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Research Paper

Passive skeletal muscle response to impact loading: Experimental testing and inverse modelling

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ABSTRACT

Appropriate mechanical representation of passive muscle tissue is crucial for human body impact modelling. In this paper the experimental and modelling results of compressive loading of freshly slaughtered porcine muscle samples using a drop-tower testing rig are reported. Fibre and cross-fibre compression tests at strain rates varying from 11,600%/s to 37,800%/s were performed. Experimental results show a nonlinear stress–stretch relationship as well as a clear rate dependency of the stress. The mean (standard deviation) engineering stress in the fibre direction at a stretch of 0.7 was 22.47 kPa (5.34 kPa) at a strain rate of 22,000%/s and 38.11k Pa (5.41 kPa) at a strain rate of 37,800%/s. For the cross-fibre direction, the engineering stresses were 5.95 kPa (1.12 kPa) at a strain rate of 11,600%/s, 25.52 kPa (5.12 kPa) at a strain rate of 22,000%/s and 43.66 kPa (6.62 kPa) at a strain rate of 37,800%/s. Significant local strain variations were observed, as well as an average mass loss of 8% due to fluid exudation, highlighting the difficulties in these kinds of tests. The inverse analysis shows for the first time that the mechanical response in terms of both applied load and tissue deformation for each of the strain rates can be captured using a 1st order Ogden hyperelastic material law extended with a three-term quasilinear viscoelastic (QVL) expansion to model viscoelastic effects. An optimisation procedure was used to derive optimal material parameters for which the error in the predicted boundary condition force at maximum compression was less than 3% for all three rates of testing (11,600%/s, 22,000%/s and 37,800%/s). This model may be appropriate for whole body impact modelling at these rates.

1. Introduction

Skeletal muscle accounts for about 40% of body mass (Chomentowski et al., 2011; Salem et al., 2006) and an understanding of its mechanical properties is therefore important for impact biomechanics research. However, the multi-axial compressive properties of fresh skeletal muscle at rates experienced during typical sports and automotive impacts are not well understood (Van Loocke et al., 2009). Therefore, although finite element human body models including muscle tissue are routinely used in impact biomechanics research, their utility remains limited mainly by uncertainties in the constitutive representation of the soft tissues.

Skeletal muscle is composed of about 70–80% water, 3% fat and 10% collagen (Vignos and Lefkowitz, 1959). It exhibits a fibre-oriented structure, with each muscle composed of fascicles

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containing bundles of fibres which themselves are composed of parallel bundles of myofibrils. There is a dense network of connective tissues which surrounds groups of fibres and this has a hierarchical structure. Arising from this structure, skeletal muscle presents a nonlinear elastic, non-homogeneous, anisotropic and viscoelastic behaviour (Van Loocke et al., 2006, 2008, 2009).

The need for experimental data for impact biomechanics applications can be assessed by the following order of magnitude comparison: in a 48 km/h unrestrained frontal vehicle impact, assuming the occupant's body strikes the vehicle interior at 48 km/h, the tissue compression rate is 13.33 m/s. For a mid-body region muscle thickness of 5 cm, this represents a compression rate of around 25,000%/s (Van Loocke et al., 2009), but combined fibre/cross-fibre data at these rates are not generally available.

Very high rate tests (up to 100,000%/s) have been performed by several researchers, but they have not assessed the effects of fibre direction (Van Sligtenhorst et al., 2006; McElhaney, 1966)). Song et al. (2007) did assess the influence of fibre direction for load ranges between 0.7%/s and 370,000%/s (Song et al., 2007), but their data surprisingly show identical responses for strain rates of 0.7%/s and 7%/s. Others have tested volunteers and cadavers (Dhaliwal et al., 2002; Muggenthaler et al., 2008) but, in the absence of inverse modelling, this approach only yields structural parameters. Chawla et al. (2009) performed compressive impact tests and inverse analysis on surgical scraps of human tissue which had been frozen and then thawed and the state of rigor mortis is unfortunately unknown in these tests.

Finite element analysis (FEA) models for whole body impact simulations have been developed in order to understand injury mechanisms (Maeno and Hasewaga, 2001; Kuwahara et al., 2007; Han et al., 2012). However these models depend on the chosen constitutive models for the tissues and their parameters, and both of these remain a significant limitation for whole body impact simulations. In most cases, an isotropic linear elastic model is used with parameters based on transverse compressive testing only. However, this is insufficient to model the anisotropic response observed in quasi-static testing (Van Loocke et al., 2006) as well as dynamic testing (Van Loocke et al., 2008, 2009). Thus, compressive testing in multiple directions at the relevant rates is required to guide appropriate model development. The attendant rate dependency of skeletal muscle also needs to be incorporated into these FEA models. Accordingly, measuring the rate dependent multi-axial compressive properties of skeletal muscle and providing input data for constitutive models at rates relevant to automotive accidents are important research goals and these form the focus of this paper.

2. Methods

2.1. *Experimental procedures*

The following sections describe the specimen preparation and the experimental procedure. More details of the experimental procedure, test-rig validation and the raw experimental data are available in Takaza and Simms (2012). However, in addition to the inverse modelling, significant additional experimental data for model verification are presented here.

2.1.1. *Specimen preparation and mechanical loading*

Freshly harvested porcine *Longissimus dorsi* samples were prepared from 3 month old female pigs. Preparation of specimens with a specific defined uniform shape was difficult due to the soft nature of the tissue and the interaction of the microstructural components of matrix and fibres, and this problem has been previously noted (Van Loocke et al., 2006). Samples were compressed either in the cross-fibre (face A or C in Fig. 1 Left) or the fibre direction (face B in Fig. 1 Left). Due to the difficulties involved in accurately cutting fresh muscle tissue, cross-fibre samples were approximately cubic, and for the fibre direction tests, the samples were cuboid with width and depth nominally at 20 mm to reduce buckling effects

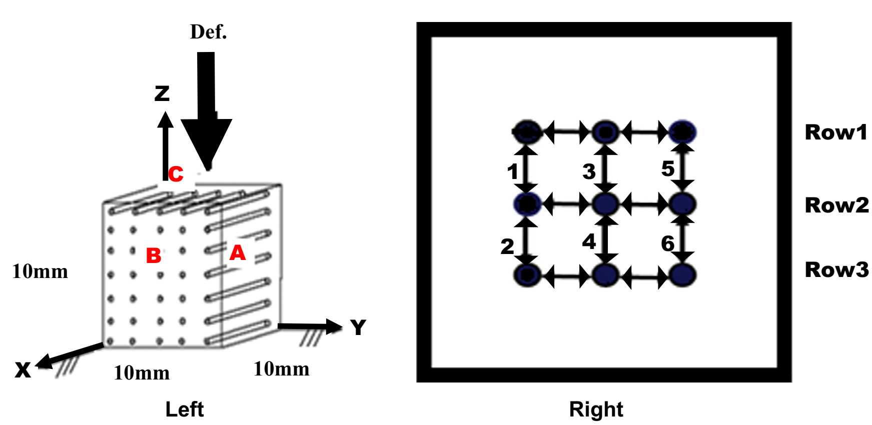

Fig. 1 – Left: schematic of a specimen illustrating the cross-fibre and fibre directions. The surfaces are marked A (cross-fibre direction), B (fibre direction) and C (cross-fibre direction). Right: schematic for the regions of interest used for strain analysis (reproduced from Takaza and Simms (2012)).