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A validated model of passive muscle in compression

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Abstract

A better characterisation of soft tissues is required to improve the accuracy of human body models used, amongst other applications, for virtual crash modelling. This paper presents a theoretical model and the results of an experimental procedure to characterise the quasi-static, compressive behaviour of skeletal muscle in three dimensions. Uniaxial, unconstrained compression experiments have been conducted on aged and fresh animal muscle samples oriented at various angles from the fibre direction. A transversely isotropic hyperelastic model and a model using the theory of transverse isotropy and strain dependent Young's moduli (SYM) have been fitted to the experimental data. Results show that the hyperelastic model does not adequately fit the data in all directions of testing. In contrast, the SYM gives a good fit to the experimental data in both the fibre and cross-fibre direction, up to 30% strain for aged samples. The model also yields good prediction of muscle behaviour at 45° from the fibre direction. Fresh samples show a different behaviour than aged tissues at 45° from the fibre direction. However, the SYM is able to capture this difference and gives a good fit to the experimental data in the fibre, the cross-fibre and at 45° from the fibre direction. The model also yields good prediction of muscle behaviour when compressed at 30° and 60° from the fibre direction. The effect of the time of test after death has also been investigated. Significant stiffening of muscle behaviour is noted a few hours after death of the subject.

Keywords: Skeletal muscle; Passive properties; Compressive behaviour; Transversely isotropic model

1. Introduction

The use of virtual human modelling has increased in the last few years, in particular in the field of impact biomechanics (Forbes et al., 2005; van Rooij et al., 2003; Verver et al., 2004; Ward et al., 2005). Finite element models of the human body can be used to predict deformations during transient loading. These models require both a description of hard and soft tissues geometries, as well as the determination of their material properties under large deformations, as appearing during impacts. However, the properties of soft tissues continue to be poorly classified. In particular, muscle tissue presents difficulties. As with most biological soft tissue, muscle presents a viscoelastic behaviour and has anisotropic properties due to its fibre-oriented structure. It is subject to large deformations in vivo and the stress state in a given muscle is the result of passive and active contributions. Furthermore, parameters such as age, gender and species are also influential. During the past century, numerous investigations have been conducted to determine muscle structure and function. Experimental studies have been performed to determine muscle properties. However, most of these studies have considered deformation in one dimension only (the longitudinal direction) and significant variability between data sets is present (see Fig. 1). In this, passive force–length curves obtained by tensile tests conducted variously on rat, cat or rabbit muscles are presented (Davis et al., 2003; Gareis et al., 1992; Hawkins & Bey, 1994, 1997; Muhl, 1982; Woittiez et al., 1984). Grieve and Armstrong (1988) and Bosboom et al. (2001a, b) published experimental data on the compressive behaviour of passive skeletal muscle in the transverse direction. To the authors knowledge, these are the only experimental study characterising muscle

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compressive behaviour. One-dimensional models have been developed to reproduce muscle global mechanical behaviour observed in the experiments. Most of these models (see for example Best et al., 1994; Cole et al., 1996; de Jager, 1996) are based on the famous model proposed by Hill (1938).

Three-dimensional constitutive models of nonlinear anisotropic tissue exist (see Humphrey et al., 1990; Li et al., 2001; Limbert and Middleton, 2004; Weiss et al., 1996). They have been developed to reproduce the tensile behaviour of soft tissues with fibre-oriented structure such as tendons and ligaments. However, only the model proposed by Humphrey et al. (1990) has been used to reproduce muscle behaviour (Martins et al., 1998) and the paucity of experimental data does not permit one to determine the validity of the model.

The objectives of this study are to provide an experimental as well as a theoretical characterisation of passive skeletal muscle properties in three dimensions. Particular attention is placed on muscle compressive behaviour, as this is the relevant deformation mode in human/machine impacts and seated postures. Emphasis is also placed on model simplicity and ease of parameter determination.

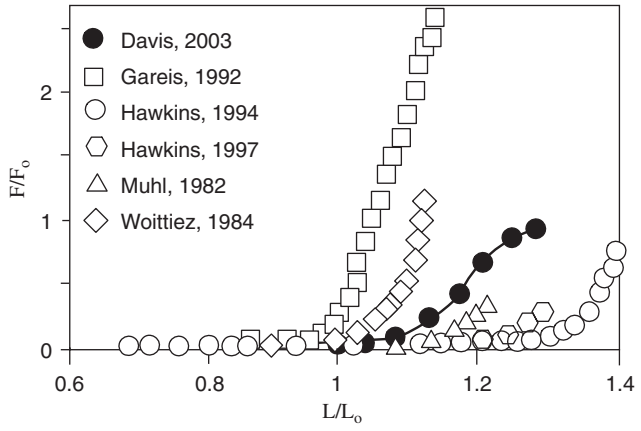

Fig. 1. Comparison of published passive muscle characteristics. F_0 is maximum isometric force generated by muscle; L_0 is the rest length of muscle.

This paper presents the results of an experimental procedure developed to determine the quasi-static properties of muscle tissue, as well as a simple model which reproduces the behaviour observed during the experiments.

2. Theoretical models

Skeletal muscle has a fibre-oriented structure: it is composed of fascicles containing bundles of fibres, themselves composed of parallel bundles of myofibrils (Gaudin, 1997). From a modelling point of view, muscle tissue is thus often considered as a nonlinear, unidirectional composite consisting of parallel fibres embedded in a matrix (see Fig. 2 for illustration). Unidirectional composites are transversely isotropic; however, applying engineering material technology to biological material is not without difficulty due to the complex behaviour of biological tissues.

