Technologies for Modelling Fibrous Muscle in Motion

J Kohout[1], G J Clapworthy[2], Y Tao[2], F Dong[2], P Kellnhofer[1], D Cholt[1], Y Zhao[2]

[1] University of West Bohemia, [2] University of Bedfordshire

Correspondence: besoft@kiv.zcu.cz, Department of Computer Science and Engineering, University of West Bohemia, Univerzitní 8, Plzeň, Czech Republic

1. Introduction

Millions of people around the world are affected by musculoskeletal diseases such as osteoporosis and neuromuscular disorders which, in most cases, are incurable. Treatment is generally ameliorative, involving a change in lifestyle supported by rehabilitation to improve the quality of patient life. To improve the effectiveness of the treatment, we need to gain a deeper understanding of the physiology of muscles and, for this, the EC-funded Integrated Project VPHOP: the Osteoporotic Virtual Physiological Human [1] is developing a modelling technique that simulates the dynamic behaviour of muscles and muscle fibres during different motion activities.

A companion paper [2] provides an overview of the various aspects of the model being developed; this paper looks in more detail at novel features of the model that make it such a promising tool for use in the clinical environment.

2. Context

Traditionally, biomechanics modelling of muscle action during motion has represented the muscles by action lines; these are formed of straight or piecewise straight lines joining the points at either end of the muscle where it is attached to the skeletal bones. While this representation allows for the actions of multiple muscles to be demonstrated without occlusions, it does not provide information about essential features of the muscle dynamics, one of which is the changing orientation of the muscle fibres during the motion; this is important because of the anisotropic nature of muscle tissue, which can apply force only by contraction along the fibre length.

Existing methods to provide volumetric musculoskeletal models represent a muscle by a B-spline solid in which the iso-lines correspond to muscle fibres [3] and movement is made possible by manipulation of its control points, or by a 3D finite-element mesh in which the cells contain information about the direction of the muscle fibres present in its volume [4] and the vertices move in reaction to the external force induced by the movement of the bones. Although results from these methods show good agreement when compared with static MRI images taken in different postures, the use of these models in the clinical context is highly impractical because generating the meshes is a complex process easily requiring several days for a highly skilled operator, and computing the solution requires several hours on a supercomputer [4].

Our approach, which is inspired by techniques commonly used in computer graphics, is completely different. In the basic model [2], a muscle is represented by a triangulated surface mesh extracted from MRI images. It is associated with two sets of landmarks denoting the sites at which the muscle is attached to the bone by a tendon and with a predefined unit cube template containing a description of the fibre geometry – these templates were proposed by Blemker & Delp [4]; see Figure 1. To define the fibres of the muscle, two different strategies can be applied: *surface-first* or *fibres-first*. In the former, as the skeleton moves, so the shape of the muscle surface mesh changes in such a way that the muscle volume is conserved and inter-penetration with bones and other muscles is avoided so that the muscles wrap properly. The deformed muscle is then decomposed into muscle fibres as specified by the template of the fibre geometry – details are given below. The latter strategy starts with the decomposition and then transforms the set of fibres produced into a mass-spring system whose end-nodes are fixed to the bones so that when the bones move, the equilibrium of the system is violated and this triggers a recalculation of the positions of the inner nodes; again, penetration between a node and a bone surface is avoided.

Both strategies have been implemented in C++ (MS Visual Studio 2010) under the Multimod Application Framework (MAF) [5], which is a visualisation system based mainly on VTK [6], integrated into LHPBuilder software being developed within the VPHOP project [1] and tested on a data of a female walking.

3. Muscle Decomposition

Our decomposition method [7], which is used in both strategies, starts with the production of poly-line muscle fibres of the requested number and resolution within a unit cube according to the fibre geometry template [4] specified for the muscle being decomposed. Next, the unit cube (with all its poly-lines) is subjected to an affine transformation such that the transformed cube is an oriented bounding box (OBB) of the muscle, and the attachment sites of the fibres in cube are aligned as well as possible with those specified for the muscle (as landmarks). After this, the transformed cube is sliced and the contours that arise from the slicing are morphed on to the contours of the muscle obtained by the same slicing,

employing the technique described by Ju et al. [8] to ensure consistency between the slices. This maps the fibre vertices into the interior of the muscle.

In the initial model developed, the paths generated for the muscle fibres were often unrealistic in the proximity of the attachment areas if these areas were large – the fibres tended to meet at a common point instead of spreading over the whole area. To correct this, we remove the part of the fibre that is close to the attachment area and replace it by a line segment whose end-point lies on the surface of the muscle in the region defined by the attachment area and that has the direction derived from the trimmed fibre. Finally, the muscle fibre is smoothed to eliminate any noise that might be present. An example of a decomposed Gluteus Medius muscle is given in Figure 2.

