Modelling of fascia lata rupture during tensile tests via the discrete element method

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1. Introduction

Musculoskeletal models are often used to better understand the behaviour of the many components of the body, to predict injuries and design safety devices. These models become increasingly more detailed by including connective tissues. However, their implementation is often challenging due to the small amount of data regarding their mechanical properties.

Fascia lata is a connective tissue wrapped around the muscles of the thigh. Histological studies on goat fascia lata (Pancheri et al. 2014) have shown its heterogeneous structure composed of two layers of collagen fibres connected by a proteoglycan matrix. Both layers have a specific fibre orientation, and are not orthogonal.

The discrete element method (DEM) makes it possible to model complex structures by discretizing them into simple elements, associated to a mass, and cohesive bonds, such as springs. This allows for a better understanding of the relationship between micro and macroscale properties of materials. While this method is usually used to model composite materials, it has been adapted to model biological materials, such as the muscle-tendon complex (Roux et al. 2016).

The aim of this study is thus to reproduce numerically the macroscopic behaviour of the fascia lata submitted to a tensile test using DEM.

2. Methods

2.1. Geometrical construction of the model

GranOO (Granular object oriented, www.granoo.org) software was used to build the fascia lata model using microstructural dimensions extracted from the microscopic view and histological sections made by Pancheri et al. 2014. The microstructure of fascia lata was discretized using spherical elements linked with spring bonds arranged in two layers. Spherical discrete elements were put at each crossing of fibres, with an interval of 143 µm along the longitudinal orientation and an interval of 132 µm along the transverse orientation. As the longitudinal fibres are larger and arranged on a different layer than the transverse ones, elements on the longitudinal layer have a diameter of 162 µm, while those on the transverse layer a diameter of 75 µm.

The collagen fibres and the proteoglycan matrix were modelled as follows. Each element is linked by a spring to its 17 closest neighbours: the eight neighbouring elements on the same layer and the nine elements facing them on the opposite layer. Among these 17 spring bonds, two of them model a portion of collagen fibre, linking the element to its neighbours along the corresponding fibre orientation, while the 15 other bonds model the proteoglycan matrix (Figure 1(a)).

2.2. Mechanical properties

Each spring stiffness was computed from the Young modulus, cross-sectional area and initial length of links. The Young modulus of the matrix was the same as that used by Roux et al. to model the extracellular matrix in muscle, 0.1 MPa. A Young modulus of 44 MPa for collagen fibres was computed using the apparent Young modulus of fascia lata assessed by Henderson et al. (2015) and the previously mentioned modulus for the matrix via an estimation of fibre-matrix ratio in a section. Longitudinal springs are given a broader cross-section than transverse springs and are thus stiffer.

The strain limit causing the failure of the springs for collagen fibres was set at 0.4, corresponding to some of
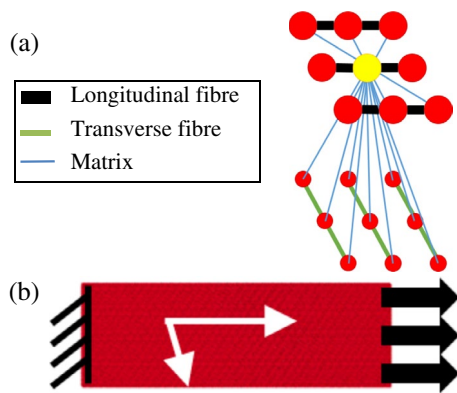
Figure 1. (a) Illustration of all bonds for the yellow element and the neighbouring fibres. (b) Model of fascia lata sample and boundary conditions. The arrows indicate fibre orientation.

Figure 2. Force-displacement curve from uniaxial tensile test simulation along longitudinal fibres.

the lowest strain limit to rupture in literature (Shen et al. 2008). Moreover, due to the lack of data on the strain limit of the matrix, it was once again set at the same value used by Roux et al. (2016), i.e. 0.3.

2.3. Numerical tensile test simulation

Numerical simulations were carried out to replicate the tensile tests up to rupture performed by Henderson et al. 2015; The model dimensions were thus taken according to their samples, meaning 3 × 1 cm rectangular samples. One extremity of the model was fixed, allowing no displacement, while a linear displacement was imposed to the other one at a quasi-static strain rate (Figure 1(b)). The force induced in the springs was measured at each step.

3. Results and discussion

Evolution of the force induced in the springs with regards to the imposed displacement is provided on Figure 2. This curve shows that fascia lata is a rather brittle material, hardly showing any yield portion, much like the results obtained by Henderson et al. in their study of canine fascia lata. However, as the current model does not take into account any slack for the fibres and considers them ready to be stretched, the curves do not show any toe/heel region, which is characteristic of most biological materials. Furthermore, while the force at rupture of 35 N was about 2/3 lower than the one found experimentally by Henderson et al., the maximum displacement of 7 mm is of the same order of magnitude. The lack of data on the mechanical properties of the proteoglycan matrix led to using the properties of similar tissues and may explain this underestimation of the force.
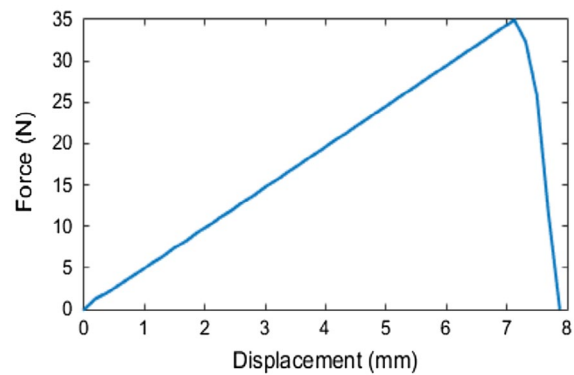
The biggest advantage of using DEM is to assess the rupture phenomenon at a microstructural scale. Failure occurred following the delamination between the fibres once the shear stress was sufficient to tear them from one another.

4. Conclusions

While the results of these simulations are still preliminary, they are encouraging as they seem to show a similar behaviour as the one found experimentally. Use of DEM is promising and our next step is to analyse the influence of various microstructural parameters, such as fibre orientation or mechanical properties distribution, on the global mechanical properties of fascia lata. The model should be able to be adapted to other fasciae as well, but it still needs to take into account slack length of the fibres and more accurate material properties.

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