The most popular models developed to characterise nonlinear transverse isotropy are based on the theory developed by Spencer (1984) for fibre-reinforced composites. He developed a constitutive hyperelastic model in which the strain energy function depends on a vector representing the material-preferred direction (a_0). Others further developed Spencer's theory to determine particular forms of the strain energy function to accommodate soft tissues with transversely isotropic properties such as tendons and ligaments (Weiss et al., 1996), arterial walls (Holzapfel et al., 2002) and cardiac tissue (Humphrey et al., 1990). These models have been developed and applied to represent soft tissues tensile behaviour. The models, respectively, developed by Weiss et al. and Holzapfel et al. assume that the fibres do not bear any load under compression, resulting in an isotropic behaviour in compression. This limits their applicability to represent skeletal muscle compressive behaviour, as our tests (see Section 4) show that, under compression, the muscle cross-fibre direction is stiffer than the fibre direction (anisotropic behaviour) and the stiffness is not equal to zero in the fibre direction. This restriction is not present in the model

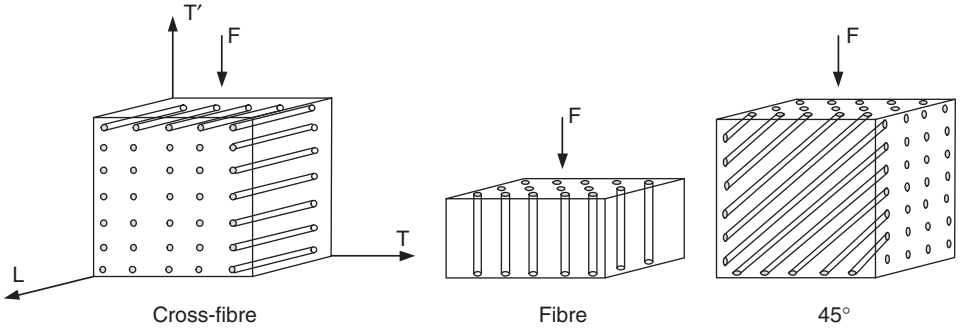

Fig. 2. Unidirectional composite and samples used for testing. F is the force applied. L is the fibre or longitudinal direction. T & T' are directions perpendicular to the fibres (transverse direction).

developed by Humphrey et al. who determined a constitutive relation for passive myocardium with a strain energy function (W) containing five parameters ($c_1, \ldots, c_5$):

$$W(I_1,\lambda) = c_1(\lambda-1)^2 + c_2(\lambda-1)^3 + c_3(I_1-3) + c_4(I_1-3)(\lambda-1) + c_5(I_1-3)^2, \quad (1)$$

where λ is the stretch ratio along the fibre direction and I_1 is a coordinate invariant measure of the deformation. Humphrey's model has been used successfully by different authors to model cardiac muscle behaviour in tension (see Novak et al., 1994; Sacks and Chuong, 1993). We have tested its applicability to model skeletal muscle behaviour in compression.

Li et al. (2001) used another approach to model nonlinear transverse isotropy in porcine aortic heart valves: they adapted the general theory of linear elasticity for transversely isotropic bodies, and introduced nonlinearity by means of strain dependent Young's moduli (SYM).

For a linear elastic, transversely isotropic material, with the L direction as the axis of transverse isotropy (see Fig. 2), the relationship between stress (σ) and strain (ε) can be expressed in the following form:

$$\begin{vmatrix} \varepsilon_L \\ \varepsilon_T \\ \varepsilon_{T'} \\ \varepsilon_{LT} \\ \varepsilon_{LT'} \\ \varepsilon_{TT'} \end{vmatrix} = G \begin{vmatrix} \sigma_L \\ \sigma_T \\ \sigma_{T'} \\ \sigma_{LT} \\ \sigma_{LT'} \\ \sigma_{TT'} \end{vmatrix}$$

with

$$G = \begin{vmatrix} 1/E_L & -\nu_{LT}/E_L & -\nu_{LT}/E_L & 0 & 0 & 0 \\ -\nu_{LT}/E_L & 1/E_T & -\nu_{TT'}/E_T & 0 & 0 & 0 \\ -\nu_{LT}/E_L & -\nu_{TT'}/E_T & 1/E_T & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/(G_{LT}) & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/(G_{LT}) & 0 \\ 0 & 0 & 0 & 0 & 0 & 2(1+\nu_{TT'})/E_T \end{vmatrix}. \quad (2)$$

The compliance matrix, G, has five independent elastic constants: two Young's moduli (E_L and E_T), two Poisson's ratios ($\nu_{TT'}$and ν_{LT}) and one shear modulus (G_{LT}). Index L denotes the longitudinal direction, indices T and T′ denote the two transverse directions.

According to Lemaitre and Chaboche (1990), the five characteristic coefficients can be obtained by uniaxial tension or compression tests on specimens cut out along (see Fig. 2):

- *the cross-fibre direction*: need to measure σ_T, ε_T, $\varepsilon_{T'}$; can determine E_T, υ_{TT}
- *the fibre direction*: need to measure σ_L, ε_L, ε_T ; can determine E_L, υ_{LT}
- *at 45° from the fibre direction*: need to measure σ_{45}, ε_{45} ; can determine G_{LT}

Soft tissues usually present a nonlinear relationship between stress and strain, and undergo large deformations in vivo. To introduce nonlinearity in the model, strain-dependent moduli have been introduced by Li et al. (2001):

$$E_L = k_1 \exp(k_2\,\varepsilon) \text{ and } \quad E_T = k_3 \exp(k_4\,\varepsilon), \quad (3)$$

where k_1, k_2, k_3, k_4 are the model parameters.

The properties in other directions are different from those in the longitudinal and transverse directions, but related to them. The following law can be used to determine the properties at any angle from the longitudinal direction:

$$\frac{1}{E_\alpha} = \frac{1}{E_L}\cos^4\alpha + \left(\frac{1}{G_{LT}} - 2\,\frac{\nu_{LT}}{E_L}\right)\sin^2\alpha\cos^2\alpha + \frac{1}{E_T}\sin^4\alpha. \quad (4)$$

Li et al. applied this model to reproduce the tensile behaviour of porcine aortic heart valves. In this paper, we have adapted it to represent skeletal muscle tissue compressive behaviour.

3. Test samples and experimental set-up

Uniaxial unconfined compression experiments have been conducted on porcine, bovine and ovine muscle samples. The bovine and a first set of porcine samples were obtained from a local butcher and had, therefore, been bled and hung (aged samples). No control over age, gender, time of death or methods of storage was available for these samples. In contrast, the ovine and the second set of porcine samples were obtained immediately after death from unbled animals (fresh samples). The ovine samples were excised from the thoracic limb of male sheep aged 10 months. The porcine fresh samples were excised from the pelvic limb of male pigs aged 3 months.

The compression tests, performed on a uniaxial test machine (Zwick ZO05, Zwick GmbH & Co. Ulm, Germany), were ramp functions applied to cubic samples of characteristic lengths between 5 and 10 mm. The samples were compressed up to 30% at the quasi-static strain rate of 0.05% s^{-1} and the tests were performed principally in three different orientations of the muscle: the fibre, cross-fibre and 45° from the fibre direction (see Fig. 2).

