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Viscoelastic properties of passive skeletal muscle in compression—Cyclic behaviour

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ABSTRACT

Skeletal muscle relaxation behaviour in compression has been previously reported, but the anisotropic behaviour at higher loading rates remains poorly understood. In this paper, uniaxial unconfined cyclic compression tests were performed on fresh porcine muscle samples at various fibre orientations to determine muscle viscoelastic behaviour. Mean compression level of 25% was applied and cycles of 2% and 10% amplitude were performed at 0.2–80 Hz. Under cycles of low frequency and amplitude, linear viscoelastic cyclic relaxation was observed. Fibre/cross-fibre results were qualitatively similar, but the cross-fibre direction was stiffer (ratio of 1.2). In higher amplitude tests nonlinear viscoelastic behaviour with a frequency dependent increase in the stress cycles amplitude was found (factor of 4.1 from 0.2 to 80 Hz).

The predictive capability of an anisotropic quasi-linear viscoelastic model previously fitted to stress-relaxation data from similar tissue samples was investigated. Good qualitative results were obtained for low amplitude cycles but differences were observed in the stress cycle amplitudes (errors of 7.5% and 31.8%, respectively, in the fibre/cross-fibre directions). At higher amplitudes significant qualitative and quantitative differences were evident. A nonlinear model formulation was therefore developed which provided a good fit and predictions to high amplitude low frequency cyclic tests performed in the fibre/cross-fibre directions. However, this model gave a poorer fit to high frequency cyclic tests and to relaxation tests. Neither model adequately predicts the stiffness increase observed at frequencies above 5 Hz.

Together with data previously presented, the experimental data presented here provide a unique dataset for validation of future constitutive models for skeletal muscle in compression.

1. Introduction

Finite element models of the human body in compression are used in impact biomechanics, rehabilitation engineering and for simulating surgical procedures (Forbes et al., 2005; Linder-Ganz et al., 2007; Guccione et al., 2001). The models require a good knowledge of the tissue mechanical properties, but the rate dependent three-dimensional compressive behaviour of muscle tissue remains poorly understood.

Skeletal muscle has a fibre-oriented structure consisting of about 80% water, 3% fat and 10% collagenous tissues. It therefore displays anisotropic elasticity as well as time and history dependency. In Van Loocke et al. (2006, 2008), the elastic and viscoelastic properties of passive skeletal muscle were investigated using uniaxial unconfined compression tests performed on fresh porcine muscle tissue *in vitro* and a model was developed to represent the properties observed experimentally. In Van Loocke et al. (2006), quasi-static compression tests at various fibre orientations showed that muscle elastic behaviour is nonlinear and transversely isotropic. In Van Loocke et al. (2008), ramp-and-hold tests at various rates and fibre orientations showed that, above a very small compression rate, the viscoelastic component plays a significant role (approximately 50% of total stress at $0.5\,\%\mathrm{str\,s^{-1}}$). A stiffening effect with compression rate was observed, especially in directions close to the muscle fibres (factor of 5 from 0.05–10 $\%\mathrm{str\,s^{-1}}$ in the fibre direction at 30% compression). Skeletal muscle viscoelastic behaviour therefore depends on compression rate and fibre orientation.

In Van Loocke et al. (2006) a transversely isotropic strain dependent Young's moduli (SYM) model was proposed for muscle elasticity. The model yielded excellent fits to the experimental data in the fibre, cross-fibre and 45° directions ($R^2 = 0.99$) up to 30% strain and mean prediction errors for 30° and 60° tests were 3.5% and 9.5%, respectively. In Van Loocke et al. (2008) the model

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was extended with Prony series to discretise viscoelasticity, which provided a good fit to experimental data in the fibre, cross-fibre and 45° directions at compression rates of 0.5, 1 and 10 %str s^{-1} (errors < 20%). The model also yielded good predictions of muscle behaviour at 0.05 and 5 %str s^{-1} (errors < 25%). However, the predictive capabilities at higher compression rates and during more complex loading were not evaluated.

Muscle compressive properties at very high strain rates (up to 3.7.10^5 %str s^{-1}) have been previously investigated (McElhaney, 1966; Van Sligtenhorst et al., 2006; Song et al., 2007). In the two more recent studies, a split Hopkinson pressure bar apparatus was used. However, the level of control and sensitivity of this apparatus can be questioned.