4. Surface-First Strategy

This strategy assumes that each muscle is associated with an action-line, that is, a poly-line that serves as the muscle "skeleton" and to which the surface model of the muscle is bound. For each time frame (current-pose position), the path of action line is recalculated using any of the action-line techniques (e.g., the obstacle set method described by Garner & Pandy [9] or the global method of Audenaert & Audenaert [10]). Our gradient domain deformation method [11] transforms the positions of the surface mesh vertices to accommodate the difference between the new and previous paths. This transformation is subject to Laplacian linear constraints to preserve the local shape of the mesh, and non-linear volume constraints to preserve the volume of the mesh. These constraints form a system of equations that is solved using an iterative Gauss-Newton method with Lagrange multipliers. For any of the muscles tested, 200 iterations were enough to achieve a volume error below 0.04%. The sequential processing of a typical mesh requires about 400 ms on commodity hardware. When parallelized using OpenMP, the required time may drop to 250 ms depending on the hardware configuration. We have found that the mechanism that prevents inter-penetration of muscles and bones can occasionally corrupt the results if it should happen that the coarse outer hulls of objects intersect significantly in their initial configuration; work on a more robust mechanism is in progress. An example of the results produced is given in Figure 3.

5. Fibres-First Strategy

This strategy starts from a decomposition of the muscles into fibres in their rest-pose positions. Within this decomposition, particles are generated by uniform sampling along the fibres, as shown in Figure 4 left, and the neighbouring relationships between particles are then identified – the 6 neighbour particles of particle *i* in the fibre are particles *i-1* and *i+1* on the same fibre, and particle *i* on the 4 adjacent fibres. The link between two neighbouring particles is considered a simple spring, with the springs along the fibre, and between the fibres, having different strengths to reflect the anisotropy of the muscle tissue. We note that spring stiffness constants were chosen empirically.

Each particle on the boundary of the fibres is checked to find out if it lies in the proximity of any bone. If this test is positive, the particle is set as fixed to its closest bone (see Figure 4 right) – when this bone moves, so does this particle. This induces the movement of other particles in order to keep the mass-spring system in balance.

For the purpose of speeding up this process, particles are first transformed rigidly based on the transformation matrix of the associated bone. Each unfixed particle on the boundary of the fibres is further checked for penetration with the bones for the current time frame; the positions of any penetrated particles are adjusted to place them on the boundary surface of the bone. This enables muscle-bone inter-penetration to be avoided.

Finally, particle-based simulation is performed to automatically adjust the position of other particles based on the mass-spring system, i.e., unfixed particles are successively translated in order to minimize energy of the whole mass-spring system and restore its equilibrium. As this process does not involve any deformed boundary surfaces of muscles, fibre inter-penetration between two muscles can occasionally occur during motion. Work to increase the robustness of the method is in progress.

6. Conclusion

The approach presented in this paper produces, in a convenient time (muscle decomposition runs in hundreds of ms, surface-first strategy without penetration avoidance takes hundreds of ms, fibres-first strategy can process all muscles of thighs and pelvis in a couple of minutes), an arbitrary number of patient-specific muscle fibres that can be used instead of action lines for biomechanical predictions with an expected accuracy somewhere between those provided by action-line methods and the more accurate, but impractically slow, finite-element methods. The work is in progress and, when completed, will need to be biomechanically verified (currently, no preliminary biomechanical results are available). The proposed approach is an interesting solution for the clinical application of musculoskeletal models since realistic patterns are produced, within clinically acceptable time limits.

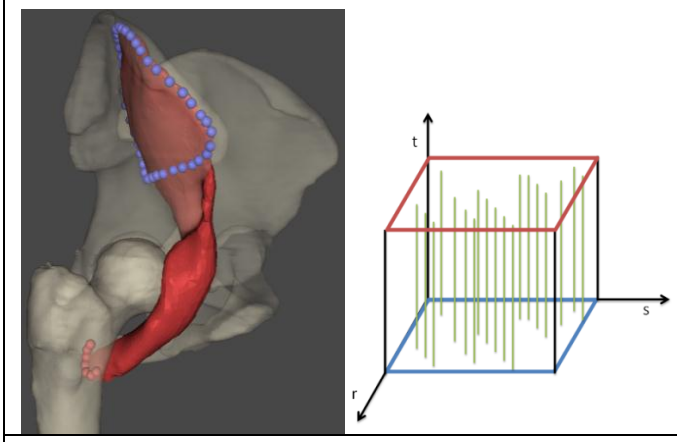

Figure 1. Iliacus, its attachment areas (origin is blue, insertion is red) and its template for the fibre geometry

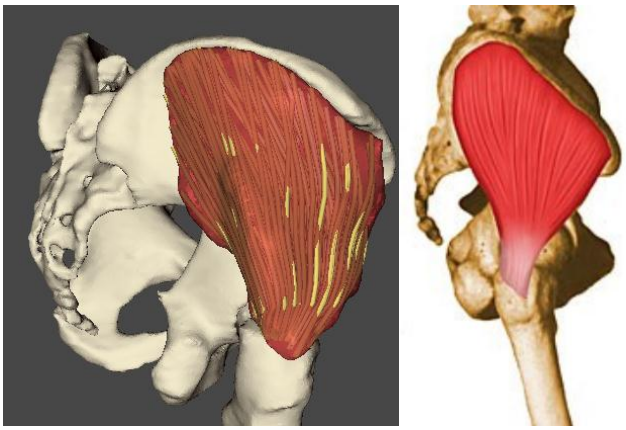

Figure 2: Gluteus Medius decomposed into fibres, compared with the illustration in Richardson [12].

