The soft tissues tension-compression asymmetry: challenges and solutions

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1. Introduction

Skeletal muscles behave in tension when activated. It is often supposed that they do not resist in compression (Liu et al. 2019) whereas such muscles and their surrounding soft tissues exert a passive resistance when compressed. Therefore, characterizing the muscle tissues' behavior under compression is essential. This was done for example by Nagle et al. (2014) who showed that the response of skeletal muscle tissue in tension/compression plays an important role in biomechanical simulations. Soft tissues have nonlinear and anisotropic elastic properties along with viscoelastic behavior (Takaza et al. 2014). Existing data from skeletal muscle tissue under tensile and compressive deformation has shown a significant tension-compression asymmetry (TCA) (Böl et al. 2022). Many researchers have worked on tensile deformations of soft tissues but only a few of them studied TCAand TCA has not been yet captured by current constitutive models using a unique set of material parameters (Takaza et al. 2014).

The aim of this paper is to study the deformation response of chicken pectoralis muscle under tension and compression and to investigate the capability of various hyperelastic models to represent the corresponding asymmetric behavior.

2. Methods

2.1. Experimental setup

Six specimens were harvested from pectoralis muscle of a two-month-old female chicken with an average body weight of 3 kg within 20 h post-mortem. Samples were cut to a proper size along the fiber directions for the corresponding tests. The three compression samples were cut to the size of 10 × 10 × 5 mm to prevent buckling and the three tension samples were cut to the size of 27 × 10 × 10 mm for having slender specimens. Load-to-failure tests were conducted on Santam STM-1 testing machine. Tests were performed at 1% per minute strain rate to minimize viscoelastic effects. No pre-tension was applied on tensile samples. Force data was recorded using a 6-kg ZEMIC load cell, MODEL L6D Class C3.

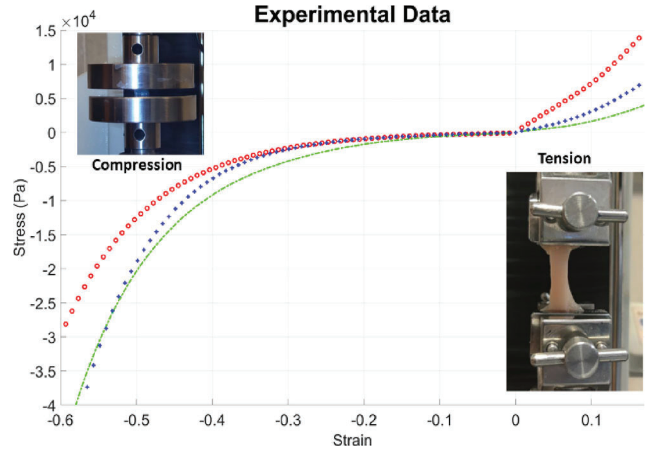

Figure 1. A sample uniaxial test data of pectoralis muscle along muscle fibers (tension and compression loadings were combined).

Ink lines were drawn at the muscle-clamp interface to confirm that the tests were performed without sample slipping.

2.2. Data processing

The strain was calculated as the elongation divided by the initial gauge length and the first Piola-Kirchhoff stress was computed as the force divided by the initial cross-sectional area. Ultimate stress and ultimate strain were taken as the stress and strain values at the point of material failure. The tensile data were combined with compression data. This combination was done for the samples harvested at same location and along the same fiber directions. The experimental stress-strain data is shown in Figure 1.

To find a suitable constitutive law, capturing the tissue behavior, different hyperelastic models available in the ANSYS© library were studied. This library is limited to isotropic models.

3. Results and discussion

As it can be seen on Figure 1, there is an obvious tension-compression asymmetry in uniaxial test. We picked up the most prominent asymmetric data to have their mechanical properties.

