

A MUSCLE MODEL THAT CAPTURES EXTERNAL SHAPE, INTERNAL FIBRE ARCHITECTURE, AND PERMITS SIMULATION OF ACTIVE CONTRACTION WITH VOLUME PRESERVATION

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1. ABSTRACT

A parametric B-spline solid model is used to produce a combined model of muscle that captures both its shape and physical properties. Initially, a data-fitting procedure is designed to allow reconstruction of virtual models of muscle from cadaveric specimens of human soleus, capturing its external boundary shape and internal fibre architecture. In contrast, muscle models created from cross-sectional image slices often only contain boundary information. Modelling internal fibre arrangements permits novel visualizations of changing pennation patterns within a volume of muscle, as well as more detailed analysis of pennation angle distribution and fibre lengths within a single muscle. This allows finer resolution of muscle tissue than possible with lumped-sum parameter models that represent an entire muscle with a single representative pennation angle or fibre length. Traditionally, finite element analysis can be used to perform physical simulation of soft tissue. However, these simulations become computationally prohibitive as accuracy needs increase. The B-spline solid model produces a compact representation for muscle that allows faster simulations of muscle contraction. A Lagrangian dynamic formulation for the equations of motion can model both global deformation forces, such as volume preservation, as well as locally-different material properties such as aponeurosis tissue adjacent to muscle fibres. We discuss several proposed experiments to compare our model with real, active contracting muscle *in vivo*.

2. INTRODUCTION

In order to gain understanding and insights into muscle function, parameterized models are developed that can be used in simulations to attempt to reproduce observed

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behaviour. Validations of the model with experiments can help to confirm basic relationships between muscle parameters and the observed performance of the model in various situations. The Hill model has been used by biomechanists to study musculo-skeletal simulations of various activities in humans and other animals by incorporating

observed dependencies between muscle force, length and contraction velocity (Zajac, 1989). These models have been successful in qualitatively matching measured EMG signals for a variety of human motions, but resolve muscles to a few lumped-sum parameters and incorporate muscle architecture only in the form of a single pennation angle. Others have used idealized volumetric models such as parallelepipeds that are limited to the specific situation of an individual muscle (Woittiez *et al.*, 1984). We develop a general model for muscle that combines volumetric shape and structural properties with physical force-generating capability. A methodology is introduced that can be used to create simulated functional muscle models from actual cadaveric muscle specimens. This can eventually allow comparisons between an actual muscle and its virtual counterpart on shape and force-generating criteria. By selecting volumetric B-spline solids as our mathematical foundation, we are able to represent a rich variety of shapes while capturing correct fibre architecture.

Although finite element approaches can be used to represent detailed muscle simulations (Chen and Zeltzer, 1992), they can be computationally expensive because of the large number of elements that are needed to capture muscle contraction behaviour. Equations of motion of the parameters of the B-spline model can be formulated to produce simulations of contracting muscle tissue while undergoing global shape constraints such as volume preservation.

3. B-SPLINE SOLID MATHEMATICAL FORMULATION

The B-spline solid model (Fig. 1) allows smooth shapes to be defined with a compact set of control point parameters, \mathbf{q} :

$$\mathbf{x}(u, v, w) = \sum_i \sum_j \sum_k B_i^u(u) B_j^v(v) B_k^w(w) \mathbf{q}_{ijk}, \quad (1)$$

where i, j, k index the control point lattice and basis functions of the triple B-spline tensor product. The evaluated points, $\mathbf{x}(u, v, w)$ are the Cartesian coordinates that make up the boundary and volume of the solid being modelled. Given n control points, we can select n material coordinates $\mathbf{u}_{max,I} = (u, v, w)_I, I = 0, \dots, n-1$, that map to a set of n spatial points, \mathbf{x} :

$$\mathbf{x} = \mathbf{B}(\mathbf{u}_{max})\mathbf{q}. \quad (2)$$

The matrix \mathbf{B} contains the triple tensor products of Eq. (1) while the coordinates of \mathbf{u}_{max} are chosen to coincide with the parameters that produce the maxima of each of the B-spline basis functions. Intuitively, the position of a spatial point in \mathbf{x} , would be most influenced by the control point in \mathbf{q} , in the same corresponding rows of Eq. (2).