In this paper, viscoelastic properties of skeletal muscle at strain rates up to 3200 %str s^{-1} are investigated using cyclic loading. Cyclic tests were used instead of airgun/SHPB tests because, despite lower loading rates, cyclic tests allow better control. The predictive capabilities of the quasi-linear viscoelastic (QLV) model (Van Loocke et al., 2008) with parameters based on stress-relaxation tests and a new nonlinear viscoelastic (NLV) model are assessed using these cyclic data. This paper complements the experiments and models in Van Loocke et al. (2006, 2008) which together provide a unique dataset on the mechanical behaviour of skeletal muscle in compression.

2. Materials and methods

2.1. Experimental tests

Uniaxial unconfined compression tests were performed on fresh porcine muscles using the same protocol as in Van Loocke et al. (2008). Muscles were excised from the pelvic limb of male pigs aged 10–12 weeks. From these, cubic samples oriented in the fibre and cross-fibre directions and at 45° and 60° from the fibre direction were cut and kept at room temperature in airtight containers prior to testing, which started within two hours post-mortem. Low and high amplitude cyclic tests were conducted on a Zwick Z005 machine (Zwick GmbH & Co. Ulm, Germany) up to a mean compression level of 25% and saw-tooth cycles of 2% or 10% amplitude were then performed for a duration of 250 s at 0.22 and 0.4 Hz (limited by the Zwick). For higher frequencies, a custom rig was developed (Fig. 1). In this device, samples are compressed between stainless steel platens. Sinusoidal displacements are provided to the bottom platen by an electro-dynamic shaker and various levels of mean compression are achieved by displacement of the top platen. An LVDT measures the displacement of the bottom platen; a static strain gauge load cell on the top platen measures the force; a dynamic load cell (piezoelectric force sensor) is also mounted on the bottom platen to account for inertial effects. Muscle samples were cyclically tested at 5, 20 and 80 Hz. Cycles of 10% amplitude were performed around a mean compression level of 25% for 250 s. Displacement and force signals were acquired using a 16-bit data acquisition card (National Instruments PCI6036E) and Labview. The raw data were digitally low-pass filtered using Matlab (cutoff frequencies of 25, 100 and 400 Hz, respectively, for 5, 20 and 80 Hz tests). Table 1 summarises the tests performed. Six samples were tested in each direction.

2.2. Mathematical modelling

The data obtained were first compared to theoretical predictions from the QLV-SYM model developed in Van Loocke et al. (2008), with parameters derived from fitting to quasi-static and stress-relaxation data (Van Loocke et al., 2006, 2008). See details in the Appendix A.

As shown in the results section, the QLV approach could not fully capture the nonlinear behaviour observed for muscle tissue during high amplitude cyclic tests. Therefore, the generalisation by Poon and Ahmad (1998) of Schapery's nonlinear viscoelastic approach (Schapery 1969) was adopted

$$\sigma_{ij}(t) = h_e\left[\sum_k\sum_l G^{\infty}_{ijkl}\varepsilon_{kl}\right] + h_1\int_0^t\sum_k\sum_l\left[\Delta G_{ijkl}(\xi_{ijkl}-\xi'_{ijkl})\frac{d}{d\tau}(h_2\varepsilon_{kl})\right]d\tau \quad (1)$$

where G^{∞}_{ijkl} and ΔG_{ijkl} represent, respectively, the equilibrium and transient components of the relaxation modulus. The scalar functions h_e, h_1, h_2 depend on the strain tensor and the reduced time is influenced by the mechanical strain via

$$\xi_{ijkl} = \int_0^t \frac{dt}{a_{\tau ijkl}(E(t'))} \quad (2)$$

This NLV model was adapted to include the SYM formulation for the elastic behaviour of muscle tissue—the SYM formulation taking the role of h_e and h_2 in Eq. (1). The terms h_l and a_τ were then introduced to account for viscoelastic nonlinearities (h_l and a_τ are, respectively, denoted h_j and a_j in the following

Fig. 1. Custom built high speed cyclic testing rig. (A) Top platen; (B) bottom platen; (C) LVDT transducer; (D) static load cell; (E) dynamic load cell; and (F) connection with shaker.

Table 1
Summary of cyclic tests performed on porcine skeletal muscle.

Type of test	Cycles amplitude (%)	Frequency (Hz)	Compression rate (%str s^{-1})	Test direction
Low amplitude, low frequency	2	0.22	1	F, XF
High amplitude, low frequency	10	0.22 and 0.4	5 and 10	F, 45°, 60°, XF
High amplitude, high frequency	10	5, 20 and 80	200, 800 and 3200	F, 45°, 60°, XF