The chicken pectoralis muscle shows nonlinear behavior as most biological soft tissues. The apparent stiffnesses in tension and in compression at the same strain show large different orders of magnitude. For

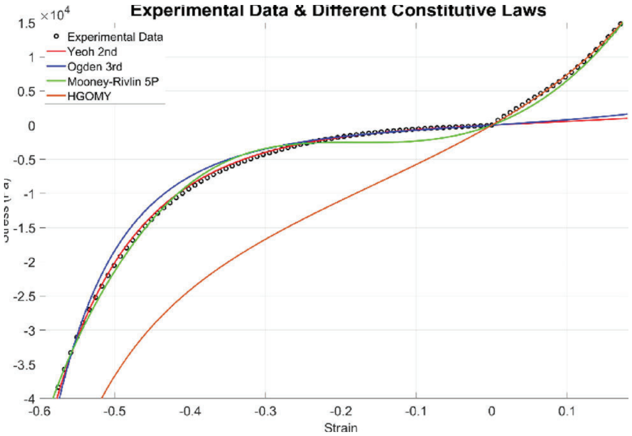

Figure 2. Three isotropic hyperelastic models to represent the TCA: 2nd order Yeoh, 3rd order Ogden and 5-parameter Mooney–Rivlin and a proposed transversely isotropic model (HGOMY).

example, at the maximum tensile stretch, the stress is about 12.4 kPa while at the equivalent compressive stretch, the magnitude of stress is about 0.68 kPa. This shows that the apparent stiffness in tension is about (12.4/0.68≈) 18 times higher than for compression.

To capture the tissue' behavior and its mechanical properties, we tried to fit our experimental data with most isotropic hyperelastic models available in ANSYS© library. Whereas hyperelastic models can well represent the tension and compression deformation responses separately, it seems necessary to check if such models can capture the tension-compression asymmetry with a unique set of material parameters. It is however evident that none of them can fully capture the slopes asymmetry observed at the origin (Figure 1) since such models have first order continuity at all points.

For fitting experimental data with different constitutive models, ANSYS© Workbench 2021 R1 was used. Among all the models tested, three are reported in this paper: 2nd order Yeoh, 3rd order Ogden and 5-parameter Mooney–Rivlin. Most of these models cannot capture the tension and compression behaviors at the same time with a unique set of parameters (Figure 2). Only the 5-parameter Mooney–Rivlin model provides a nice account of the tension-compression asymmetry.

The 5-parameter Mooney–Rivlin constitutive law is described by the following equation for strain energy density with respect to initial configuration:

$$\psi = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) + C_{11}(I_1 - 3)(I_2 - 3) + C_{20}(I_1 - 3)^2 + C_{02}(I_2 - 3)^2$$

Table 1. 5-Parameter Mooney–Rivlin material constants.

C_{01} [kPa]	C_{02} [kPa]	C_{10} [kPa]	C_{11} [kPa]	C_{20} [kPa]
−49.0	14.5	54.4	−56.5	73.4

where I_1 and I_2 are the first and second invariants of Cauchy-Green strain tensor respectively. The corresponding obtained material parameters are given in Table 1.

In fitting the experimental results, the program assumed a fully incompressible hypothesis. This assumption needed to be verified. For this purpose, we have computed the Poisson ratio using DIC imaging and assuming isotropy. This ratio was found to be equal to 0.3 which seemed not realistic since muscular soft tissues are mostly composed of water with a ratio that should be near to 0.5. This meant that the hypothesis of isotropy is probably wrong. To overcome this difficulty, a new method was proposed to model the material behavior, with a transversely isotropic material and an exponential behavior along fibers, inspired from the HGO strain energy density:

$$\psi = c_1(I_1 - 3) + c_2(I_1 - 3)^2 + \frac{k_1}{2k_2}\left(e^{k_2(I_4-1)^2} - 1\right)$$

in which I_4 is the square of stretch ratio along fiber directions. The first term represents the matrix behavior and the second term shows the fiber properties. It is known that the fibers do not resist much in compression. Therefore, to estimate the parameters of ψ, the values c_1 and c_2 were first estimated using compression data only. Then, while keeping these values constant, the parameters describing fiber direction were estimated using tension data only.

Using the above algorithm, c_1 and c_2 were found equal to 1.12 kPa and 0.705 kPa respectively assuming full incompressibility. Keeping these constants fixed and using tension experimental data, the material constants k_1 and k_2 were found equal to 14.1 kPa and 0.79 respectively.

4. Conclusions

Fitting experimental data with constitutive laws that include many parameters is not classical. This can end to an oscillating behavior as it can be seen with the 5 parameters Mooney–Rivlin law. The fitted models first assumed an isotropy of the tissue. This assumption had to be revoked since the quasi-incompressibility of the tissue could not be guaranteed in that case. A new anisotropic constitutive law was introduced to take into account muscle fibre directions, but this law still lack to represent the strong tension-compression asymmetry observed during the tests.

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