Each sample was tested once as the high strain level leads to some permanent tissue deformation, and no preconditioning was applied to the samples for the same reason. As illustrated, the samples used for compression in the fibre direction have a smaller height than the two

other types of samples. This is adopted in order to diminish buckling of muscle fibres that can be observed in samples compressed in the fibre direction. Samples with a ratio L/D (height/transverse dimension) approximating 1/2 are used.

Muscle tissue is quite mobile, and manual cutting leads to specimens with inconsistent geometry and orientation. In order to improve the cutting process, a cutting bench was developed (Fig. 3). The bench is equipped with a cutting blade (disposable single edge razor blade) mounted perpendicular to a stainless steel rotary table, on which samples are held in place by superficial freezing (Peltier Effect device mounted on the obverse of the table). Using the device, a 'cuboid' specimen with correct geometry and orientation can be obtained quickly.

Experiments were also conducted on fresh samples in order to characterise muscle sensitivity to time of test after death. Repeated non-destructive tests were performed on fresh ovine and porcine tissue at regular intervals after death of the animal.

4. Results

4.1. Aged tissue

Fig. 4 presents experimental stress–strain curves obtained by compression of aged porcine samples oriented in (a) the cross-fibre, (b) fibre and (c) 45° from the fibre direction. Four samples were tested in each direction and mean curves with standard deviation are presented. Fig. 4(d), in which mean curves in the three directions of testing are plotted, shows that the cross-fibre direction is stiffer than the 45° direction, which in turn is stiffer than the fibre direction.

The theoretical curves obtained with the Humphrey's model are represented in Fig. 5 (symbols). These parameters are determined by fitting to experimental data using nonlinear regression based on the Levenberg–Marquardt method (Marquardt, 1963). Using the method proposed by Humphrey et al. (1990), fitting is performed in two steps: in the first step, parameters c_3, c_4 and c_5 are determined by fitting to data obtained in the cross-fibre direction. Using this first set of parameters, parameters c_1 and c_2 are then determined by curve fitting of data obtained in the fibre direction. (Table 1)

Data obtained at 45° from the fibre direction are then used to check the predictive capabilities of the model. The results show that Humphrey's hyperelastic model does not adequately fit the data in all three directions of testing: good fit is obtained in the cross-fibre direction up to 30% strain ($R^2 = 0.99$). However, the model diverges from experimental data in the fibre direction after 20% strain, and cannot predict at all the behaviour of the muscle at 45° from the fibre direction.

Fig. 6 presents the same experimental results fitted with the SYM.

We choose a polynomial form for the two Young's moduli, E_L and E_T:

$$E_L = k_1 + k_2\varepsilon + k_3\varepsilon^2 \text{ and } \quad E_T = k_4 + k_5\varepsilon + k_6\varepsilon^2. \tag{5}$$

Parameter k_1, k_2, k_3 are determined by fitting to the data in the fibre direction. k_4, k_5, k_6 are determined by fitting to the data in the cross-fibre direction. Their values are presented in Table 2. Poisson's ratios of 0.45 are assumed and the shear modulus, G_{LT}, is determined

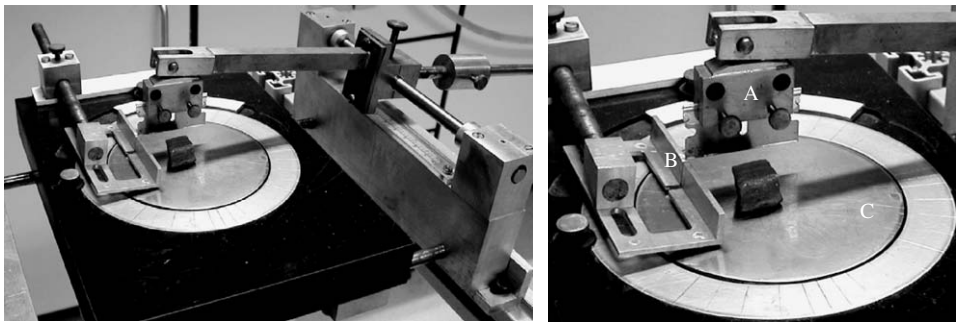

Fig. 3. Cutting bench: (A) blade, (B) support fence, (C) rotary table.

Table 1
Parameters for Humphrey's model fitted on aged porcine data (see Eq. (1))—mean ± standard error

c_1	c_2	c_3	c_4	c_5
0.948 ± 0.024	11.360 ± 0.085	0.880 ± 0.009	−0.070 ± 0.176	2.873 ± 0.039

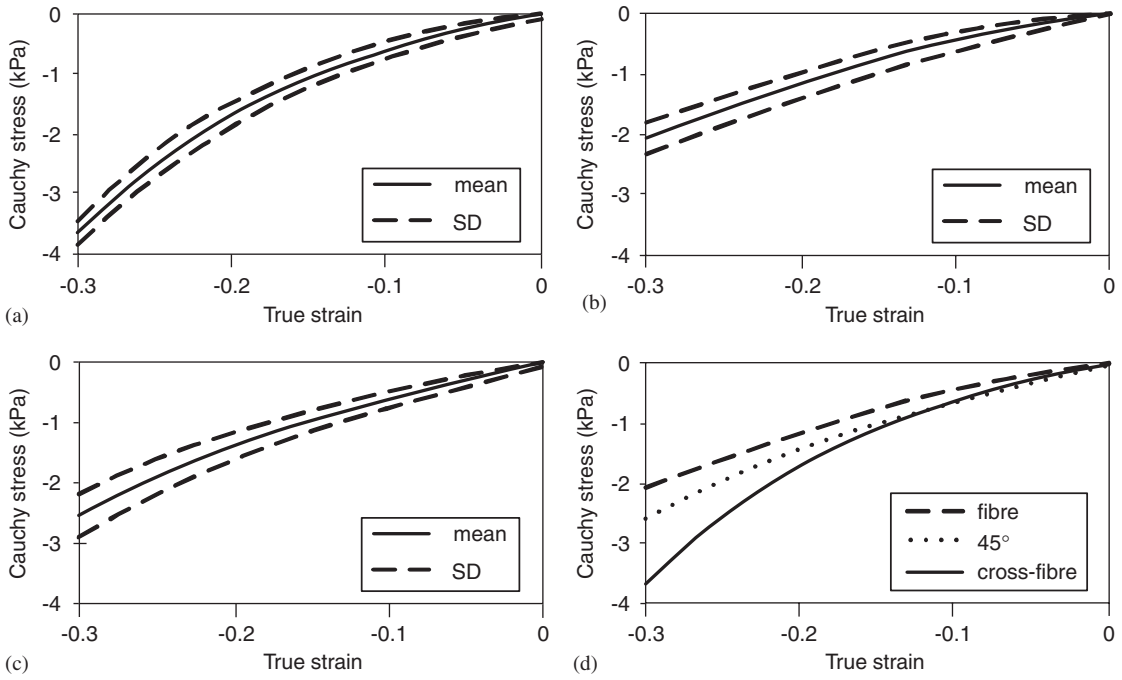

Fig. 4. Compression tests on aged porcine samples in the cross-fibre direction (a), fibre direction (b) and 45° from the fibre direction (c)—mean curves with standard deviation. (d) Mean curves for the three directions of testing.