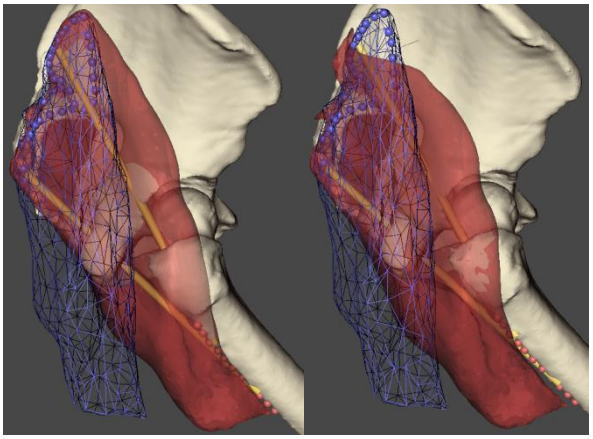

Figure 3. Gluteus Maximus before (wire) and after (solid) the deformation; without (left) and with (right) the inter-penetration avoidance mechanism.

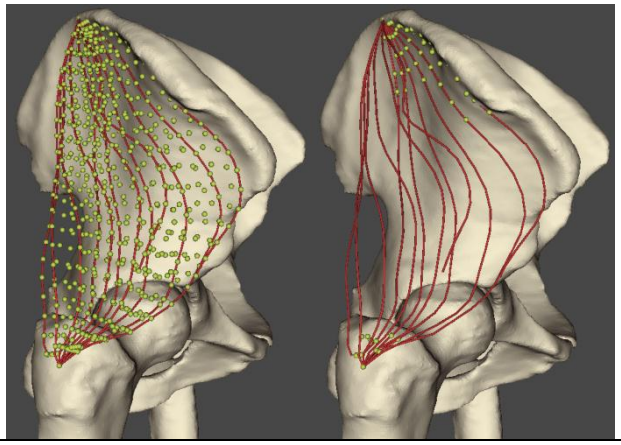

Figure 4. Gluteus Medius decomposed into particles (left) and fixed particles (right). Some fibres are displayed for comparison.

Acknowledgements

This work was supported by the Information Society Technologies Programme of the European Commission under the project VPHOP (FP7-ICT-223865). The authors would like to thank the various people who provided conditions under which the work could be done.

References

1. VPHOP: the Osteoporotic Virtual Physiological Human. http://vphop.eu
2. Kohout J, Kellnhofer P, Cholt D, Kohoutová E, Clapworthy GJ, Zhao Y, Tao Y, Gonzalez-Garcia G, Dong F, Fibre-based Models of Muscle Wrapping. *Submitted for VPH 2012*
3. Ng-Thow-Hing V, Anatomically based models for physical and geometric reconstruction of animals, PhD Thesis, University of Toronto, Department of Computer Science, 2001
4. Blemker SS, Delp SL, Three-dimensional representation of complex muscle architectures and geometries, Annals of Biomedical Engineering 33 5 (2005) 661-673
5. Viceconti M, Zannoni C, Testi D, Petrone M, Perticoni S, Quadrani P, Taddei F, Imboden S, Clapworthy GJ, The Multimod Application Framework: a rapid application development tool for computer aided medicine, Comp Meth Prog Biomed 85 2 (2007) 138-151
6. Schroeder W, Martin K, Lorensen B, The Visualization Toolkit: an object-oriented approach to 3D graphics 1998, Prentice Hall
7. Kohout J, Clapworthy GJ, Martelli S, Viceconti M, Muscle Fibres Modelling, In: Proc. GRAPP 2012, Rome, Feb 2012, pp. 58-66.
8. Ju T, Schaefer S, Warren J, Mean value coordinates for closed triangular meshes, ACM Transactions on Graphics 243 (2005)561–566.
9. Garner BA, Pandy MG, The obstacle-set method for representing muscle paths in musculoskeletal models, Computer Methods in Biomechanics and Biomedical Engineering 3 1 (2000) 1-30
10. Audenaert A, Audenaert E, Global optimization method for combined spherical-cylindrical wrapping in musculoskeletal upper limb modelling, Computer Methods and Programs in Biomedicine 921 (2008)8-19.
11. Kohout J, Kellnhofer P, Martelli S, Fast Deformation for Modelling for Musculoskeletal System, In: Proceedings of GRAPP 2012, Rome, Italy, February 2012, pp. 16-25.
12. Richardson M, Muscle Atlas of the Extremities, Amazon Whispernet, 2011.