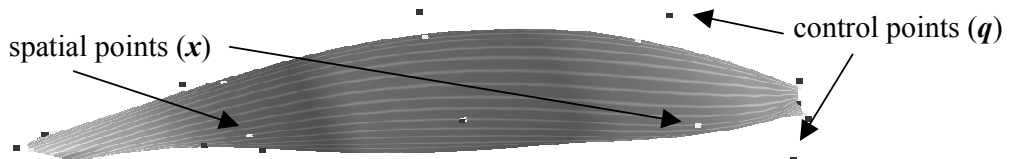


Fig. 1: B-spline solid muscle with control points (dark) and spatial points (light).

4. DATA-FITTING PROCEDURE

We have developed a methodology for allowing the construction of deformable models of muscle from various sources of actual muscle data. A *continuous volume sampling function*,

$$CVSF(\tilde{u}, \tilde{v}, \tilde{w}), \text{ where } (\tilde{u}, \tilde{v}, \tilde{w}) \in [0, 1] \times [0, 1] \times [0, 1] \quad (3)$$

is defined from digitized 3-D data from actual muscle specimens to generate the spatial points of \mathbf{x} . The normalized coordinates, $(\tilde{u}, \tilde{v}, \tilde{w}) \in ([0, 1] \times [0, 1] \times [0, 1])$, span the entire volume of the muscle model, allowing any arbitrary sample to be generated from the *CVSF* function, with the property of interpolating the original digitized 3-D data points. These spatial points are used to solve for \mathbf{q} using the inverse of Eq. (2) to define the B-spline solid model in Eq. (1):

$$\mathbf{q} = \mathbf{B}^{-1}(\mathbf{u}_{max})\mathbf{x}. \quad (4)$$

In previous studies (Ng-Thow-Hing *et al.*, 1998), we have reconstructed models from transverse image slices of the Visible Human data-set. Since only the boundary curves of the cross-section of muscle can be reliably extracted, image slices cannot be used to reconstruct internal representations of fibre architecture.

The procedures for fitting a B-spline solid to digitized fiber sets are outlined in this section. A serial dissection and digitization procedure describe previously (Ng-Thow-Hing *et al.*, 1998) was used to systematically reveal and digitize the end-points of representative muscle fibres of human soleus muscle by performing a series of dissections that reveal layer by layer the muscle's fibre arrangement within its volume. The result of this procedure revealed three architecturally-distinct fibre regions which are termed *posterior*, *anterior*, and *marginal* regions (Fig. 2).

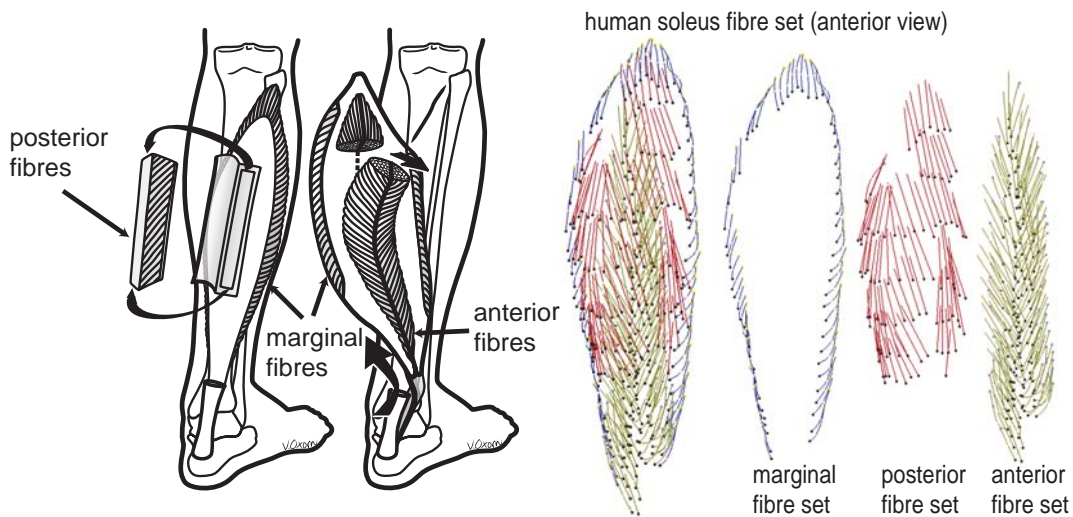


Fig. 2 Left: The three architectural regions of human soleus (illustration by Valerie Oxorn) . Right: Digitized fibre sets for different architectural regions of the human soleus (anterior view).