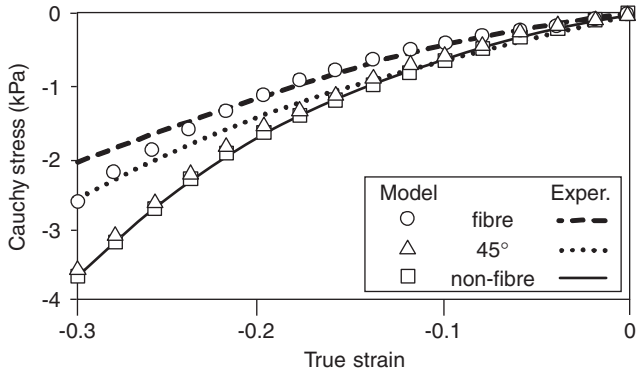

Fig. 5. Compression tests on aged porcine samples fitted with Humphrey's model. Lines denote experimental data and symbols denote the model.

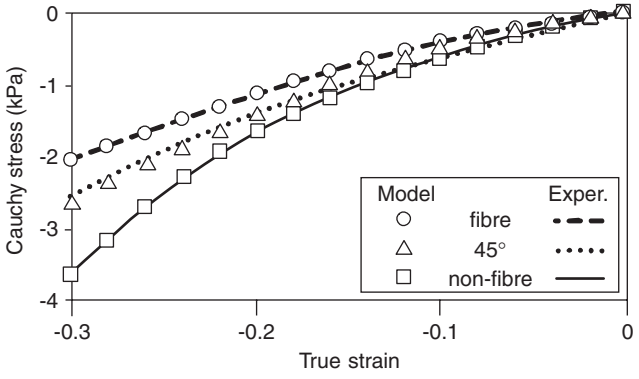

Fig. 6. Compression tests on aged porcine samples fitted with SYM. Lines denote experimental data and symbols denote the model.

by the following empirical expression, taken from Lekhnitskii (1981):

$$G_{LT} = \frac{E_L E_T}{E_T(1 + 2\nu_{LT}) + E_L}. \quad (6)$$

This formula was established for transversely isotropic rocks and is applied here as a first approximation.

The SYM gives a good fit to the experimental data in both the fibre and cross-fibre direction up to 30% strain ($R^2 = 0.99$ for each fitting). The model also yields a very good prediction of the behaviour of the muscle at 45° from the fibre direction (a mean error of 7% is observed between the experimental results and the model prediction). Similar results are obtained for aged bovine samples.

4.2. Fresh tissue

Data from the fresh porcine samples show a markedly different behaviour of the tissue under compression. Fig. 7 presents experimental stress–strain curves obtained by compression of fresh porcine samples oriented in (a) the cross-fibre, (b) fibre and (c) 45° from the fibre direction. Tests were performed on four pigs, aged 10–12 weeks. In each direction 15 samples were tested and the mean curves with standard deviation are presented. The tests were started within 2 h after animal

Table 2
Parameters for SYM fitted on aged porcine data (see Eq. (2) and (5))—mean ± standard error

k_1	k_2	k_3	k_4	k_5	k_6
2.825 ± 0.013	−20.150 ± 0.138	−1.290 ± 0.341	6.107 ± 0.034	15.790 ± 0.351	182.200 ± 0.865

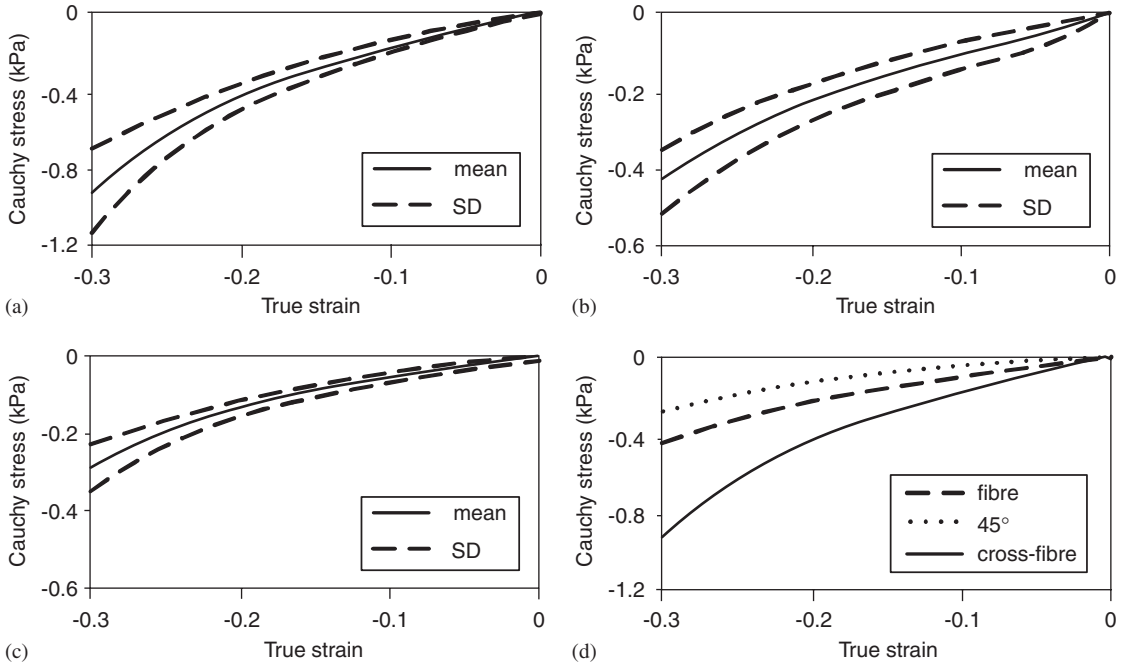

Fig. 7. Compression tests on fresh porcine samples in the cross-fibre direction (a), fibre direction (b) and 45° from the fibre direction (c)—mean curves with standard deviation. (d) Mean curves for the three directions of testing.

death. Fig. 7(d), in which mean curves in the three directions of testing are plotted, shows that as for the aged tissue, the cross-fibre direction is still stiffer than the 45° and fibre directions. However, the fibre direction is this time stiffer than the 45° direction. These results confirm preliminary data obtained on fresh ovine samples (Van Loocke et al., 2005).

