) whose distortion directly reflects tissue motion. SPAMM tagged MRI data was acquired in 3 orthogonal directions using full 3D volume read-outs (Transient Field Echo read-out, $T_R/T_E=2.9/1.8$ ms, flip angle 8° , resolution $0.88 \times 0.88 \times 3$ mm, field of view $253 \times 253 \times 45$ mm). A single acquisition took place in less than 1.5s. In order to derive deformation from the MRI data spline surfaces were fitted at the tag locations using a newly developed step-wise sheet marching algorithm. The intersections of tag surfaces from all 3 directions over time provide a 3D grid of trackable points. Standard high resolution anatomical MRI was also recorded providing the geometry necessary for FEA (e.g. using iso-surfaces such as those shown in figure 2). The deformation measurement techniques were validated against marker tracking in a silicone gel phantom (figure 2B) using the methods outlined in [8] and in [9]. The phantom shape represents an idealised arm geometry and as such similar deformation modes could be tested, as is evident from the high congruence between the iso-surfaces shown in figure 2. Sub-voxel accuracy and precision with a mean displacement difference of $72\mu\text{m}$ and a standard deviation of $289\mu\text{m}$ was achieved. In addition displacement magnitude precision could be analysed both *in-vivo* and in the phantom demonstrating that the standard deviations of displacement magnitude with respect to the average displacement magnitude were $75\mu\text{m}$ and $169\mu\text{m}$ respectively for the phantom and volunteer data.

RESULTS

Figure 3 shows several ghosted SPAMM tagged MRI slices and overlain the segmented tag surfaces (A), whose intersections in the initial and deformed configurations yielded the 3D displacement field (B) which allows for the calculation of soft tissue strains (C-D). As is shown in figure 3 the relatively simple indenter movement resulted in non-uniform muscle tissue deformation and strain. The results show an anisotropic response with regions of highest tensile strain around the sidewalls of the indenter, as well as highest compressive strains below the indenter (indicated by the white arrows in figure 3).

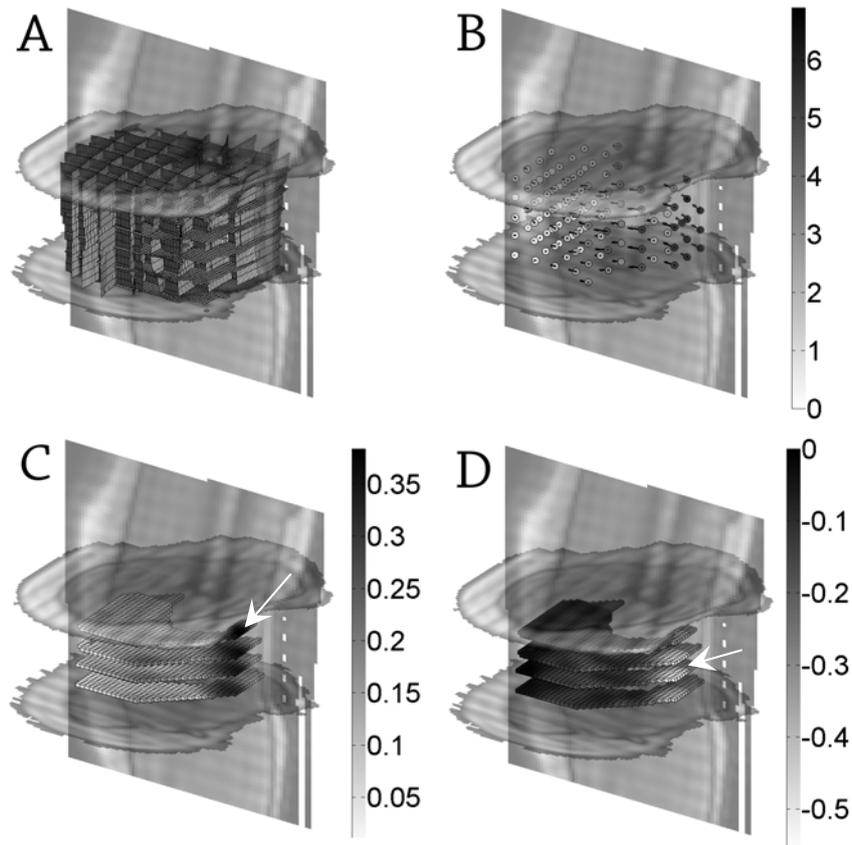


Figure (3): The ghosted SPAMM tagged MRI data and overlain the segmented tag surfaces (A), the displacement field (initial coordinates coloured towards displacement magnitude) (B) and the logarithmic maximum (C) and minimal (D) principal strain. The peak strain locations are indicated by the white arrows.