4.1 Fibre set CVSF construction

Each fibre visualized in Fig. 2 has three indices, $F_{i,j,k}$ representing the k th point of the j th fibre in the i th layer of serial dissection. Using this indexing scheme and the 3-D coordinates of fibre points, we design the *CVSF* from Eq. (2) so that it contains the sampled fibre points and can generate novel points that are interpolations of the original data fibres. The *CVSF* is used to generate the necessary number of spatial points to solve a linear system for the number of control points in the given B-spline solid model.

4.2 Fibre measurements

With iso-parametric streamlines representing fibres, we can compute the arclength to estimate fibre length using numerical integration. Fig. 3 illustrates computed lengths of fibre streamlines generated in a model of the posterior soleus region. The pennation angle a fibre's orientation has with respect to the tendon is traditionally approximated as a single, lumped average for the entire muscle (Zajac, 1989). Observations taken from the dissected soleus muscle show that pennation angle can vary significantly within a single muscle. The existence of a 3-D muscle model motivates the possibility of developing a more accurate method to measure pennation angle. Pennation angle is usually measured by examining the fibre angle within a planar cross-section of muscle relative to the tendon attachment area. Mechanically, the pennation angle of the fibre should be measured relative to the local tangent plane of the tendon aponeurosis that it is attached as seen in the right of Fig. 3.

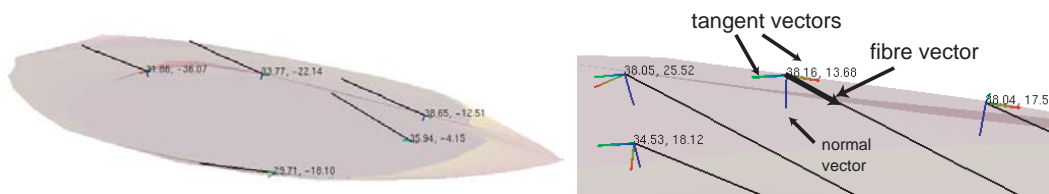


Fig. 3 Fibre measurements such as fibre length and pennation angle can be computed analytically on the virtual muscle model. Sampled fibres are measured for fibre length and pennation angle and the measurements are labelled as a pair, (fibre length, pennation angle). The right image illustrates how the pennation angle can be calculated by measuring the angle the fibre vector makes with the local tangent plane.

Once the B-spline solid is fit to the fibre set data, we can visualize arbitrary fibres within the solid by generating streamlines from iso-parametric curves in the solid (Fig. 4). To distribute fibres evenly throughout the muscle's volume, a Sobol sequence is used to generate the fibre samples (Press, 1992).

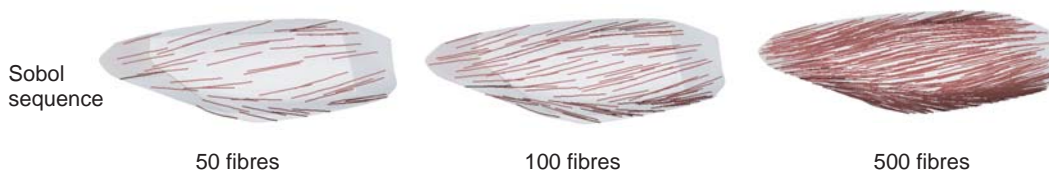


Fig. 4 An arbitrary number of fibres can be generated in the constructed virtual muscle model.