Using the results of tests performed at 45° from the fibre direction and expression (4), a new form was determined for the shearing term of the SYM:

$$\frac{1}{G_{LT}} = 4\frac{\varepsilon_{45}}{\sigma_{45}} - \frac{1-2\nu_{LT}}{E_L} - \frac{1}{E_T}. \quad (7)$$

The experimental set-up was extended during these tests to calculate values of the Poisson's ratios: the displacement of markers placed on the samples (using quick-drying ink) was recorded with a video camera. The values presented in Table 3 were obtained for the different Poisson's ratios characterising the material.

As illustrated in Fig. 8, this model gives a very good fit to the experimental data in the fibre, the cross-fibre and at 45° from the fibre direction ($R^2 = 0.99$ for each fitting). The values of the model parameters are presented in Table 4. In order to assess the predictive capabilities of the model, further tests were also performed at 30° and 60° from the fibre direction (Fig. 9). Fig. 10 shows that the model also yields good prediction of muscle behaviour in these two directions (3.5% and 9.5% of mean error are observed between the experimental results and the model predictions, respectively, at 30° and 60° from the fibre direction).

Table 3
Values of Poisson's ratios—mean ± standard deviation

$\nu_{LT} = \nu_{LT}'$	ν_{TT}'	ν_{TL}
0.50 ± 0.02	0.64 ± 0.04	0.36 ± 0.03

To investigate the influence of time of death, samples kept at room temperature in airtight containers were compressed up to 10% strain at regular intervals after death of the animal. These tests were performed on fresh ovine and porcine muscle samples. Significant stiffening of muscle behaviour is noted a few hours after death of the subject: the maximum compressive stress measured

at 10% strain is increased approximately eight-fold for the ovine samples 10.5 h after death (Fig. 11). The same type of behaviour was observed in fresh porcine samples, with a maximum compressive stress at 10% strain increased approximately five-fold. We believe this is due to rigor mortis. After death, the slow formation of adenosine triphosphate (ATP) and its rapid consumption cause a total decrease of ATP level. When the ATP level falls to a specific level, the actin and myosin filaments combine irreversibly, and rigor mortis onsets (Kobayashi et al., 1996). In order to correlate this increase in stiffness with rigor mortis, the levels of ATP in tests samples will be determined at regular intervals in future tests.

5. Discussion

Three-dimensional models describing the behaviour of nonlinear, transversely isotropic soft tissues exist, but little experimental data was previously available to judge their applicability to model passive skeletal muscle, as existing data is mostly limited to tensile longitudinal behaviour. In this study, we provide an experimental procedure and a nonlinear model to characterise the three-dimensional compressive behaviour of passive skeletal muscle. Data on porcine muscle tissue tested in five orientations of muscle fibres (0°, 30°, 45°, 60° and 90°) are provided along with fitting parameters for the three-dimensional model. Our tests clearly demonstrate the non-isotropy and nonlinearity of mammalian skeletal muscles properties in compression. The model developed (SYM) provides a very good fit to the data in the cross-fibre direction, fibre direction and at 45° from the fibre direction ($R^2 = 0.99$). It also predicts well the behaviour of muscle tissue at 30° and 60° from the fibre direction. Moreover, the SYM uses engineering

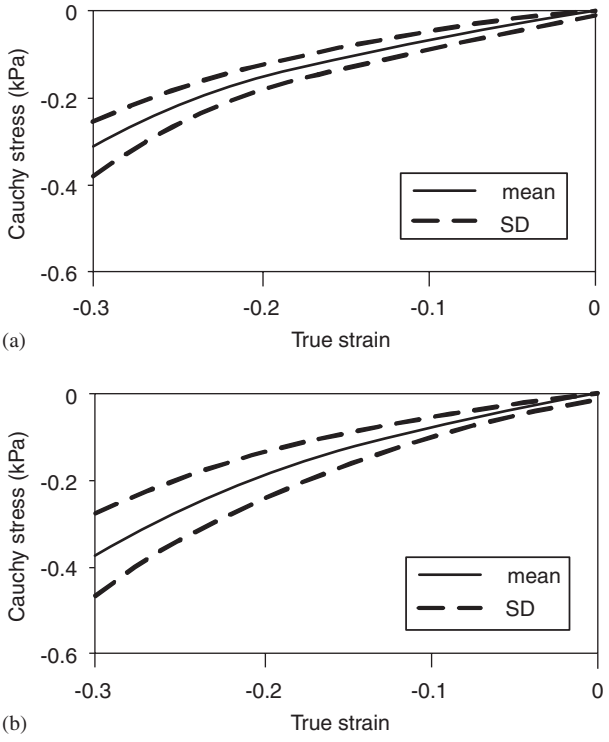

Fig. 9. Compression tests on fresh porcine samples at 30° (a) and 60° (b) from the fibre direction—mean curves with standard deviation.

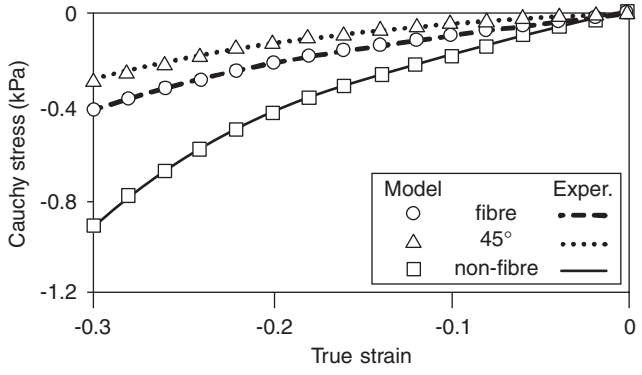

Fig. 8. Compression tests on fresh porcine samples fitted with SYM. Lines denote experimental data and symbols denote the model.