The FBG derived force and the indenter displacement time histories for 7 consecutive indentations for the volunteer are shown in figure 4A. These demonstrate the ramp and hold displacement pattern and the corresponding viscoelastic force response which decays during the hold phase. The load response also shows evidence of preconditioning as the peak force is reduced in additional motion cycles which is a common empirical finding for biological soft tissue. Figure 4B shows an approximate and preliminary force-displacement curve (some time matching was required for the two sensor outputs). This curve, although approximate, shows clear evidence of hysteresis, but it also shows apparent plateauing of the load for indenter displacements between 9 and 13mm, which obviously requires further analysis.

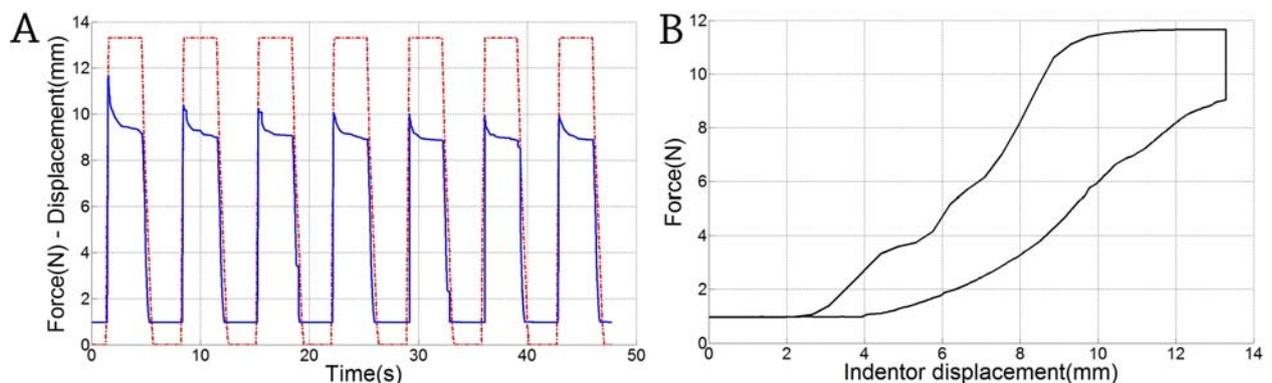


Figure (4): Force (solid curve) and displacement (dotted curve) histories (A) and an approximate force-displacement curve (B)

DISCUSSION

Determination of living human bulk muscle tissue response in compression is crucial for the bio-fidelity of computational models in impact biomechanics, this is however a difficult task. In particular, very little work has been presented on *in-vivo* human bulk tissue deformation during external compression, especially for large strain cases. This paper presents indentation tests on the

upper arm of a volunteer using a custom designed MRI compatible force sensor. The boundary conditions such as geometry and 3D tissue deformation were recorded using anatomical MRI and a novel SPAMM tagged MRI technique respectively, while the indenter force was measured using a novel FBG based optical force sensor incorporated in the indenter. The effectiveness of the MRI method for gaining full field deformation has been previously proven using a silicone gel phantom [9]. Application of the method to the human body showed the approach to be a robust means to extract full field internal deformation due to external loading, see figure 3.

In the future, these experimental data will be combined with iterative inverse FEA to optimize the parameters of the recently proposed constitutive model for muscle tissue based on the premise of transversely isotropic Strain-dependent Young's Moduli (SYM) [2-4]. The FEA model geometry is available from high-resolution anatomical MRI scans. The indenter displacement time-history (figure 4) will be used as input to the simulations, and the previously proposed transversely isotropic strain-dependent Young's modulus model for muscle [2-4] will be implemented. The SYM model was previously used to predict porcine muscle behaviour *in-vitro*, and the derived material parameters from those tests will be used as initial estimates for optimising the *in-vivo* human muscle parameters, using the measured indenter force and soft tissue deformation as a comparison for optimisation.

To our knowledge this is the first time full-field deformation of human soft tissue during external compression has been measured *in-vivo*. In future work the sample size will be increased to evaluate variability within the human population, as this will clearly influence the material parameters derived for muscle tissue.

Currently only a single indenter shape has been used thus future work will analyse the influence of indenter shape. The volunteer testing to date has been limited to a quasi-static regime. However the constitutive model [2-4] is formulated to represent more dynamic cases. Therefore, in the future (low rate) dynamic loading regimes will also be performed. However, limitations in the existing hydraulic device, the current MRI techniques and obvious ethical considerations will prevent very high rate and depth compression testing using human volunteers. Nonetheless, the results presented here for the first time provide *in vivo* data required for the calibration of constitutive models for soft tissue which are essential for the impact biomechanics community.

CONCLUSIONS

This paper presents a non-invasive method of determining the mechanical response of passive living human muscle tissue to external loading, which is highly relevant for impact biomechanics. The response to this kind of loading has been shown to be a non-homogeneous 3D strain field which in the future can be modelled using appropriate finite element analysis. This level of detail has not been previously available, and the nonlinearity, viscoelasticity and anisotropy of muscle tissue under external compression were clearly evident. Further work is required to fully integrate these experimental results with iterative finite element modelling to provide actual parameters for muscle modelling.

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