5. PHYSICAL MODELLING

The B-spline solid's compact control point representation allows us to express the equations of motion of these control points, \mathbf{q} , using the Lagrangian formulation:

$$\frac{d}{dt} \frac{\delta T}{\delta \dot{q}_i} - \frac{\delta T}{\delta q_i} = Q_i - \frac{\delta V}{\delta q_i}, i = 0, \dots, n-1, \quad (4)$$

where the q_i are the components of the vector \mathbf{q} . Alternatively, we can also express equations of motion in terms of spatial points, \mathbf{x} . T and V correspond to kinetic and potential energy expressions, with Q_i as a generalized force acting on the control or spatial points. By modelling various generalized force expressions and potential energy functions, physical behaviour and response can be specified for muscle. The volume of B-spline solids has a closed-form expressed as a function of control points or spatial points (Ng-Thow-Hing, 2001): $Volume = V(\mathbf{q}) = V(B^{-1}\mathbf{x})$. Given a target volume, V_0 , with the gradient of the volume, a volume-preserving force can be computed:

$$F_{volume-preserving} = k(V(\mathbf{q}) - V_0)\nabla_{\mathbf{q}}V(\mathbf{q}). \quad (5)$$

The coefficient k can be used to control the relative strength of volume preservation. Various force models for Q_i can be specified to add damping terms, collision-contact forces or fibre contractile forces to the muscle model. In particular, viscoelastic links between spatial points can be added for muscle contractile forces to simulate contraction. The links can be added to generate forces in the same directions as the fibre architecture. Simulation details are described in the next section.

6. RESULTS

We have generated simulations of muscle contraction on virtual models of posterior soleus obtained from Section 4. Fig. 5 illustrates simulations of posterior soleus contractions with and without an aponeurosis (or tendon plate) attached. The muscle force of each contractile fibre has its own parameterized Hill model, allowing a nonuniform distribution of contractile forces within the same muscle. The aponeurosis is modelled as an elastic sheet that restricts deformation of the soleus along the surfaces where it is attached. Global volume-preserving forces defined in Eq. 5 are added to preserve volume. With volume preservation, the fibres changed orientation as they contracted, matching behaviour seen with real soleus muscle using ultrasound imagery.

7. DISCUSSION

The B-spline solid model allows both the form and function of muscle to be integrated within a single mathematical framework. Its relatively compact representation can be used to accelerate simulations because of the smaller size of the equations of motion compared to finite-element methods. However, the value of parameters of the various force models that can be applied to the muscle model needs to be estimated by observing actual muscle specimens undergoing contraction. Currently, this can only be done with medical imaging technology that can be recorded fast enough to get a time series of contracting muscle, such as CINE MRI.

Although the serial dissection procedure we used is time-consuming and tedious, it guaranteed that we could properly capture the complete architectural structure of the soleus muscle we modelled. The lack of adequate non-invasive imaging technology to discriminate internal fibre structure makes serial dissection the only feasible solution that unfortunately is restricted to cadaveric specimens. Nevertheless, the ability to embed active, physical properties to produce animated versions of muscle from static specimens is a useful approach to overcoming the limitations of the original, non-active muscle specimens we used to create the model.

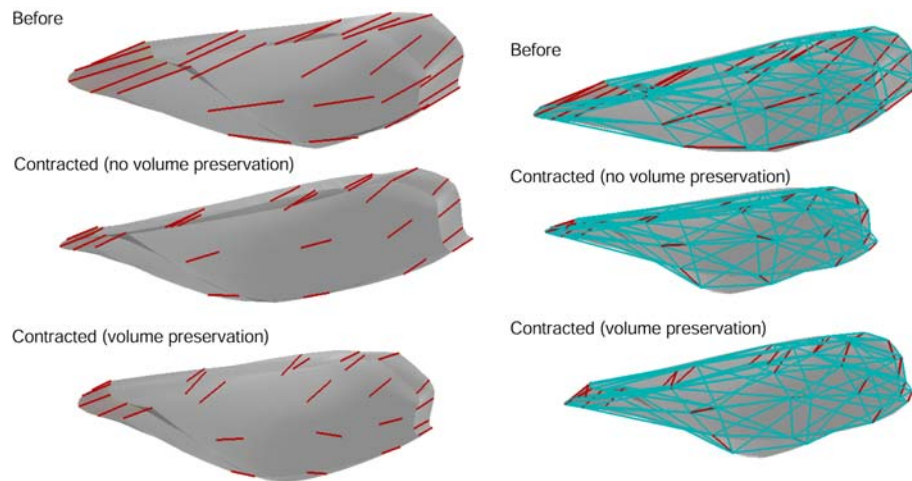


Fig. 5 Left: Soleus model with no aponeurosis. Right: Soleus model with aponeurosis. Simulation runs at a rate of 14.5 Hz on a Pentium III 400 MHz CPU.

8. REFERENCES

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