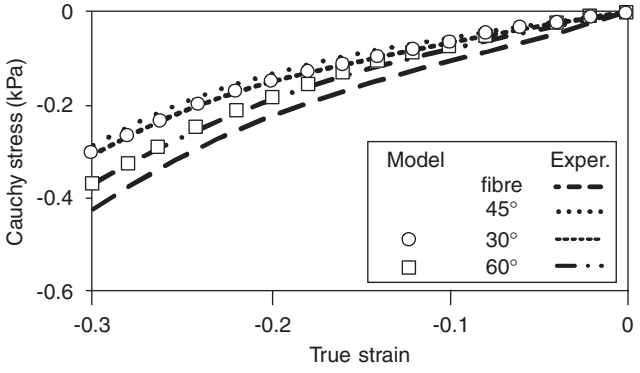

Fig. 10. Compression tests on fresh porcine samples fitted with SYM. Lines denote experimental data and symbols denote the model.

Table 4
Parameters for SYM fitted on fresh porcine data—mean ± standard error

k_1	k_2	k_3	k_4	k_5	k_6
1.495 ± 0.002	6.773 ± 0.023	28.60 ± 0.056	2.250 ± 0.007	10.850 ± 0.069	61.880 ± 0.170

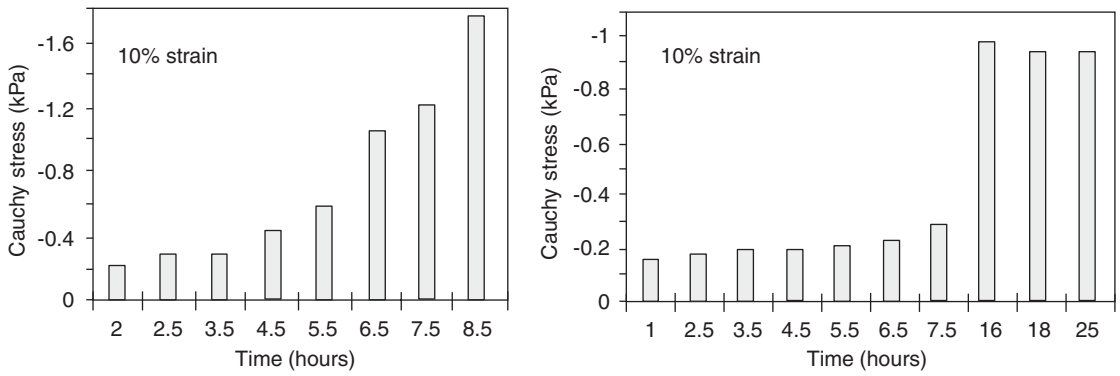

Fig. 11. Evolution of the maximum compressive stress with time after death for fresh ovine (left) and porcine (right) samples.

constants which have a direct physical interpretation, and the parameters of the models can be determined readily from uniaxial tests performed in three orientations of muscle fibres.

Fig. 12 presents a comparison between our results and data published in the literature. The two graphs present the same curves but different scales are adopted for clarity. The thick solid and the dashed curves were obtained by Bosboom et al. who conducted, respectively, in vitro Bosboom et al. (2001a) and in vivo Bosboom et al. (2001b) compression tests on rat tibialis anterior muscles in the transverse direction. The stress–strain curves represented on Fig. 12 were obtained from the quasi-static part of the model adopted by Bosboom et al. to fit their experimental results. The short-dashed curve was obtained by Zheng et al. (1999) from indentation tests on human forearm in vivo using an ultrasound indentation system with a pen-size hand-held probe. These experiments take into account the compressive properties of skin, fat and muscle tissues. The solid curve was published by Grieve and Armstrong (1988) who performed unconfined compression tests on aged porcine samples in vitro. Our results are also plotted on these graphs. Finally, the stress–strain curve obtained by Hawkins and Bey (1994) during extension tests performed on rat tibialis anterior muscles in the fibre direction is represented, to give an idea of the range of values where muscle properties are situated.

The results obtained by Grieve and Armstrong (1988) are in good agreement with our results. The data provided by Zheng et al. (1999) are slightly stiffer than our results, but this can be attributed to the fact that, in addition to muscle tissue, the properties of the skin and fat surrounding the muscle were also measured during these experiments. Besides, the tests were performed in vivo on adult human subjects. The results obtained by Bosboom et al. suggest, however, a much stiffer behaviour, which cannot be fully explained in the context of other researchers work. It could be partly due to the small relaxation time allowed for the tissues during the experiments (12 s for in vitro tests, 20 s for in vivo tests). Moreover, the difference observed between the in vivo and in vitro results leads one to wonder if, although the test animal was anesthetised, some involuntary muscle activity was present during the in vivo tests. The difference in stiffness observed between our post-mortally aged and our fresh tissue tests can be attributed to dehydration, period of storage after death, etc. of the tissues used for the aged tests, and also to the fact that young animals (10–12 weeks old) were used for the tests on fresh tissues. More tests will be performed in future to investigate the effect of age (before death) of the subject on muscle stiffness.

The difference in behaviour that was also observed with fibre orientation between aged and fresh tissue might be explained by the greater amount of fluid present in the fresh (unbled) samples, leading to a greater mobility of the connection between the fibres and resulting in shearing between adjacent fibre fascicles during compression. However, this is only based on experimental observations and requires further investigation. Aged muscles, where less fluid is present, show an increase in stiffness with an increase of angle from the fibre direction. Fresh tissues, containing a significant amount of fluid, show however that the weaker direction is at 45° from the fibre direction, where shearing has its most significant contribution. This difference in behaviour, as well as the stiffening noticed soon after death, emphasise the need to perform tests within a few hours after death of the subject in order to obtain data representative of living tissue properties.

Even with the variability inherent to biological structures, we manage to obtain data (on fresh porcine muscle) presenting relatively little scatter (10–15% on average). This is due to the fact that tests were performed on animals of the same origin, same sex, same age and that the same muscle was tested each time. Besides, the cutting device developed enables one to obtain samples with repeatable geometry, and care was taken to obtain samples with consistent structure, i.e. samples presenting a lot of fatty tissue were excluded from the tests.

To explain the values obtained for the Poisson's ratios characterising the material ($\nu_{LT} = 0.5$; $\nu_{TT'} = 0.64$; $\nu_{TL} = 0.36$) muscle anisotropy and incompressibility

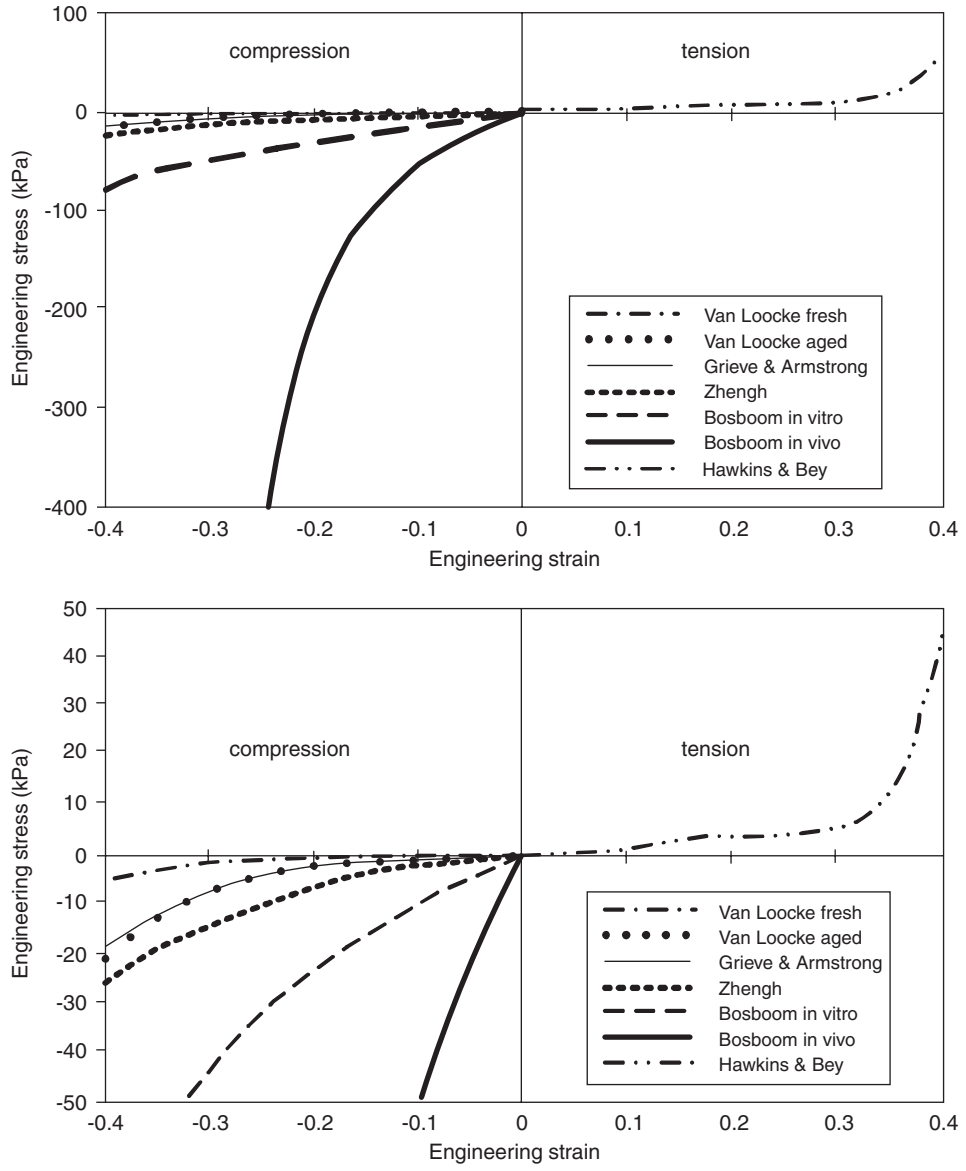

Fig. 12. Compressive properties of muscle tissue—comparison with published data.

have to be considered. During compression in the cross-fibre direction (T), both the fibre (L) and the other cross-fibre direction (T′) will expand. If the material was isotropic and incompressible, the amount of expansion would be the same in the fibre and in the cross-fibre direction, and the Poisson's ratios would all have a value of 0.5. However, for muscle tissue, the fibre direction is stiffer than the cross-fibre direction in expansion, so the material will expand less in the fibre than in the cross-fibre direction, leading to $v_{TL} < v_{TT'}$, and to a value greater than 0.5 for $v_{TT'}$.

The relationship, valid for linear transversely isotropic material, $v_{LT}/E_L = v_{TL}/E_T$ cannot be applied here as: $E_L < E_T$ in compression $\rightarrow v_{LT} < v_{TL}$ which is different from the behaviour observed during the tests. So, for muscle tissue in compression the compliance matrix becomes

$$G = \begin{vmatrix} 1/E_L & -v_{TL}/E_T & -v_{TL}/E_T & 0 & 0 & 0 \\ -v_{LT}/E_L & 1/E_T & -v_{TT'}/E_T & 0 & 0 & 0 \\ -v_{LT}/E_L & -v_{TT'}/E_T & 1/E_T & 0 & 0 & 0 \\ 0 & 0 & 0 & 1/(G_{LT}) & 0 & 0 \\ 0 & 0 & 0 & 0 & 1/(G_{LT}) & 0 \\ 0 & 0 & 0 & 0 & 0 & 2(1+v_{TT'})/E_T \end{vmatrix}. \quad (8)$$

The symmetry between off-diagonal terms does not apply here. It must also be added that the parameters of the model are not valid in tension at present.

The use of a PTFE lining diminished friction between the testing platens and the samples, but barrelling of the samples was still observed. The contribution of this

effect to the measured stress has not, as yet, been taken into account and will be characterised in the near future.

Only the quasi-static behaviour of muscle tissue has been investigated. However, in order to fully characterise the mechanical behaviour of muscle under impact, its viscoelastic properties must also be determined. This forms part of the future work to be undertaken in this project.

One of the important challenges of our research will be to relate the data obtained on animal muscle to human muscle properties. Due to the difficulty to obtain human tissue, in vivo tests using a non-invasive technique will be performed.

It is envisaged that the data provided in this paper will considerably improve the predictive capabilities of future finite element models.

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References

Best, T.M., McElhaney, J., Garrett Jr, W.E., Myers, B.S., 1994. Characterization of the passive responses of live skeletal muscle using the quasi-linear theory of viscoelasticity. Journal of Biomechanics 27, 413–419.

Bosboom, E.M.H., Thomassen, J.A.M., Oomens, C.W.J., Bouten, C.V.C., Baaijens, F.P.T., 2001a. A numerical experimental approach to determine the transverse mechanical properties of skeletal muscle. In: Middleton, J., Jones, M.L., Shrive, N.G. (Eds.), Computer Methods in Biomechanics and Biomedical Engineering-3. pp. Gordon and Breach Sciences Publishers, pp. 187–192.

Bosboom, E.M.H., Hesselink, M.K.C., Oomens, C.W.J., Bouten, C.V.C., Drost, M.R., Baaijens, F.P.T., 2001b. Passive transverse mechanical properties of skeletal muscle under in vivo compression. Journal of Biomechanics 34, 1365–1368.

Cole, G.K., van den Bogert, T., Herzog, W., Gerritsen, K., 1996. Modelling of force production in skeletal muscle undergoing stretch. Journal of Biomechanics 29, 1091–1104.

Davis, J., Kaufman, K.R., Lieber, R.L., 2003. Correlation between active and passive isometric force and intramuscular pressure in the isolated rabbit tibialis anterior muscle. Journal of Biomechanics 36, 505–512.

de Jager, M., 1996. Mathematical head-neck models for acceleration impacts. PhD. Thesis, Technische Universiteit Eindhoven, Eindhoven.

Forbes, P.A., Cronin, D.S., Deng, Y.C., Boismenu, M., 2005. Numerical human model to predict side impact thoracic trauma. In: Gilchrist, M.D. (Ed.), IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications. Springer, New York, pp. 441–450.

Gareis, H., Solomonow, M., Baratta, R., Best, R., D'Ambrosia, R., 1992. The isometric length–force properties of nine different skeletal muscles. Journal of Biomechanics 25, 903–916.

Gaudin, A.J., 1997. Muscle, anatomy. In: Encyclopedia of Human Biology, vol. 5, second ed. Academic, London, pp. 823–830.

Grieve, A.P., Armstrong, C.G., 1988. Compressive properties of soft tissues. In: de Groot, G., Hollander, A.P., Huijing, P.A., van Ingen Schenau, G.J. (Eds.), Biomechanics XI-A, International Series on Biomechanics. Free University Press, Amsterdam, pp. 531–536.

Hawkins, D., Bey, M., 1994. A comprehensive approach for studying muscle–tendon mechanic. Journal of Biomechanical Engineering 116, 51–55.

Hawkins, D., Bey, M., 1997. Muscle and tendon force–length properties and their interactions in vivo. Journal of Biomechanics 30, 63–70.

Hill, A.V., 1938. The heat of shortening and the dynamic constants of muscle. Proceedings of the Royal Society London 126, 136–195.

Holzapfel, G.A., Gasser, T.C., Stadler, M., 2002. A structural model for the viscoelastic behavior of arterial walls: continuum formulation and finite element analysis. European Journal of Mechanics A/Solids 21, 441–463.

Humphrey, J.D., Strumpf, R.K., Yin, F.C.P., 1990. Determination of a constitutive relation for passive myocardium. Journal of Biomechanical Engineering 112, 333–346.

Kobayashi, M., Takatori, T., Iwadate, K., Nakajima, M., 1996. Reconsideration of the sequence of rigor mortis through post-mortem changes in adenosine nucleotides and lactic acid in different rat muscles. Forensic Science International 82, 243–253.

Lekhnitskii, S.G., 1981. Theory of Elasticity of an Anisotropic Body. Mir Publishers, Moscow.

Lemaitre, J., Chaboche, J.-L., 1990. Mechanics of Solid Materials. Cambridge University Press, Cambridge.

Li, J., Luo, X.J., Kuang, Z.B., 2001. A nonlinear anisotropic model for porcine aortic heart valves. Journal of Biomechanics 34, 1279–1289.

Limbert, G., Middleton, J., 2004. A transversely isotropic viscohyperelastic material. Application to the modelling of biological soft connective tissues. International Journal of Solids and Structures 41, 4237–4260.

Marquardt, D.W., 1963. An algorithm for the least-squares estimation of nonlinear parameters. SIAM Journal on Applied Mathematics 11, 431–441.

Martins, J.A.C., Pires, E.B., Salvado, R., Dinis, P.B., 1998. A numerical model of passive and active behaviour of skeletal muscles. Computational Methods in Applied Mechanics and Engineering 151, 419–433.

Muhl, Z.E., 1982. Active length–tension relation and the effect of muscle pinnation on fibre lengthening. Journal of Morphology 173, 285–1292.

Novak, V.P., Yin, F.C.P., Humphrey, J.D., 1994. Regional mechanical properties of passive myocardium. Journal of Biomechanics 27, 403–412.

Sacks, M.S., Chuong, C.J., 1993. A constitutive relation for passive right-ventricular free wall myocardium. Journal of Biomechanics 26, 1341–1345.

Spencer, A.J.M., 1984. Continuum Theory of the Mechanics of Fiber Reinforced Composites. Springer, New York.

Van Loocke, M., Lyons, C.G., Simms, C.K., 2005. Three dimensional passive properties of muscle tissue in compression. In: Gilchrist, M.D. (Ed.), IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications. Springer, New York, pp. 313–320.

van Rooij, L., Bours, R., van Hoof, J., Mihm, J.J., Ridella, S.A., Bass, C.R., Crandall, J.R., 2003. The development, validation and application of a finite element upper extremity model subjected to airbag loading. Stapp Car Crash Journal, 47.

Verver, M.M., van Hoof, J., Oomens, C.W.J., Wismans, J.S.H.M., Baaijens, F.P.T., 2004. A finite element model of the human

buttocks for prediction of seat pressure distributions. Computer Methods in Biomechanics and Biomedical Engineering 7, 193–203.

Ward, E.E., Kleinverger, M., Lennon, A.M., Roberts, J.C., 2005. Modelling the effects of blast on the human thorax using high strain rate viscoelastic properties of human tissue. In: Gilchrist, M.D. (Ed.), IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications. Springer, New York, pp. 17–24.

Weiss, J.A., Maker, B.N., Govindjee, S., 1996. Finite element implementation of incompressible, transversely isotropic hyperelasticity. Computational Methods in Applied Mechanics and Engineering 135, 107–128.

Woittiez, R.D., Huijing, P.A., Boom, H.B.K., Rozendal, R.H., 1984. A three-dimensional muscle model: a quantified relation between form and function of skeletal muscles. Journal of Morphology 182, 95–113.

Zheng, Y., Mak, A.F.T., Lue, B., 1999. Objective assessment of limb tissue elasticity: development of a manual indentation procedure. Journal of Rehabilitation Research and Development 36, 2.